UNIVERSITY OF SOUTHAMPTON

FACULTY OF ENGINEERING AND APPLIED SCIENCE

DEPARTMENT OF ELECTRICAL ENGINEERING

FEEDBACK MECHANISMS FOR HUMAN POSTURAL COMPENSATION

A THESIS SUBMITTED FOR THE DEGREE OF

DOCTOR OF PHILOSOPHY

BY

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(DR.ING. - M.SC. BIO-ENGINEERING)

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"IT IS NOT POSSIBLE TO DEFINE
WHAT LIFE IS BUT ONLY TO
ANALYSE ITS MANIFESTATIONS"

Galileo Galilei

CONTENTS

Page No.

CONTENTS

CONTENTS

CONTENTS

CONTENTS

CONTENTS

UNIVERSITY OF SOUTHAMPTON

ABSTRACT

FACULTY OF ENGINEERING AND APPLIED SCIENCE
ELECTRICAL ENGINEERING DEPARTMENT

Doctor of Philosophy

FEEDBACK MECHANISMS FOR HUMAN POSTURAL COMPENSATION

By Armando Ferraioli

The thesis describes an experimental study performed using simple engineering techniques, to investigate the ability of the human operator to achieve postural compensation for gross body movements when performing a manual task. These investigations were also used as the basis of a structural model which can be quantified by the evaluation of the frequency responses of its component elements.

As with most non-invasive biological tests the difficulties were to measure suitable variables and to isolate the various elements in the system. A number of experiments are described in which vision, proprioception and vestibular sensation are variously disabled when arm posture is controlled in the presence of vertical whole-body vibrations. The results are subjective since no model can be conclusively verified. However physiological measurement of relevent EMG activity in the deltoid muscle shows that compensation is an active neurological response under all the test conditions. Furthermore, assumption of superposition of the sensory information appears justified given the variability of the experiments conducted.

While much experimental refinement is necessary to identify the control parameters numerically, the model does seem to agree with the conceptual form which has been developed for the control of a mechanical arm with a fully-flexured tactile hand-manipulator. Adaptation within this model consists merely of the selection of appropriate feedback channels using simple criteria based on sensory data.

ACKNOWLEDGEMENTS

Due to the interdisciplinary nature of this dissertation, a large number of people have aided in its evolution. The author would like to express his thanks to:

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Technical assistance during the design and testing of the vibrating platform was provided by technicians from both the Electrical and Electronics Departments.

The author is grateful to all members of the Control Engineering Group and wishes to thank all the volunteers who acted as subjects, particularly to *Ian Swain* and *Sam Lagumina* for their endurance.

Thanks are due to *Joanne Harley* for her typing of this thesis and to my wife *Maria Teresa Kindjarsky* for drawing the many figures and diagrams included.

Finally, I gratefully acknowledge my personal support under a "Higher of Europe Scholarship" sponsored by the British Council.

The author wishes to dedicate this thesis to his parents and his family for their constant encouragement.

LIST OF FIGURES

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LIST OF FIGURES

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LIST OF FIGURES

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LIST OF FIGURES

LIST OF TABLES

CHAPTER ONE

INTRODUCTION

Interest in control engineering descriptions of neuromuscular system originates not only from academic study of physiological mechanisms but also from a requirement for better understanding of the basic dynamics of the human operator.

Feedback concepts are recognised as having important implications for the investigation of the mechanisms of overall postural control and this may be of some significance in understanding problems concerned with the design of artifical limbs.

The study of the neuromuscular control system also overlaps the study of manual control systems. The precision of the latter is ultimately limited by the properties of the physiological part of the system. The dynamics of the postural control system may be observed in many different ways. Since the input-output investigations must be of a peripheral nature, in human experimentation, different experimental methods of analysis provide information on different aspects of the system. Collectively, such experiments complement each other.

In such a complex system determining a set of experiments which can completely elucidate the nature of the system is not apparent and probably is not even feasible as a theoretical concept.

The work undertaken in this thesis has been motivated by a somewhat different approach to the same problem. When manual tasks are undertaken it is possible to compensate, largely subconsciously, for gross body movements and it is the role of the vestibular and proprioceptive sensors in this adaptation which we seek to model.

It is hoped that such an understanding might lead to a significant improvement in upper limb prosthesis. At present no such compensation is available, often with quite disastrous results.

The most general purpose of this work was to investigate the mechanisms involved in human motor control by means of adequate stimulation of the feedback pathways involved.

The problem has been restricted by studying only the control of the hand in a stable position despite gross body movements and by limiting the sensory feedbacks, as much as possible through the vestibular and proprioceptive pathways.

The various sections of the thesis may appear somewhat discontinuous. This is mainly due to the nature of the problems discussed. In some cases the status of present knowledge allows a sufficient formulation of the problems for a mathematical model to be formed. But in other cases only qualitative understanding is possible. It has been felt necessary to include an extensive review of the neuromuscular system and in the control of upper extremity function. This may prove useful to the reader with an engineering background. The contents of the thesis may be summarised as follows:

Chapter 2 and 3 review the physiological aspects of the work.

Chapter 4 reviews the control of skeletal muscles in the light of feedback concepts. In Chapter 5 the design of the vibrating platform used in experimental work is discussed together with its control system, while in Chapter 6 the measurement system is outlined.

Chapter 7 reports a measurement of the transmissibility of low frequency vertical vibrations to the shoulder and head. In Chapter 8 an attempt is made to model certain aspects of the overall control structure in a way which is useful to the particular functions being investigated. A block diagram structure is proposed in which individual sensory feedback paths may be isolated and described, at least conceptually, by transfer functions.

In Chapter 9 an experimental program is outlined which attempts to quantify the characteristics of the transfer function dynamics by non-invasive methods.

Chapters 10 and 11 report the experimental results from the many tests performed.

Finally Chapter 12 discusses the results obtained and makes suggestions for a further study.

CHAPTER TWO

REVIEW OF THE NEUROMUSCULAR SYSTEM AND ITS COMPONENTS

2.1. INTRODUCTION

Animals can produce a wide range of highly co-ordinated movements under a variety of conditions. This ability requires the interaction of central commands with sensory feedback, and several hypotheses have been suggested for integrating central commands and sensory feedback to produce co-ordinated movements.

The neuromuscular system referred to here is the output or actuation element of the human controller. It is a composite of neural and muscular components situated in the spinal cord and the periphery - typically a limb and its neural connections - operating on commands sent from higher centres.

A complete study of the neuromuscular system would be far beyond the scope of this thesis. However an overall view will be given in order to provide the reader with a physiolgical background.

2.2. SOME PHYSIOLOGICAL BACKGROUND

The nervous system of higher organisms are, in effect, communication systems transmitting electro-chemical impulses (or signals). The structural unit of these communication systems is the individual nerve cell or neuron. Although sensors are highly diversified in both structure and function, each sensor is made up of the same basic parts: the soma, the axon, the dendrites and the synapses. Figure 2.1 shows a reconstructed motor neuron. Each neuron receives signals from many other neurons through terminal contacts referred to as synapses. Synapses are found on the neuron's cell body or soma, and on its dendrites, the tree-like structures which are routed in the soma. The geometry of the dendritic trees varies greatly, and is often crucial to the neuron's function. At the soma, ionic currents produced at the synapses on the soma and the dendrites are combined or integrated. While considerable physiological work has been done related to this integrative process, the details of the process are not fully known. In most neurons synaptic potentials are summed in a nonlinear fashion for synapses located within the dendritic trees, but are summed nearly linearly for synapses located on the soma membrane itself, since here summation occurs only over a limited range of potentials below the threshold for spike generation. In the motor neuron, initiation of spikes occur near the axon hillock, where the soma and the axon are joined, when the summed ionic currents at the axon hillock produce a depolarization to the threshold level (the threshold is not necessarily constant), the neuron fires, transmitting an output signal (action potential, discharge or spike) down the axon to connecting neurons. Spikes can also be initiated in dendrites themselves, but we do not consider such situation here.

We can classify neurons in three types according to their function: sensory, interneuron, and motor. Sensory neurons either transduce stimuli directly into spike trains, or are activated synaptically by specialised epithelial cells such as those in the auditory or olfactory systems; interneurons connect neurons with other neurons; and motor neurons transmit signals to motor mechanisms such as muscles and glands. Some motor neurons are simultaneously interneurons. The physiological features of neurons differ both among and within these classes.

The neuron consists of a well-conducting electrolyte gel, surrounded by a somewhat insulating membrane (a leaky membrane). Changes in the permeability of the membrane lead to the action potential referred to above. More specifically, the cell membrane differentiates between different types of ions and, as a result, a potential difference exists across the membrane, the size and sign of which are a function of the relative permeability of the membrane to various ions (such as Na^+ and K^+ ions). This potential difference serves as a source of stored energy, the partial release of which constitutes an action potential. This propagated impulse which travels along the axon is an all-or-none event. The signals received by the neuron via its synaptic contacts are labelled as excitatory or inhibitory depending upon whether the membrane potential is depolarised or hyperpolarised upon the receipt of the signal. Excitatory and inhibitory potentials vary in magnitude and are functions of the membrane potential. In some cases, inhibitory synapses are located closer to the soma than excitatory ones. In motor-neurons, however, inhibitory and excitatory synapses appear to be distributed uniformly over the cell. It is important to note that when a neuron emits an action potiential, it becomes temporarily incapable of integrating further input signals. For a brief period of time, in the order of one, or two milliseconds, the neuron does not respond to any stimulus (the absolute refractory period) and for several milliseconds thereafter its relative threshold (i.e. the total amount of excitation required to trigger an action potential) is higher than usual (the relative refractory period). The relative refractory period is not fixed. The absolute refractory period is essentially of fixed length. Sequence of neuron firings, referred to as spike trains, arise either from so-called spontaneous activity or as responses to stimuli. A stimulus which effects the firing of the neuron results in a spike train that is either a mixture or an interaction of the two. When stimuli reach the neuron at high rates the contribution of the spontaneous activity typically becomes small.

2.3. BASIC ANATOMICAL AND FUNCTIONAL FEATURES OF MUSCLES

Skeletal muscle is the ultimate effector mechanism of all our mechanical activities. They are unilateral force generator which act across joints and function either as prime movers, synergist, or antagonists.

Skeletal muscle is made up of individual muscle fibres, each a single, multinucleated cell which is cylindrical in shape. The muscle fibres are made up of fibrils, and the fibrils are divisible into individual filaments (Figure 2.2). The filaments are made up of a series of segments called sarcomeres, which are the basic contractile elements of skeletal muscle. The sarcomere includes myosin and actin filaments which slide relative to each other during shortening as illustrated in Figure 2.3. The basic element for modification of skeleton muscle tension or length is the motor unit. (Figure 2.4). This consists of a single motor neuron in the spinal cord, the axon through which it transmits impulses to the periphery, and all muscle fibres to which the axon is connected. As the motor axon reaches the muscle it innervates, it divides into multiple branches, or collaterals, each of which makes contact with a single muscle fibre; conversely, each muscle fibre is activated by only one input axon. The region at which the axon makes contact with the muscle fibres is called the "motor end-plate" or "neuromuscular junction". This is a region of very small dimensions (a few square microns) where the membranes of the axon and the muscle come into very close contact. Upon reaching the end-plate, a pulse travelling from the motor cell in the cord down the axon to the muscle , triggers the release of a minute quantity of a chemical agent, stored at the end-plate, which diffuses across the junction to react with the muscle fibre membrane in such a way as to initiate a comparable pulse in the muscle. By a process that is not well understood, this electrical pulse, or sudden depolarisation sweeping across the muscle membrane, triggers a set of energy-releasing chemical reactions in the muscle protein which cause contraction or development of tension. Thus, each unitary (pulse) event in the motor neuron elicits a unitary mechanical response in every muscle cell it reaches; the response is called the "muscle twitch". Like the electrical pulse, it is all-or-none, i.e. it is basically an invariant quantised event. Each muscle fibre contributes its individual twitch response to the summated over-all response of the motor unit. A generalised tension time plot of a twitch response can be seen in Figure 2.5, but considerable differences in its details will be manifested between different motor units. These arise because a motor neuron does not always innervate the same number of muscle fibres, nor do individual muscle fibres necessarily generate twitches with the same amplitude or time course. Indeed, it is this variability in the size of the motor unit and the details of its unitary events that contribute to the richness and complexity of neuromuscular control. If a sequence of impulses reaches the muscle fibre each impulse will trigger a twitch response, and if the impulses are close enough together in time a summation of tensions will result. As can be seen from Figure 2.5 the total tension developed will depend on the timing or frequency of the arriving nerve impulses, their time span, and the temporal characteristics of the muscle twitch. When the nerve impulse frequency is high enough, however, a saturation effect develops and the individual twitch response fires into a continuous maximum tension. This condition is called "tetanus", the neuron frequency necessary to produce this (the "tetanic fusion frequency") will of course depend on the twitch time of the muscle, which, as already pointed out, is variable from unit to unit. A few general statements on the organisation of neuromuscular elements should be made. First it is important to note that all the muscle fibres in a motor unit lie within the same parent muscle, i.e. they are not distributed between various muscle. There may be anywhere from about 10 - 10.000 fibres in a unit and from rather few to several million units within a muscle. The fibres in a muscle are organised into a hierarchy of bundle called "fascicles" and these in turn are bound together to form the total muscle.

2.3.1. PARAMETERS OF CONTROL FOR MOTOR UNITS

There are several motor system parameters which are relevant to the control of muscle tension at the peripheral level of the motor unit. The first important parameter is the size of the motor unit itself, i.e., the mumber of muscle fibres innervated by a single motor neuron. There is a large variation in the size of motor units sometimes referred to as the "innervation ratio". For a large and powerful muscle such as the gastrocnemius as many as several thousand muscle fibres are directly innervated by a single neuron in the lumber region of the cord; at the other extreme, a motor cell in the oculo-motor nucleus in the brain stem may control as few as ten muscle fibres which serve to rotate the eyeball.

A second obvious parameter is the total number of fibres available in the entire muscle. Again, a powerful muscle such as the gastrocnemius, as would be expected, has a very, large number (in the order of a million), while the smallest muscle the tensor tympany has only about 10000 fibres. Taken together, the innervation ratio and the total number of muscle fibres enables us to compute the total motor neuron pool size (i.e. the number of motor neurons for each muscle). Assuming that all muscle fibres of comparable diameter have the capacity of generating the same amount of tension, it follows that the size of the motor unit determines the minimum quantum of tension or force which muscle can "negotiate" in the control of movement.

A third factor of extreme importance in the graduation of muscle tension is the firing frequency of the motor neuron. In general, tension in the motor unit is a monotonically increasing function of firing frequency. Another variable, which appears to be intrinsic to the muscle fibre itself, is a characteristic property known as the "twitch time" (the time required for a twitch to develop and subside), it being an important phenomenon either "fast" or "slow". Generally, the "slow" fibres are tonically active postural-type muscles which maintain fixed tensions over relatively long time period. "Fast" muscles are those related to phasic, i.e., short term kinetic movements. Refractory periods and action potential durations for muscle also tend to parallel twitch times, slow fibres having long recovery times and larger action potentials (which is unimportant because their longer twitch times mask their inexcitability). The duration of the twitch will naturally influence the firing frequency / tension relation.

An understanding of the significance of particular patterns of the impulse activity generated by motor neurons in the spinal cord depends on an understanding of the electromechanical properties of the muscle cells to which these impulses are directed. It is often convenient in studying the physiological properties of muscle to restrict attention to two extreme modes of muscle operation, namely, those cases in which during contraction, the over-all length of the muscle remains constant (isometric contraction) and those in which the load on the muscle remains constant (isotonic contraction). Many natural movements, of course, are intermediate between these, but some simplifications in the treatment of experimental data are possible in these two restricted conditions. From experiments using a combination of isometric, isotonic, and abrupt load reduction contractions, it can be shown that muscle can be represented by a combination of a series elastic component plus a contractile component where only the latter is influenced by motor nerve stimulation. One can further distinguish the active from the passive properties of the contractile component - the latter consist of the mechanical properties of muscle which is not being stimulated, while the former are those which are observed during stimulation, to which the passive properties are usually added. The total tension a muscle can exert is the sum of the active tension and passive tension and varies with muscle length as illustrated in Figure 2.6.

The active tension characteristics are ascribable to sarcomere behaviour, e.g. in the fully contracted and fully stretched states the sarcomere can generate no force, whereas, at about resting length, the sarcomere has its greatest force capability. Although physical damage does not necessarily occur in the fully stretched state, this level of stretch is not attained under normal physiological conditions.

The passive tension curve is obtained by stretching the passive muscle and recording the force. Thus, in the musculoskeletal system the force capability of a muscle is affected by the position of the joint(s) across which the muscles extends. In a dynamic situation the force a muscle can exert is dependent on the velocity of shortening, determined by force - velocity curves for different levels of excitation which illustrate that, for a constant velocity, muscle viscosity is directly proportional to the level of activation. Thus, typical force-velocity curves for skeletal muscle have the characteristics show in Figure 2.7 where $\underline{\alpha}$ max indicates maximum activation and FISO is the maximum isometric force.

2.4. MUSCLE SPINDLES AND OTHER PERIPHERAL NEUROMUSCULAR SENSORY ELEMENTS (MUSCLE PROPRIOCEPTORS)

An important aspect of movement control is the existence of multiple feedback signals originating from the moving limb. The receptors which signal the relevant information to the various control centres under normal conditions are the low-threshold mechano-receptors in skin, joint, tendon and muscle. The responses of these receptors are determined not only by the characteristics of their sensory nerve endings, but also by the transfer functions of their mechanical structures.

The receptors of interest are:

(a) Muscle Spindles These are probably the most important sensory organs for the control of movement and muscular activity. The muscle spindle is a complicated receptor organ whose function will be described later.

(b) Golgi Tendon Organ sensitive to muscle force.

(c) Joint Capsule Receptors There is evidence that these are the receptors which signal actual joint angle and in contrast to the two previous organs, their signals reach consciousness. Although joint receptors may be involved in some reflex activity, little is known about their actions besides their importance for the position sense.

(d) Touch and Pressure Receptors located in skin and muscles. These include free nerve endings and various specialised structures like Pacinian corpuscles.

2.4.1. MUSCLE SPINDLES

Muscle spindles constitute the most important and most numerous sensory organs of vertebrate muscle. There may be, typically, from 50 to 100 such organs in a single mammalian muscle and each may have a quite complex organisation in terms of its motor and sensory innervation. A spindle has a length of several millimeters and is known to consist of several distinct and specialised regions. Within the fluid-filled capsule enclosing the organ are two distinct types of special muscle fibres, the thicker bag fibres, in which there is an enlarged nuclear region, and the thinner, simpler, but more numerous, chain fibres. A highly simplified diagrammatic view of a muscle spindle is shown in Figure 2.8. This shows the central axis of the spindle which consists of a globular nuclear bag region connected to either pole of the spindle by means of a pair of nuclear bag fibres which are themselves typical striated muscle fibres. These nuclear bag fibres are known as "intrafusal fibres", and do not contribute significantly to the development of tension in the muscle. Rather they appear to be motor fibres related solely to control within the spindle itself. Chain fibres lack the nuclear bag region. They have smaller diameters and are connected in series and in parallel with the nuclear bag fibres. All spindles have at least one nuclear bag fibre, but may not have any chain fibres. The motor nerve supplying the intrafusal fibres have their cell bodies situated in the spinal cord. Their axons have a diameter somewhat smaller than that of the axons of the nerve supply to the extrafusal muscle fibres. To distinguish these two classes of motor cells, axons, and muscle fibres, the terms "gamma" (referring to the spindle motor system) and "alpha" (referring to the main muscle motor system) are used.

Two distinctly different sensory fibres arise from the muscle spindle; the larger of these, the primary ending (also referred to as the "annulospiral ending" or type Ia afferent) apparently can have multiple origins in the spindle, i.e. can have its endings imbedded in both the nuclear region of the bag fibre and the chain fibres. Each spindle usually has a single primary ending, and this afferent innervates but a single spindle. The secondary endings (also called "flower-spray" or type II afferents) arise from the nuclear chain fibres, with occasional branches to the bag fibres, but these never terminate at the bag region. Secondary fibres may be traced back to more than one spindle. The axons arising from secondary endings are smaller in diameter than those from primary endings. With regard to the motor innervation of the intrafusal fibres, considerable disagreement exists with regard to their anatomical distribution. It is agreed that two types of small diameters "gamma" fibres contribute to the motor innervation as well as a larger size fibre of the "Beta" class. The Beta fibres and one group of gamma fibres terminate in "plate" type endings, the other gamma fibres terminate in "trail" type endings. The actual distribution of these endings over nuclear bag and nuclear chain fibre is controversial. However, agreement has been reached about their functional significance. The gamma plate fibres and beta fibres are believed to be associated with the control of the "static" response of spindle primary and secondary endings, the trail endings with the control at the "dynamic" response, acting almost exclusively on the primary endings.

The sensory fibres leaving the muscle spindle enter the spinal cord and form direct synaptic connections with the motor neurons supplying the muscle in which the spindle is inbedded. The nature of this monosynaptic coupling is such that increases in spindle firing frequency, generate increases, in the corresponding motor neuron firing frequency, and hence produce increasing motor unit contractile forces or resistance to stretch. The spindle axon also makes more complex connections which effectively inhibit the antagonist pool of motor neurons. This reveals the underlying importance of the spindle/motor neuron feedback loop in stabilising the length of the muscle. Influences tending, for example, to increase the length of the muscle, such as sudden increases in load, augment spindle activity which reflexively generates motor command signals tending to resist changes in length.

2.4.2. GOLGI TENDON ORGANS

Unlike the muscle spindle which is connected parallel with the muscle fibres and responde to muscle length, the Golgi Tendon Organ is connected in series with the muscle and respond to tension. It is located in the tendinous insertion of the muscle. As yet there is no clear-cut evidence that the afferents arising from these receptors form distinct classes; therefore, we can consider only one single class of afferent fibres, the type I_b axons. Although the threshold of response of this sensory organ is indeed high to passive muscle stretch, under conditions of active muscle contraction, activation of a single-muscle fibre is sufficient to elicit tendon organ response. An individual tendon organ is connected in series with about five to twenty-five fibres from several different motor units. These in turn are connected in parallel with the other extrafusal fibres of the muscle that are joining at the insertion. There is no voluntary control of the tendon organ that is the equivalent of the spindle's fusimotor system.

The physiological role of the Golgi tendon organ is still not clear. Between them, the spindle and the tendon organs generate a continuous stream of length and tension information to the spinal cord and higher centres.

2.5 GENERAL FEATURES OF THE SPINAL CORD ORGANISATION

The central nervous system consists of the brain and the spinal cord which is attached to it in the region of the medulla. The spinal cord, throughout its length, is surrounded by the bony vertebral column a union of separate bones which effectively define a sequence of spinal cord segments. At the side of each vertebrae there are two openings through which all the sensory nerves enter and all the motor nerves leave that section of the cord. A cross section through the cord shows the internal pattern to consist of butterfly shaped grey matter surrounded by white matter. In the dorsal horn of the grey matter are cells primarily concerned with or under the domination of sensory input pathways from the dorsal root fibres. (The cell bodies of all incoming sensory axons are actually outside the CNS, strung along the vertebrae in swellings, or ganglia, at each spinal level). From some of these cells arise axons which pass into the white matter, aggregating to form pathways connecting local spinal regions (propriospinal pathways) or pathways relaying information to higher centres. At more intermediate regions of the grey matter are located interneurons, cells lying in the integrative pathways between sensory input and motor output. They may receive input from the periphery (for example, from skin receptors, joint position receptors, or muscle receptors), as well as from the sensory centres in the midbrain (e.g. the vestibular system), the higher integrative centres (e.g. the cerebellum) or the cerebral cortex. In addition to propriospinal pathways located in the white matter, a number of very important long pathways, both ascending and descending occupy standard locations in the cord. They are the "dorsal column" pathways which carry touch, pressure, and joint position information to cortical levels; the "spinocerebellar" pathways which carry a variety of proprioceptive information (including that from spindles) to the cortex of the cerebellum; and the "spinothalamic" pathways which carry touch, pressure, pain, and temperature information to the midbrain, the thalamus, and, ultimately, the cerebral cortex.

There are a considerable number of descending pathways whose functions are not clearly understood. The most important of these are the "vestibulospinal" path through which, apparently, spinal motor systems are enabled to compensate for motions of the body in space; the "cerebellospinal" path which therefore provide a closed spinocerebellar loop; the "subrospinal" paths from the red nucleus in the midbrain, an area implicated in the control of gamma activity; the "reticulospinal" paths from the midbrain and medullary reticular formation, a complex integrating centre also concerned with gamma control and the "corticospinal" paths arising from the sensory and motor cortices and concerned with control of sensory and motor functions at the spinal level.

2.6. ORGANISATION OF THE NEUROMUSCULAR SYSTEM

The principal anatomical pathways of a neuromuscular control system are outlined in Figure 2.9. Efferent signals proceed from the spinal cord to the muscle and its receptors (muscle spindles) via the ventral spinal root. Incoming sensory axons arrive via the dorsal root, while outgoing motor fibres leave via the ventral root.

There are two main pathways from higher centres to peripherals descending the spinal cord, namely alpha and gamma efferent paths. Alpha motoneuron in the spinal cord receives signals not only from alpha efferents but also from peripheral receptors via afferent fibres. Impulse discharges of alpha motoneuron makes the muscle contract and generate movements. As already said, muscle spindle, located in parallel with the muscle fibres, detect the length of muscle and the rate its change. Muscle spindle has two types of afferent sensory fibres, the group I_a and group II afferent fibres. Note that the sensitivities of muscle spindle are regulated by gamma system impulses.

Tendon organ, located in the muscle tendon, detects the muscle tension and afferent fibre, group I_b fibre, arises in tendon organ.

I_a fibres excite their own motoneuron and inhibit the motoneuron of antagonist muscle. I_b fibres inhibit the motoneuron of synergistic muscle and excite the motoneuron of antagonist muscle.

Type II fibres are known to produce some reflex patterns for example, they produce excitation of flexor motoneurons while inhibiting extensor motoneuron (flexor reflex). Thus, alpha motoneuron (spinal ventral horn cell) receives a consequence of signals coming from both higher nervous centres and from sensory receptors and muscle fibres, innervated by alpha motoneuron, work together to provide fine control of limb movement.

CHAPTER 2

LIST OF FIGURES

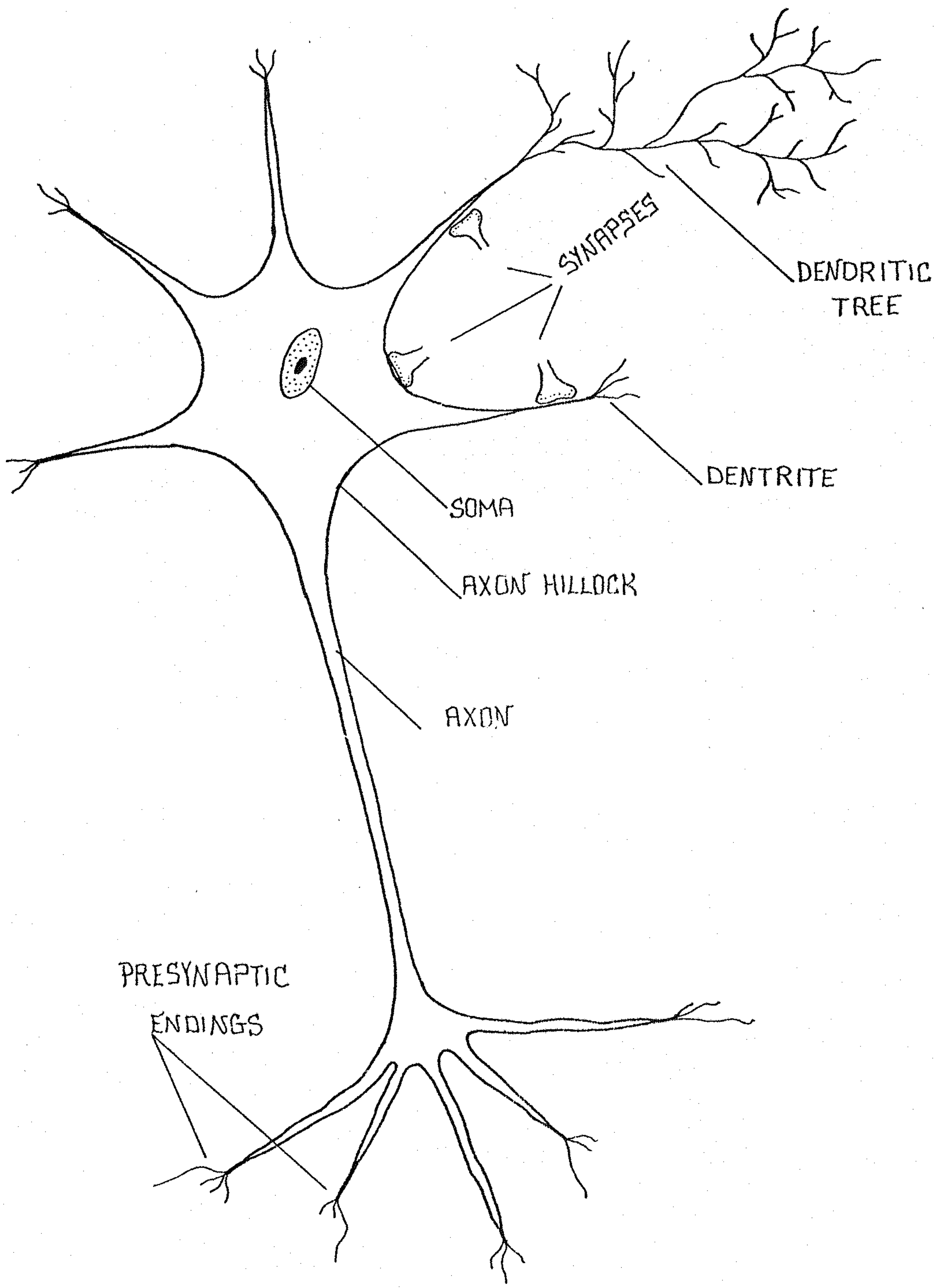


FIG. 2.1 A TYPICAL MOTOR NEURON

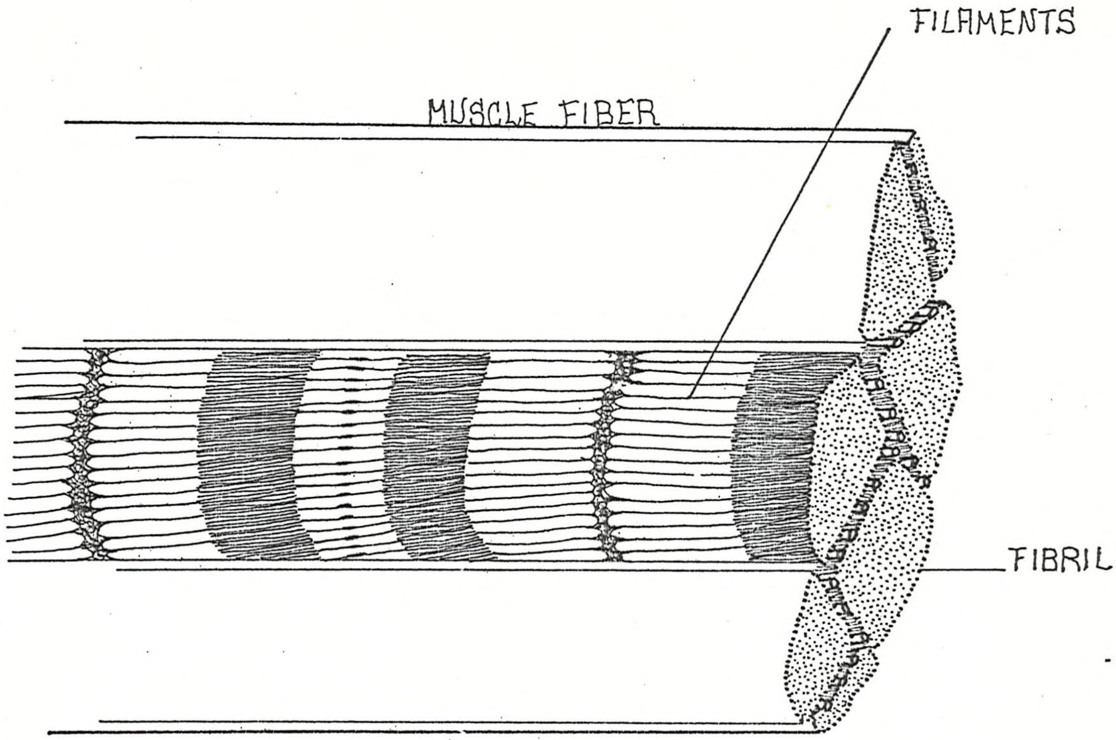


FIG. 2.2 STRUCTURE OF SKELETAL MUSCLE

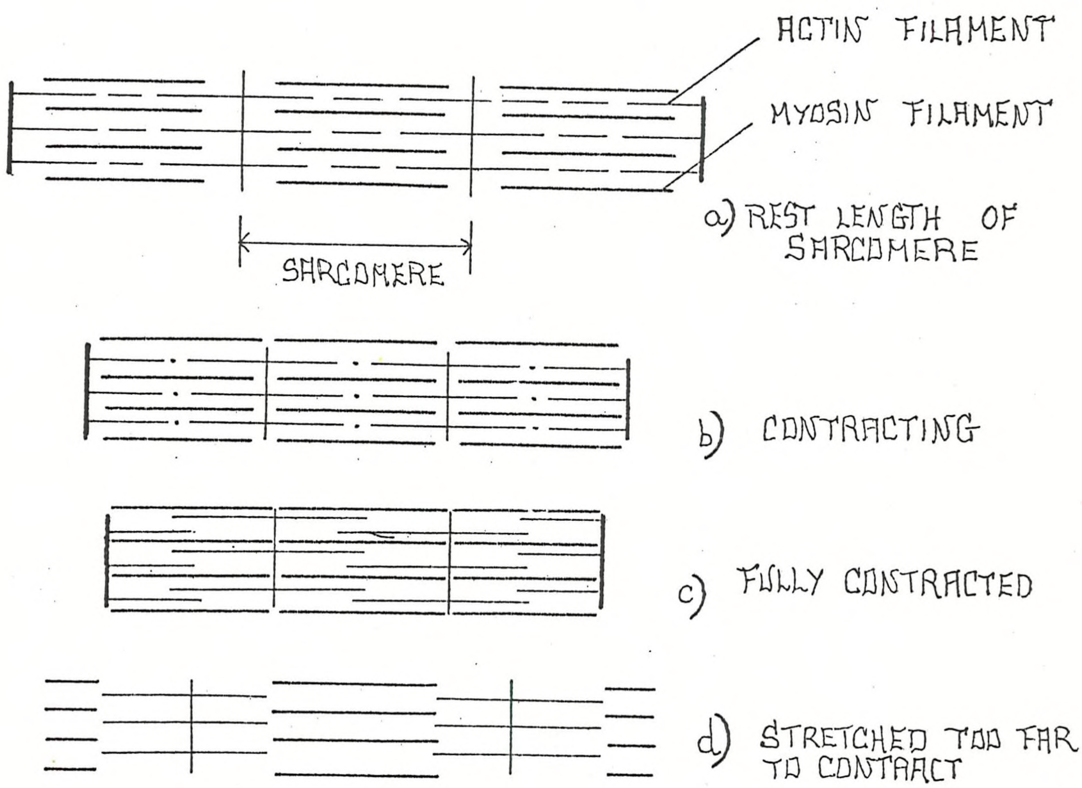


FIG. 2.3 SCHEMATIC OF SLIDING FILAMENTS OF SKELETAL MUSCLE

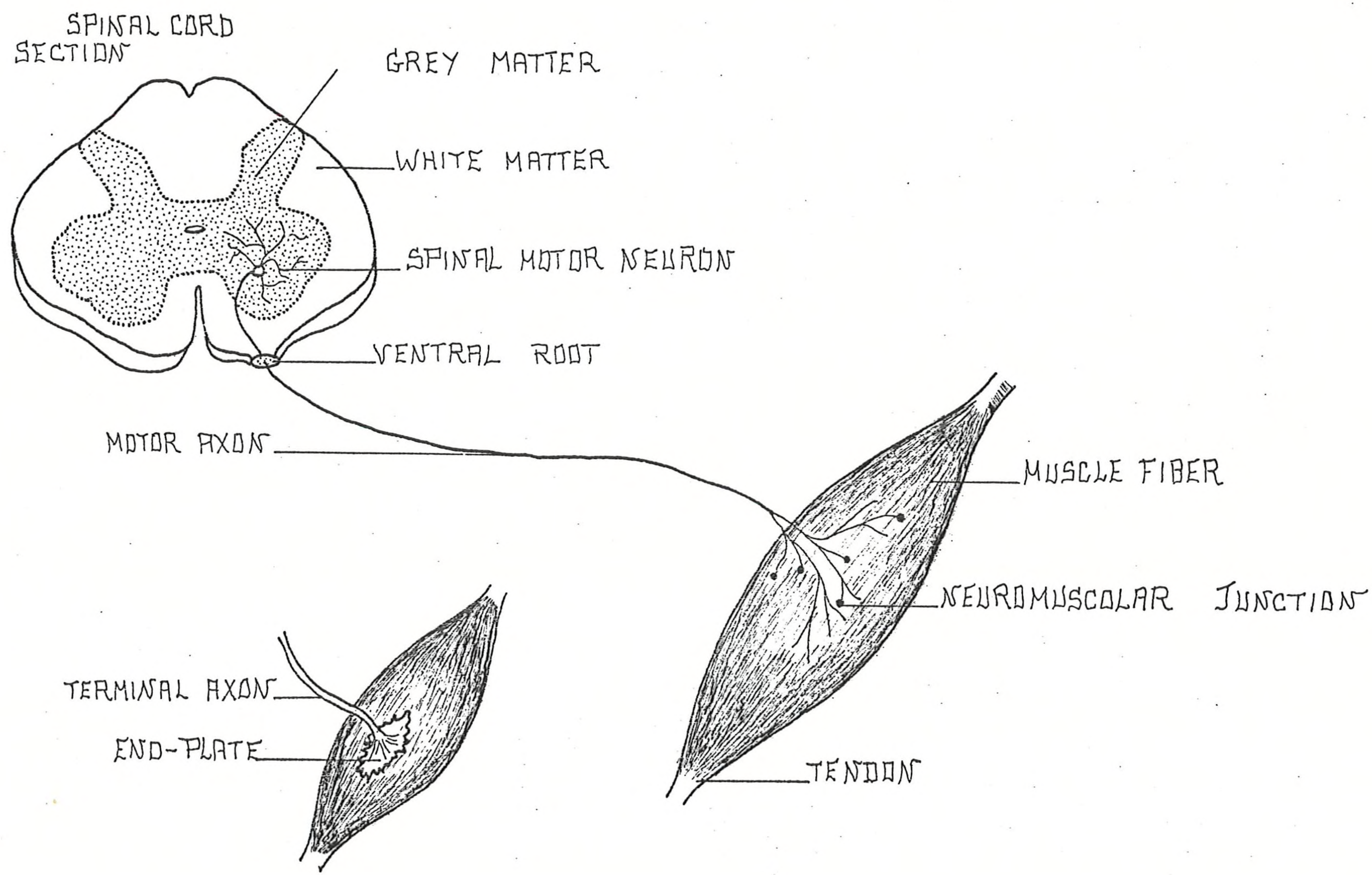


FIG. 2.4 MOTOR UNIT COMPONENTS

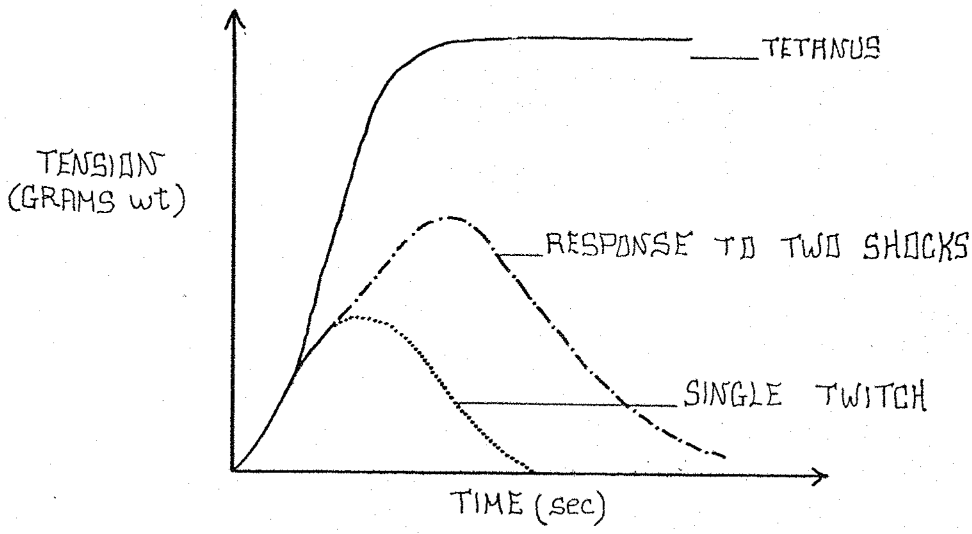


FIG. 2.5 MUSCLE TWITCH RESPONSE

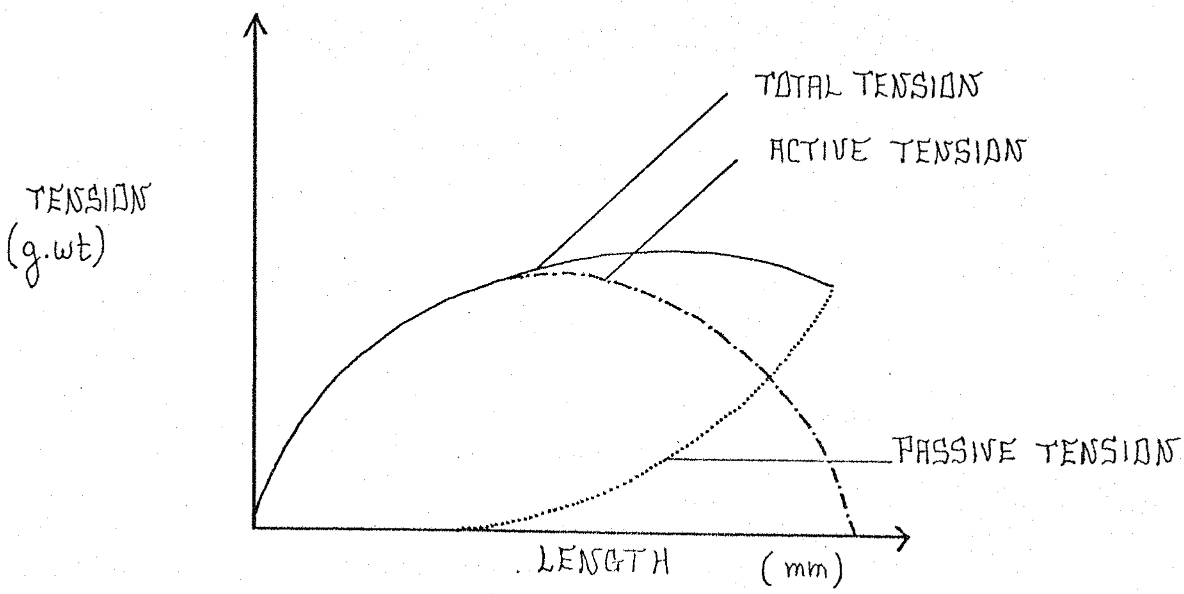


FIG. 2.6 LENGTH/TENSION RELATION FOR SKELETAL MUSCLE

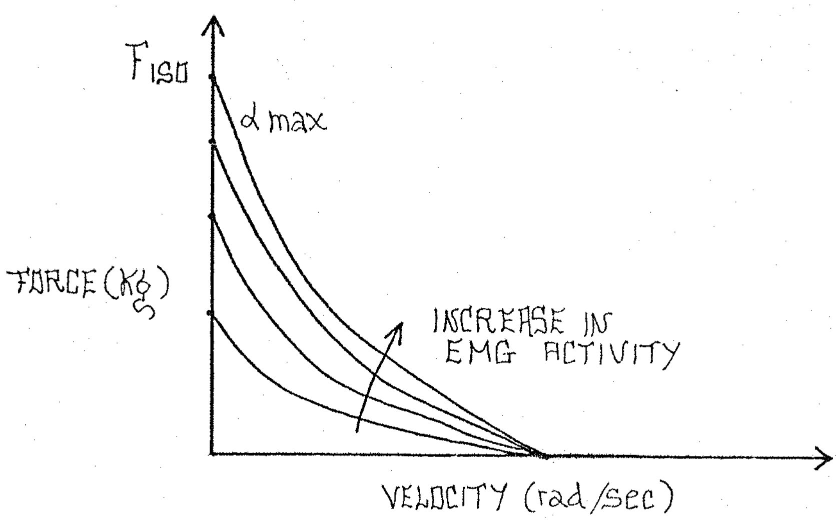


FIG. 2.7 FORCE-VELOCITY RELATION OF HUMAN SKELETAL MUSCLES

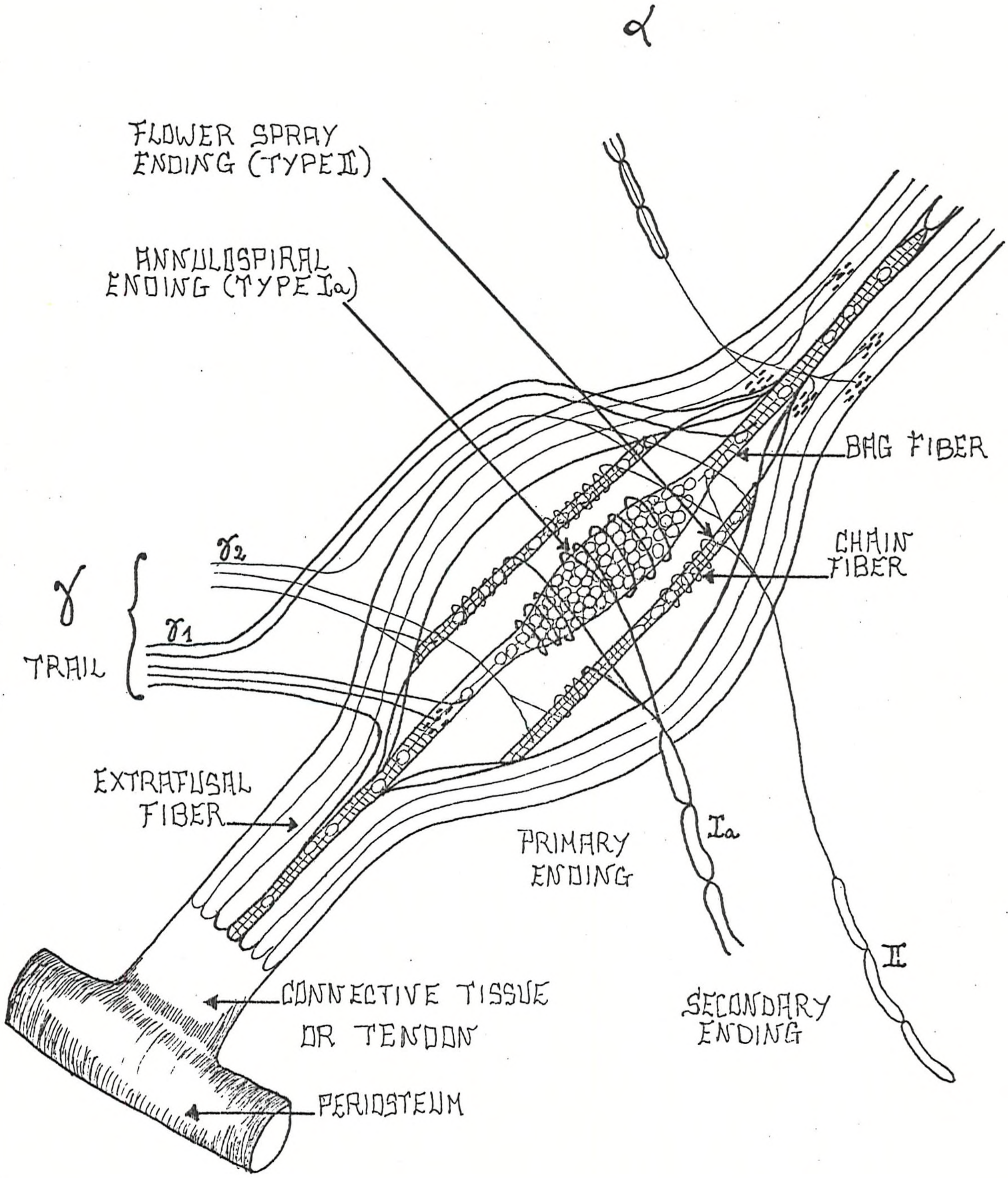


FIG. 2.8 DIAGRAM OF MUSCLE SPINDLE

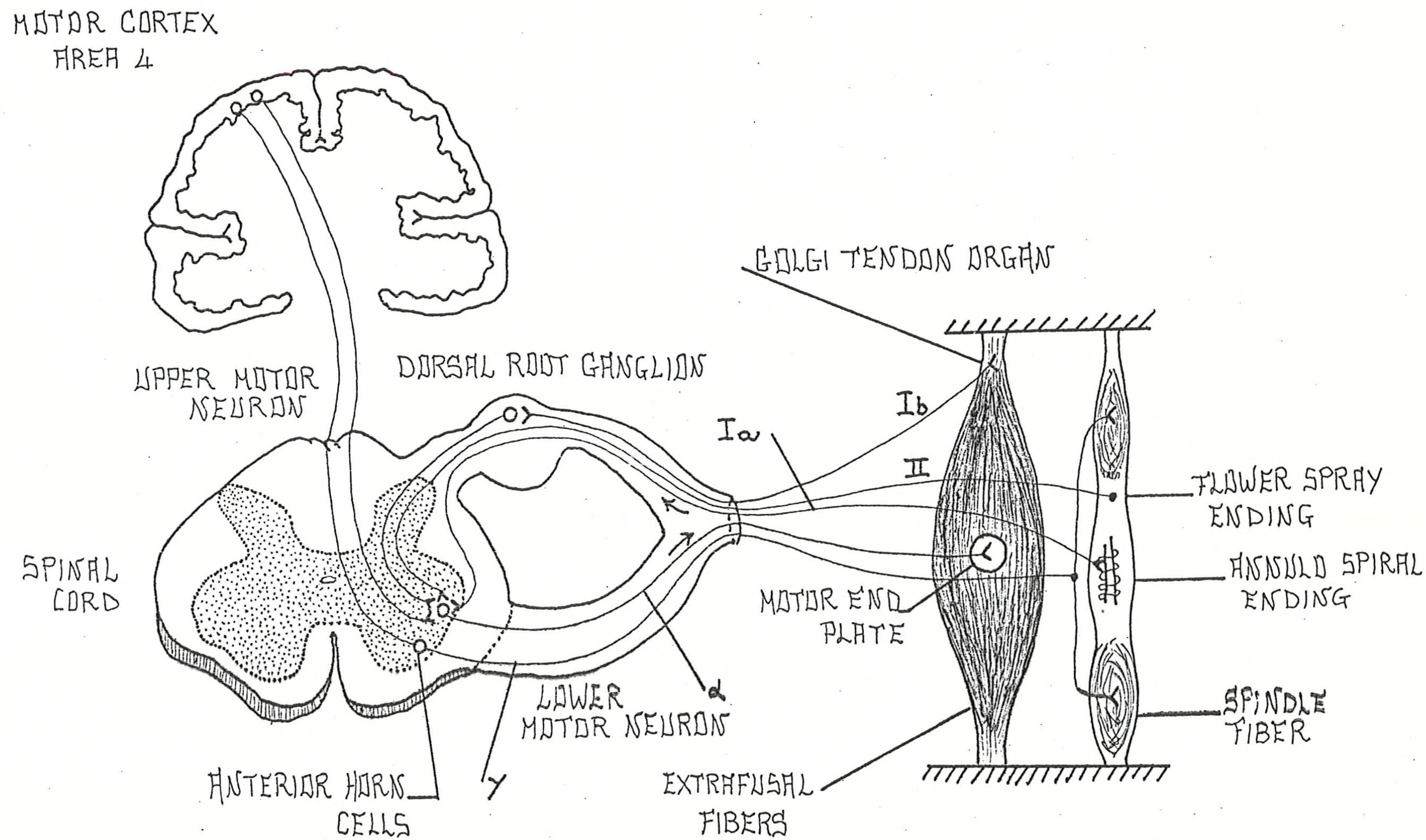


FIG. 2.9 PHYSIOLOGICAL CONTROL OF SKELETAL MUSCLE

CHAPTER 3

THE UPPER-EXTREMITY SYSTEM

3.1. INTRODUCTION

Man is a biped mammal whose upper limbs differ from his lower limbs in that they are adapted to mobility and prehension rather than stability and locomotion. The skeletal and neuromuscular structures of shoulder, arm, forearm, and hand constitute the upper extremity.

Clearly, this is a biological system of such complexity that, if it were to be considered from all possible points of view, volumes would be required to describe its anatomy, physiology, and manifold performances. However, anatomical, physiological and performance data have been selected only where they are pertinent for the present work. Most of the material reported in this Chapter has been summarised from WELLS [1971] and TAYLOR [1968].

3.2. THE SEGMENT, ANGLES, AND AXES OF THE UPPER-EXTREMITY SYSTEM

It is important at the outset to define carefully the principal component motions of the upper-extremity system. The motion of each part upon its proximal joint may be described with respect to the principal planes which intersect at the joint.

Figure 3.1 shows the standard position and body reference planes: frontal, sagittal and horizontal. In this standard position, the trunk is erect, the arms hang with their axes vertical, the elbows flexed to 90 degrees and the wrist planes are vertical to assume the "shake-hands" position.

Table 3.1 lists the bones, angles, and axes of the system with chief reference to this standard position, although it is evident that, as the parts move, their reference axes should be shifted so that the angular motions made by each segment are properly referred to the new position of the joint with which the segment articulates. The angular movements are further displayed in Figure 3.2. The shoulder-on-chest, arm-on-shoulder, and hand-on-wrist actions take place through two angles, as if these articulations were moving about a universal joint.

3.3. THE FUNCTIONAL ANATOMY OF THE SHOULDER AND ARM

3.3.1. THE SHOULDER AND ARM BONES

The shoulder is articulated with the thorax anteriorly by the clavicle, which forms the sternoclavicular and costoclavicular joints with the breastbone and with the first rib. These are the only direct skeletal attachments of the shoulder, and therefore of the arm, to the torso. The clavicle then extends laterally and, passing superior and posterior to the coracoid process, articulates with the acromion of the scapula to form acromioclavicular joint. The body of the scapula hangs in a muscular suspension on the postero-lateral aspect of the torso and, in addition, receives lateral support from the clavicle at the acromioclavicular joint. The body of the scapula is flat, and triangular with two lateral projections, the acromion and the coracoid process which are important attachments points for muscles and ligaments. The humerus, with its rounded proximal head articulating in the glenoid cavity of the scapula, complete the shoulder-arm complex.

Both the shoulder and arm may, to a first approximation, be considered as linked segments, each having a two-angle motion field about its proximal articulation.

