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TRANSFER CHARACTERISTICS OF
LOAD MOVING SKELETAL MUSCLE

by
John Willis Gesink

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of the requirements for the degree of
Doctor of Philosophy
(Bioengineering)
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1973

Doctoral Committee:

Associate Professor William J. Williams, Chairman
Associate Professor Robert B. Keller
Associate Professor James H. McElhaney
Professor J. Raymond Pearson
Dr. James L. Cockrell Research Engineer

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TABLE OF CONTENTS

	PAGE
ACKNOWLEDGEMENTS	ii
LIST OF TABLES	v
LIST OF FIGURES	vi
LIST OF APPENDICES	viii
CHAPTER 1 INTRODUCTION	1
1.1 Foreword	1
1.2 Background	2
1.3 Research Objectives	15
1.4 Implications and Applications	16
CHAPTER 2 EXPERIMENTAL APPROACH	20
2.1 General Considerations	20
2.2 Animal Preparation	26
2.3 Apparatus and Instrumentation	27
2.4 Experimental Procedure	34
2.5 Data Acquisition	39
2.6 Data Analysis	44
2.7 Data Presentation	49
CHAPTER 3 RESULTS	55
3.1 Response to Modulation Frequency	55
3.1.1 Magnitude Response	55
3.1.2 Phase Response	63
3.2 Response to Inertial Loading	68
3.3 Response to Depth of Stimulus Modulation	71
3.3.1 Magnitude Response	71
3.3.2 Phase Response	75
3.3.3 Resonance Response	75

	PAGE
3.4 Response to Mode of Stimulation	76
3.5 Harmonic Analysis	77
3.5.1 Coherence Measurements	77
3.5.2 Signal to Noise Ratio Measurements	87
CHAPTER 4 DISCUSSION	89
4.1 General Remarks	89
4.2 Muscle Model Development	91
4.2.1 The Single Motor Unit Model	92
4.2.2 Multiple Motor Unit Model	97
4.2.3 Nonlinear Time Variant Model	103
4.3 Discussion of Models and Results	108
4.3.1 Form of the Frequency Response	108
4.3.2 Depth of Modulation	112
4.3.3 Inertial Loading	121
4.3.4 Stimulation Mode Sensitivity	122
4.4 Functional Interpretations	128
4.4.1 The Low Pass Filter Characteristic	128
4.4.2 The Non-minimum Phase Characteristic	129
4.4.3 Stimulation Mode Sensitivity	131
4.5 Muscle as a Control System Element	132
CHAPTER 5 SUMMARY	139
REFERENCES	144
APPENDICES	150

LIST OF TABLES

TABLE	PAGE
1. Summary of experimental results.	110

LIST OF FIGURES

FIGURE	PAGE
1. Visco elastic model and mechanical properties of skeletal muscle.	5
2. Contractile element response to neural excitation.	8
3. Diagram of experimental apparatus.	24
4. Diagram of neural stimulator.	30
5. Typical input and output signals of the neural stimulator.	33
6. Isometric force response of the muscle to three different modes of stimulation.	36
7. Inertial load response to discrete frequency sinusoidal modulation signals.	41
8. The swept frequency method of data acquisition.	43
9. Time series analysis techniques used to process the swept frequency data.	47
10. Bode plots of typical describing function data.	51
11. Bode plots comparing experimental and modelling results.	53
12. Bode plots of experimental results with zero inertial load.	57
13. Bode plots of experimental results when the inertial load was set to an intermediate value.	59
14. Bode plots of experimental results with large inertial loads.	61

FIGURE	PAGE
15. Bode plots showing phase discrepancy between the experimental and modelling results.	65
16. Modelling and experimental results showing the effect of inertial loading on the response.	70
17. Static transfer characteristic of the muscle-load system.	74
18. Comparison of experimental results showing the effects of mode of stimulation.	79
19. Coherence functions of experimental data for zero and intermediate inertial loads.	81
20. Coherence functions of experimental data for large inertial loads.	83
21. Signal to noise ratios of experimental data.	85
22. Single motor unit model of the muscle load system.	94
23. Bode plot of the response of the single motor unit model.	96
24. Multiple motor unit model of muscle.	99
25. Nonlinear parametric model of muscle.	105
26. Contributions of the passive muscle and the inertial load to describing function phase lag.	115
27. Passive elastic characteristic of muscle.	117

LIST OF APPENDICES

APPENDIX	PAGE
I Stimulator block diagram	150
II Cat body moment of inertia	155
III Describing function definition	156
IV Time series analysis techniques	157
V Describing function modelling	161
VI Harmonic Analysis	163

CHAPTER 1

INTRODUCTION

1.1 Foreword

Skeletal muscle is the effector organ of the neuro-muscular system. It's function is to convert the neural information flow from the central nervous system (CNS) into muscular contractions. Many studies directed at examining muscle function have produced information relative to muscle energetics and mechanical properties. The sliding filament theory which relates the structure of contractile proteins to gross muscle force probably best represents current thinking regarding muscle energetics and contractile processes. This branch of muscle physiology, although of major significance, contributes little toward describing muscle as an interacting element in the complex closed loop neuro-muscular control system. As a key component in this system, muscle performs an important information processing role. When carrying out motor actions, muscle receives and processes massive amounts of neural information. Muscle decodes the neural signals arriving from the CNS and utilizes its energy converting contractile function to produce an appropriate mechanical output. In this sense, muscle may be examined using the input-output approach of control

system theory. The input to the muscle is a neural signal and the output of the muscle is the mechanical version of this signal after it has been transformed by the muscle.

The problem of characterizing muscle as a dynamic neural information processor is the subject of this study. Of particular interest is insight into the nature of the transformations muscle imposes on neural signals as it converts them into mechanical signals. In short, the input-output transfer characteristic of muscle is sought.

1.2 Background

A considerable amount of information is already available concerning the information processing role of muscle. Over the last half century there have been numerous investigations directed at determining muscle energetics, thermodynamics, and mechanical properties (1,2,3,4). These studies, although not explicitly directed at examining the transfer characteristics of muscle, reveal significant factors affecting this study. Most important, for this investigation, various conceptual muscle models have been produced which consolidate the results of a diverse spectrum of muscle studies. Further, after making several simplifying assumptions,

it is possible to express the conceptual model in a mathematical form which may be used to examine the transfer characteristic of muscle. Figure 1a illustrates a version of this model while the rationale for selecting this configuration is thoroughly covered elsewhere (1,5). The model, consisting of viscous and elastic components, is a version of the single motor unit, four element, lumped parameter model widely used in the literature (1,5,6,7). While the model is conceptually quite simple, inclusion of all the time varying and nonlinear properties of muscle render the model mathematically intractable. Figures 1b through 1e illustrate some of the general nonlinear characteristics of the total muscle that the model must account for (8, 9,10). Figure 1b shows the elastic characteristics of the muscle. Here the passive elasticity is nonlinearly related to muscle length. In the active state, the tensile force produced by the muscle is nonlinearly related to both muscle length and level of stimulation (Figures 1b and 1c). Figures 1d and 1e indicate the nature of velocity dependent or viscous components of the active muscle. Figure 1d shows a viscous component which has a saturation type of nonlinearity in the positive velocity direction and a component which is

FIGURE 1

- a. Four element visco-elastic model of muscle¹.
 CE - contractile element; K_p - parallel elastic element; C_p - parallel viscous element;
 F_m - muscle force; F_c - viscous element force;
 $V(t)$ - neural excitation; MN - motor neuron;
 τ - conduction activation delay; $f(y,t)$ -
 force response of CE to neural excitation; s -
 Laplace operator.
- b. Isometric (constant length) force response of
 muscle v:s muscle length (x) for various levels
 of neural excitation (F_s).
- c. Isometric force (F_m) response v:s level of
 neural excitation (F_s) for various muscle
 lengths.
- d. Isotonic (constant force) force response v:s
 rate of change of muscle length ($\dot{x}$) for
 various levels of stimulation (F_s).
- e. Force-velocity characteristic of parallel
 viscous element for various levels of steady
 state neural stimulation (F_s).

1 Many models of muscle found in the literature combine the contractile and parallel elastic elements into either a neurally controlled length dependent contracting element or a neurally controlled elastic element. This model separates these two functions by using both contractile and elastic elements under neural control.

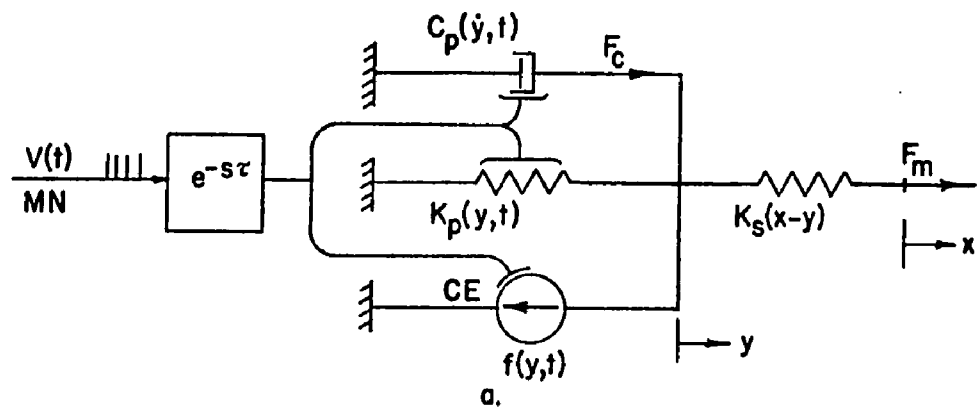
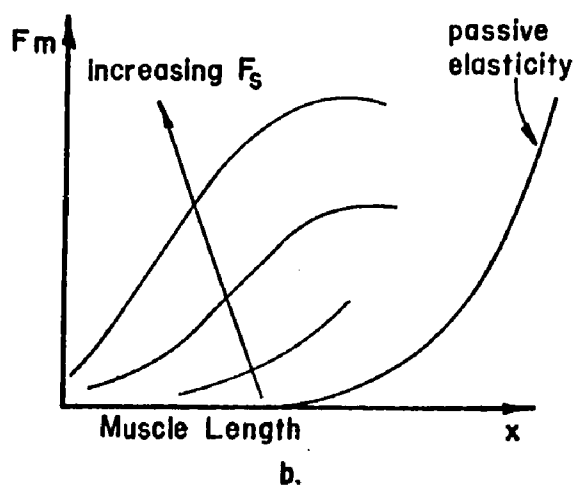
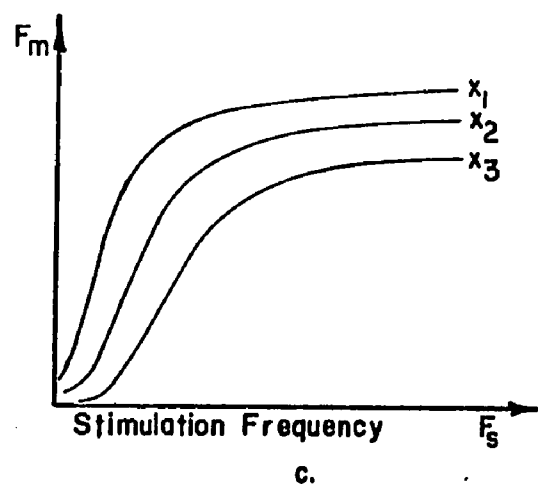
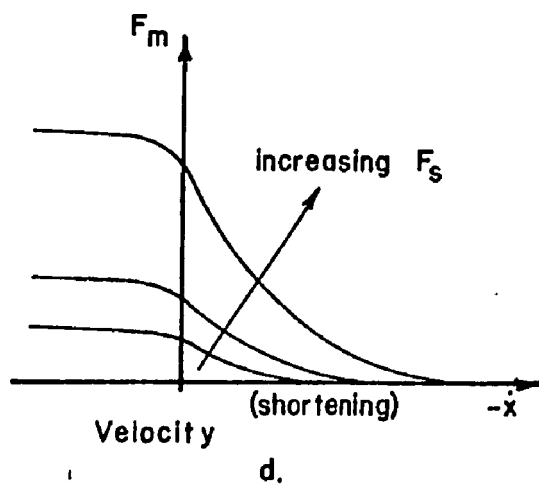
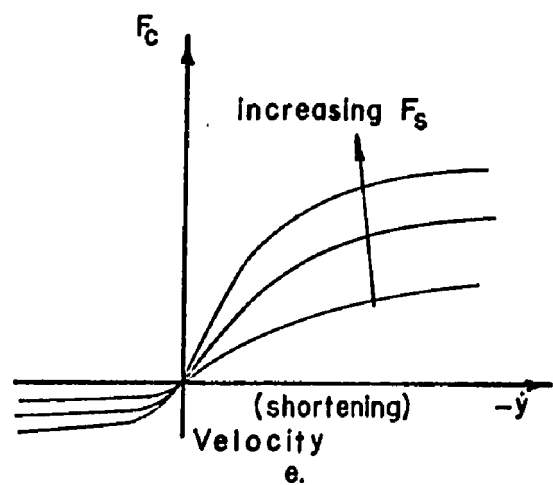


FIGURE 1

relatively independent of velocity in the reverse direction. Examining the response of the individual components of the model indicates various detailed characteristics of muscle which influence its transfer characteristics. Figure 2 illustrates the response of the contractile element to both a single stimulus pulse as well as to a train of pulses. Upon arrival of a neural pulse, the muscle enters an "active state". In the case of a single pulse, the response of the contractile element (CE) is characterized by a brief tension force pulse (11,12,13). The tension rises very rapidly to a peak, dwells momentarily, then decays exponentially with a time constant of the order of 10's of milliseconds. The CE displays an exponentially decaying type of memory for a single stimulus pulse. When the muscle is subjected to a train of pulses however, it immediately sustains a loss of memory with the arrival of each new pulse, i.e., sequential pulses do not superimpose on one another (14,15,16). Thus under constant frequency stimulation, the CE tensile force soon saturates as stimulus frequency increases (Figure 2). This saturation effect partially explains the stimulus level non-linear saturation characteristics seen in the isometric response of Figure 1c. The isometric response reflects

FIGURE 2

General form of the response of the contractile element to neural excitation at fixed CE length. $f(t)$ is the normalized "active state" force pulse response of the CE to a single stimulus pulse. $F(t)$ is the composite force response to repedative stimulation (F_s). $F(t) = f(t)$ in the case of a single stimulus pulse.

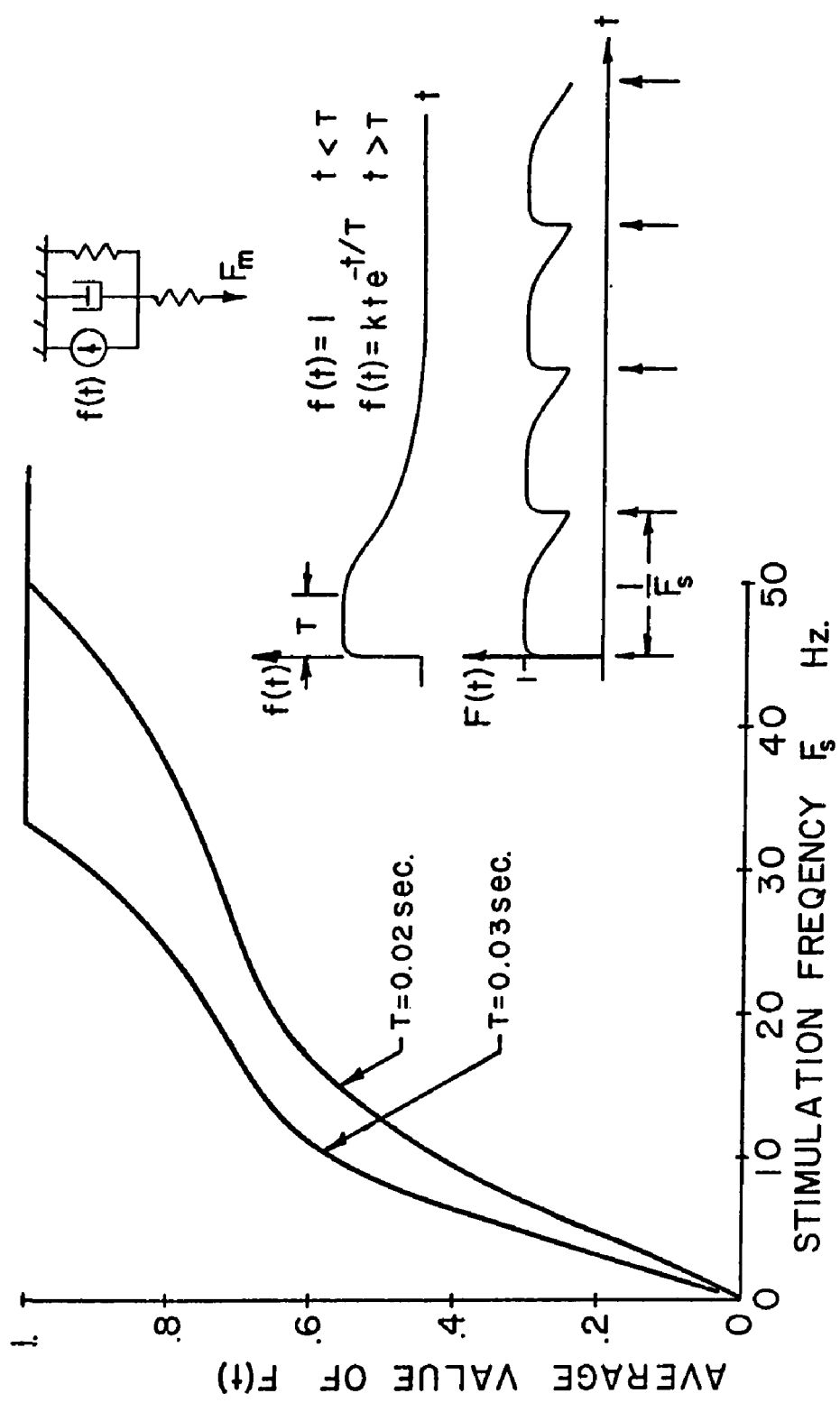


FIGURE 2

the summed responses of all the CE's. As the individual CE's saturate with increasing stimulation frequency, the isometric force of the whole muscle must also saturate.

The CE force pulse, in addition to being influenced by stimulus level, is also sensitive to muscle length. Because both the shape and amplitude of the pulse vary with CE length, the overall characteristic of the muscle exhibits the nonlinear effect seen in Figure 1b.

An additional relevant characteristic of muscle is its latency in activation. Upon arrival of a neural pulse at the motor endplate, the muscle exhibits a latent period (0 to 10 msec.) or transport delay prior to entering the active state (17,18). In the intact closed loop neuromuscular system this delay along with other conduction and transmission delays in the system produces a de-stabilizing effect on the dynamic performance of the system (19). This activation delay is modelled by the e^{-sT} block in Figure 1a and may be significant in investigating the transfer characteristic of the muscle.

Another feature of muscle performance which must be accounted for by the model is the neural control of both the parallel viscous and elastic components. It has been suggested (20,21) that these two components have both a nonlinear passive component and also a neurally

controlled active component, i.e., $C_p(\dot{y}, t)$ and $K_p(y, t)$ may be formulated as $C_p(\dot{y}, t) = C_p(\dot{y})(1 + \epsilon_1 F(t))$ and $K_p(y, t) = K_p(y)(1 + \epsilon_2 F(t))$ where: $F(t)$ is the time course of the active state, the ϵ_i 's are small positive constants. $C_p(\dot{y})$ and $K_p(y)$ are the passive nonlinear viscosity and elasticity. Describing the parallel components of the model in this manner conceptually accounts for the increase in viscosity and decrease in compliance which occurs experimentally with increasing levels of stimulation (20,7).

While the conceptual model is not complete, it does pinpoint several characteristics of muscle which may be directly related to the information processing role of muscle. The components of the muscle exhibit both position and velocity nonlinearities, a time variance which depends on level of stimulation, and a transport delay in activation.

A significant anatomical feature of muscle relevant to its information processing role is the parallel motor unit structure of muscle. The muscle is composed of hundreds of parallel coupled motor units, each unit having its own channel of communication with the central nervous system. While all of the motor units respond in an all or none manner, each is distinct in the detail features

of its dynamic response (22,23). Further, the motor units in a muscle may be grouped into several populations where the grouping is based on histochemistry and size (22,23,24). The CNS, through selective spatial and temporal recruitment of the individual units precisely adjusts the output of the muscles to each specific motor task. Thus in addition to possessing multidimensional nonlinearities and time varying components, the muscle is a multi-input system with nonhomogeneous spacially distributed dynamic characteristics.

Additional factors relating to the information processing role of muscle deal with the order of recruitment and the distribution of sizes of motor units in the muscle. It has been shown that there is a direct relationship between the size of a motoneuron and the associated motor unit's maximum force producing capability (25). Large motor units producing large forces are subserved by large motoneurons, while small motor units are innervated with small motoneurons. It has also been shown that the excitabilities of motoneurons in a given motor neuron pool are inversely proportional to the motoneuron sizes. The smaller a neuron is the more easily it is fired; the larger it is the greater is the amount of excitatory input required to discharge it.

Henneman et al. (25,26,27) have used these facts to formulate and verify a general rule defining the order of motor unit recruitment in any motor task. This rule states that in a graded motor task, the motor units of a given muscle can only be fired in an order determined by their sizes, small units, being the most excitable, are recruited first and most frequently and large units last and relatively infrequently. This ordered pattern of motor unit recruitment by the CNS is probably a major factor in the information processing role of muscle.

Among the literature, primarily in more recent contributions, some studies specifically investigate muscle as an information processing component. Several investigators have interpreted the isometric twitch to be the impulse response of muscle and have successfully modelled it with linear transfer functions containing several time constants (5,28,29). This interpretation, although somewhat valid for single twitches, has been shown to be completely inadequate when the input to muscle is a complex, naturally occurring neural signal (20,30,31).

Several frequency response studies of the isometric force response of muscle to various modes of sinusoidally modulated neural signals have been performed. Sinusoidal

modulation of the neural pulse rate (simulating temporal recruitment of motor units) results in a low pass filter type of frequency response characteristic (32,33,34).

When the number of motor units is sinusoidally modulated (simulating spacial recruitment of motor units) the isometric frequency response characteristic also shows a low pass filter type of response (5). A third case involves simultaneous modulation of both the neural pulse rate and the number of motor units (simulating a combination of spacial and temporal recruitment). The frequency response of this case also displays the low pass filter characteristic (34).

Partridge (35) did a comprehensive study of the information processing characteristics of load moving skeletal muscle. He constrained muscle to work into an inertial load of physiological size and forced the muscle with a neural pulse train whose frequency was sinusoidally modulated. Then considering the inertial load position as output, a frequency response study was made. Assuming that muscle acts primarily as a force generator, his predictions indicated its response would be very sensitive to inertial load size. Instead the frequency response was found to be very similar to a positioning servo, the response being relatively independent of inertial loading.

Partridge attributed this inertial compensation to some internal feedback mechanism located within the body of the muscle. Later, he realized that the inertia compensating mechanism was due to the viscoelastic properties of the muscles structural components and contractile machinery (36,37). His major contribution was his study and description of muscle under a very general set of realistic conditions. A majority of the studies previously reviewed were performed under a less realistic but experimentally more convenient conditions. In them, many of the muscle's response modifying variables of force, velocity of shortening, length, and stimulation parameters were constrained while attention was focused on the few that were free to vary. If the muscle were a linear device, the results of such studies might be used to predict performance in more realistic conditions such as the positioning of a limb (i.e. moving an inertial load) where most of the variables are changing simultaneously. Muscle is multidimensionally nonlinear making such prediction extremely difficult if not impossible. Partridge recognized this in the design of his experiment. Not only was a dynamic neural signal used; but also the muscle was loaded in a manner to allow its force, length, and rate of shortening to vary simultane-

ously. Thus realistic test conditions were used and the nonlinearities associated with each of the muscle's variables were acting together to produce a physiological movement.

1.3 Research Objectives

In his frequency response study of skeletal muscle, Partridge (35) considered two of several rather fundamental determinants of muscle performance. The first of these was the mode of neural information transmission. By modulating the pulse frequency of the neural signal the temporal recruitment mode of information transmission was simulated. Pulse frequency modulation is only one of at least three modes used by the nervous system to transmit information to muscle. The second mode is spacial recruitment and the third is a combination of both spacial and temporal recruitment. Spacial recruitment is effected by activation of different numbers of motor units (26,38). When a combination of spacial and temporal recruitment is used, the muscle machinery is modulated by changes in activity of a single motor unit as well as by changes in the number of active units (38).

The second determinant of performance considered by Partridge was muscle loading. A single muscle was coupled to both an inertia disc and a weight. The muscle

provided an active torque for displacement of the inertia of both the disc and the weight while the weight produced a passive restoring torque. In the intact animal, because of their unidirectional properties, muscles or groups of muscles are arranged in antagonistic pairs. This "push-pull" configuration, a feature essentially ignored by Partridge, provides a means of actively displacing a load in either direction. In addition, the antagonistic member of the pair may contribute to the load on the agonist, either actively or passively, by supplying a retarding or braking torque (39).

In this study the "push-pull" property as well as the three modes of neural information transmission are considered. The objective of this work is to obtain the open loop frequency response of a pair of muscles which are antagonistically coupled to a common inertial load and stimulated in ways approximating the known modes of neural information transmission. Further, an attempt to interpret and evaluate this response in terms of the physiology of the neuromuscular control system will be made.

1.4 Implications and Applications

While the major thrust of this study will be directed toward basic research, its results might be of some

significance on a more practical level. One possible area of application is FNS (Functional Neuromuscular Stimulation). FNS has been defined as the use of electrostimulation in the process of carrying out effective movements by activating muscles which are paralyzed as the result of a motor neuron lesion (40). Ever since Liberson (40) first employed FNS in restoring some function to a patient, considerable effort has gone into researching FNS orthotic systems (41). The motivation for pursuing this type of system, as opposed to methods using external power sources and technological actuators, is clear. The denervated muscle is not only ideally located and coupled to the skeleton, but is also capable of deriving its energy through normal biological processes. Additionally, some control over the muscle may be obtained simply by electrostimulation via external electrodes. The problems, around which much of the research on FNS is centered, are equally clear. Intact muscle is a motor or actuator that has a motor unit discriminating multi-input information channel, the motor nerve. In re-establishing an information link, as would be necessary in an FNS system, it is necessary to accept the fact that the delicate motor unit discriminating information channel cannot be duplicated with techno-

logical components (e.g. electrodes) and state of the art technology. Even if a technological multichannel link could somehow be established, generation of the necessary control information would itself constitute a monumental task. There have been some efforts which have attempted to restore the information channel and information source (41,42). These efforts have relied on schemes using a limited number of electrodes carrying processed electromyographic signals from muscles still under volitional control (41,42,43). The limited number of control channels and control sites in this scheme necessarily dictated a crude degree of muscle control as compared to that provided by the intact neuromuscular system.

Along with these fundamental problems, there are innumerable detail problems. For example, electrode materials and placement, stimulus current parameters, feedback between control and stimulus sites, and control information processing and transmission (42,43,44). Although the problems are numerous, the potential advantages of an FNS system continue to generate a considerable amount of research.

This study hopefully will contribute to a better understanding of one of the problem areas encountered in

the development of practical FNS systems. It is assumed that the technology of the near future will not provide a communication link to the denervated muscle which is significantly improved over the electrode pair types currently used. This constraint necessarily limits the possible degree of control over the muscle to a crude level. Within this framework, one of the problems to be solved is illustrated in the following question. What kind of performance, as a dynamic actuator, can be expected from denervated skeletal muscle under the conditions of electrostimulation by feasible methods?

In this study we will examine and attempt to characterize the dynamic performance of electrostimulated muscle. It is hoped that this study will contribute insight into the answer of the above question as well as provide information which will be useful in the design of improved FNS systems.

CHAPTER 2

EXPERIMENTAL APPROACH

2.1 General Considerations

In designing the experiment, it was necessary to make certain assumptions and approximations. The first of these concerns the pattern of the nerve signal delivered to the muscle. The intact animal uses asynchronous temporal and spacial recruitment of motor units (26, 31,38), but it is not clear how these two modes are scheduled for different motor tasks. Some evidence indicates that temporal recruitment dominates in volitional movement while spacial recruitment is operative during involuntary activity such as reflex postural control (45). Other evidence suggests that the grading of muscular activity is accomplished mainly through spacial recruitment (38). The work of Henneman et al. (25,26,27) showing that there is an ordered pattern of recruitment of motor units in a given motoneuron pool is a further factor of significance.

These factors presented considerable difficulty in designing synthetic neural signals to simulate those in the intact animal. The relative contributions of spacial and temporal recruitment used in an intact animal for the type of task under consideration are not precisely known.

Even if they were, to simulate them would require an ensemble of motor unit discriminating electrodes as well as a good deal of computing equipment to generate the required temporal sequence of pulses for each motor unit electrode.

Because of these factors, no attempt to exactly duplicate the naturally occurring neural signals was made. Instead the following procedure which approximates invivo recruitment was followed. The motor nerve subserving each muscle was severed to open the neuromuscular control system loop. To each nerve was attached a single stimulating electrode pair. This procedure provided a means of generating any combination of essentially two independent pseudo-physiological synthetic nerve signals. The two were obtained by independent amplitude (AM) and frequency modulation (FM) of the stimulus pulse train applied to the nerve. FM produces an essentially synchronous pulse frequency coded neural signal and thus approximates naturally occurring temporal recruitment, its major shortcoming being that it is synchronous. Amplitude modulation of the stimulus pulse train above threshold level produces a distinct type of spacial recruitment which, unfortunately, is essentially the reverse of the pattern which has been

shown to occur in the intact animal by Henneman et al. (25,26,27). This reverse pattern is also a result of the relationship between size and excitability of motor neurons and their associated motor units. It has been shown that the excitability of an electro-stimulated neuron in a motor nerve trunk is directly related to the neuron's diameter (46). Thus, as the amplitude of the stimulus pulse is increased, large axons innervating large motor units are the first to respond and the smaller units follow in the order determined by their axon diameters. Thus the recruitment pattern resulting from electrostimulation of the motor nerve is the reverse of that which, as shown by Henneman, occurs naturally.

Another approximate procedure used in this study concerns the pair of muscles chosen for the study. Rather than using a naturally occurring extensor-flexor antagonistic pair, two identical extensor muscles from opposite limbs were coupled to a common inertial load (see Figure 3). This procedure was followed for several reasons. Coupling two identical muscles in an antagonistic push-pull configuration has certain advantages from a systems analysis point of view. Each muscle is an unidirectional nonlinear actuator. Coupling them

FIGURE 3

Diagram of experimental apparatus showing the preparation table and the instrumentation platform. The load cells monitor cable tendon tension. The potentiometer monitors inertial load position.

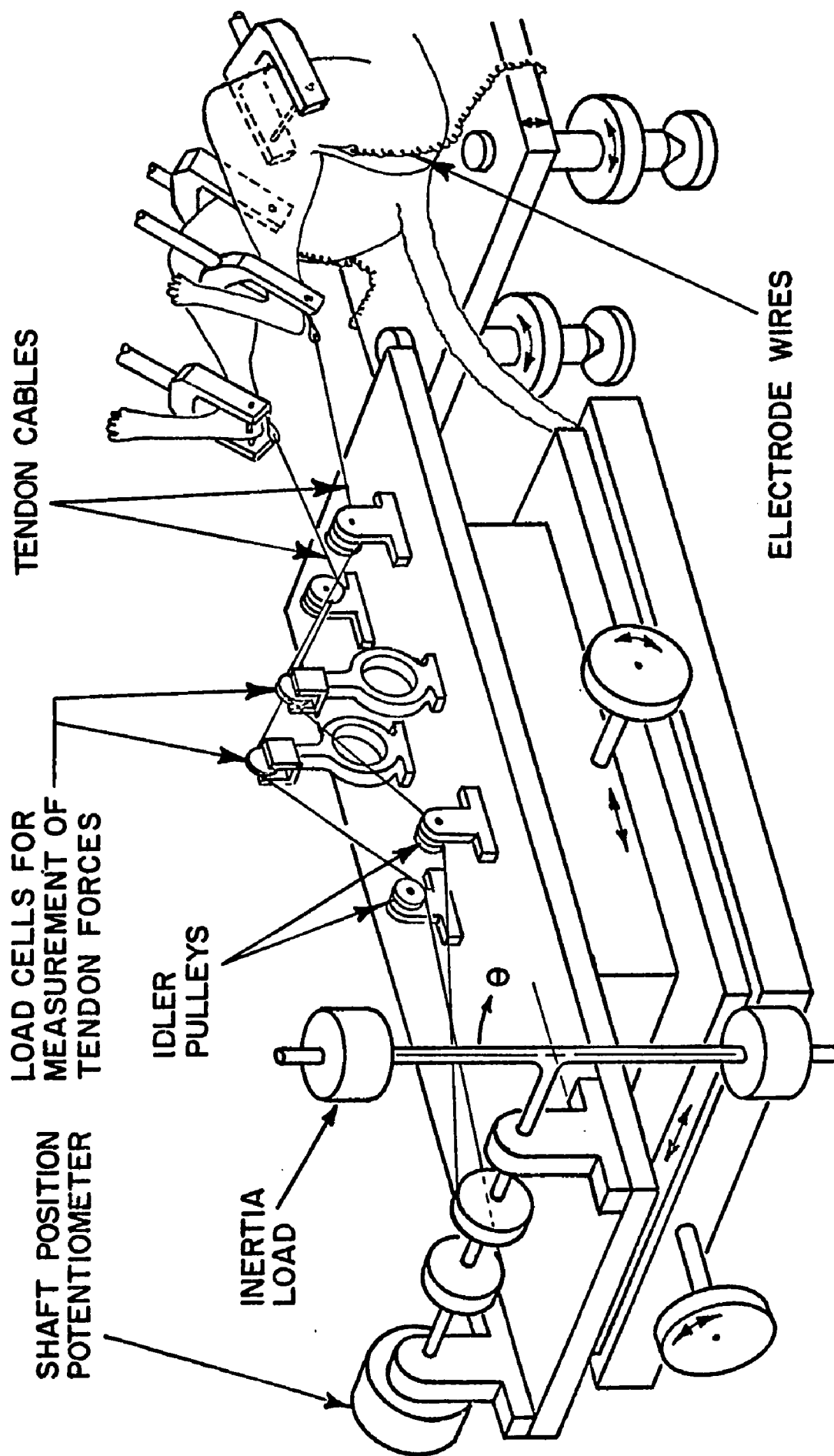


FIGURE 3

antagonistically results in a bilateral actuator with symmetrical nonlinearities. Techniques for analyzing the frequency response of symmetrical nonlinear systems have been well developed and produce an analysis from which some of the nonlinearities may be identified (47). Using these techniques, identification of the characteristics of a "matched" pair of muscles was possible. A naturally occurring antagonistic muscle pair likely would not be matched. Interpretation of results would be more difficult in the case of a natural pair of muscles. In addition, symmetrical nonlinear devices, when driven with an unbiased sinusoid, (as is to be used in this study), suppress even harmonic distortion. This eases the task of data acquisition, analysis, and interpretation (47).

Another factor is methodological. There are several technical problems associated with mechanically coupling tendons in a manner such that their structural and mechanical properties are preserved. As will be discussed later, the anatomical details of tendon insertion of either of the muscles chosen for this study suggest a coupling technique that was both convenient and mechanically reliable.

2.2 Animal Preparation

The data to be reported here was obtained in 13 experiments. Both triceps surae (gastrocnemius and soleus groups) muscles of adult acute cats were used. The animals ranged in weight from 2.6 to 5.1 Kg. and were anesthetized with sodium pentobarbital (30mg/kg). Smaller sustaining doses were administered intravenously as needed to maintain the level of anesthesia. In all experiments, the animal's temperature was continuously monitored and maintained at 38°C. Experiments ranged in length from 18 to 28 hours, the first 6 to 10 hours being devoted to animal preparation and the balance of the time to data collection.

The triceps surae muscles were prepared by freeing the Achilles tendons along with the dorsal portion of the calcaneus (heel) on which the tendons insert. Bone pins were placed in the freed portion of the calcaneus. Cable clamps were then attached to the bone pins. This procedure preserved the mechanical and structural characteristics of the tendons and also proved to be a convenient and mechanically reliable method of coupling to the tendons. Potential contributions by the plantaris muscles were eliminated by severing their tendons.

Each hind limb was rigidly coupled to a preparation

table with a femur pin and an ankle clamp (see Figure 3). Both tibial nerves which innervate the triceps surae muscles were transected at their points of junction with the sciatic nerves. The nerves were then threaded through circular, epoxy encased, silver stimulating electrodes. The internal diameter of the electrodes was slightly smaller than the nerve diameter to insure complete circumferential contact between nerve and electrode.

During all surgery, care was taken to preserve the blood supply of the muscle. After surgery, all wounds were closed with wound clips and that portion of the tendon extending through the skin was wrapped in gauze and frequently bathed in isotonic saline to prevent dehydration. At the end of each experiment the animal was sacrificed by administration of an overdose of Nembutal.

2.3 Apparatus and Instrumentation

The mechanical apparatus, illustrated in Figure 3, consisted of a preparation table and an instrument platform. A rotary inertial load, consisting of a pair of matched weights, was mounted as illustrated in Figure 3 on a common shaft supported by instrument ball bearings. The freed tendons were coupled to the inertial load by

cables (Bergen Wire Rope BWR 73.6) anchored to take-up pulleys on the load shaft. The cables were threaded through load cells (Statham UC3) which continuously monitored tendon tension. A device for locking the inertial load shaft provided a way of holding muscle length fixed while isometric force characteristics were determined.

Inertia load angular position was monitored with a potentiometer and the size of the load could be changed by adjusting the distance between the two weights. The tendon take-up pulleys on the load shaft were circular with a 2.54 cm diameter. Thus the normal variation in muscle lever arm that occurs with changing joint angle was neglected in these experiments. A preliminary experiment showed, however, that in cats as in humans (48), the variation in the Achilles tendon lever arm is small.

A schematic diagram of the nerve stimulator is shown in Figure 4. The stimulator delivered trains of constant current 1.0 msec. duration pulses to the nerves. The pulse trains were sinusoidally modulated either in frequency (FM), in amplitude (AM), or in both frequency and amplitude¹. In all three modes the pulse train was

1 Note that two frequencies, the pulse train frequency and the frequency of the modulating sinusoid, are required to describe a given stimulation condition. Thus to eliminate any ambiguity the two will henceforth be designated as the pulse frequency (f) and the modulation frequency (ω).

FIGURE 4

Schematic diagram of nerve stimulator, inertial loading configuration and describing function definition. See Appendix I for a more complete description of the stimulator

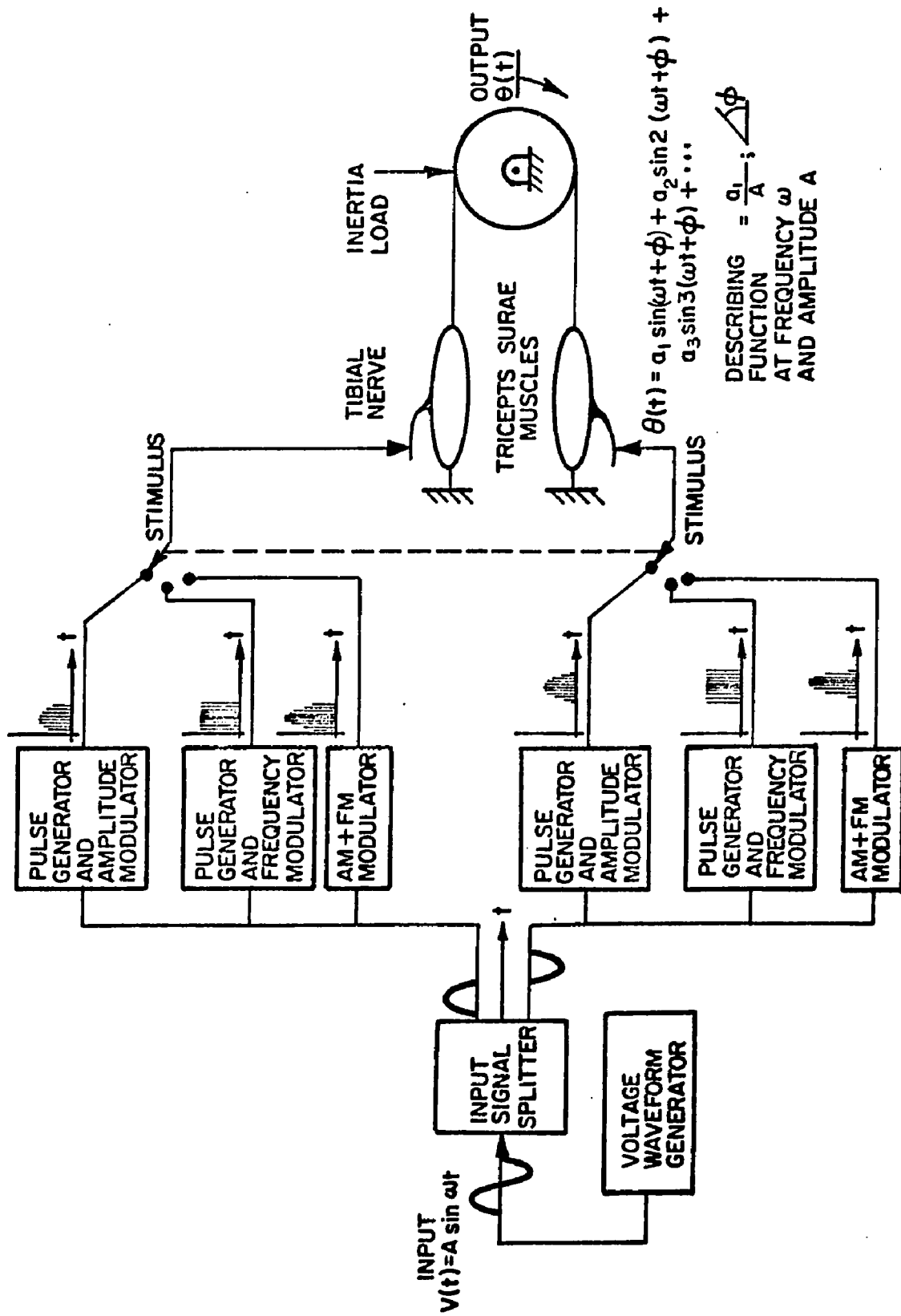


FIGURE 4

commutated, depending on the sign of the modulating sinusoid, between the muscles in a 'push-pull' manner. Figure 5 illustrates a strip chart record of each of these modes. In the FM mode, pulse amplitude was supramaximal while in the AM case stimulus frequency was held constant. In both the AM and AM+FM cases a constant amplitude pulse train was added to the modulated signal on each side and the amplitudes of these biasing pulse trains were adjusted to compensate for differences in nerve threshold properties on the two sides.

It was necessary to construct several special purpose circuits to implement the stimulator. A voltage to frequency converter using a unijunction transistor controlled resetting integrator was the primary part of the pulse frequency modulator. Amplitude modulation was accomplished with electronic multipliers. A full wave rectifier voltage comparator and reed relays implemented the commutation of the stimulus between the two muscles. Two voltage-to-current converters using field effect transistors provided the output constant current stimulus pulses. Block diagrams of the interconnections of these circuits may be found in Appendix I.

FIGURE 5

Typical records of the three modes of neural stimulation. The upper-most record is the input modulating signal to the stimulator. In the AM and AM+FM cases, the pulse amplitudes were biased to a threshold level to accommodate differences in the threshold properties of the motor nerves of the two muscles.

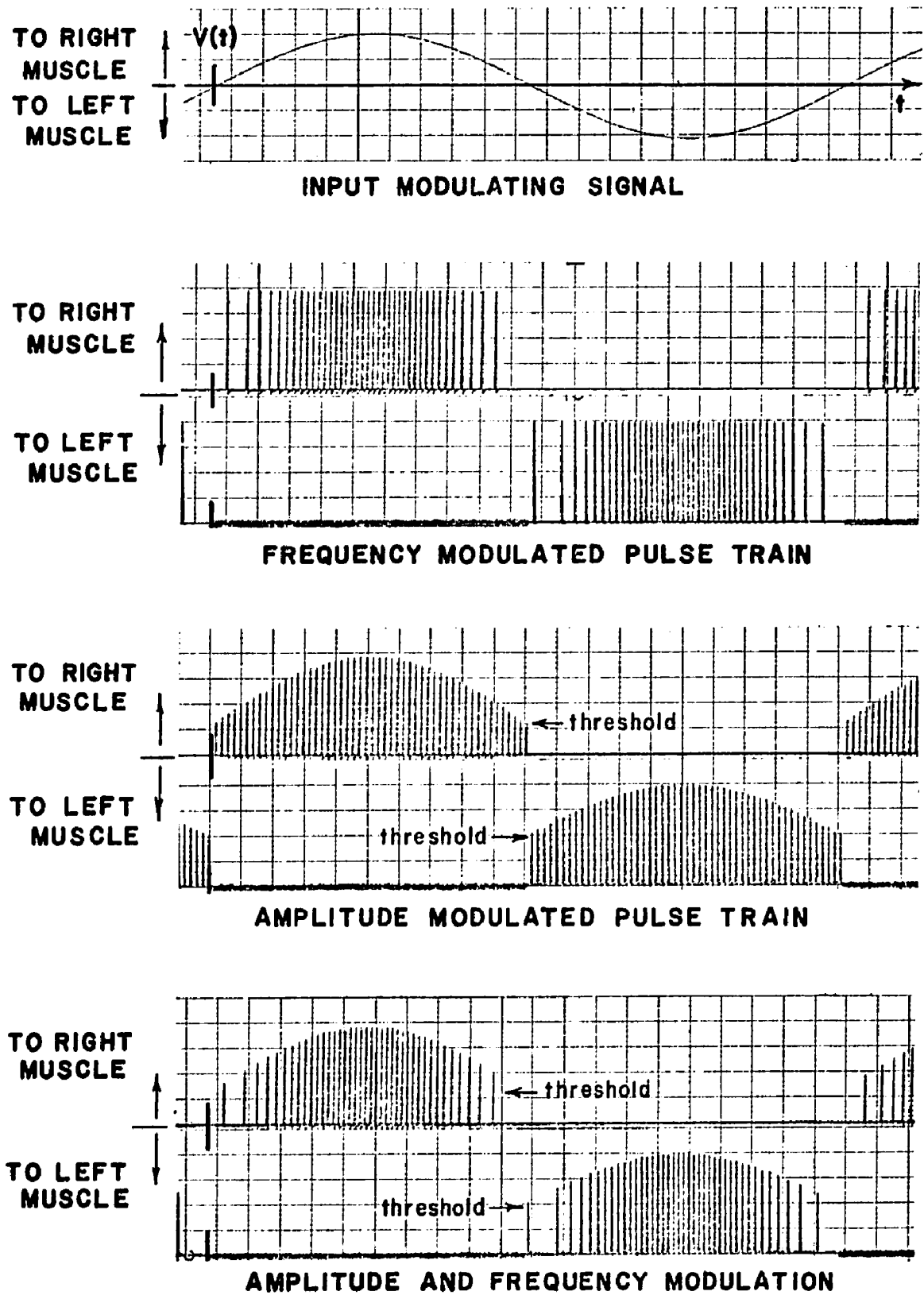


FIGURE 5

2.4 Experimental Procedure

Frequency response data, consisting of the modulating signal and the resulting inertia load motion were collected over a 0 to 8 Hz modulation frequency bandwidth. Individual trials were run for various combinations of the following three parameters:

- 1) depth of modulation, 2) size of inertial load, and
- 3) mode of stimulation.

Depth of modulation, expressed as a percentage, was based on the isometric force response of the muscle. Figure 6 illustrates typical isometric responses for the various modes of stimulation. These curves were obtained by slowly sweeping the pertinent stimulation parameters over a range large enough to produce a symmetrical saturated isometric force characteristic. In the FM case a stimulus pulse frequency of 70 Hz usually produced sufficient isometric force saturation and was assigned the 100% modulation level for all experiments. At the low end 0% modulation corresponded to 0 Hz pulse frequency. In the AM case, the pulse frequency was fixed at 60 Hz. The zero reference threshold and the maximum stimulation amplitudes were set on the basis of muscle response. A stimulation level producing a just noticeable isometric response

FIGURE 6

Typical isometric force response of a pair of muscles to the three modes of stimulation. The abscissa parameter, percent modulation, is explained in the text.

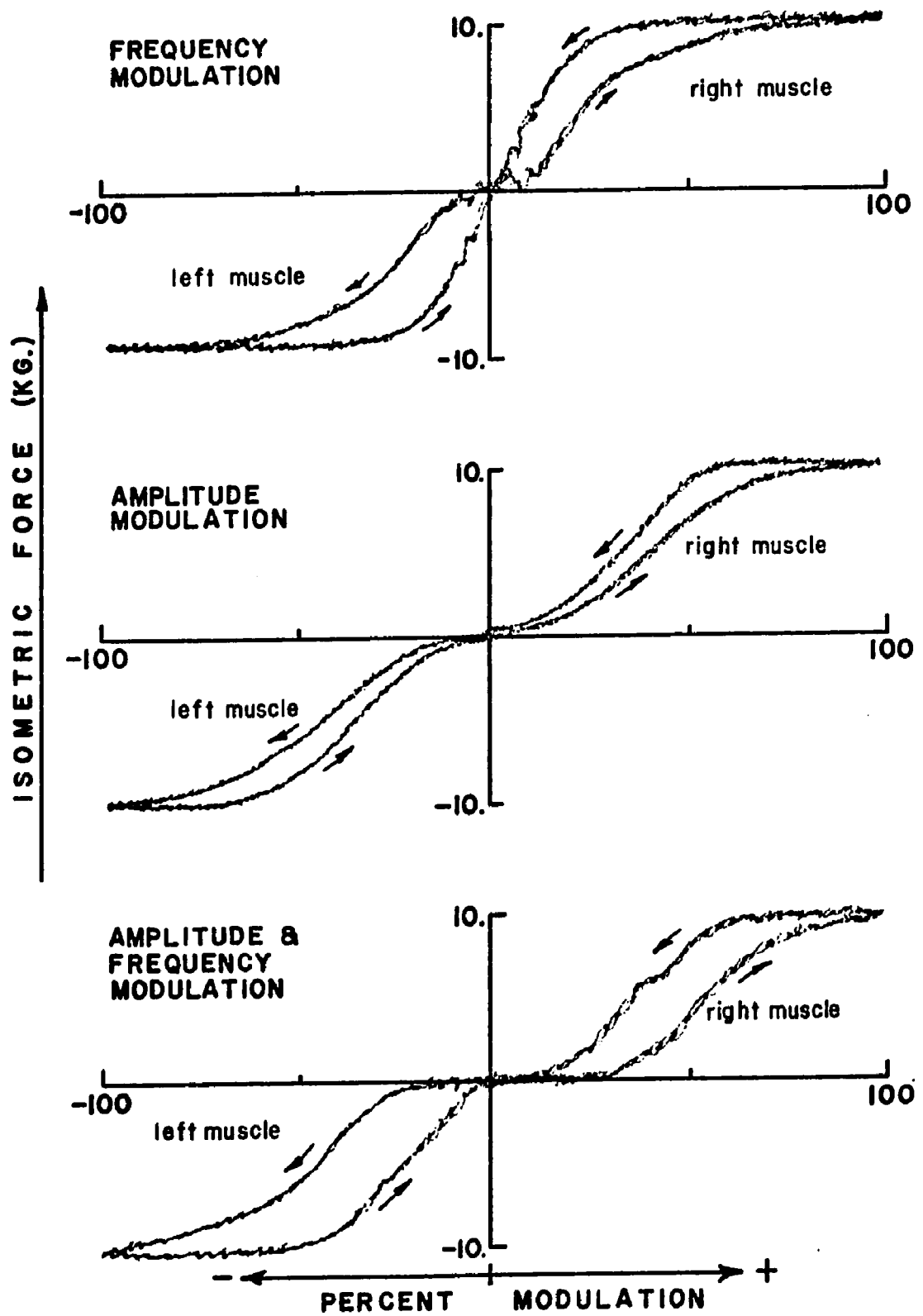


FIGURE 6

was taken as threshold. The criterion for maximum amplitude was that the degree of saturation of the isometric force curve be qualitatively the same as in the FM case. The maximum and threshold amplitudes thus obtained for the two muscles were taken as the 100% and 0% modulation levels respectively. In the AM+FM case, the pulse frequency was modulated between 0 and 70 Hz while the pulse amplitude was modulated between the threshold and the maximum value established in the AM case. Thus 100% modulation corresponded to a pulse frequency of 70 Hz at maximum amplitude while 0% corresponded to 0 Hz and threshold amplitude. This total procedure of establishing modulation levels was performed at the beginning of each experiment and frequently throughout its duration. If the isometric force saturation levels at any point during the experiment varied significantly from their initial values, the muscles were given a rest period after which changes in stimulation parameters necessary to restore the isometric curve to its original form were made.

The various inertial loads used in these experiments were all within the physiological range. The upper limit of this range is the moment of inertia of the cat's whole

body about its ankles in the sagittal plane. A numerical estimate of this limit was obtained in a preliminary experiment using compound pendulum techniques (49) on the body of a cat frozen in an upright position and pivoted about its ankles. The upper limit for all other animals was extrapolated from this measured value by assuming that the other animals were different only in size but not in shape or density. A graph of these data is shown in Appendix 2. All inertial loads will be referred to by a number representing the ratio of the experimental load to the animal's moment of inertia about its ankles. Several frequency response data gathering trials were run with the inertia load weights removed from the apparatus leaving only the inertia of the load shaft, cable pulleys, and potentiometer shaft. The inertia of these components was generally less than one thousandth of the inertia of the cat's body and thus will be referred to as a zero load.

For both stimulus level setting procedures and frequency response data gathering trials, the resting length of both muscles was usually set to the length at which maximum isometric force was produced. The resting tendon tension corresponding to this length was usually in the range of .25 to .45 kg wt. This

resting tension, initially set at the beginning of each experiment, was continuously monitored and maintained at its initial value throughout the experiment.