3.3.2. THE SHOULDER-ARM MUSCULATURE

The muscular anatomy of the shoulder arm system is highly complex, and a considerable degree of simplification is necessary to gain an elementary understanding of the basic actions of the muscles as well as of their interactions in the various phases of motion. For clarification, the schematic view of Figure 3.3. show the skeletal members and principal muscle groups. The muscles can be divided into three groups: those joining scapula to thorax, those joining arm to scapula, and those running from arm to thorax. The neuromuscular organisation is capable, within the limits set by the musculoskeletal mechanism, of performing shoulder motions alone, arm motion alone or co-ordinated motions of both. The last is the normal mode.

3.3.3. THE SHOULDER JOINT (GLENOHUMERAL ARTICULATION)

3.3.3.1. STRUCTURE

The shoulder joint is formed by the articulation of the spherical head of the humerus with the small, shallow, somewhat pear-shaped glenoid fossa of the scapula (Figure 3.4). It is a ball-and-socket joint. The structure of the joint and looseness of the capsule (permitting between 2.5 and 5 cm. of separation between the two bones) account for the remarkable mobility of the shoulder joint.

The joint is completely enveloped in a loose sleevelike articular capsule which is attached proximally to the circumference of the glenoid cavity, and distally to the anatomic neck of the humerus. The capsuleis reinforced both by ligaments and the muscle tendons. The latter are particularly important in preserving the stability of the joint. Apparently they do not prevent downward dislocation however.

3.3.3.2. MOVEMENTS

The movements of the humerus are as follows:

Flexion and hyperflexion (Figure 3.5-A)

A forward-upward movement in a plane at right angles to the plane of the scapula. If the movement exceed 180 degrees, it is hyperflexion.

Extension (Figure 3.5-B)

Return movement from flexion.

Hyperextension

A backward movement in a plane at right angles to the plane of the scapula.

Abduction (Figure 3.5-C)

A sideward upward movement in a plane parallel with the plane of the scapula (some authorities interpret this movement as the sideward movement of the arm away from the body, thus including the action of the muscle gridle).

Adduction (Figure 3.5-D)

Return movement from abduction.

Outward Rotation

A rotation of the humerus around its mechanical axis so that, when the arm is in its normal resting position, the interior aspect turns laterally.

Inward Rotation

A rotation of the humerus around its mechanical axis so that when the arm is in its normal resting position, the interior aspect turns medially.

Horizontal Flexion

A forward movement of the abducted humerus in a horizontal plane (i.e. from a plane parallel to the plane of the scapula to a plane at right angles to it).

Horizontal extension

A backward movement of the flexed humerus in a horizontal plane (i.e. from a plane at right angles to the plane of the scapula to a plane parallel to it).

Circumduction

A combination of flexion, abduction, extension, hyperextension and adduction, performed sequentially in either direction so that the extended arm describes a cone, and the finger tips a circle.

3.3.3.3. MUSCLES

The muscles of the shoulder joint may be classified according to their position in relation to the joint.

Superior

Middle deltoid - Supraspinatus

Inferior

Latissimus dorsi-Teres major - Long head of triceps (primarily a muscle of the elbow joint).

Anterior

Pectoralis major - Coracobrachialis - Anterior Deltoid - Subscapularis - Biceps (primarily a muscle of the elbow joint).

Posterior

Posterior deltoid - Infraspinatus - Teres mionr.

Table 3.2 summarises the muscles acting at the shoulder joint.

3.3.4. THE SHOULDER GIRDLE (ACROMIOCLAVICULAR AND STERNOCLAVICULAR ARTICULATIONS

3.3.4.1. STRUCTURE OF ACROMIOCLAVICULAR ARTICULATION

The articulation between the acromion process of the scapula and the outer end of the clavicle belongs to the diarthrodial classification, (Figure 3.6-a). Within this group it is further classified as an irregular (arthrodial) joint. A small wedge-shaped disk may be found between the upper part of the joint surfaces, but this is frequently absent.

The articular capsule is strengthened above by the acromio-clavicular ligament, and behind by the aponeurosis of the trapezius and deltoid muscles. The clavicle is further stabilised by means of the coracoclavicular ligament, which, as the name suggests, binds the clavicle to the coracoid process.

3.3.4.2. STRUCTURE OF STERNOCLARICULAR ARTICULATION

The sternal end of the clavicle articulates with both the sternum and the cartilage of the first rib. It is classified as a double arthrodial joint because there are two joint cavities, one one either side of the articular disk. (Figure 3.6-b) This round flat disk of white fibrocartilage is attached above to the upper and posterior border of the articular surface of the clavicle, and below to the cartilage of the first rib near its junction with the sternum. The movements of the clavicle at this joint are as follows: elevation and depression which occur approximately in the frontal plane about a sagittal - horizontal axis; horizontal forward - backward movements which occur in the horizontal plane about a vertical axis; and rotation forward and backward, very limited movements which occur approximately in the sagittal plane about the bone's own longitudinal axis. (In forward rotation the top of the clavicle revolves forward - downward). The sternoclavicular articulation is of great importance in the movements of the shoulder girdle and of the arm as a whole. It permits limited motion of the clavicle in all three planes and because of the bone's attachment to the scapula at its distal end, is partially responsible for the latter's movements.

3.3.4.3. MOVEMENTS

It is customary to define the movements of the shoulder girdle in terms of the movements of the scapulae. It is well to emphasise the fact that every movement of the scapula involves motion in both joints. The movements of the shoulder girdle, expressed in terms of the composite movements of the scapula, are as follows:

Elevation (Figure 3.7-A)

An upward movement of the scapula with the vertical border remaining approximately parallel to the spinal column.

Depression

The return from the position of elevation. There is no depression below the normal resting position.

Abduction or Protraction (Figure 3.7-B)

A lateral movement of the scapula away from the spinal column, with the vertebral border remaining approximately parallel to it.

Adduction or Retraction

A medial movement of the scapula toward the spinal column combined with a reduction of lateral tilt.

Upward Tilt (Figure 3.7-C)

A turning of the scapula on its frontal - horizontal axis so that the posterior surface faces slightly upward and the inferior angle protrudes from the back.

Reduction of Upward Tilt

The return movement from upward tilt.

Upward Rotation (Figure 3.7-D)

A rotation of the scapula in the frontal plane so that the glenoid fossa faces somewhat upward.

Downward Rotation

The return from the position of upward rotation.

3.3.4.4. MUSCLES

The muscles of the shoulder girdle are classified as anterior or posterior muscles, according to their location on the trunk:

Anterior

Subclavius - Pectoralis minor - Serratus anterior.

Posterior

Levator scapulae - Trapezius - Rhomboids.

Table 3.3 summarises the muscles acting on the shoulder girdle.

3.3.5. MOVEMENTS OF THE ARM ON THE TRUNK

The movements of the arm on the trunk do not take place at the shoulder joint alone, but they involve movement of the shoulder girdle at both the acromioclavicular and sternoclavicular joints. In order to analyse correctly the movements of the upper extremity, it is necessary to understand the co-operative action of the shoulder joint and shoulder girdle. The fundamental movements of the arm, analysed in terms of shoulder joint and shoulder girdle action, are stated below:

Sideward Elevation

Shoulder joint: abduction of the humerus; outward rotation if the palms are turned to face each other in the overhead position.

Shoulder girdle: upward rotation and slight elevation of the scapula, particularly after arm passes above the horizontal.

Sideward Depression

S.J.: Adduction of the humerus, reduction of outward rotation.

S.g.: Downward rotation of the scapula.

Forward elevation

S.J.: Flexion of the humerus; slight outward rotation; abduction if scapula is laterally tilted (in which case glenoid fossa will be facing partly forward).

S.g.: Slight upward rotation of the scapula; abduction and lateral tilt, unless inhibited.

Forward - Upward elevation

(From the horizontal to the vertical and beyond)

S.j.: Flexion and possibly hyperflection of the humerus; slight outward rotation.

S.g.: Upward rotation and some elevation of the scapula; reduction of abduction and lateral tilt.

Forward - Downward depression

(From the vertical to the starting position)

S.j.: Extension of the humerus; adduction; reduction of outward rotation.

S.g.: Downward rotation of the scapula; passes through position of abduction and lateral tilt, unless prevented by strong action of the scapular adductors.

Backward elevation

S.j.: Hyperextension of the humerus

S.g.: Upward tilt of the scapula; elevation if movement is carried to extreme.

Outward rotation

S.j.: Outward rotation of the humerus; tendency toward hyperextension.

S.g.: Adduction of the scapula; reduction of any lateral tilt which may have been present.

Inward rotation

S.j.: Inward rotation of the scapula

S.g.: Abduction and lateral tilt of the scapula; tendency toward elevation.

Horizontal forward - Swing from position of sideward elevation

S.j.: Horizontal flexion of the humerus; slight inward rotation.

S.g.: Abduction and lateral tilt of the scapula; unless inhibited.

Horizontal sideward - Backward swing from positon of forward elevation

S.j.: Horizontal extension of the humerus; slight outward rotation.

S.g.: Adduction and reduction of lateral tilt of the scapula.

CHAPTER 3

LIST OF TABLES

LIST OF FIGURES

TABLE 3.1

PRINCIPAL COMPONENTS AND MOTIONS

OF THE UPPER EXTREMITY SYSTEM

SEGMENT	BONES	AXIS	ANGULAR MOVEMENTS
SHOULDER	CLAVICLE, SCAPULA	OH	FLEXION - EXTENSION, ELEVATION - DEPRESSION, ROTATION NEGLECTED
ARM	HUMERUS	H.E	FLEXION - EXTENSION, ELEVATION - DEPRESSION MEDIAL AND LATERAL DEPRESSION
FOREARM	RADIUS, ULNA	EW	FLEXION - EXTENSION, PRONATION - SUPINATION (WRIST ROTATION)
HAND	CARPALS, METACARPALS, PHALANGES	WK	RADIAL AND ULNAR FLEXIONS. VOLAR AND DORSAL FLEXIONS

TABLE 3.2

MUSCLES ACTING AT THE SHOULDER JOINT

X Definitely active in the movement

O An assistant or emergency muscle

? Action uncertain in the movement.

MUSCLES AND LOCATIONS	UPPER ARM								
	FLEXION	EXTENSION	HYPEREXTENSION	ABDUCTION	ADDUCTION	ROTATION OUTWARD	ROTATION INWARD	HORIZONTAL FLEXION	HORIZONTAL EXTENSION
SUPERIOR									
Middle deltoid				X					X
Supraspinatus				X		?			
INFERIOR									
Latissimus dorsi		X	X		X		X		X
Teres major		X	X		X		X		X
ANTERIOR									
Anterior deltoid									
Upper	X			?			X	X	
Lower					O				
Pectoralis major									
Upper or clavicular	X						X	X	
Lower or sternal		X			X		X	X	
Coracobrachialis	O				O		O	X	
Subscapularis							X	X	
POSTERIOR									
Posterior deltoid		X			O (Lower fibres)	X			X
Infraspinatus						X			X
Teres minor						X			X
PRIMARY ACTIONS AT OTHER JOINTS									
Biceps brachii	O				O			O	
Triceps brachii (Long head)		O	O		O	O		O	

TABLE 3.3

MUSCLES ACTING ON THE SHOULDER GIRDLE

X Definitely active in the movement

O An assistant or emergency muscle

MUSCLES AND LOCATION	SCAPULA							
	ELEVATION	DEPRESSION	ABDUCTION	ADDUCTION	UPWARD TILT	REDUCTION OF UPWARD TILT	ROTATION UPWARD	ROTATION DOWNWARD
ANTERIOR								
Subclaris		clavicle						
Pectoralis minor		X	X		X			X
Serratus Anterior								
Upper			X				O	
Lower			O				X	
POSTERIOR								
Levator scapulae	X							X
Trapezius								
Part I	X							
Part II	X			O			X	
Part III				X				
Part IV		X		O			X	
Rhomboids								
Major and minor	X			X				X

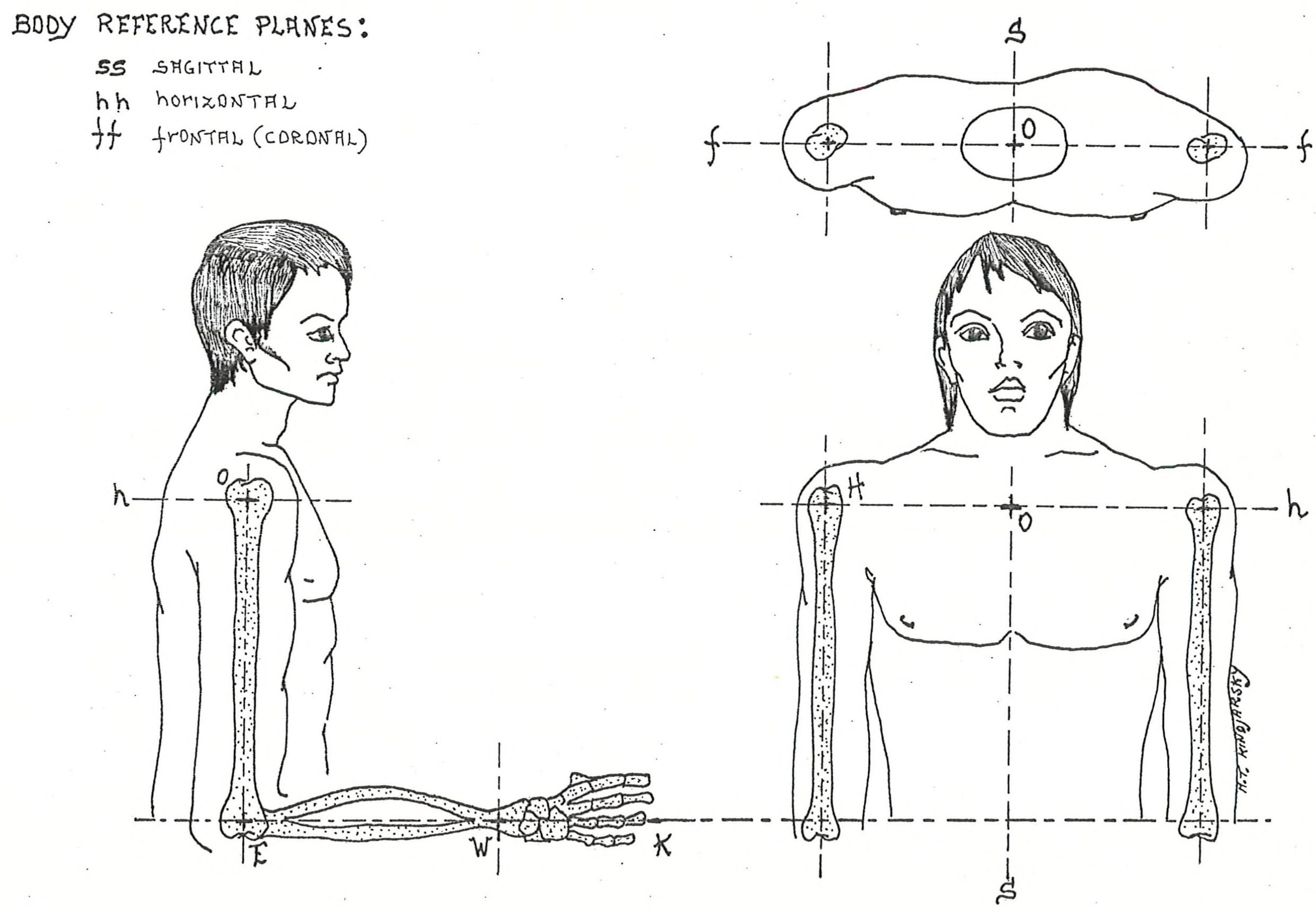


FIG.3.1 UPPER EXTREMITY SYSTEM IN STANDARD POSITION

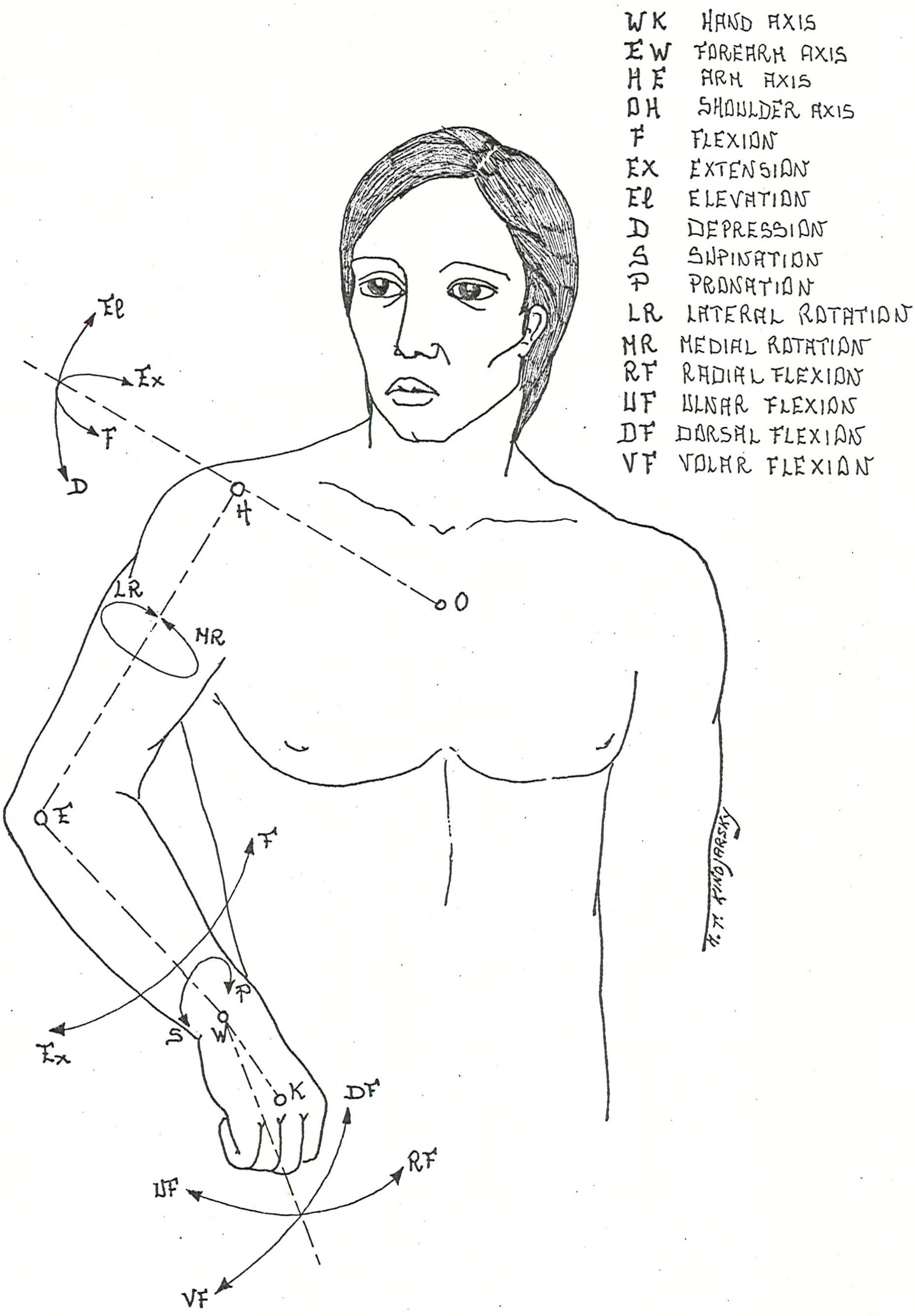


FIG. 3.2 MAJOR MOTIONS OF THE UPPER EXTREMITY SYSTEM

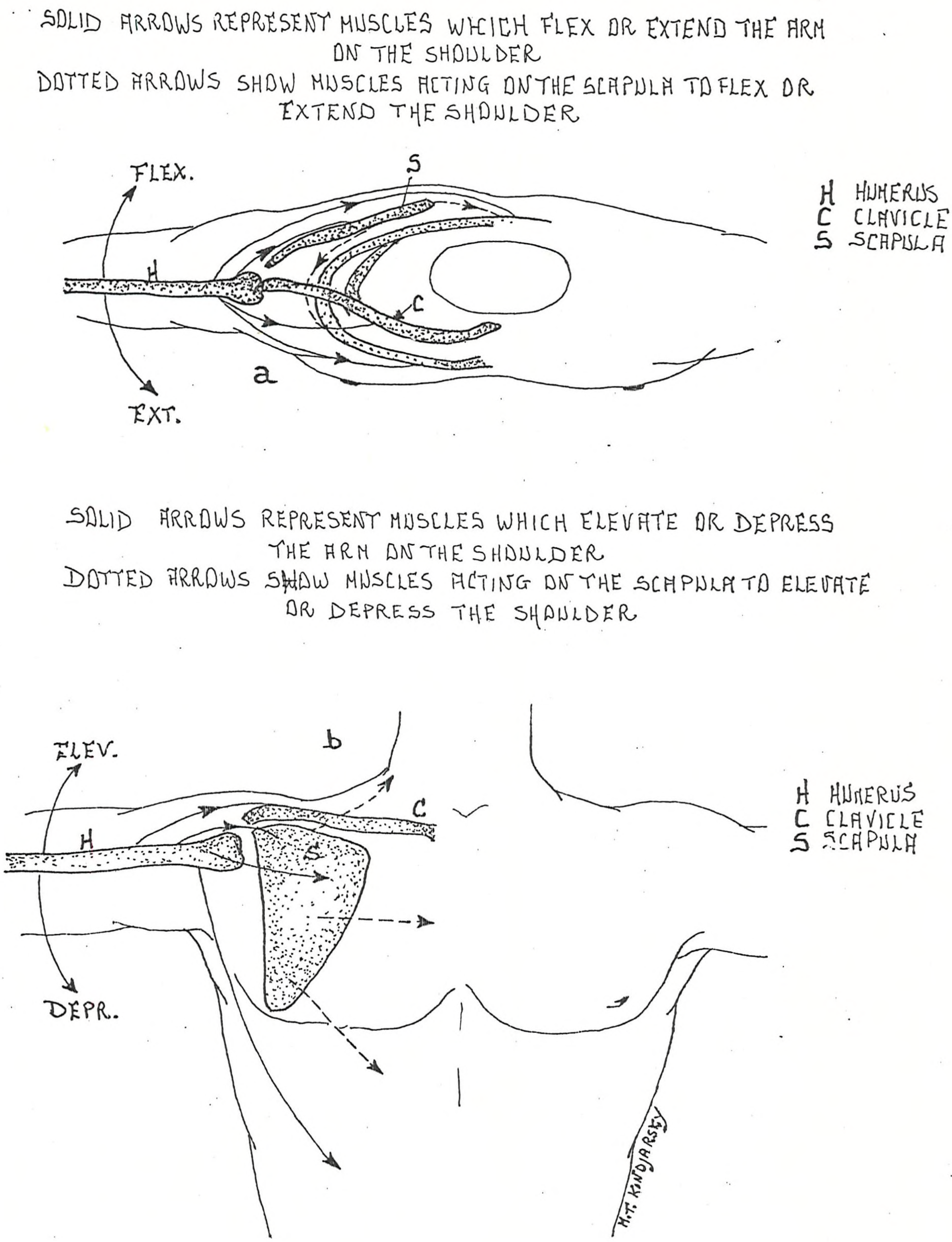


FIG. 3.3 SCHEMATIC VIEW OF MUSCLES ACTING ON THE SHOULDER-ARM SYSTEM

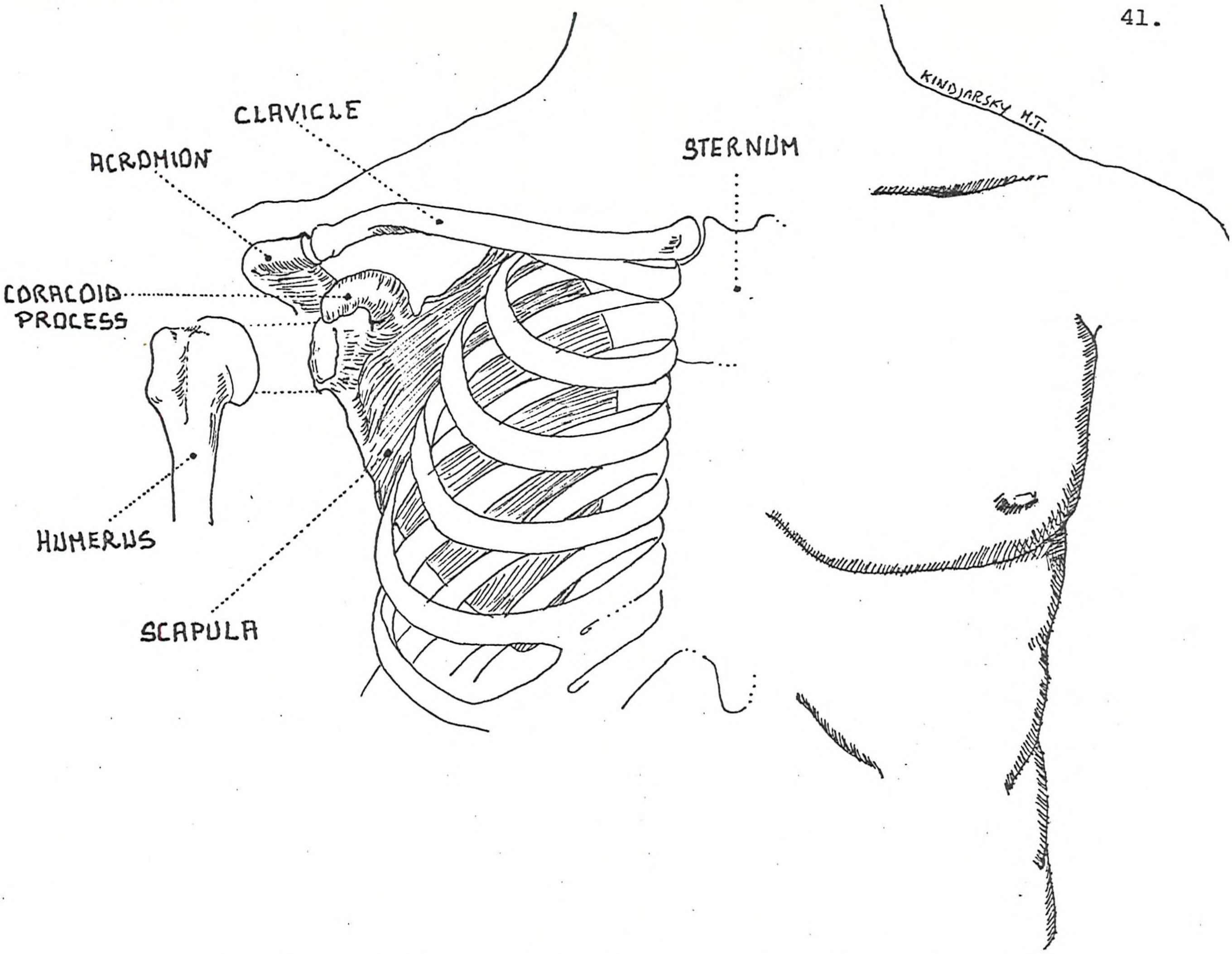


FIG. 3.4 SHOULDER JOINT AND SHOULDER GIRDLE

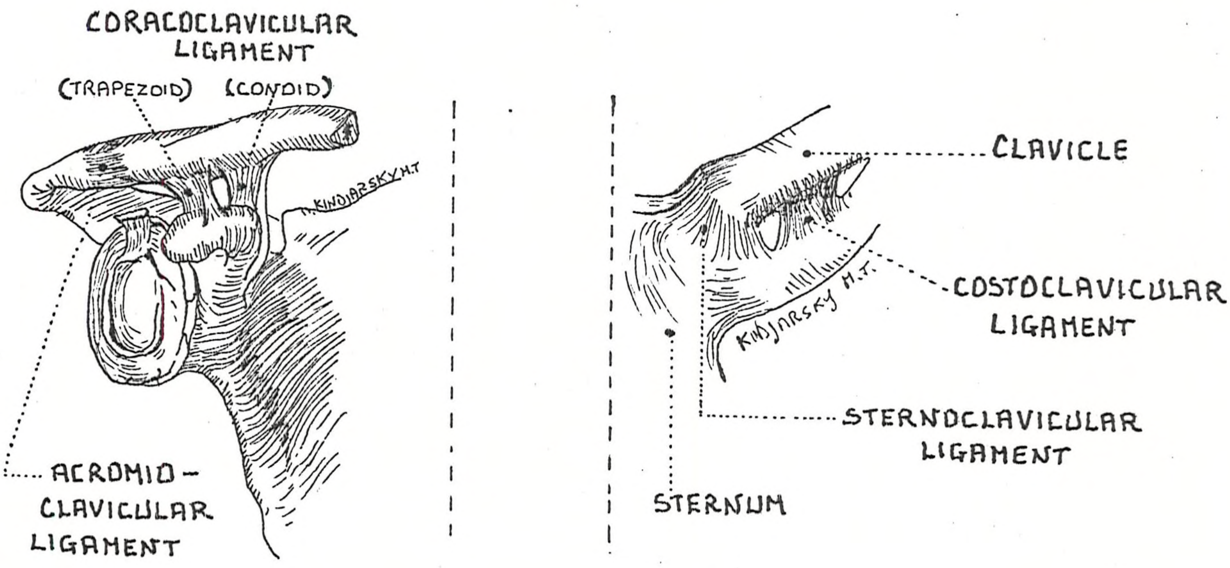


FIG. 3.6 a) ACROMIOCLAVICULAR ARTICULATION(Ant. view)

b) STERNOCLAVICULAR ARTICULATION (Ant. view)

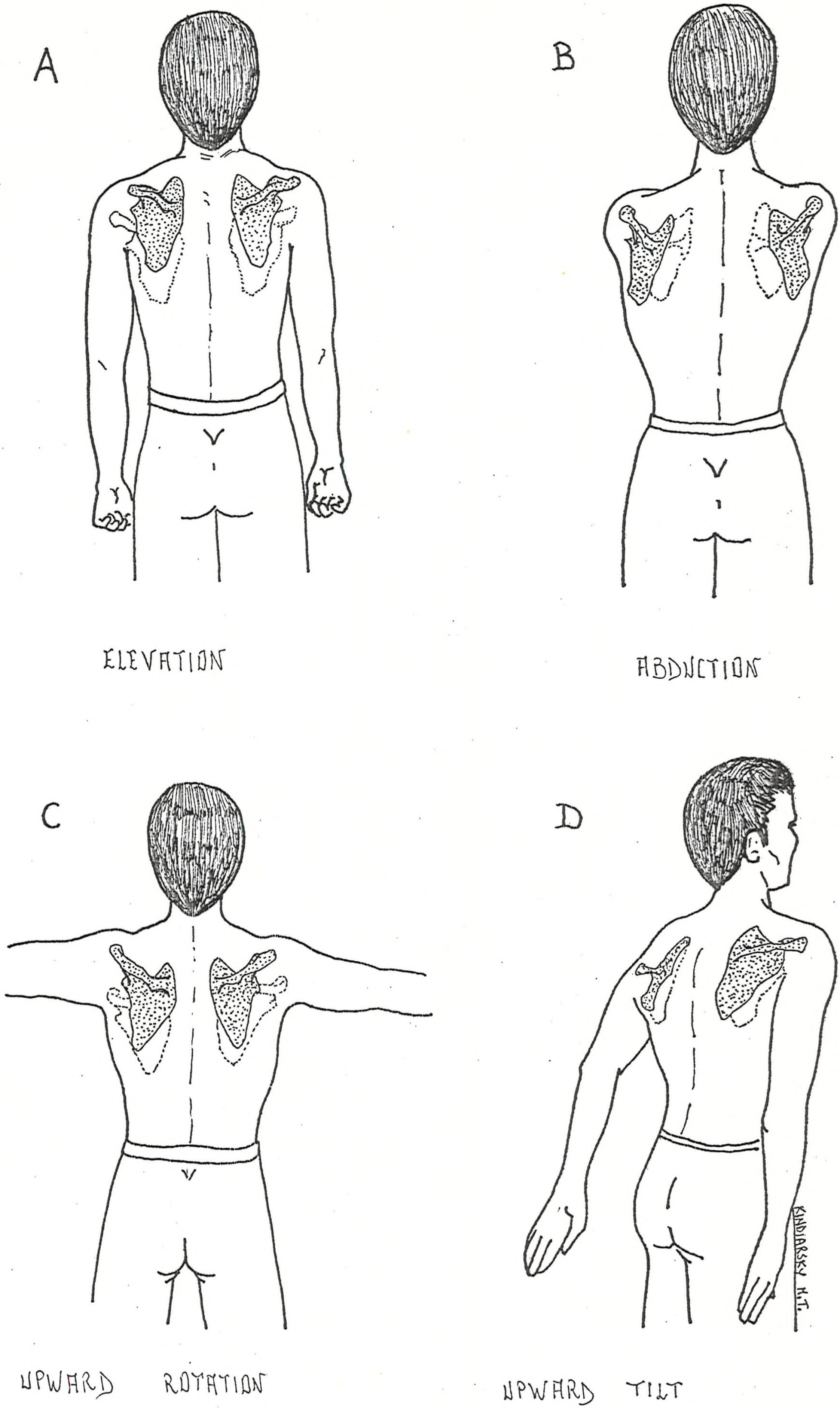


FIG. 3.7 PRINCIPAL MOVEMENTS OF THE SHOULDER GIRDLE

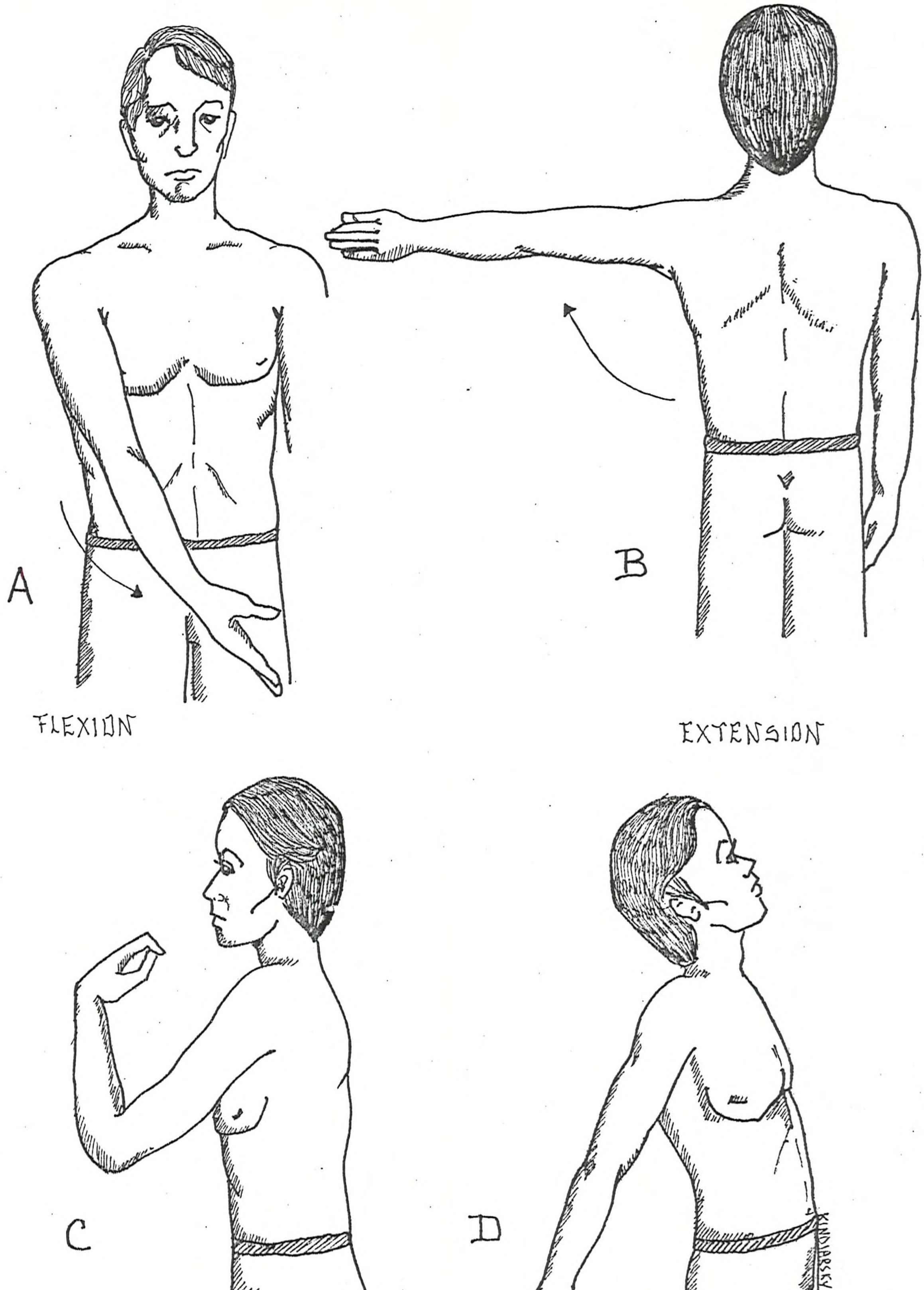


FIG. 3.5 PRINCIPAL MOVEMENTS OF THE HUMERUS

CHAPTER FOUR

FEEDBACK CONTROL OF SKELETAL MUSCLE: A REVIEW

4.1. INTRODUCTION

It is well known that the human nervous system represents a very complicated and sophisticated feedback system. Such a control system in the human body participates in a variety of functions such as regulation of body temperature, sugar level in the blood, carbon dioxide in the tissues, etc.

The neuromuscular control system comprises many reflexes and reflex pathways, and this system is not only used for maintaining posture. In principle the same system with its neural reflex pathways is used when different types of movements are performed. A reflex in the control system of posture or movement implies that a signal carrying information concerning a part of the muscular-skeletal system is fed back to a control centre. This information is then used in regulating the activity. The neuromuscular system comprises a complex hierarchy of feedback control mechanisms and may be likened to an extensive, multiloop servo system. Motor control is one of the most important functions of the central nervous system. One of the basic difficulties handicapping all research on the motor system, is the lack of unifying theory of motor function. With good reason it has been felt that the available information is too fragmentary for the formulation of any overall theory and most investigation proceeds without it.

In studies on spinal reflexes, for example, muscles, receptors and neurons are usually treated as though they were isolated elements with little regard for their roles in coherent reflex systems. This is somewhat like studying a finger without being aware that it is part of the hand. A comprehensive approach relating all studies on motor function is needed, but is still far from feasible.

The discipline of control theory offers an approach which is suited to these problems and it offers some advantages in the study of motor functions. Motor control in man is realized by a complex set of neural structures. The course of evolution has created a hierarchy of these structures, each contributing certain characteristics and capabilities to the motor system, but none of which can be designated as sole source or organizer of any specific motor function.

4.2. STRETCH REFLEX SYSTEM AND SERVOMECHANISM HYPOTHESIS OF NEUROMUSCULAR CONTROL

There is little question that feedback plays a dominant role in the muscular control of posture and movement, even though many details are still unknown. One of the most important feedback loops is the stretch reflex, also termed myotatic reflex, which produces a muscular force in opposition to a lengthening of a muscle. The stretch reflex is involved in the regulation of posture as well as in the control and, in all likelihood, the initiation of movements.

The anatomical components of this reflex and its mechanism have been briefly reviewed in Chapter 2.

Figure 4.1 shows a simplified schematic diagram of the stretch reflex system. The components are interconnected by afferent axons from spindle receptors which form, in the spinal cord, excitatory connections with agonist motor neurons and inhibitory connections with antagonist motoneurons, and by efferent axons of motor neurons which innervate the muscle.

That this interconnection of physiological components can be recognized as a feedback control system is apparent from the block diagram of Figure 4.2 which shows some of the peripheral components of the motor system.

The alpha motoneuron (α) summates signals that originate centrally (Δ) and peripherally (only I_a afferents shown) and activates the muscle. The force F_M generated by the muscle, plus external forces, act on a mechanical load (which includes the intrinsic viscosity, elasticity; and in ertia of the muscle itself) and movement occurs as governed by the basic laws of physics. Length changes of the muscle (x) are sensed by the muscle spindle and transmitted back to the alpha motoneuron to restore the muscle to its previous length. The set point of the spindle error detector, as well as the dynamic behaviour of the spindle, are under control by the dynamic and static fusimotor system (γ). The stretch reflex is looked upon as a powerful servo, automatically holding the muscle at the particular length "demanded" by the fusimotor discharge, and thus unaffected by any disturbances produced by variations in the loading of the muscle or in its strength.

A servomechanism may be defined as any automatic control mechanism which is actuated by an error signal occurring in a closed-loop control cycle and which possesses power amplification.

The stretch reflex, being independent of the cortex, is clearly automatic and its "power amplification" is provided by the contraction of the muscle.

The closed-loop control cycle may be identified as being: tension on muscle, stretch of muscle, excitation of spindle endings, excitation of the α motoneurons of the same muscle, contraction of muscle, increase in muscle tension. This is a negative feedback loop, for the increase in tension in the muscle on its contraction opposes the applied tension, and so tends to maintain the length of the muscle constant in spite of the disturbance. Equally, the muscle tends to maintain the same length when the load on it is reduced. Therefore, the length control system functions as an automatic regulator which quickly compensates for disturbances. Another manifestation of the system is its ability to execute any changes in length dictated by the control signal.

Higher centres in the nervous system have only to initiate a new control signal and then may go on to other tasks, knowing that the length control system will automatically move the arm (the load) to the desired position. Furthermore, if any disturbances should occur, the control system would also automatically regulate against them.

A further basic premise of the servo hypothesis is that the stretch reflex is sufficiently powerful (i.e. sufficiently stiff, has enough gain) to ensure that the shortening commanded by the γ bias occurs irrespective of the value of any load and the variation of the strength of extrafusal fibres with their length; the servo will then produce the appropriate increase in α motor discharge required to compensate entirely for such things.

If a movement were to be produced solely by the α route, then the command signal to the α motoneurons from the higher centres would of itself have to take account of these factors, whereas a command issued via the α route does not. Thus the use of the γ route of excitation would save the higher centres a certain amount of effort. The circumstances under which use is made of one route or the other, or (as seems more likely) various mixtures of the two, are not known, although, for example, it is probable that posture would mainly employ the α route, and that rapid movements with minimum reaction time go through the α route. If there is truth in these suppositions, it is clear that some mechanism must exist to adjust activity quickly between one route and the other, and to vary the proportions of the two routes in use. Granit speculates that, in ordinary life, both pathways, direct and servo, are employed together (γ-α linkage), perhaps with the proportions of the two in use varying according to circumstances. Matthews forcefully urges contractions controlled by such a watered-down follow-up servo, are properly referred to as "servo-assisted".

We may conclude that the central nervous system is offered two pathways through which neural commands may be transmitted to the length control system, one by way of the alpha motor neurons and the other by way of the gamma motor neurons. Both pathways act through the length control system and therefore enjoy the advantages of feedback which has already been emphasized; however, each also has subtle advantages of its own. The alpha route to muscle is faster. Alpha commands encounter only a brief delay between motor neuron and muscle. Gamma commands are additionally delayed due to transmission in slowly conducting gamma efferents and in afferent fibres from spindle to alpha motor neuron. On the other hand, inputs via the gamma route have the advantage of causing muscle shortening without a steady-state change in the output of the spindle, whereas large inputs via the alpha route may drive the spindle below threshold or into saturation, thus hindering reflex regulation against disturbances. One might speculate that alpha inputs are employed where speed is essential and gamma inputs are employed for smooth, continuous control. However, we must await further experimental evidence to elaborate the full possibilities of these two modes of control.

4.3. OTHER FEEDBACK SYSTEMS

In addition to the reflexes seen before, other spinal reflexes are involved in the task of combining the motions of individual joints into purposeful movements of the whole limb, or even the whole body. Balance, for example, is maintained by associated reflexes.

Two categories of proprioceptive feedback are specifically designed. Peripheral feedback includes signals from spindle receptors and tendon organs originating in the primary muscle system as well as others which, through mechanical linkage, are coupled to movement at the primary joint. This feedback operates with short latency and is apparently "continuous" in nature. Central feedback includes signal from joint and visual receptors specifically, but one may also consider those cutaneous signals which are involved in the more automatic decisions affecting voluntary control. Experiments with human operators suggest that this feedback is "discontinuous". Muscular control is, therefore, managed by a hierarchy of feedback control systems built one upon another. Every one of the levels occurring in this hierarchical organization is a set composed of material things, each of which is characterized by peculiar properties dictated by its particular laws.

At the bottom of the hierarchy are the positional control systems, which must serve as the final common mechanism of control for all joints and muscles which participate in the control of movement. Complex movements require the action of many types of feedback. The Golgi tendon organ responds to an increase in tension rather than length. The tendon organ in the muscle provides the necessary feedback by sending to the motor neurons a signal proportional to the actual tension being produced in the muscle. If this tension is not the value specified by the control signal, then the discrepancy is detected by motor neurons which generate an error signal which modifies muscle tension so as to reduce this error. If the length of the muscle or its velocity should change, or should fatigue set in, the tension feedback mechanism will automatically compensate for these disturbances.

Feedback from receptors in the skin of the moving-parts is required for exploratory behaviour and grasping. Maintenance of upright posture and balance requires vestibular feedback which contributes to postural stability and assists in updating spatial awareness. The vestibular system also operates an eye tracking system so that visual contact is maintained despite head movement. If posture becomes unstable, this sytem will take over all motor functions until posture is restored and, under those conditions, the vestibular apparatus is at the head of hierarchy.

A large part of the all voluntary motor activity is guided by visual feedback, the basic role of which is clearly to provide a measure of the error between the actual and the desired position of a limb. It may also provide a measure of the rate at which the error is being reduced and so improve the dynamic performance of the task by enabling corrective action to be initiated earlier when the rate of change of error is great.

The manner in which all of the various control mechanisms are co-ordinated, with proper regard for their priority in different situations and for the sequence and timing of movements, is not fully understood.

Figure 4.3 summarizes the higher centres and their inter-communication paths. It is not a straightforward hierarchical system of controls stacked one upon another, neither are the elements equally interconnected as in a net. Most important is the motor cortex, which can initiate movements and has direct access to the motoneurones and cord mechanisms though the cortico-spinal or pyramidal track. This pathway is of particular importance for fine and discrete movements. All other motor signals to the spinal cord originate from cells in the reticular formation, which receives dominant transmissions from the cerebellum, the vestibular apparatus and the cerebral motor cortex.

The subcortical structures, the thalamus and the basal ganglia, can influence the reticular formation, but their major transmissions are upwards to the cortex. The cerebellum, which receives signals from all of the known feedback systems, unquestionably plays a major role in co-ordination. It is reciprocally linked with both sensory motor cortex and positional control systems and may be regarded as a feedback centre common to both.

It is generally believed that in some way the cerebellum functions as a type of computer that is particularly concerned with the smooth and effective control of movement.

It is assumed that in the cerebellum there is integration and organization of the information flowing into it along the various neural pathways and that the consequent cerebellar output either goes down the spinal cord to the motoneurones, and so participates directly in the control of movement, or else is returned to the basal ganglia and to the cerebral cortex, there to modify the control of movement from the higher centres.

The cerebellum receives muscle length and tension information directly through very fast conducting pathways, and the circuitry of the cerebellum seems designed to produce an output immediately. Impulses can reach the cerebellum, be processed and transmitted back to the spinal cord in the time it takes for a motoneurone to discharge and for the muscle fibres to twitch.

The cerebellum is important for the correct timing of movements. Its precise function, however, has not been defined yet.

CHAPTER 4

LIST OF FIGURES

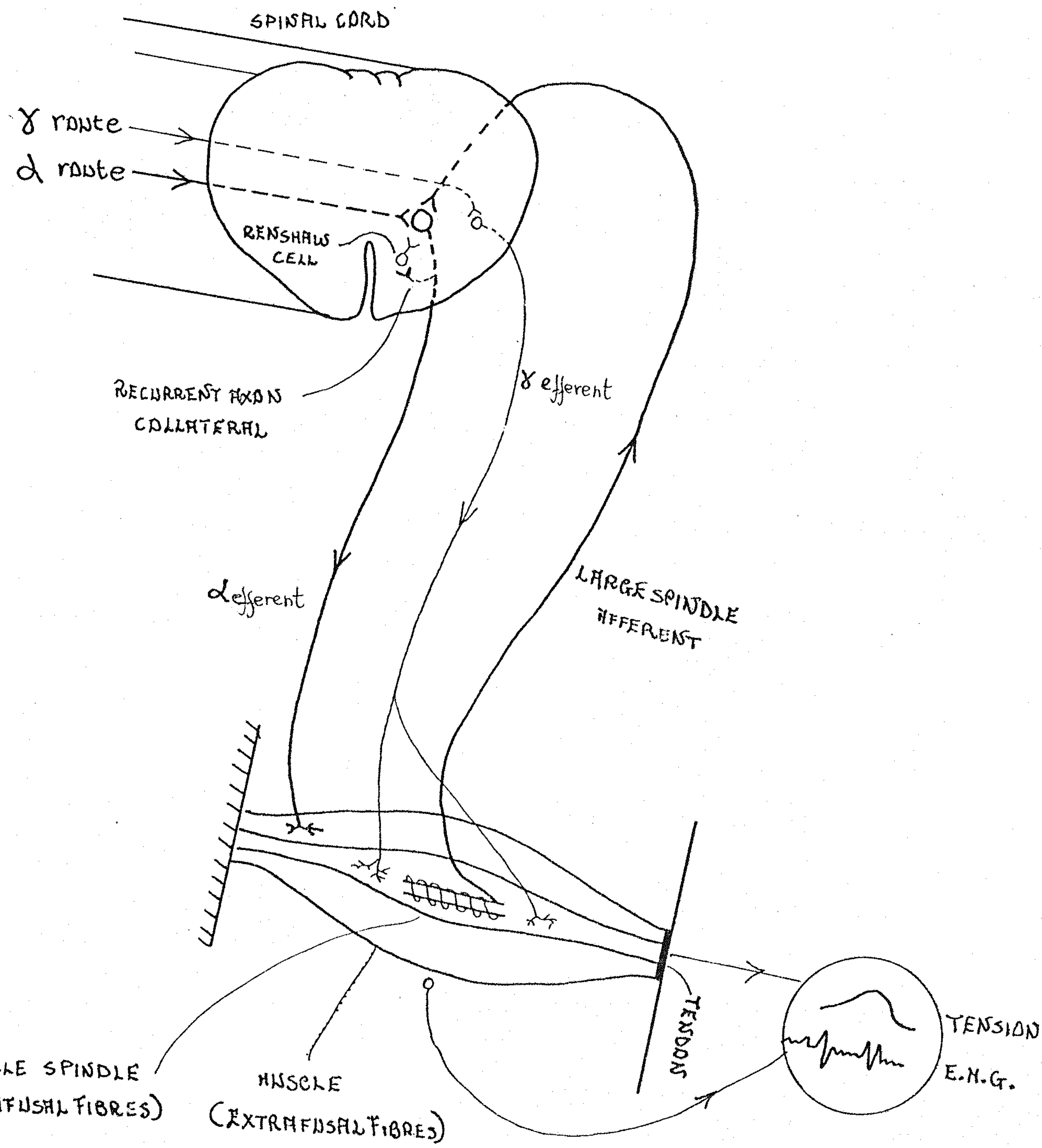


FIG. 4.1 SIMPLIFIED SCHEMATIC DIAGRAM OF STRETCH REFLEX SYSTEM

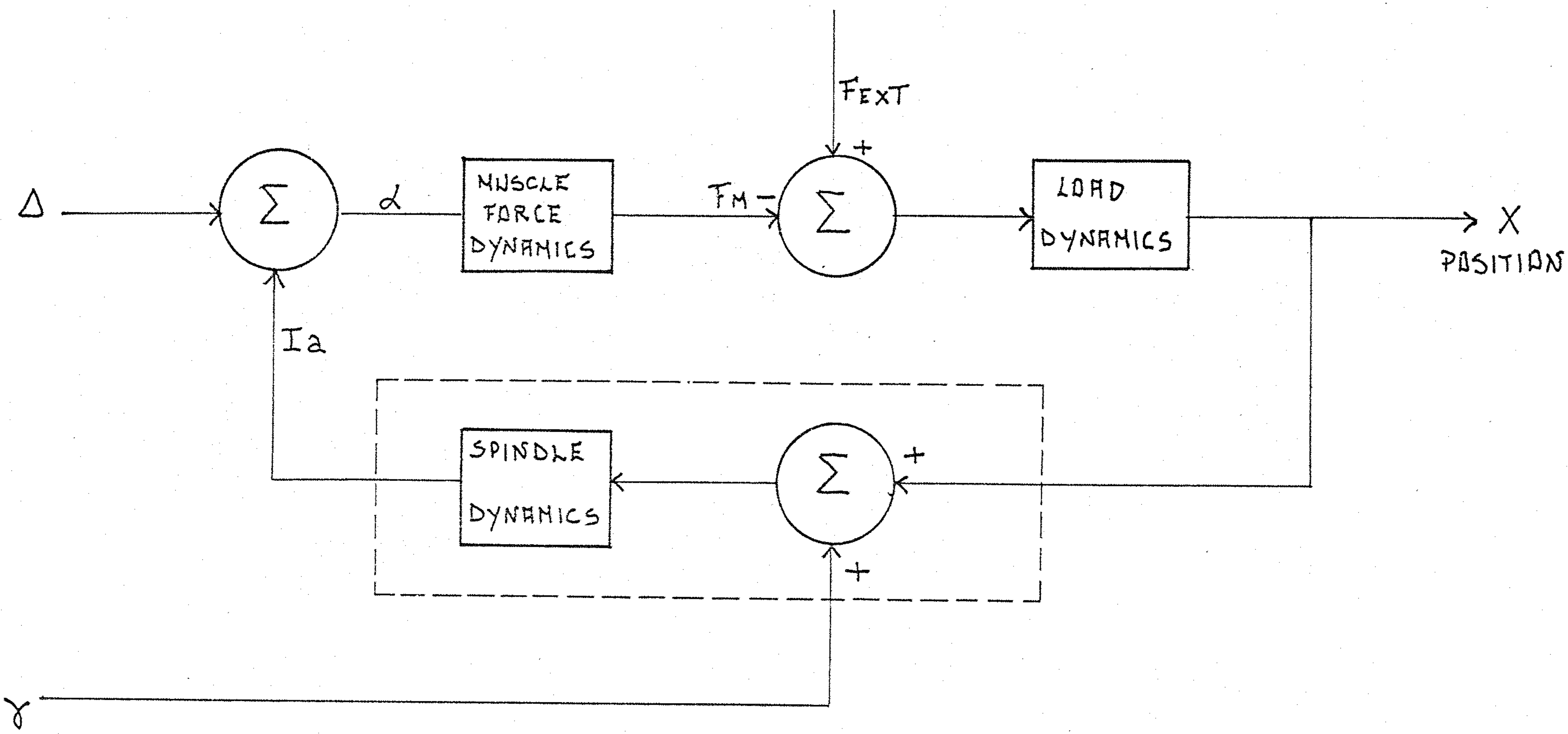


FIG. 4.2 A SIMPLIFIED BLOCK DIAGRAM OF THE PERIPHERAL MOTOR SYSTEM

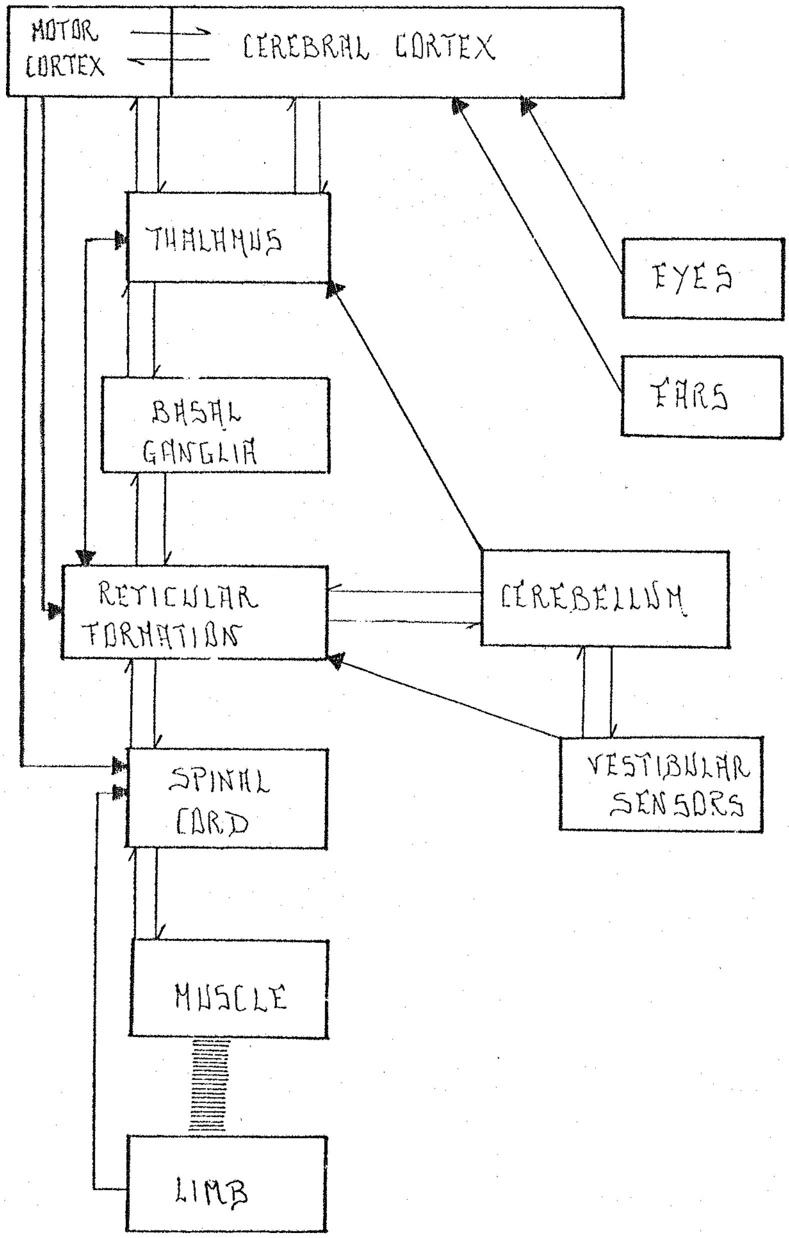


FIG. 4.3 ANATOMICAL COMPONENTS OF THE MOTOR CONTROL SYSTEM AND THE PRINCIPAL INTERCONNECTIONS

CHAPTER FIVE

DESIGN OF A VIBRATING PLATFORM AND ITS CONTROL SYSTEM

5.1. INTRODUCTION

The main apparatus used in the experimental work described in the thesis is a pneumatically actuated platform capable of vertical movement, (Figure 5.1). A special purpose device was constructed because of the low range of frequencies needed with comparatively large amplitudes which exceed the normal operating range of vibration shakers.