2.5 Data Acquisition

Two procedures were used in collecting the frequency response data. The first employed the classical frequency response technique which uses a series of discrete frequency input sinusoids spaced at intervals over the modulation frequency range of interest. Measurements of the amplitude and phase (relative to the input sinusoid) of the system's periodic output were made after start up transients had decayed. Figure 4 illustrates this scheme and Figure 7 is a sample of the data collected using this technique. In order to obtain sufficient data, it was necessary to collect 15 to 30 complete cycles of response at each frequency. This procedure proved to be prohibitively time consuming as it required, including a muscle rest period and apparatus adjustments, approximately ten minutes for each modulation frequency investigated. Thus, although the data proved reliable, this procedure was abandoned after one experiment.

The second data collection procedure that was used is illustrated in Figure 8. Instead of supplying discrete sinusoids as inputs to the stimulator, a sinusoid

FIGURE 7

Typical inertial load position (θ) response to the indicated input modulating sinusoid (V). The input is a discrete frequency sinusoid at frequency ω . The mode of stimulation is FM.

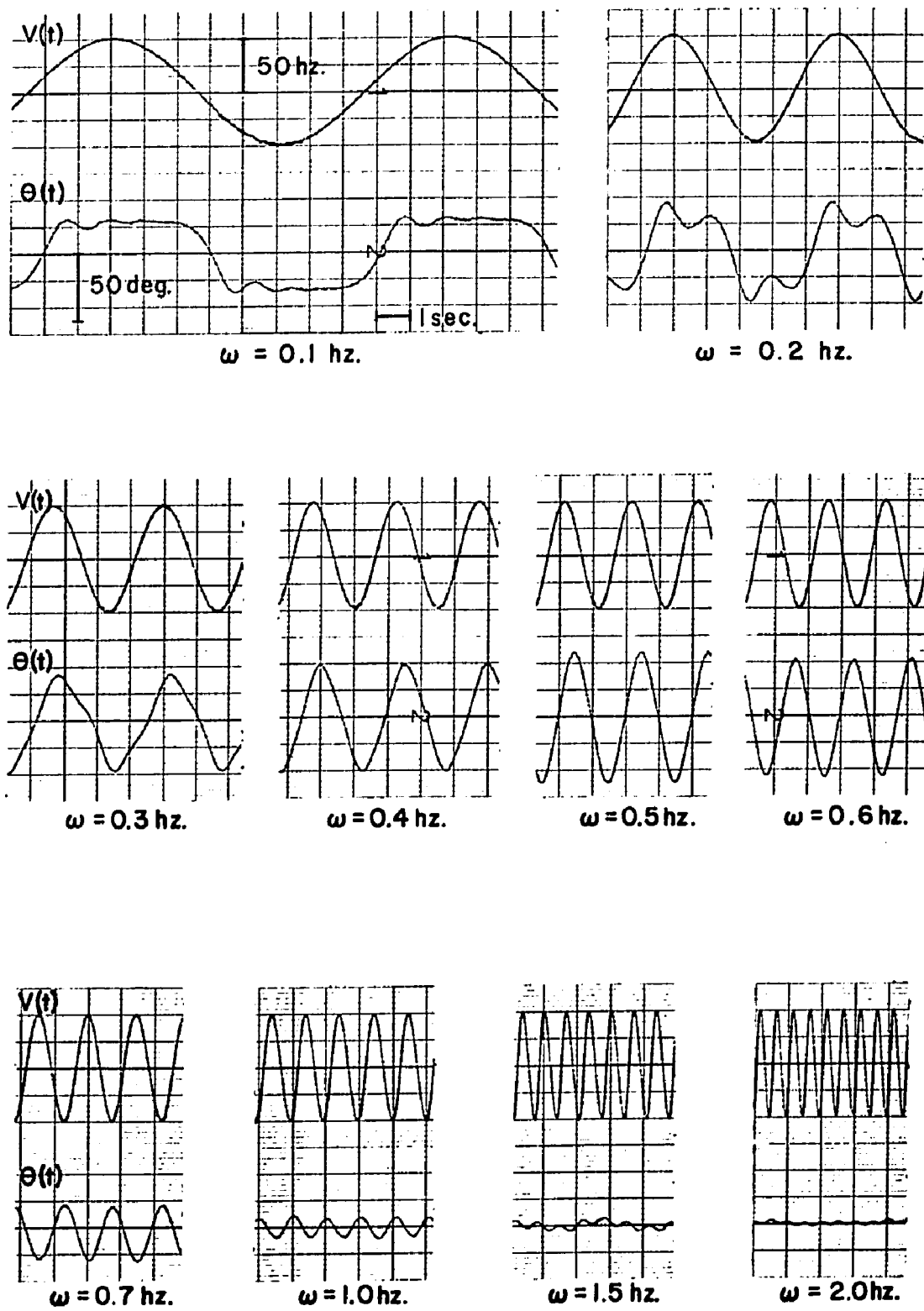


FIGURE 7

FIGURE 8

Typical data illustrating the swept frequency method of data collection. The input of the stimulator is a sinusoid (V) whose frequency is swept of the bandwidth of interest. The mode of stimulation in this case is FM.

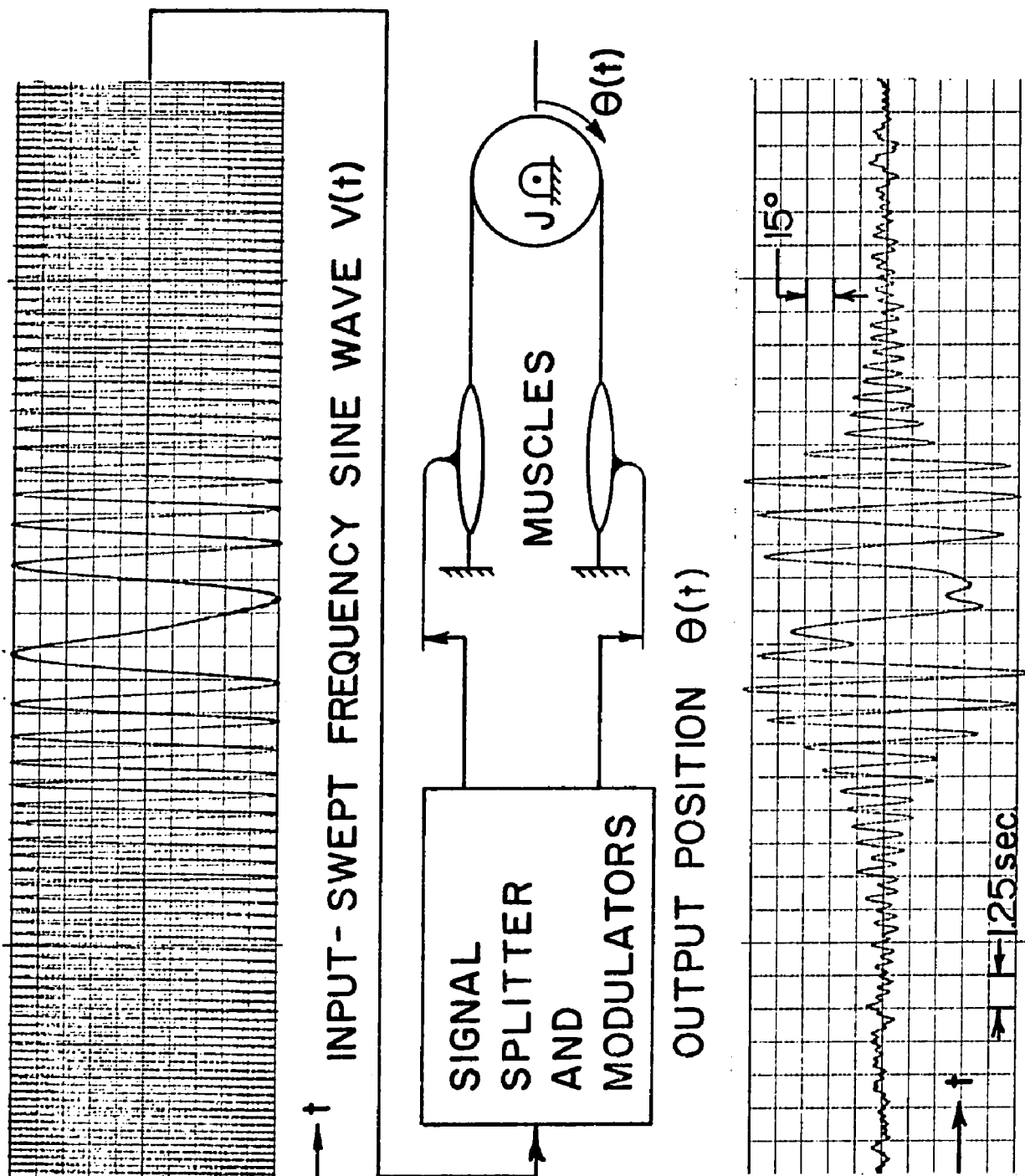


FIGURE 8

whose frequency was swept over the range of modulation frequencies of interest was employed. This input signal was generated by electronically controlling the frequency of the input modulating sine wave with a slowly varying triangular wave form obtained from a second oscillator (Hewlett Packard 3300). This procedure, which produces the same results as the classical method (50), required approximately 15 minutes per trial (including apparatus adjustments, a rest period of 6 to 7 minutes, and at least 7 uninterrupted minutes of data acquisition). However, final results at up to 128 discrete modulation frequencies could be extracted from these data. By using this method it was possible to obtain from one animal, high resolution, complete frequency response profiles of the muscles for at least 36 different combinations of stimulus modes, stimulus amplitudes, and inertial loads.

2.6 Data Analysis

The input modulating signal, $(V(t))$, and the inertial load position signals $(\theta(t))$, were recorded on magnetic tape with an FM instrumentation tape recorder (Hewlett Packard 3960). These records were later processed with a digital mini-computer programmed to perform the desired frequency response analysis. The results of this initial

processing were further processed, plotted and modelled using IBM 360/67 general purpose digital computer.

As illustrated in Figure 7 the motion of the inertial load is periodic at the input modulating frequency with some higher harmonic content (particularly at low modulation frequencies). The harmonics, indicative of system nonlinearities, precluded the use of frequency response techniques since, in the strict sense, frequency responses applies only to linear systems. Instead describing functions (47, also see Appendix III), which ignore higher harmonics, were used to describe the response of the muscle-load system.

The techniques used to extract the describing functions from the recorded data are summarized in Figure 9. The describing function, $H(j\omega)$, was obtained by dividing $S_{\theta V}(j\omega)$, the output-input cross-power spectral density by $S_{VV}(\omega)$, the input power spectral density. $S_{\theta V}(j\omega)$ and $S_{VV}(\omega)$ were obtained by Fourier transforming $R_{\theta V}(\tau)$ and $R_{VV}(\tau)$, the cross- and auto-correlograms of $\theta(t)$ and $V(t)$. This method, in the strict sense, is valid only for linear systems (51). However, if the higher harmonics are removed in the process of calculating $R_{\theta V}(\tau)$, the method may also be used to extract describing functions of nonlinear

FIGURE 9

Schematic diagram of the time series analysis technique used to extract the describing functions of the muscle-load system from the input (V) and output (θ) signals. The output-input ($R_{\theta V}$) and input (R_{VV}) correlograms were first computed from $V(t)$ and $\theta(t)$. R_{VV} and $R_{\theta V}$ were then Fourier transformed to obtain the input (S_{VV}) and output-input ($S_{\theta V}$) power spectral densities. Finally, the magnitude of the describing function, $H(j\omega)$, was obtained by dividing the magnitude of $S_{\theta V}$ by S_{VV} while the phase was given by the angle of $S_{\theta V}$ minus the angle of S_{VV} . Note that S_{VV} is real and therefore its phase angle is zero.

Source: Williams, W.J., Gesink, J.W., and Stern, M.M.
 Biological Transfer Function Extraction Using
 Swept Frequency and Correlation Techniques.
 Med. & Biol. Eng. 10:609-620, 1972.

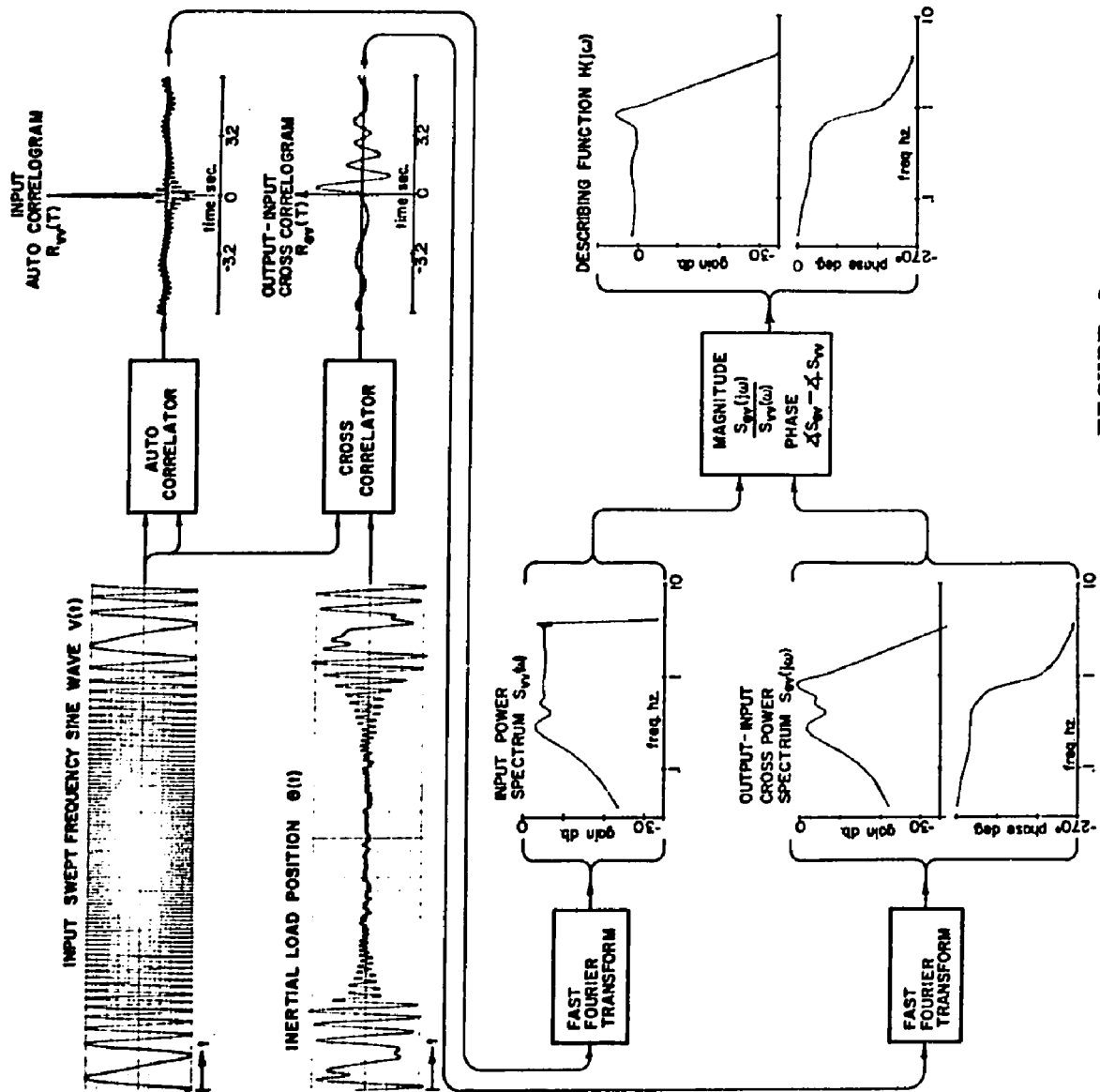


FIGURE 9

systems (50).

All steps indicated in Figure 9 were performed with a digital mini-computer (Hewlett Packard 2115A) which used standard time series analysis programs. The programs, using a Fast Fourier Transform (FFT) algorithm (52), calculated 256 point correlograms in real time and 128 point spectra off line. Appendix IV reviews the correlation and spectral analysis techniques that were used as well as the procedures of applying them to describing function extraction.

Although the higher harmonics were ignored in extracting the describing functions, an investigation of the degree of harmonic contamination in $\theta(t)$ was made. This was necessary because the describing function technique assumes a reasonably small higher harmonic content. Determination of the amount of harmonic contamination provided a measure of the validity of the describing function. Two measures of harmonic distortion, the coherence function ($\gamma^2(\omega)$) (53) and signal to noise ratio (S/N), were used in analyzing $\theta(t)$. Appendix VI discusses these measures at some length.

A portion of the data analysis dealt with modeling. The experimentally measured describing functions were fit with transfer functions models which were

variations of

$$M(s) \Big|_{s=j\omega} = \frac{K e^{-s\tau} \prod_{i=1}^k (s+z_i)}{\left[\prod_{i=1}^j (s+p_i) \right] \left[\prod_{i=1}^m (s^2 + 2\zeta_i \omega_{ni} s + \omega_{ni}^2) \right]}$$

where the model parameters are: z_i and p_i are real poles and zeros, ζ_i and ω_{ni} define pairs of complex poles, τ is a transport delay, K is a gain constant and s is the Laplace operator. In some cases models were fit to the magnitude portion of the describing function data while in others to both magnitude and phase. Fitting was accomplished with a computer program which, by perturbing the model's parameters, found the minimum of the integral of the square of the difference between the data and the model. The program used a subroutine, FMFP, available in the IBM Scientific Subroutine Package (54). Details of the fitting procedure may be found in Appendix V.

2.7 Data Presentation

The conventions established by Bode (55) for illustrating frequency response data will be used to present both the experimental describing functions, their coherence functions, and their models. Figures 10 and 11 are typical examples. The abscissa of both the

FIGURE 10

Experimental describing functions plotted according to Bode conventions. Individual curves represent describing functions obtained with the three indicated depths of frequency modulation. In all three cases the inertial load was 0.11.

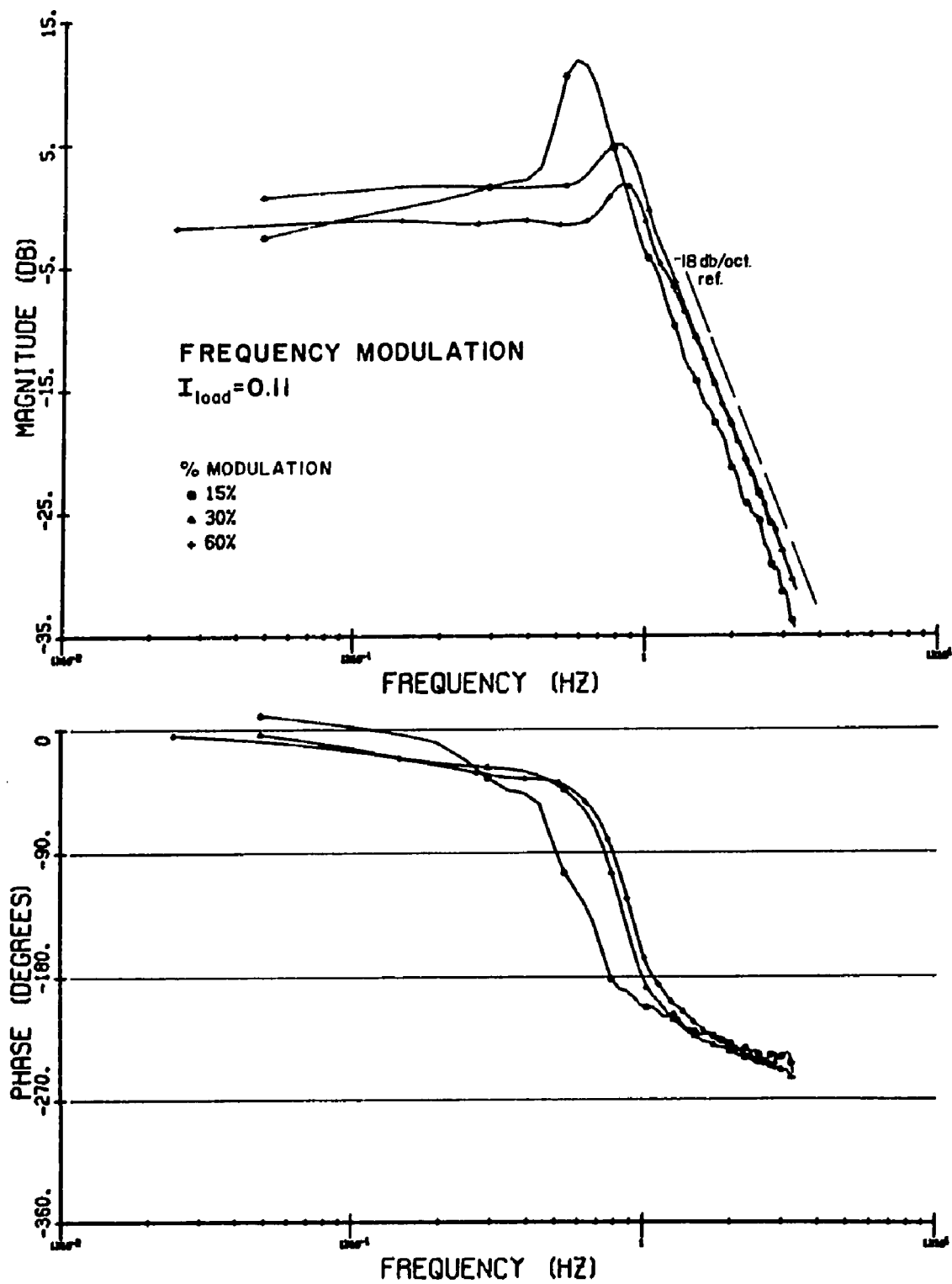


FIGURE 10

FIGURE 11

Bode plots of typical experimental describing functions and fits of these data that were obtained for the indicated model. The values of the model parameters whose units are in Hz are indicated by arrows placed at the proper abscissa locations. In these examples the model was fit to both the magnitude and phase of the describing function data. The model parameter K_0 is a normalizing factor and is equal to

$$\frac{\omega_n^2 p_1 p_2 p_3}{z_1 z_2} .$$

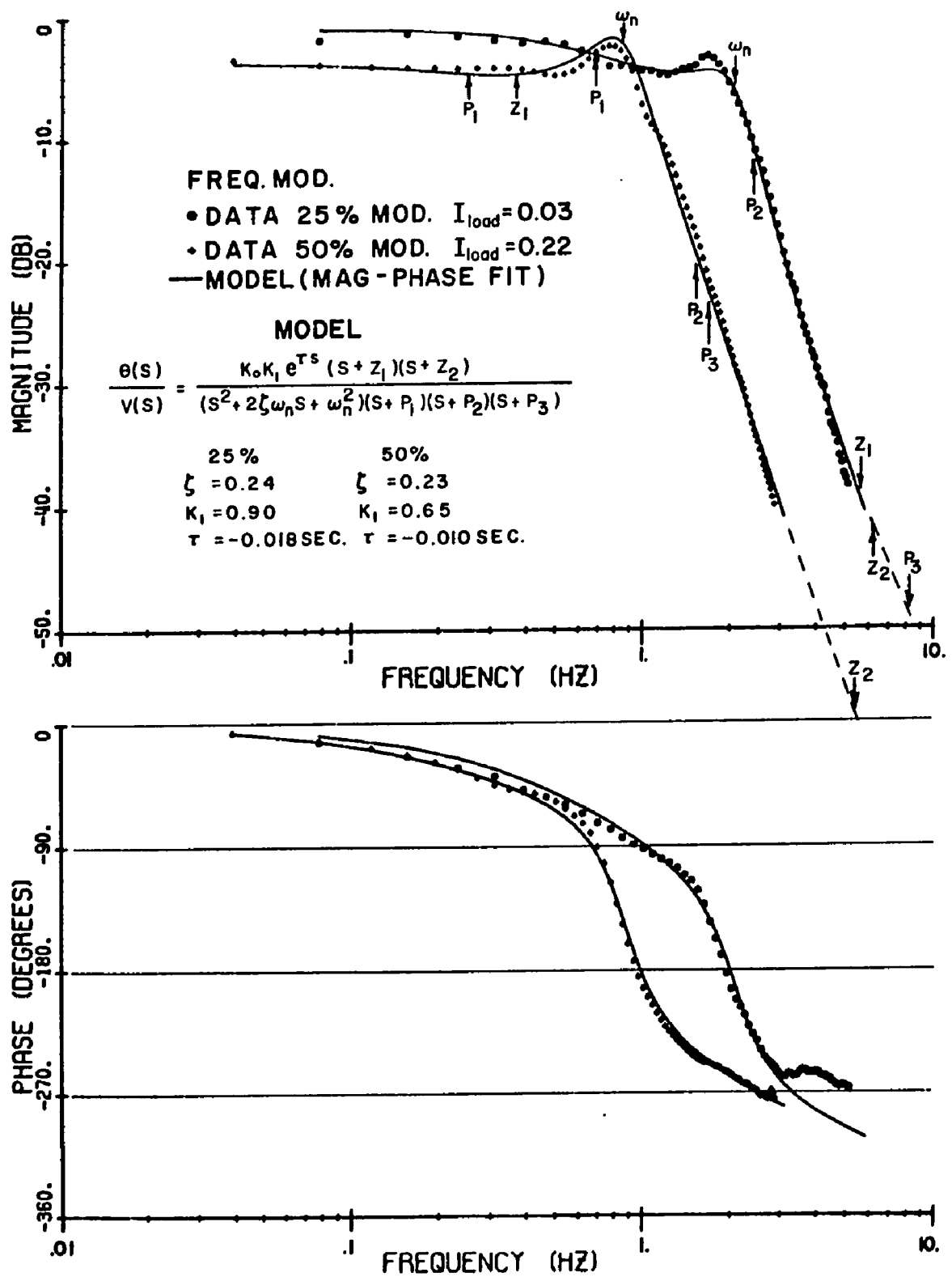


FIGURE 11

magnitude and phase plots is modulation frequency in Hz. The ordinate of the phase plot is Phase (ϕ) in degrees, ϕ being the angle of the fundamental component of load position (see Figure 4) relative to the modulating signal. The ordinate of the magnitude portion of the graphs is the ratio, in decibels (db), of the amplitude of the fundamental component of the load position (in degrees) to the amplitude of the modulating signal (in %). When data alone is presented (as in Figure 10), a symbol is placed around every fifth point with straight line segments joining each of the individual points. When data and modelling results are presented together (Figure 11), data is indicated by a point plot and the model by a solid line. As indicated on the graphs, some models were fit to magnitude data only, while others were fit to both magnitude and phase. The values of model parameters whose units are Hz (z , p , ω_n) are sometimes indicated by an arrow head placed at the proper abscissa location. When applicable, reference asymptotic slope lines are placed beside the high frequency "skirt" of the magnitude plots. All Bode plots were generated using CalComp in conjunction with the IBM 360/67 digital computer.

CHAPTER 3

RESULTS

3.1 Response to Modulation Frequency

Figures 12, 13, and 14 illustrate a complete set of thirty six experimental describing functions of the muscle-load system. These Figures show the effects of modulation frequency, depth of modulation, mode of stimulation and inertial load size on the response of the muscles.