The main problem with very low frequency mechanical oscillators is that of stick-slip motion which is caused by a combination of friction characteristics and backlash in gearing and similar mechanisms. Any electrical drive designed to meet the specifications would necessarily employ a gear box and for this reason and because of poor torque characteristics at low speeds in most electrical drives it was decided to use direct mechanical actuation. A pneumatic system was chosen rather than a hydraulic one for reasons of economy.

The pneumatic actuator consists of a large air-bag accumulator placed under a moving platform. In this way large forces could be generated with modest variations in the air pressure. However, since the volume of the air bag is large and considerable variations are required, it is necessary to generate a large volumetric flow-rate of air at normal temperatures. A centrifugal turbo blower was chosen to meet the flow requirement. A disadvantage of such devices is that they are dynamic rather than flow generators and their flow/pressure characteristics are both time and load varying. However since air is an inexpensive material it was possible to design very conservatively with considerable margins in both pressure and flow.

There are also problems of resonance due to compressibility in pneumatic control systems and these will be briefly discussed further on. Here the operating frequencies are so far below any resonance that no special stability features were required.

In addition to the actuator and platform a control valve is required together with its actuating system and an instrumentation system for measuring appropriate subject variables. The individual performance and that of the complete system are discussed in this chapter.

5.2 THE VIBRATING PLATFORM

The vibrating platform carries a chair supporting the subject who then undergoes displacements controlled by an independent signal generator. A range of signal characteristics were used but initially the design was based on values of acceleration for which the vestibular apparatus is known to respond. (see Appendix 5.1). This led to a design specification for the platform dynamics as shown in Table 5.1. The platform was conveniently designed as a hollow box containing the airbag accumulator. (Appendix 5.2). The effective cross section area, S is about $0.70 m^2$. The maximum vertical velocity is

$$V_v = A_M \cdot W_{MAX} = 5.65 \text{ m/sec}$$

where: A_M is the maximum displacement and $W_{MAX} = 2\pi f_{MAX}$ (f_{MAX} = maximum operating frequency).

Hence the maximum flow rate required is:

$$F_{R_{MAX}} = V_{v_{MAX}} \cdot S = 3.95 \text{ m}^3/\text{sec} \simeq 66 \ell/\text{sec}$$

where: $F_{R_{MAX}}$ = flow rate max.

$V_{v_{MAX}}$ = Max vertical velocity

S = Cross section area

This must be accomodated by the turbine and the control valve. The other control problem which directly affects the platform is that of resonance. It is known, Appendix 5.3, that pneumatic servos tend to oscillate at the natural frequency of the output mass supported by half the volume of air between the valve and the actuator. This is (see Appendix 5.4)

$$\omega_o = 2\pi f_o = \sqrt{K/M} \doteq \sqrt{\frac{2S^2 \frac{P}{\gamma}}{VM}}$$

where:

K = stiffness (N/m)

f_o = natural frequency

S = cross section area (m^2)

P = air pressure (Kg/m^2)

V = air volume (m^3)

M = total mass (Kg)

γ = ratio of specific heats

Normal values for P and V give:

$$\omega_o \simeq 6.6 \text{ rad/sec}$$

This is a rather pessimistic estimate of ω_o, since adiabatic conditions have been assumed. Even so it estimates the resonant frequency below the common operating frequency. In practice friction in the box supports is somewhat large and in free suspension the natural oscillations of the box decay after few cycles. This is confirmed by frequency responses measured for the closed loop system.

5.2.1. THE PRESSURISED SUPPLY SYSTEM

The pump unit chosen (Appendix 5.5) was a centrifugal turbine blower as shown in Figure 5.2. The drive is a 415 V, 0.56 KW variable speed a.c. motor. The maximum air supply is about 9000 ℓ/m at 1494 N/m^2 pressure. It can be seen from the last section that even at the extreme pressure requirement (1494 N/m^2) the maximum delivered flow is greatly in excess of that required. Thus the demands made on the pump during the test run will not drastically affect its performance and it may therefore be regarded as a constant pressure source. Infact under low flow conditions, the pressure delivered by the pump was so great that it could damage the air bag and for this reason a variac was used to regulate the motor speed so that the operating pressure was kept within acceptable limits.

It was impractical to vary the flow by changing the turbine speed continuously and so to vary the flow of air into the bag a by-pass system using a "butterfly" valve was used. This is described in more detail in section 5.3.

5.2.2. THE SEAT AND SUPPORT

The vertical displacement of the platform is to be transmitted to the torso of the subject. To eliminate spurious lateral movement it was decided to seat the subject on a rigid chair.

In addition a back support was provided to restrain relative lateral motion of the head. The seat could be adjusted relative to the platform in both vertical and horizontal planes while the back support has three directional adjustments which allow variations in torso position. Both the seat and the support could be adjusted to suit the dimensions of the subject while inhibiting unwanted lateral movements. Problems of body compliancy were great and some experimentation was necessary to ensure both reduction of unwanted movement and subject comfort and distraction during prolonged tests. Eventually the design shown in Figures 5.1 and 5.2 was evolved.

5.3. THE CONTROL SYSTEM

A schematic diagram of the complete control system is shown in Figure 5.3. The flow of air to the airbag actuator is to be regulated by a flow control valve. With sealed pneumatic valve systems operating at low velocities about the null position there are stability problems since the valve cannot generate sufficient flows to compensate for compression effects in the actuator. Stability can be achieved by allowing leakage through the valve although this can result in loss of up to 1/3 of the total energy from the source. Here loss of power or fluid is not a problem and so a by-pass system has been used with the valve modulating the leakage to atmosphere, see Figure 5.4. In this way there are small fluctuations in pressure about a fixed value which balances the weight of the platform.

The butterfly valve was constructed largely empirically since there was little available literature on flow-pressure characteristics for such devices with air as the operating medium. A diameter was chosen so that full closure produced air flow variations consistent with the required maximum platform velocities. In view of the large capacity of the bag it was considered best to aim at bang-bang control of the butterfly valve, see Figures 5.4 - 5.5. Accordingly a subsidiary feedback loop was provided to control the small D.C. motor driving the valve. Angular displacement of the valve is sensed by a potentiometer and the gain of the loop was adjusted to give a maximum closing time of 100 m sec. The time constant of this loop is negligibly small when considering the response of the overall system.

Position feedback from the platform was obtained from a 12.5 cm linear potentiometer attached to one side of the platform, see Figure 5.6. This potentiometer had a sensitivity of 0.5% which therefore matched the design specification. Input signals were obtained from a signal generator and compared with the feedback signal in an operational amplifier. However, the main problems in such systems were avoided by providing an excessive power margin in the pneumatic power drive.

Because of the bang-bang control loop around the main valve, and the integrating properties of this valve, spurious inputs of power interruptions would cause drift in the platform displacement. Since the platform travel is limited, a "soft" constraint was provided to prevent damage.

This safeguard consisted of a switch which caused the valve to open fully thus venting the supply to atmosphere and returning the platform to its neutral position. During start-up, also, care had to be taken to gradually increase the pressure in the bag to avoid damage. Figure 5.7 shows the circuits of the complete control system.

5.4. MEASURED RESPONSE

A feature of the system is that the demanded platform displacements can easily be selected and varied by appropriate choice of an electrical signal. Both sinusoidal and random signals were used in the experimental program. The latter signals were first-order Markov signals of break-frequencies up to 10 Hz, but predomnantly lower than this value. RMS amplitude variations of 8 cm were selected.

Frequency response tests on the system showed its response to be flat within the operating range of frequencies. Similarly transient response characteristics, Figure 5.8, show well-damped characteristics with a rise time of 1.4 sec. and nearly no oscillations.

Only few adjustments, e.g. to loop gain, were necessary to bring about a completely satisfactory performance. Table 5.2 shows values of platform displacements and accelerations for sinusoidal inputs of different amplitude and frequency. Figures 5.9 and 5.10 show input and output waveforms obtained from pen recorders. Figure 5.11 compares platform displacement and acceleration. It can be seen that there is little harmonic distortion. The system behaves in a linear fashion on open-loop operation over the test range 0.1 - 9 Hz.

Table 5.3 compares the rms input and platform displacement for random signals of different amplitude and cut-off frequencies. The random generator used was a Solartron Random Signal Generator Mod. RO 1227. To produce a smooth continuous output from the generator a suitable low pass filter was designed, see

Appendix 5.7. Figure 5.12 shows recordings of input and output displacements for a range of input parameters. Although there are no apparent resonances within the loop it must be remembered that the platform load is not a rigid one. There are compliance, damping and inertia effects between the various degrees of freedom in the body of the subject. A study of the human vibration response in the range of frequencies of interest is discussed in Chapter 7. There does not, however, appear to be any serious interaction with the platform control system.

CHAPTER 5

LIST OF TABLES

TABLE 5.1

DESIGN SPECIFICATIONS FOR THE PLATFORM DYNAMICS

OPERATING FREQUENCY (f)	0.001 - 9 Hz
FREQUENCY SENSITIVITY $\left(\frac{\Delta f}{f}\right)$	5%
AMPLITUDE RANGE (A)	0 - 10 cm
AMPLITUDE SENSITIVITY $\left(\frac{\Delta A}{A}\right)$	1%
ACCELERATION RANGE $(\lvert\alpha\rvert)$	0.0001 - 0.25 g
ACCELERATION SENSITIVITY $\left(\frac{\Delta\alpha}{\alpha}\right)$	5%
MAXIMUM TOTAL MASS (M)	100 Kg
MAXIMUM VERTICAL FORCE (F_{MAX})	1250 N
MAXIMUM VARIATION IN VERTICAL FORCE (ΔF_{MAX})	500 N

TABLE 5.2

Displacement of the platform and relative accelerations for different frequencies and amplitudes of the input reference signal (sinusoidal input).

FREQUENCY	REFERENCE INPUT 20%		REFERENCE INPUT 40%		REFERENCE INPUT 60%		REFERENCE INPUT 80%		REFERENCE INPUT 100%	
Hz	MAX DISPLAC. PLATFORM cm	MAX ACCEL. PLATFORM g	MAX DISPLAC. PLATFORM cm	MAX ACCEL. PLATFORM g	MAX DISPLAC. PLATFORM cm	MAX ACCEL. PLATFORM g	MAX DISPLAC. PLATFORM cm	MAX ACCEL. PLATFORM g	MAX DISPLAC. PLATFORM cm	MAX ACCEL. PLATFORM g
0.1	0.8	0.00032	1.6	0.00032	2.4	0.00096	3.4	0.0013	4.8	0.00192
0.5	0.9	0.009	1.6	0.009	2.4	0.024	3.6	0.036	5.2	0.052
1	1	0.040	2	0.080	2.7	0.108	3	0.12	3.2	0.128
2	1	0.16	1.2	0.192	1.2	0.192	1.2	0.192	1.4	0.224
3	0.6	0.216	0.6	0.216	0.6	0.216	0.6	0.216	0.7	0.252
4	0.4	0.256	0.4	0.256	0.4	0.256	0.4	0.256	0.4	0.256
5	0.2	0.201	0.2	0.2	0.2	0.2	0.2	0.2	0.2	0.2
6									0.1	0.144

TABLE 5.3

Displacement of the platform for different frequencies and amplitudes of the input reference signal (Random signals).

FREQUENCY C/S	AMPLITUDE REFERENCE SIGNAL (RMS)	MAX DISPLACEMENT PLATFORM CM
1	0.5	1
	1	2.5
	2	4
	4	6.5
	6	6
	8	5.8
3.2	0.5	1.2
	1	2.2
	2	4
	4	6.2
	6	6
	8	5.8
10	0.5	1
	1	1.8
	2	3.3
	4	5.8
	6	6
	8	6.5

CHAPTER 5

LIST OF FIGURES

FIG. 5.1 OVERALL VIEW OF THE VIBRATING APPARATUS DESIGNED

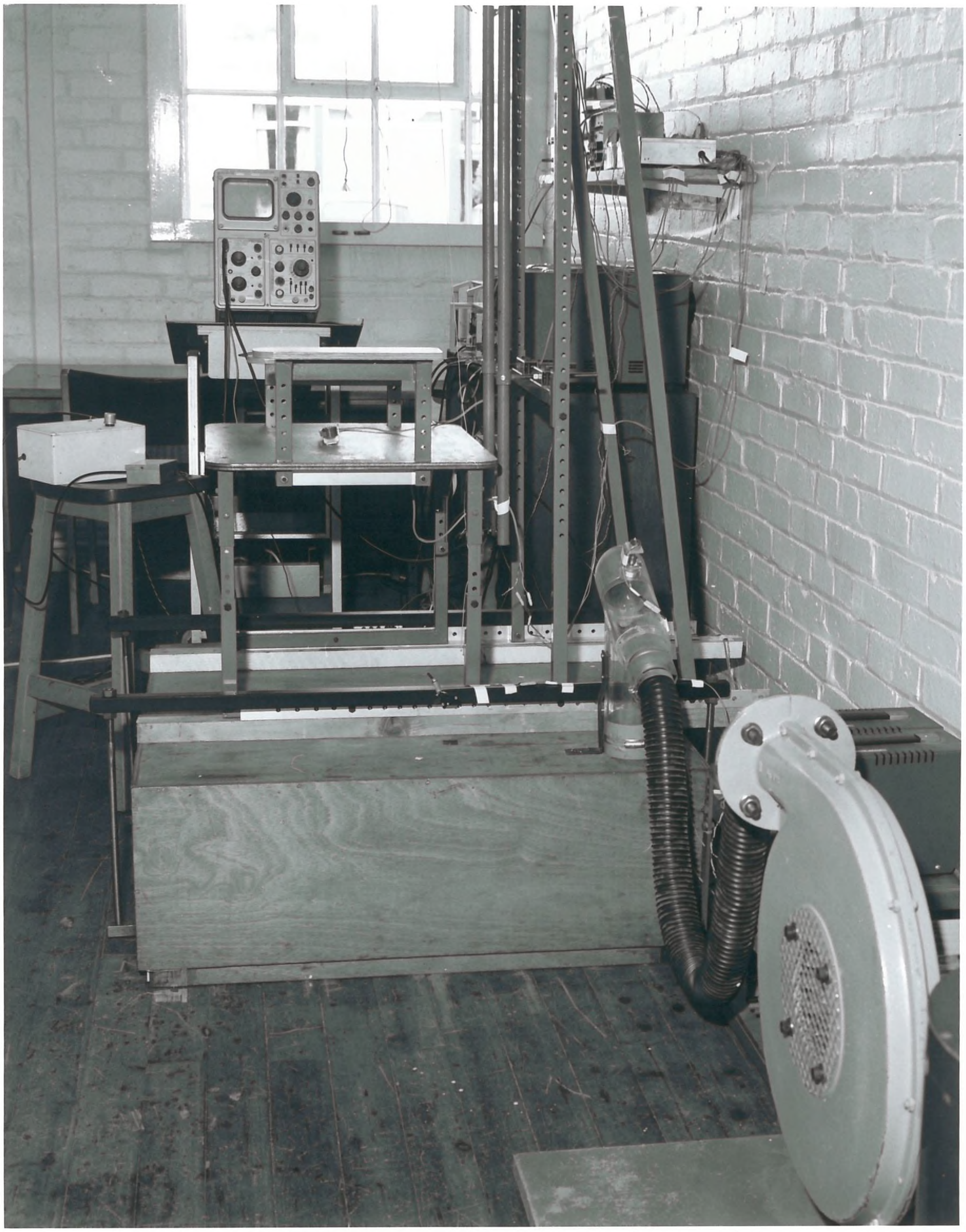


FIG. 5.2 OVERALL VIEW OF THE VIBRATING PLATFORM AND THE CENTRIFUGAL BLOWER

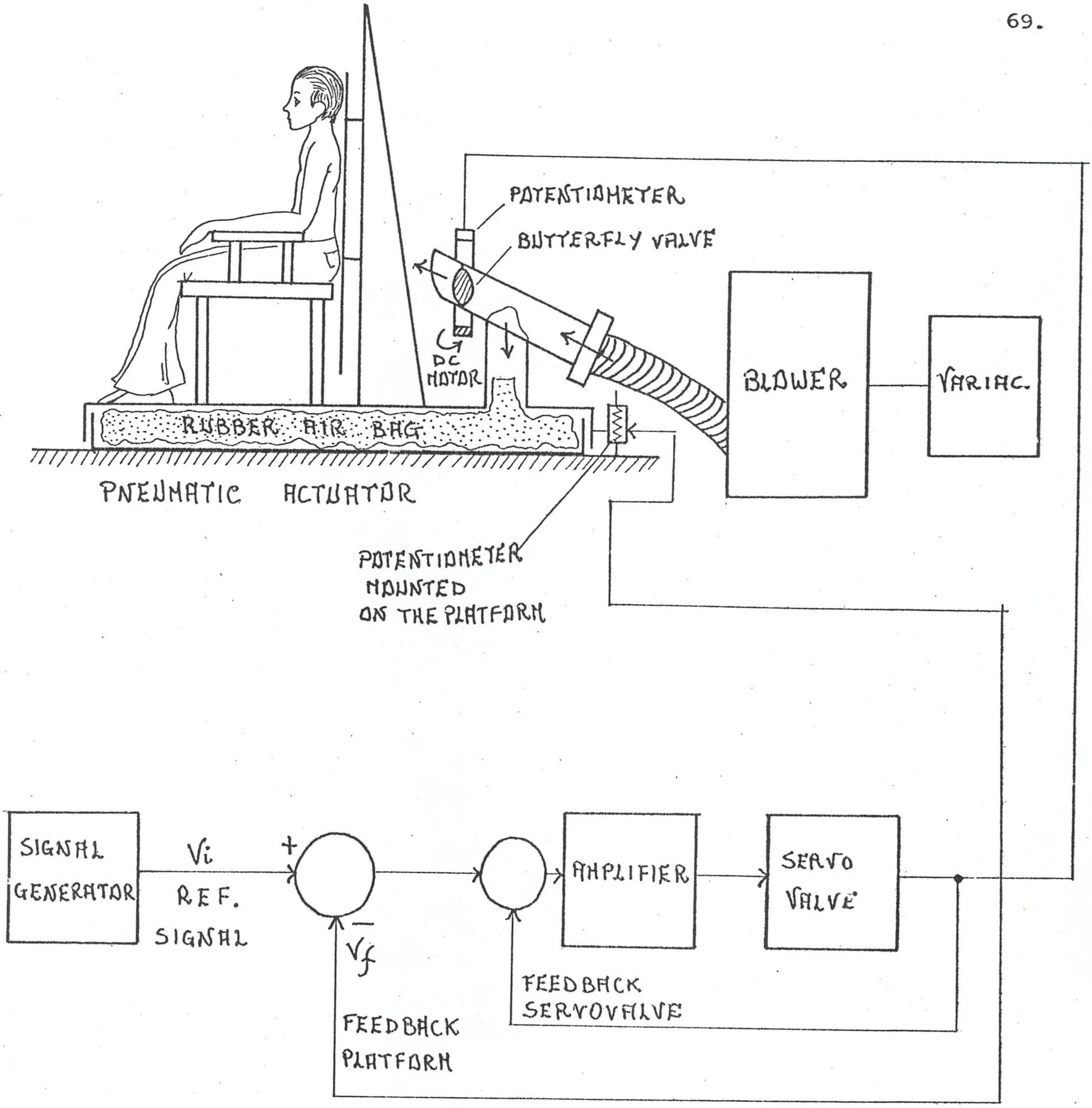


FIG. 5.3 SCHEMATIC DIAGRAM OF THE PNEUMATIC PLATFORM CONTROL SYSTEM

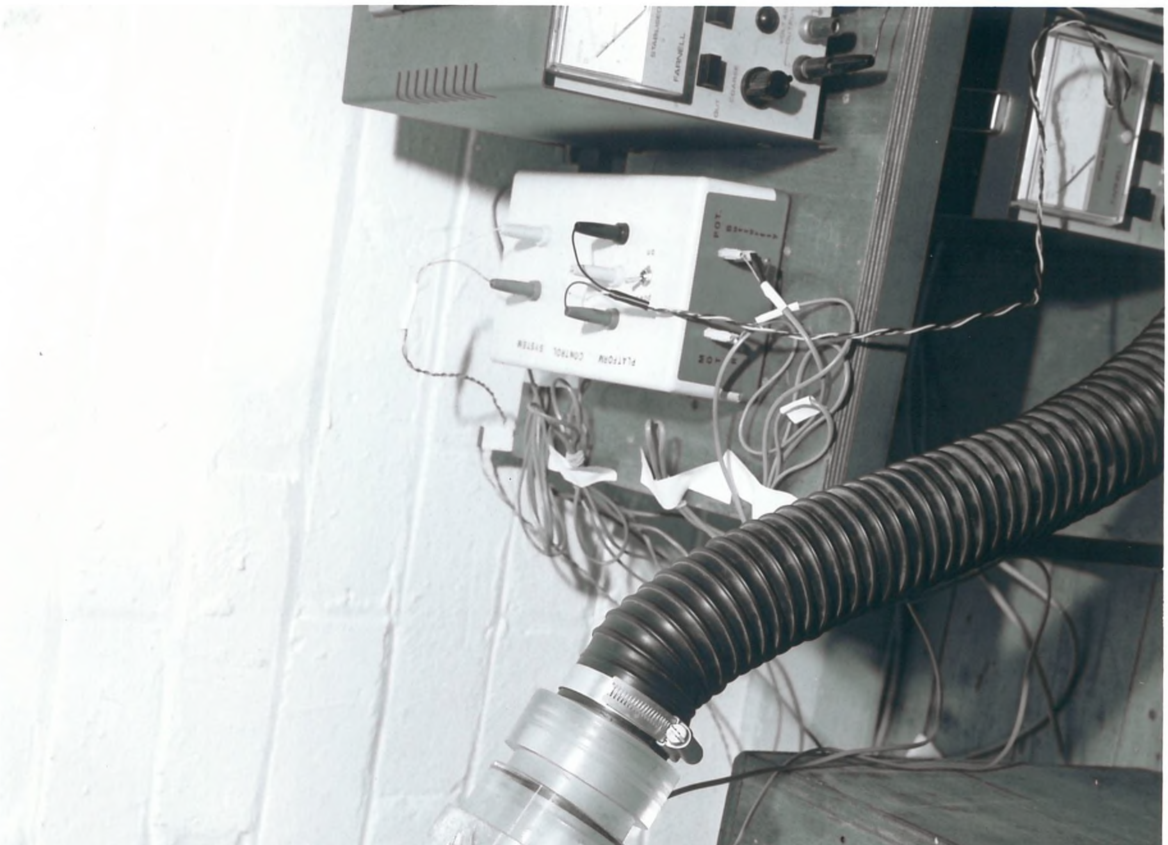


FIG. 5.4 PNEUMATIC PLATFORM CONTROL SYSTEM

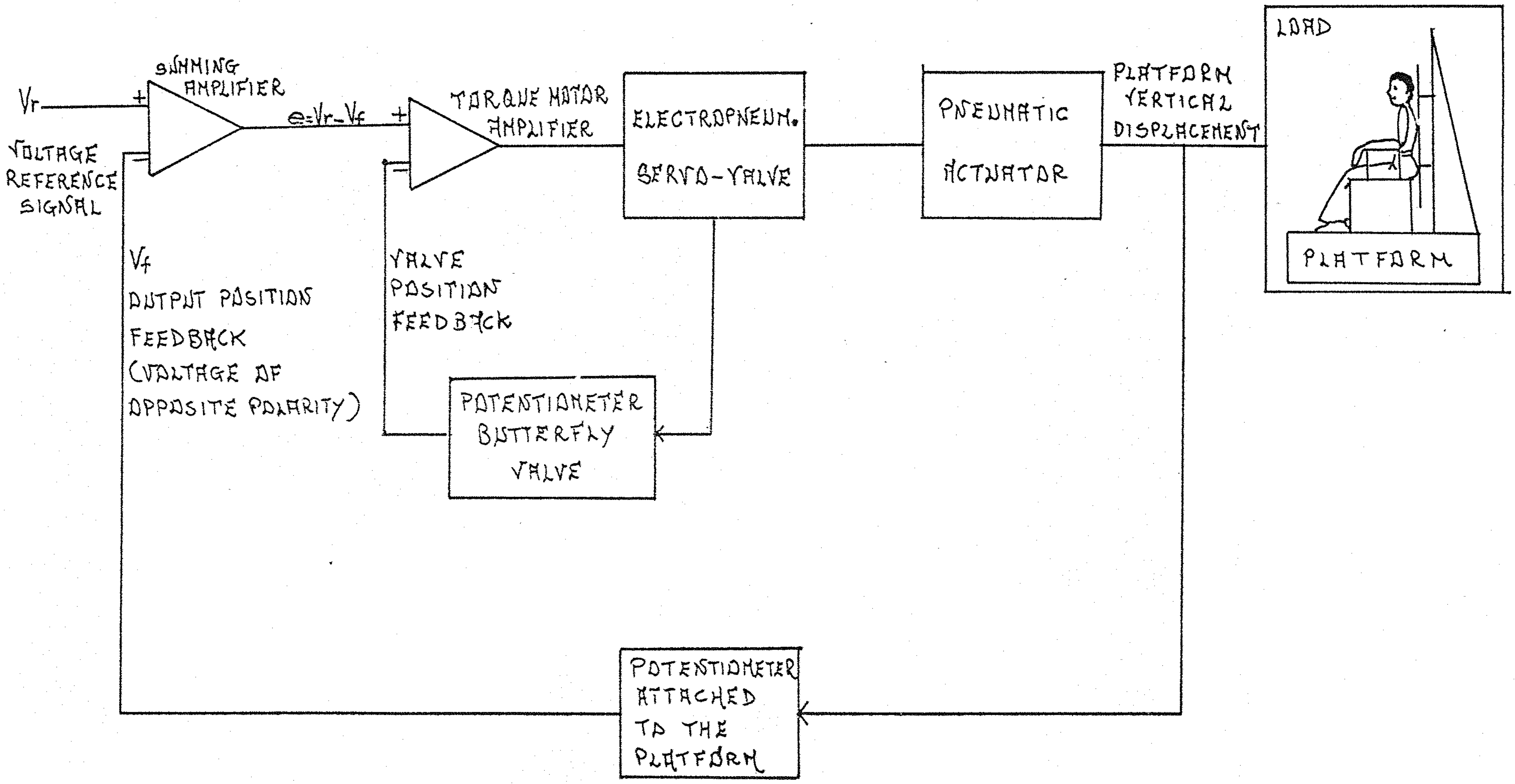


FIG.5.5 BLOCK DIAGRAM OF THE ELECTRO-PNEUMATIC POSITION CONTROL SYSTEM

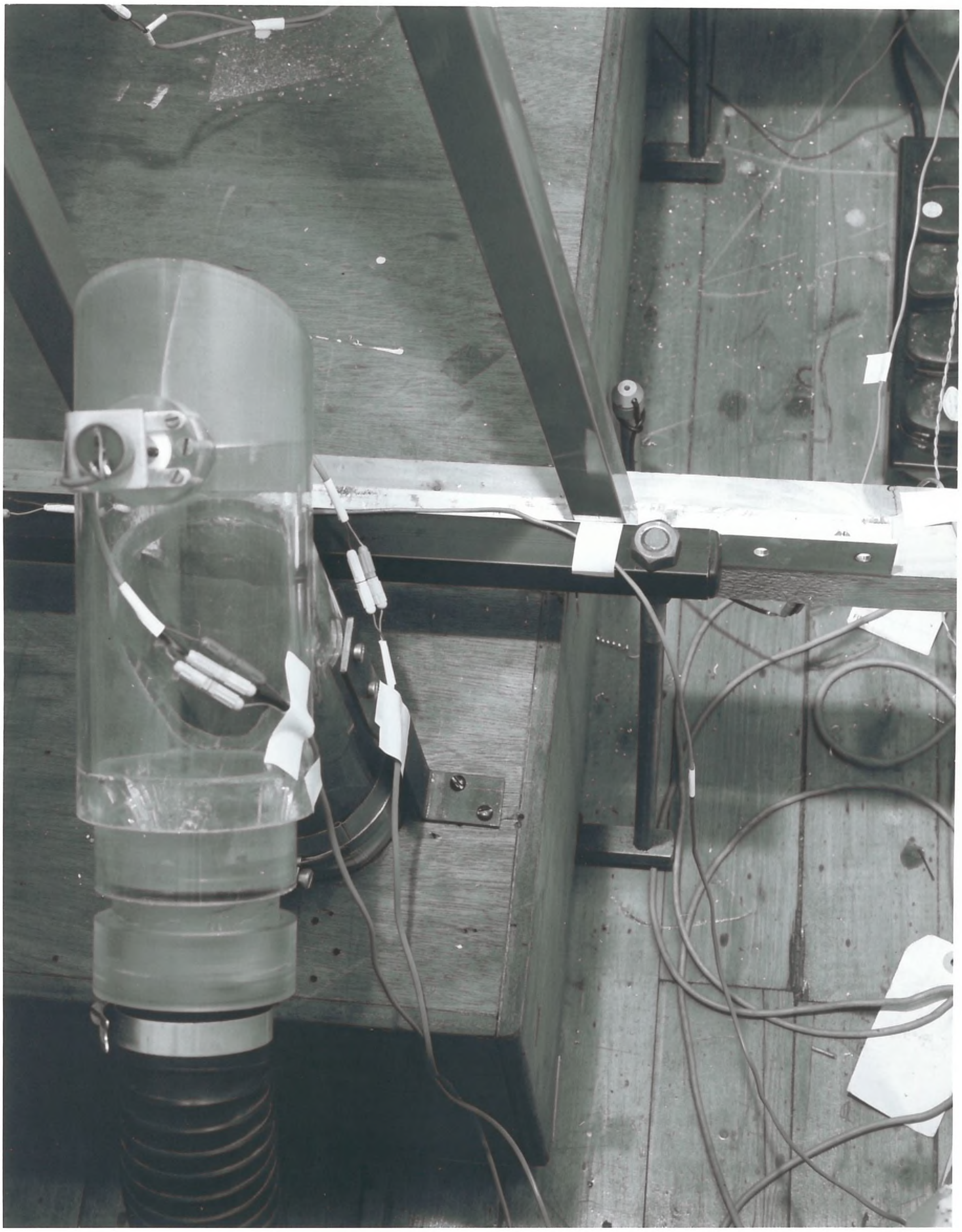


FIG. 5.6 PNEUMATIC PLATFORM CONTROL SYSTEM WITH THE POTENTIOMETER ATTACHED TO ONE SIDE OF THE PLATFORM

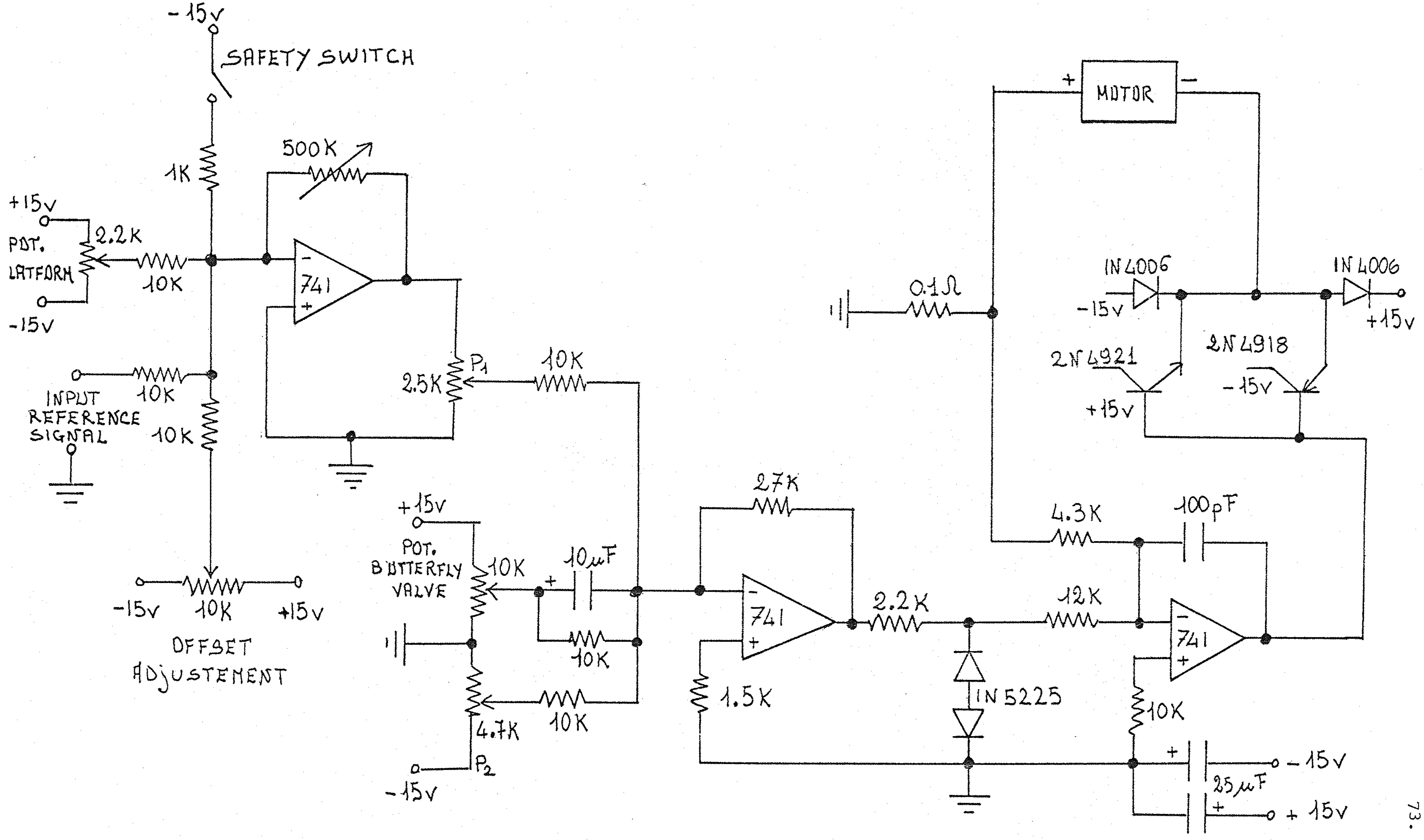


FIG.5.7 CIRCUIT DIAGRAM OF THE CONTROL SYSTEM

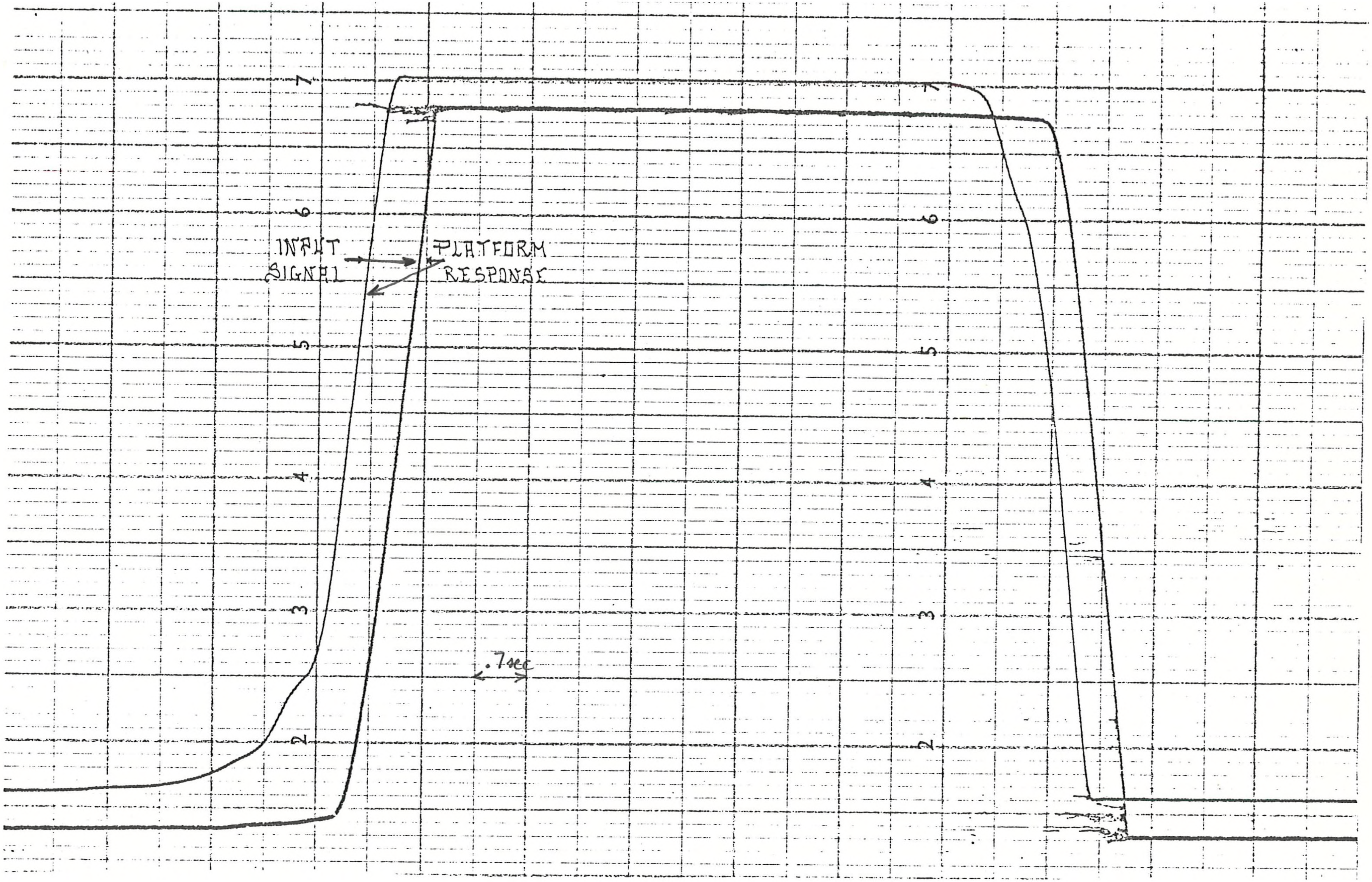


FIG. 5.8 TRANSIENT RESPONSE OF THE SYSTEM

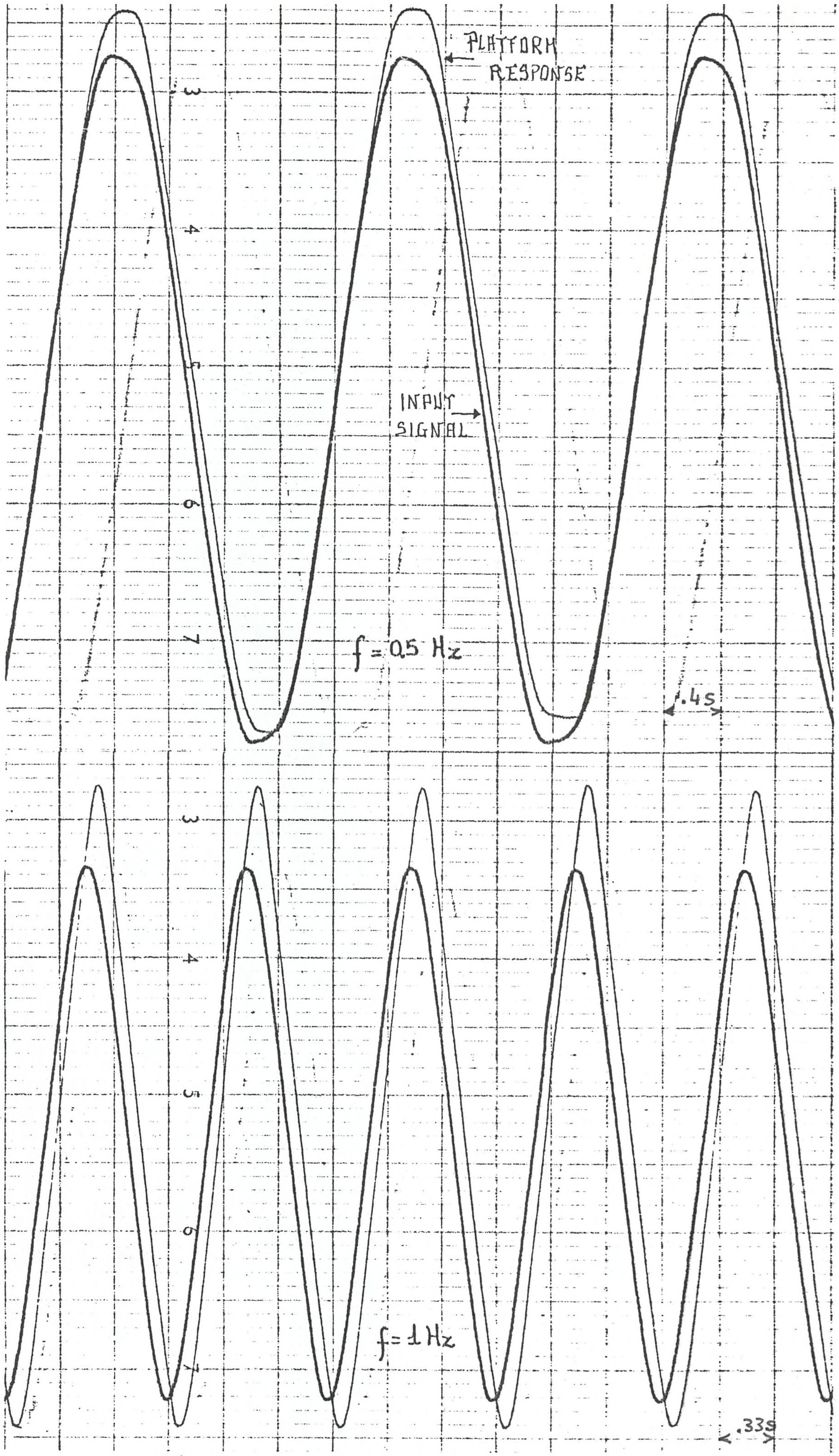


FIG. 5.9 PLATFORM RESPONSE TO A SINUSOIDAL INPUT

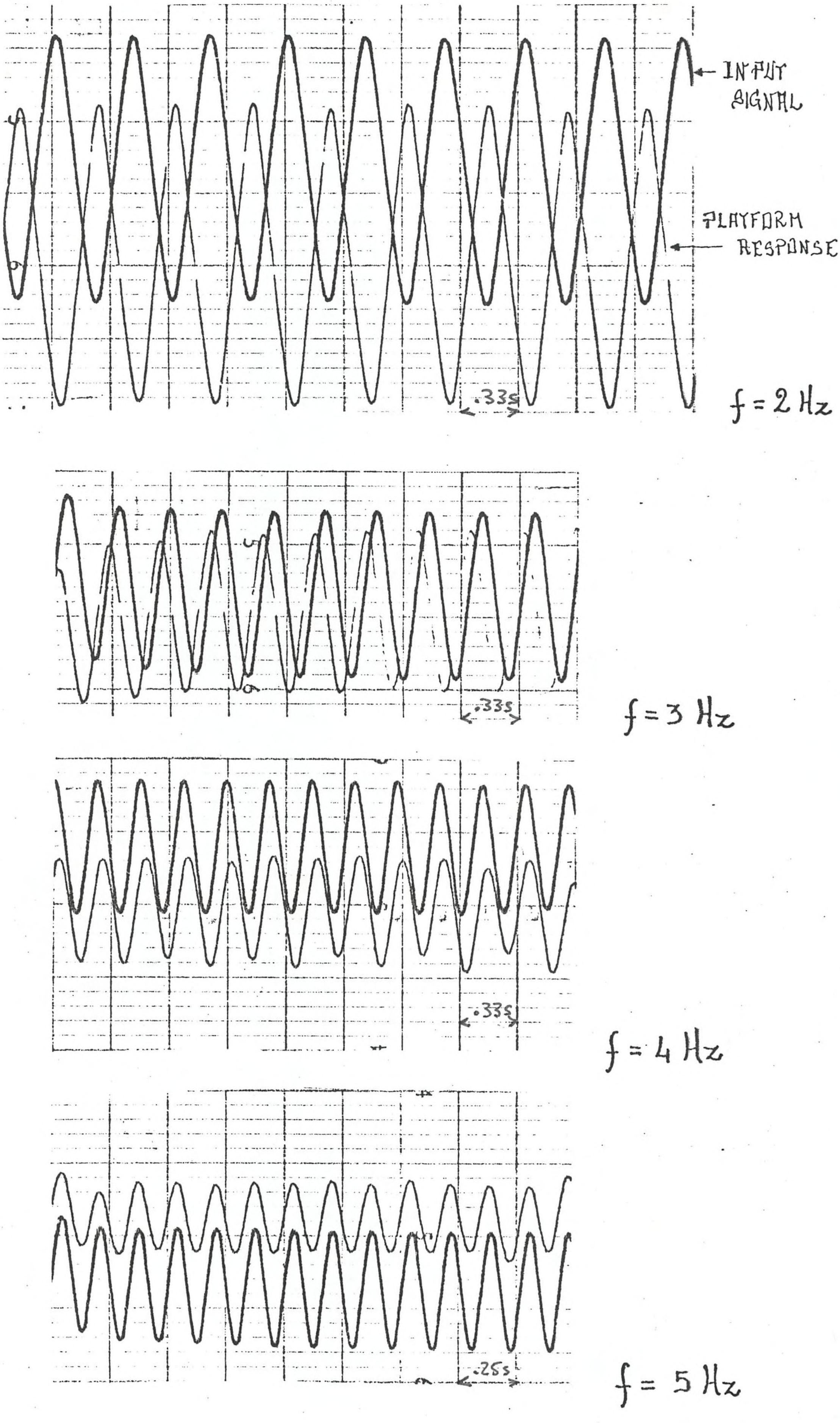
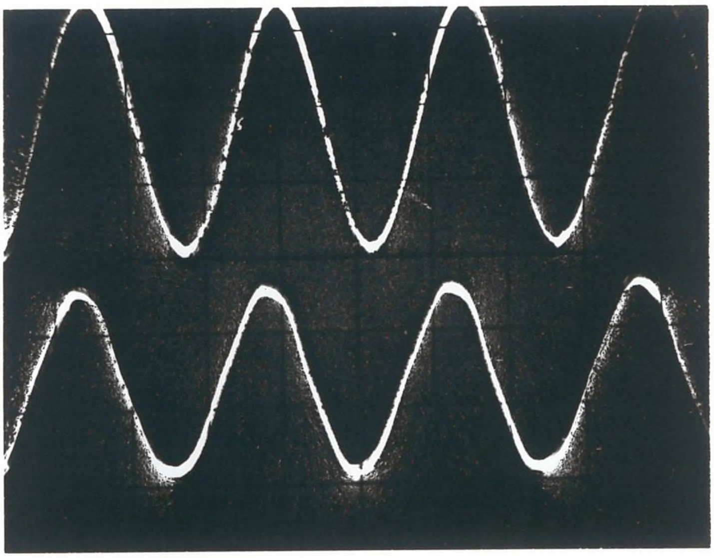


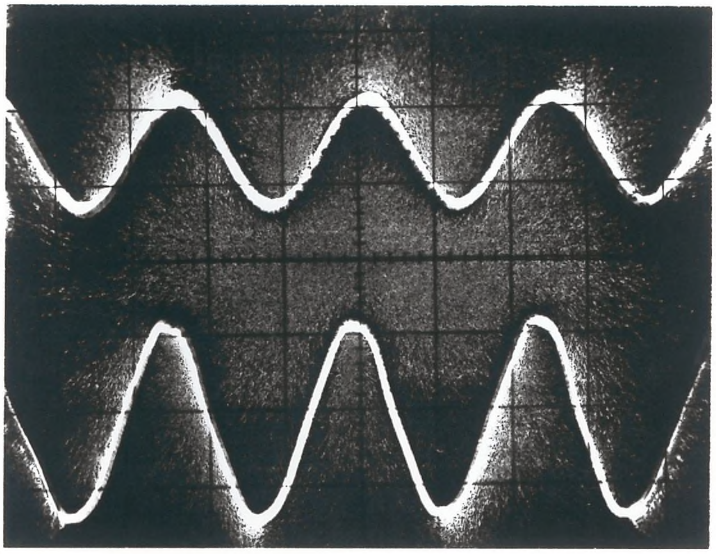
FIG. 5.10 PLATFORM RESPONSE TO A SINUSOIDAL INPUT



Displacement

Acceleration

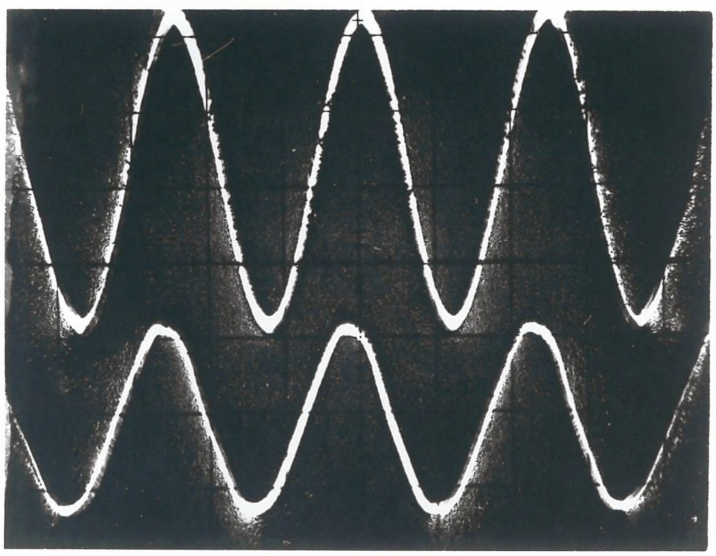
f = 1 Hz



Displacement

Acceleration

f = 3 Hz



Displacement

Acceleration

f = 2 Hz

FIG. 5.11 PLATFORM DISPLACEMENTS VERSUS ACCELERATIONS

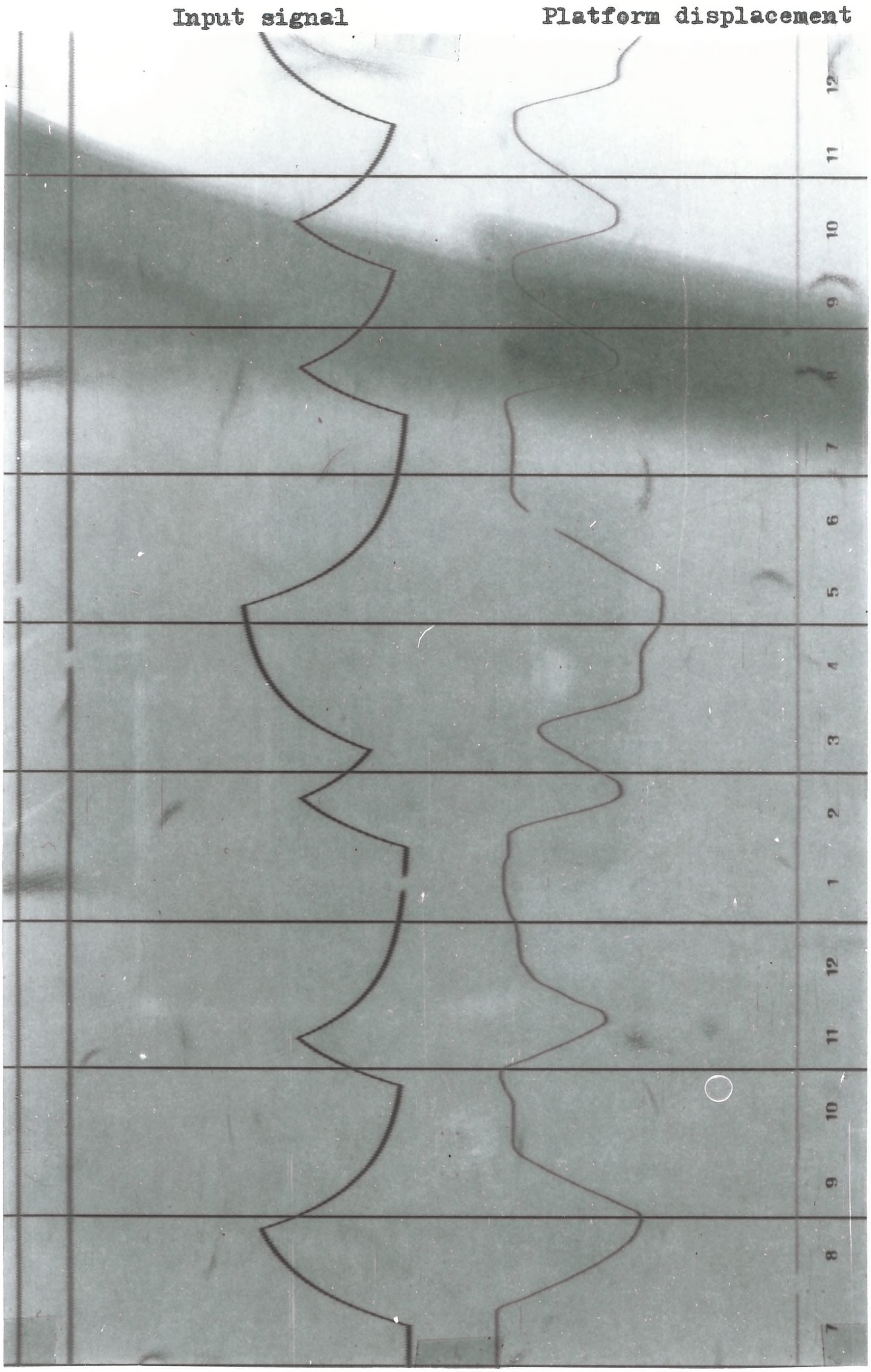


FIG. 5.12 PLATFORM RESPONSE TO A RANDOM SIGNAL

APPENDIX 5.1

VESTIBULAR APPARATUS

(Range of Acceleration Response)

The non-auditory section of the human inner ear is the recognised centre of motion sensors. This centre, the vestibular system, is one of the sensory system which provides information of body orientation and balance. The vestibular system, also called the labyrinth, lodges the sensors associated with maintenance of balance and orientation in three-dimensional space.

The principal organs of balance are the mutually perpendicular semicircular canals (which sense rotational velocities about 3 axes in the head) and the maculae of the utricle that, again in three dimensions, signal head position with relation to the terrestrial vertical, see figure A5.1.1. These two groups together with the saccule, whose role is not so clearly established, form the vestibular apparatus. There are two complete and independent systems, one on either side of the head. Their biological importance may be judged from the fact that they are protected by being deeply encased within one of the hardest bones of the skeleton, the mastoid process.

The semicircular canals are neither semicircular nor mutually at right angles, neither does any one canal lie in a single plane. This is of no functional significance, because the perception of motion includes a corrective programme invoked by the brain and developed since birth. The canals contain a liquid, endolymph, and a gelatinous obstruction, the cupola. When a canal is angularly accelerated about an axis perpendicular to its plane the endolymph moves with relation to the canal wall, distorting and moving the cupola, which in bending stimulates the nerve endings embedded in it. These nerves send frequency-modulated pulses to the brain, which represent, because of the physical time constant of the canal-endolymph system, the angular velocity of the head. The utricle, contains a gelatinous body in which small "stones", the otoliths, are embedded; hence if the head is moved, gravitational forces will act upon the otoliths, distort the utricula maculae, and as with the cupulae of the semicircular canals, neural signals will be sent to the brain.

Otolith organs have been shown to have potentially a dual function. They can serve as a pure gravity receptors, with an associated function as receptors for linear movement and centrifugal force, or as receptors for vibrations, or both - as in the case of the elasmo branch sacculus. In fact the otolith organ is fundamentally designed as a receptor for linear accelerations of all kinds.

The vestibular sensors are sensitive to accelerations, thus the input variable to the vestibular system is a vector having direction and magnitude. The output quantity, however, is not a vector in the strict sense. Information from the semicircular canals and the otoliths is sent to the central nervous system and thereafter an awareness of the sensation of motion is perceived. This perceived output variable preserves the indication of direction like a vector. However, sense of magnitude is applicable only on a comparative basis, where motions are faster or slower without an absolute scale. Perceived orientation is the most inportant sensation the human experiences in response to an input acceleration.

To summarise, the anatomy of the labyrinth displays two different groups of sensors. The semicircular canals are the human angular accelerometers while the otoliths are the linear motion sensor of the vestibular system. It must, however, be said that recent studies by Goldberg-Fernandez, suggest that the semicircular canals may also respond to linear accelerations.

Since the ability of human to orient himself and preserve equilibrium with the surrounding environment is a basic prerequisite for normal existence, a sizable research literature is devoted to various aspects associated with the function of the vestibular system.

In the following part of this appendix the dynamic characteristics of the vestibular sensors will be reviewed briefly. [MEIRY, 1966]. The semicircular canals are known to be heavily damped, angular accelerometers. Consequently, they respond as angular velocity meters over the frequency range from 0.02 cps to about 1.5 cps. The threshold of perception of angular acceleration are 0.14^{o}/sec for rotation about the Y_h (yaw) axis figure A5.1.2 and $0.5^{o}/sec^2$ about the X_h (Roll) axis. A similar value of threshold ($0.5^{o}/sec^2$) is presumably valid for rotation about the Z_h (Pitch) axis.

The otoliths are planar linear accelerometers sensitive to the specific force applied to the skull. Young and Meiry's (1968) revised dynamic otolith model using subjective response data described the otoliths as heavily damped linear accelerometers with an acceleration threshold. Statically they measure head orientation with respect to the apparent vertical quite accurately. This sense of orientation to the vertical is modified by habituation during prolonged stay in tilted positions. Dynamically the otoliths function as linear velocity metres and the computed bandwidth over which the cupula-endolymph system functions as a velocity transducer extends from 0.025 - 25 Hz a range encompassing the bandwidth of physiological head movement. The human threshold for perception of linear acceleration is near $2 \times 10^{-2}g$ representing an estimated displacement of 10^{-6}cm.

Table A5.1 summarises the results for the vestibular sensors as found in Meiry (1966).

TABLE A5.1

THE VESTIBULAR SYSTEM [MEIRY, 1966]

SENSOR	SEMICIRCULAR CANALS	UTRICLE
Input Variable	Angular acceleration	Specific force in the plane of the otolith
Sensitive axis	Sensitive to angular acceleration about an axis perpendicular to the plane of the canals	Sensitive to acceleration in the plane of the otolith
Output variable	Subjective sensation of angular velocity; vestibular nystagmus	Subjective sensation of Tilt and linear velocity counteroling, Eye movements
Sensor Transfer function	$H(s) = \frac{\text{subjective angular velocity}}{\text{input angular velocity}}$ Rotation about the sagittal Head Axis X_h (Roll) $HX_h(s) = \frac{7s}{7s + 1)\ (0.1s + 1)}$ Rotation about the vertical Head Axis Y_h (Yaw) $HZ_h(s) = \frac{10s}{(10s + 1)\ 0.1s + 1)}$ Rotation about the Lateral Head Axis Z_h (Pitch) $Hz_{(h)}s = \frac{7s}{(7s + 1)\ (0.1s + 1)}$	$\frac{\text{Subjective velocity}}{\text{input velocity}} =$ $\frac{Ks}{(10s + 1)(0.66s + 1)}$
Threshold of Perception	Angular acceleration $\alpha X_h = 0.5^o/sec^2$ $\alpha Y_h = 0.14^o/sec^2$ $\alpha Z_h = 0.5^o/sec^2$	Acceleration in the Plane of the otolith $\alpha_o = 0.005$ g

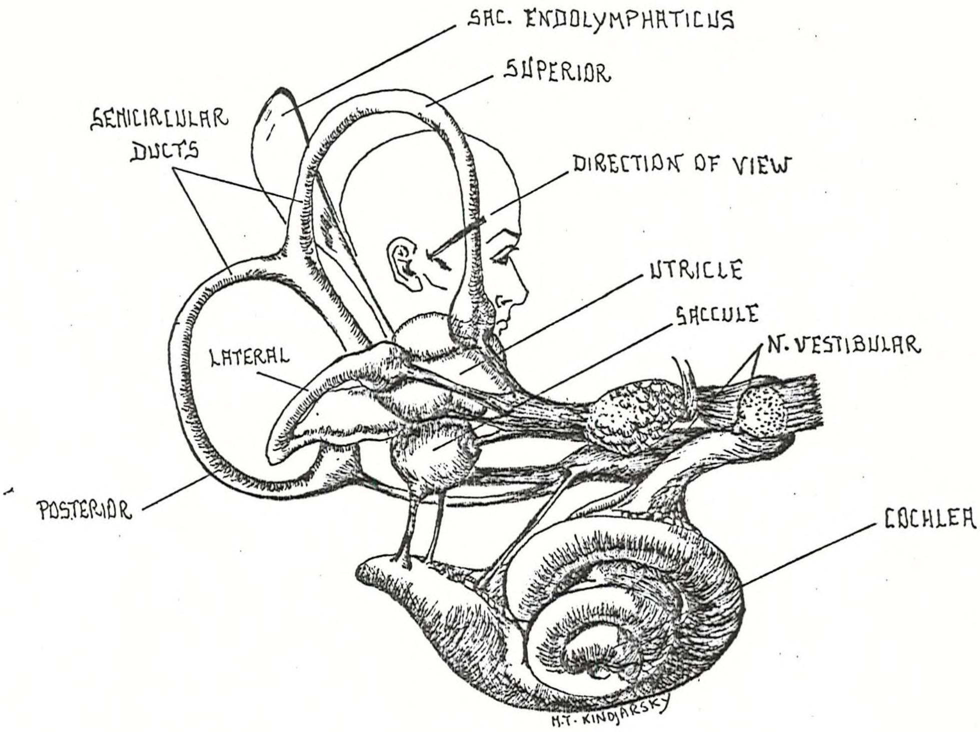


FIG. A5.1.1 THE VESTIBULAR APPARATUS

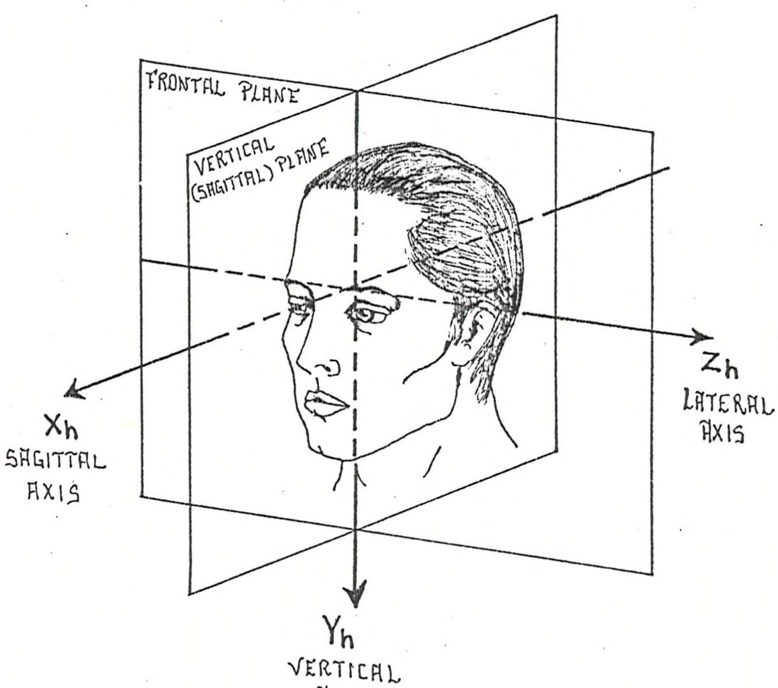
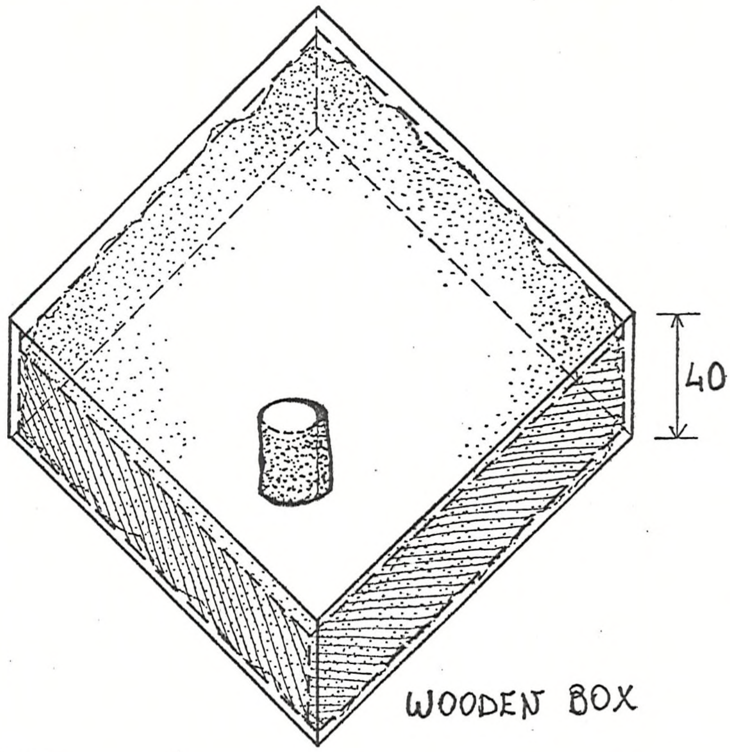
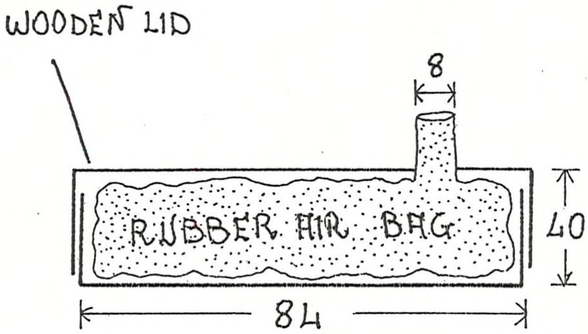


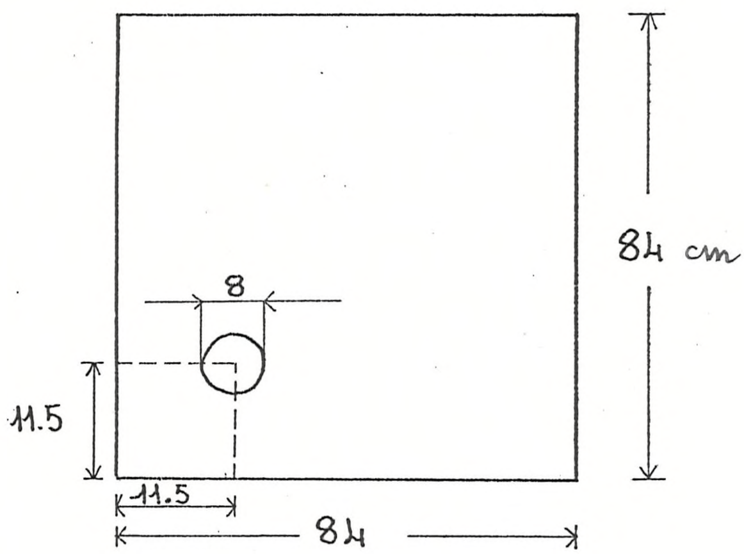
FIG. A5.1.2 THE HEAD PLANES AND THE HEAD AXIS SYSTEM

APPENDIX 5.2

DIMENSIONS OF THE PLATFORM WOODEN BOX







APPENDIX 5.3

PERFORMANCE AND STABILITY OF THE PNEUMATIC SERVO

The drive for the platform is a variant of the pneumatic servo. It is slightly unorthodox in that its power source is a centrifugal turbine pump, its control valve is a butterfly exhaust valve and the motor is a large volume bellows. Under critical operating conditions there would be a number of factors which would complicate the analysis of the system and aggravate stability considerations. However, there is such an abundance of pressure and power in the drive that variations about the quiescent conditions are so small that linearisation techniques may be used reliably and the system will behave like a more conventional system.

The following assumptions are made [BURROWS, 1972; McCLOY-MARTIN, 1973]:

(1) The supply gas pressure and temperature are constant

(2) Heat transfer to the gas is negligible

(3) The air obeys the perfect gas law

(4) No temperature change across the valve

(5) Volume changes in the bag are relatively small

(6) All flows occur under adiabatic conditions

(7) Pressure variations are small

(8) The external load has small variations

(9) The valve characteristics may be linearised in terms of a gain and internal resistance terms.

The most adverse stability conditions occur when the valve is in its neutral position (zero steady flow).
The open-loop transfer function may be written:

$$\frac{X_o}{e} = \frac{2GS'\gamma\ P/mV}{S\left|S^2 + \frac{C\gamma RT}{V}\ S + \frac{\gamma P\zeta S'^2}{mV}\right|}$$

$$= \frac{K}{S\ (S^2 + 2\zeta\omega_o S + \omega_o^2)}$$

where:

G = mass rate/valve travel slope of valve

C = mass rate/pressure drop slope of valve

S' = platform area

P = quiescent bag pressure

m = mass of load inertia

V = volume of air in bag

R = gas constant

γ = ratio of specific heats

T = Air temperature

At the neutral position : C = O. Thus the system is unstable or at best lightly damped. Three factors are known to improve stability ÷ leakage past the valve, compliance between valve and bag and a viscous friction component in the load. In practice these effects combined to give a reasonably well-damped system. The natural frequency of the load inertia on the air volume in the bag is given by:

$$\omega_o^2 = \frac{\zeta S'^2\gamma P}{VM}$$

APPENDIX 5.4

VIBRATING PLATFORM NATURAL FREQUENCY

Assumptions made in the analysis of the pneumatic system are given in Appendix 5.3.

From the ideal gas equation:

$$pV = RT$$

where:

p = pressure
V = volume
R = gas constant
T = absolute temperature

In adiabatic conditions, pV^{γ} = const: Where γ is the ratio of specific heats. It follows:

$$\gamma p dV + V dp = 0$$

i.e.

$$dp = \frac{-\gamma \cdot p}{V} dv$$

Since

$$dp = \frac{dF}{S},$$

$$dv = S\ dx$$

it follows:

$$dF = \frac{-\gamma \cdot p}{V} S^2\ dx = -K\ dx$$

where

$$K = \frac{\gamma p S^2}{V} = \text{stiffness } (N/m\cdot)$$

In our case:

S	=	Platform cross section	=	$0.70 m^2$
h	=	height of the platform	=	0.40 m
γ	=	ratio of specific heats	=	1.4
F_{MAX}	=	maximum vertical force	=	1250 N

Therefore:

$$K = 4375 \text{ kg/sec}^2$$

since

$$P = \frac{F_{MAX}}{S} \simeq 1800 \quad (N/m^2)$$

$$V = S \cdot h = 0.28 \quad (m^3)$$

For a single degree of freedom system:

$$\omega_o = \sqrt{\frac{K}{M}} = 2\pi f_o$$

where f_o = natural resonant frequency
M = mass

it follows:

$$\omega_o = \sqrt{\frac{4375}{100}} = 6.6 \text{ rad/sec}$$

and:

$$f_o = \frac{\omega_o}{2\pi} = 1.05 \text{ Hz}$$

APPENDIX 5.5

SPECIFICATIONS OF THE AIR BLOWER

Keith Blackman Limited

A member of the GEC-Woods Ltd Group
Registered Office Registered No. 31964 London
Mill Mead Road Tottenham London N17 9ND Telephone 01-808 4522 Telex 25259

O 16

ITEM NO.	1		
NO. OFF	ONE		
FAN SIZE	14/9 Series 23 Direct Drive	No series	25N
VOLUME	320 CFM	with 2½"	outlet.
PRESSURE	6.0 swg		
TEMPERATURE	68°F		
FAN R.P.M.	2900		
FAN B.H.P.	.47		
CASING	Mild Steel		
OUTLET	5¼ x 3½		
OUTLET VELOCITY	2500 Ft/Min		
INLET	5½" Dia with Guard		
CASING DOOR			
DRAIN IN CASING			
IMPELLER	Backward Curve		
FINISH	Sprayed Enamel		
BEARINGS			
BEARING SUPPORT			
PRICE PER SET £			
RECOM'D MOTOR B.H.P.	.75		
MOTOR R.P.M.	2900		
MOTOR & RAILS	D71		
MOTOR ENCLOSURE	TEFC		
ELECTRIC SUPPLY	415/3/50		
V-DRIVE			
DRIVE GUARD			
MILD STEEL BASE	Under Motor		
ANTI-VIBS.			

APPENDIX 5.6

LOW-PASS FILTER

The Solartron Random Signal Generator Mod. BO 1227, available in our department, gave at its output a random signal continuously changing between different levels. In order to obtain a continuous random signal which could be followed by the vibration platform, the signal generator output signal was passed through the circuit shown in figure A5.6.

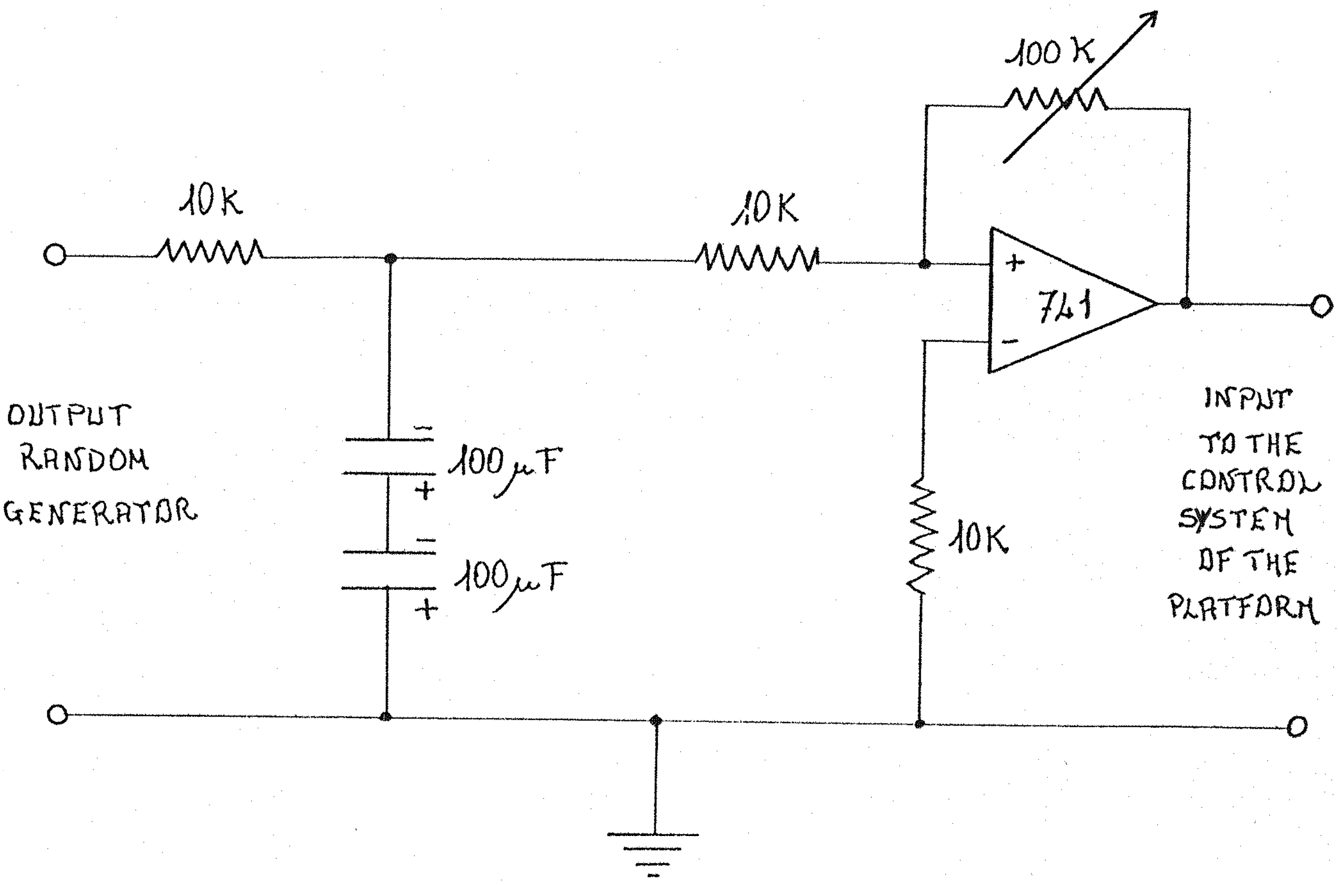


FIG. A5.6 LOW-PASS FILTER

CHAPTER SIX

THE MEASUREMENT SYSTEM

6.1. INTRODUCTION

This chapter discusses the measuring devices and techniques to be used in subsequent tests. Initially the following variables were selected as being of particular significance.

1. Vertical displacement of the seat.
2. Vertical displacement of the shoulder.
3. Vertical displacement of the hand (arm outstretched).
4. Vertical acceleration of the seat.
5. Vertical acceleration of the shoulder.
6. Vertical acceleration of the hand.
7. Electromyograms from relevant muscle groups.

Some of the equipment described here was of an exploratory nature and was not used extensively in the experiments. However, an account of its development is included for completeness and because there were original aspects.

6.2. PLATFORM AND SEAT DISPLACEMENT

The platform was mechanically constrained to move vertically and so the displacement was simply measured by a linear potentiometer. Details are:

Type: Bourns Long Life 12.5 cm. Potentiometer Sensitivity: 0.5%.

The reference voltage was supplied from a ± 15 Volt stabilized D.C. supply. The potentiometer served the double purpose of feedback sensor and displacement transducer.

6.3 ACCELERATION

In physiological experiments involving human subjects performing manual functions under near normal conditions, it is essential that sensors and leads used to record accelerations, etc. be as small and light as possible so as to cause as little loading or distraction to the subject. Care must also be taken when siting the instruments and attaching them if the output is to be meaningful. Several kinds of inertia measuring devices are possible but small strain gauge accelerometers were preferred for their response characteristics, size, interference rejection and cost. In particular, it is important that orthogonal vibrations of the platform do not produce significant response from the transducers.

Accelerometer type: TML-AM2 made by Tokyo Sokki Kenkyujo Co., Japan was selected.

Their dynamic characteristics are shown in Figure 6.1 and manufacturer's details are given in Appendix 6.1.

The devices use the suspended cantilever principle with resulting forces being converted to electrical outputs by a strain gauge bridge. Each accelerometer weighs 13 grams only and has flat frequency responses up to 20 Hz where a slight resonance (2 db) occurs. This is well within the dynamic specification required in the tests.

The associated circuits are shown in Figure 6.2. Two legs of the strain gauge bridge were connected to a bridge driver while the other two legs were connected to a synchronous detector thus giving a rectified D.C. output voltage proportional to the acceleration. Balancing and calibration are achieved by means of the potentiometer P_1. When the device was tested in situ by impact methods it was confirmed that the frequency response was flat up to 20 Hz.

The calibration was achieved by comparison with the known gravitational field. The full procedure is outlined in Appendix 6.2. Calibration, see Appendix 6.1, also showed the input/output characteristic to be linear throughout the frequency range of interest. Phase-shifts were negligible. It was concluded, therefore, that the instrument was suitable for use even with non-sinusoidal inputs.

6.4. HAND DISPLACEMENT BY ULTRASOUND METHOD

Use of ultrasound methods for detecting vertical hand displacement was considered ideal since no physical linkage between the transducer and its recording system is required. An experimental system was devised involving a transmitter and a receiver, one of which was attached to the hand. The time taken for impulses to travel between these elements gave a measure of the vertical displacement. Fig. 6.3 shows schematically the layout of the complete transducer system. Of course it is essential that the velocity of ultrasound in the separating medium be known and constant.

Complete circuit diagrams for each transducer are shown in Figures 6.4 and 6.5. The transmitter, Figure 6.4, consists basically of a free-running multivibrator and a square wave shaping circuit which drive the ultrasonic generator. The receiver, Figure 6.5, consists of an amplifier with the ultrasonic pick-up, a pulse width generator, mono-stable multivibrator, a ramp generator and a sample-and-hold clamping circuit whose output is proportional to the transmission time. This output can be fed to any suitable recording instrument. The characteristics of the ultrasound transducers are given in Appendix 6.3. They are bi-directional, linear and resonate at 40 kHz. The electronic elements in the circuit are in cascade and so their sensitivities add. Design calculations and measurements, Appendix 6.4, show that the detector was capable of 2 mm resolution which was considered adequate for the experiments. Figure 6.6 shows the complete experimental sensor system.

6.5. SHOULDER DISPLACEMENT BY CAPACITANCE METHOD

Again it is essential to have free movement and light loading and it was an ideal application for a capacitive position transducer devised by TODD (1970) and CODD (1975). In this transducer a spring, stretched by a light cord, moves freely inside a brass sheath as shown in Figure 6.7. The resulting change in capacitance between the spring and the sheath is used in a resonant circuit tuned at a fixed frequency of 10 kHz. Attenuation in the transducer output then gives a measure of the change in capacitance and hence of the spring extension.

Figure 6.8 shows a photograph of the transducer, while Figure 6.9 shows the associated electronic circuitry. A plot of measured detector output voltage against spring displacement shown in Figure 6.10 is surprisingly linear over the range of interest and the sensitivity is considered adequate.

6.6 ELECTROMYOGRAPHIC (EMG) SIGNAL PROCESSORS

Electromyographic (EMG) signals give an indication of the electrical activity in a muscle whenever there is a voluntary or involuntary contraction. EMG signals, usually detected by surface electrodes, are gross records of the firing of individual motor units. The waveform consists of a large number of impulsive spikes varying in amplitude from a few μvolts up to several hundred μvolts and ranging in average frequency from a few Hz up to several kHz. When a substantial muscle mass between recording electrodes is active, the signal produced is influenced both by the number of fibres in the muscle mass and their rate of contraction. The lowest signal level occurs during virtually complete relaxation and the highest during a tetanic (sustained) contraction or spasm. An average value of the waveform is considered to give a measure of the mechanical state of the muscle. Ideally an infinite integration time would give a good measure of a constant state but in practice, since the mechanical state changes, the averaging time must be a compromize between smoothing out the random fluctuations and following dynamic changes in the muscle. In practice, a balanced output is achieved by comparing differentially the EMG voltages from two recording electrodes placed on the opposite sides of the muscle mass, a suitable neutral electrode providing a reference potential.

Two forms of EMG processors have been used here. The EMG integrator shown in Figure 6.11 consists of an ideal resettable integrator (CHANDLER, 1973) while the EMG amplifier-filter of Figure 6.12-6.13 consists of a differential follower input stage, a high gain difference amplifier circuit, 50 Hz notch filter and an averaging filter (WILLIAMS et al, 1977; GARLAND 1972; FERRAIOLI 1973; 1974). Details of the averaging filter are reported in Appendix 6.5.

In the EMG integrator the averaging time could be varied between fast and slow regimes.

In this work only empirical methods have been used to determine suitable filter characteristics. However, there have been attempts by other workers (BRODY et al 1974; BATTYE et al 1955; BASMAJIAN 1958; HERTZ et al 1973; GOTTLIEB et al 1970; SCOTT 1967; HARBA, 1981) to define the frequency content of the useful information in the EMG signal and hence to design optimum filters. The author considers that spurious noise caused by unwanted movement and particularly mains interference causes the most significant problems and accordingly the system shown in Figures 6.12-6.13 was preferred. Frequency response curves are shown in Appendix 6.6.

Figure 6.14 shows a view of the EMG processors and the other instrumentation described in this thesis.