3.1.1 Magnitude Response

Inspection of the magnitude portion of these data reveal a sensitivity to frequency which is characteristic of most actuators. All thirty-six trials exhibit an over (Figure 12) or under-damped (Figures 13 and 14) "low pass" filter type of frequency response. At low modulation frequencies the magnitude of the response is nearly independent of frequency. At these frequencies most of the contractile force of the muscle is used in producing a load motion which essentially follows the sinusoidal input signal with little amplitude attenuation. Above the corner frequency of approximately 1 Hz, the amplitude of the inertial load motion, in all cases, drops rapidly

FIGURE 12

Describing functions of the muscle-load system showing the relationship of the magnitude and phase to modulation frequency. In all twelve cases the inertial load was zero. Results at four different depths of modulation for each of the three modes of stimulation are illustrated. Note that the curves are constructed of straight line segments connecting successive data points and that symbols are placed around only every fifth point.

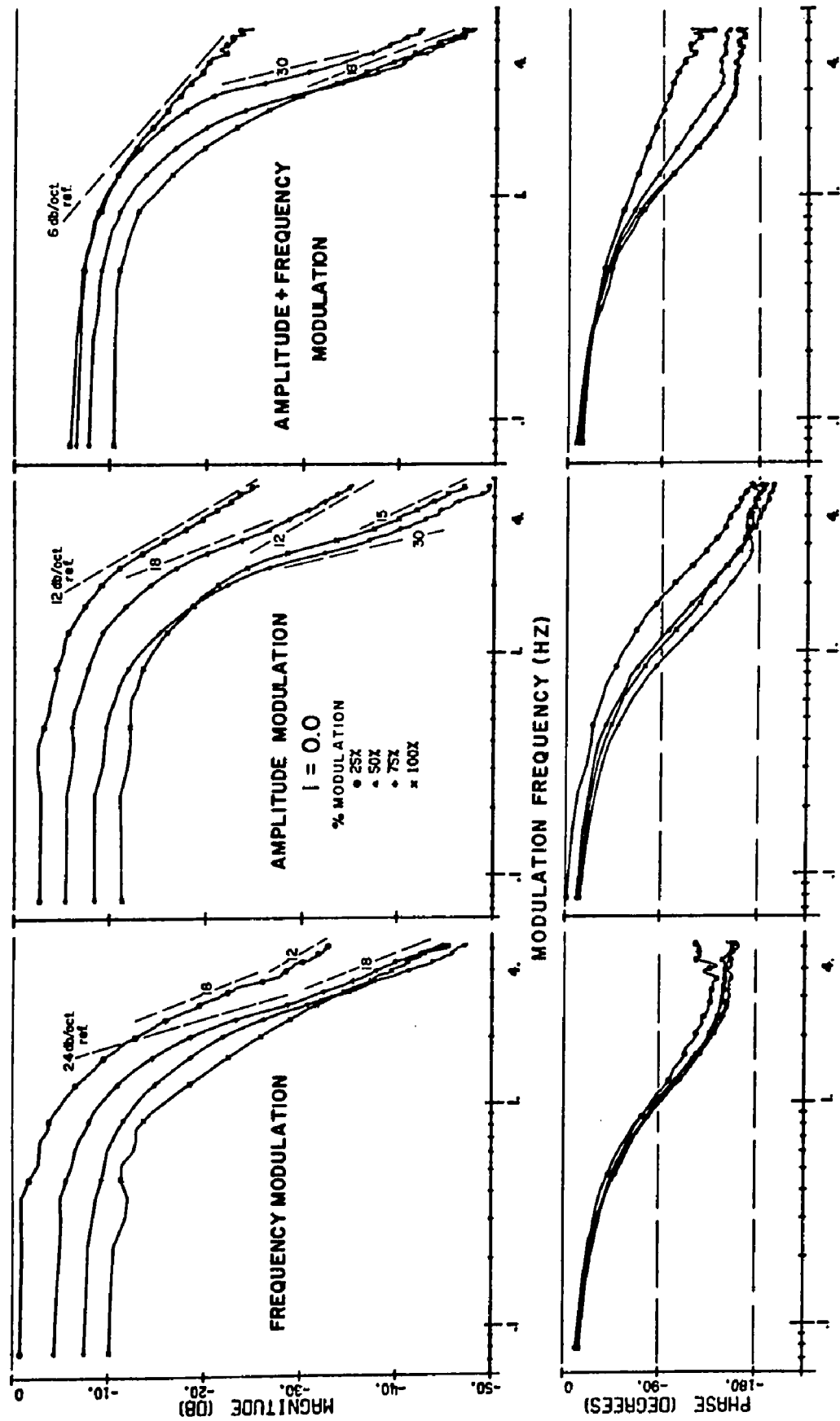


FIGURE 12

FIGURE 13

Describing functions of the muscle load system showing the magnitude and phase as a function of modulation frequency. The inertial load ratio for these data is 0.03. Results at four different depths of modulation for each of the three modes of stimulation are illustrated.

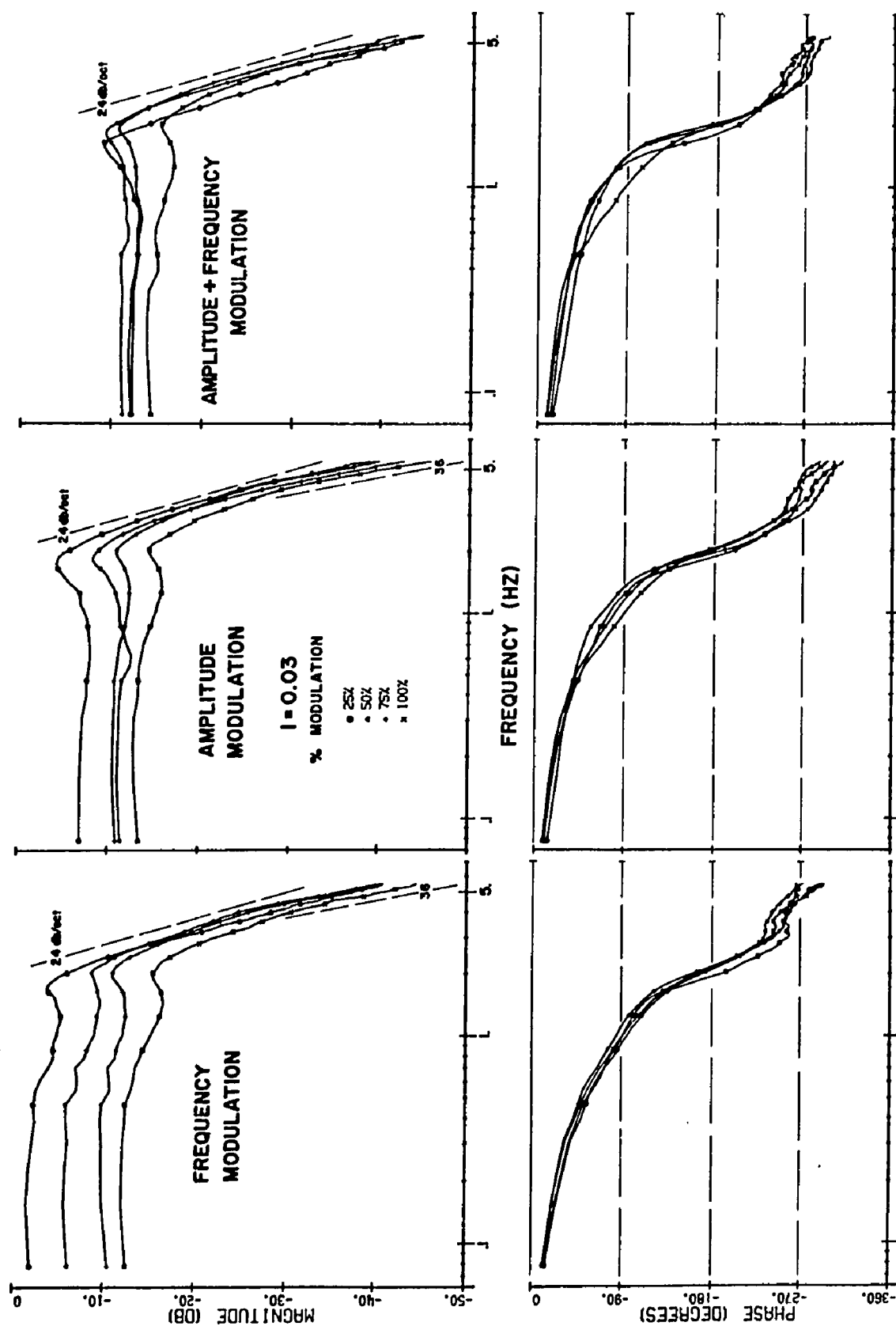


FIGURE 13

FIGURE 14

Describing functions of the muscle-load system showing the magnitude and phase as a function of modulation frequency. The inertial load ratio for these data is 0.21. Results at four different depths of modulation for each of the three modes of stimulation are illustrated.

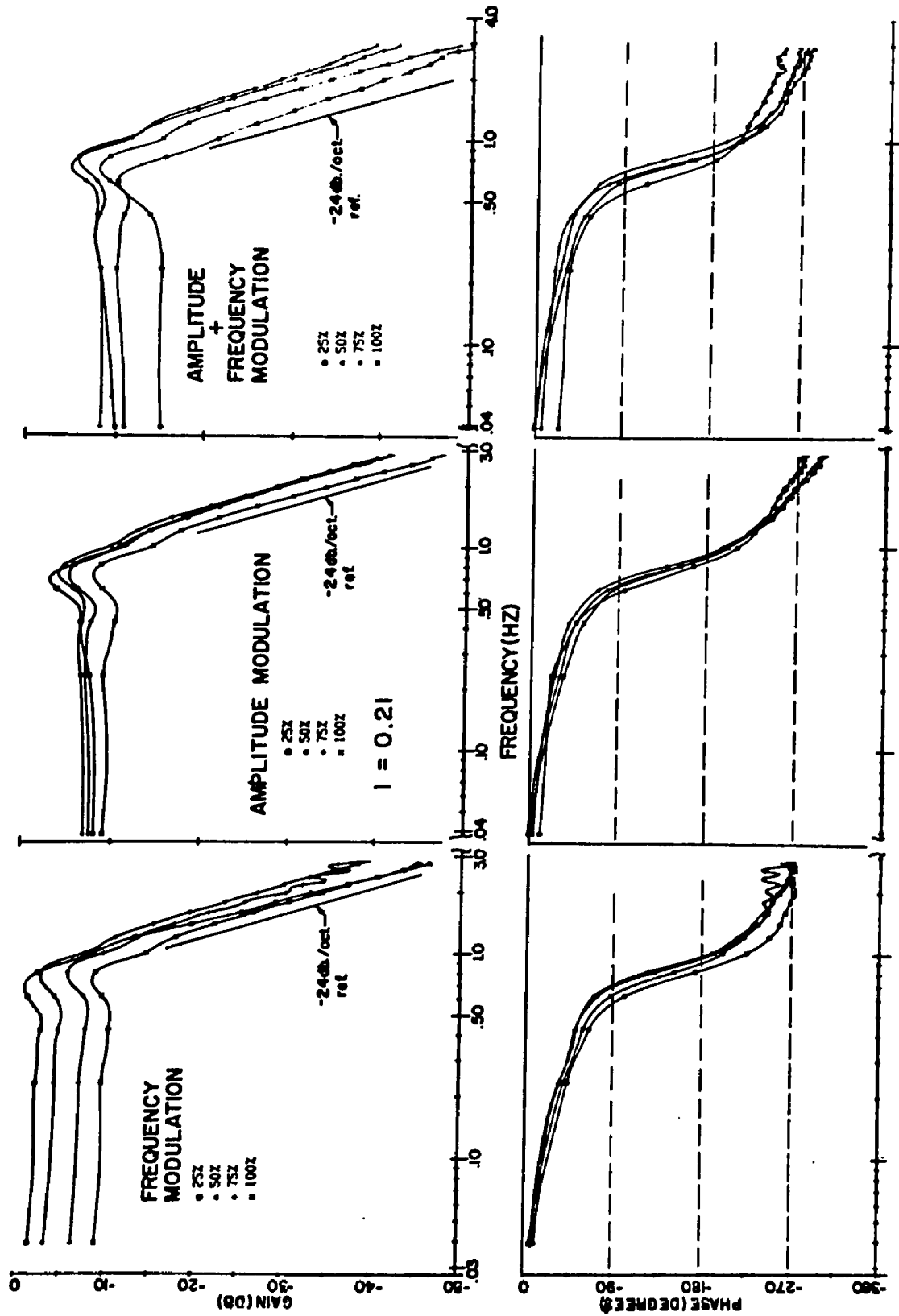


FIGURE 14

with increasing frequency. At these high frequencies the viscous properties of the muscle absorb much of the contractile force leaving only a minor portion to move the load.

Use of the asymptotic approximation technique of linear control theory (56) is helpful in further characterizing the frequency response of the muscle-load system. This technique fits the magnitude of experimental frequency response data with a series of straight line segments. The slopes of the segments are integer multiples of ± 6 decibels per octave (db/oct). The segments are related to the poles and zeros of a linear transfer function model and provide a means of estimating a linear model of the data. Subjecting the data in Figure 12 to this method requires straight line segments with slopes varying over the range from a -6db/oct (25% AM+FM) to -30db/oct (75% AM and 50% AM+FM). These fits imply that the magnitude response is similar to a linear system containing as many as 5 poles (5th order) or by as few as a single pole (1st order). Also, in some cases there is a change in slope from a greater to a lesser negative value (e.g. 25% FM and 25% AM) at modulation frequencies between 2 and 5 Hz. This change suggests that a model fitting these data might contain

zeros. Figure 15 illustrates the results of fitting a fourth order transfer function model containing a single zero to the magnitude portion of one of the data sets of Figure 12 (50% FM). This model, based on the asymptotic approximation, is seen to fit the magnitude portion of the data within 0.4%.

Using the asymptotic method on the data in Figures 13 and 14 suggests up to a sixth order (-36db/oct) model for the case of small loads (Figure 13) and a fourth order (-24db/oct) characteristic for those cases where a large inertial load was used (Figure 14). Rates of a -18db/oct were also evident in some cases where large inertia loads were used (Figure 10).

3.1.2 Phase Response

Inspection of the phase angles of the responses in Figures 12, 13, and 14 also reveal a characteristic typical of most actuators. The phase, as did the magnitude, exhibits a low pass filter type of response. The phase is always negative and decreases monotonically with increasing frequency. At both low and high frequencies the rate of decrease is rather gradual while near the corner frequency the phase changes rather rapidly. With zero inertial load (Figure 12) the phase lag, in most cases, appears to be approaching an

FIGURE 15

Comparison of modelling and experimental results. The indicated linear model was fit to only the magnitude portion of the describing function data. The average error per data point in the magnitude fit is 0.4%. The error in the phase is considerably larger with the model showing a larger lag than the data.

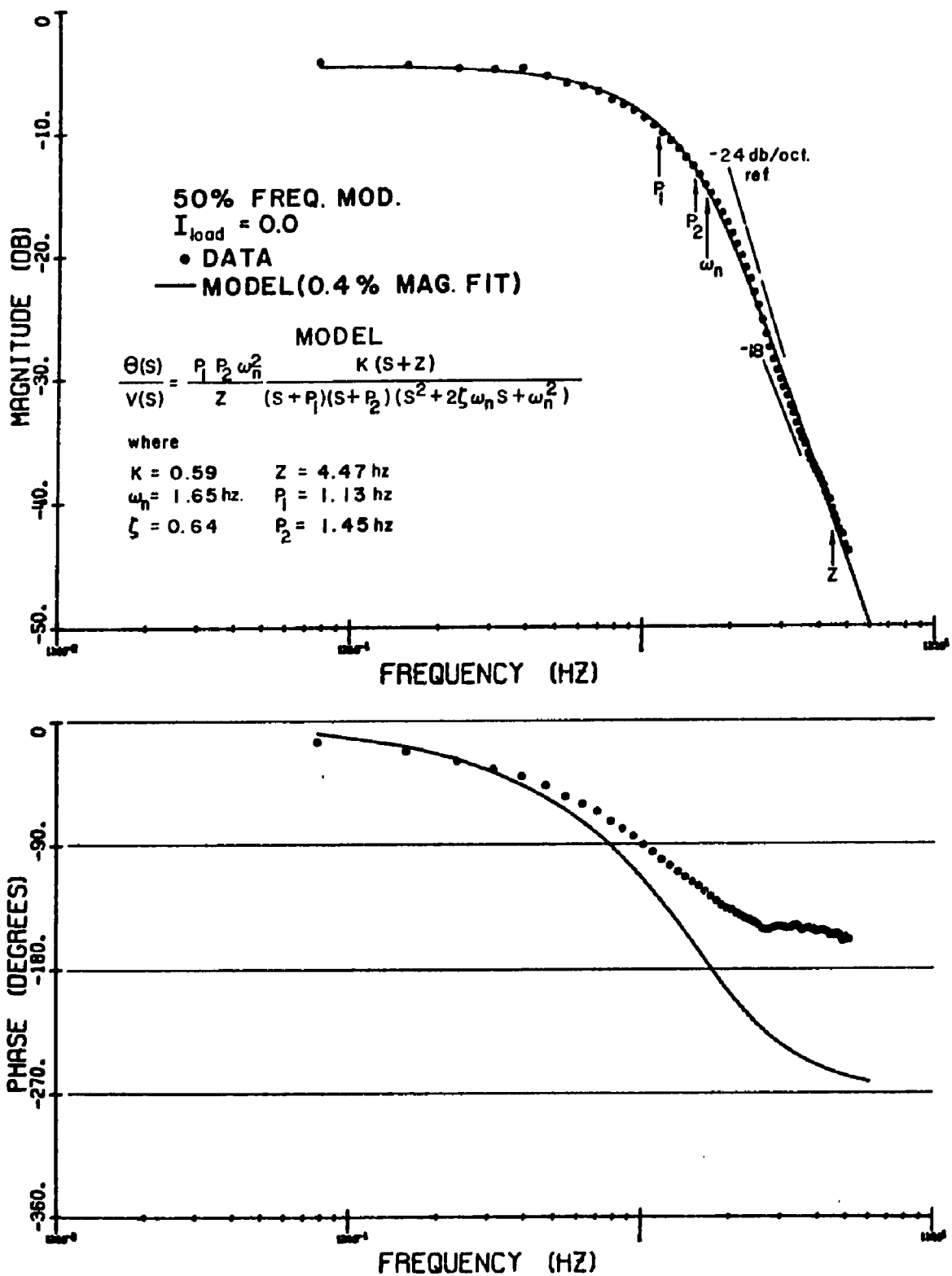


FIGURE 15

asymptotic value in the neighborhood of 180° while in the cases with finite load (Figures 13 and 14), the lag reaches values considerably greater than 180° .

The asymptotic methods of linear control theory, as applied to phase response, also were used to aid in characterizing the phase response. In a linear system, a ± 6 db/oct slope change in the magnitude characteristic corresponds to a $\pm 90^\circ$ phase angle change. This method, when used to examine the data in Figures 12, 13, and 14, revealed an apparent nonlinearity in the response of the muscles. In most cases the phase lag attained at high frequencies is far less than would be predicted by linear theory based on the slopes of the magnitude response. For example, consider the 50% FM case in Figure 12. The high frequency asymptote of approximately -18 db/oct implies a phase lag of 270° . The data however exhibits a phase lag which approaches a value of a little less than 180° . Figure 15 also shows the phase discrepancy. Here the model, based on the asymptotic approximation, fits the magnitude data to within 0.4%. However, at high frequencies, the linear model exhibits a phase lag considerably larger than that shown by the data. Those cases for which the inertial load was finite (Figures 13 and 14) also show

phase lags which are considerably less than predicted by an asymptotic fit of the magnitude data. The asymptotes predict phase lags in the neighborhood of 540° (Figure 13, -36db/oct) and 360° (Figure 14, -24db/oct) while the data displays maximum phase lags around 270° . For twenty-four cases of frequency modulation a quantitative measure of the difference between the experimental and the predicted phase lags was made using a two step model fitting procedure. First a linear transfer function model was fitted to the experimental phase data. This model was then modified by including only the magnitude of the number of poles which yielded a good fit of the magnitude portion of the data. It was found, in all cases, that the magnitude contribution of two poles was sufficient to produce close modelling of the magnitude data. These results then indicated, at least for the FM case, that the phase discrepancy was nearly equivalent to the phase angle contribution of two poles.

A similar quantitative measure was not made in AM and AM+FM cases. Aside from a few exceptions however, qualitative inspection of the data indicated an experimental phase lag considerably less than the magnitude asymptotes might predict.

3.2 Response to Inertial Loading

The sensitivity of the muscle system to changes in inertial load size is also evident from the data presented in Figures 12, 13, and 14. The changes that occur in the responses are characteristic of most mechanical actuators when they are subjected to changing inertial loads. The typical actuator responds to an increasing inertia load with a decrease in the amount of damping, an increase in the phase lag, and, if resonance occurs, with a decrease in the resonant frequency. All three of these factors appear in the data of Figures 12, 13, and 14. When the load is zero, (Figure 12) the response may be characterized as overdamped. Increasing the load to an intermediate value (Figure 13) produces a resonant peaking effect around 2 Hz and causes an increase in the asymptotic phase lag of about 90° . A further increase in load (Figure 14) has little additional effect on the phase lag but produces a greater amount of resonant peaking while the frequency of the resonance decreases from 2 Hz to about .8 Hz. Figure 16 clearly illustrates the effect. Here the results for two cases with different loads but identical modes and depths of stimulation show the changes in both resonant frequency and amount of

FIGURE 16

Models and experimental data showing the effect of inertial loading on the resonant frequency and the amount of resonant peaking. As the inertial load increases, the resonant frequency decreases and the amount of peaking increases. The indicated sixth order model was fit only to the magnitude portion of the data. K_0 and K_1 are gain constants.

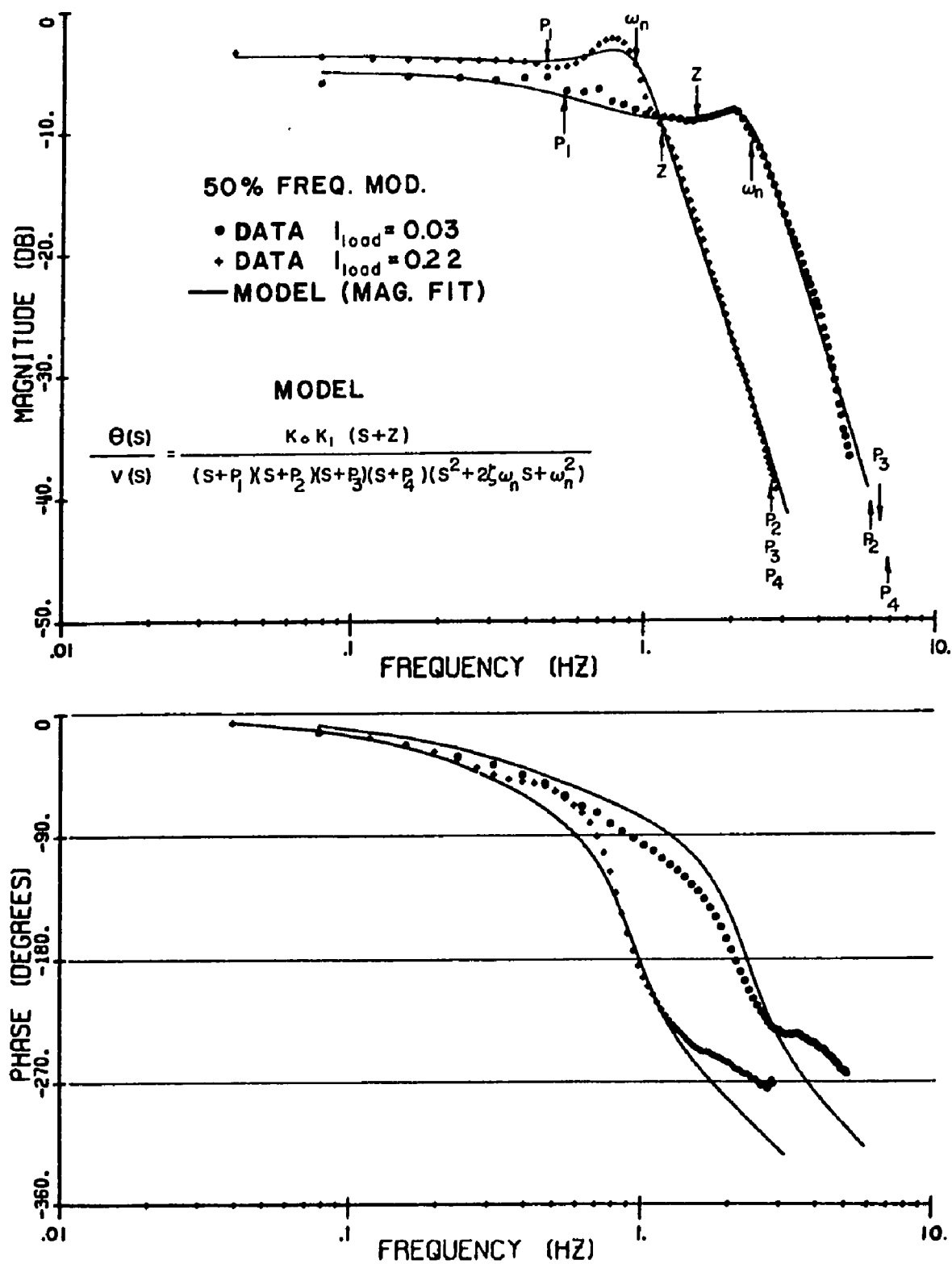


FIGURE 16

resonant peaking. Note that the ratio of these two frequencies ($2.0/0.8 = 2.5$) is approximately equal to the inverse of the square root of the inertial loads ($\sqrt{.21/.03} = 2.7$). This equivalence would be the expected result from a linear visco-elastic system where the resonant frequency is determined by the characteristic mass and elasticity of the system.

An additional minor effect can be observed in Figures 13 and 14 from inspecting the rate of "roll off" of the magnitude response at high frequencies. With intermediate loads (Figure 13) the rate starts at -24db/oct and tends toward -36db/oct with increasing frequency. In contrast, the large loads (Figure 14) produce a nearly constant rate (-24db/oct) independent of frequency. Figure 10 illustrates an additional large load case where the rate soon reaches a constant value of -18db/oct. This effect would also be expected of a linear system. Increasing the inertia load tends to reduce the corner frequency, masks the effect of high frequency poles, and dominates the rate of attenuation of the "skirt" of the magnitude response.

3.3 Response to Depth of Stimulus Modulation

3.3.1 Magnitude Response

Several effects of the depth of modulation on the describing functions of the muscle are also evident in Figures 12, 13, and 14. Of the various effects, the most prominent is the sensitivity of the response magnitude to modulation depth. In the cases illustrated in the Figures, four different depths of modulation (25%, 50%, 75%, 100%) were used and the magnitude response is different for each depth. The general trend of the effect on magnitude, particularly at low frequencies, is one of decreasing magnitude with increasing modulation depth, i.e., the increase in load motion decreases with each incremental increase in modulation depth. Figure 17 further illustrates the effect. Here the low frequency amplitude is plotted against depth of modulation. These curves, when interpreted in terms of control theory, reveal that the relationship between the input and output is the typical non-linear effect of saturation. In the cases where the mode of stimulation was FM, the saturation effect was very consistent from one load to the next throughout the low frequency range. This consistency is not as apparent in the cases of AM and AM+FM stimulation and some times occurred only in a limited frequency band near the resonant peak. In general, the major effect of modulation depth on

FIGURE 17

Transfer characteristic of muscles at low frequency showing relationship between amplitude of inertial load motion and amplitude of the modulating signal. Note that above about 40% modulation the output saturates.