CHAPTER 6

LIST OF FIGURES

FIG. 6.1 ACCELEROMETER SELECTED

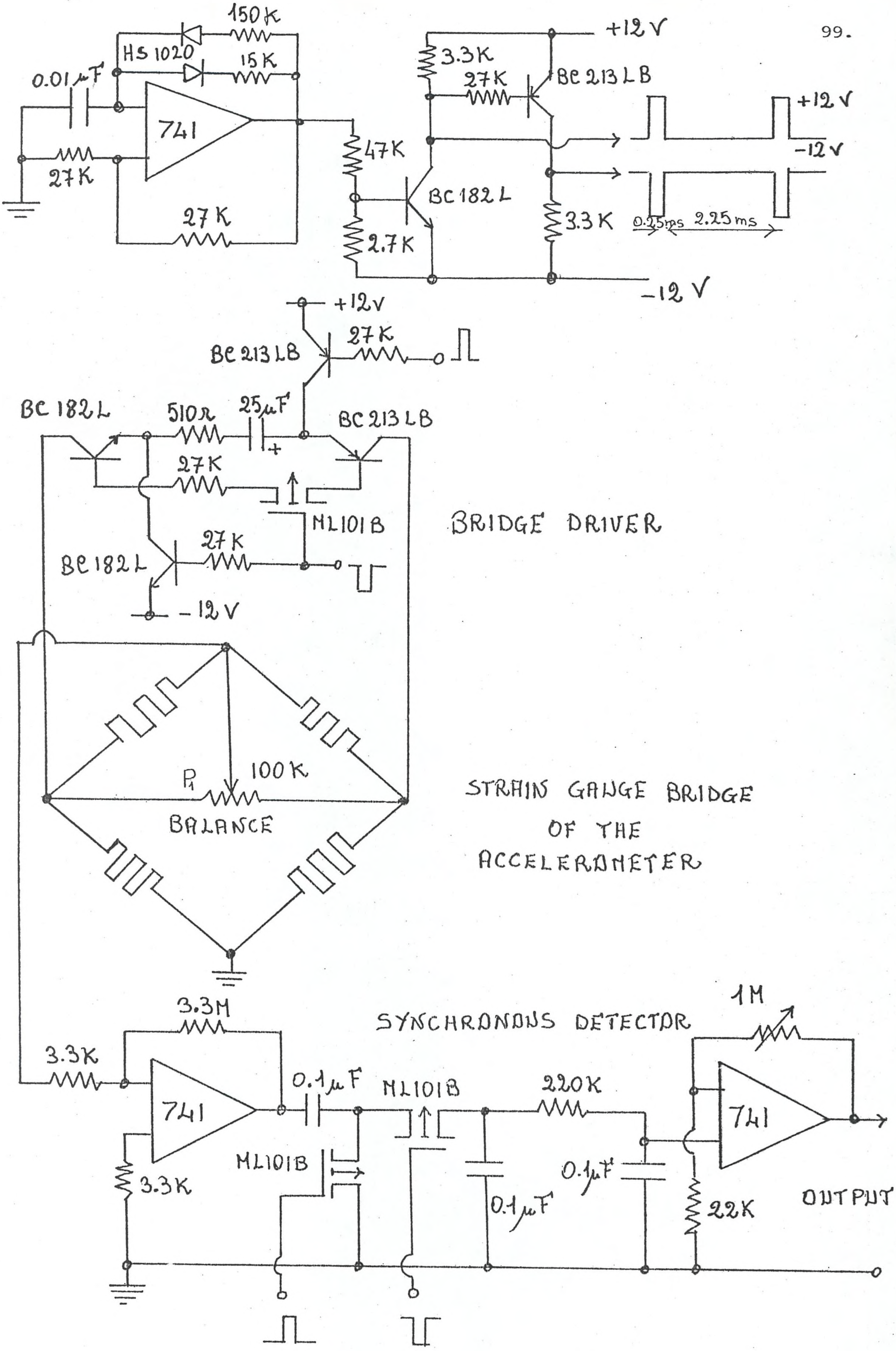


FIG. 6.2 CIRCUIT DIAGRAM FOR THE ACCELEROMETER

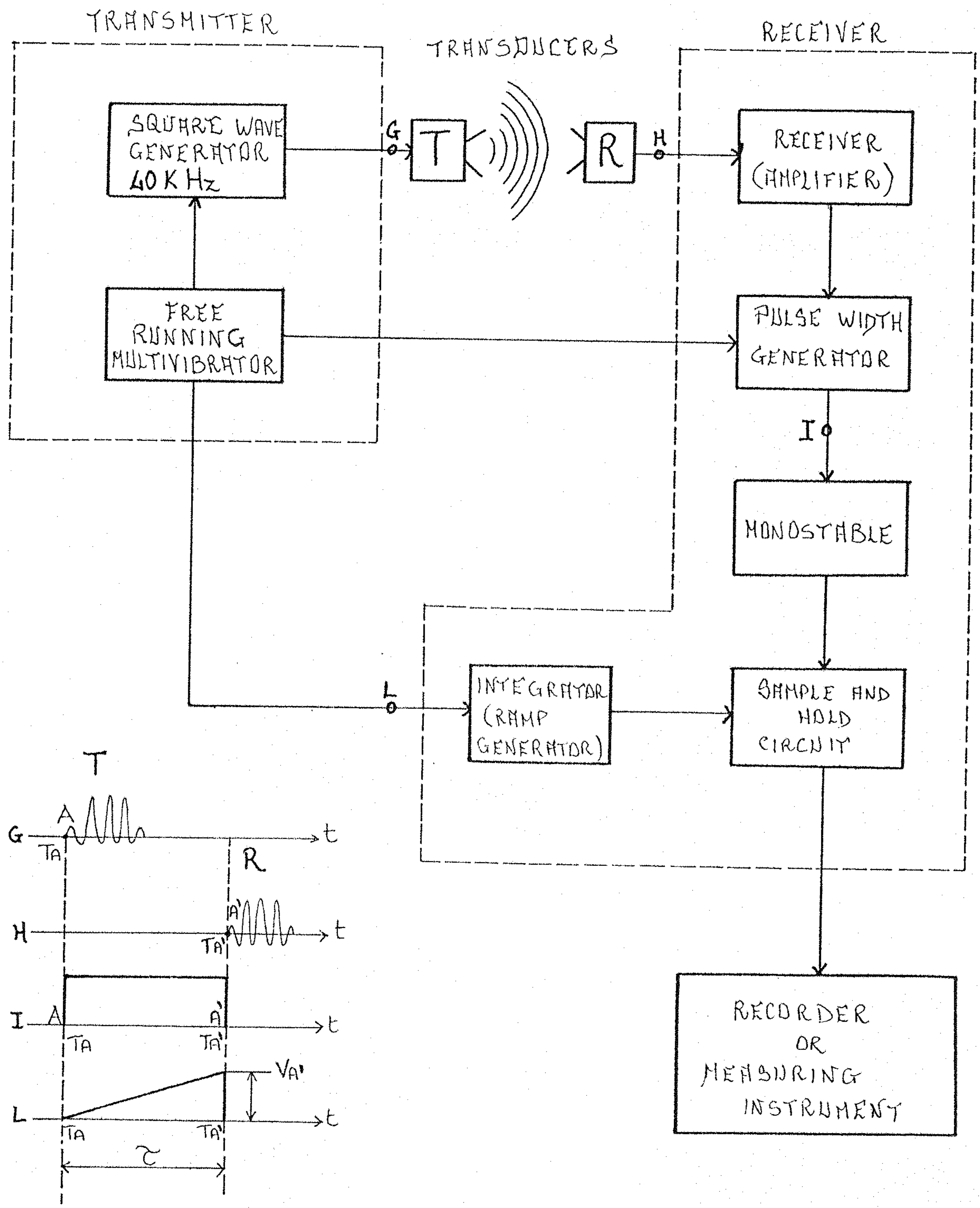


FIG. 6.3 SCHEMATIC DIAGRAM OF THE ULTRASOUND DEVICE

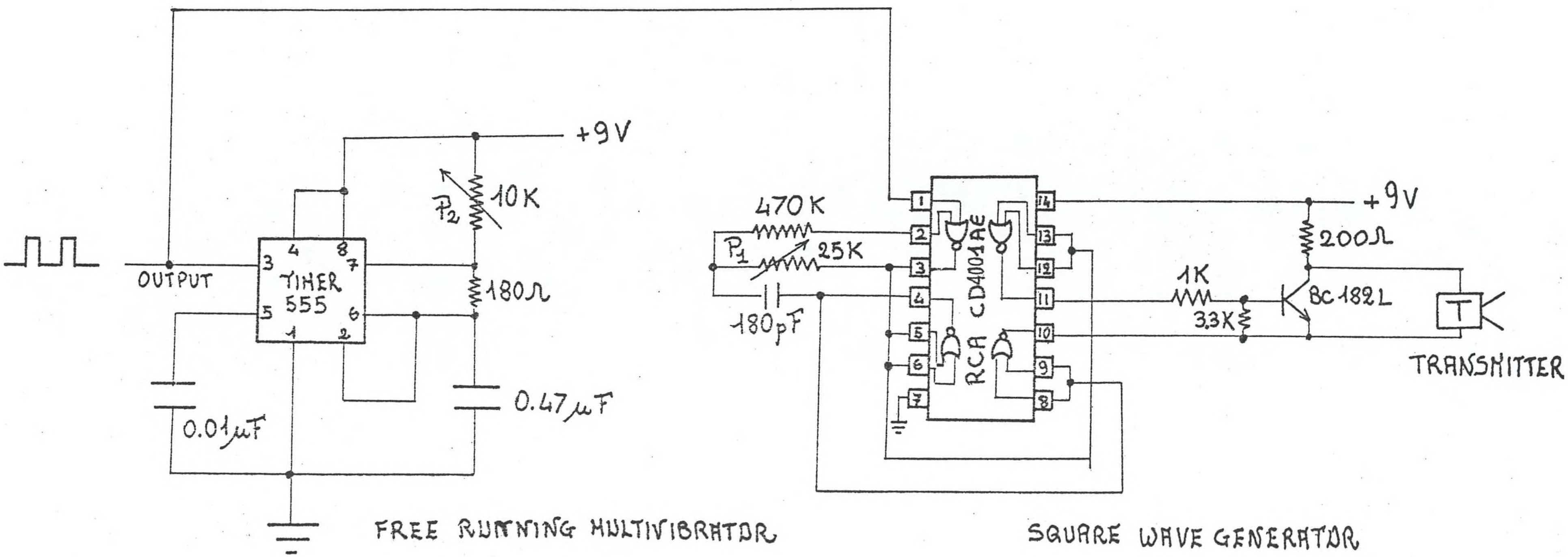


FIG.6.4 CIRCUIT DIAGRAM OF THE TRANSMITTER

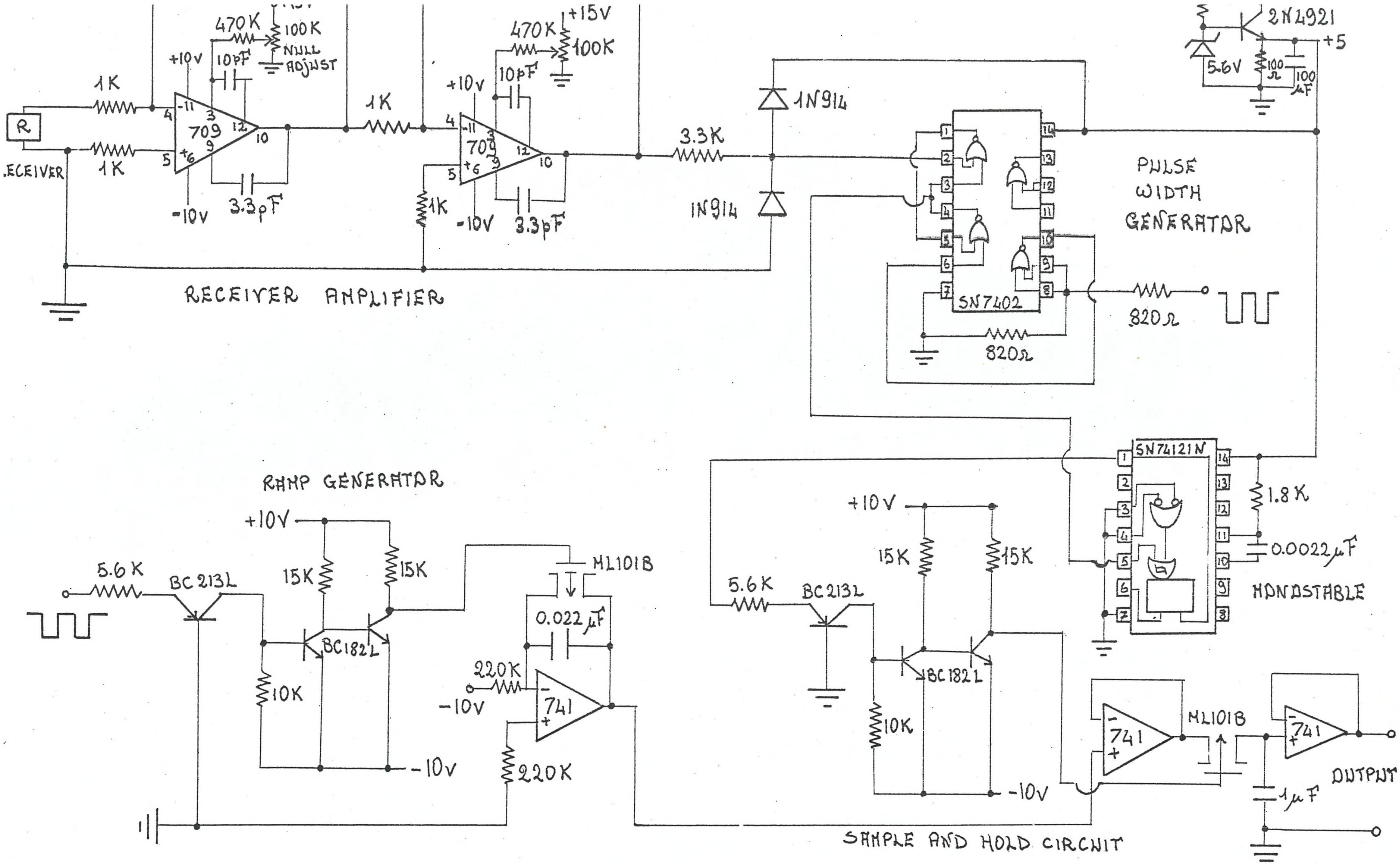


FIG. 6.5 CIRCUIT DIAGRAM OF THE RECEIVER

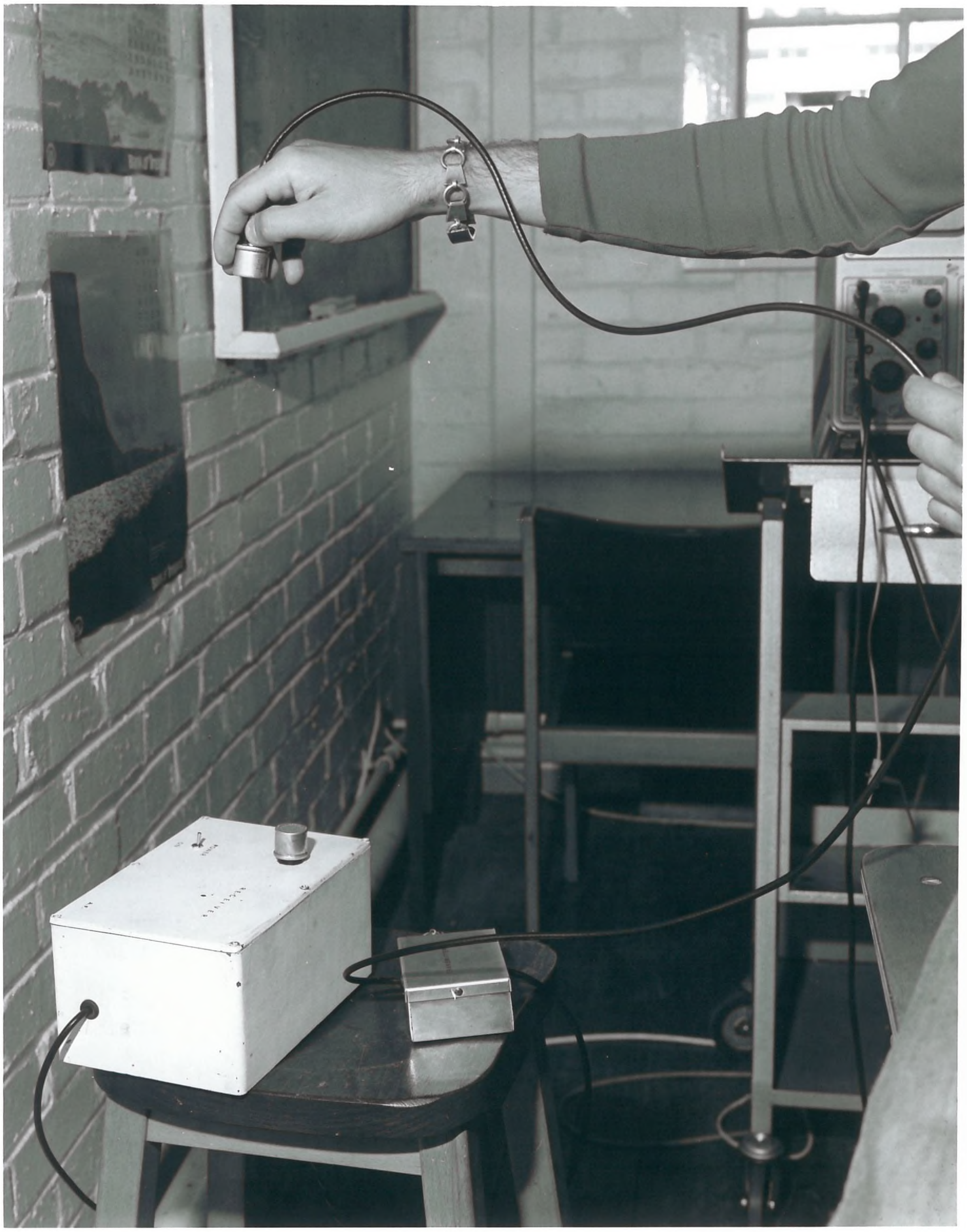


FIG. 6.6 ULTRASOUND DEVICE

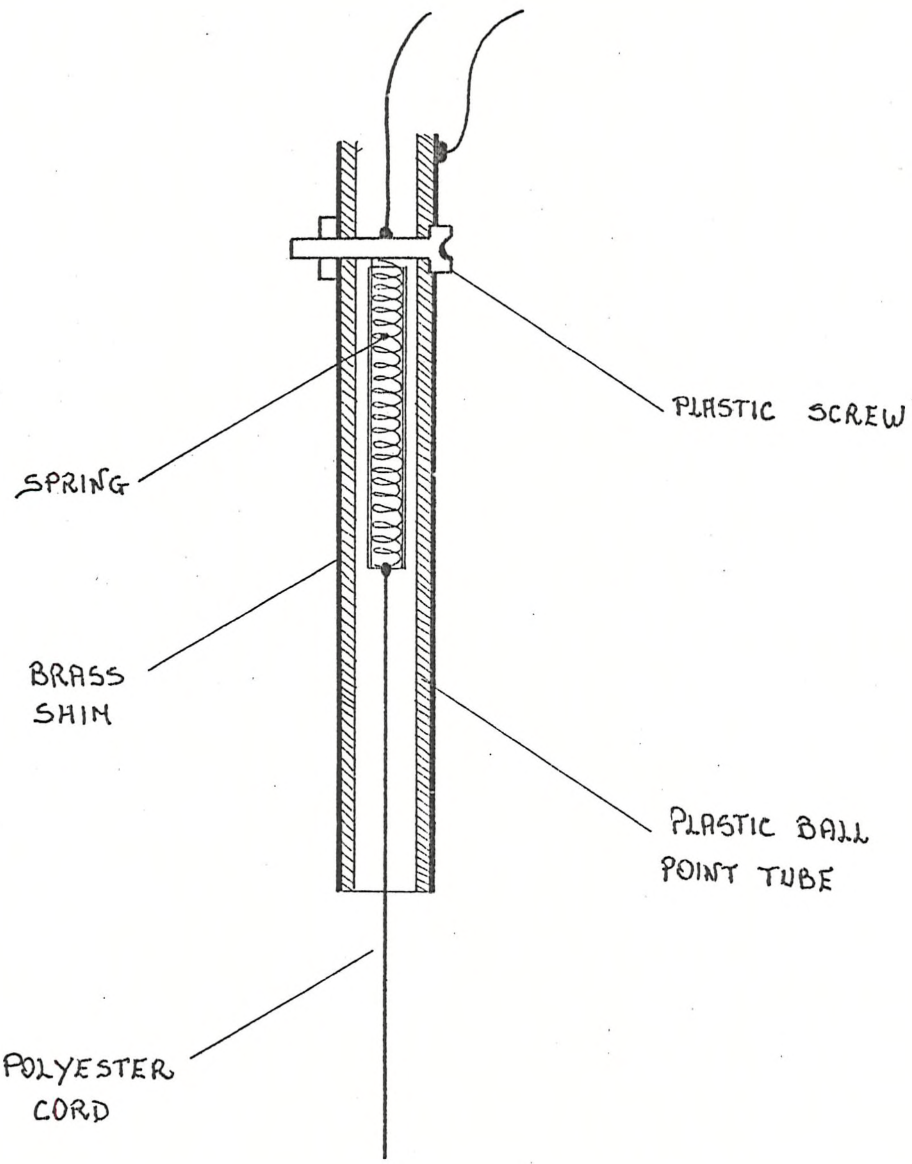


FIG. 6.7 SECTIONAL VIEW OF THE POSITION TRANSDUCER FOR THE SHOULDER DISPLACEMENT

FIG. 6.8 POSITION TRANSDUCER FOR THE SHOULDER DISPLACEMENT

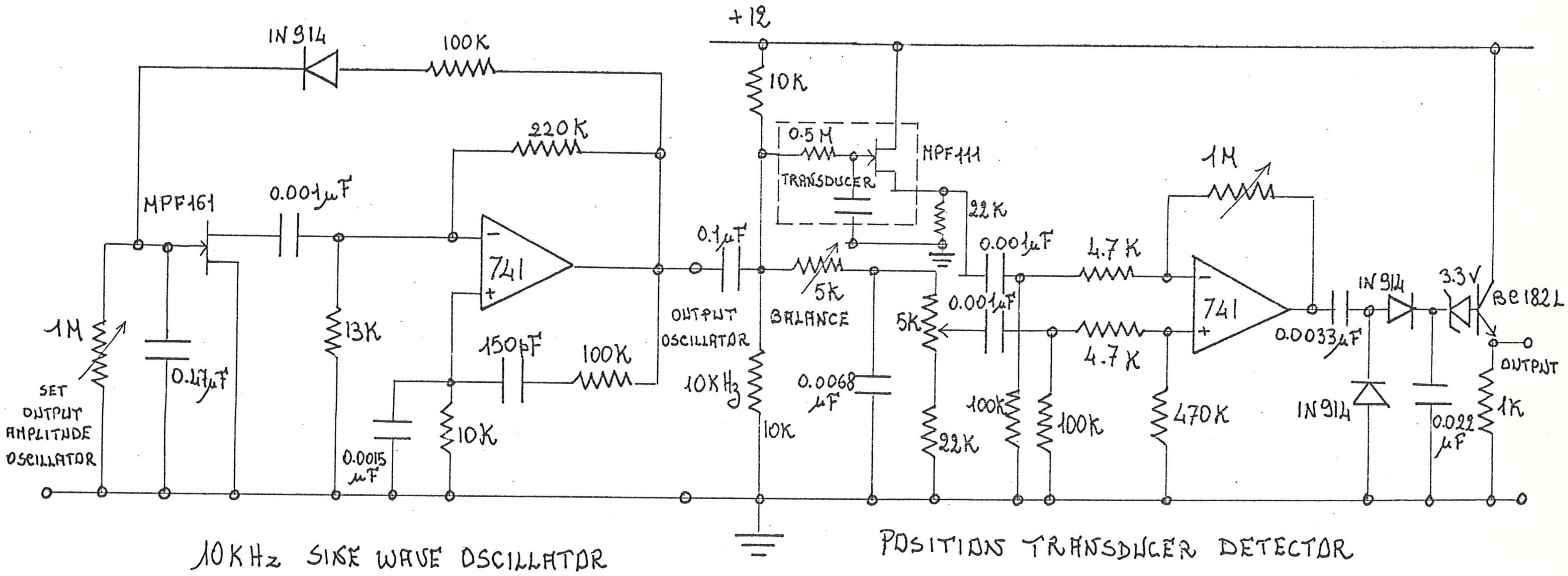


FIG.6.9 CIRCUIT DIAGRAM OF THE POSITION TRANSDUCER

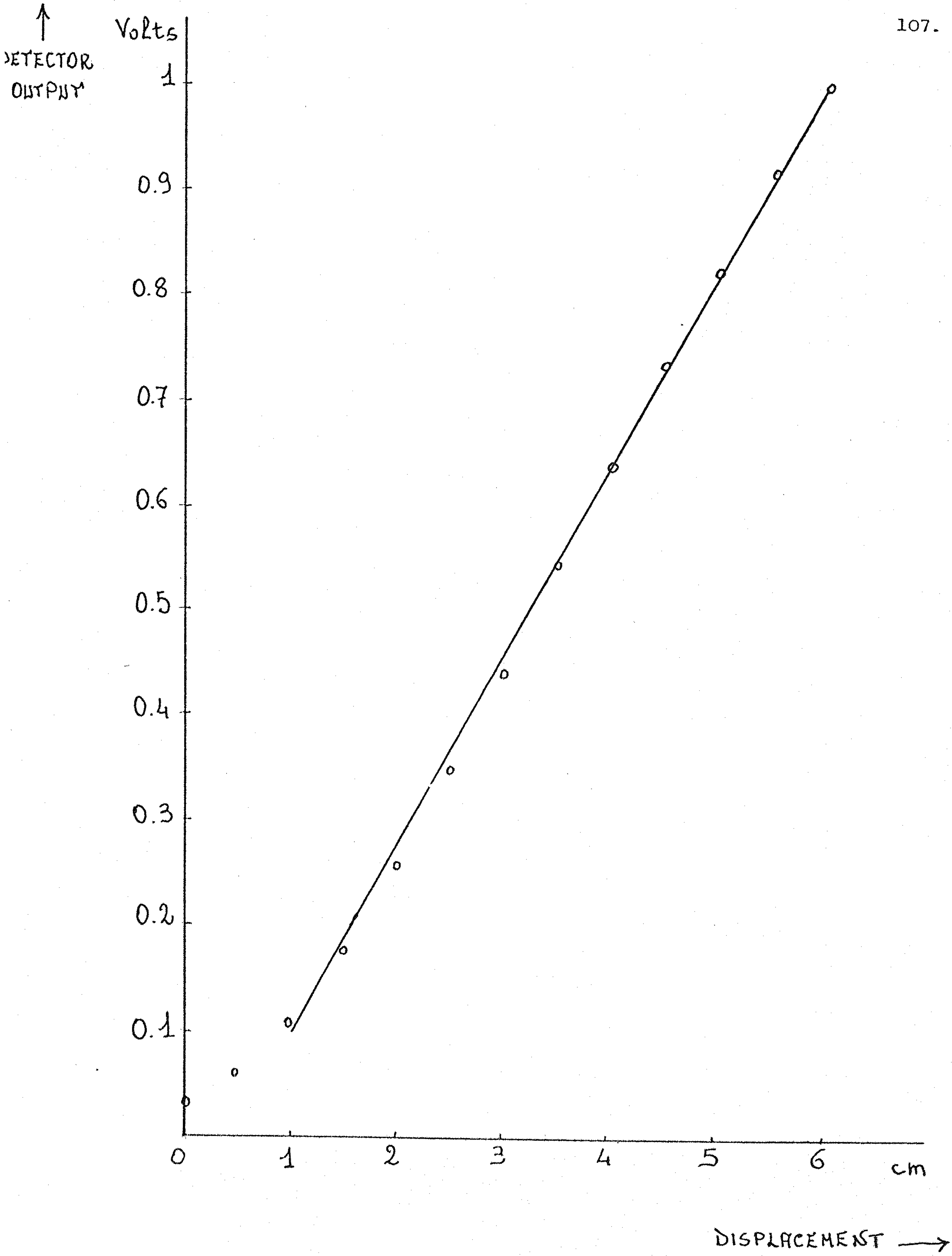


FIG. 6.10 PLOT OF THE DETECTOR OUTPUT VERSUS EXTENSION FOR THE POSITION TRANSDUCER

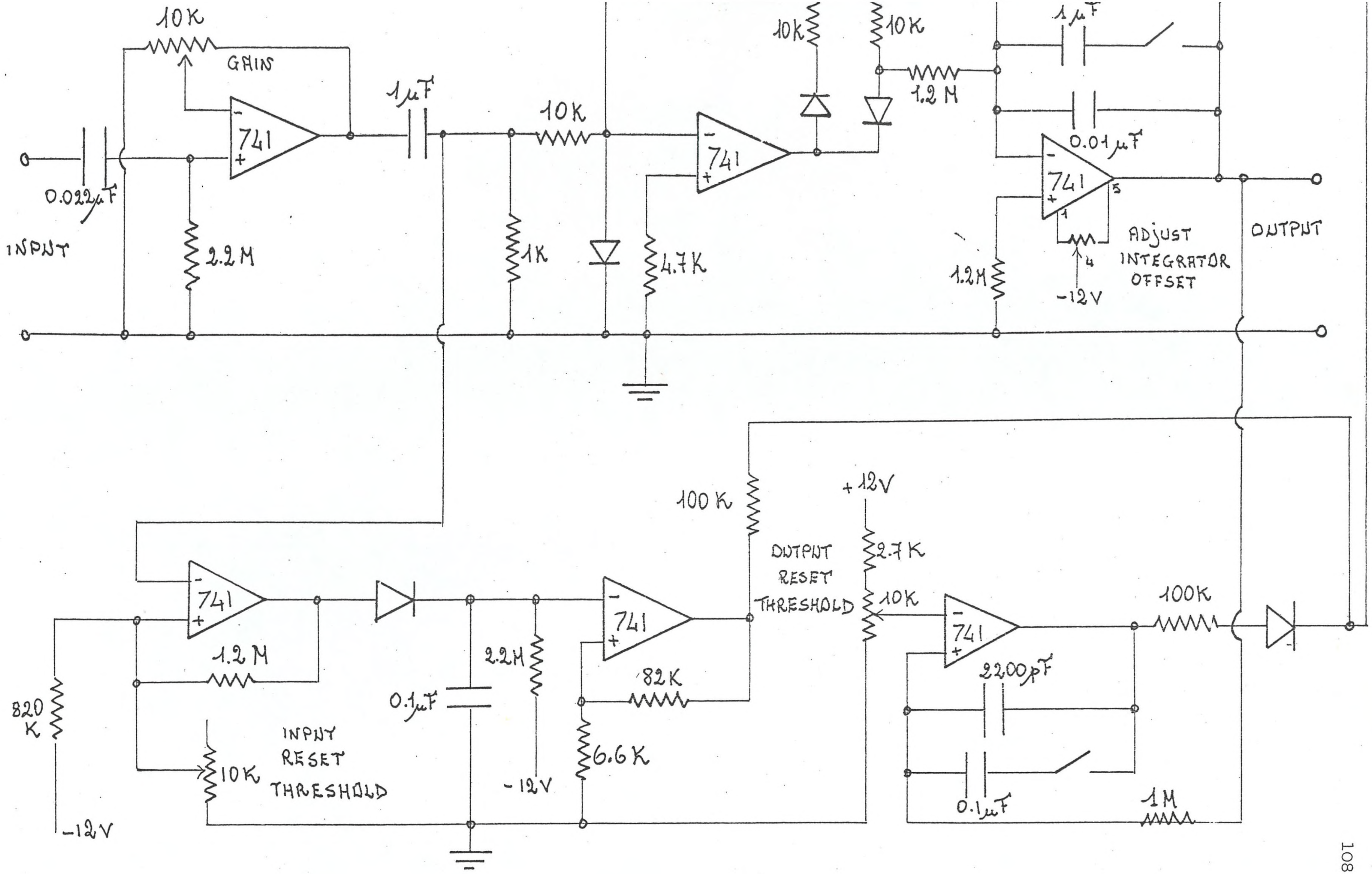


FIG.6.11 CIRCUIT DIAGRAM FOR THE EMG INTEGRATOR

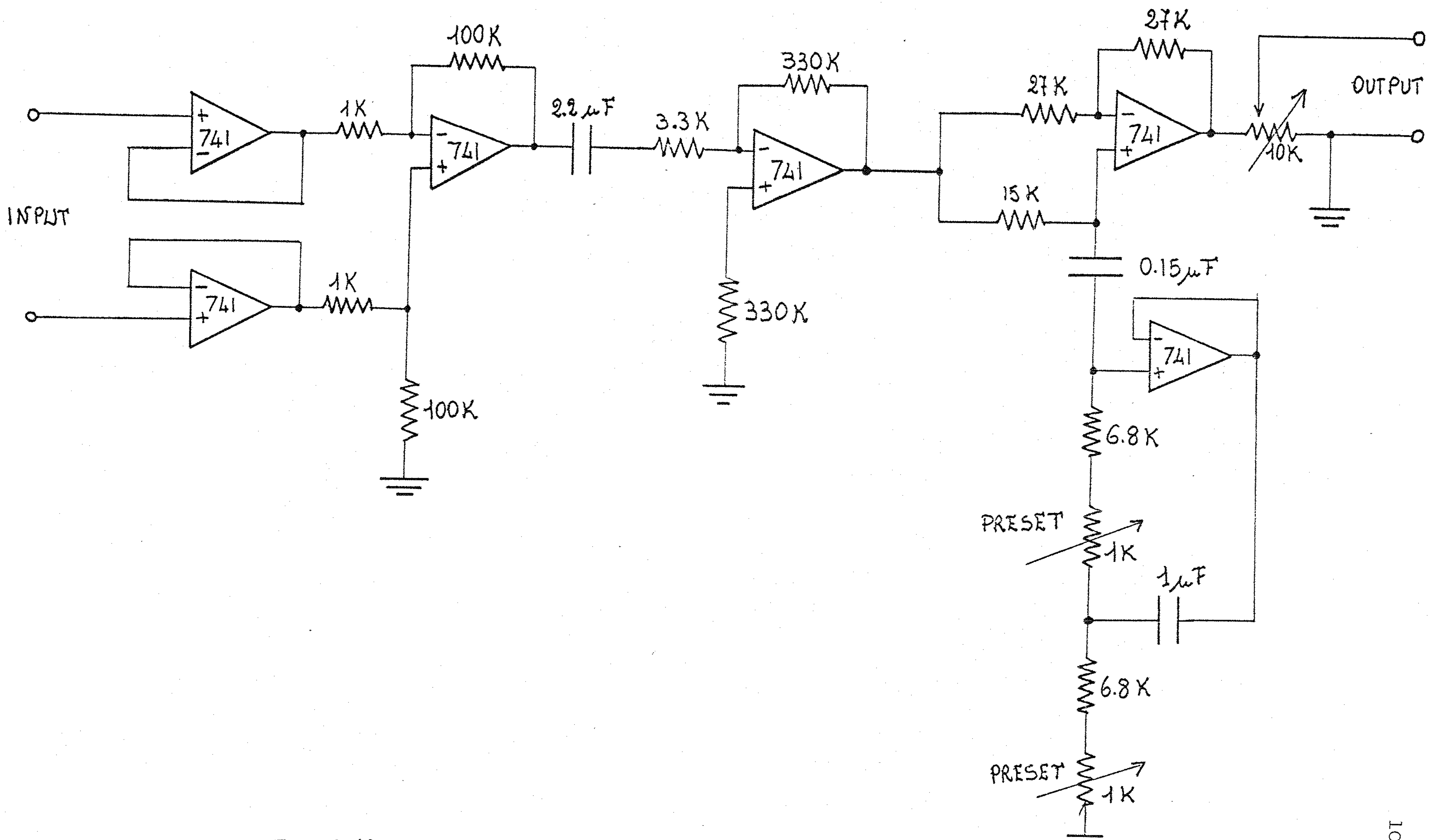


FIG. 6.12 EMG AMPLIFIER-FILTER

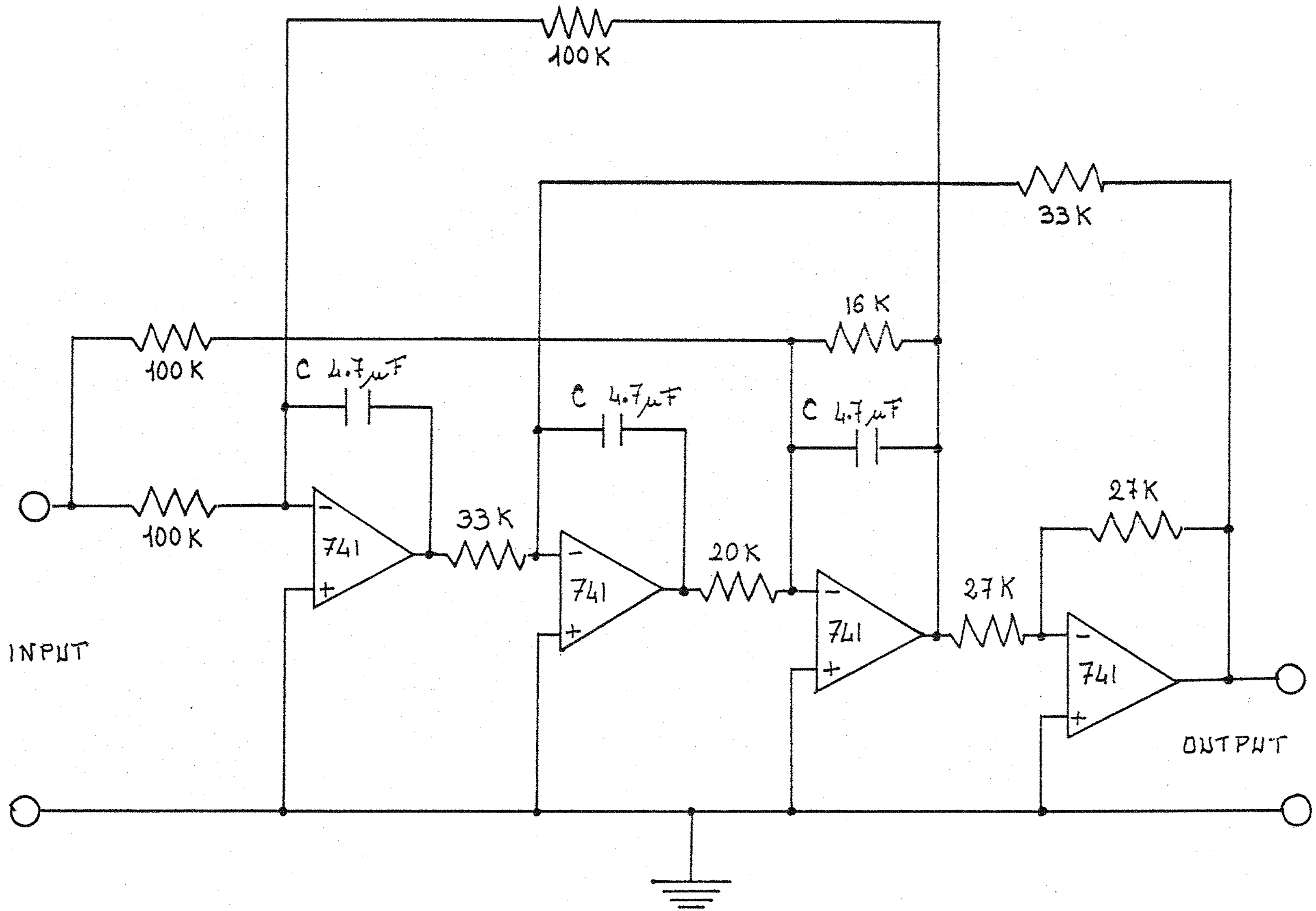


FIG. 6.13 EMG AVERAGING FILTER

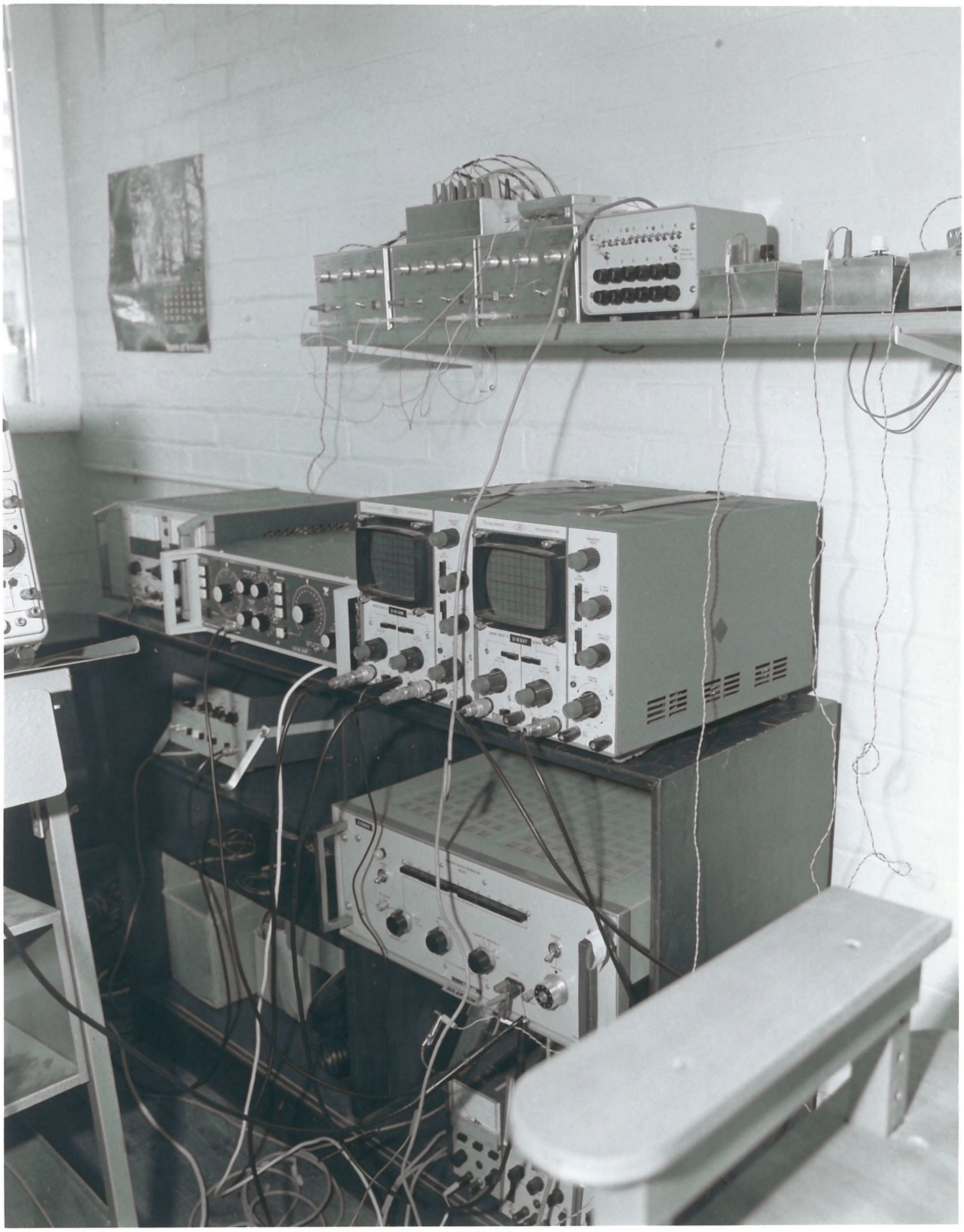


FIG. 6.14 AMPLIFIERS AND EMG PROCESSORS AND OTHER INSTRUMENTATION

APPENDICES

CHAPTER 6

APPENDIX 6.1

CHARACTERISTICS OF THE ACCELEROMETERS USED AND THEIR TEST AND INSPECTION DATA

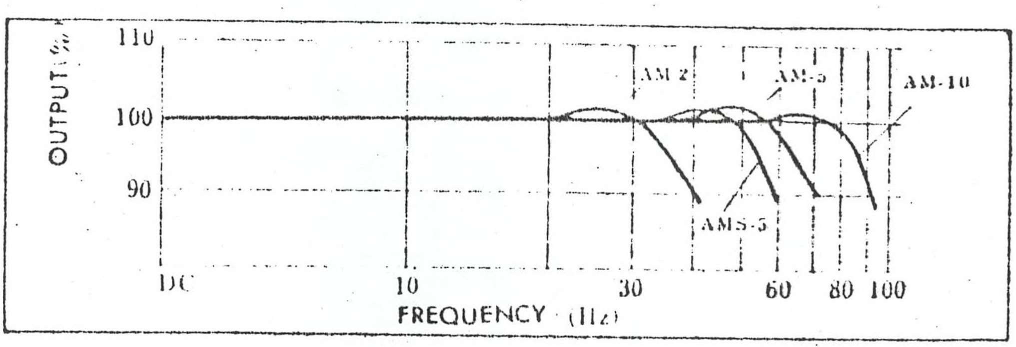
SPECIFICATION

ITEM \ TYPE	AM-2	AM-5	AM-10	AMS-5
RANGE (G)	2	5	10	5
RESPONSE (Hz)	30	60	80	60
NATURAL FREQ. (Hz)	50	90	130	90
BRIDGE CONFIGURATION	120Ω FULL BRIDGE			
RATED OUTPUT (mV/V)	1			25
CROSS SENSITIVITY	less than 0.5% F.R.O.			
TEMPERATURE EFFECT ON ZERO	less than 0.1% FRO/°C			
DAMPING	0.7 at 20°C Silicon Oil			
OVERLOAD RATING	150% F.R.O.			
WEIGHT (Gr.)	13			
OUTPUT CONNECTION	1m. of 0.08mm^2(0.12dia x 7) x 4 Core Shielded Cable			

LINEARITY ±0.5% FS
CROSS SENSITIVITY 0.5% FS.
EXCITATION VOLTS 2V NOM. 5V MAX.
TEMP RANGE 0-40°C.
RESIDUAL UNBALANCE $\pm 2000 \times 10^{-6}$

FREQUENCY RESPONSE

CASE MATERIAL PHENOLIC RESIN.



Customer Techni Measure Type AM-2

Customer's Order No. T/219 Serial No. 00815

APPLICABLE SPECIFICATIONS ☑ TML Standard
☐ Special Order

===

TEST DATA

Sensitivity 1096 $X10^{-6}/G$ (G.F.=2.00)

Rated Output ☑ 1.096 mV/V

Non-linearity ☑ 0.5 %/F.S.

Rated Acceleration 2 G

Allowable Acceleration 3 G

Bridge Configuration 120Ω Full Bridge

Electric Connections

☑ Input - RED & BLACK

☑ Output - GREEN & WHITE

===

INSPECTION ☑ Workmanship

[illegible]

===

REMARKS

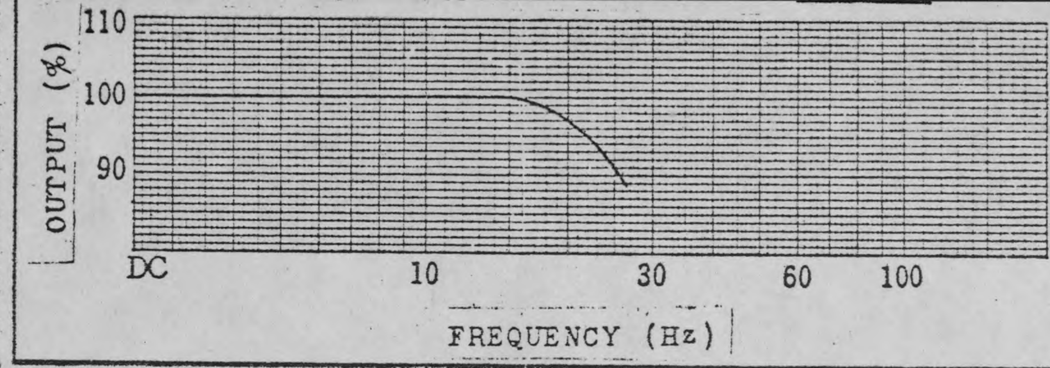
===

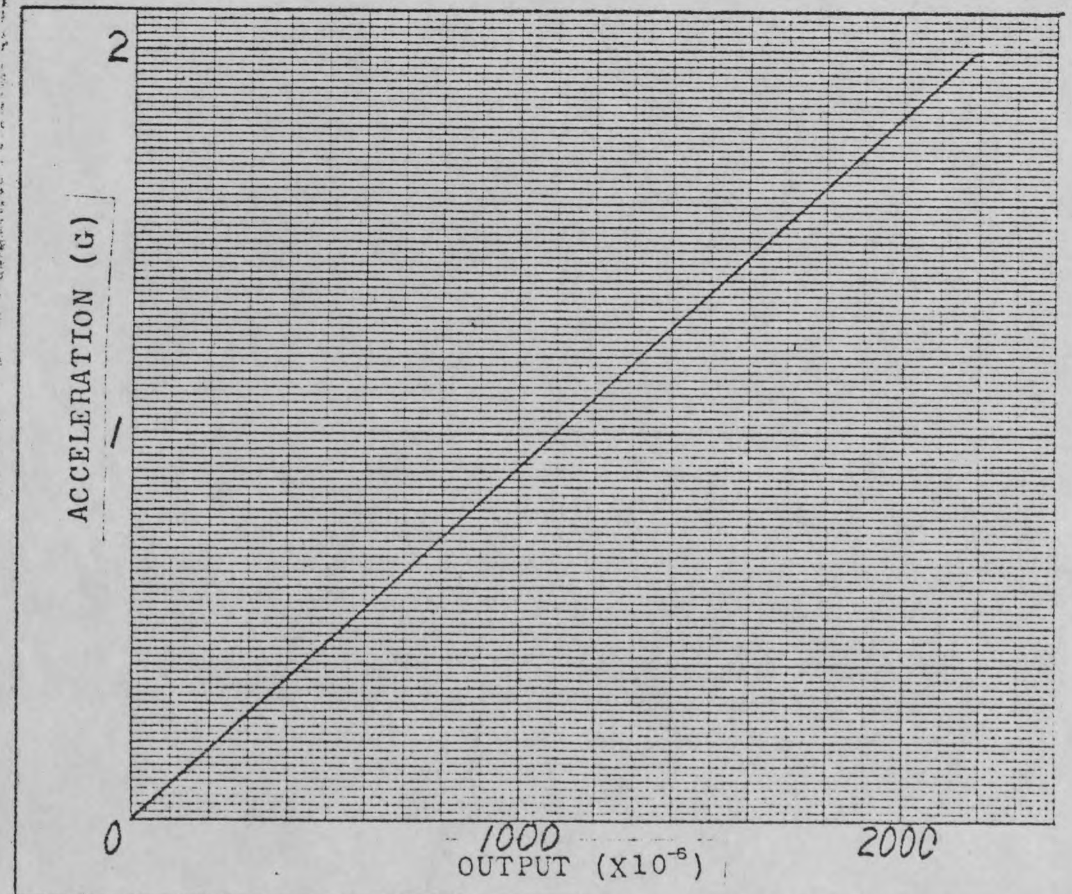
ACCEPTANCE Tested by T. Kubota Date 31.10.'75

Inspected by M. Suzuki Date 31.10.'75

Tokyo Sokki Kenkyujo Co., Ltd.

8-ban 2-go, 6 chome, Minami-ohi, Shinagawa-ku, Tokyo





Customer Techni Measure Type AM-2

Customer's Order No. T/219 Serial No. 00816

APPLICABLE SPECIFICATIONS ☑ TML Standard ☐ Special Order

TEST DATA

Sensitivity 914 $X10^{-6}$/G (G.F.=2.00)

Rated Output ☑ 0.914 mV/V

Non-linearity ☑ 0.5 %/F.S.

Rated Acceleration 2 G

Allowable Acceleration 3 G

Bridge Configuration 120Ω Full Bridge

Electric Connections

☑ Input - RED & BLACK

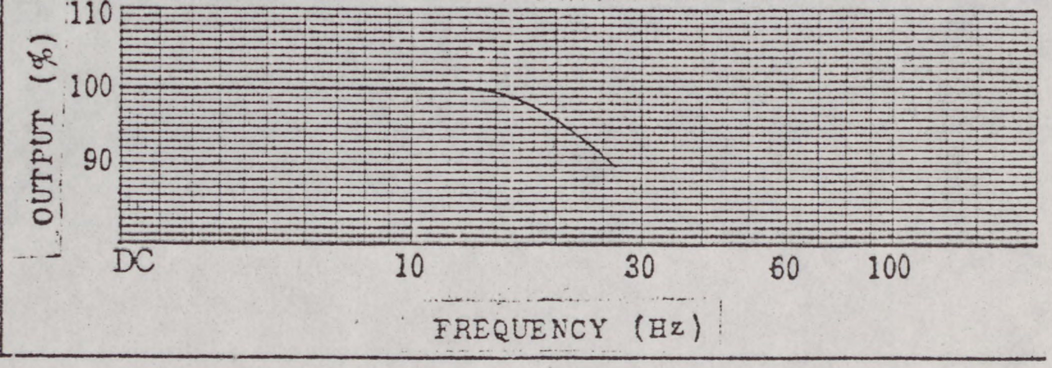
☑ Output - GREEN & WHITE

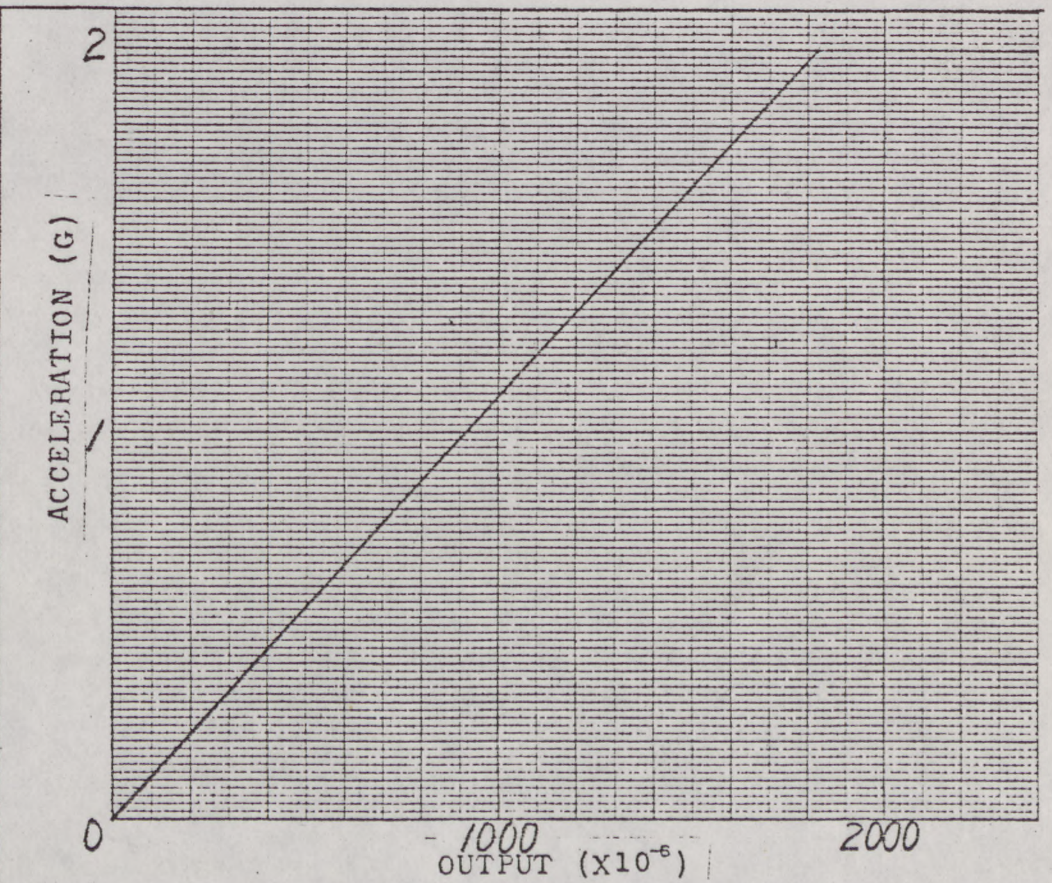
INSPECTION ☑ Workmanship

REMARKS

ACCEPTANCE Tested by T. Kubota Date 31.10.'75

Inspected by M. Suzuki Date 31.10.'75





TML Tokyo Sokki Kenkyujo Co., Ltd.

8-ban 2-go, 6-chome, Minami-ohi, Shinagawa-ku, Tokyo
Cable Address: STRAINGAUGE TOKYO
Telex: 0246-8083 STRAIN

APPENDIX 6.2

ACCELEROMETER CALIBRATION

Calibration was based on the ambient Earth gravitational field. Accelerometers sensitive only to vertical components are used. Initially the accelerometer is placed orthogonal to the earth field and the output adjusted to zero by the balancing potentiometer.

The accelerometer is then placed in its working position, i.e. axially co-incident with gravitational direction, and the output measured. It is then known that this output corresponds to an acceleration of 9.81 m/sec^2. As a check the accelerometer is then turned upside down when a negative output equal in magnitude to the latter value should be obtained, corresponding to an acceleration of - 9.81 m/sec. The interval between these outputs is then scaled linearly. Typical readings are as follows:

Accelerometer No. 00816

Input acceleration 1g - Output potential 10.71 V

Accelerometer No. 00817

Input acceleration 1g - Output potential 12.76 V.

APPENDIX 6.3

CHARACTERISTICS OF THE ULTRASOUND TRANSDUCERS

VERNITRON LIMITED

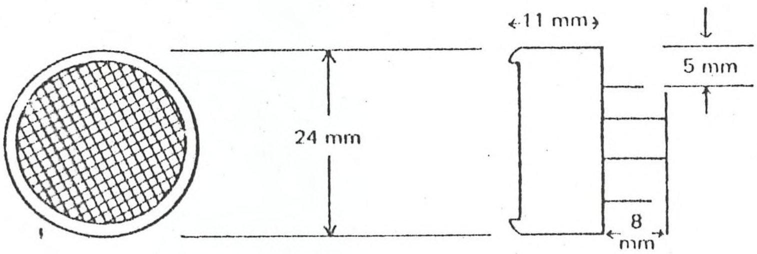
BULLETIN

93002/A

REVISED NOVEMBER 1972

AIRMASTER 40 kHz AIR TRANSDUCER

Part Number: 264001/40 kHz



NOMINAL DIMENSIONS

CONNECTIONS

Should be made to one short pin and one long pin. The long pins are common to each other, as are the short pins.

SPECIFICATION

	Transmitter:–	*Receiver:–*
Resonant frequency, F_R	40 kHz ± 1 kHz†	——
Anti-resonant frequency, F_A	——	40 kHz ± 1 kHz†
Capacitance	1300 pF	1300 pF
Electrical impedance at F_R	Approx. 1 kΩ	——
Receiving sensitivity (nominal)	——	–58 dB ref: 0 dB @ 1V/microbar (untuned 10 kohms load)
Transmitting sensitivity (nominal)	+16 dB ref: 1 microbar @ 1 foot with 1 volt drive	——
Beam angle	Approx. 60° at 3 dB down power points	——
Maximum drive	Continuous 10V RMS (250 mW) Pulsed 30V RMS	——

†Units are supplied in pairs with the transmitter (marked T_X) having an F_R of 40 kHz ± 1 kHz and the Receiver (marked R_X) an F_A at the same frequency.

These low cost, open grill ultrasonic air transducers are primarily intended for use in indoor conditions. They lend themselves to operation either singly as combined transmitter/receiver or in pairs with one unit transmitting and the other receiving, depending on the application.

APPENDIX 6.4

DESIGN CALCULATIONS OF THE ULTRASOUND DEVICE

TRANSMITTER CIRCUIT

Figure 6.4 shows the detailed circuit of the transmitter. The free running multivibrator sends at time T_A (figure A6.4-A) a starting pulse to the square wave generator, figure A6.4-B to the ramp generator, figure A6.4-F and the pulse width generator, of the receiver, figure A6.4-D. The amplitude of the pulse has to be equal to the amplitude of the burst of pulses generated by the square wave generator, figure A6.4-B.

The frequency of oscillation of the circuit is normally adjusted by means of the, potentiometer P_1 figure 6.4, for maximum sensitivity. That is, P_1 is adjusted for maximum meter deflection in the receiver when the input is so low that no amplifier stage is limiting. The free running multivibrator resets to zero the square waves sent to the transducer of the transmitter during the time interval T_{BC} figure A6.4-B. In this way the received pulses occur at the receiver transducer without overlapping the following burst of pulses.

Provided the velocity of the ultrasound in the air remains constant and equal to 331 m/sec, TBC has to satisfy the condition:

$$T_{BC} > \frac{dOTT}{331} \text{ [sec]}$$

where dOTT is the initial distance between the two transducers. TBC is adjusted by means of the preset P_2, figure 6.4. The time interval T_{AB} has to be chosen equal to:

$$T_Y \leqslant T_{AB} < T_{BC}$$

where T_γ is the duration of a single pulse.

In order to increase the sensitivity of the receiver transducer, T_{AB} is about 2 ÷ 3 times T_γ.

In T_B, figure A6.4-A, the free running multivibrator resets to zero the burst of pulses.

RECEIVER CIRCUIT

Figure 6.5 shows the circuit of the receiver. The integrated circuit used in the ultrasonic receiver to amplify the 40 KHz signal has a high gain since great sensitivity is required.

The output of the cascade of two operational amplifiers is connected to the pulse width generator (TTL positive NOR gates) whose output is related to the distance to be measured.

The pulse width generator output signal is sent to the monostable SN 74121, whose output is a pulse of about 20μs. The monostable output is then connected to the sample and hold circuit which gives at its output a dc voltage proportional to the distance.

Use of the monostable is necessary because the duration time of the pulse width generator output signal would be too long to be sent directly to the sample and hold circuit. The output pulses of the pulse width generator trigger the monostable. The free running multivibrator syncronises both the square wave generator and ramp generator (integrator) whose output is reset to zero at TA' figure A6.4-E by means of the monostable. The voltage VA' figure A6.4-G reached at TA' is proportional to the distance to be measured.

CALIBRATION CURVE

A plot of the measured output voltage against distance is shown in Figure A6.4-1. It can be seen that the characteristics of the device is linear. The initial distance (dOTT) between transmitter and receiver transducers has been set at 70 cm.

This distance, which is critical for the resolution of the measurement, has been chosen in order to minimise the error introduced in the measurement due to the movement of the transducer in the horizontal plane. The accuracy of this method is limited by the rise time of the burst of pulses and the dispersion of the pulse as it travels to the receiver. With a 10μs pulse, the resolution is in any case better than 2 mm.

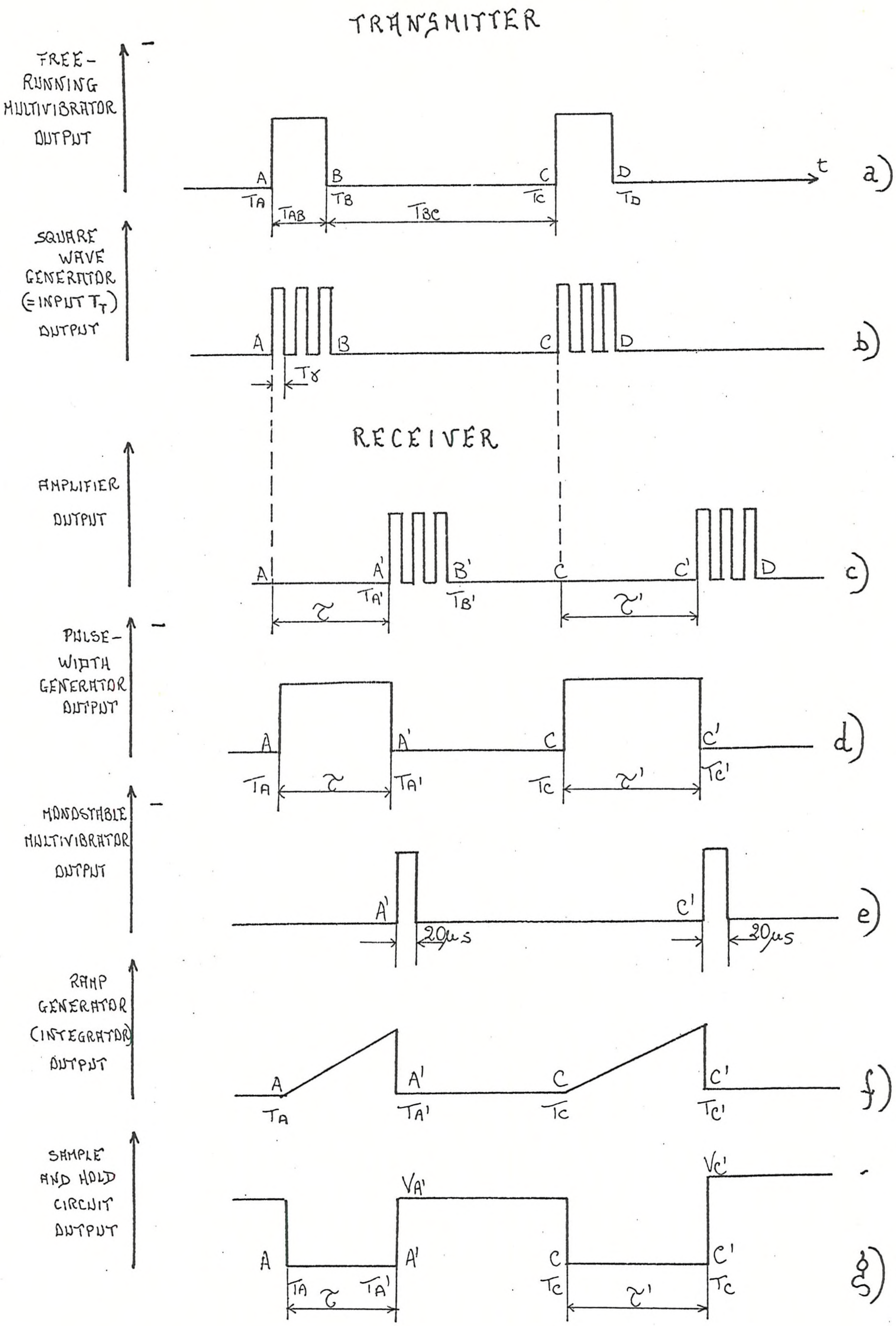


FIG. A6.4 OUTPUT WAVEFORMS OF THE ULTRASOUND DEVICE

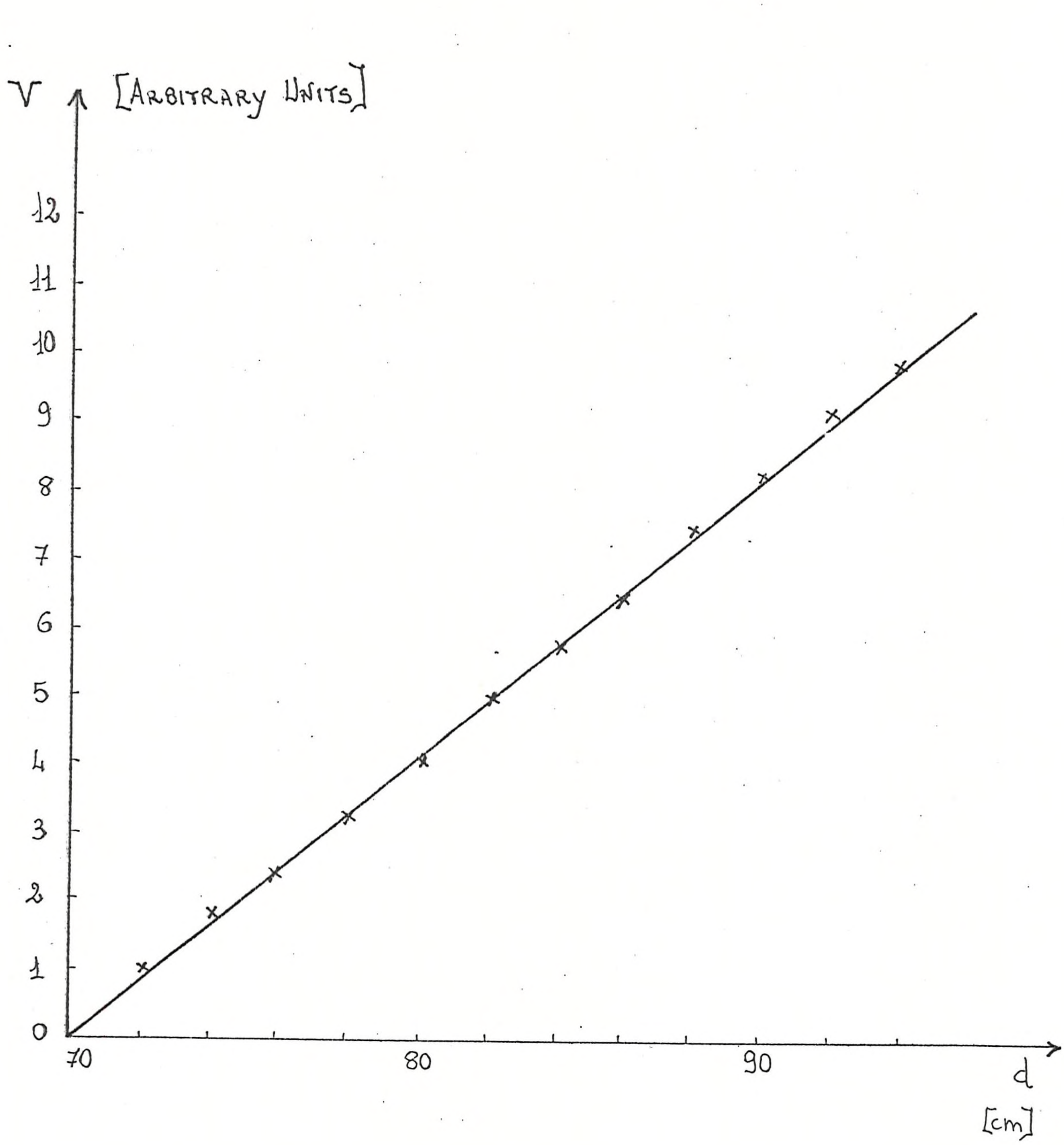


FIG. A6.4.1 ULTRASOUND DEVICE CALIBRATION PLOT

APPENDIX 6.5

AVERAGING FILTER

The averaging filter is a good low-pass filter for EMG signals because of its rapid dynamic response and since it is the mean or average value of the rectified EMG that is to be extracted. The schematic diagram of the state-variable averaging filter is shown in figure 6.13.

The frequency and step responses are shown in figure A6.5. The time response of the ideal averaging filter is given by:

$$Y_{(t)} = \frac{1}{T} \int_{t-T}^{t} X(t)\, dt$$

where T is the averaging period or "window". The corresponding transfer function is:

$$W_{(s)} = \frac{1 - e^{-st}}{ST}$$

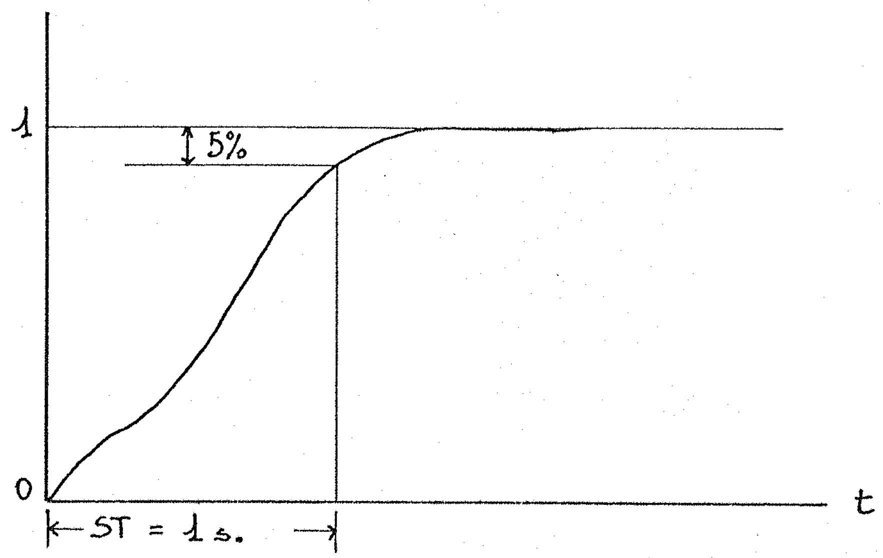
By using a third order Pade expansion for e^{-st}, a third order approximation to this transfer function can be obtained:

$$W_{(s)} \simeq \frac{2T^2s^2 + 120}{T^3s^3 + 12\, T^2s^2 + 60\, Ts + 120}$$

The filter used in the experiment was originally designed by GARLAND et al. (1972). It is a third-order state variable approximation of the averaging filter. Such an approximation uses four inexpensive integrated-circuit operational amplifiers and can easily be designed with an averaging period of any desired size, while the averaging filter typically requires components with values that are not readily available. The filter averaging period is given by:

$$T = 200\ C$$

where T is the averaging period in ms and C the capacitance in μF. This filter has a wide dynamic range, limited only by the power-supply voltage, and produces an output signal which is the non inverted average of the input signal.



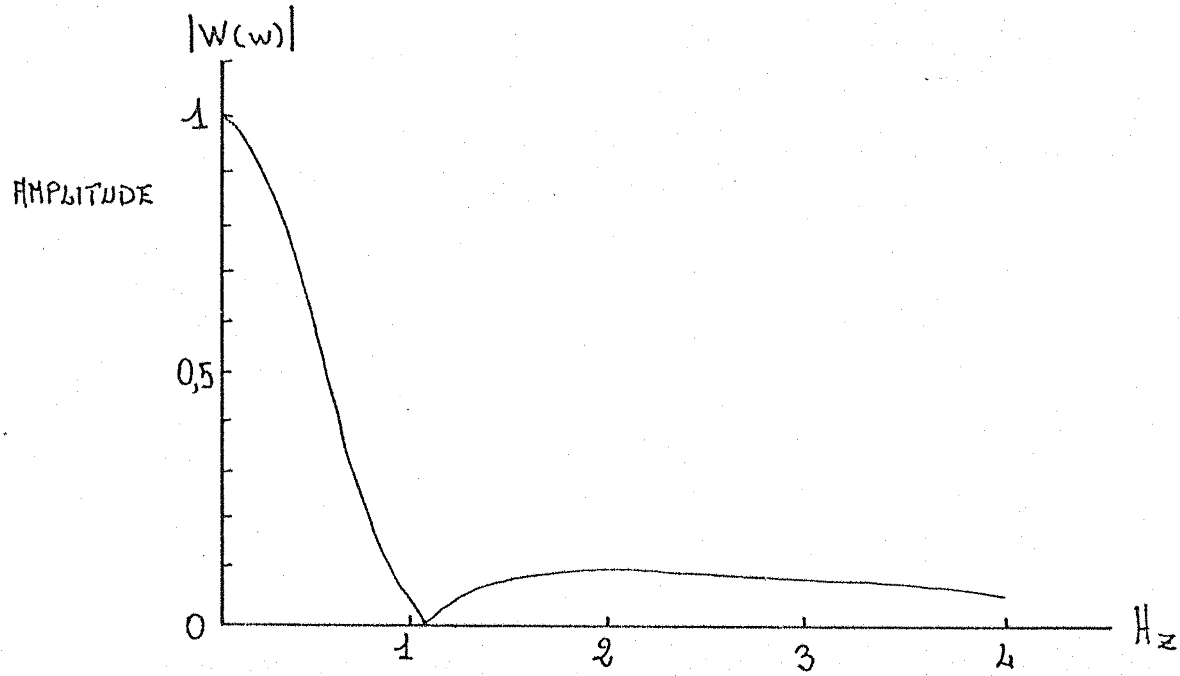


FIG. A6.5 AVERAGING FILTER STEP AND FREQUENCY RESPONSES

APPENDIX 6.6

EMG PROCESSOR'S FREQUENCY RESPONSES

Figure A6.6.1 reports the frequency response curve of the EMG integrator shown in Figure 6.11. While Figure A6.6.2 reports the frequency response curve of the EMG amplifier-filter shown in Figure 6.12 - 6.13.

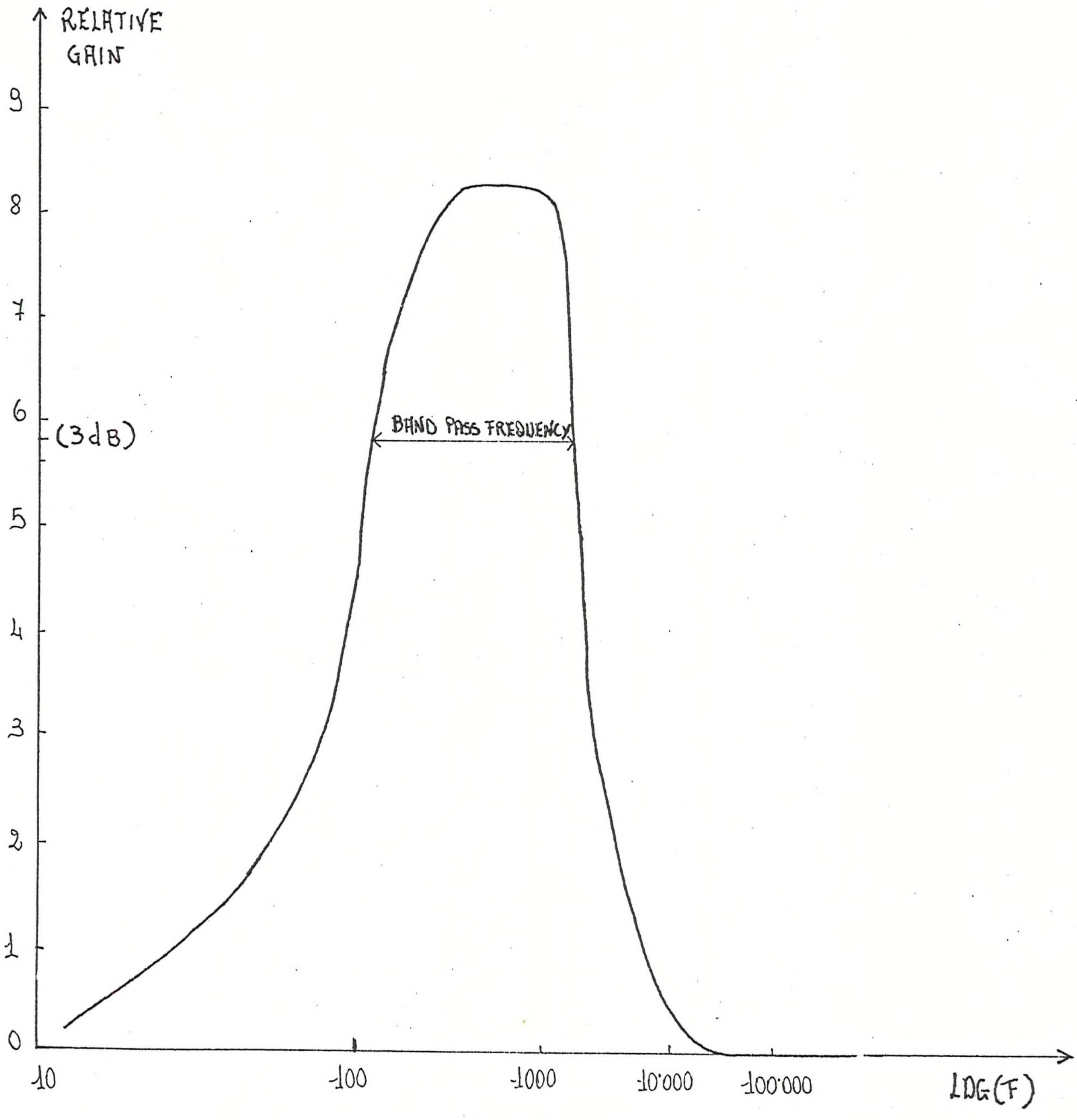


FIG. A6.6.1 EMG-INTEGRATOR FREQUENCY RESPONSE

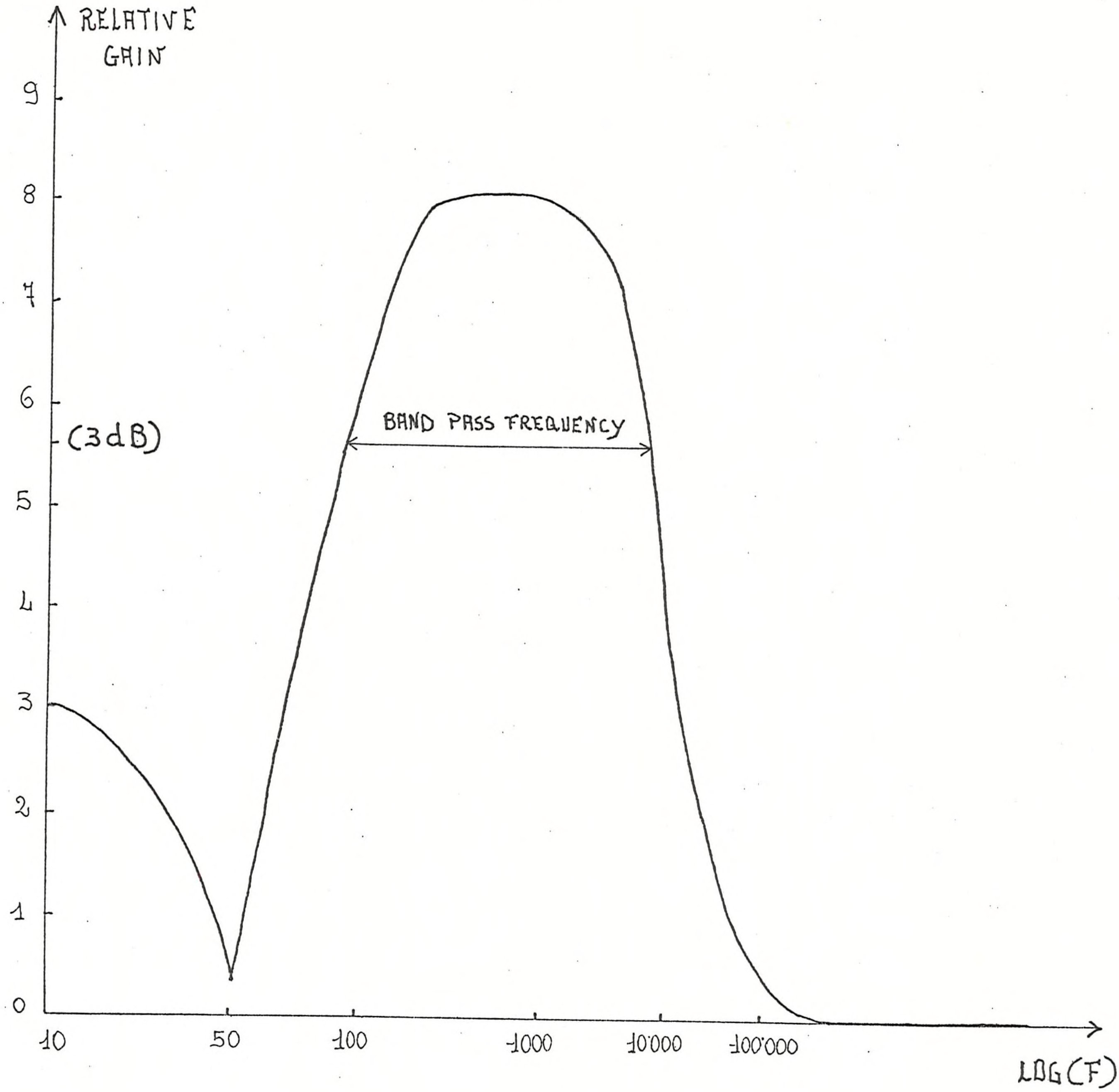


FIG. A6.6.2 EMG-FILTER FREQUENCY RESPONSE

CHAPTER SEVEN

HUMAN VIBRATION TRANSMISSION AT LOW FREQUENCIES

7.1. INTRODUCTION

Whole body vibration can be evaluated and assessed by many criteria. These consider such factors as physiological effects, subjective tolerance, mechanical body responses and task performance. Performance of manual tasks is generally degraded by vibration. The exact mechanisms of this performance loss are complex but one can identify factors such as direct interference with muscular action, indirect interference through the central nervous system, blurring of vision, interference in audio communication channels and actual occurrence of fatigue and disorientation. In addition, of course, the actual implements and displays involved in the manual tasks may themselves be adversely affected by the vibrations. Thus, for example aircraft displays may become illegible, factory workers may suffer loss of efficiency or even physical injury. Thus body vibration is an area of vital interest to physiologists, psychologists, clinicians and engineers. Most vibration studies involve fairly small amplitude but wide frequency band signals or specific periodic or pseudo-periodic random signals at frequencies near known body resonances. Whereas in some fields, e.g. maritime studies, the effect of slow, large amplitude vibrations have been investigated in relation to human task efficiency there appears to have been little attempt to quantify the results in terms of specific attributes of the motor control system. It is certainly recognised that stimulation of the balance organs and other receptors involved in the control of posture leads to increasingly severe demands of concentration in order to maintain a given posture while attempting to carry out the task at hand. Thus a part of the task which normally is achieved purely involuntarily, now becomes a part, often dominant, which requires severe conscious effort thereby disrupting the normal operating mode of the CNS.

This latter is perhaps an extreme example but there certainly is a threshold beyond which the balance organ signals are routed to the conscious brain (motor cortex) and demand conscious intervention. Below this threshold response to the vestibular and other exteroceptors appears to be autonomous.

The work undertaken in this thesis has been motivated by a somewhat different approach to the same problem. When we undertake manual tasks we are able to compensate, largely subconsciously, for gross body movements and it is the role of the vestibular sensors in this adaptation which we seek to model. It is hoped that such an understanding might lead to a significant improvement in upper-limb prosthesis, where at present no such compensation is available, often with quite disastrous results. In the next chapter a proposal for some contrived tracking problems, in the presence of extraneous body vibration, will be described. A necessary point in attempting to devise and analyse the results of these experiments is to model the vertical movements of the human body. Of course the physical body is a complex distribution of mass, elastic and viscous material but, for necessary simplicity, we consider the masses to be distributed at head and shoulder levels with visco-elastic supports. The experiments reported in this chapter were primarily intended to provide measures of the transmissibility of vibrations, applied to the seat, to the torso and head in order to partially quantify the vestibular feedback system. However, since the proposed frequency range spans some of the human body resonances the results may also be of more general interest. Generally the experiments involve seated subjects, approximately in the car-driver position with vibrations applied vertically to the seat.

7.2. TRANSMISSIBILITY TO THE SHOULDER

7.2.1. EXPERIMENTAL DATA

Ten male adults were tested. Their personal data are given in Table 7.1. They can be seen to be of average build, aged between 16 and 33 years, reasonably fit and able to withstand a vibration experiment. In all tests the subject rested his hands on his thighs and was instructed to look straight ahead. No mental or physical task had to be performed. A sketch showing the posture of the subject on the platform is shown in Figure 7.1, which also indicated the positions (1) and (2) of the two accelerometers. These were required to measure respectively:

(1) vertical seat acceleration

(2) vertical shoulder acceleration

N.B. The latter accelerometer was attached to wedge shaped block designed to cancel the natural slope of the shoulders. This transducer was strapped tightly to the top of the right shoulder just medial to the tip of the acromion*. It was fixed so that it did not lift the shoulder during vibration tests.

The accelerometers were calibrated before each run. The transducer outputs were recorded on either a 6-channel/U.V. Recorder (SE LAB Type 300 6/DL) or an FM Tape Recorder (ES LAB T-3000). Each subject was instructed to maintain a normal relaxed erect position. No seat harness was used. Tests were not commenced until the subject was seated comfortably.

---

* Acromion: point or summit of the shoulder; the triangular process at the extreme outer end of the spine of the scapula.

7.2.2. SINUSOIDAL EXCITATION

Five subjects were subjected to sinusoidal accelerations. The frequencies were changed by discrete increments in the range 0-9 Hz. When the transducer output looked reasonably steady a recording lasting 30 seconds was made from the accelerometers.

The duration of the tests did not impose unnatural constraints on the subject. During tests emphasis was placed on frequencies near resonance where smaller frequency intervals were chosen. The principal results for these experiments have been combined and illustrated in Figures 7.2 and 7.3. The ratio of acceleration amplitudes are recorded for each frequency.

The results, Table 7.2, show a broad resonant peak, nearly twice the value of the forcing vibration around the frequency 6 Hz.

7.2.3. SWEPT FREQUENCY EXCITATION

Many difficulties were encountered in performing the sinusoidal excitation tests. These mainly arose from the prolonged period of observation for each amplitude and frequency. Not only was this fatiguing for the subjects but it was very difficult to maintain a constant posture and there was a slight drift also in the d.c. platform position.

It was therefore decided to try to assess the response using a quicker transient method. One possibility is to use step, impulse or other discontinuity functions. However, apart from the obvious discomfort caused to the subject there are other disadvantages. Firstly these functions generally give a very uneven distribution of energy across the frequency spectrum and secondly the imposed shocks are likely to excite neurological reflexes which would give unwanted components in the response.

In order to preserve an inert subject it was decided to use a gradually varying continuous function. The most easily generated is the swept sine wave. This has the advantage that a steady condition could be achieved before the transient frequency sweep is introduced, the instant of introduction being unknown to the subject. Thus the excitation takes the form

$$X_1(t) = A \cos \Omega t \qquad t \leqslant 0$$

$$= A \cos \Omega t \left(1 + \frac{\alpha}{2} t\right) \qquad 0 < t \leqslant T$$

where the transient is introduced at t = 0.

In the tests performed, the initial frequency $\frac{\Omega}{2\Pi}$ was 1Hz, the duration T was 25 seconds and the maximum frequency $\frac{\Omega}{2\Pi}\left[1 + \frac{\alpha}{2} T\right] = 7$ Hz. Figure 7.4a.

It can be shown, Appendix 7.1, that the energy spectrum takes the form:

$$|X(j\omega)| = A\sqrt{\frac{1 + \sin\left(\frac{\Omega^2+\omega^2}{\alpha\Omega}\right)}{\alpha\Omega}}$$

The form of this distribution is shown in Figure 7.4b. It will be noticed that the spectrum is broader than the range of swept frequencies, but obviously the amplitude is reduced due to the spread of energy.

The spectrum shown is continuous, assuming T is very large. Since T is finite, the spectrum actually is discrete occurring at frequencies:

$$\frac{2\Pi}{T}, \frac{4\Pi}{T}, \frac{6\Pi}{T} \ldots \text{ect.}$$

In this case the excitation is fairly "flat" in the frequency range of interest, but in determining the frequency response of the shoulder care should be taken to perform the calculations at the above frequencies. The responses were actually computed by a standard FFT Techniques (See Appendix 7.2), the effect of interpolation between the spectral lines accounts for the spiky nature of the frequency spectra.

The tests were repeated decreasing the frequency at the same rate. Data was continuosly recorded on an SE LAB Type T 3000 magnetic FM tape recorder.

FFT and other computations were performed, off-line, on a Digital Equipment Corporate PDP 11/50 Digital Computer (See Appendix 7.3).

If the output acceleration spectrum is given by $\ddot{X}_2(j\omega)$ then, by division, the frequency response of interest is given by:

$$F_1(j\omega) = \frac{\ddot{X}_2(j\omega)}{\ddot{X}_1(j\omega)}$$

Figures 7.5 - 7.7 show the resulting frequency response curves for three of the subjects. The continuous curves are related to the sweep with increasing frequency while the dotted curves to the sweep with decreasing frequency.

The characteristics show general agreement with those drawn for steady sinusoidal excitation, although the resonant frequency has an average value, 5 Hz, which is somewhat lower. It is felt that the more controlled conditions of the swept frequency experiments make these results more reliable. The results of the test are tabulated in Table 7.3. There is no pronounced difference when the direction of frequency sweep is reversed.

7.3. TRANSMISSIBILITY TO THE HEAD

7.3.1. EXPERIMENTAL DATA

Five male subjects were used and their personal data are given in Table 7.4. Two accelerometers were mounted, see Figure 7.8. The seat accelerometer is mounted directly to the structure while the head accelerometer is strapped firmly on the head and aligned by an aluminium wedge. The subjects were seated and given trial runs so that a comfortable resting posture could be adopted.

During the actual tests the subject rested his hands on his thighs and kept his head in an upright position. No physical or mental tasks were performed by the subjects during the actual experimental runs.

7.3.2. SWEPT FREQUENCY EXCITATION

The applied platform acceleration was a swept sinusoid with the same parameters as those in the previous tests. In some experiments a restraining neck collar made from polystyrene was fitted to eliminate relative head/shoulder displacement and to keep the head in the vertical plane. Figure 7.9 and 7.10 show typical results with the neck collar, while Figures 7.11 and 7.12 show the results for the same subjects without neck collar. The continuous curves are related to the sweep with increasing frequency while the dotted curves to the sweep with decreasing frequency. The data was recorded and analysed using the techniques outlined in the last section. The results are summarised in Table 7.5.

When the subject is wearing the neck support there is a resonant peak between 4 and 5 Hz. Very similar results were obtained when the collar was not worn. This suggests that neck flexibility is only of secondary importance. Generally, the results are stationary under statistical averaging and bear a good relationship to the results of other workers [GARG et al. 1976; WOODS, 1967; MACDUFF, 1971; GUIGNARD, 1972; GIERKE, 1968; DIECKMANN, 1958;].

Comments on the frequency spectra and FFT calculations made in the previous section also apply here.

TABLE 7.1 PERSONAL DATA OF THE SUBJECTS TESTED

Subject	Age ys	Weight Kg	Height cm	Shoulder Height cm
A.T.	29	64	170	65
S.L.	33	58	167	63
T.D.	17	78	192	88
C.N.	15	49	171	80
B.F.	23	81	180	93
A.B.	20	58	170	68
I.S.	22	72	172	87
M.T.	23	69	182	90
M.A.	31	65	175	85
M.D.	16	60	170	85
Average	22.90	65.40	174.9	50.40
St.Dev.	6.31	9.82	7.65	11

TABLE 7.2 FREQUENCY RESPONSE RESULTS BY SUBJECT

SINUSOIDAL TECHNIQUE

	100% Displacement MAX.Peak		70% Displacement Max.Peak	
Subject	Frequency	Magnitude	Frequency	Magnitude
B.F.	6	2.6	6	2.25
S.L.	5.9	2	6	1.85
T.D.	7	1.75	8	1.65
C.N.	5.5	1.8	5	1.65
A.T.	5	1.9	5.5	1.75
Average	5.88	2.01	6.1	1.83
St.Dev.	0.74	0.34	1.14	0.25

TABLE 7.3 FREQUENCY RESPONSE RESULTS BY SUBJECT

SWEPT TECHNIQUE

Subject	Swept 1 → 7 Hz Max. Peak Frequency	Magnitude	Phase	Swept 7 → 1 Hz Max. Peak Frequency	Magnitude	Phase
A.B.	4.5	1.65	60	4.6	1.7	48
I.S.	5.5	2	64	5.8	1.8	72
M.T.	4.9	1.85	62	4.7	1.7	56
M.D.	4.8	1.95	68	5	1.9	62
M.A.	5	1.7	58	4.8	1.8	66
Average	4.94	1.83	62.4°	4.98	1.78	60.8°
St.Dev.	0.36	0.15	3.85	0.48	0.08	9.23

TABLE 7.4 PERSONAL DATA OF THE SUBJECTS TESTED

Subject	Age ys	Weight kg	Height cm
A.B.	20	58	170
I.S.	22	72	172
T.D.	17	78	192
M.D.	16	60	170
A.T.	29	64	170
Average	20.80	66.40	174.80
St.Dev.	5.17	8.41	9.65

TABLE 7.5 FREQUENCY RESPONSE RESULTS BY SUBJECT

SUBJECT WEARING A RESTRAINING SUPPORT FOR THE NECK

	Swept 1 → 7 Hz Maximum Peak			Swept 7 → 1 Hz Maximum Peak		
Subject	Frequency	Magnitude	Phase	Frequency	Magnitude	Phase
A.B.	4.9	2.05	57.	4.6	2.0	56
I.S.	4.	1.35	35.	3.6	1.2	28
T.D.	4.6	1.70	60.	4.4	1.65	62
M.D.	4.4	1.55	48.	4.	1.50	55
A.T.	5	1.62	36.	4.7	1.65	34
Average	4.58	1.65	47.2	4.26	1.60	47
St.Dev.	0.40	0.26	11.56	0.46	0.29	15

TABLE 7.6 FREQUENCY RESPONSE RESULTS BY SUBJECT

SUBJECT NOT WEARING THE RESTRAINING NECK SUPPORT

	Swept 1 → 7 Hz Maximum Peak			Swept 7 → 1 Hz Maximum Peak		
Subject	Frequency	Magnitude	Phase	Frequency	Magnitude	Phase
A.B.	4.6	1.80	46	4.9	2.10	60
I.S.	4.	1.30	44	4.	1.25	42
T.D.	4.8	1.40	50	4.7	1.55	52
M.D.	3.9	1.35	39	4.2	1.40	45
A.T.	4.5	1.55	46	4.8	1.45	50
Average	4.36	1.48	45	4.52	1.55	49.80
St.Dev.	0.39	0.20	4	0.40	0.33	6.94

CHAPTER 7

LIST OF FIGURES

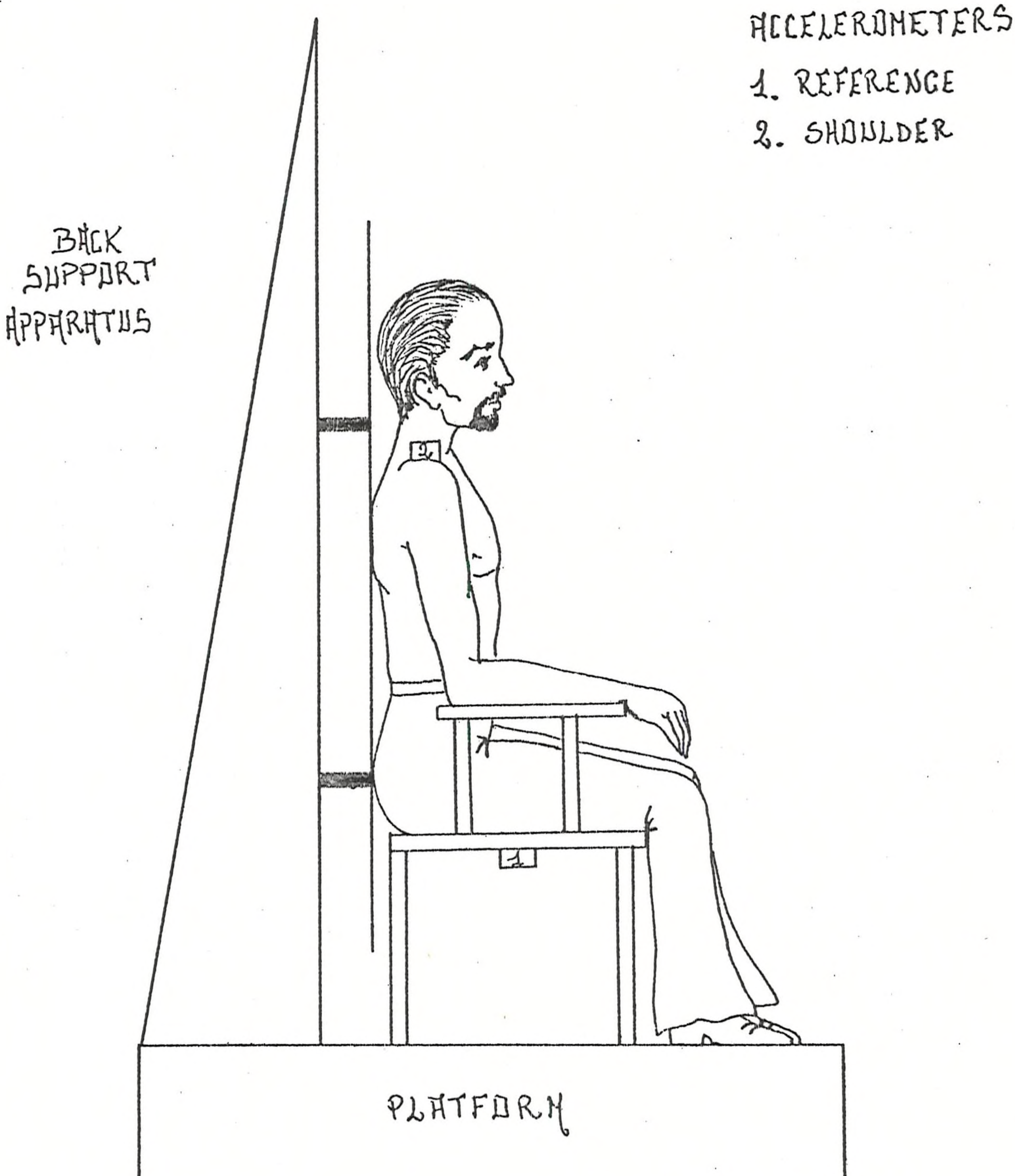


FIG. 7.1 SKETCH OF THE SUBJECT UNDER TEST SHOWING THE LOCATIONS OF THE ACCELEROMETERS

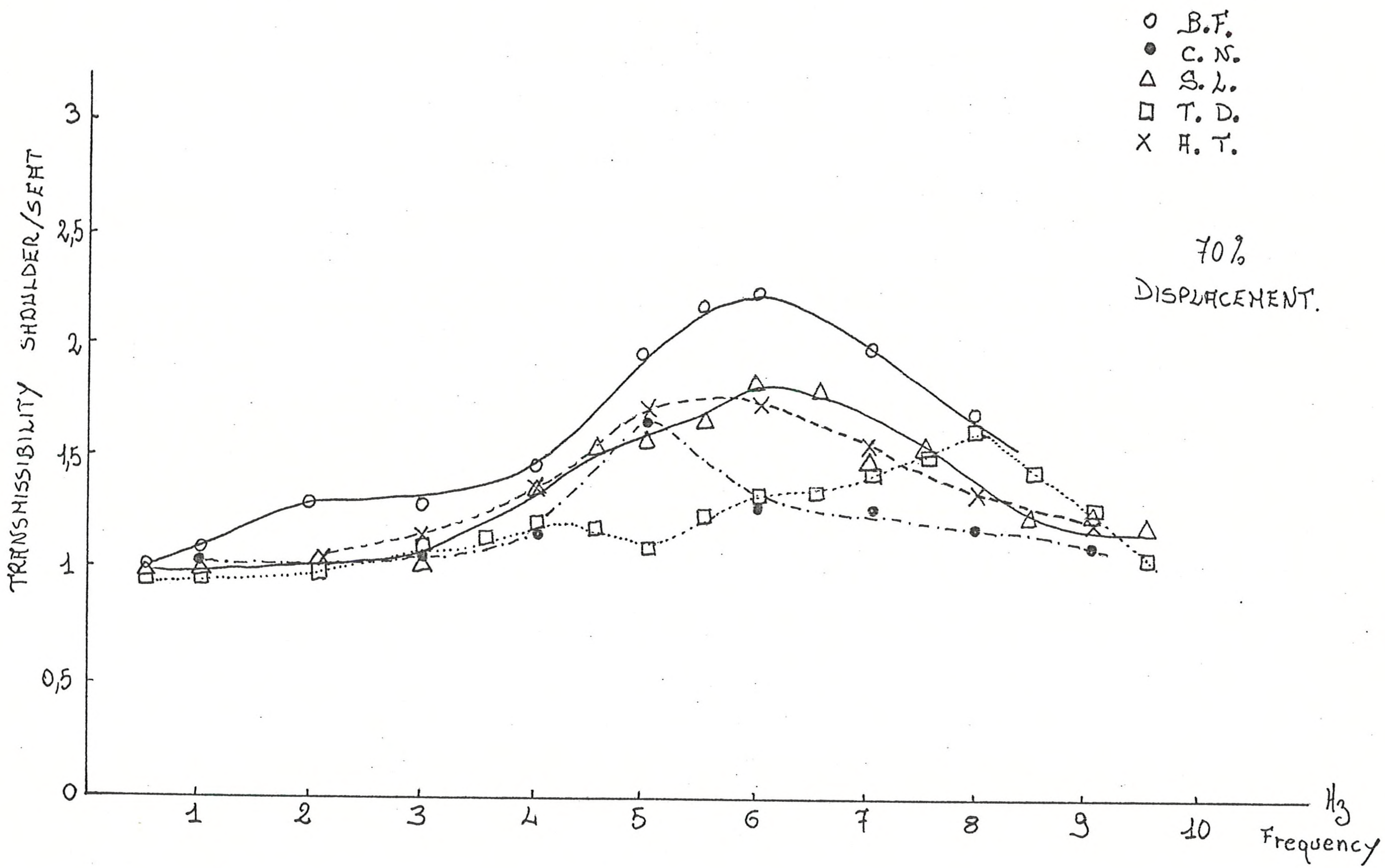


FIG.7.2 FREQUENCY RESPONSE SINUSOIDAL EXCITATION TESTS

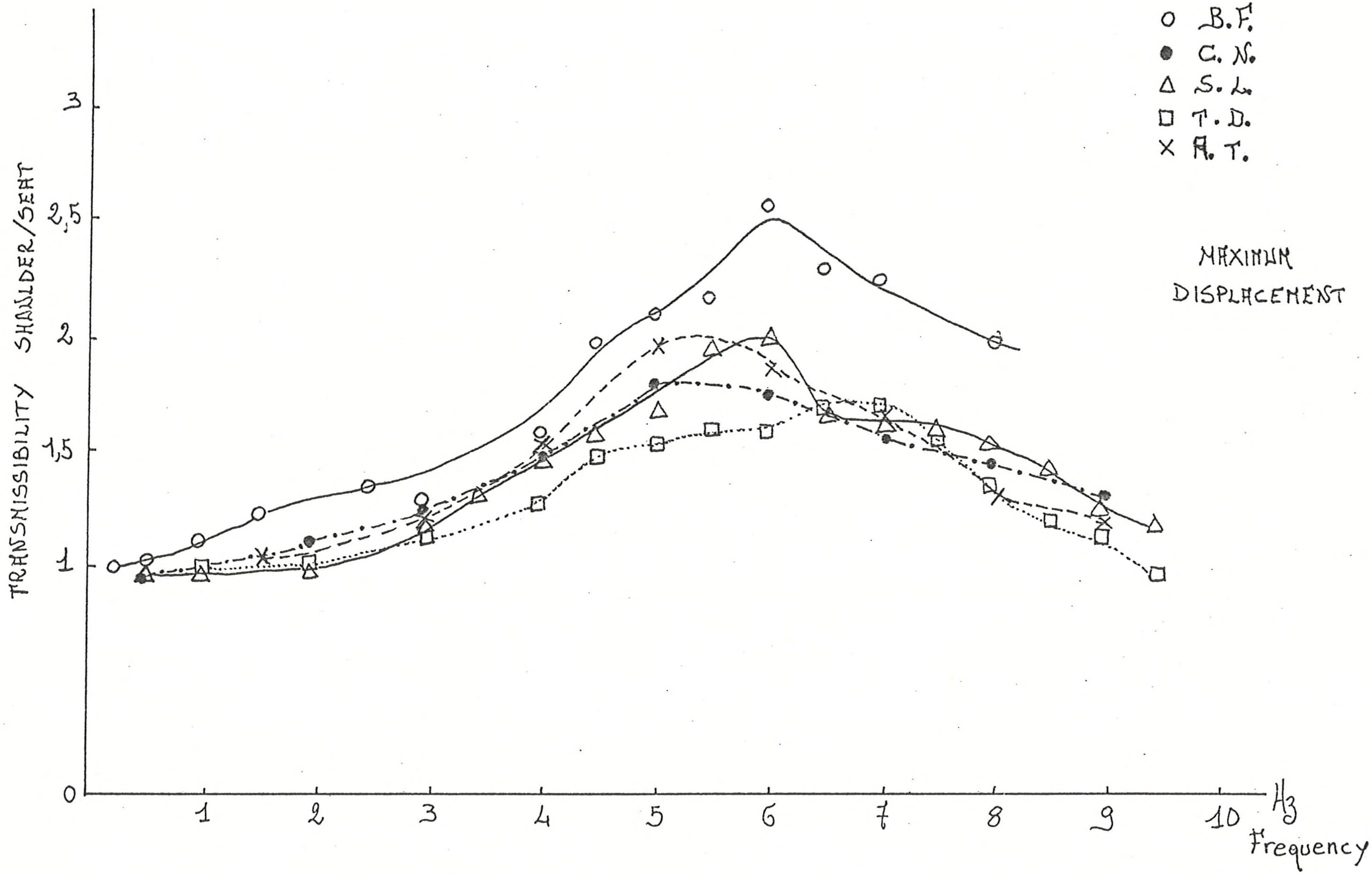


FIG.7.3 FREQUENCY RESPONSE SINUSOIDAL EXCITATION TESTS

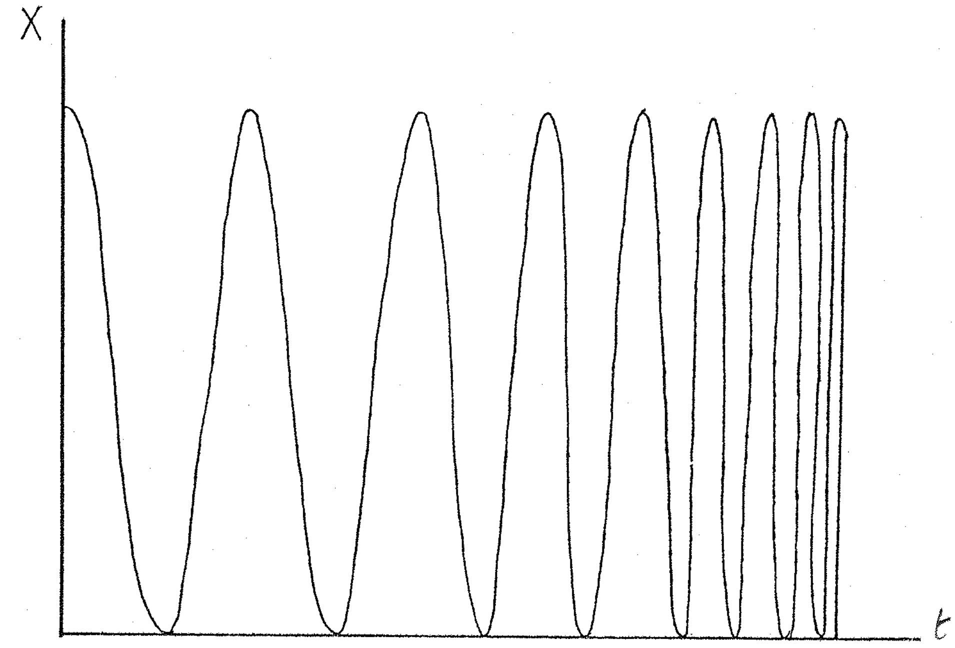


FIG. 7.4 a) SWEPT SINE WAVE

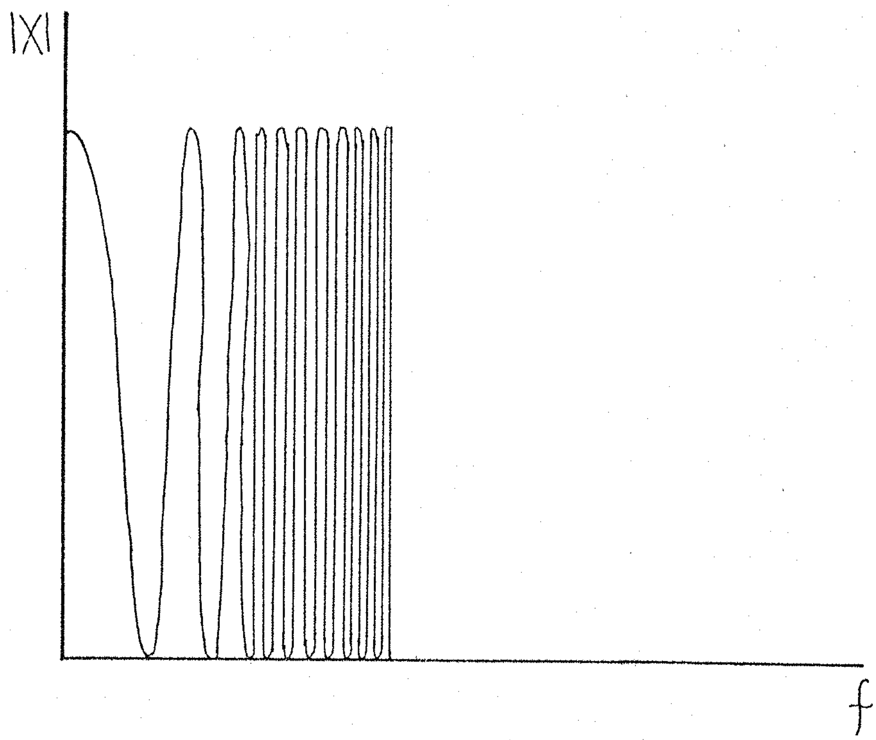


FIG. 7.4 b) SHAPE OF THE ENERGY SPECTRUM

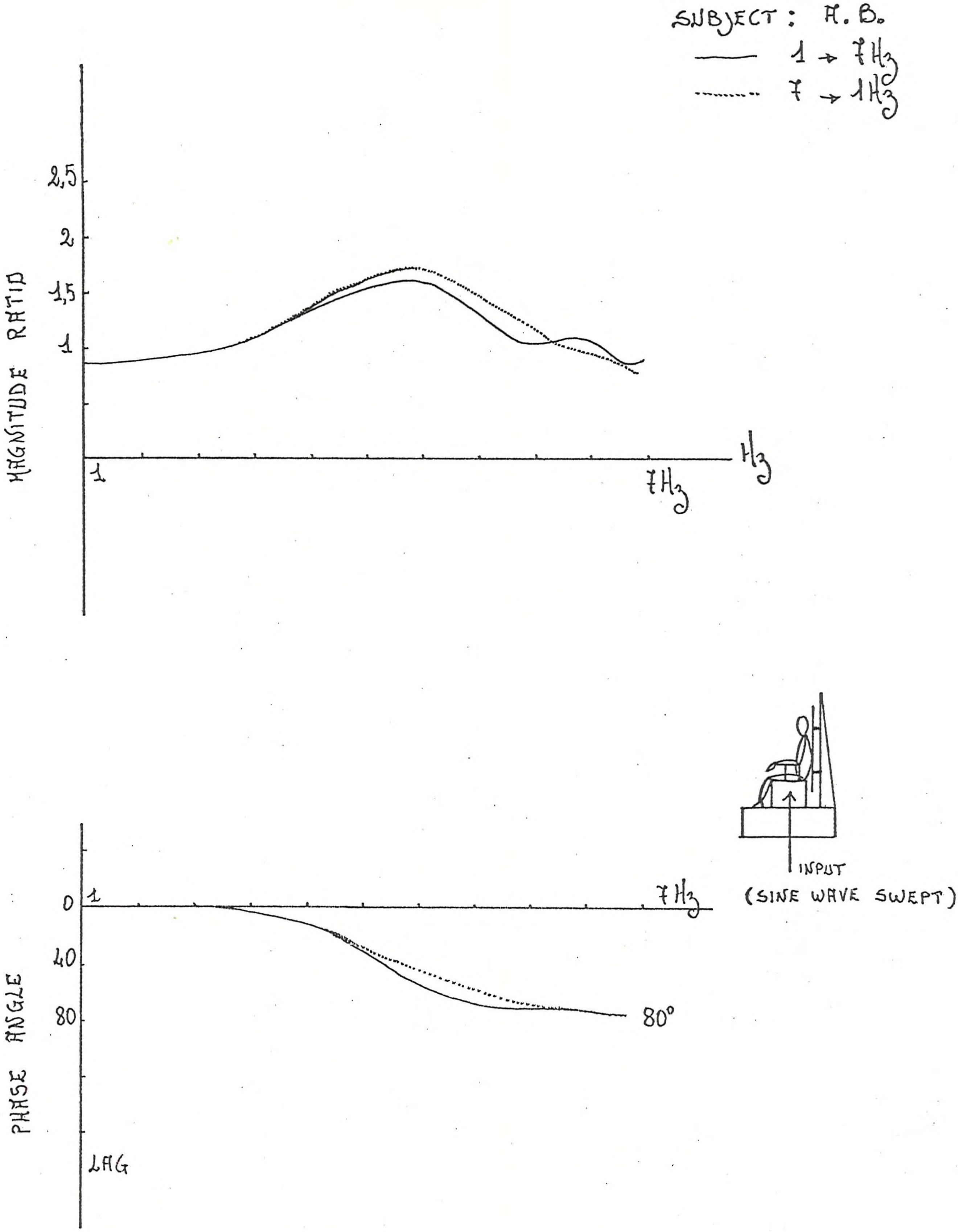


FIG. 7.5 FREQUENCY RESPONSE SWEPT EXCITATION TEST

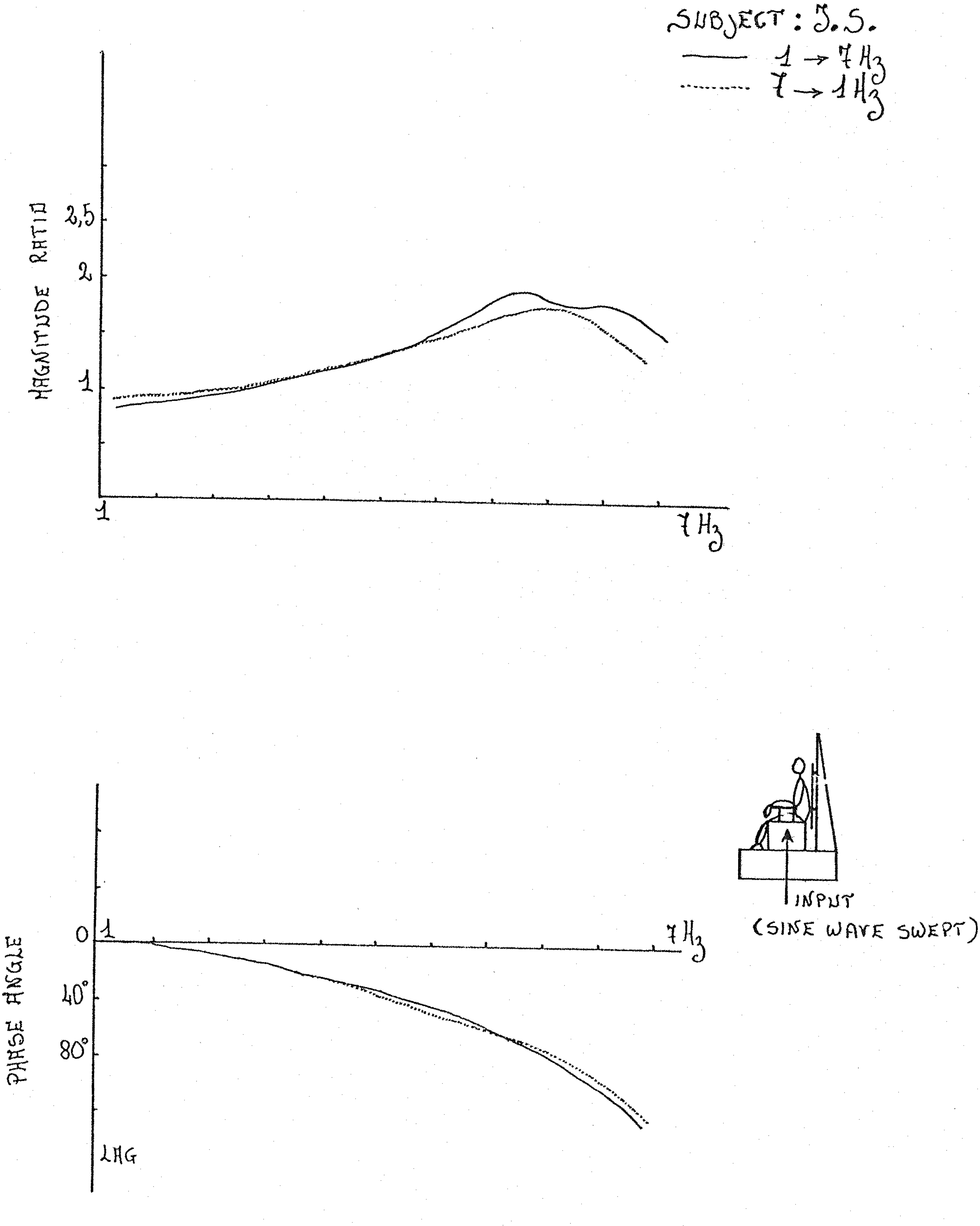


FIG. 7.6 FREQUENCY RESPONSE SWEPT EXCITATION TEST

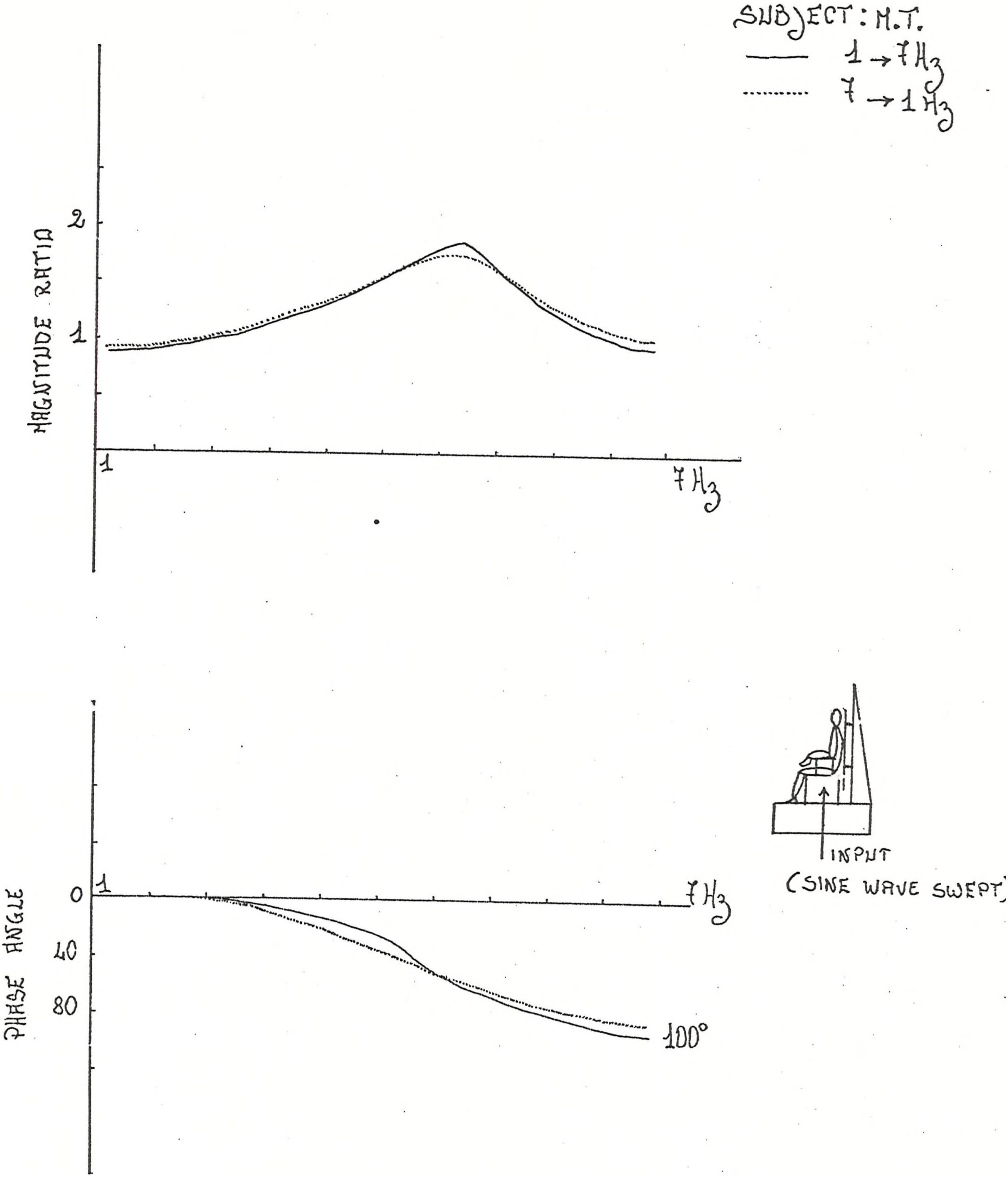


FIG. 7.7 FREQUENCY RESPONSE SWEPT EXCITATION TEST

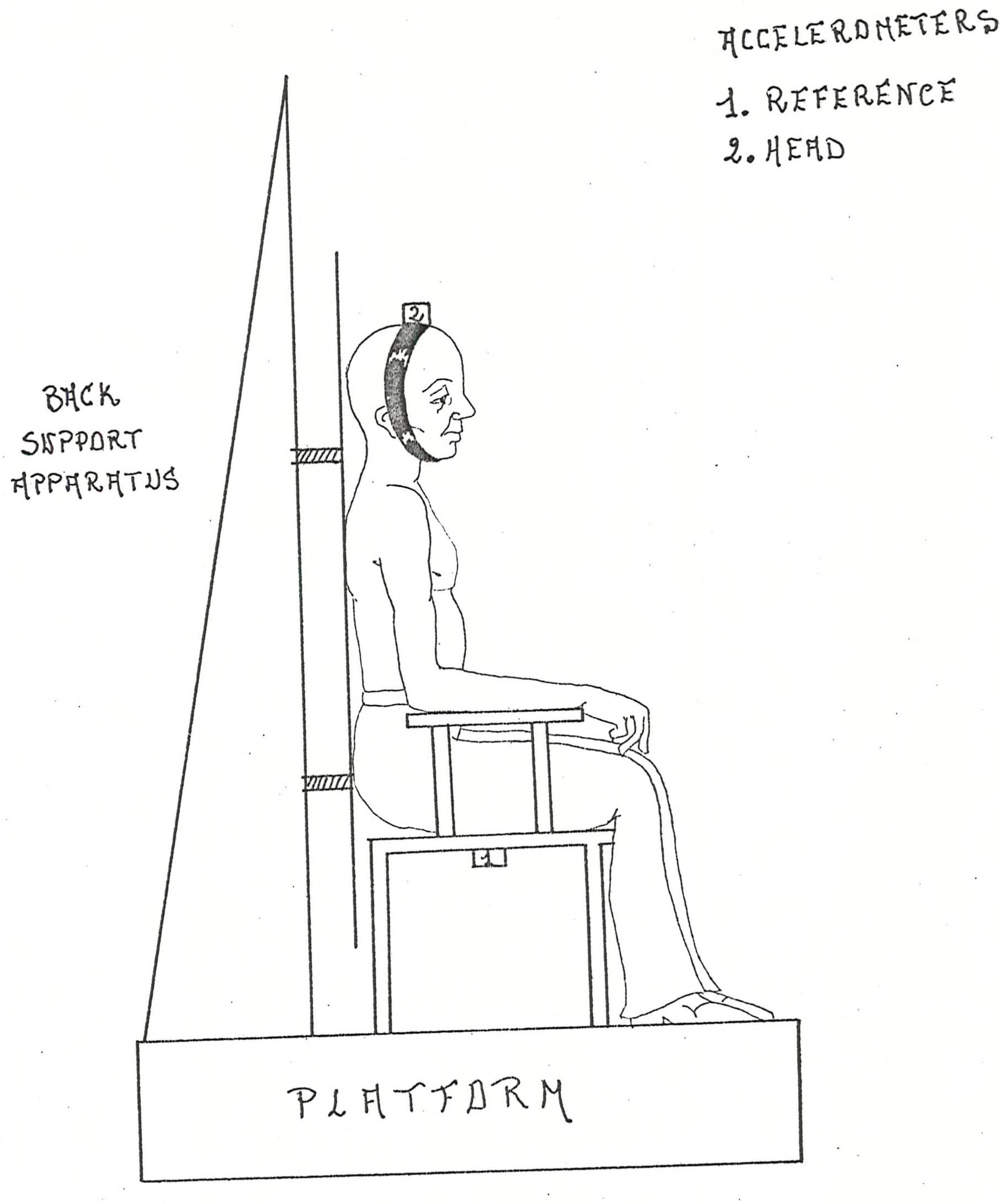


FIG. 7.8 SKETCH OF THE SUBJECT UNDER TEST SHOWING THE LOCATIONS OF THE ACCELEROMETERS

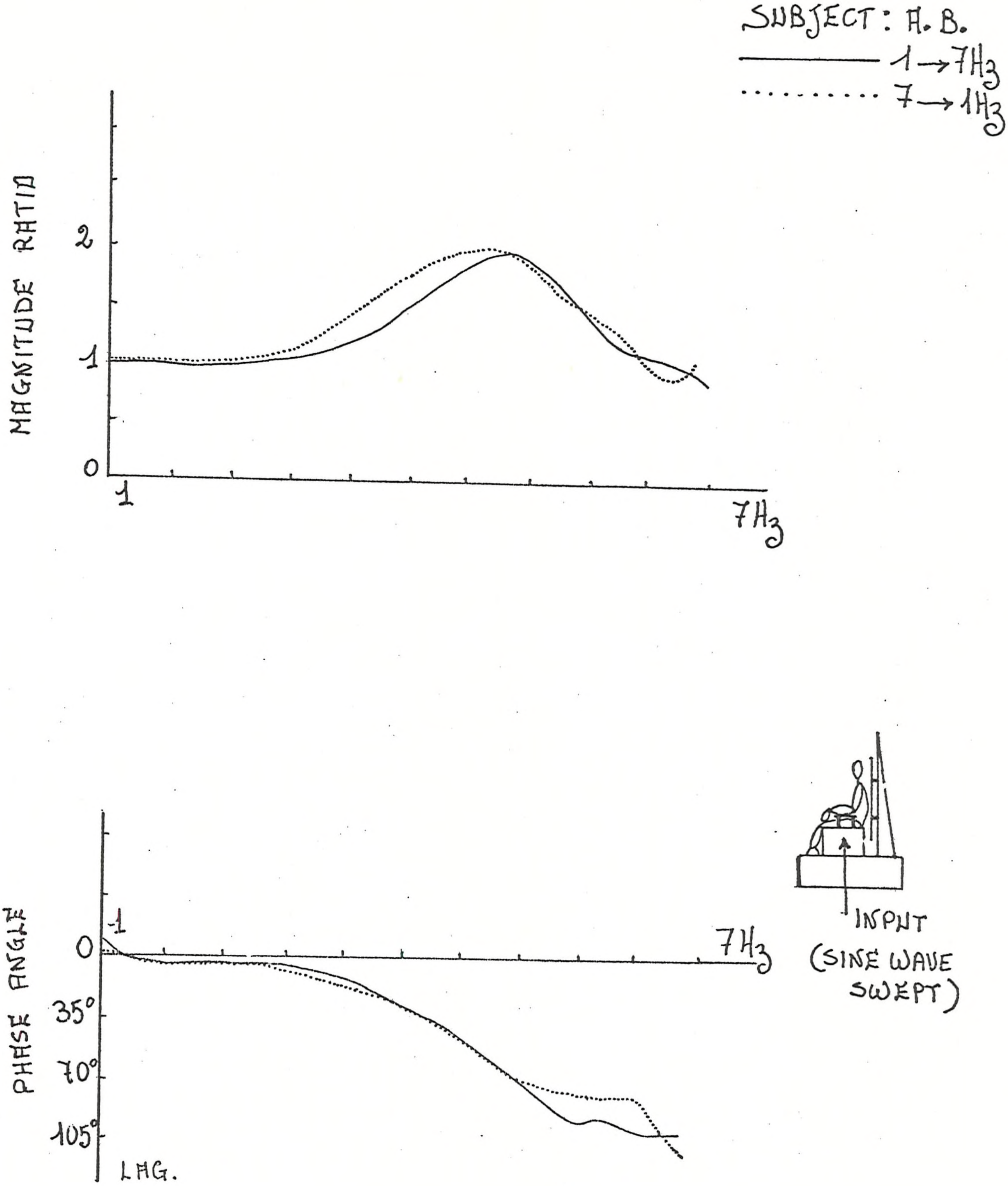


FIG. 7.9 FREQUENCY RESPONSE SWEPT EXCITATION TEST
(Subject wearing the restraining neck support)