▷, FM, I load=0.012; ◇ FM, I = 0.0; ⊙, FM, I = 0.03;
 Δ, FM, I = 0.21; □, FM, I = 0.0; ⊖, AM, I = 0.0;
 ⊖, AM+FM, I = 0.0

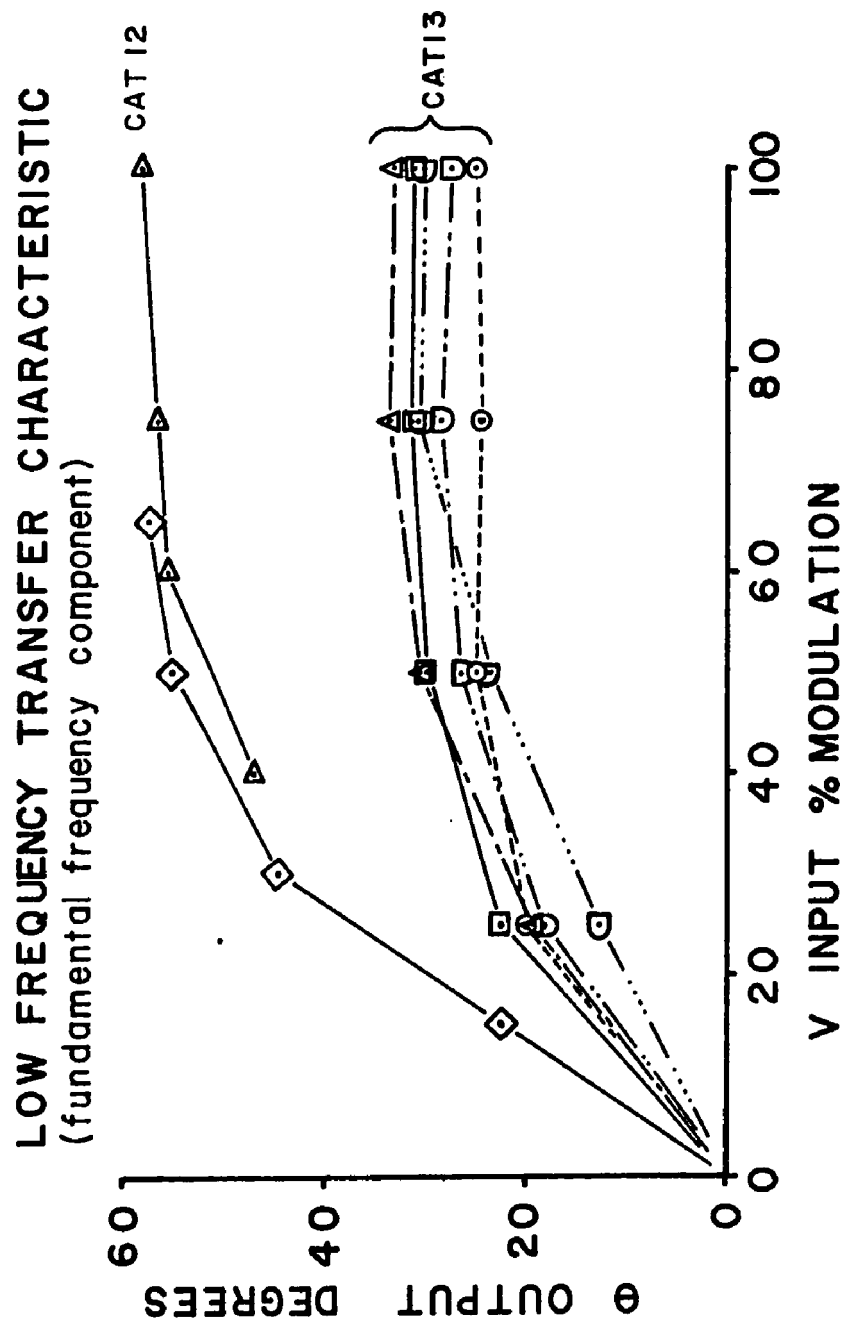


FIGURE 17

magnitude of the muscle response may be characterized as saturation.

3.3.2 Phase Response

Figures 12 through 14 show that the effect of modulation depth on the phase response of the muscle-load system was, in general, small. In most cases the phase curves cross and frequently are superimposed. Only in the cases of zero inertial load (Figure 12), where the modes were AM and AM+FM, is there significant separation between the phase response for different amplitudes of modulation.

3.3.3 Resonance Response

There are two additional minor effects of depth of modulation which warrant some attention. Both effects are associated with the resonant peaking phenomena and occurred to a significant degree only at reduced depths of modulation. At reduced depths, the height of resonant peaking is usually larger than that which occurs in the other cases while the resonant frequency is lower. Both of these effects are apparent in Figures 13 and 14 when the response at 25% modulation is compared with the other three responses. Figure 10 clearly illustrates the two effects. When the depth of modulation

is 15% the height of the resonant peak is better than twice that of the 30% and 60% cases. Also, the frequency at which the peak occurs is approximately .6 Hz while the resonant frequency of the other two cases is near .9 Hz.

3.4 Response to Mode of Stimulation

Examination of the results in Figures 12, 13, and 14 for differences in response due to the three modes of stimulation (AM, FM, AM+FM) reveals some minor effects. The existence of major changes however is decidedly lacking. The frequency responses for all modes at a given inertial load are, in all their general characteristics, quite similar in both magnitude and phase. All responses display low pass filter characteristics with approximately the same corner frequency. Further, the phase responses at high frequencies tend to approach the same values, the response magnitude at low frequency are all clustered in the same range, and the attenuation rate of the skirts of the magnitude response are also very similar from one mode to the next at a given inertial load. In those cases where resonant peaking occurs, the amount and frequency of the peaking depends primarily on inertial load size

and shows little dependency on the stimulation mode. Figure 18 serves to further emphasize the similarity between the responses to the different modes. This Figure shows a direct comparison of responses to the different modes at two different levels of modulation. While there are some differences in the cases at 75% modulation, the 100% cases illustrate a high degree of similarity.

3.5 Harmonic Analysis

Figures 19, 20, and 21 illustrate the results of the harmonic analysis performed on the data sets presented in Figures 12-14. Figures 19 and 20 show the coherence functions (γ^2) while Figure 21 displays the signal to noise (S/N) ratio. Both of these functions measure the amount of the input modulating signal that appears in the inertial load position, or, in other words, gives some measure of the degree of linearity of the muscle load system (see Appendix IV).

3.5.1 Coherence Function Measurements

Inspection of the coherence functions shows that they generally hover close to zero decibels at modulation frequencies below the describing function corner frequency. With the exception of the 25% AM case in

FIGURE 18

Comparison of the responses of the muscle-load system obtained with the three different modes of stimulation. Each Bode plot illustrates three experimental describing functions obtained with the FM, AM, and AM+FM modes of stimulation. In one case the depth of modulation was 75% while in the other it was 100%. In both cases the inertial load was 0.012. Note that at either depth of modulation the data, both magnitude and phase, are frequently indistinguishable from each other.

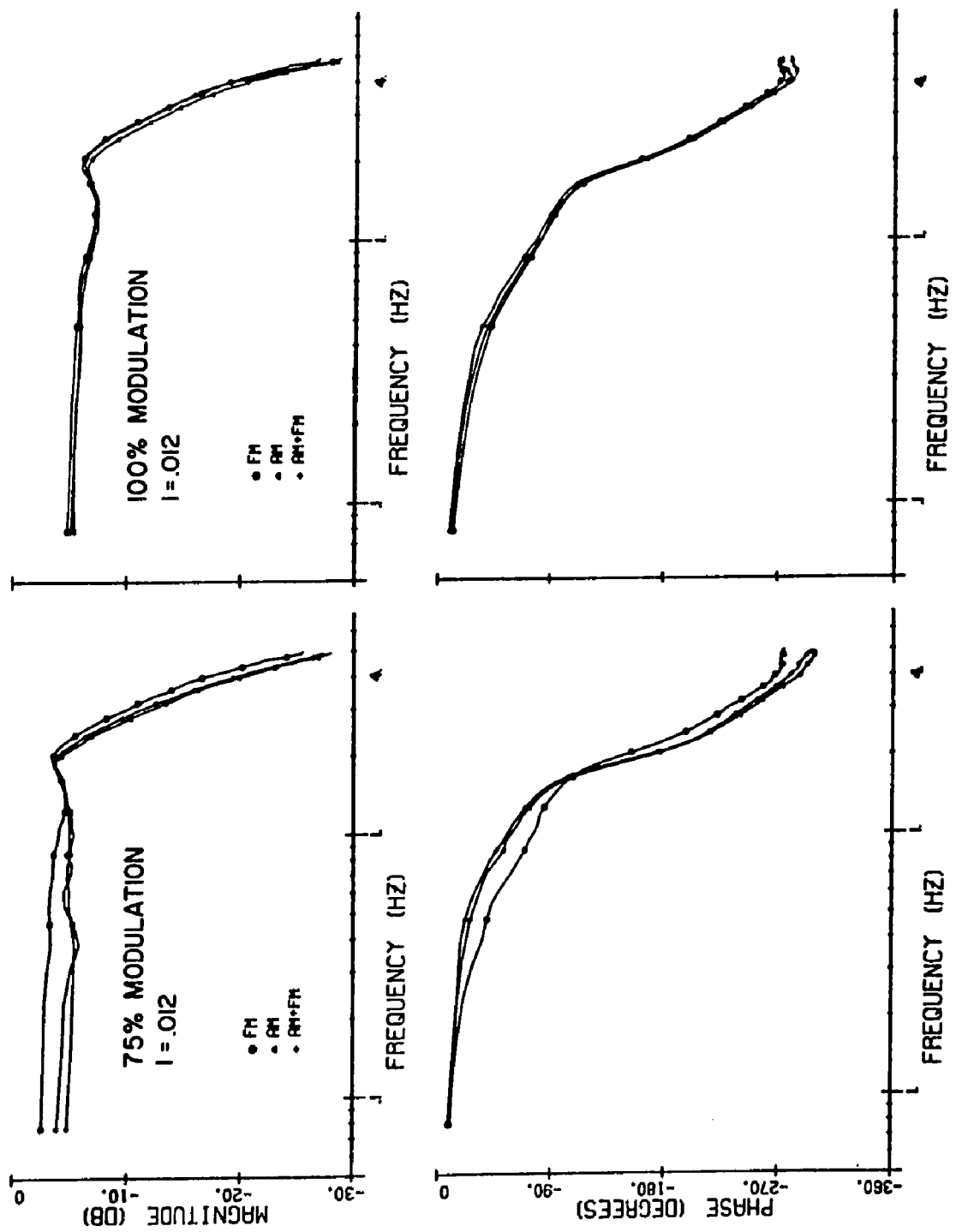


FIGURE 18

FIGURE 19

Coherence functions corresponding to the describing function data presented in Figures 12 and 13. The upper set of data shows coherence functions for the case where the inertial load was 0.0; the lower for the case where the load was 0.03. Each set contains four coherence functions corresponding to the four depths of modulation. The mode of stimulation is indicated on each set. Note that for any given stimulation mode, the coherence functions at different depths of modulation are frequently indistinguishable, particularly in the .06 to 3.0 Hz range. Note also that at frequencies below the corner frequency of the describing functions, the coherence functions hover close to 0 db.

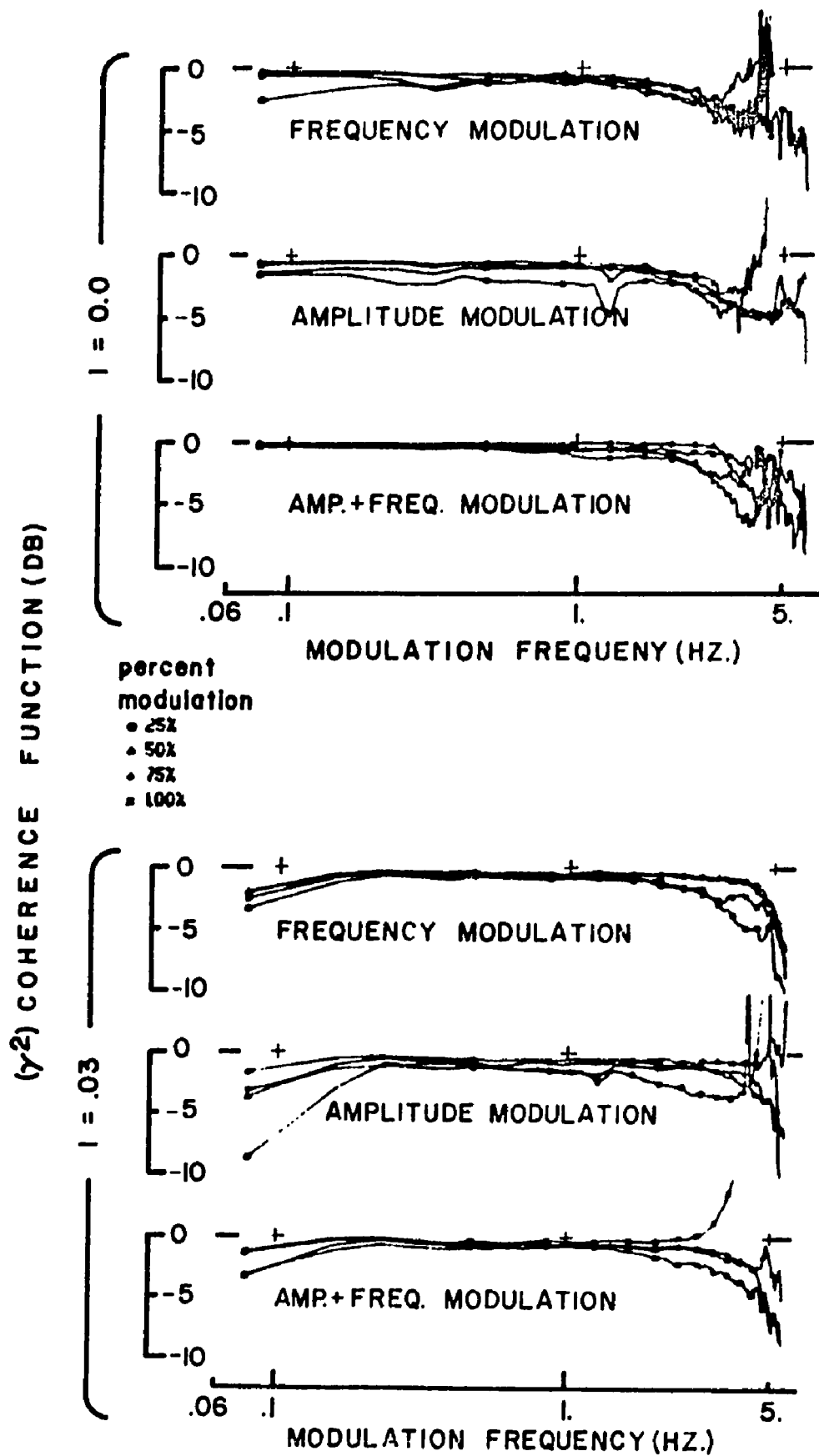


FIGURE 19

FIGURE 20

Coherence functions of the experimental data presented in Figure 14. Each set shows the results for a different mode of stimulation for each of four depths of modulation. The inertial load was 0.22. At frequencies below 1 Hz (the corner frequency of the describing functions), the coherence functions are all near zero db and show little dependance on depth of modulation. Above 1 Hz the coherence functions deviate noticeably from 0 db indicating a significant amount of harmonic distortion.

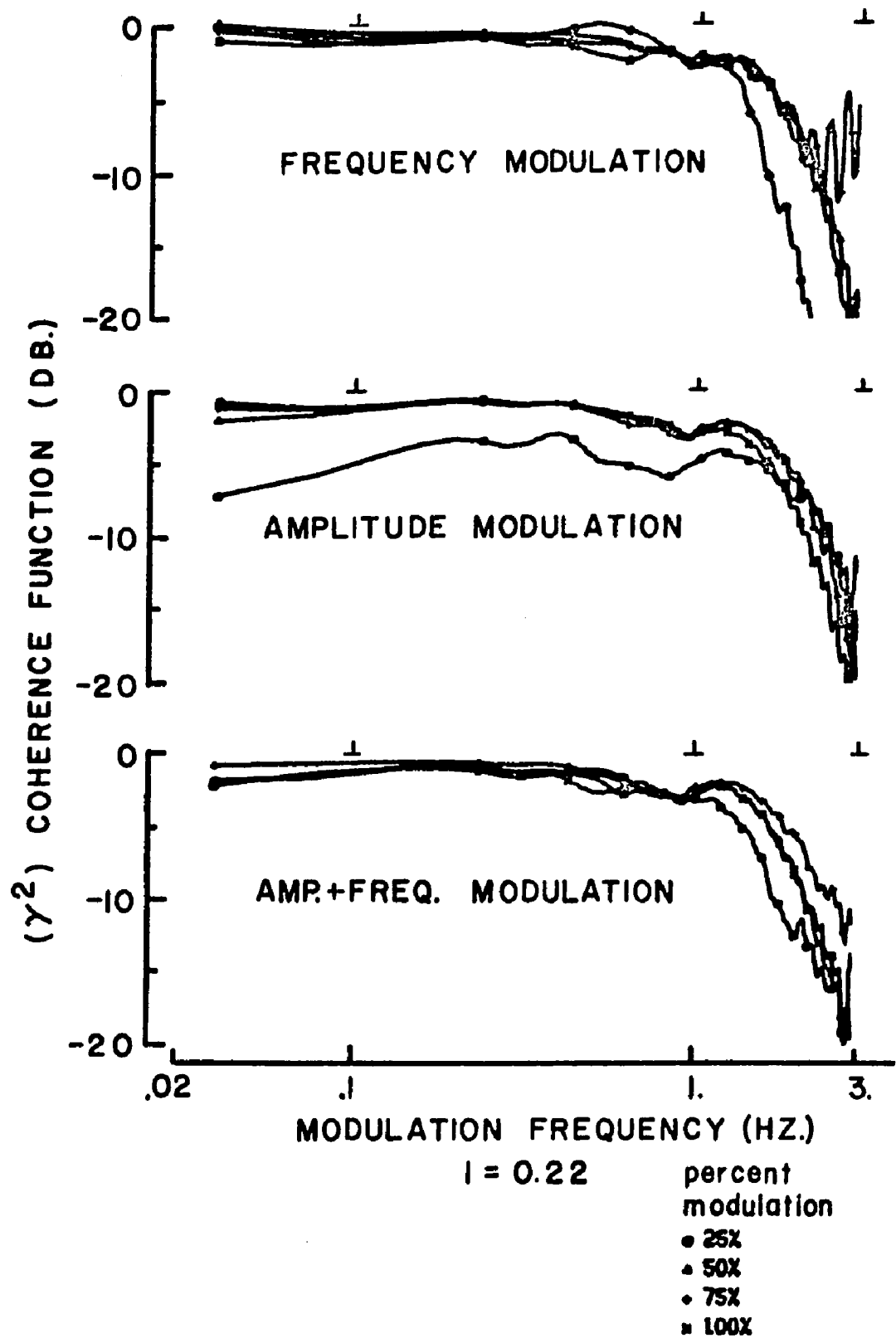


FIGURE 20

FIGURE 21

Total signal power to noise power (S/N) in the inertial load signal at all modulation frequencies. These data correspond to the results presented in Figures 12, 13, and 14. These measurements show the effects of mode of stimulation (FM, AM, and AM+FM), percent modulation, and inertial load (I) on the S/N ratio. A complete description of the S/N ratio measurement procedure may be found in Appendix V.

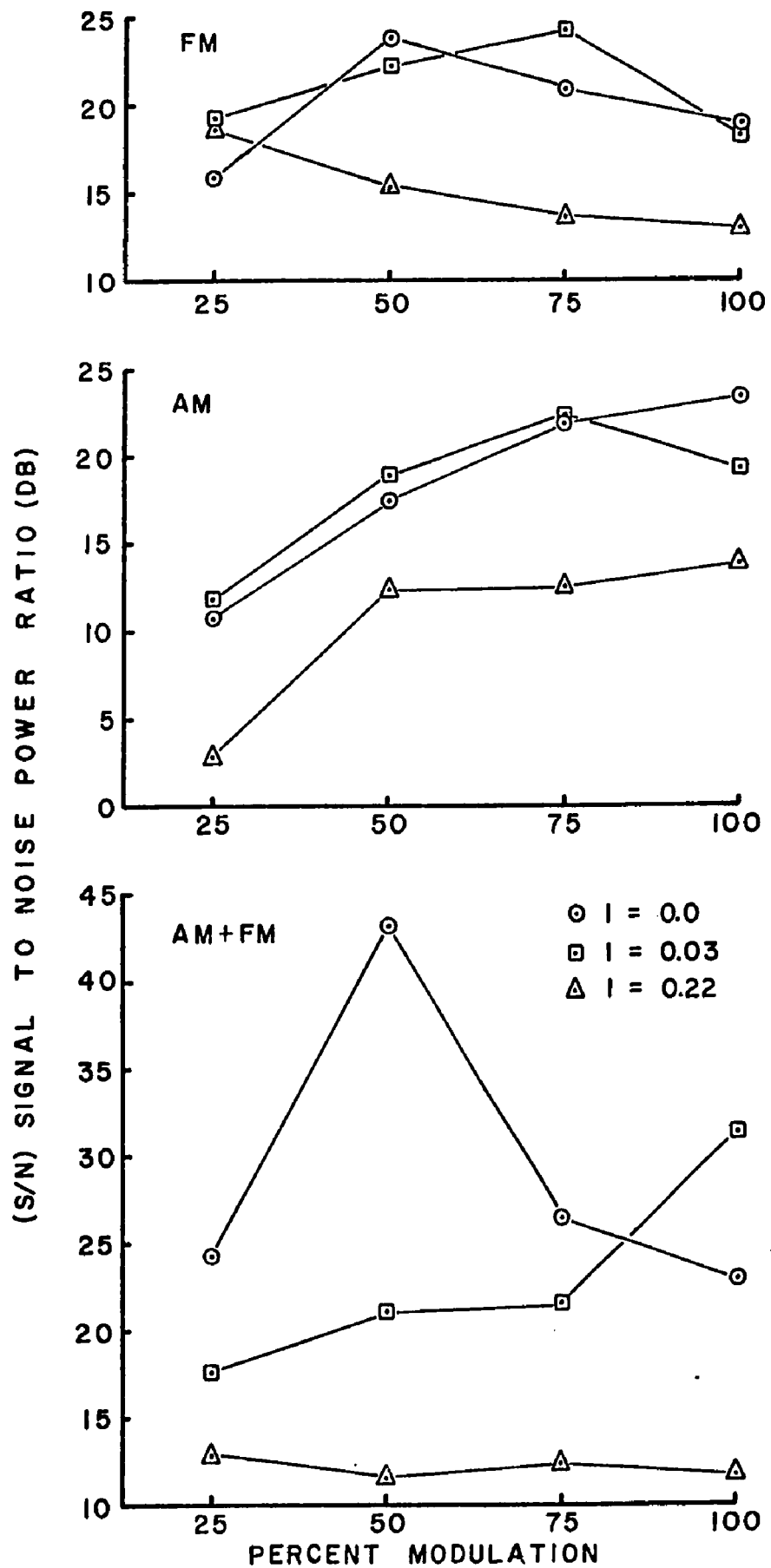


FIGURE 21

Figure 21, the coherence functions of all 36 cases are frequently indistinguishable from each other below the corner frequency. Also, in all thirty-six cases, the coherence functions depart significantly from 0 db at frequencies above the corner frequencies. This departure of the coherence function from 0 db is most prominent in the large inertial load cases (Figure 20) and appears to occur almost independently of stimulation mode or depth of modulation. Lack of coherence at high frequencies in both the zero and intermediate load cases (Figure 19) also appears to occur with no obvious pattern of heavy dependence on stimulation mode or depth of modulation.

These results generally imply that the muscle quite faithfully transforms the pulse coded sinusoidal shape of the modulation signals into inertial load positions at frequencies below the describing function corner frequencies and does so almost independently of load, mode of stimulation or depth of modulation. Above the corner frequency, where the load amplitude is rapidly decreasing, the muscle apparently introduces noticeable amounts of noise and harmonic contamination into the load's position. Here the amount of contamination seems to be primarily dependent on load size while mode of

stimulation and depth of modulation influence it only in a minor way.

3.5.2 Signal to Noise Ratio Measurements

Figure 21 illustrates the effects of load size, depth of modulation, and mode of stimulation on the S/N ratio of the response of the muscles. Although there are some local trends in these data, a general pattern relating the three parameters to the S/N ratio is not evident. Of the local trends that do exist, one concerns the inertial load size. Large loads generally produced the poorest S/N performance. However, performance didn't show a progressive improvement with decreasing load size. In the AM and FM cases the S/N ratios for zero and intermediate loads are quite similar at a given modulation depth. In the AM+FM cases however, the ratios for the zero and intermediate loads are considerably different at a given modulation depth.

An additional trend is evident in the AM cases of Figure 21. Here a general improvement in S/N ratio with increasing modulation depth at all loads is evident. This trend is lacking in both the FM and AM+FM cases where the ratios both increase and decrease with

changing modulation depth. Note also that with the exception of three cases, the S/N ratios for all thirty-six cases grouped in the band between 11 and 27 db and that the large load cases tend to dominate the lower portion of this band. These factors substantiate, in part, the findings of coherence measurements; the noise introduced by the muscles is relatively independent of mode of stimulation and depth of modulation but it shows noticable dependence on inertial load size.

CHAPTER 4

DISCUSSION

4.1 General Remarks

The results suggest that there are several types of transformations that muscle imposes on neural signals. In many ways the results are similar to what might have been expected had the study been performed on a linear mechanical actuator. The basic form of the frequency response of the muscles was similar to that of a low pass filter. Mechanical actuators, limited at high frequency by viscous and inertial effects also demonstrate a low pass filter frequency response. The muscles demonstrate an insensitivity to the mode of neural coding. The filtering action of the low pass filter would tend to smooth the high frequency fluctuations in the pulse coded input. Only the low frequency portions of the input signal which code the modulating signal would pass through the filter. Changing the inertial load on the muscles produces changes in the frequency response which are also quite similar to the changes that would occur in the response of a linear mechanical actuator. Increasing the inertial loading produces a resonant peaking, reduces the resonant frequency. The inertial effect dominates the

rate of attenuation of the skirt of the magnitude response, and significantly increases the amount of phase lag. The saturation in the output amplitude that occurs with increasing depth of modulation is also somewhat reminiscent of technological actuators. All real linear actuators are only linear over a finite range. Outside of this range they typically show an output-input saturation similar to that which was observed in the muscles.

While the response of the muscles and of mechanical actuators is similar in many respects, the muscle demonstrates some nonlinearities which are not typically found in actuators. The most noticeable of these is the relatively large inconsistency between the magnitude and the phase of the frequency response of the muscle. The muscles, together with the load, demonstrate a nonminimum phase characteristic¹ with the phase lag being considerably less than the magnitude might predict. Additional nonlinear characteristics were evident in the resonant peaking and in the way in which the peaking depended on depth of modulation. The resonant peaks

¹ For minimum phase systems the amplitude characteristic is completely determined when the phase characteristic is prescribed, and vice versa (57).

were always sharper than those found in linear systems. Further, the frequency at which the resonance occurred as well as the amount of damping associated with the response appeared to increase with depth of modulation.

In further examining the results of this study, it will be useful to consider them in terms of the analytically predicted response of various models of muscle found in the literature. Since these models consolidate the results of a broad spectrum of muscle studies, they provide a convenient reference point for evaluating the results of this study. Three models of muscle will be considered. All are based on the visco-elastic properties of muscle. Two of the models are linearized while the third attempts to incorporate both nonlinear and stimulation level dependent characteristics.

4.2 Muscle Model Development

The models of muscle to be presented in this section are developed using viscous and elastic components coupled in a manner which incorporates many of the known mechanical properties of muscle. Differential equations describing the performance of the models in terms of frequency response characteristics are then

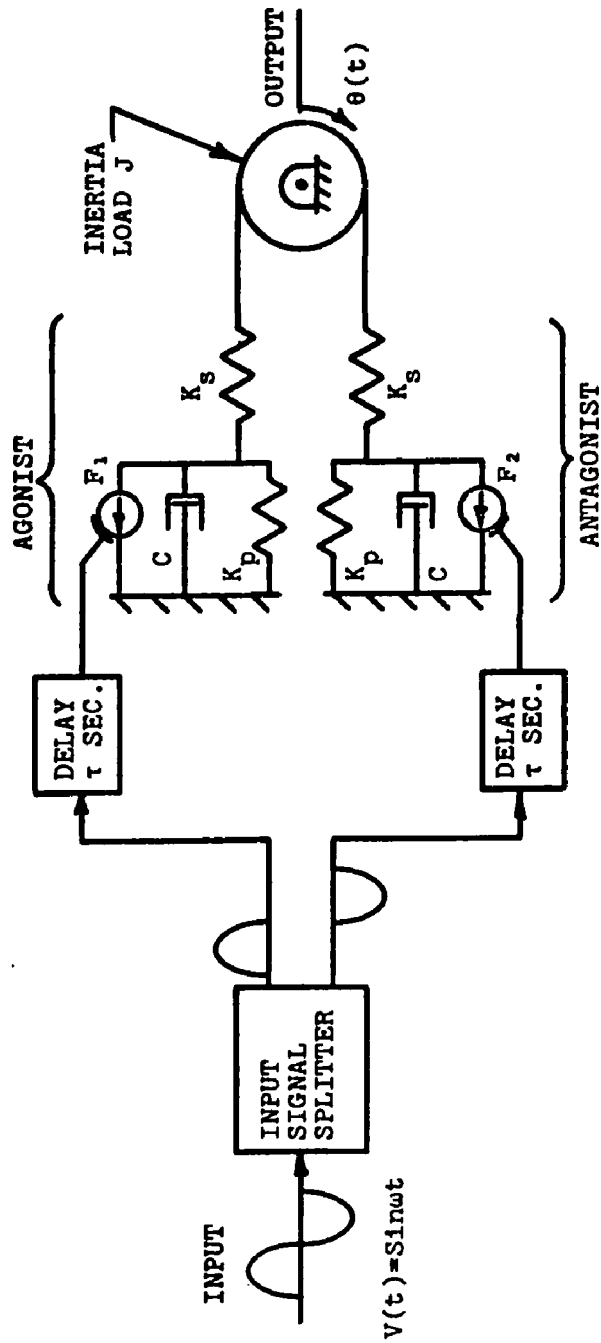
developed. Numerous simplifying assumptions are made. For this reason the models are in no way exact. They do, however, provide a framework within which the results of this study may be evaluated.

4.2.1 The Single Motor Unit Model

Perhaps the simplest model of muscle is the lumped parameter single motor unit model of muscle that was referred to in the introduction. Figure 22 illustrates a scheme wherein this model is incorporated into a configuration similar to that used in this study. Two motor unit models are coupled in an antagonistic manner to a common inertial load. As a first approximation, insight into the performance of the muscle-load system may be gained by examining the analytical transfer function obtained when this configuration is linearized. Linearization is achieved by making the coefficients of the three visco-elastic components constant. A further simplification is made by assuming that the force response of the CE is linearly related to the modulating signal amplitude. The analytical transfer function obtained with these assumptions is expressed in general form and in terms of the models parameters in Figure 22. Figure 23 illustrates a Bode plot of the transfer

FIGURE 22

Linearized lumped parameter model of muscle incorporated into a configuration similar to that used in the study. τ is the conduction and activation delay, F_1 and F_2 are the contractile elements, C is the viscous component, K_p and K_s are the parallel and series elastic components, J is the inertial load, $V(t)$ is the modulating signal, $\theta(t)$ is the inertial load position, and s is the Laplace operator. The equations show the transfer function for the model both in terms of the parameters and in general form. When J is large the transfer function is third order but reduces to first order when J is small.



LINEARIZED MODEL OF MUSCLE-LOAD SYSTEM

MODEL EQUATIONS

$$\frac{\theta(s)}{V(s)} = \frac{K_0 e^{-st}}{JCs^3 + J(K_s + K_p)s^2 + 2K_s Cs + 2K_s(K_s + K_p)}$$

WHEN J IS LARGE:

$$\frac{\theta(s)}{V(s)} = \frac{K_1 e^{-st}}{(s+a)(s^2 + 2\zeta\omega_n s + \omega_n^2)}$$

WHEN $J=0$:

$$\frac{\theta(s)}{V(s)} = \frac{K_2 e^{-st}}{s+a}$$

FIGURE 22

FIGURE 23

Bode plots of the linearized muscle-load model. Plots with four different sets of parameters ($J=0.0$, $J=$ a large value, $\tau=0.0$, and $\tau=.015$ sec.) are shown. When $J=0.0$, the magnitude asymptotes at -6db/oct and the phase at -90° (when $\tau=0.$). When $J=$ a large value the magnitude asymptotes at -18db/oct and the phase at -270° (when $\tau=0.$). When τ is finite the phase continues to increase with modulation frequency.

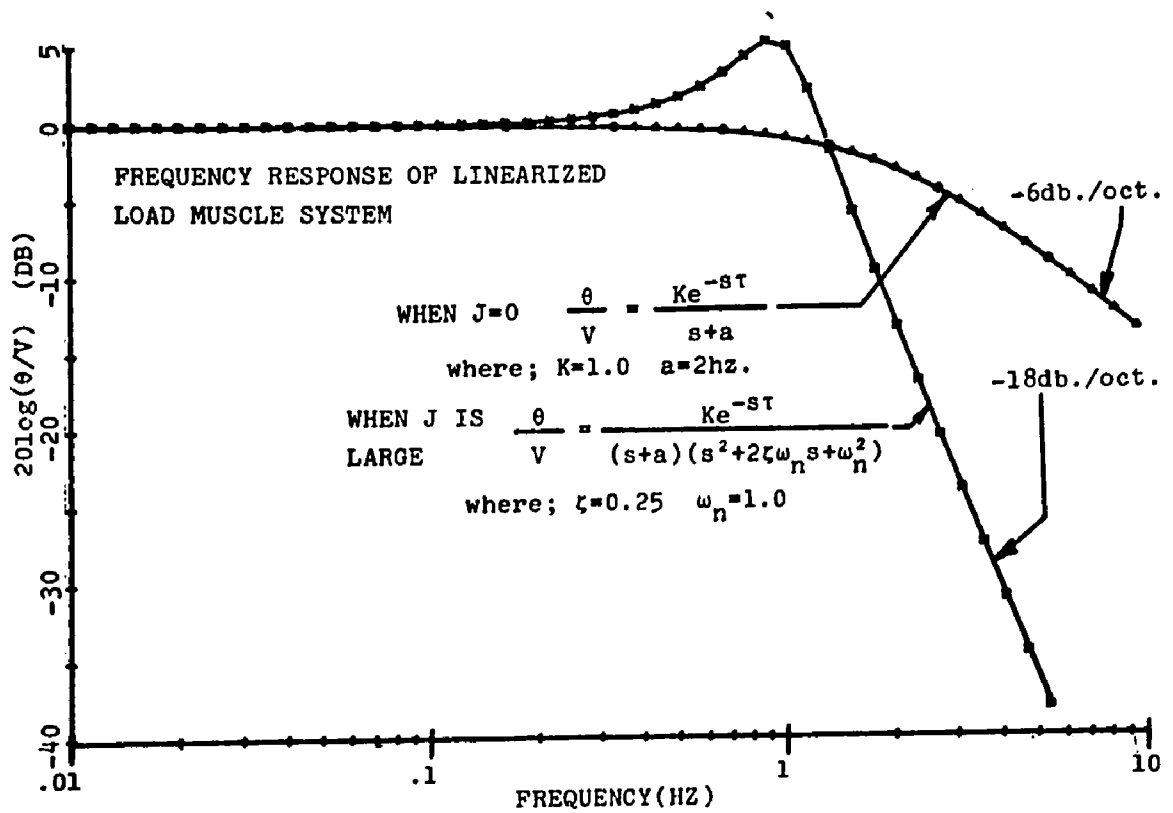
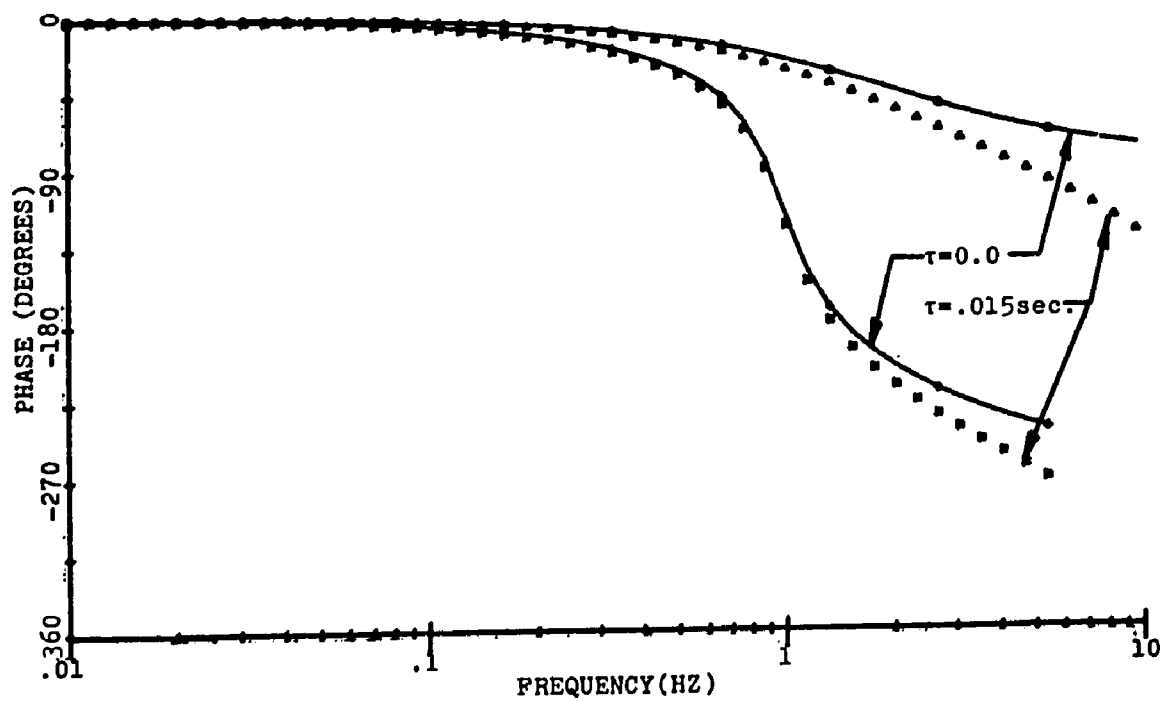


FIGURE 23

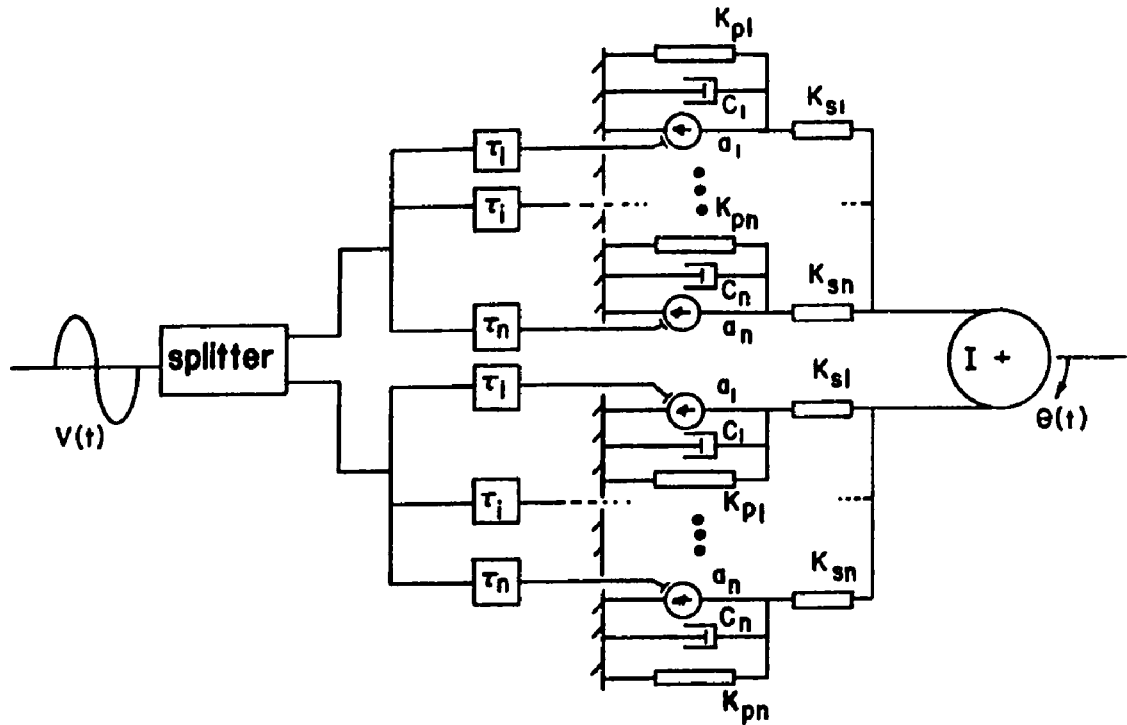
function for two values of inertia load and two values of τ , the activation-conduction delay. Let us direct our attention to several features in both the analytical and graphical expressions of the transfer function. The dominant feature is the low pass filter characteristic. Because the model is linear, the form of the filter characteristic is insensitive to depth of modulation. The form of the response does however depend on the size of the inertial load. A zero load produces a first order filter response. Increasing the load can result in a third order characteristic wherein two of the poles are complex. With complex poles, the large load case produces a resonant peaking, a sharp magnitude attenuation (-18db/oct) above the resonant frequency (ω_n) and a 180° increase in phase lag at high frequencies. Many of these features are seen in the experimental data.

4.2.2 Multiple Motor Unit Model

While useful in examining the basic characteristic of the performance of the muscles, the model in Figure 22 ignores the multiple parallel motor unit structure of muscle. Figure 24 illustrates a configuration where an attempt has been made to incorporate multiple motor

FIGURE 24

Linearized parallel motor unit models of muscle coupled in a configuration similar to that used in the study. $V(t)$ is the modulation signal, the τ_i 's are the conduction and activation transport delays, the K_{pi} 's and K_{si} 's are the parallel and series elastic components, the C_i 's are the parallel viscous components, n is the number of pairs of motor units, s is the Laplace operator, I is the inertial load, and $\theta(t)$ is the inertial load position. The first equation shows the relationship between the output $[\theta(t)]$ and the input $[V(t)]$ in series form. If the τ 's associated with each motor unit are nearly equal, then the transfer function, in general form, is given by the second equation. Note that the number of real poles is equal to n and that there are $n-1$ real zeros. The two terms in the denominator containing I represent a pair of complex poles for large I but reduce to constants when I is zero.



$$V(S) \sum_{i=1}^n \frac{a_i e^{-\tau_i s} K_{si}}{C_i S + K_{pi} + K_{si}} + 2 \theta(S) \sum_{i=1}^n \left(\frac{K_{si}^2}{C_i S + K_{pi} + K_{si}} - K_{si} \right) = \theta(S) I S^2$$

$$\text{If } \tau = \tau_1 = \tau_2 = \tau_3 = \dots \tau_n$$

$$\frac{\theta(S)}{V(S)} = \frac{K_o e^{-\tau s} (S + A_1)(S + A_2) \dots (S + A_{n-1})}{(a_1 I S + \beta_1)(a_2 I S + \beta_2)(S + B_1)(S + B_2)(S + B_3) \dots (S + B_n)}$$

FIGURE 24

units with different mechanical properties into a model. Again, if the individual elements of each motor unit are linearized, a finite series form of the transfer function for the model may be determined in terms of the visco-elastic characteristic of the individual motor units. Further, if an equivalence between the conduction-activation delays of the motor units is assumed, the transfer function of the model may be determined in standard closed form. Both the series and closed form representations of the transfer function are shown in Figure 24. Examination of the closed form reveals several factors of interest. While the single motor unit model contained only poles, this model has both poles and real zeros (A_1 's). Further, the number of real poles (B_1 's) is equal to n , the number of pairs of motor units, while there are $n-1$ real zeros. The two factors in the denominator containing inertial terms represent a complex pole pair for large loads but reduce to a constant at zero load. At very high frequencies in the zero load case, the magnitude ultimately attenuates at -6db/oct while the phase approaches -90° plus the lag due to the activation delay, τ . In the large inertial load case, the addition of the complex pair of poles results in a high frequency

attenuation rate of -18db/oct and a phase lag which tends toward -270° plus the lag due to τ . Between the low and high frequencies however, both inertial load cases may produce attenuation rates and phase angles which are greater or less than the asymptotic values, the rates being dependent on the number and relative locations of the poles and zeros. Thus, in this mid-frequency range, attenuation rates and phase lags considerably different than those occurring at very high frequencies would be possible. This effect is illustrated in Figure 11. There the frequency response of the linearized multiple motor unit model for two sets of parameters and three motor units are illustrated. For these cases the parameters were determined by fitting the model to the indicated experimental data in the Figure¹. Note that while the model predicts an ultimate high frequency magnitude asymptote of -18db/oct , it fits the -24db/oct slope of the data in the 1 to 5 Hz

1 The choice of three motor units was made on the basis of the work of McPhedron et al. (22) and Wueker et al. (23). While the triceps surae contains hundreds of motor units, they found that the dynamic characteristics of the individual units could be classified into three populations. The three units in the model may be thought of as representing the mean dynamic properties of each of these three groups.

range quite well. The model achieves this fit by locating some of the poles and zeros above 5 Hz. Close examination of the model's fit of the 25% modulation case in Figure 11 illustrates a further characteristic of the model. In the neighborhood of 4 Hz the magnitude response exhibits a gradual change from a greater to a lesser rate of attenuation with increasing modulation frequency. This change is achieved in the model by proper placement of the various poles and zeros and suggests a mechanism to explain the decrease in attenuation rate which frequently occurs in the data.

4.2.3 Nonlinear Time Variant Model

While the single and multiple motor unit models are useful guides in relating known properties of muscle to its information processing role, they ignore two of the most analytically difficult properties of the muscle, its nonlinearities and its neural excitation dependency or time variant properties. To include these factors, a model of the muscle would have to include viscous and elastic components which were both nonlinear and time dependent. Since both the nonlinearities and the time dependency of these components have not completely been determined, formulation of an

exact model is not possible. However, if certain assumptions are made, it is possible to construct a nonlinear time variant model which may provide insight into the performance of the muscles. As mentioned earlier, the parallel viscous and elastic components of the single motor unit model have been considered as being composed of a fixed nonlinear component and a time varying component which depends on the time course of the active state. Including these factors as well as the stimulus level saturation effect of the active state (Figures 1C and 2) in the model of Figure 22 produces the differential equation shown in Figure 25 (eq. 3). This equation is similar to the equation for the linearized single motor unit model (Figure 22) in that it is an ordinary third order nonhomogeneous equation. The time varying coefficients of the parallel viscous and elastic components are modelled by the product of the time course of the active state and the absolute value of the modulating signal. The rationale for using the absolute value comes from the bilateral antagonistic configuration of the experiment. The input signal ($A \sin \omega t$) generates the active state. When the signal goes negative, the antagonist muscle is activated thus producing an increase in the viscous and elastic

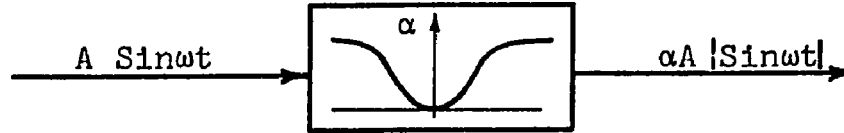
FIGURE 25

Possible form of the differential equation of a lumped parameter model of the muscle when the nonlinear and time varying factors of the viscous, elastic, and contractile components are included. The equation is derived from the transfer function of the linearized single motor unit model of muscle (EQ. 1). The activation-conduction delay τ is assumed to be zero. Equations 2 and 3 illustrate the model in parametric and general form. Note that the coefficients of all the derivatives in EQ. 3 are nonlinear and time variant.

$$JC\ddot{\theta} + J(K_s + K_p)\ddot{\theta} + 2K_s C\dot{\theta} + 2K_s(K_s + K_p)\theta = A \sin \omega t \quad \text{EQ. 1}$$

Assumptions:

- α = a rectifying and saturating function which accounts for;
 a) the active state saturation and
 b) the active state never going negative i.e.



$F(t)$ = the time course of the active state

" = $\alpha A |\sin \omega t|$

$C = C(\dot{\theta}, t) = C(\dot{\theta})[1 + \epsilon_2 F(t)]$; $0 < \epsilon_2 < 1$

$K_p = K_p(\theta, t) = K_p(\theta)[1 + \epsilon_1 F(t)]$; $0 < \epsilon_1 < 1$

$K_s = K_s(\theta)$

Substituting in EQ.1

$$\begin{aligned} & \overbrace{JC(\dot{\theta})[1 + \epsilon_2 \alpha A |\sin \omega t|]}^C \ddot{\theta} + J(K_s(\theta) + \overbrace{K_p(\theta)[1 + \epsilon_1 \alpha A |\sin \omega t|]}^{K_p}) \ddot{\theta} + \\ & 2K_s(\theta) \overbrace{C(\dot{\theta})[1 + \epsilon_2 \alpha A |\sin \omega t|]}^C \dot{\theta} + \\ & 2K_s(\theta) \overbrace{[K_s(\theta) + K_p(\theta)[1 + \epsilon_1 \alpha A |\sin \omega t|]]}^{K_p} \theta = A \sin \omega t \end{aligned} \quad \text{EQ. 2}$$

or in general form:

$$\begin{aligned} & \overbrace{f_3[\theta, |\sin \omega t|]}^C J \ddot{\theta} + \overbrace{f_2[\theta, |\sin \omega t|]}^{K_s + K_p} J \ddot{\theta} + \overbrace{f_1[\theta, \dot{\theta}, |\sin \omega t|]}^{2CK_s} \dot{\theta} + \\ & \overbrace{f_0[\theta, |\sin \omega t|]}^{2K_s(K_s + K_p)} \theta = A \sin \omega t \end{aligned} \quad \text{EQ. 3}$$

FIGURE 25

properties of the whole system rather than a decrease as would have been the case had the absolute value not been used. While this model is in no way exact, it does have merit. The time variant concept of muscle has been examined previously and several of the known properties of muscle are well described by such formulation (20). In addition to serving as a framework within which many of the known dynamic properties of muscle may be considered, it emphasizes the complexity of muscle. Attempts to gain insight into the form of the solution of this equation lead to the literature of nonlinear oscillators and parametric excitation (58,59). While the literature in these areas did not suggest a general form of the solution to the equation, it did provide some insight, particularly the literature on parametric excitation. A fundamental principle of parametric oscillators is that they are driven into oscillation at frequency ω_n when the parameter is made to vary periodically at $2\omega_n$. Here ω_n is the natural frequency of the system when the coefficients are constant. This principle, in view of the model, is of particular interest. In the model, because of the absolute value operation, the fundamental frequency component of the parameters is always at twice the frequency

of the forcing function. Thus, when the forcing frequency is in the neighborhood of ω_n of the system, the fundamental component of the parameter frequency will be at twice ω_n . Thus there are two resonance producing effects acting simultaneously and there should be a sharp resonant peaking effect. Also, as the forcing frequency is swept toward values above ω_n the effect of two resonance producing factors are receeding. It seems likely that this condition might produce high frequency attenuation rates which exceed the rate due to either of the effects acting individually.

Another interesting characteristic of parametric amplifiers relates to stability. In general, linear parametric oscillators are unstable. However, if they contain nonlinearities, as does the muscle, they tend to produce an output which becomes more stable as amplitude increases (58).

A final point to note about parametric oscillators is that under certain conditions they may be used as very narrow band pass filters, i.e., the rates of attenuation of their skirts are extremely large. Recall that large rates of attenuation occurred frequently in the data.

4.3 Discussion of Models and Results

The major results of this study are the frequency response characteristics of the muscle and these results are presented in the form of describing functions. The effect of three parameters, mode of stimulation, depth of modulation, and inertial load on the form of the frequency response were investigated. Some of the effects are quite linear and may be interpreted in terms of the linearized visco-elastic models of muscle. Others are nonlinear and may be examined in the light of the nonlinear model of the muscle. This section will be devoted first to the discussion of the general form of the experimental describing functions and secondly to the effect of the three parameters on various features of the describing function. Table 1 summarizes this section by tabulating the effects of the three parameters on the various features of the describing functions.

4.3.1 Form of the Frequency Response

The most dominant characteristic of the transformation that the muscle imposes on the neural signals it receives is the low pass filter effect. Considered in light of the visco-elastic models of muscle, the low pass filter response is quite expected. The elastic, viscous and inertial components act together to produce

TABLE 1

Summary of the effects of varying the depth of modulation, inertial load size, and mode of stimulation on the various features of the describing function. The table entry briefly describes the effect that the parameter has on the indicated feature of the describing function. Also, the table indicates whether the effect was primarily linear or nonlinear. In many cases the parameter had little or no effect. This usually is an indication that the effect is linear.

Feature Parameter	Low Frequency Magnitude	Low Frequency Phase	Resonant Frequency	Amount of Resonant Peaking	High Frequency Magnitude Asymptote	High Frequency Phase Asymptote
Increasing Depth of Modulation	Decreases (nonlinear)	Little or no effect (linear)	Decreases for small modulation depths (nonlinear)	Decreases (nonlinear)	Little or no effect (linear)	Little or no effect (linear)
Increasing Inertial Load	Little or no effect (linear)	Little or no effect (linear)	Decreases (linear)	Increases (linear)	Masks effects of higher order poles and zeros (linear)	Increases lag by 90° (nonlinear)
Mode of Stimulation	Mode of stimulation had little or no effect on the describing function features.					

TABLE 1

a typical spring-mass-damper low pass filter.