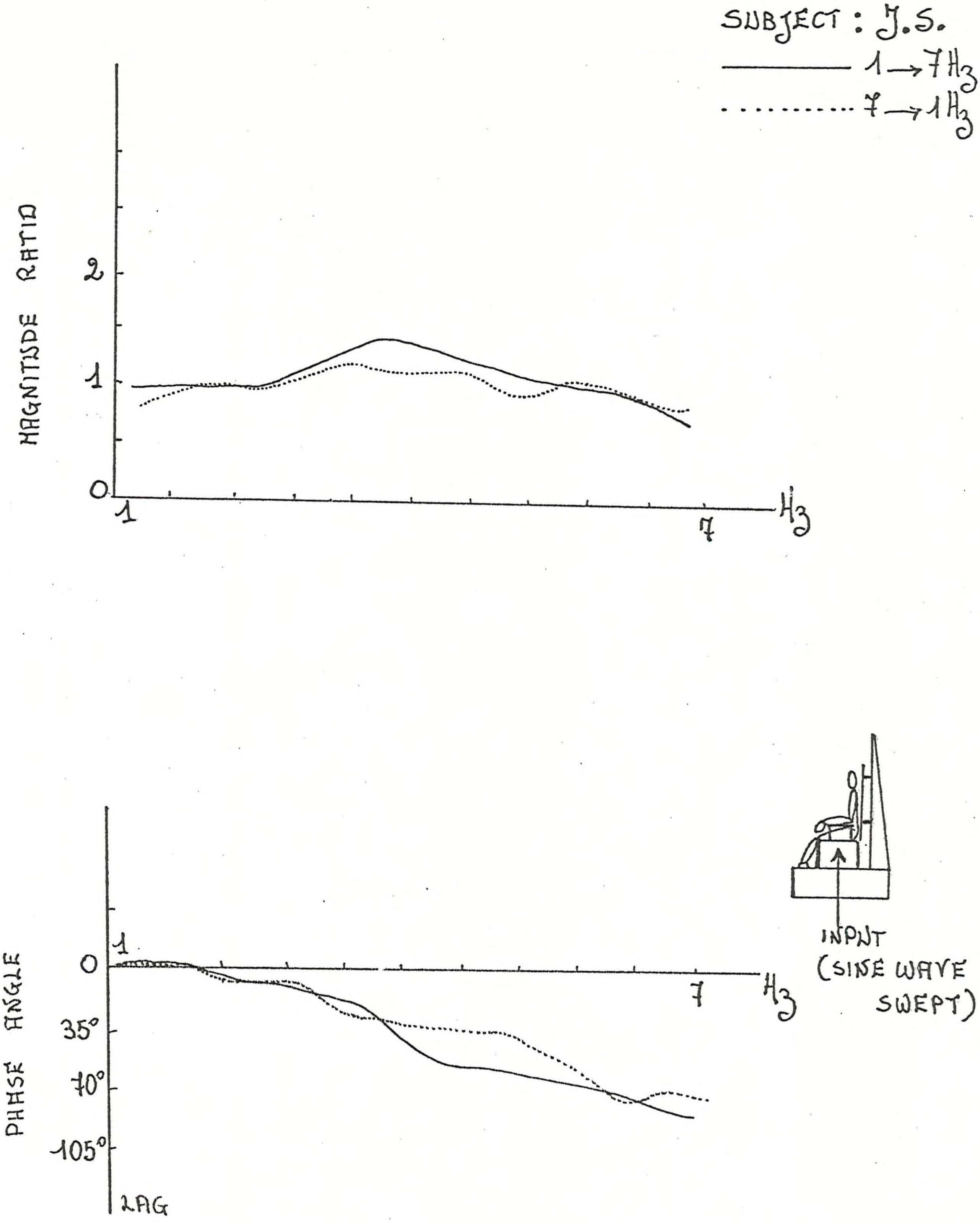
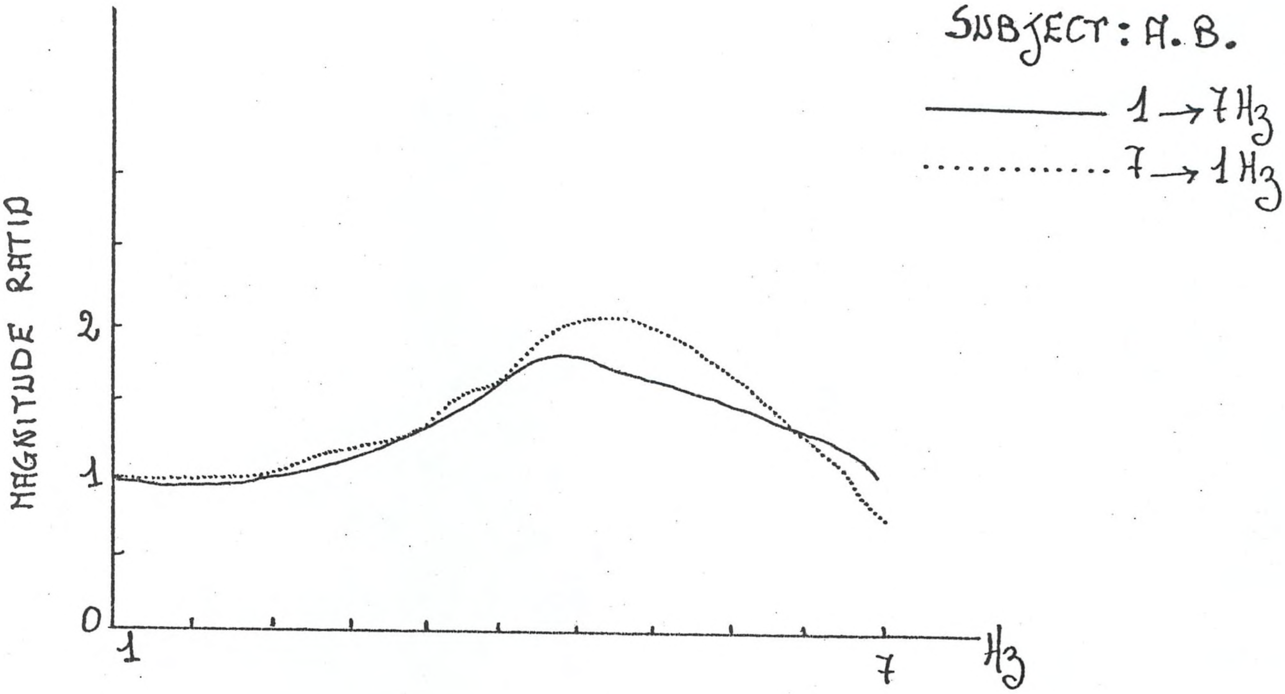


FIG. 7.10 FREQUENCY RESPONSE SWEPT EXCITATION TEST
(Subject wearing the restraining neck support)



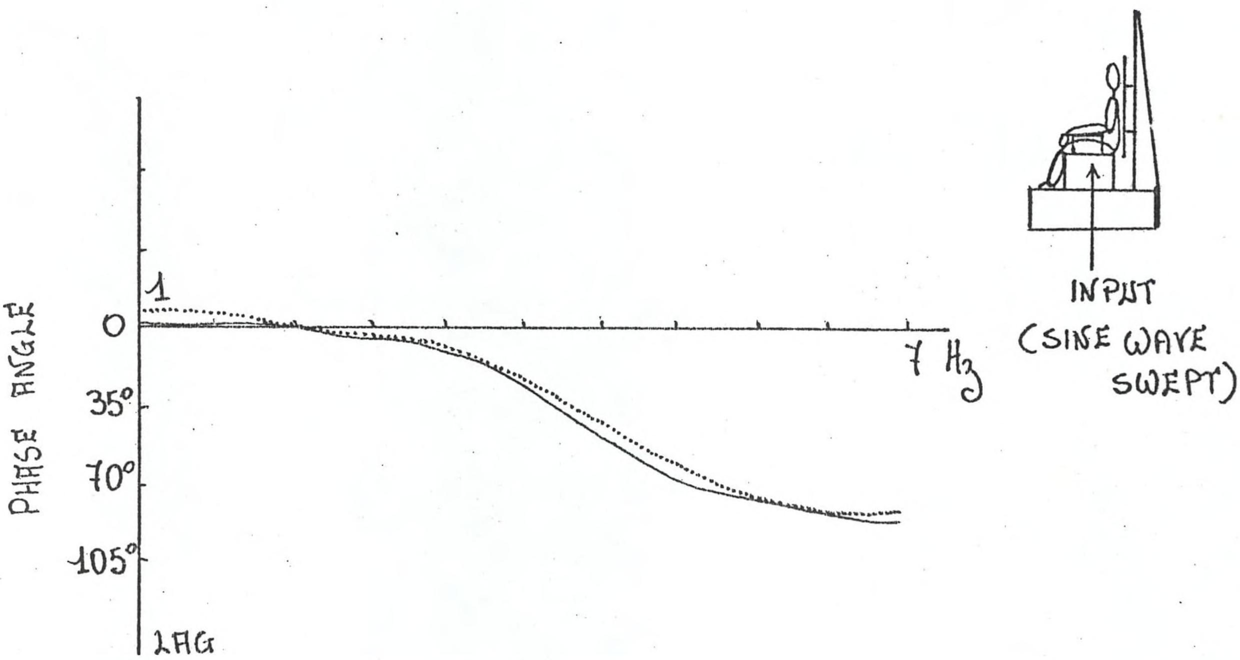


FIG. 7.11 FREQUENCY RESPONSE SWEPT EXCITATION TEST
(Subject not wearing the restraining neck support

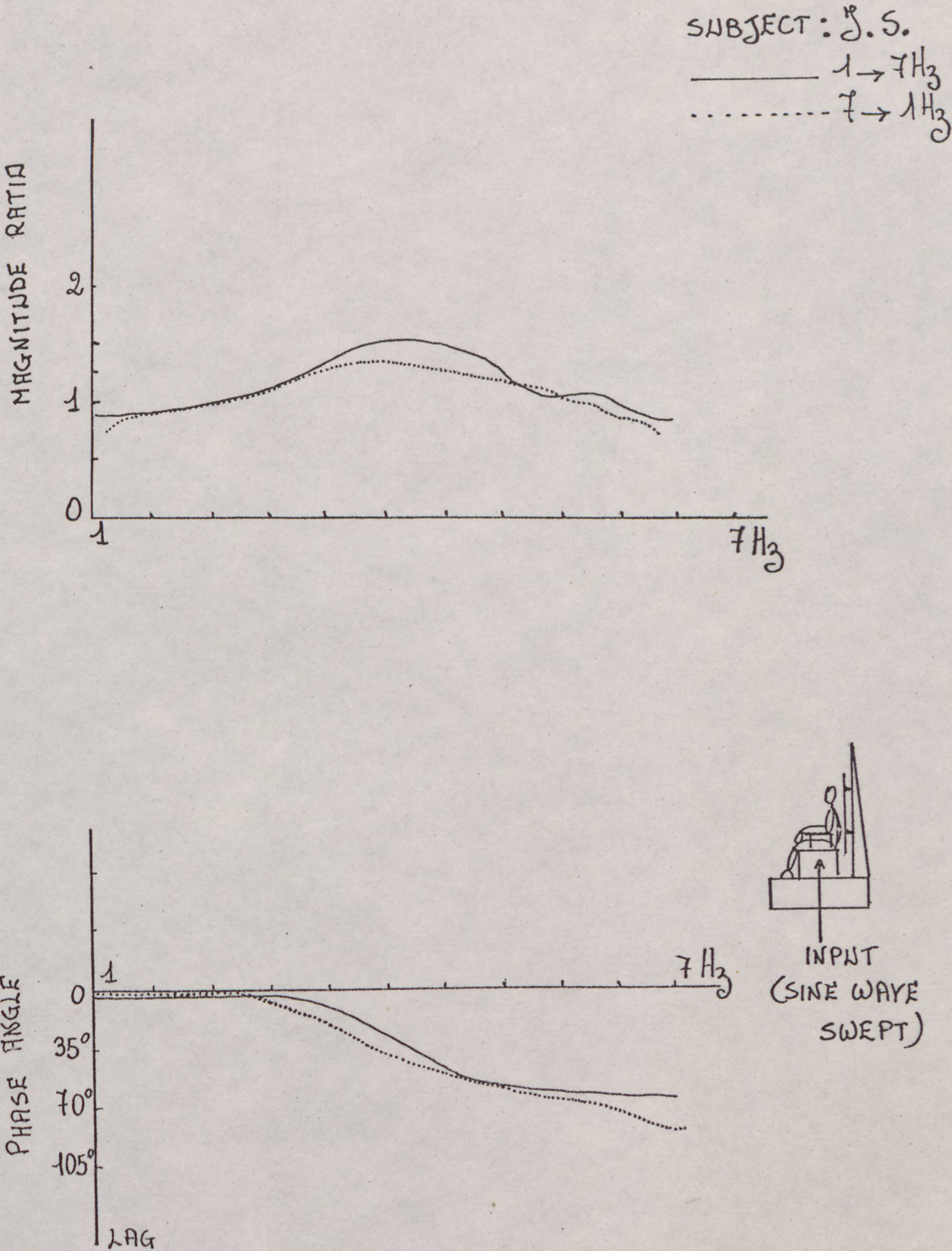


FIG. 7.12 FREQUENCY RESPONSE SWEPT EXCITATION TEST

(Subject not wearing the restraining neck support

APPENDIX 7.1.

SPECTRUM OF A SWEPT SINE WAVE

Let $x(t) = A \cos \Omega t \left(1 + \frac{\alpha}{2} t\right)$ for $t>0$ ... (1)

Define

$$X(j\omega) = \int_0^\infty x(t)\, e^{-j\omega t}\, dt =$$

$$= \int_0^\infty A \cos\left(\frac{\alpha\Omega t^2}{2} + \Omega t\right) e^{-j\omega t}\, dt \qquad \text{... (2)}$$

Making the substitutions:

$$\frac{\omega}{\sqrt{\alpha\Omega}} = \mu \qquad \sqrt{\frac{\Omega}{\alpha}} = \beta$$

gives:

$$X(j\omega) = \frac{A}{\sqrt{\alpha\Omega}} \int_0^\infty \cos\left(\frac{t^2}{2} + \beta t\right) e^{-j\mu t}\, dt$$

$$= \frac{A}{\sqrt{\alpha\Omega}} \int_0^\infty \left\{\cos\frac{t^2}{2}\cos\beta t - \sin\frac{t^2}{2}\sin\beta t\right\} e^{-j\mu t}\, dt$$

$$= X_1(j\mu) - j\,X_2(j\mu)$$

where

$$X_1(j\mu) = \frac{A}{\sqrt{\alpha\Omega}} \int_0^\infty \left\{\cos\frac{t^2}{2}\cos\beta t - \sin\frac{t^2}{2}\sin\beta t\right\} \cos\mu t\, dt$$

$$X_2(j\mu) = \frac{A}{\sqrt{\alpha\Omega}} \int_0^\infty \left|\cos\frac{t^2}{2}\cos\beta t - \sin\frac{t^2}{2}\sin\beta t\right\} \sin\mu t\, dt$$

Now

$$\sqrt{\alpha\Omega}\, X_1 = \frac{A}{2} \int_0^\infty \cos\frac{t^2}{2}\,(\cos\gamma t + \cos\delta t)\, dt\ -$$

$$\frac{A}{2}\int_0^{\infty} \sin\frac{t^2}{2}\,(\sin\gamma t + \sin\delta t)\,dt$$

$$\sqrt{\alpha\Omega}\;X_2 = \frac{A}{2}\int_0^{\infty} \cos\frac{t^2}{2}\,(\sin\gamma t - \sin\delta t)\,dt +$$

$$\frac{A}{2}\int_0^{\infty} \sin\frac{t^2}{2}\,(\cos\gamma t - \cos\delta t)\,dt$$

where

$$\gamma = \beta + \mu \qquad \delta = \beta - \mu$$

It follows that

$$\sqrt{\alpha\Omega}\;X_1\,(j\omega) = A\int_0^{\infty} \cos\frac{t^2}{2}\,(\cos\gamma t + \cos\delta t)\,dt \qquad \ldots (3)$$

$$\sqrt{\alpha\Omega}\;X_2\,(j\omega) = A\int_0^{\infty} \sin\frac{t^2}{2}\,(\cos\gamma t - \cos\delta t)\,dt \qquad \ldots (4)$$

These integrals are standard integrals [SNEDDON - The use of integral transforms]

Thus

$$\sqrt{\alpha\Omega}\;X_1 = \frac{A}{\sqrt{2}}\left\{\cos\frac{\gamma^2}{2} + \sin\frac{\gamma^2}{2} + \cos\frac{\delta^2}{2} + \sin\frac{\delta^2}{2}\right\} \qquad \ldots (5)$$

$$\sqrt{\alpha\Omega}\;X_2 = \frac{A}{\sqrt{2}}\left\{\cos\frac{\gamma^2}{2} - \sin\frac{\gamma^2}{2} - \cos\frac{\delta^2}{2} + \sin\frac{\delta^2}{2}\right\} \qquad \ldots (6)$$

These are the quadrature components of the frequency spectrum. The amplitude of the spectrum $|X\,(j\omega)|$ is given by:

$$|X\,(j\omega)| = \sqrt{\frac{X_1\,(j\mu)^2 + X_2\,(j\mu)^2}{2\alpha\Omega}} \qquad \ldots (7)$$

Substituting from (5) and (6) gives

$$|X\ (j\omega)|^2 = \frac{A^2}{\alpha\Omega}\ (1 + \cos\frac{\gamma^2}{2}\ \sin\frac{\delta^2}{2} + \cos\frac{\delta^2}{2}\ \sin\frac{\delta^2}{2})$$

$$= \frac{A^2}{\alpha\Omega}\left\{1 + \sin\left|\frac{(\beta+\mu)^2}{2} + \frac{(\beta-\mu)^2}{2}\right|\right\}$$

$$= \frac{A2}{\alpha\Omega}\left\{1 + \sin\ (\beta^2+\mu^2)\right\} \qquad \ldots (8)$$

Substituting back to the original parameters gives:

$$|X\ (j\omega)| = A\sqrt{\frac{1 + \sin\left(\frac{\Omega^2+\omega^2}{\alpha\Omega}\right)}{\alpha\Omega}} \qquad \ldots (9)$$

The shape of the spectrum is shown in Figure 7.4b CHAPTER 7. If α is small, i.e. the rate of change of frequency is small then the ripples in the spectrum coalesce and effectively the spectrum becomes "flat". If this signal is used as a test signal to identify a system, the system response will have ripples due to the above spectrum, but if α is small these will not be detectable.

APPENDIX 7.2

FAST FOURIER TRANSFORM PROGRAM

```
KILL  (1, ........., 12)

ACQUIR (20.48, 512 2, 1)

CONV (1, *, 4)

CONV (2, *, 4)

NORM (*, 4, 1)

NORM (*, 5, 1)

FFTA (3, 5, -1)

FFTA (4, 6, -1)

MOPH (5, 7, 0, 1)

MOPH  (6, 8, 0. 1)

ARITH (6, 10, 4, 5)

MOPH (10, 11, 0, 1)
```

```
∫ DISPLY (11, 0, 11, 1000)

END
```

Parameters:

1. Input file name
2. Output file name
3. Mode

   0 = forward with mirror image

   1 = inverse

   -1 = forward without mirror image

Notes:

1. The maximum number of samples, FFTA accepts is 4096 complex pairs. The number acquired is rounded down to the nearest power of two.

2. The sampling rate must be chosen at least twice the highest frequency of the data.

3. The data is reduced to zero mean to cancel any DC offset that the data may contain.

4. The data is then converted to have zeros in the imaginary part.

APPENDIX 7.3

PDP 11/50 DIGITAL COMPUTER SYSTEM

The system is based on PDP 11/50 digital computer having 96K of memory, 2 x 256K fixed head disc storage, 2 x 20 M moving head disc storage, 9 track magnetic tape, paper tape reader and punch, line printer, 35 cm incremental plotter, 2 x 15 bit word analogue to digital converters 2 x 16 channel multiplexer, 2 x 12 channels of sample and hold amplifiers, 2 x 2 channels of analogue filtering (24 dB/oct) 6 on-line terminals type. Tektronix 4010 having alpha-numeric and graphic capabilities 3 x LA 36 terminals and 2 x 14 track, analogue, tape recorders.

CHAPTER EIGHT

EVALUATION OF CONTROL

OF MOVEMENT

8.1. INTRODUCTION

Engineering methods have been applied to three major areas of medicine : diagnosis, therapy, and assistive devices. The latter include orthotic systems to aid weakened or paralysed limbs and prosthetic systems to replace missing limbs.

The main characteristics of biological systems which make them more difficult to study than engineering systems can be summarised as follows : [ALBERGONI-COBELLI-FRANCINI, 1974]

- evolutionary (instead of designed) system structures;
- hierarchical organisation;
- large dimensionality (number of variables very large and sometimes not well known)
- great interaction at all organisation levels (often varying with time);
- large dispersion from individual units to other units
- variability of the individual characteristics (dispersion with time);
- non stationarity in statistical problems;
- linearity assumptions generally represent only rough approximation;
- considerable limitation in the number of experiments repeatable in sufficiently similar conditions;
- limits to the experimentation in order to prevent damage;

- experimentation on animals can give useful information on the human structure and its working, but seldom on numerical values; in engineering simulation studies with models can also give useful numerical estimates;
- the study of biological problems often necessitates logical procedures based on pattern recognition procedures.

Feedback control aspects of physiological systems have been the object of much study.

Investigations of biological control have ranged from experimental studies which have applied the more standard tools of control system analysis, to theoretical studies of the general properties of systems. The main difficulty is to combine the knowledge of the physiologist with the methods of the engineer. The physiologist is trained to study complex systems which are the result of a long evolutionary history known to him only partially. Therefore he employs qualitative more often than quantitative methods and his procedures may not appear to be well defined to the engineer with a mathematical background. The engineer is trained to analyse systems which have been designed and built by man; they have been generally developed through the history of technology together with systematic methods of investigation. Therefore these systems are comparatively simple and well defined in their structure.

8.2. PERIPHERAL FEEDBACK MECHANISMS - OVERALL CONTROL

In chapters 2 and 4 we have looked at the physiological processes which are present in the motor system and in particular at the feedback elements associated with the skeletal muscle. We now turn to the question of motor function. Lack of a unifying theory has handicapped all experimental work on the motor system and especially in the field of rehabilitation, progress is largely empirical.

We are mainly concerned with the upper limb where there are two distinct problems - movement and prehension. The hand, which is essentially a gripping mechanism, uses many feedback loops and the stretch reflex, involving spindle feedback, has been identified as a vital element. However, the Golgi tendon organ which provides force feedback also is thought to be essential to the control of gripping forces. While it has been possible to establish feedback models of these two mechanisms, their simple character is unable to explain the wide variability of gripping characteristics.

It is believed that a threshold non linearity in the tendon organ characteristic and adaptation of the sensor parameters allows the grip to adapt to a wide range of load characteristics. This adaptation is achieved at a subconscious level and therefore does not involve excessive mental activity by the wearer.

Also present in the skin and the limb joints are sensors which give information on the shape and weight of an object, its dynamic state and also indicate the configuration of the limb in terms of its joint angles and their derivatives. It is believed that these play a vital role in co-ordinating the diverse muscle activities involved in any specific arm/hand function. This co-ordination reflects the skill that we acquire in performing manipulative tasks. There is of course an element of learning in such activities and this suggests that somewhere in the upper centres of the central nervous system we have the ability to store programmes of activity patterns. As yet this is not understood and there is little evidence to support the idea that numerical programming occurs. Indeed, knowing the response times of cells, the functioning time of the brain in certain activities is inconsistent with an algorithmic device. Recent experiments on animals also suggests that much more of the learned and rhythmic activity in the motor system is controlled by the peripheral feedback mechanisms.

It is possible to classify the basic limb functions into a set of important categories. [NIGHTINGALE-SEDGEWICK, 1979]

Thus:

(a) Ballistic movements - often surprisingly precise.

(b) Stereotype rhythmic patterns of movements - often involuntary in which only the general pattern is important and where exact timing may be of little consequence.

(c) Precise voluntary movements - where pattern, sequence and timing are important and often critical.

(d) Voluntary control of hand forces involving limited and often constrained movements by means of which objects are held and manipulated.

Some of the inherent limitations in the response of the motor system have been investigated by performing contrived tracking experiments. The tasks are manually very simple and thus eliminated the need for adaptation in the prehensile mechanisms and for co-ordination among the arm muscles since their relative activities are constrained by the linkage to be operated. In this case the dominant feedback mechanisms involve the outer loops which are closed by feedback from the exteroceptors and in particular by vision. A visual input then allows a single input-output model to be investigated and conventional control techniques, e.g. sinusoidal response, random inputs etc. allow a transfer function description to be formed.

In section 8.3 a short review of tracking studies and transfer function descriptions are given. Results show the dead - time associated with the human reaction time and also that there is a strong predictive component (phase lead) which effectively stabilises otherwise unstable situations. More impressive is the fact that the loop sensitivity (gain) and the derivative action vary with time indicating an ability to tune the loop to a given load and to improve the performance with repetition (learning). Also performance, and hence the model parameters vary with the type of input, being more favourable for smooth, near periodic signals but conservative for random or discontinuous inputs.

Most of the experiments have involved a subject whose body is stationary with only the arm moving. However this is not typical of the way that we perform tasks where there is often gross body movement associated with locomotion or with external disturbances. Effectively this introduces a new input variable in the loop and it is this situation that we are attempting to describe. Simple observation shows that despite these disturbances which, transmitted to the torso produce unwanted movement at the shoulder, we are able to keep the hand relatively stationary in space and to continue to perform a manipulative task. This is an acquired skill and is believed to involve the vestibular apparatus as well as proprioceptive feedback. The skill is maintained for a time in the absence of visual feedback but is impaired by disease or trauma which disables th vestibular mechanism.

In studying the tracking loop with motional disturbances and with visual proprioceptive and vestibular inertial feedback, we seek to investigate its effect on the predictive element in the operator response and the extent to which the three feedbacks are linear and obey the superposition rule. In particular we seek to find the extent to which random motions disrupt the operator response. To achieve better understanding of the mechanisms and the role of the various feedbacks, it is opportune to examine the overall system involved in the control of posture and movement as shown in Figure 8.1. Details of the different blocks are summarised as follows:

Block 1: HIGHER CENTRES - Include the sensory-motor cortex (from which originate voluntary movements), the cerebellum, the basal ganglia and the reticular formation. Higher centres receive signals from muscle tendon, joint, and skin receptors and from visual, auditive, vestibular and muscle spindles, Tendon organs, joint receptors (these are all proprioceptors), Visual Auditory (often referred to as exteroceptors) organs. All this enormous flow of sensory data conveyed to the higher contres, is analysed, processed and utilized for posture and motor functions and it co-ordinates the activity of the circuits involved at all levels of the CNS and down to the spinal motoneurons which excite the motor units fibres.

Block 2: SPINAL CENTRES: These are connected to the higher centres through the ascending and discending pathways in the spinal cord. It must be remembered that if the spinal centres are destroyed, all the regulatory loops are interrupted and no voluntary or reflex movement can occur. Though the spinal centres can function automatically, their activity is mostly controlled by the higher centres. Infact the latter not only impose the reference signal to the internal loops, but they also control the gain of the different loops, adjusting the muscle tone, by means of signals sent to block 3 obtaining in this way parametric changes of the postural system without modifying the configuration.

Block 3: MUSCLE: The inputs to this block are the commands coming from the spinal centres through α motoneurons. The outputs are the forces exerted from the muscular apparatus. Forces depend on tension, length, velocity and fatigue of the muscle.

Block 4: MECHANICAL SYSTEM: This block contains the overall dynamic and static characteristics of the controlled system.

Block 5: MUSCLE SPINDLES: A description of their role was given in Chapter 2.

Block 6: GOLGI TENDON ORGANS: Also described in Chapter 2.

Block 7: PERCEPTIVE SENSORY SYSTEMS: This block summarises the functions of the various sensory systems such as the visual, vestibular and proprioceptive systems, directly or indirectly involved with the control and organisation of posture and mevement.

In the experiments to be performed, which involve vertical disturbances only, the arm will be kept horizontal to avoid co-ordination problems and relatively simple tasks will be involved.

A simple mechanical representation is shown in Figure 8.2. and a block diagram in Figure 8.3.

We especially wish to investigate the operator response to vestibular input (head motion) V. Not only is there, little or no quantitative evidence of its character but it is a vital component which is missing in the fabrication of prosthetic limbs. If these are to be used in a natural way, without excessive conscious effort, then some form of spatial stabilisation of the hand is necessary in some conditions. However the design of an inertial system is complex and its realisation would certainly be expensive. For these reasons it is highly desirable to know the importance of vestibular feedback in humans and to attempt to quantify it.

There are two main difficulties in an analysis of the model shown in Figure 8.3 from our planned experiments. Firstly, it is very difficult to obtain direct physical measures of the signal involved. This is partly a conceptual one since overall effort is the sum of a very large number of motor units firing randomly and partly the practical one of obtaining measurements by non-invasive techniques (essential for manual responses). Secondly, it is difficult to isolate the particular elements of the loops and to eliminate other spurious feedback loops such as audio channels which give unwanted clues to the task. We shall try, by introducing visual artefacts and more especially by physiological suppression techniques, to break some of the feedback loops. There is an obvious difficulty in breaking the vestibular loop which should contain the equivalent of double integration with respect to time since the sensor gives a measure of acceleration. There are of course many sources of proprioceptive feedback which would be relevant but we shall try to derive experiments which will particularly enhance feedback of head motion relative to the torso from sensors in the neck.

8.3. TRACKING STUDIES: A REVIEW

For the present study, man is viewed as an interrelated group of anatomical components. Techniques for evaluation of central nervous system performance are developed, based on control engineering principles and past studies on the evaluation of human performance under dynamic conditions.

These are reviewed in the following two sections. The first section is devoted to studies in man-machine control systems, which are concerned with our ability to interact with complex control systems. The second section reviews those studies concerned with man's ability to perform in isolation.

8.3.1. TRACKING STUDIES: MAN-MACHINE SYSTEMS

The literature on man-machine systems is quite extensive. The studies reviewed here were chosen only for their general value in understanding basic concepts. [OTIS, 1974]. In a tracking task the subject usually employs a handle by which he controls the position of a cursor. A moving target is presented to the subject, and he is expected to match the position of the target (CRATTY, 1967). By relating output to input, i.e. the target position to cursor position, one can determine a mathematical relationship which describes the behavior of the human operator during the tracking task. By utilizing randomly appearing inputs, one need not be concerned with the subject's learning capability, i.e. that capacity which permits him to modify his response with practice.

In general, human-operator models of compensatory tracking tasks consist of a quasilinear portion and a remnant portion as illustrated in Figure 8.4. The concept of a remnant term was introduced by TUSTIN (1947) to account for the fact that the human operator generates frequencies beyond the bandwidth of his input. The remnant term represents that portion of the output which cannot be ascribed to a linear operation on the input.

Lumping the nonlinear components into the remnant term permits the utilization of a linear model to describe the input-output relation for those frequencies contained in the input. The term quasilinear is applied to this portion of the human operator model because different linear models are obtained for different input functions (GRAHAM AND MCRUER, 1971). Tustin (1947) suggested that the linear portion of the human operator can be represented by the transfer function

$$H_t(s) = \frac{K_t(1 + T\alpha s)\ e^{-T ds}}{s}$$

where K_t is the human operator gain constant, T_d is the transport delay time, and T_α is the numerator time constant. The transfer function is a lumped model which includes effects of the visual system, higher centres, and peripheral neuromuscular system. Tustin's classical work demonstrates how tracking models may account for those output characteristics which are linearly correlated with the input and those which are not.

McRuer and Krendel (1959) suggested a linear adaptable model of the human operator for randomly appearing visual inputs and motion outputs, which is applicable to the results of investigators prior to them. The most general form for this comprehensive model is

$$Y_p(s) = \frac{K_p\ e^{-T_r S}\ (\lambda\ T_i\ s + 1)}{(T_i S + 1)(T_n S + 1)} \qquad (8.1)$$

where

T_r is the reaction time delay

T_n is the neuromuscular lag time constant

K_p is the gain

$\frac{(\lambda\ T_i s + 1)}{(T_i s + 1)}$ represents the equalising ability of the operator.

The optimizing capability of the human operator is reflected in the works of McRuer and Krendel (1959): "The operator transfer function for a given task is very similar to the one that a servo engineer would select if he were given an element to control together with a black box having within it, elements making up the describing function given by the equation 8.1. and knobs on the outside for adjustment of λ_1 T_i, and K_p".

They conclude that the human will modify his transfer characteristics until he has attained an acceptable level of performance, or reached his performance limit.

Magdaleno and McRuer (1971) have modelled a pilot's neuromuscular subsystem during a visual tracking test to obtain describing function data for the muscle/manipulator actuation element and the whole human, for both hand manipulators and rudder pedals, respectively. Their general control situation is illustrated in Figure 8.5. and an illustration of the neuromuscular subsystems considered is shown in Figure 8.6. The parameter values for the muscle/manipulator dynamics were determined by correlating EMG activity with limb position. The parameter values assumed for the muscle spindle were based on earlier literature. The joint receptors were modelled with a pure gain and delay term. The human operator's tracking capabilities were then evaluated for first - and second order controlled element dynamics, respectively. The subject displayed a low frequency lead when second-order process dynamics were involved in the tracking task. It was concluded that the peripheral neuromuscular system remained essentially unchanged during the two tracking conditions and, consequently, that the central nervous system had adapted to the change in process dynamics.

8.3.2. TRACKING STUDIES - MAN

The dynamic characteristics of the wrist rotation system has been evaluated for freewheeling (open-loop) conditions and during a visual tracking task (STARK, 1968). The maximum freewheeling frequency obtained was approximately 8 Hz, whereas, during tracking conditions, with unpredictable inputs, the subject could not track beyond 3 Hz. The results suggest that the frequency limitation during tracking is due to the processing of sensory information and the determination of an appropriate motor response.

Tracking studies which are independent of the visual system have also been conducted. Neilson (1972) performed tests for rotation at the elbow joint. During the freewheeling experiments his subjects attained maximum frequencies of 4 - 6 Hz. When one elbow was passively moved in an irregular fashion and the blindfolded subject was instructed to match the movement with his other elbow, the maximum attainable following frequency was 2 Hz. Based on the results of these studies, it is reasonable to assume that during these proprioceptive tracking tasks the major factors limiting performance are central processing mechanisms. Thus, a subject's central nervous system characteristics are reflected in the overall frequency response measurements obtained from the tracking test and, consequently, the tracking tests cited above are useful techniques for assessing the dynamic performance of human subjects.

Attempts have been made to relate the parameters of transfer functions describing abstract models, to physical processes; for example, time constants associated with the mechanical processes in muscles, time delays in neural processing and transmission, and so on. However the parameters are found to vary with the task and the environment. Also the response may contain frequencies not present in the input, thus casting doubt on the validity of a linear model. Other experimental evidence (Bekey, 1963) suggests that a) the operator is discontinuous; b) the operator has the ability to predict the response when the input is briefly removed; c) the operator adapts to changes in the dynamics or kinematics of the tracking problem.

While all the experiments above are a tribute to the flexibility and power of the central nervous system and perhaps give an indication of the general strategies employed, they really do little to throw light on the control structures involved in our particular task and still less on the general system organisation.

CHAPTER 8

LIST OF FIGURES

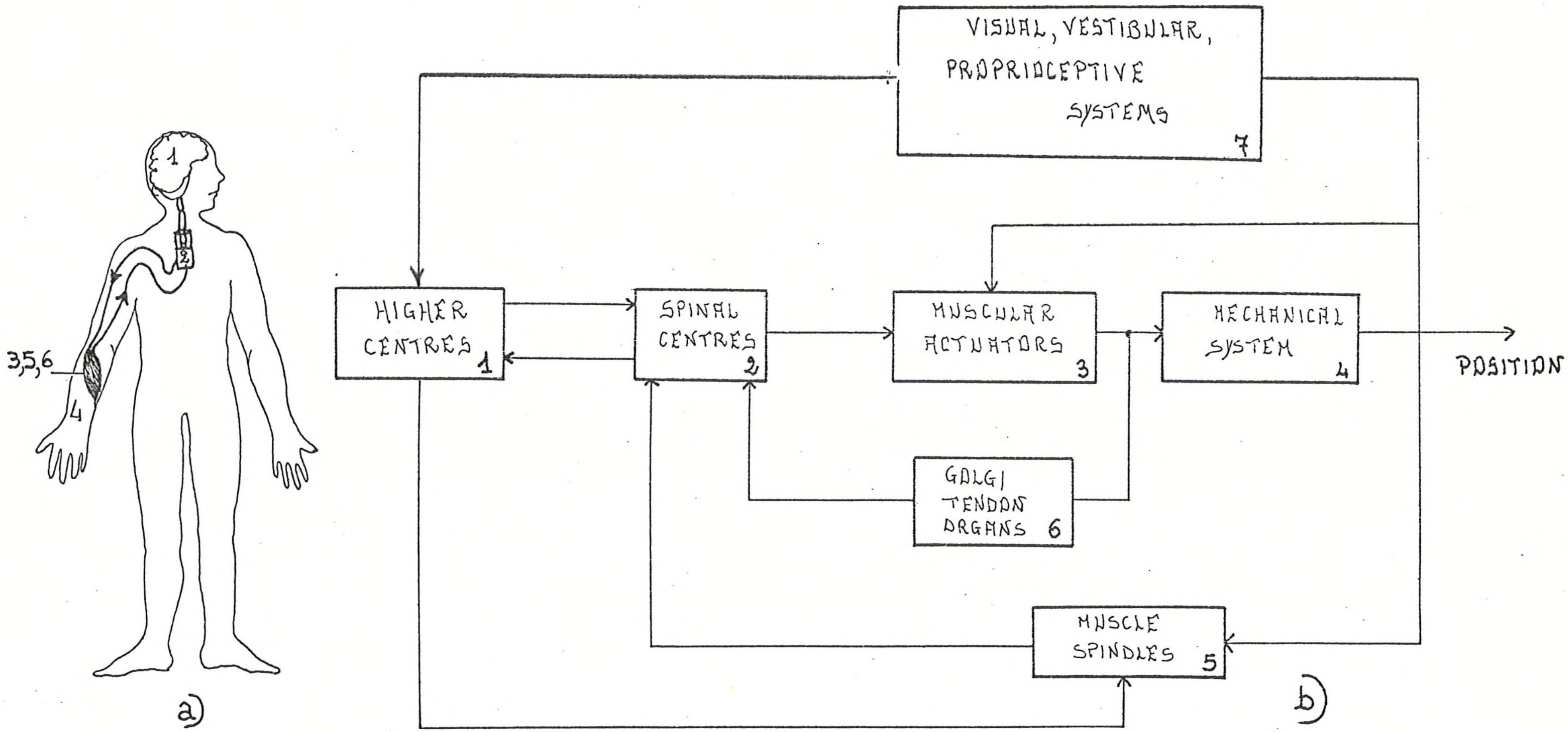
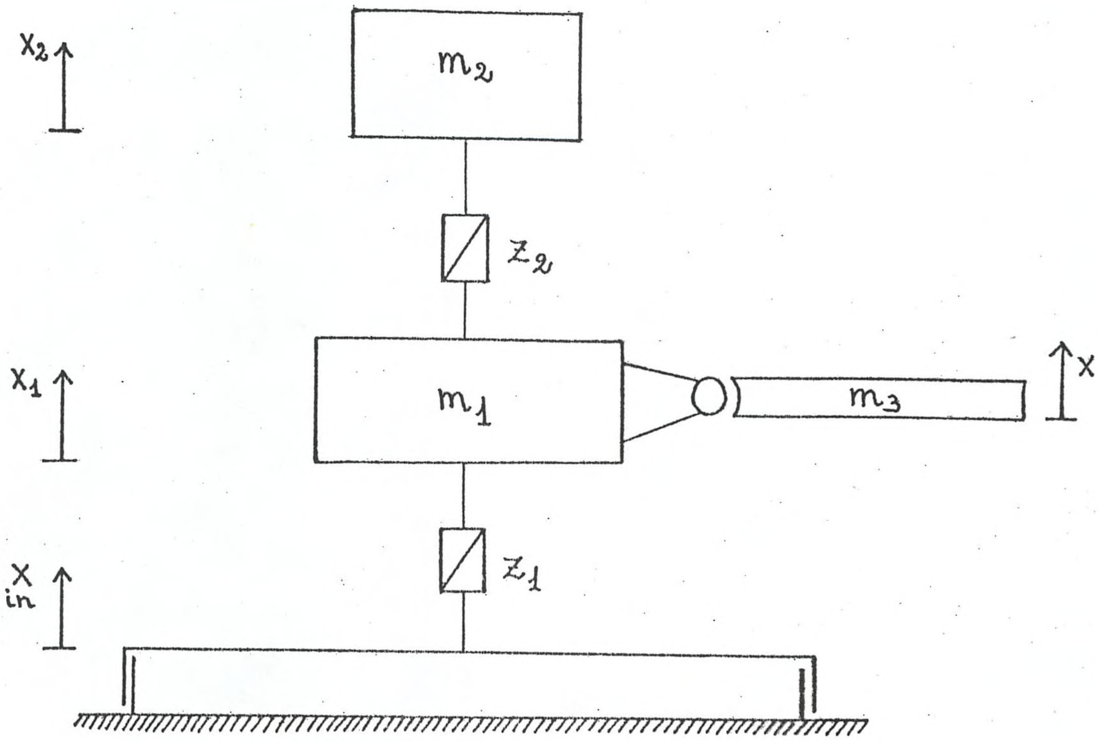


FIG. 8.1 POSTURE AND MOVEMENT CONTROL SYSTEM : a) ANATOMICAL SITUATION
b) BLOCK DIAGRAM



X_{in} = INPUT (EXTERNALLY APPLIED MOTION)

m_1 = EFFECTIVE MASS OF THE TORSO

m_2 = EFFECTIVE MASS OF THE HEAD

m_3 = EFFECTIVE MASS OF THE UPPER ARM

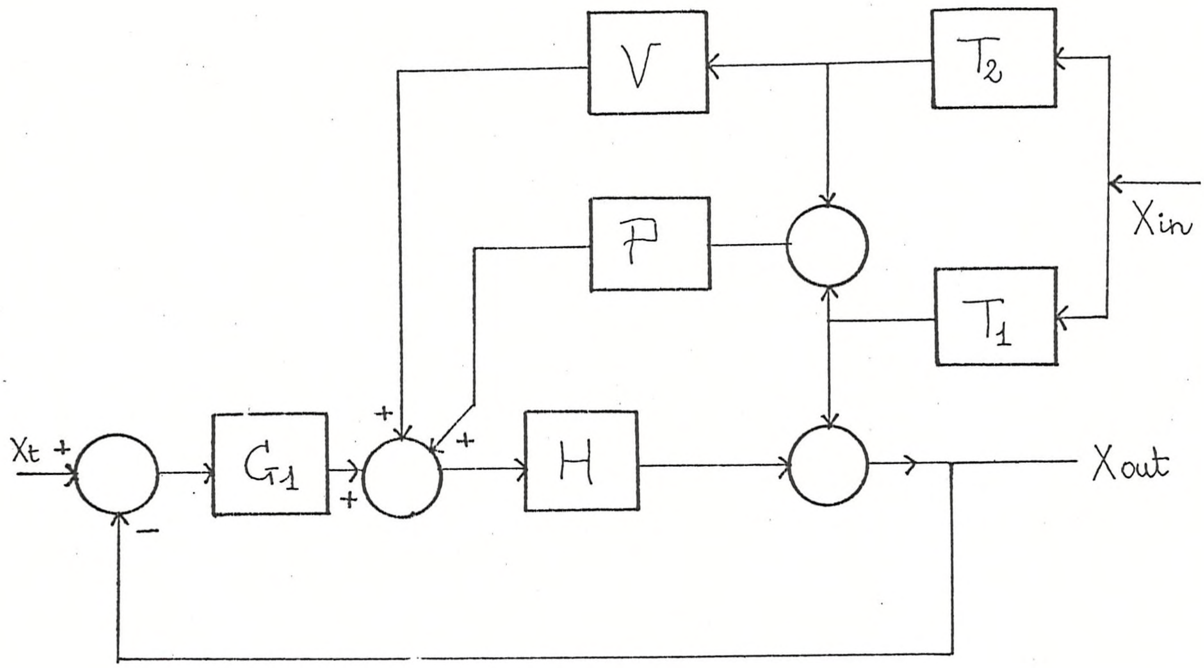
X_1 = RELATIVE DISPLACEMENT OF THE TORSO

X_2 = RELATIVE DISPLACEMENT OF THE HEAD

X = RELATIVE DISPLACEMENT OF THE UPPER ARM (OCCASIONED BY MUSCLE ACTIVITY)

Z_1, Z_2 = EFFECTIVE IMPEDANCES OF THE SUPPORTING STRUCTURES

FIG. 8.2 SIMPLE MECHANICAL REPRESENTATION OF THE HUMAN BODY



X_t = TARGET MOTION

X_{in} = MOTION OF SEAT (EXTERNAL DISTURBANCE)

T_1 = MOTION TRANSFER FUNCTION GIVING MOTION OF TORSO

T_2 = MOTION TRANSFER FUNCTION GIVING MOTION OF HEAD

G_1 = OPERATOR RESPONSE TO VISUAL INPUT

P = OPERATOR RESPONSE TO NECK PROPRIOCEPTION

V = OPERATOR RESPONSE TO VESTIBULAR INPUT (HEAD MOTIONS)

H = GROSS LIMB RESPONSE TRANSFER FUNCTION

FIG. 8.3 BLOCK DIAGRAM OF THE CONTROL SITUATION UNDER INVESTIGATION

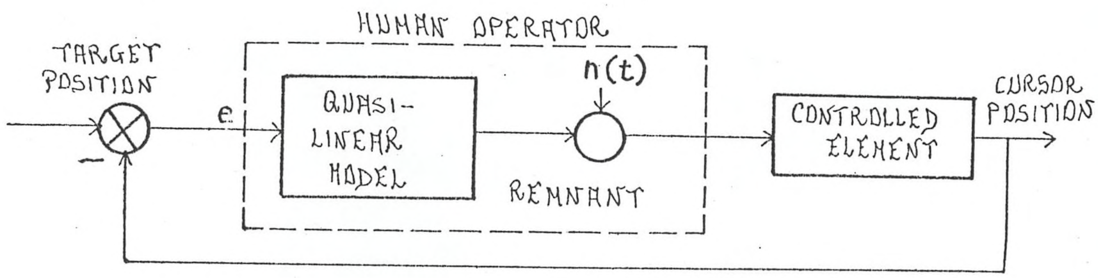


FIG. 8.4 GENERAL SCHEME FOR HUMAN OPERATOR TRACKING MODEL

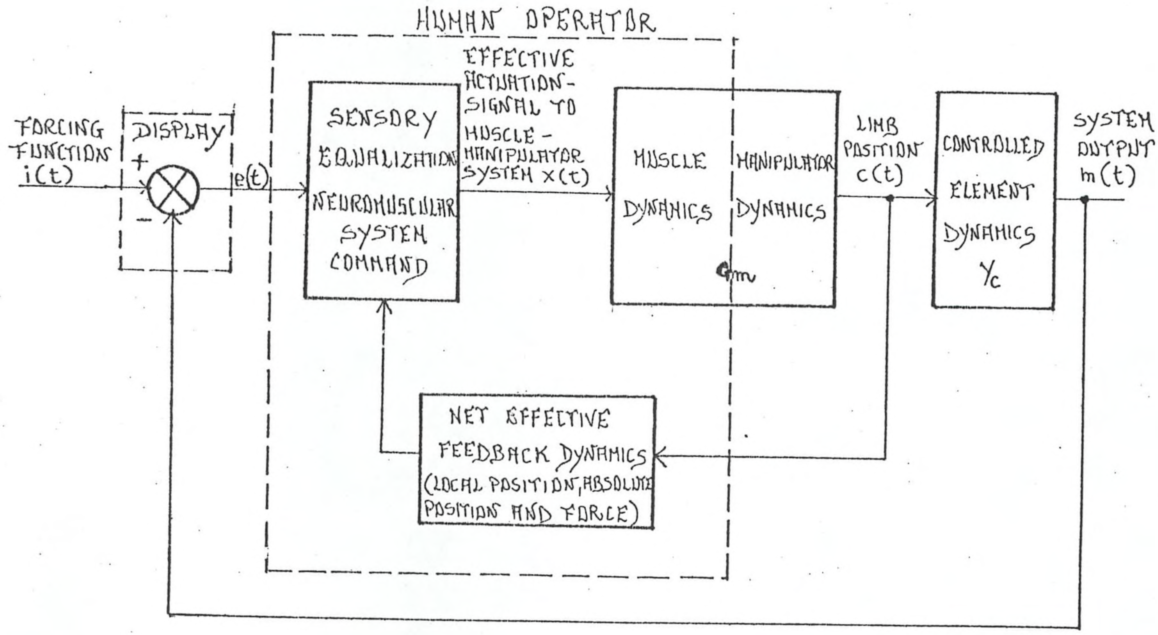


FIG. 8.5 MODEL FOR PILOT'S CONTROL SITUATION (Magdaleno and McRuer 71)

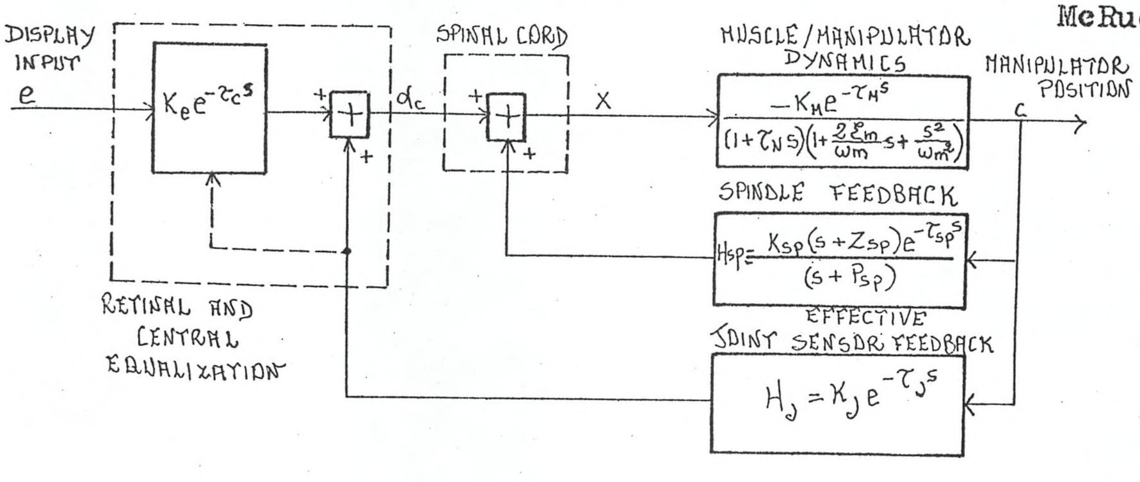


FIG. 8.6 MODEL FOR PILOT'S NEUROMUSCULAR SUBSYSTEMS (Magdaleno and McRuer, 1971)

CHAPTER NINE

THE EXPERIMENTAL PROGRAM

9.1. INTRODUCTION

This chapter will describe the experimental program devised to investigate the characteristics of the control model proposed in the previous chapter, involving the vestibular and proprioceptive apparatus.

Figure 9.1. shows fully the block diagram model we propose, while Figure 9.2. shows the proposed control model structure superimposed on the physiological model. It must be stressed, that the diagram shown can only be interpreted as a conceptual model though it partly resembles that put forward by other workers. Since it is not possible to find physiological equivalents for all the variables in such a model and, since those which we can identify cannot always be accessed during an experiment, we must try to contrive separate experiments to isolate, whenever possible, the individual components in the system.

Finally some attempts must then be made to determine if the values obtained for the parameters of the individual components are valid under conditions where the system performance is considered to depend on all of its mechanisms.

As can be seen, the block diagram of Figure 9.1. shows some slight differences from Figure 8.3. These are structured so that an estimate $\hat{X}_4$ of the vertical hand movement can be obtained from the vestibular sensors and the neck and shoulder proprioceptors. This particular form seems reasonable in the absence of a visual input where we need to model our undoubted ability to estimate the position of the hand despite gross body movement. In this model we need to determine the following characteristics:

$H(s)$ - gross limb response transfer function. This is a closed-loop response since we are not interested in separating the spindle, Golgi tendon organs etc. We also suppose that the external load does not vary substantially during tests. We will not investigate this function in detail.

$G_1(s)$ - operator response to visual input. This describes the central processing of a visual error. A number of studies have been concerned with this function but we will not investigate it in detail.

$G_2(s)$ - transfer function describing the response of the central processing unit to an estimated hand position error.

V(s) - assumed transfer function describing the response of the vestibular sensors.

P(s) - assumed transfer function describing the neck proprioceptors.

S(s) - assumed transfer function which relates the transmission of length sensory information in the shoulder muscles to the motor cortex.

$T_1(s)$ - transfer function describing the motion of the torso.

$T_2(s)$ - transfer function describing the motion of the head.

It must be noticed that V(s), P(s), S(s) are not identifiable mechanisms (i.e. simple mechanisms) but they are functions which we propose be there.

Two different kinds of experiments, open-loop and closed loop, were performed and these will be described in sequence. In the open-loop experiments, the limb was not moved and so the vestibular and proprioceptive sensors were not used in a limb feedback loop. Thus S(s) is eliminated from our considerations and $G_1(s)$ and H(s) are separable. These tests then enable us to determine V(s) and P(s) directly. The closed loop experiments involve the vestibular and proprioceptors essentially in a feedback situation and it is necessary to separate out their effects as well as other spurious, but useful feedback signals. This is a quite difficult task and the results which are offered would be difficult to corroborate exactly.

Since the main purpose of our model is to investigate the possible benefits of an inertial sensor system in prosthetic arm control, the success of a system based on the results of the experiments reported here may be the only justification feasible.

The models proposed are linear and there is of course no justification for assuming that exteroceptor signals add linearly. Finally, when proposing reduced models to describe the conditions which apply during particular experiments, the assumptions made have been somewhat severe. The reason for this approach is simplicity. In such a speculative venture there does not appear to be justification for undue complexity.

The individual experimental configurations will now be described in detail together with the particular model structure assumed. The experimental results will be described in subsequent chapters and then the results interpreted.

One important verification of these models would result from a study of the deltoid muscle activity. EMG signals give a crude indication of the neurological input to the muscle and we will describe our attempts to use EMG to verify our models.

9.2. OPEN-LOOP EXPERIMENTS

These experiments all involve the subject being displaced vertically by seat vibrations. The arms were in a resting position and the subject response was voluntary adjustment of a small potentiometer. This was held in the left hand and the slider adjusted by the right hand, Figure 9.3. In front of the subject, at head level, was an oscilloscope display. Two channels were displayed one being the input, in this case the seat displacement, the other being a voltage proportional to the potentiometer displacement which was desired to follow the seat displacement.

In some early experiments the output was the force exerted by the subject under test, on a load-cell but the results here were disappointing since subjects found it difficult to control force with sufficiently accuracy.

As mentioned before spurious sensory signals such as the noise generated in the motor drive tended to give unwanted clues to the platform position these effects were masked out either by the use of headphones producing fairly high intensity broad-band audio noise or by putting earphones or mufflers on the subject.

In trying to remove proprioceptive feedback from muscles in the neck, several methods were tried. Of these the neck collar, see Chapter 7, was probably the most justifiable,

In trying to impair the vestibular feedback attempts were first made to achieve this by indirect methods where cancellation feedback was artificially introduced in the output display. This method was not very successful and so reluctantly we turned to physiological methods to inhibit the sensors. In this work the author is indebted to Dr. Sedgewick and Dr. Snashall for their guidance and clinical supervision. The technique of caloric stimulation is well-known to physiolcgists but a brief summary follows:

9.2.1. CALORIC STIMUATION OF VESTIBULAR APPARATUS

The procedure is based on the known observation that prolonged application of cold substances in the ear drum causes loss of balance and nystagmus always occurs automatically when the semicircular canals are so stimulated. The external semicircular canal lies adjacent to the ear, and cooling the ear can transfer a sufficient amount of heat from this canal to cool the endolymph. This increases the density of the endolymph, thereby causing it to sink downward, resulting in slight movement of fluid around the semicircular canal. This stimulates the canal, giving the individual a sensation of rotation and also initiating nystagmus.

Before each test involving caloric stimulation, a clinical examination of the subject's ear was made with an othoscope to verify that there was no damage or disease processes which could render the experiment either dangerous or misleading. The ear was then cleaned of wax. From a preparation of iced-water at a measured 0°C a quantity of approximately 120 cc was perfused through the auditory canal using a syringe. The irrigation was performed with the subject in a sitting position and the head tilted sidewards 60°. Tests using left and right ears were made alternatively.

The total duration for the cold water to be in the ear was about 30 seconds. The major effect of the immersion lasts about 60 seconds and this is just long enough to perform meaningful tests. The effect disappears completely after 3-5 minutes.

Early experiments used only a cold water bolus for 20 seconds duration and were somewhat less effective. The following are the variations of open-loop tests performed with the potentiometer output.