In most of their general features, the describing functions are very similar to the predictions of the linearized models. There are, however, some features that can not be readily explained using the linearized models. The most prominent of these is the non-minimum phase nature of the response. The rate of attenuation of the magnitude response implies a much greater phase lag than that which occurs in the data. This discrepancy was measured and found to be equivalent to the phase angle associated with two poles. The mechanisms responsible for this discrepancy are not clear. The linearized single and multiple motor unit models provide no insight relative to the mechanisms. The suggested nonlinear time variant model, while it emphasizes the general complexities of the muscle and the difficulties of predicting a response, also contributes very little in the way of explaining the mechanism. On a superficial level, the phase discrepancy might be attributed to a complex interplay between the multidimensional nonlinearities and the suggested double frequency parametric forcing. Verification of this would require further study. There are however some other features of the data which suggest that parametric excitation has

a significant role in determining the response of the muscles. The high rates of magnitude attenuation that occur in parametric oscillators are reminiscent of the attenuation rates in the data. These high rates occurred at frequencies where both resonance producing effects, the forcing function and the double frequency parametric excitation are exceeding frequencies which produce resonance. The attenuation rate under these circumstances might exceed the rate which would occur if only one of the factors were present. Another feature of the data which relates to parametric excitation is the sharpness of the resonant peaking. Figure 10 illustrates this result particularly when it is compared to the linearized response (Figure 23). While neither the mechanism nor function of this sharp peaking is clear, the effect does give some support to the suggested nonlinear time variant model, which, in turn provides insight into the information processing aspects of the muscle.

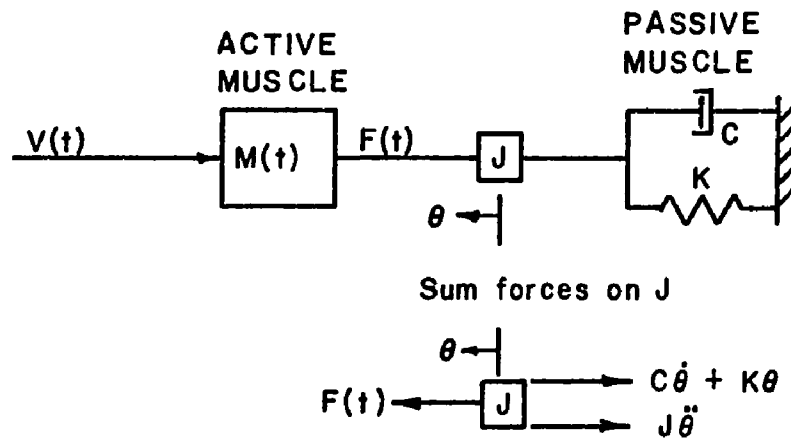
4.3.2 Depth of Modulation

Table 1 summarizes the effects of the depth of modulation on the frequency response characteristic of the muscle. Three features of the response, (low frequency magnitude, resonant frequency, and amount of

resonant peaking) are significantly affected by the depth of modulation and all of these effects are essentially nonlinear. The other three describing function features show little or no dependence on depth of modulation and thus, are essentially affected in a linear manner. Of the three features showing a dependence, the most prominent was the magnitude saturation which occurred at low frequencies. Two factors contributing to the nonlinear saturation effect are readily identifiable. The first of these results from the relationship between isometric force response of the muscle and depth of modulation or level of stimulation. As level of stimulation increases, the tensile force response of the muscle is seen to saturate. Beyond about 40% (Figure 6) modulation the isometric force response is almost constant and independent of modulation depth. The second factor contributing to the nonlinearity is the passive elastic characteristic of the muscle (Figure 27). The stimulus was alternately shifted from one muscle to the other depending on the sign of the modulating signal and at any given time, one muscle was active while the other was passive. The load on the currently active muscle consisted of two components (Figure 26); 1) the inertial load and,

FIGURE 26

Linearized model of the muscle system showing the contribution which the currently passive muscle and inertial load (J) make to the net phase angle (ϕ) of the muscle-load system. $M(t)$ is the impulse response of the active muscle. $-\beta$ is the phase angle contribution of the inertial load and the passive muscle. $F(t)$ is the tensile force response of the active muscle. Note that when the modulation frequency (ω) is equal to the resonant frequency of the passive muscle-load combination, then $J\omega^2 = K$, or the elastic and inertial forces are equal in magnitude but out of phase.



$$F(s) = M(s)V(s) = (Js^2 + Cs + K)\theta(s)$$

$$\text{or } \frac{\theta(s)}{V(s)} = \frac{M(s)}{Js^2 + Cs + K} = \frac{M(j\omega)}{Cj\omega + K - J\omega^2} \Big|_{s=j\omega}$$

$$\text{or } \frac{\theta(j\omega)}{V(j\omega)} = \frac{\sqrt{(M(j\omega))^2} \angle \alpha}{\sqrt{(C\omega)^2 + (K - J\omega^2)^2} \angle \beta} = \text{Magnitude } \angle \phi$$

$$\text{where: } \tan \alpha = \frac{\text{Im}[M(j\omega)]}{\text{Re}[M(j\omega)]}$$

$$\tan \beta = \frac{C}{K - J\omega^2}$$

$$\phi = \alpha - \beta$$

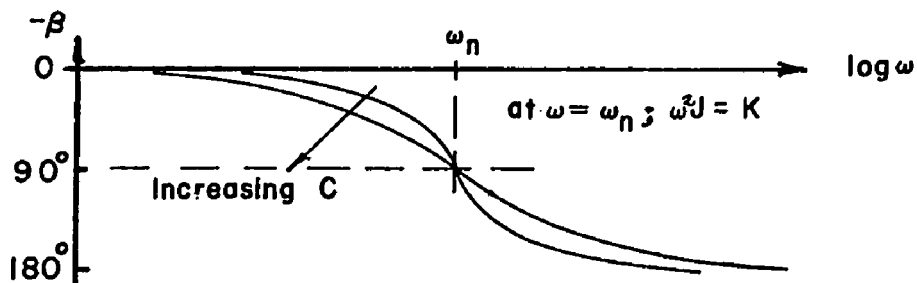


FIGURE 26

FIGURE 27

Typical passive elastic characteristic of the muscle used in this study. Dotted and dashed lines show the characteristic of each muscle individually with origins of the plots displaced. Solid line shows composite characteristic when muscles are coupled (as shown in the inset) with a preload of 0.2kg wt.

PASSIVE ELASTIC CHARACTERISTIC

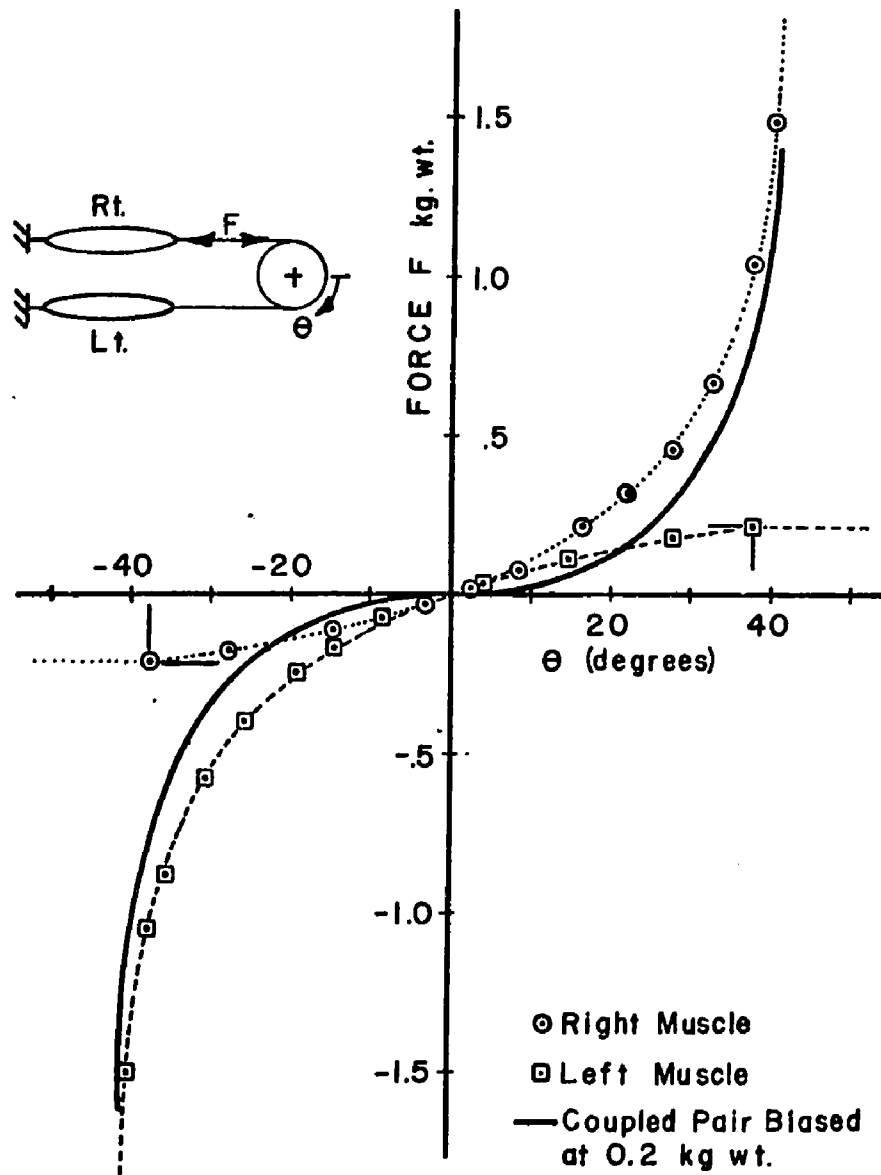


FIGURE 27

2) the mechanical load of the passive muscle. Figure 27 shows that the force required to extend the passive muscle rapidly increases with increasing displacement. Thus the active muscle, limited by its own stimulus level dependent force saturating characteristic, is further limited in producing large displacements by the nonlinear characteristic of the passive muscle. These two factors acting together contribute to the describing function saturation characteristic. Since neither of these two factors is significantly dependent on the mode of stimulation or inertial load size, the occurrence of the saturation characteristic in the results for all three stimulation modes is expected.

While the mechanical properties of the passive muscle contribute to the amplitude saturation nonlinearity, they are relatively ineffective in altering the phase response. Figure 26 illustrates how the properties of the passive muscle contribute to the net phase lag in a linearized situation. The contribution is seen to be that of a typical second order system with a resonance condition occurring when the elastic forces equal the inertial forces ($K = J\omega^2$). Although the development in Figure 26 is based on linearity, it may be used to examine the case where the elasticity of the passive

muscle is nonlinear. In the nonlinear case, K , the effective elastic rate of the passive muscle increases with increasing displacement. An increased K will have the effect of increasing ω_n and thus of pushing the phase characteristic to the right along the frequency axis. The saturation data of Figure 17 shows however that above 40% modulation, displacement amplitude and therefore the effective elastic rate K , changes very little. Thus the nonlinear elastic characteristic should produce only minor shifts of the phase characteristic along the frequency axis. This prediction is in agreement with the describing function data of Figures 13 and 14. The phase data shows a minor trend of shifting of the phase characteristic along the frequency axis as percent modulation increases.

The other two features of the frequency response that were affected by depth of modulation were the amount and frequency of resonant peaking. Both of these effects occurred at low depths of modulation. Figure 10 illustrates the effects. At levels of modulation well below the point where isometric force saturation occurs, the resonant peaking is quite narrow in bandwidth, relatively large in amplitude, and occurs at a reduced frequency. At higher levels

of modulation, where the isometric force characteristic is saturating, the various changes in resonant peaking were not as noticeable from one level of modulation to the next. The above factors together lend support to the concept of the viscous and elastic characteristics of the muscle being time varying or stimulus level dependent. The isometric force characteristic (Figure 6) may be taken as a measure of the level of activity of the active state and thus as a measure of the stimulus level dependent viscosity and elasticity. At higher levels of stimulation the elasticity and viscosity are saturated at their maximum values and thus a resonant peaking change is not evident with changing levels of stimulation. At low modulation levels, e.g., 15% in Figure 10, the stimulus level dependent elasticity and viscosity are significantly below their maximum values. This reduced elasticity contributes to reducing the frequency at which resonance occurs while the reduced viscosity makes the response more oscillatory thus producing the rather large resonant peak. The parametric excitation effect may be responsible for producing the narrow bandwidth of the resonant peak. Note also that the phase profile of the 15% modulation case in Figure 10 is shifted downward

along the frequency axis. This shift, as previously discussed, will occur as a result of a decrease in the effective elastic constant of the passive muscle. The reason for the reduction in elastic constant in this case follows from a lower modulation depth which in turn produces a smaller amplitude displacement and thus the lowered effective elastic constant.

4.3.3 Inertial Loading

Table 1 indicates that inertial loading had an effect that was primarily linear on five of the six describing function features. Apparently, the viscous, elastic, and inertial forces, under some circumstances, interact in a quasi-linear manner. The feature showing a definite nonlinearity was the high frequency phase asymptote. Adding the inertial load to the muscles generally increased the phase asymptote from a lag of 180° to a lag in the neighborhood of 270° , a net change of about 90° . Both linearized models show that addition of an inertial load introduces a complex pole pair into the transfer function and should add 180° to the asymptote. Here as with the non-minimum phase non-linearity, the models contribute little toward explaining the factors responsible for this deviation from

linearity. The feature itself however is probably associated with the non-minimum phase nonlinearity. Both nonlinearities demonstrate a phase lag which is considerably less than might be predicted using linear modelling results. The unexplained factors responsible for the non-minimum phase nonlinearity are thus also probably responsible for this nonlinearity associated with inertial loading.

4.3.4 Stimulation Mode Sensitivity

The relative invariability of muscle performance with changing modes of stimulation was the most pronounced result of the mode sensitivity portion of the study. Considering the different mechanisms that are employed by the three modes in modulating muscle output, the insensitivity to stimulation mode was not initially expected. In the FM case the whole muscle responds as a single motor unit to each stimulus pulse, and increases in force output are achieved through temporal summation of successive tension pulses. In the AM case, temporal summation is a minor transient factor since the stimulus frequency is constant and supra-tetanic. Instead, increases in muscle output are achieved by spacial summation, i.e., increasing

numbers of the parallel motor units are activated. A further characteristic of the contraction mechanism in the AM case is the order of recruitment of the motor units. As depth of modulation is increased the order of recruitment is from large easily fatigued fibers to small less easily fatigued fibers, essentially the reverse of that occurring naturally. Thus, the increment in force generated by the CE's, decreases with each increment in modulation depth. In the AM+FM mode, both spacial and temporal summation factors combine to increase muscle output.

Considering the individual motor units and their contractile elements, it would appear that the three different stimulation modes would have significantly different effects on the timing and magnitude of the CE force response throughout a given modulation cycle. In the AM mode, the various CE's are activated, depending on their size, at different times into the modulation cycle and activation is always at a supratetanic frequency thus saturating the active state. In the FM mode, all CE's are activated simultaneously but the frequency of stimulation varies throughout the modulation cycle producing varying degrees of active state saturation. Because the phase and magnitude of

the inertial load motion reflects the combined effect of all the contractile elements, it is not immediately clear why the timing and magnitude differences on the CE level failed to appear in the describing function data.

Differences between the stimulation modes may also be examined in terms of the linearized single and multiple motor unit models of the muscle. The FM case may be considered in terms of the single motor unit model of the muscle. Since all parallel motor units are activated simultaneously, the net effect is a single motor unit whose dynamics, as a first approximation, might be characterized by a single pole, the value of the pole being determined by some weighted average of the dynamics of all of the motor units. Alternatively, the AM case might be considered in terms of the multiple parallel motor unit model of the muscle. As the depth of modulation is increased, motor units with different dynamic characteristics are activated. If the dynamics of the individual motor units are thought of in terms of individual poles and zeros, then the dynamic performance of the whole muscle might be thought of as a system whose pole-zero locations change with the amplitude of the input signal. Contrasting this with modelling the FM

case as a single motor unit would suggest that there be a difference in the overall dynamic performance between the FM and AM cases as depth of modulation changed. It would appear that the FM case would produce a frequency response whose pole-zero location was independent of depth of modulation while the net pole-zero locations in the AM case would change with depth of modulation. The data, however, do not exhibit these differences between the two cases. This implies that the pole-zero changes, if they occur at all in the AM case, do so only to a minor degree.

There are several factors which possibly contribute to masking differences due to mode of stimulation on the describing function level. As already discussed, the inertia, viscous, and elastic components of the muscle-load system produce a very effective low pass filter with a corner frequency near 1 Hz. Many of the differences between the three modes of stimulation at the CE level occur at frequencies near the instantaneous stimulation frequency which, in general, is well above 1 Hz. The low pass filtering properties of the muscle will attenuate these high frequency differences to a level below detection. Another factor which may be significant is the saturation that occurs in the active

state at larger depths of modulation. If the average stimulation frequency in any of the three modes is sufficient to drive the active state well into saturation for a majority of the modulation cycle, the response of the CE's would show very little difference from one stimulation mode to the next. In this case the total muscle would be working in an essentially "on-off" mode. The isometric force response curves of Figure 6 suggest that at higher levels of modulation the muscles may have been operating, at least to some degree, in an "on-off" mode thus partially masking differences due to mode of stimulation.

A third factor which may have contributed to the masking of differences in response was the data analysis scheme. The time series data analysis methods produced a describing function, i.e., only the fundamental frequency component of the output was considered. Differences due to mode of stimulation may have occurred in the higher harmonics of the output which are ignored in the describing function. However, as indicated by the coherence measurements, differences in the higher harmonics could not have been excessively large. The coherence functions at a given inertial load show little change from one mode of stimulation to the next.

Further, below the describing function corner frequency, the coherence function values range primarily between 0 db and -5 db indicating a small higher harmonic content in this frequency range. Thus while differences in response due to mode of stimulation may have occurred as harmonics in the 0 to 8 Hz range, the coherence functions suggest that these differences are relatively small.

Of the various factors influencing the masking of differences in response, it appears that the low pass filter characteristics of the muscles are the most effective. The motor unit-contractile element structure of muscles requires that the muscle respond to a stimulus train on a pulse by pulse basis. The temporal events associated with each of these responses must have frequency content near the instantaneous value of the stimulation frequency. In these experiments the instantaneous frequency went as high as 70 Hz. The viscous, elastic and inertial properties of the muscle and load severely attenuate events with frequency content above 2 or 3 Hz, however. Differences due to mode of stimulation that occur on a pulse by pulse basis are completely filtered out. Only the summed effect of numerous motor units which have frequency content on

the same order as the modulating signal finally appear at the output.

4.4 Functional Interpretations

This study has identified several facets of the transformation that muscle imposes on the neural signals it receives. Among these were: 1) the basic low pass filter form of the characteristic, 2) the nonlinear non-minimum phase characteristic, 3) the insensitivity to mode of stimulation, and 4) the saturation that occurs with increasing depth of modulation. In this section an attempt will be made to interpretate the functional significance of the first three of these factors. Interpretation of the fourth is included in the following section which considers muscle as a control system element.

4.4.1 The Low Pass Filter Characteristic

The functional significance of the basic low pass filter form of the response relates to the mode of neural information coding used by the CNS. The CNS uses variations of the pulse coded form of information transmission. The organ receiving this information, the muscle in this case, must be capable of decoding or demodulating this signal. The low pass filter

characteristic of the muscle provides this capability. By low pass filtering, the muscle extracts the encoded information (amplitude and frequency of the modulating signal) from the low frequency portion of the neural signal and severely attenuates all of the meaningless high frequency side bands. The low pass filter characteristic thus acts to demodulate the incoming pulse coded neural signal.

From a purely functional aspect, it is also clear that muscle must demonstrate a low pass filter characteristic. The low frequency response enables the animal to engage in normal low frequency motor tasks such as standing, walking, and running (cats typically exhibit periodic motions in the 0 to 5 Hz range (60)). At high frequencies, the severe magnitude attenuation tends to suppress meaningless high frequency limb movements and oscillations (such as tremor) which might otherwise result from the high frequency side bands in the neural signals.

4.4.2 The Non-minimum Phase Characteristic

The rate of attenuation of the magnitude responses of the describing functions implied a much greater phase lag than was actually measured. While we were only

able to speculate about the mechanism responsible for this characteristic, the characteristic itself could have considerable significance in determining the stability of the closed loop neuromuscular system. It is well known that transport delays are a destabilizing influence in control systems. Transport delays produce phase lags which are proportional to frequency and thus tend to lower the frequency at which the open loop phase response crosses the critical 180° value. The synaptic, neural conduction, and muscle activation delays that occur in the loop all contribute phase lags which tend to destabilize the system. The two additional phase angle contributing components of the system are the muscle and the mechano-receptors. The receptors, in general, tend to exhibit phase lead characteristics and thus act as a stabilizing influence since they cancel some of the lag produced by the transport delays. The muscles tend to produce a considerable amount of lag, but the amount, as we have seen, is considerably less than might be expected. Thus, the muscle may be considered as contributing a net phase lead which, by compensating for some of the transport lags, may produce a significant stabilizing influence on the closed loop neuromuscular system. The non-minimum phase

characteristic may thus be thought of as an inherent phase lag compensating characteristic of muscle.

4.4.3 Stimulation Mode Sensitivity

Interpretation of the functional significance of the stimulus mode insensitivity of the muscles must be made with consideration of the pseudo-physiological nature of the stimulus pulse trains that were used. In all three modes, the synthetic coding failed in various ways to exactly simulate physiological neural coding. The data suggest, however, that the mode of neural coding may not be of critical importance since the motion produced by the muscles depended little on the detailed features of the coding. Instead, the motion is dependent upon the lower portion of the frequency spectrum (below 10 Hz) of the net neural traffic impinging upon the muscle. The muscle filters out all of the high frequency side bands produced by the AM, FM, or AM+FM coding processes and shows significant response only to the fundamental frequency component of all the efferent neural traffic on the motor nerve. In terms of the CNS, this mode insensitivity characteristic of the muscles could represent an additional degree of flexibility in the neuromuscular

system. The CNS, in synthesizing a particular control signal to be transmitted to a muscle group, would not be constrained to use a particular type of coding. Instead it would be free to select any scheme, its only constraint being that the low frequency portion of the neural traffic carry the necessary information.

4.5 Muscle as a Control System Element

Thus far we have examined muscle as an information processor by inspecting in detail various facets of its dynamic performance. In the neuromuscular system, these facets work together to produce coordinated muscular activity. In examining muscle as a control element, the rules or criteria governing the coordination of the muscles characteristics are of major interest. What are these rules, or, what are the criteria governing the performance of the neuromuscular system? The neuromuscular system is a component of the total organism and its performance is dictated by the overall goals of the organism. Since the total organism is an adaptive system its immediate goals are frequently in a state of flux, the particular state at any time being dependent on its immediate needs. If the immediate goal of the total organism, for example, is escape, the

performance criteria of the neuromuscular system will no doubt be one of full effort, rapid response. In other situations (e.g. walking) the performance criterion may be one which optimally conserves energy. In analyzing the open loop performance of muscle in a control theory context, we have implicitly selected, out of a variety of possibilities, a set of conditions associated with two particular performance criteria. The two criteria are those which are usually associated with control systems; 1) high speed error reduction with finite steady state error and 2) low speed error reduction with zero steady state error (56). In typical control systems the first is frequently achieved with large loop gains and position plus derivative feedback. The second is achieved with position and integral control and reduced loop gain. It is often possible to attain some of the benefits of both types of control through modification of the open-loop gain with lead-lag and lag-lead networks. Of these two types of control, the muscle spindle receptors together with the myotatic reflex connections suggest that proportional plus derivative control plays a significant role in the performance of the closed loop neuromuscular systems (61). The muscle spindle

receptors are located in the body of the muscle and provide sensory output which is approximately proportional to muscle length and to rate of change of muscle length (62). The myotatic reflex connections provide a neural path for negative feedback of this sensory information to the motoneurons. Further, through efferent control, the gain of these receptors and consequently the gain of the whole neuromuscular control system may be set by the CNS (62). In the signal flow pattern of the neuromuscular system, the spindle receptors are in the feedback path and thus, in an open loop configuration, they are in series with the muscle. Insight into their effect on the open loop response may be obtained by assuming that they are linear elements. Placing them in series with the muscles amounts to adding a zero to the experimentally derived describing functions. Further, because the spindle receptors are under CNS control, the amount of gain (or feedback) they contribute to the describing function and the placement of the zeros are variables. Recall that the muscles, when loaded, had a phase lag which approached -270° . Addition of the 90° phase lead of a zero at the proper frequency could reduce the maximum open loop phase lag to a value near 180° . The significance of

this, in terms of control theory, is that very large loop gains may be used without fear of driving the system unstable. Thus, the CNS, by setting the gain and zero of the spindle receptor, could set the loop gain very high. This action would produce both a rapid error reducing type of performance and small steady state errors. It might appear that the loop gain could be increased indefinitely. There are however at least two factors which limit its value. The first of these is the neural conduction and transmission delay around the loop. These delays add a phase lag to the open loop response which is proportional to frequency. At larger modulation frequencies total lags greater than 180° would occur. Once the phase lag is greater than 180° large loop gains are more likely to drive the system into an unstable region. The second factor does not actually place an upper limit on the value of gain but instead limits its effectiveness in cases with large errors. This is a consequence of the nonlinear saturation response of the muscle to depth of modulation. The saturation characteristic results in a "clipping" of large amplitude error signals and thus effectively reduces the loop gain as error signal amplitude increases.

While this action tends to reduce the speed of response for some error signals, it has two beneficial side effects. First, the effective reduction in gain which occurs with large error signals drives the system into a more stable or less oscillatory region. The second effect affects performance criteria. One of the constraints on all real control systems is that the output of their actuators is limited to a finite value. Within this constraint, it can be shown that maximum speed of response can be achieved when the control system is operated in a "bang-bang" or maximum effort mode (63). In this mode the error is reduced by properly switching the actuator between maximum positive and negative effort. In the neuromuscular system large loop gains together with the saturation characteristic might be instrumental in implementing a maximum effort control mode. When the gain of the muscle spindle is set to a very large value, any error signal will immediately saturate the muscle characteristic thus producing a maximum effort response. Studies with human subjects show that the neuromuscular system frequently operates in a maximum effort mode (64). Subjects instructed to move an inertial load in minimal time produced displacement and velocity patterns which showed,

after a preliminary learning period, a response nearly identical to that of an ideal "bang-bang" servo. It should be emphasized that in the full effort mode of operation, position and velocity feedback signals are not used in the conventional manner. In this mode, the system's performance criteria, while it is still one of rapid response, has been altered by virtue of the mechanism used to achieve the performance. The control problem in this case is one of switching, that is, of predicting the time at which muscle effort should be reversed in order to attain the desired final position. The position and velocity information in this case may be used by the CNS in establishing a switching curve.

Complementary to the optimal performance that may occur in the case of large errors, it appears that the characteristics of the muscle and spindle receptor may also be utilized in a rather optimal fashion in the case of small errors. With small errors, the large loop gains are fully effective in reducing steady state error. This action might be of considerable significance in tasks requiring the movement of a limb to, and maintaining it at, a precise location. As the limb approaches the vicinity of the target, the error

approaches zero. At this point large increases in the loop gain would be very effective in reducing the steady state error toward zero. Note that the non-minimum phase characteristic of the muscle allows loop gains of much larger value than might be used had the characteristic of the muscles been one of minimum phase. As a consequence, much smaller steady state errors may be obtained.

As mentioned earlier, the criteria governing the neuromuscular system's performance changes with the immediate goals of the total organism. The criteria discussed above are probably only a few of a broad spectrum of possibilities. However, while they do represent only a small sample, they serve to illustrate the significance of some of the experimentally derived characteristics of the muscle. The saturation that occurs with increasing levels of stimulation may be a significant factor in both the maximum effort mode of control and in maintenance of stability in the case of large error signals. The non-minimum phase properties may be significant in maintaining loop stability, in rapidly reducing errors toward zero, and in producing a small steady state error.

CHAPTER 5

SUMMARY

The open loop transformation that muscle imposes on sinusoidally modulated neural signals as it converts them into inertial load motions has been studied. The study included a determination of the effects of depth of stimulus modulation, mode of neural coding and size of inertial load of the transformation. Inertial loads of physiological size were used. Three modes of neural coding, frequency, amplitude and amplitude plus frequency modulation of the stimulus pulse train, were used to simulate the physiological recruitment modes of temporal, spacial and combined temporal and spacial summation. Modulation depths were within the physiological range and were based on the isometric response of the muscle. The stimulation signals were applied to the distal portion of the sectioned motor nerve.

The basic feature of the transformation was found to be a low pass filter effect, ie., the inertial load displacement signal was the same as that which would have been produced by low pass filtering the stimulus modulating signal. In addition, the filter characteristic of the muscle demonstrated several types of nonlinear

features. The findings of this investigation may be summarized as follows.

1. Describing functions relating the output inertial load motion to the input stimulus modulating signal took the form of a non-minimum phase low pass filter with a corner frequency in the range of 0.8 to 2.0 Hz. The low pass filter response is a consequence of the viscous, elastic and inertial properties of the muscle-load system. The form of the filter may readily be predicted from simple visco-elastic models of muscle. The non-minimum phase nature of the characteristic was evidenced by a phase lag which in general was considerably less than the magnitude portion of the describing function might predict. This discrepancy between magnitude and phase was interpreted as a phase lag compensating characteristic of the muscle. In most cases the compensation was equivalent to the amount of lead produced by two zeros. While it is only possible to speculate about the mechanisms responsible for the non-minimum phase performance, it is suggested that this compensation effectively places two zeros of lead in the closed loop neuromuscular system and may thus be a major factor in maintaining closed loop stability.

2. The sensitivity of the transformation characteristic of the muscle to inertial loading was found to be similar to the sensitivity of most mechanical actuators to inertial loading. The muscle is a visco-elastic body. Adding an inertial component to it results in a typical oscillatory system with characteristics similar to a simple spring-mass-damper system. The response was oscillatory when inertial loads were driven by the muscle as was evident from the appearance of a resonant peaking in the response. The load also significantly increased the phase lag at higher frequencies, determined the corner frequency of the filter characteristic, tended to mask the effect of higher frequency poles and zeros, and dominated the rate of attenuation of the magnitude portion of the response. In general, the inertial loading affected the response of the muscles in the same way that the mass affects the response of the simple spring-mass-damper system.

3. The magnitude of the describing functions were sensitive to the depth of modulation, but the phase was unaffected. The magnitude nonlinearity that occurred was a typical saturation type of nonlinearity. At low frequencies the inertial load position saturated

at large modulation depths. It is suggested that this nonlinearity was due to two factors: a) the "hard" nonlinear elastic characteristic of the passive muscle and, b) the saturation which occurs in the active state as depth of modulation is increased. The resonance occurring in the response of the muscle-load system was also influenced by the depth of modulation. Both the resonant frequency and the amount of damping increased with depth of modulation. Both of these factors are attributed to the increases in muscle elasticity and viscosity which occurred with increasing levels of stimulation.

4. The transformation characteristic of the muscle-load system was generally insensitive to differences between the AM, FM, and AM+FM stimulation signals that were used in the study. Instead of responding to the detailed characteristics of the individual neural codes, the muscles responded to the low frequency component of the net neural traffic on the motor nerve. Apparently the low pass filter characteristic of the muscle attenuates all of the high frequency sidebands associated with modulating process. The remainder of the signal, which codes the amplitude and frequency of the modulating signal, is transformed by the muscles into an inertial

load position.

The significance of the insensitivity to mode of stimulation displayed by the muscle is not clear. It is suggestive however of another degree of signal processing flexibility in the neuromuscular control system. Since the motion that will result from a neural signal can be relatively independent of the detailed features of neural coding, the CNS is free to select a code on the basis of factors not explicitly related to the motion.

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APPENDIX I

Stimulator block diagrams

The stimulator circuitry used in the study is shown in block diagram form in Figures I.1 and I.2. Figure I.1 shows the configuration of circuit elements which were used to produce the FM stimulus pulse trains. Figure I.2 is a diagram of the circuit used to produce the AM and AM+FM stimulus pulse trains.

Figure I.1

Block diagram of the circuitry used to frequency modulate the stimulus pulse trains. The full wave rectifier and voltage to frequency converter are used to synthesize the FM pulse train. The comparator and reed relays constitute a commutator which switches the stimulator pulses to either the right or left muscle constant current sources.

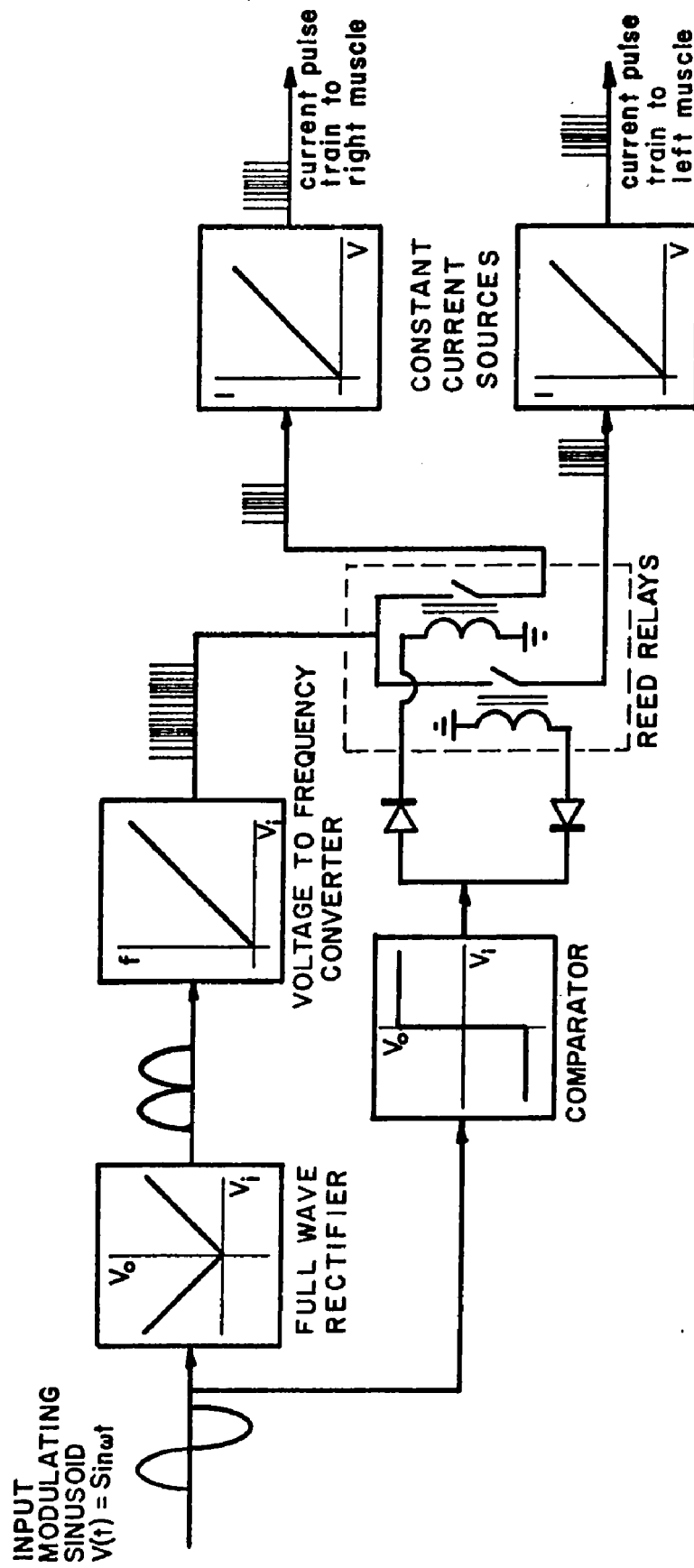


FIGURE I.1

Figure I.2

Block diagram of the circuitry used to synthesize the AM and AM+FM stimulus pulse trains. The voltage to frequency converter is used to generate either a fixed frequency pulse train or a frequency modulated pulse train. The multipliers are used to sinusoidally modulate the amplitude of the pulse trains. Threshold pulses are added to the amplitude modulated pulse trains. The commutator of Figure I.1 is used to switch the pulse train to either the right or left muscle constant current source.

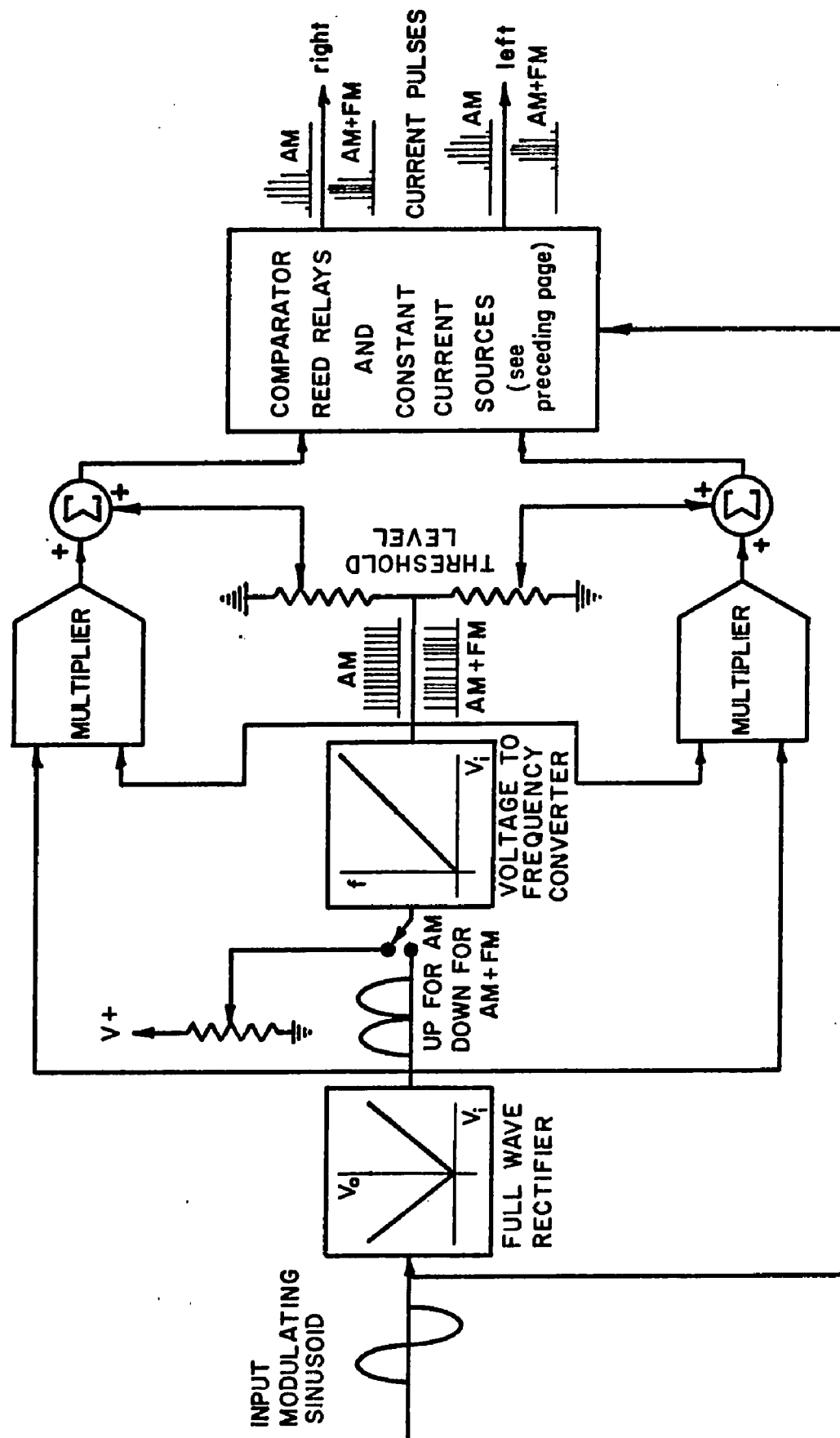
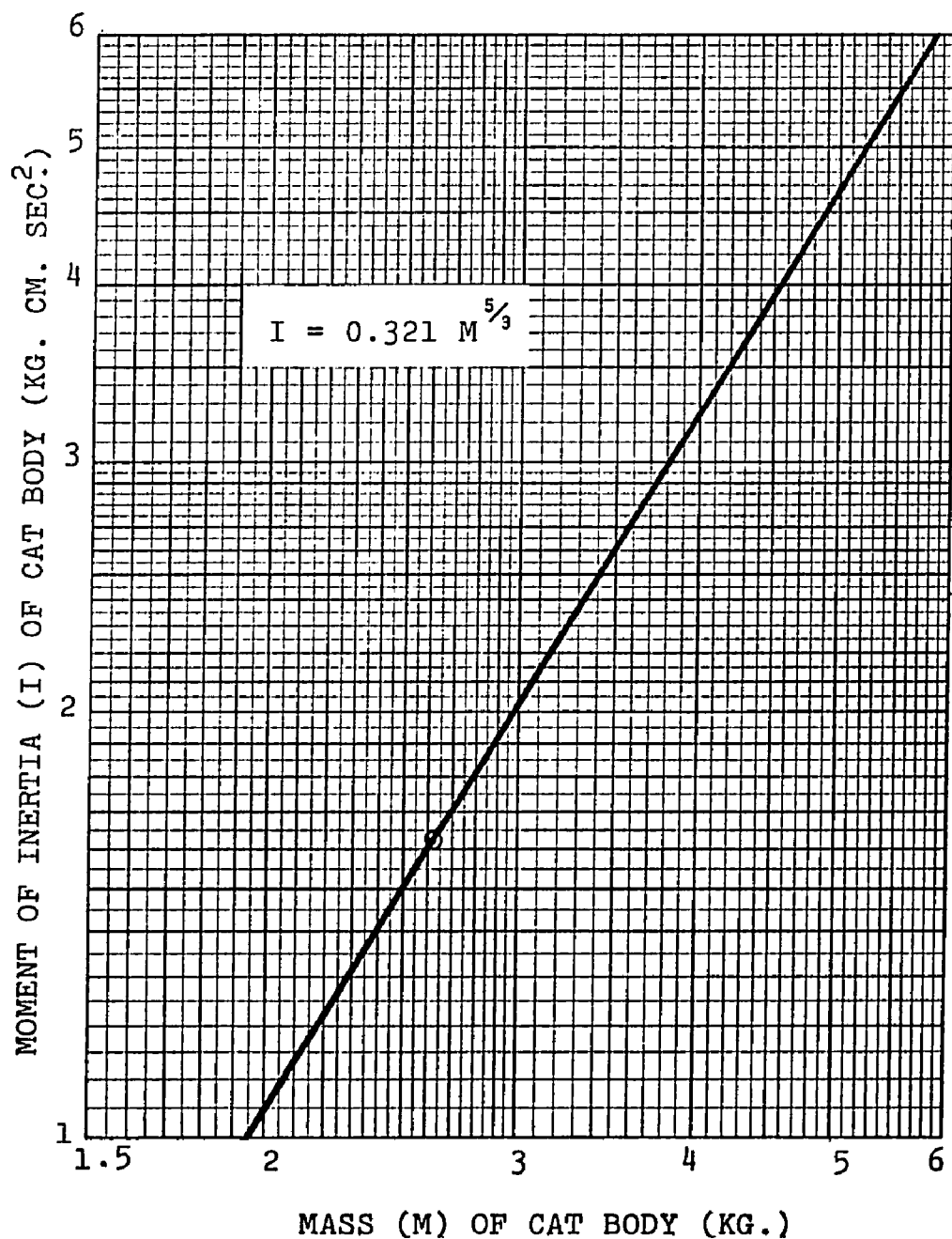


FIGURE I.2

APPENDIX II

Cat body moment of inertia

Moment of inertia of the cat body about the ankle
in the sagittal plane v:s cat body mass.



APPENDIX III

Describing Function

Let the input, $V(t)$ to a nonlinear system be

$$V(t) = A \sin \omega t.$$

If the fundamental frequency term in the Fourier series of the response is designated by

$$\text{Fundamental Component} = a_1(A, \omega) \sin[\omega t + \phi(A, \omega)]$$

then the describing function for the nonlinear system is defined as

$$\text{Describing Function} = \frac{a_1(A, \omega)}{A} \angle \phi(A, \omega)$$

where $\frac{a_1(A, \omega)}{A}$ is the magnitude and $\phi(A, \omega)$ is the phase portion of the describing function. In general a_1 , and ϕ are both functions of the amplitude and frequency of the input.

APPENDIX IV

Time series analysis techniques

Correlation Analysis

The correlation function of a signal, $\theta(t)$, with another signal, $V(t)$, is defined as:

$$R_{\theta V}(\tau) = \lim_{T \rightarrow \infty} \frac{1}{2T} \int_{-T}^T \theta(t) V(t-\tau) dt.$$

When $\theta(t) = V(t)$ the function is termed the auto correlation function and when $\theta(t) \neq V(t)$ the function is termed the cross-correlation function. This definition requires that the integration be carried out over infinite time. In practice the integration can only be performed over a finite time span and the result is an estimate of the integral. This realizable estimate is termed the auto(cross)-correlogram to distinguish it from the auto(cross)-correlation function. If $V(t)$ and $\theta(t)$ are the input and output of a linear system with impulse response $H(t)$, then

$$R_{\theta V}(\tau) = \int_{-\infty}^T H(t) R_{VV}(\tau-t) dt.$$

If $R_{VV}(t)$ is an impulse function, $\delta(t)$, then evaluation of the integral yields $R_{\theta V}(\tau) = H(\tau)$. When $V(t)$ is a sinusoid whose frequency is linearly swept over a large bandwidth, $R_{VV}(t)$ approximates an impulse function (see Figure 9 in the text) and thus $R_{\theta V}(\tau)$ approximates the impulse response of the system.

Spectral Analysis

If $V(t)$ and $\theta(t)$ are the input and output of a linear system, then $H(j\omega)$ the systems frequency response may be obtained from

$$H(j\omega) = \frac{S_{\theta V}(j\omega)}{S_{VV}(\omega)}$$

where: $S_{\theta V}(j\omega)$ is the output-input cross power spectral density and $S_{VV}(\omega)$ is the input power spectral density. $S_{VV}(\omega)$ and $S_{\theta V}(j\omega)$ may be found by Fourier transforming the appropriate correlograms, i.e.,

$$S_{VV}(\omega) = \int_{-\infty}^{\infty} R_{VV}(\tau) e^{-j\omega\tau} d\tau$$

and

$$S_{\theta V}(j\omega) = \int_{-\infty}^{\infty} R_{\theta V}(\tau) e^{-j\omega\tau} d\tau$$

Harmonic Filtering

If the input to a nonlinear system is

$$V(t) = A \cos \omega t$$

and the output is

$$\theta(t) = a_1 \cos(\omega t + \phi) + a_2 \cos 2(\omega t + \phi) + a_3 \cos 3(\omega t + \phi) + \dots$$

then the cross-correlation of $\theta(t)$ with $V(t)$ is

$$R_{\theta V}(\tau) = \lim_{T \rightarrow \infty} \frac{1}{2T} \int_{-T}^T [a_1 \cos(\omega t + \phi) + a_2 \cos 2(\omega t + \phi) + \dots] A \cos \omega(t - \tau) dt$$

Due to orthogonality considerations, all higher harmonic terms in the integral cancel and the value of the

integral reduces to

$$R_{\theta v}(\tau) = \frac{a_1 A}{2} \cos(\omega\tau + \phi)$$

which is the same result as would be obtained if the system were linear with response $a_1 \cos(\omega t + \phi)$. Thus the correlation process extracts the fundamental frequency component from $\theta(t)$. The $R_{\theta v}(\tau)$ thus obtained may be used to calculate the describing function, $H(j\omega)$ of the nonlinear system by using the methods described above for linear systems

Practical Considerations

In practice the correlograms were calculated by on-line processing of taped experimental data with a Hewlett Packard 2115 computer system programed to calculate 256 point correlograms. The analog data was digitized at a sampling rate of 20 Hz., each 256 point block representing about 12.8 seconds. The maximum frequency of the input modulating signal was always less than 8 Hz. Sampling at 20 Hz. thus satisfied the nyquist sampling criterion. For each block of 256 points, the correlogram was calculated, averaged with results of previous blocks, and displayed on a CRT. Initially the display fluctuated somewhat but as successive blocks were averaged the correlogram converged toward a stable final value. Data record lengths were

always greater than six minutes. Thus, sampling at 20 Hz., each 256 point correlogram represented the mean of at least 7200 measurements or about 28 measurements per correlogram point.

The auto- and cross-correlograms were then processed by a second computer program to obtain the respective input and output-input power spectral densities and finally $H(j\omega)$, the describing function. This second program, using a Fast Fourier Transform (FFT) algorithm, calculated describing function magnitude and phase values for up to 128 discrete modulation frequencies. Each describing point value represented the mean of at least 56 measurements.

APPENDIX V

Describing Function Modelling

Experimentally measured describing functions were fit with various models of the general form

$$M(s) \Big|_{s=j\omega} = \frac{K e^{-s\tau} \prod_{i=1}^q (s+z_i)}{\left\{ \prod_{i=1}^j (s+p_i) \right\} \left\{ \prod_{i=1}^m (s^2 + 2\zeta_i \omega_{ni} s + \omega_{ni}^2) \right\}}$$

where the models parameters are: z_i and p_i are real zeros and poles, ζ_i and ω_{ni} define pairs of complex poles, τ is a transport delay and K is a gain factor.

The error between the data and the model was defined as the absolute value of the complex difference between the data and the model at each modulation frequency, i.e.

$$\epsilon(\omega_i, X_k) = |H(j\omega_i) - M(j\omega_i, X_k)|$$

where $H(j\omega_i)$ is the describing function data at modulation frequency ω_i and X_k is the vector of the k model parameters. An integral-error-squared cost function $C(X_k)$ was defined as

$$C(X_k) = \sum_{i=1}^n [\epsilon(\omega_i, X_k)]^2$$

where the summation is over all data points. A subroutine available in IBM's Scientific Subroutines Package was used to find the minimum of $C(X_k)$. Using 'descent methods,' the subroutine, FMFP computed the unconstrained

minimum of the cost function. Given an initial vector of trial values for the parameters X_k , the subroutine incremented this vector in a direction which drove $C(X_k)$ to a minimum and the cost's gradient, $\partial C(X_k)/\partial X_k$, toward zero. The vector of parameters which fulfilled these criteria defined the best fit for the model.

In some cases the error was defined as the difference between the absolute values of the data and model, i.e.

$$\varepsilon(\omega_1, X_k) = |H(j\omega_1)| - |M(j\omega_1, X_k)|$$

Using this error as the cost function produces models which fit only the magnitude portion of the describing functions and not their phase.

The statistic that was used to measure goodness of fit of a model was percent error averaged over the data points, i.e.

$$\% \text{average error} = \frac{100}{n} \sum_{i=1}^n \frac{|M(j\omega_i, X_k) - H(j\omega_i)|}{|H(j\omega_i)|}$$

APPENDIX VI

Harmonic Analysis

Let $\theta(t)$ be the periodic response of a nonlinear system to the input signal $V(t) = A \sin(\omega t)$. Consider $\theta(t)$ to be the sum of two components i.e.

$$\theta(t) = \bar{\theta}(t) + n(t)$$

where: $\bar{\theta}(t)$ is the signal or fundamental component of $\theta(t)$ which is proportional to the input $V(t)$, $n(t)$ is the higher harmonic and noise components of $\theta(t)$.

The coherence function, $\gamma^2(\omega)$ is a measure of the effect of $n(t)$ and is defined as

$$\gamma^2(\omega) = \frac{S_{\bar{\theta}\bar{\theta}}(\omega)}{S_{\bar{\theta}\bar{\theta}}(\omega) + S_{nn}(\omega)}$$

where: $S_{\bar{\theta}\bar{\theta}}(\omega)$ and $S_{nn}(\omega)$ are the signal and noise power spectral densities of the output $\theta(t)$.

$\gamma^2(\omega)$ ranges in value from zero (when $S_{\bar{\theta}\bar{\theta}}(\omega)$ is zero) to one (when $S_{nn}(\omega)$ is zero).

A γ^2 of one results when there is only signal power in the output and implies a linear noise free system. A γ^2 of zero implies that there is no signal power in $\theta(t)$. The coherence function may also be defined as

$$\gamma^2(\omega) = \frac{|S_{\theta V}(j\omega)|^2}{S_{\theta\theta}(\omega) S_{VV}(\omega)}$$

where: $S_{\theta\theta}(\omega)$ is the power spectral density of $\theta(t)$ and $S_{\theta V}(j\omega)$ and $S_{VV}(\omega)$ are defined in Appendix IV.

When $S_{\theta V}(\omega)$ is zero, $\gamma^2(\omega)$ is zero. This implies that there is no correlation between output and input and there-

fore any describing function of the system would be meaningless.

In practice the coherence functions were generated using the following relationships (53).

$$S_{\bar{\theta}\bar{\theta}}(\omega) = S_{vv}(\omega) |H(j\omega)|^2$$

$$S_{nn}(\omega) = S_{\theta\theta}(\omega) - S_{vv}(\omega) |H(j\omega)|^2$$

Substituting these quantities in the first definition of coherence function above yields

$$\gamma^2(\omega) = \frac{S_{vv}(\omega) |H(j\omega)|^2}{S_{\theta\theta}(\omega)}$$

$S_{\theta\theta}(\omega)$, $S_{vv}(\omega)$, and $H(j\omega)$ were calculated using the correlation and spectral methods of Appendix IV. A computer program was written to carry out the necessary calculations.

Figure 19 illustrates a set of coherence functions. Note that at high frequencies the coherence function occasionally drifts to values greater than 0 db which are outside of the 0 to 1 theoretical range of the function. Efforts to identify these inconsistencies were not completely successful, however all indicators pointed toward it being the result of digital noise introduced in the process of digitizing $\theta(t)$. Due to the low pass nature of the muscles, $\theta(t)$ was sometimes attenuated to a level approaching the resolving power of the analog to digital converter. At these levels the digitized value is always

approaching the resolving power of the analog to digital converter. Thus the digitized value is always less than the actual value and can result in an error of up to 50%.

A second measure of harmonic distortion in $\theta(t)$ is the signal to noise ratio (S/N). The S/N ratio is defined as

$$S/N = \frac{\sum_{i=1}^n S_{\theta\theta}(\omega_i)}{\sum_{i=1}^n S_{nn}(\omega_i)}$$

where the ω_i are the modulation frequencies at which frequency response data was calculated. The value of this measure ranges from zero to infinity. A noise free system with only signal power in its output would have a S/N ratio of infinity while a system whose output was completely noise would result in a S/N ratio of zero.