9.2.2. TRACKING EXPERIMENT A

In this experiment the subject had access to the visual display. There was no caloric stimulation and no restriction of the head. Although the vestibular and proprioceptor mechanisms are not inhibited, it is considered that availability of a visual feedback will result in this loop being dominant. Thus the simplified block diagram of Figure 9.4, is adopted.

If we call X_1 the displayed seat displacement and y the subject's measured response, the tracker-loop transfer function will be:

$$R_1(s) = \frac{y(s)}{X_1(s)} \qquad (9.1.)$$

9.2.3. TRACKING EXPERIMENT B

The conditions here are the same as in experiment A except that caloric stimulation was applied. If, as we summarized earlier the visual loop swamps the vestibular response, then the results of experiments A and B should differ very little.

bxn

9.2.4. TACTILE EXPERIMENT C

For this experiment the oscilloscope display was not presented to the subject who was blindfolded. A neck collar was provided but no caloric stimulation was applied. The reduced block diagram is shown in Figure 9.5.

In the absence of visual feedback the subject must estimate the seat displacement from his vestibular output. If we call V(s) the assumed transfer function describing the response to vestibular sensors, $T_2(s)$ the transfer function giving motion of the head, X_3 the motion transmitted to the head, we can assume that the output of the vestibular sensors V(s). $X_3(s)$ is used as the estimate.

The response would then be:

$$Y(s) = T_2(s) \cdot V(s) \cdot R_1(s)\ X_1(s) \tag{9.2.}$$

or

$$R_2(s) = \frac{y(s)}{X_1(s)} = T_2(s)\ V(s)\ R_1(s) \tag{9.3.}$$

where $T_2(s)$ is assumed known from the transmissibility tests of Chapter 7 and $R_1(s)$ has been determined from experiments A and B.

Thus V(s) follows by division:

$$V(s) = \frac{R_2(s)}{T_2(s)\ R_1(s)} \tag{9.4.}$$

9.2.5. TACTILE EXPERIMENT D

Here the conditions are as in experiment C but now the neck support is removed. The block diagram is shown in Figure 9.6.

In this case since there is some information concerning the head position relative to the shoulder, a corrected estimate of $\hat{X}_1$ can be formed. For simplicity it is assumed here that the sensor outputs add. Thus if we call $T_1(s)$ the transfer function giving motion of the torso, P(s) the assumed transfer function describing the neck proprioceptors, X_3 the motion transmitted to the head, X_2 the motion transmitted to the shoulder, it follows:

$$\hat{X}_1(s) = V(s)\ X_3(s) + P(s)\ [X_2(s) - X_3(s)] \qquad (9.5.)$$

where:

$$X_3(s) = T_2(s) \cdot X_1(s) \,//\, X_2(s) = T_1(s) \cdot X_1(s)$$

Substituting the latter in equation (9.5.)

$$\hat{X}_1(s) = V(s)\ T_2(s)\ X_1(s) + P(s)\ [T_1(s) \cdot X_1(s) - T_2(s) \cdot X_1(s)]$$

and therefore:

$$\hat{X}_1(s) = X_1(s)\ [V(s)\ T_2(s) + P(s)\ (T_1(s) - T_2(s))] \qquad (9.6.)$$

From the equation (9.1.)

$$y(s) = R_1(s)\ \hat{X}_1(s) = R_1(s)\ [V(s)\ T_2(s) + P(s)\ (T_1(s) - T_2(s))]X_1(s) \qquad (9.7.)$$

it follows that

$$y(s) = R_3(s) \cdot X_1(s) \qquad (9.8.)$$

where

$R_3(s)$ is equal to

$$R_1(s)\ [V(s)\ T_2(s) + P(s)\ (T_1(s) - T_2(s))] \qquad (9.9.)$$

Since $V(s) = \dfrac{R_2(s)}{R_1(s)\ T_2(s)}$, equation (9.4.), substitution in equation (9.9.)

gives:

$$R_3(s) = R_1(s) \left[\frac{R_2(s)}{R_1(s)\ T_2(s)} \cdot T_2(s) + P(s)\ (T_1(s) - T_2(s)) \right]$$

Thus

$$R_3(s) = R_2(s) + R_1(s)\ P(s)\ [T_1(s) - T_2(s)] \qquad (9.10.)$$

From which it follows that:

$$[T_1(s) - T_2(s)] \; P(s) = \frac{R_3(s) - R_2(s)}{R_1(s)} \qquad (9.11.)$$

From which, in principle, the proprioceptor transfer function can be isolated.

9.2.6. TACTILE EXPERIMENT E

This test involves no visual feedback, vestibular feedback is inhibited by caloric stimulation but there is no proprioceptive inhibition. The assumed model is shown in Figure 9.7. In this case there is no spatial reference in the absence of vision and assuming complete vestibular inhibition.

The output of the proprioceptors will, for periodic stimuli, give some indication of the seat displacement X_1.

Here:

$$\hat{X}_1(s) = P(s) \; [T_1(s) \; X_1(s) - T_2(s) \; X_1(s)] =$$

$$P(s) \; [T_1(s) - T_2(s)] \; X_1(s) \qquad (9.12.)$$

and therefore:

$$y(s) = P(s) \; [T_1(s) - T_2(s)] \; . \; R_1(s) \; . \; X_1(s) \qquad (9.13.)$$

or

$$\frac{y(s)}{X_1(s)} = R_4(s) \equiv P(s) \; R_1(s) \; [T_1(s) - T_2(s)] \qquad (9.14.)$$

substituting (9.11) in to (9.14) it follows that:

$$R_4(s) = R_3(s) - R_2(s) \qquad (9.15.)$$

or, if our assumption of superposition is true, then

$$R_3(s) = R_4(s) + R_2(s) \qquad (9.16.)$$

A measure of R_3 will facilitate estimation of P(s).

9.2.7. TACTILE EXPERIMENT F

It is known physiologically that the proprioceptors in the neck are enhanced by increasing their tension. One way to achieve this is to tilt the head slightly and experiments D and E were repeated under these conditions. Different values of P(s), the sensor sensitivity are calculated.

9.3. CLOSED LOOP EXPERIMENTS

The previous experiments were performed to obtain some measure of the sensitivity of the proprioceptive and vestibular sensors. However, the tests did not involve any controlled motion of the arm and so it was not possible to investigate the resulting control sequence. Further experiments were devised to enclose the sensors in feedback loops so that a conjectured simple control structure can be modelled. In these experiments the arm is outstretched, figure 9.8, and the vertical height of the hand is monitored and compared with a reference height while the platform is subject to vertical displacements.

It was also intended to superimpose a hand tracking task on this experiment but initially the only problem investigated involved keeping the hand stationary. In these experiments the arm does move and strictly speaking we should include components of arm dynamics in the transmission equations. However, since there is no load on the arm and since movements are relatively slow, the arm dynamics are ignored in this study. The experiments and the proposed block diagram models are outlined as follows:

9.3.1. VISUAL EXPERIMENT G

In this experiment the subject can see his hand and the vertical reference. Thus visual proprioceptive and vestibular feedbacks are all present. Again this is largely a reference experiment for the muscle dynamics since we assume that the visual loop swamps out the other feedback effects. The resulting block diagram is shown in Figure 9.9.

If we call H(s) the gross limb transfer function, $G_1(s)$ the subject response to visual input, X_4 the hand response it follows

$$X_4(s) = T_1(s)\ X_1(s) - G_1(s)\ H(s)\ X_4(s) \tag{9.17.}$$

and therefore

$$R_5(s) \equiv \frac{X_4(s)}{X_1(s)} = \frac{T_1(s)}{1 + G_1(s)\ H(s)} \tag{9.18.}$$

If $\quad G_1(s)\ H(s) >> 1$

$$R_5(s) \simeq \frac{T_1(s)}{G_1(s)\ H(s)} \tag{9.19.}$$

9.3.2. BLIND EXPERIMENT H

Here the subject was blindfolded, so the visual loop was broken. However, both the proprioceptive and vestibular sensors are unimpaired. Therefore there is now no direct information on the hand position so we conjecture that the central processor in the brain compute the hand position and its error from the proprioceptor and vestibular sensor data.

Data on the deltoid muscle stretch are also required and it is thought that this is directly available from spindle receptors S, see Figure 9.10.

If r is the reference signal, $G_2(s)$ the transfer function describing the response of the central processing to an estimated hand position error, S(s) the assumed transfer function which relates the transmission of length sensory information from the shoulder muscles to the motor cortex, the hand position estimate, $\hat{X}_4$ is given by

$$\hat{X}_4(s) = \hat{X}_1(s) + S(s)\,X(s) \tag{9.20.}$$

where $X(s) = G_2(s)\,H(s)\,(r - \hat{x}_4(s))$ and (9.21.)

$\hat{X}_1$ is given by equation (9.6).

Therefore, substituting in (9.20) gives:

$$\hat{X}_4(s) = X_1(s)\,[V(s)\,T_2(s) + P(s)\,(T_1(s) - T_2(s)] +$$

$$S(s)\,G_2(s)\,H(s)\,[r - \hat{X}_4(s)] \tag{9.22.}$$

Furthermore,

$$X_4(s) = X(s) + X_1(s)\,T_1(s)$$

Thus the actual hand position is given by:

$$X_4(s) = G_2(s)\,H(s)\,(r - \hat{X}_4(s)) + X_1(s)\,T_1(s) \tag{9.23.}$$

From the equation (9.22.) it follows that:

$$\hat{X}_4(s) = \frac{X_1(s)\,[V(s)\,T_2(s) + P(s)\,(T_1(s) - T_2(s)] + S(s)\,G_2(s)\,H(s)\,r}{1 + S(s)\,G_2(s)\,H(s)} \tag{9.24.}$$

which when substituted in Equation (9.23.) leads to the following equation for the hand position in terms of the reference input r and the platform displacement X_1:

$$X_4(s) = \frac{G_2(s)\,H(s)}{1 + S(s)\,G_2(s)\,H(s)}\left\{r + X_1(s)\left[T_2(s)\,(P(s) - V(s)) + T_1(s)\left(S(s) - P(s) + \frac{1}{G_2(s)\,H(s)}\right)\right]\right\} \tag{9.25.}$$

If $G_2(s)\ H(s) >> 1$ i.e. the controlled gain is large and if the sensors are ideal (equally sensitive) i.e. $P(s) = V(s) = S(s)$ it follows:

$$X_4(s) = \frac{r}{S(s)} + \frac{T_1(s)}{G_2(s)\ H(s)\ S(s)}\ X_1 \simeq \frac{r}{S(s)} \qquad (9.26.)$$

Thus ideally the accuracy depends on the shoulder sensors while the error due to the platform vibration becomes very samll. Of course, in practice, there are many reasons why this ideal state of affairs is not achieved. One of the most likely sources of error is acceleration drift in the vestibular apparatus. This could be caused by coupling with other degrees of freedom of the head and would result in a growing error in the hand position. Such errors which worsen with time are certainly characteristic of positioning tasks performed without vision.

One feature of the block diagram, Figure 9.10. which cannot really be explained in physiological terms is the reference input. Conceptually this involves some kind of residual image generated from the visual system Certainly such image preservation is present to varying degrees of accuracy and, in cases of tracking a moving target, some degree of prediction is possible from the last visible state. However, a precise modelling of such an input process seems beyond our present understanding of the CNS.

Another feature which we have not tried to model is conscious participation in this task. Certainly initiation and termination of the response are volitional actions but there are probably adaptive activities during the task itself. To attempt to model these last two aspects would be to speculate on the internal structure and function of the motor cortex and the cerebellum. This is an aspect of biological system modelling for which we are very unprepared at present.

9.3.3. BLIND EXPERIMENT J

A series of experiments were performed which are basically similar to experiment H but in which the neck and shoulder proprioceptors are disrupted by various techniques, Figure 9.11. It was very difficult to completely inhibit such sensors and so no specific alteration to the block diagram seems to provide a reasonable basis for assessing these experiments.

Certainly a study of the transfer function and its dependence on S(s) and P(s) gives some indication of what to expect if these parameters are reduced in magnitude. Clearly since we loose some vital information on body configuration we should expect the overall performance to be degraded compared with experiment H.

9.4. CONCLUSIONS

In this chapter considerable speculation on the possible control structure which allows postural compensation has been attempted. These structures are represented by linear models and the results of the subsequent experiments will be expressed solely in terms of these.

The feedback models associated with simple hand position control are particularly speculative. In particular we meet the common problem concerning how or where a reference input is generated within a closed-loop physiological process. Even in more generally understood processes such as respiratory regulation the answers to this and similar questions are not readily available.

An essential feature of all the experiments so far described is that they have been non-physiological and non-invasive. From the model structures formulated it is possible to predict gross muscle activity in a given situation and, in subsequent experiments, see Chapter 11, it is proposed to investigate these predictions by EMG recordings of muscle activity.

CHAPTER 9

LIST OF FIGURES

Transfer Function defined on pag. 177

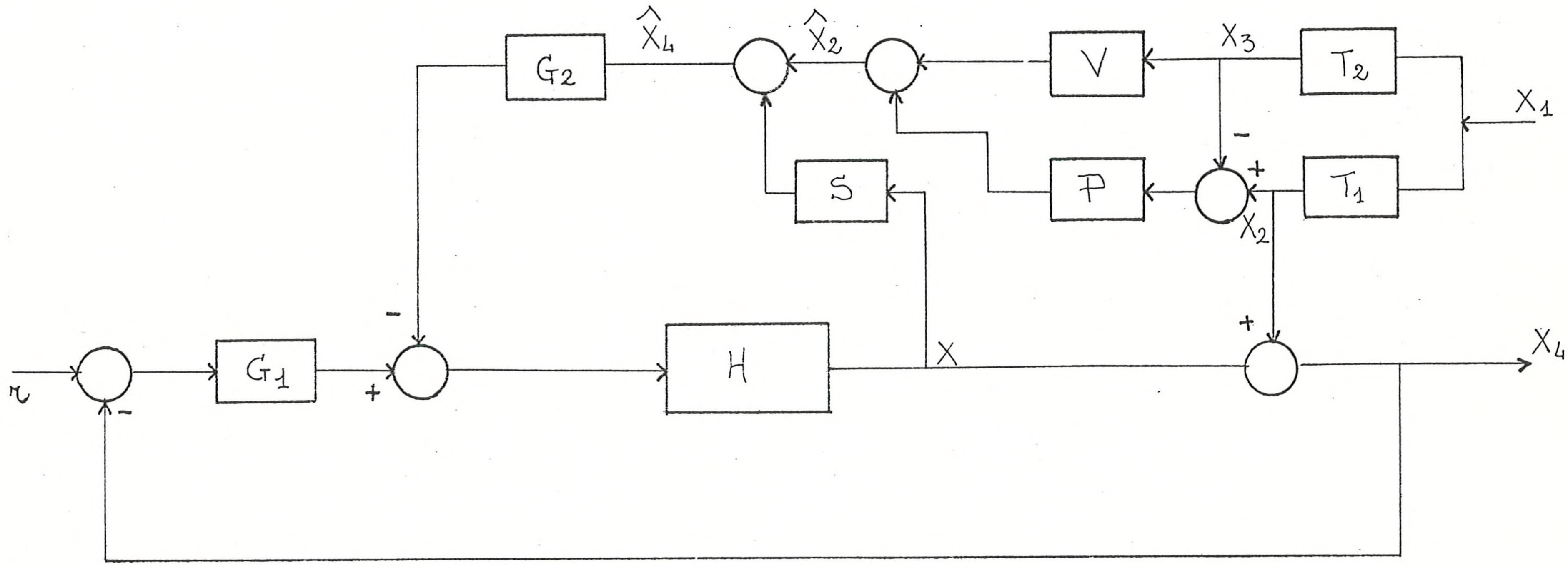


FIG. 9.1 COMPLETE BLOCK DIAGRAM OF THE PROPOSED MODEL

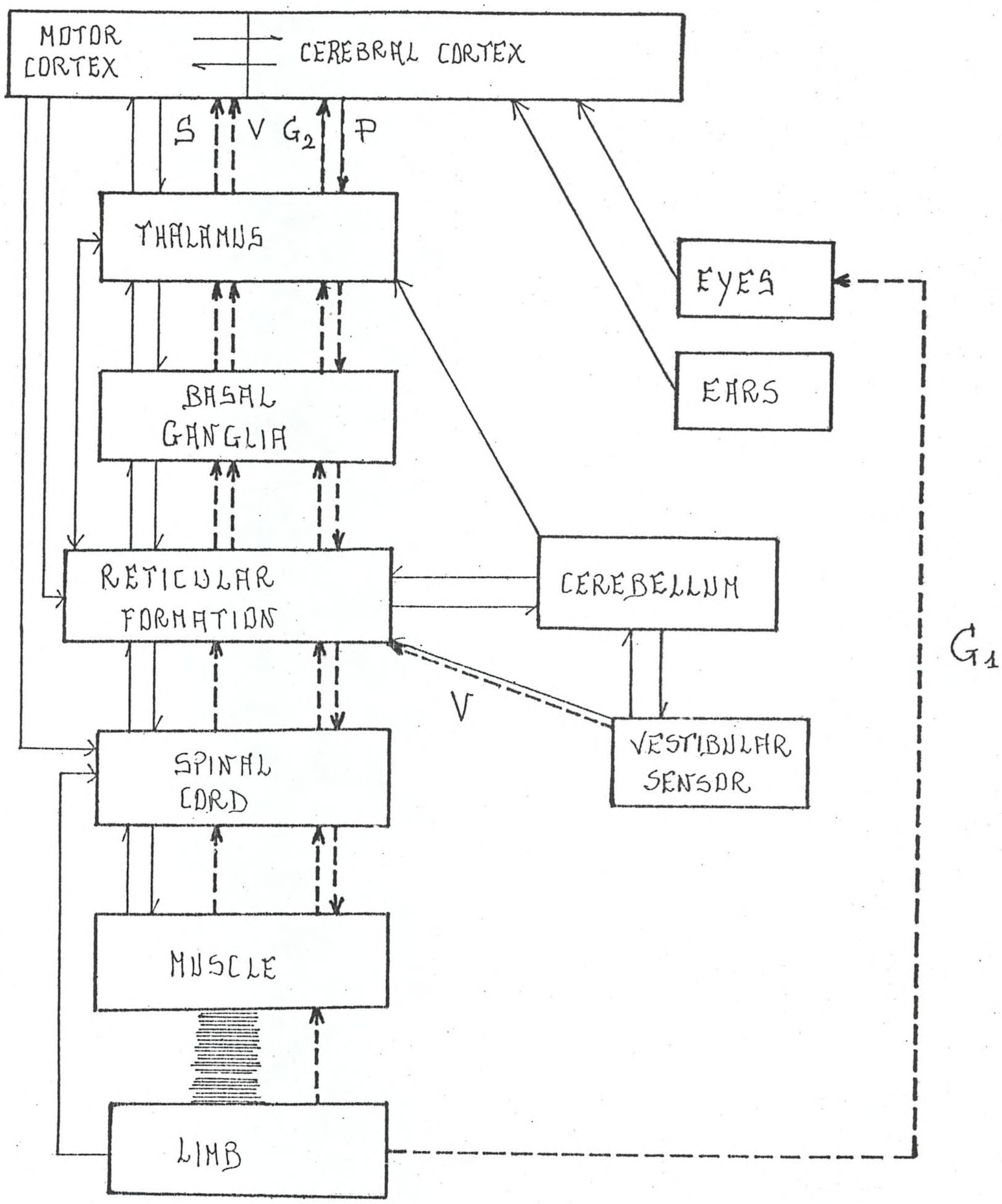


FIG. 9.2 BLOCK DIAGRAM SHOWING THE PROPOSED CONTROL MODEL STRUCTURE SUPERIMPOSED ON THE PHYSIOLOGICAL MODEL

FIG. 9.3 SUBJIECT UNDERTAKING OPEN-LOOP EXPERIMENTS

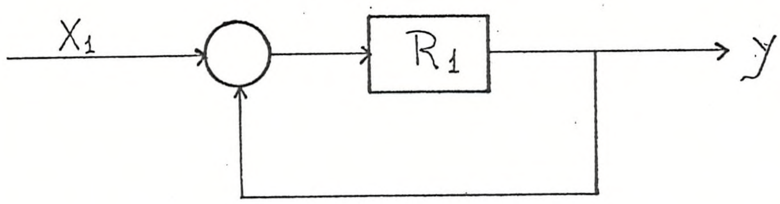


FIG. 9.4 BLOCK DIAGRAM - TRACKING EXPERIMETS A

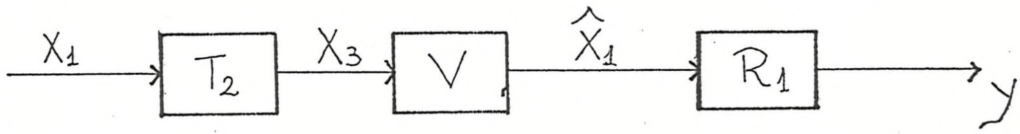


FIG. 9.5 BLOCK DIAGRAM - TACTILE EXPERIMENTS C

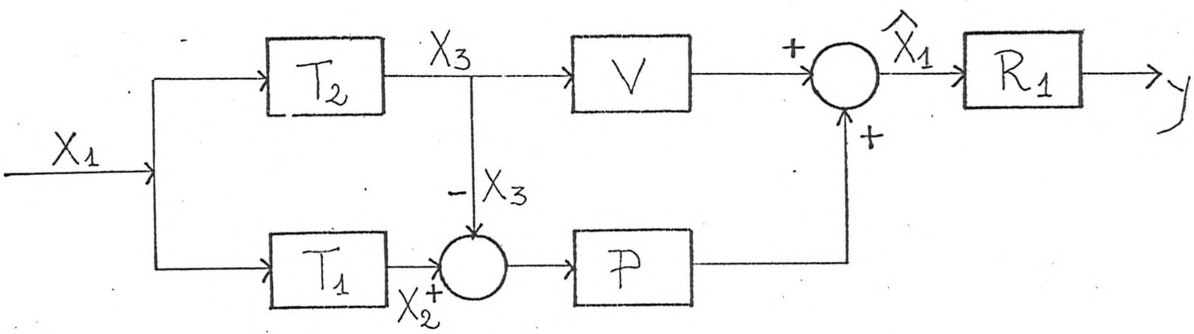


FIG. 9.6 BLOCK DIAGRAM - TACTILE EXPERIMENTS D

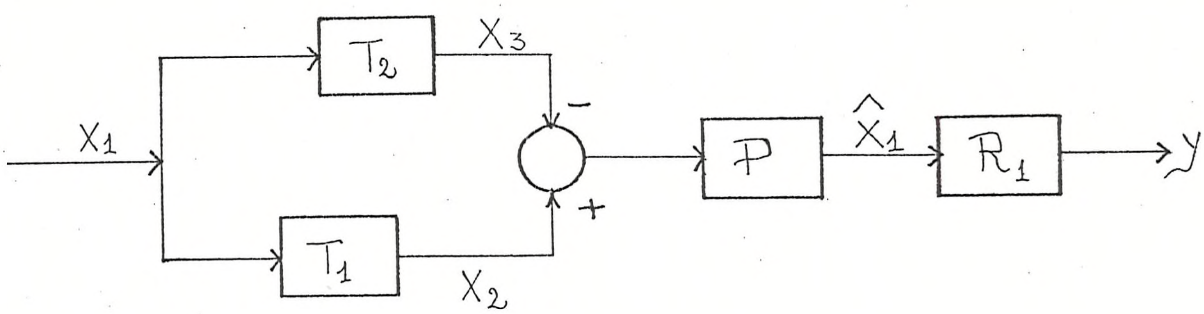


FIG. 9.7 BLOCK DIAGRAM - TACTILE EXPERIMENTS E

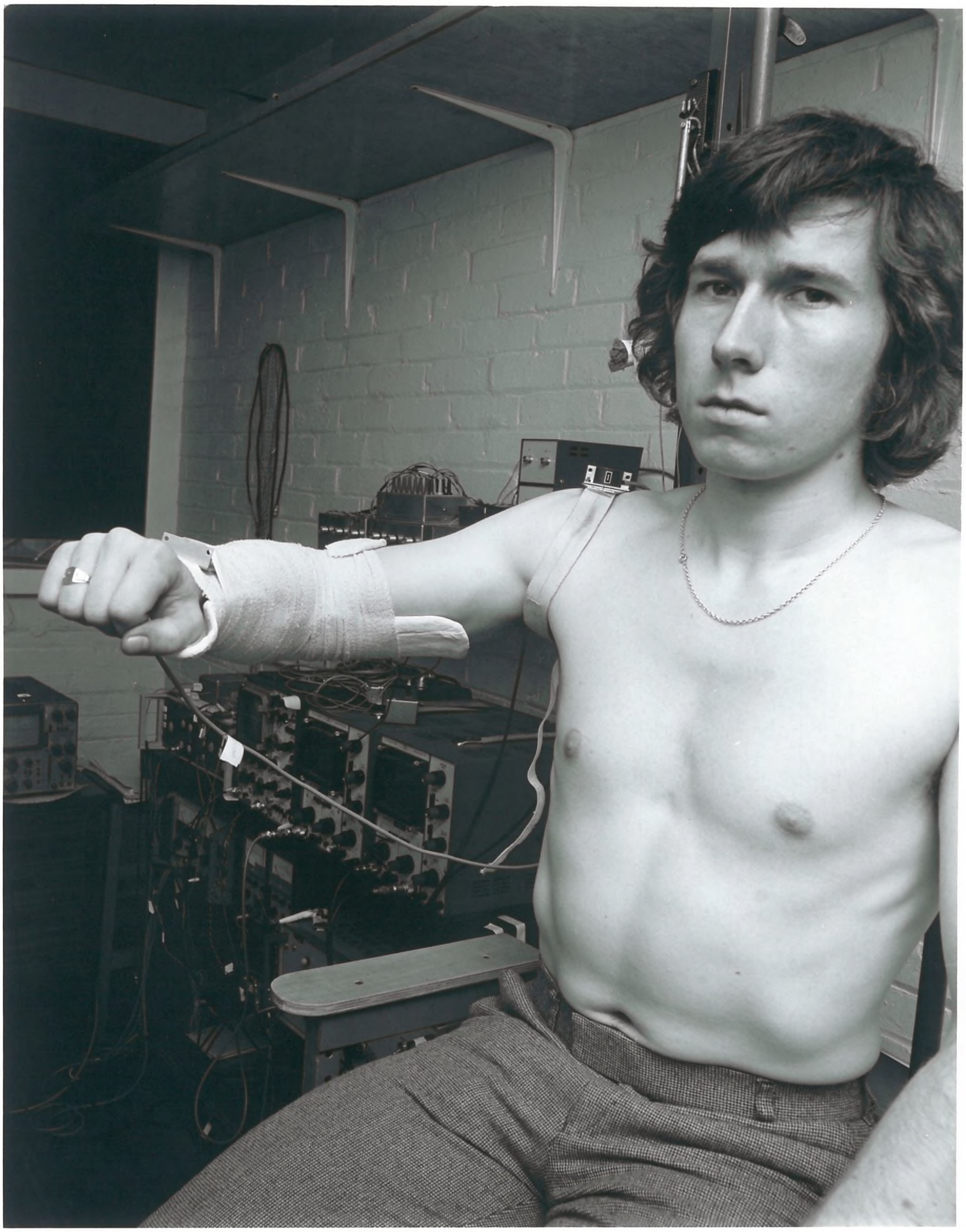


FIG. 9.8 SUBJECT UNDERTAKING CLOSED-LOOP EXPERIMENTS

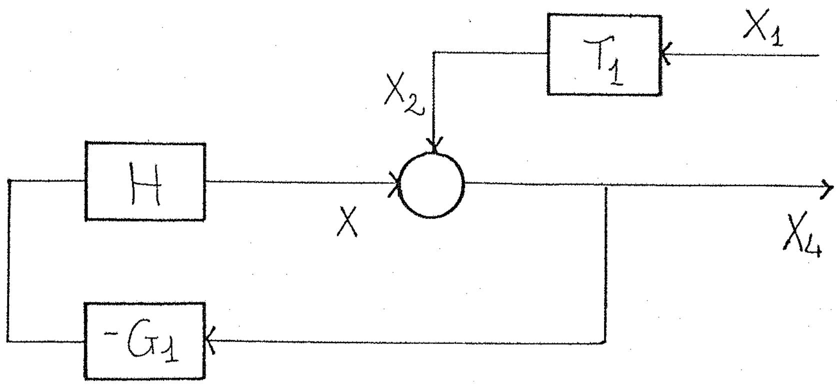


FIG. 9.9 BLOCK DIAGRAM - VISUAL EXPERIMETS G

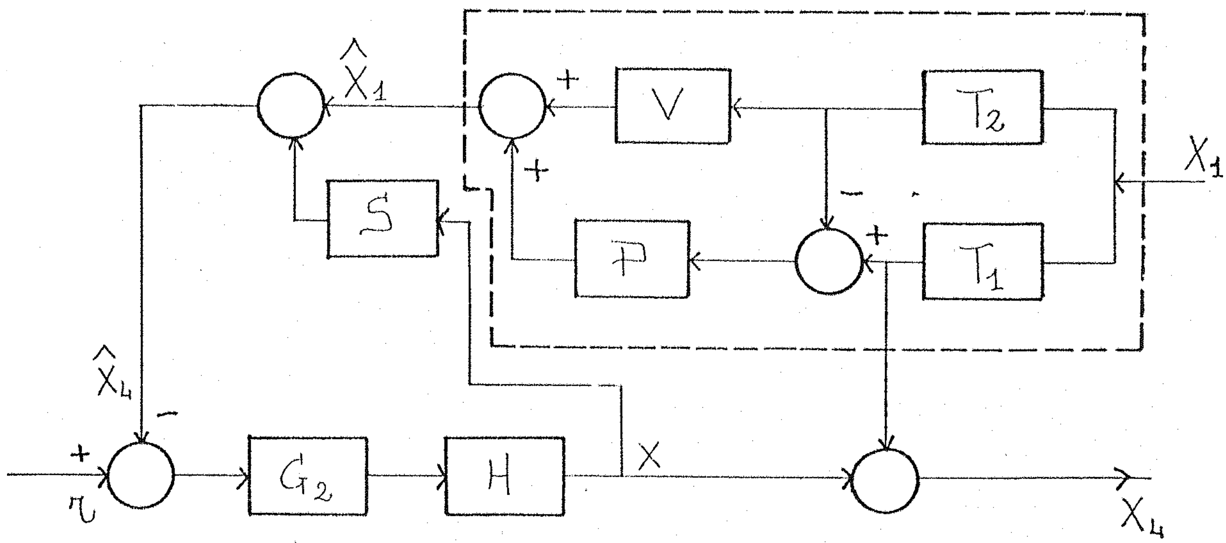


FIG. 9.10 BLOCK DIAGRAM - BLIND EXPERIMETS H

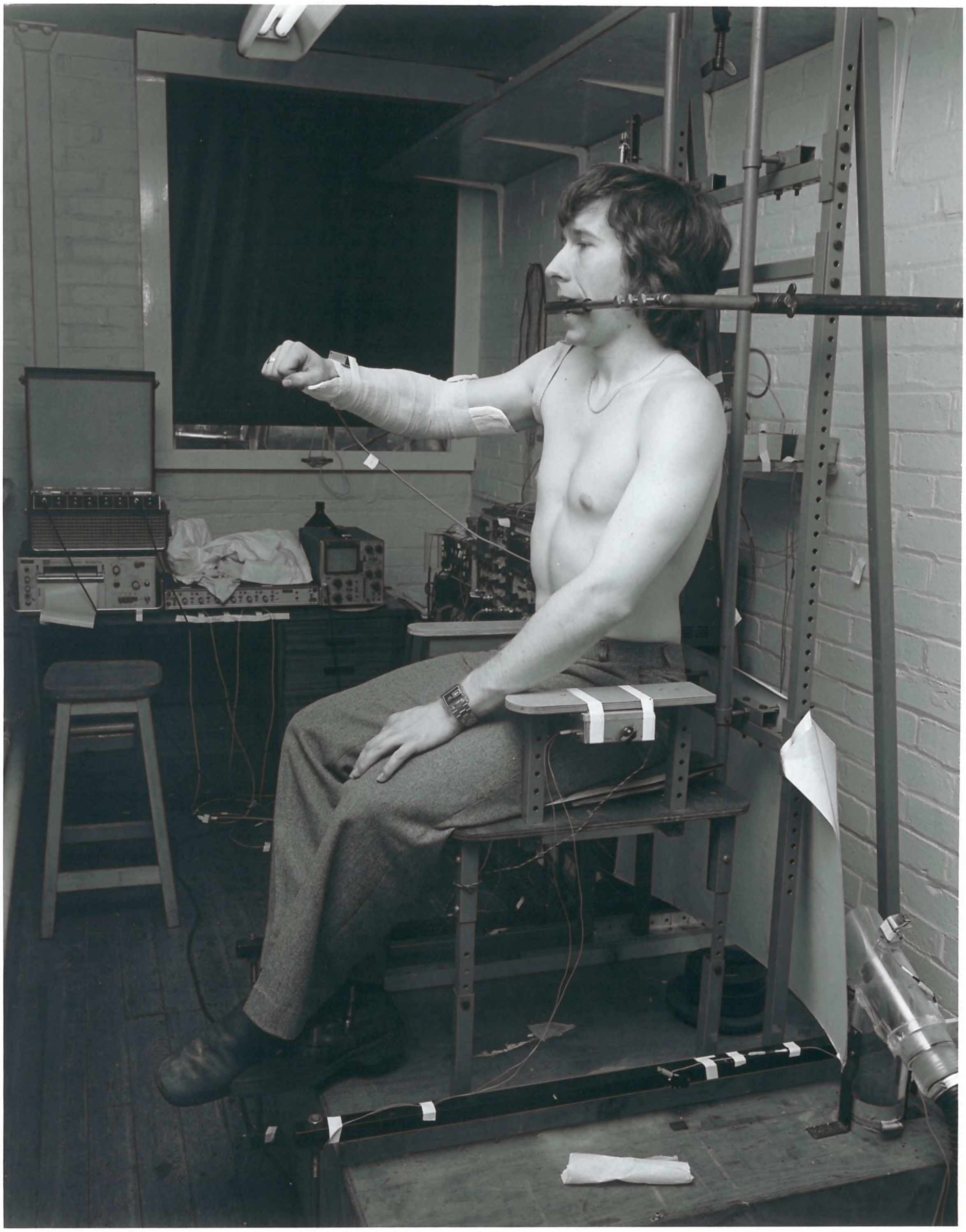


FIG. 9.11 SUBJECT UNDERTAKING BLIND EXPERIMENTS J

CHAPTER TEN

EXPERIMENTAL RESULTS

10.1. INTRODUCTION

In this chapter the results of the many experiments performed with the vibrating platform are described. Each experiment was performed with a number of subjects, with a variety of input waveforms and a number of techniques were used to analyse the results. There was extreme variety in the experimental results, not only between the performance of individual subjects but also from time to time with the same subject. The tabulated and graphical results will show that it is difficult to be precise both in determining the comportmental responses with which we have attempted to model the various feedback processes and in trying to validate the complete model in tests where all sensory systems are thought to act. It is undoubtedly true that our model should be non- linear and adaptive but, in the sense that a simple physical model is being sought, it is considered that the approach here is at least a beginning. An alternative way to interpret the results of each experiment is to use a comparative verbal method. Certainly such comments are included for completeness although they do not in themselves help to formulate a system model of the processes being studied.

10.2. ANALYSIS OF RESULTS

From transient inputs such as the swept sinusoid and random signals it is required to determine the frequency response. Three methods have been used:

(a) the FAST FOURIER TRANSFORM (FFT)

(b) the CROSS SPECTRAL DENSITY (CSD)

(c) CORRELATION TECHNIQUES

There are the usual problems of sampling frequency, window functions and record length encountered in these techniques. The programmes used are standard, see Appendix A10.1.

10.2.1. PRESENTATION OF RESULTS

With so many results and such wide fluctuations the question of presentation of results needs consideration. While one should ideally include all results there is a need to aim for conciseness. In this case it is difficult not to show only the most favourable results. In selecting a single basic set of results some attempt at standardisation has been made.

A pair of subjects were involved in a wide range of tests and were found to be fairly consistent in their performances. Their results are presented, generally as average. Statistical analysis of results is not attempted since the sample numbers involved are small, the observations are not stationary and the parameters being sought are only tentative. The results are presented in the sequence outlined in the experimental program of the previous chapter. A separate section is then devoted to determining the transfer functions which characterise the model. Finally some conclusions are drawn.

10.2.2. DATA ACQUISITION AND PROCESSING EQUIPMENT

The transient inputs to the platform were applied by means of the following signal generators.

(a) PS1 Model A104 Variable Phase Waveform Generator

(b) SOLARTRON, Model BO 1227 Random Signal Generator,

(c) PS1 Sweep Oscillator.

A block diagram of the data acquisition and processing system is shown in Figure 10-1 (a-b).

During an experimental run, the transducers outputs (position and acceleration signals) were recorded in two forms:- An eight-channel SE LAB T 3006/DV U-V Oscillograph was used to make a paper recording which afforded the opportunity for immediate visualisation of the progress of the observations. Parallel storage was afforded by a SE-LAB T-3000 four-channel, frequency - modulated magnetic tape recorder, writing at a speed of $1\frac{7}{8}$ inches/sec.

The paper recorder was used for gross visual editing and for selection of sequences of trials free from technical problems. The higher fidelity recordings made simultaneously on magnetic tape were located and played back in various suitable combinations.

After a recording session was completed, the data was reduced by digitising the recorded responses over a period of about 60 seconds starting from a prerecorded trigger signal which slightly preceeded the onset of the recorded platform movement. The signals were all digitised at a rate of 100 samples per second so that a maximum frequency component of 20 Hz could be very well reproduced. The behaviour of the subject between the digitised sections could be determined by visual inspection of the strip-chart records which were orginally used to monitor each recording session.

The strip-chart recordings were also used to spot-check the numerical averages obtained and to study the operation of the system as a function of time. All computations were performed on a digital Equipment Corporation PDP 11/50 Digital Computer, Appendix A7.3.

10.2.3. EXPERIMENTAL DATA

Different male subjects, (Table 7.1) with no previous experience of the tests were employed in the study.

To compensate for their lack of experience, the subjects were throughly briefed on their task and given ample time to familiarise themselves with the experiment.

Each group of data runs was preceded by a set of practice runs to give the subjects time to reach a stable level of performance and reduce the effects of learning during the data runs. Ample rest time was allowed between all runs to minimise the effects of fatigue.

10.3. OPEN LOOP EXPERIMENTS

(Descriptions of the tests are reported in Chapter 9 - Section 9.2).

10.3.1. TRACKING EXPERIMENT A

Only sinusoidal inputs were used for the experiment. The frequency range was 0.5 Hz to 1.8 Hz; the frequency interval 0.1 Hz; the duration time 60 seconds and the amplitude of displacement 4.5 cm.

Figure 10.2 shows the graphical response of the average subject.

FFT and correlation techniques were used to evaluate the frequency characteristics and both sets of results are shown on the same graph, Figure 10.2.

10.3.2. TRACKING EXPERIMENT B

The input signals and methods of analysis were exactly the same as in Tracking Experiment A. Caloric stimulation of the left ear was carried out as outlined in section 9.2.1. The results are shown in Figure 10.3.

10.3.3. TACTILE EXPERIMENT C

Two types of input were used:

(a) Swept sinusoid, initial frequency 0.1 Hz, final frequency 1.5 Hz, sweep time 25 seconds, amplitude 3.7. cm.

(b) Random 1st order Markov signal, cut-off frequency 1Hz and 3.2 Hz, rms amplitude 3.6 cm. at 1Hz and 4 cm. at 3.2. Hz. Duration 79 seconds.

Methods of analysis were:

(a) Swept sinusoids - FFT

(b) Random-CSD.

The results are shown respectively in Figure 10.4 (FFT), Figure 10.5. (Random - 1Hz), Figure 10.6. (Random, 3.2. Hz).

10.3.4. TACTILE EXPERIMENT D

The experimental conditions were the same as in experiment C with an additional number of sinusoidal tests. The neck support removed for this experiment.

The signal parameters were:

(a) Swept sinewave - as in 10.3.3.

(b) Random Input - as in 10.3.3.

(c) Sinusoidal Input - as in 10.3.1.

Methods of analysis were respectively:

(a) FFT

(b) CSD

(c) FFT and correlation

The swept sinusoid tests were performed with increasing and decreasing frequencies. The results are shown in Figure 10.7 (Sweep sine input), 10.8 (Random input, 1 Hz) 10.9 (Random input, 3.2. Hz), 10.10 (Sine wave input).

10.3.5. TACTILE EXPERIMENT E

Inputs were:

(a) Sinusoid - as in 10.3.1.

(b) Swept sinusoid - as in 10.3.3.

(c) Random - as in 10.3.3.

Methods of analysis were as in 10.3.4. The caloric stimulation was applied in alternate ears and some variation was found. The results are shown in Figure s 10.11. (sine wave input), 10.12 (sweep sine input), 10.13 (random input, 3.2. Hz), 10.14 (random input, 1Hz).

10.3.6. TACTILE EXPERIMENT F

Here Experement D was repeated with the subject inclining his head at 50° to the left.

The results are shown in Figure 10.15. (swept sine input).

10.4. CLOSED LOOP EXPERIMENTS

(Descriptions of the tests are reported in Chapter 9. Section 9.3.).

10.4.1. VISUAL TEST G

The subject was seated with his arm immobilised in an elbow and wrist acrylic splint. Accelerometers were mounted on the shoulder and hand to record vertical accelerations. A reference line was drawn on a board in front of the subject and his objective was to align his hand with the reference during platform motion. To eliminate the strong predictive element in the visual response only random inputs were used.

The parameters were:

(a) Cut-off frequency 3.2. Hz, rms amplitude 4 cm - Duration time 60 seconds.

(b) Cut off frequency 1 Hz, rms amplitude 3.6 cm - Duration time 60 seconds.

Method of analysis : CSD. The results are shown in Figure 10.16. (randon - 1Hz), 10.17 (random 3.2 Hz).

10.4.2. BLIND EXPERIMENT H

The subject was allowed to align his hand visually and then his vision was obscured for the duration of the experiment, i.e. 60 seconds. The objective was to keep his hand aligned with the reference mark. Only random inputs were used as in section 10.4.1. The results are shown in Figure 10.18 (random 1Hz) 10.19 (random, 3.2 Hz).

10.4.3. BLIND EXPERIMENT J

A series of experiments were performed, basically the same as experiment H but with disruptive stimulation applied to neck muscles and stimulation of the anterior deltoid shoulder muscle.

In both experiments the subject was required to bite on a fixed horizontal bar, Figure 9.11 in order to constrain vertical head motion and thus limit vestibular signals. These result are shown in Figure 10.20 (vibration of the neck muscles) and 10.21 (vibration of the deltoid).

10.5. OBSERVATIONS

In the following section a number of important observations are made from the results of the experiments. No attempt to quantify these is yet made.

10.5.1. EXPERIMENTS A AND B

(a) There is little difference between the set of responses, from which we conclude that the visual feedback mechanism dominates the other feedbacks.

(b) The human operator response has a leading phase characteristic for sinusoids over the range of frequencies in the experiments. This is confirmed by similar tracking experiments.

(c) At higher frequencies the phase becomes increasingly leading.

(d) The results from each method of analysis are similar.

10.5.2. EXPERIMENTS A,B,C,D,E

(a) When the visual feedback is not available the response indicates that other mechanisms provide compensation since proprioception has been limited in EXP. C, we conclude that vestibular feedback provides this compensation.

(b) When the neck collar is removed, the differences are slight at low frequencies, but increase at higher frequencies. Better control is obtained with the collar. It is inferred then that head movement gives a better indication of the upper arm (shoulder) displacement, and so vestibular feedback is more effective.

(c) The body transmissibility tests (Chapter 7) show little measurable relative displacement of the head and shoulder in the range of frequencies 0-2.5 Hz. This implies that the neck proprioceptors are problably force rather than displacement sensors.

(d) The overall human operator response declines when the platform displacements are random rather than sinusoidal. This is presumably because the predictive element in his response has inadequate information on the future state of the input. The response now becomes lagging rather than leading.

(e) With sinusoidal and swept inputs the difference in response with and without caloric ear stimulation is pronounced indicating that the vestibular feedback can function effectively in the conditions of experiment D. For random inputs the difference is much less pronounced showing that the vestibular apparatus does not have enough information to be so effective as in experiment D. In other words the predictability in control is lost.

(f) Results for caloric stimulation of left and right ears show some differences which suggests that the vestibular apparatus is not balanced with respect to vertical motion.

10.5.3. EXPERIMENT F

When the head is inclined, putting the neck muscles in tension, the phase angle was reduced to almost zero and very accurate control was achieved. It is suggested that proprioceptive feedback from the neck receptors becomes more effective in these conditions.

10.5.4. EXPERIMENTS G,H,J

(a) When available visual feedback appears to dominate the feedback mechanisms.

(b) The response without vision deteriorates with increasing time, showing that the central processor has a short term spatial reference only.

(c) With the bite bar, control is worse but it is not excessively poor. But when stimulation interferes with proprioceptive feedback the response is markedly worse.

10.6. CALCULATIONS AND COMMENTS

10.6.1. OPEN LOOP EXPERIMENTS

As previously explained these calculations represent an attempt to generalise the results in the form of a model. The particular responses are derived from the experiments as follows:

R_1 - Exp. A

R_2 - Exp. C

R_3 - Ext D

R_4 - Exp E.

The following model characteristics are calculated from the average curve in Figures 10.2 (R_1); 10.4(R_2); 10.7 (R_3); 10.11 (R_4).

Vestibular Sensitivity from equation 9.4.

$$V(j\omega) = \frac{R_2(j\omega)}{R_1(j\omega)\;T_2(j\omega)} \simeq \frac{R_2(j\omega)}{R_1(j\omega)} \tag{10.1}$$

since $T_2(j\omega) \simeq 1$ in the frequency range 0-2.5 Hz.

Proprioceptor Sensitivity from equation 9.11.

$$P(j\omega) = \frac{R_3(j\omega) - R_2(j\omega)}{R_1(j\omega)} \tag{10.2}$$

Proprioceptor Sensitivity from equation 9.14

$$P'(j\omega) = \frac{R_4(j\omega)}{R_1(j\omega)} \tag{10.3}$$

where $P'(j\omega) = P(j\omega)\,[(T_1(j\omega) - T_2(j\omega)]$

The results are plotted as function of frequencies in Figures 10.22 (V) and 10.23 (P and P') In the latter figure the two estimates for the proprioceptor sensitivity clearly give some justification for the assumed model, particularly in their phase values.

Table 10.1 shows the numerical values of the results.

The calculated sensor characteristics clearly have considerable fluctuations which are exagerated by the fluctuations in the measured responses. However, the following general comments may be made:

(a) The vestibular sensitivity appears much greater than the proprioceptor sensitivity for the experiments performed.

(b) Both responses show leading phase characteristics, the proprioceptors giving the greater leading angles.

(c) The same calculations for sinusoidal inputs produce even greater phase lead angles. This suggests that there is compensation within the central processor and that this is adaptive. One of the great difficulties in performing this experiment was the removal of other unwanted feedback, especially audible signals associated with the platform drive. It was considered that attempts to validate our model under these conditions were probably not justified.

(d) The variations of phase and gain characteristics for both sensors groups do not appear consistent with each other in terms of normal minimum phase compensating networks. It does not appear to be worthwhile fitting numerical transfer functions to the data while the experimental results are so widely fluctuating.

Table 10.2 reports the calculated numerical values for R_3 and R_4, figures 10.24 and 10.25 show the comparison between the measured and the calculated values.

10.6.2. CLOSED LOOP EXPERIMENTS

As previously stated the results for the feedback experiment G, H and J are extremely fluctuating. In particular it was difficult to record phase information. However apart from the obvious deterioration in control caused by loss of vision and proprioception these experiments can be used for a crude verification of the feedback model chosen.

Experiment G gives a measure of the ultimate performance of the shoulder muscle control system, since all sensory systems are active. It can be seen, Figures 10.16 and 10.17 that the performance deteriorates at a frequency of approximately 2 Hz. This may be taken as a measure of the closed-loop bandwidth of the system.

In experiment H the visual loop is suppressed and from equation 9.25, it follows that the measured response to a sinusoidal platform displacement can be written as:

$$\frac{X_4(j\omega)}{X_1(j\omega)} = \frac{G_2(j\omega)\ H(j\omega)}{1 + G_2(j\omega)\ H(j\omega)\ S(j\omega)} \left| T_2(j\omega)\ [P(j\omega) - V(j\omega)] + T_1(s) \left\{ S(j\omega) - P(j\omega) + \frac{1}{G_2(j\omega)\ H(j\omega)} \right\} \right| \qquad (10.4.)$$

Since from the transmissibility tests (Chapter 7).

$T_1(j\omega) \simeq T_2(j\omega) \simeq 1$, for frequencies in the range 0-2.5 Hz, equation (10.4) may be written as:

$$R_6(j\omega) = \frac{X_4(j\omega)}{X_1(j\omega)} = \frac{G_2(j\omega)\ H(j\omega)}{1 + G_2(j\omega)\ H(j\omega)\ S(j\omega)} \left[S(j\omega) - V(j\omega) + \frac{1}{G_2(j\omega)\ H(j\omega)} \right] \qquad (10.5.)$$

If we call

$$\frac{G_2(j\omega) H(j\omega)}{1 + G_2(j\omega)\ H(j\omega)\ S(j\omega)} = H_c(j\omega) \qquad (10.6.)$$

the closed-loop frequency response when only the vestibular and proprioceptive feedback are active,

then equation 10.5. becomes

$$R_6 = H_c(j\omega)\ [S(j\omega) - V(j\omega)] + \frac{1}{1 + G_2(j\omega) H(j\omega) S(j\omega)} \qquad (10.7.)$$

R_6 will only have a similar frequency response to R5, see 9.19, if

$$T_1 = S = V = 1$$

and $$G_1 = G_2$$

It is unlikely that this will be true since there seems to be an uncertain weighting to the vestibular data. Thus R_6 will be in error, particularly at low frequencies. Assuming however that the bandwidth of Hc is the same as that of R_5 then the deterioration in performance should be roughly the same with and without vision beyond 2 Hz, (Figure 10.16 - 10.19). Clearly from Figure 10.20 partial inhibition of the proprioceptors further deteriorates the performance.

N.B. R_5 and R_6 represent the ability to cancel out the unwanted movement caused by the seat displacement.

CHAPTER 10

LIST OF TABLES

TABLE 10.1

RESULTS

f(Hz)	0.5	0.6	0.7	0.8	0.9	1	1.2	1.4
$\|R_1\|$	.39	.42	.43	.43	.37	.45	.55	.58
$\|R_2\|$	1.8	1.7	1.3	1.0	1	1.4	1.9	2
$\|R_3\|$	1.3	1.3	1.3	1.2	1.2	1.2	1.2	1.5
$\|R_4\|$	.33	.31	.34	.36	.26	.35	.39	.40
$\|V\|$	4.61	2.38	3.02	2.32	3.51	3.11	1.45	3.45
$\|P\|$	1.01	1.22	.12	.08	.13	.37	.33	1.16
$\|P'\|$	.13	.13	.15	.16	.10	.16	.21	.23
$<R_1$	31	27	35	43	44	54	64	49
$<R_2$	-80	-24	-24	-16	-24	-40	-48	-80
$<R_3$	-20	-8	0	-16	-8	0	-88	-148
$<R_4$	22	37	48	70	83	100	86	20
$<V$	-49	3	11	27	20	14	14	-31
$<P$	83	57	113	100	134	136	124	85
$<P'$	53	64	83	113	108	154	150	69

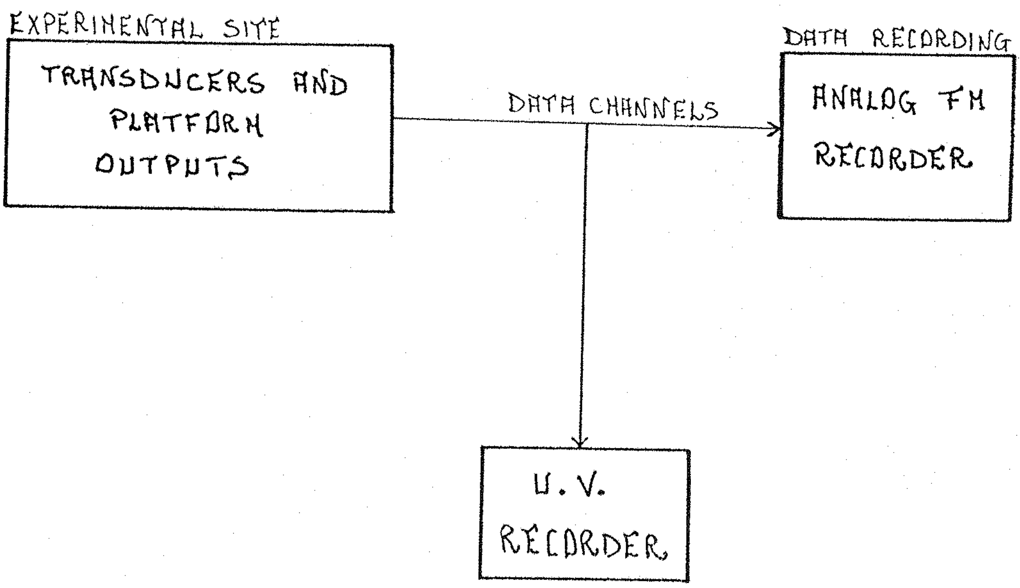
TABLE 10.2

ESTIMATED VALUES

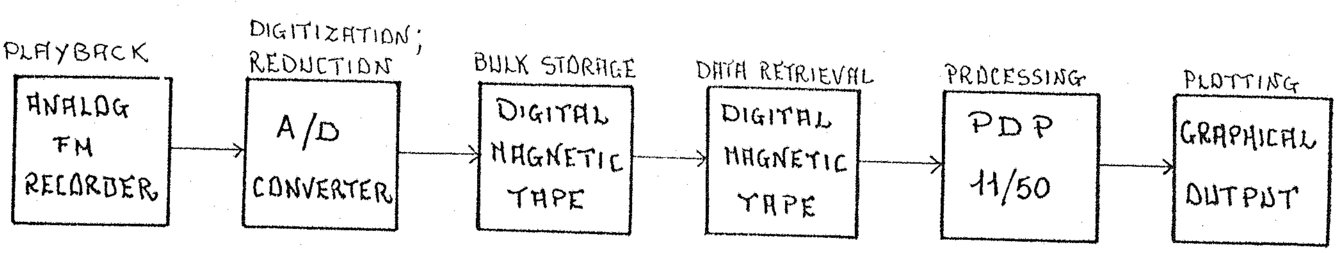
f	0.5	0.6	0.7	0.8	0.9	1	1.2	1.4
$\|R_3\|$	1.76	1.17	1.44	1.09	1.33	1.15	0.6	1.97
$<R_3$	-69.5	-10.4	-11	3.3	-12.7	-28.8	-20	-68.5
$\|R_4\|$	1.61	0.91	0.54	0.2	0.36	0.91	.78	2
$<R_4$	55.6	73	78	87	89.8	81.9	50.8	-36

CHAPTER 10

LIST OF FIGURES



a) DATA ACQUISITION



b) DATA PROCESSING

FIG. 10.1 BLOCK DIAGRAM OF THE DATA HANDLING SYSTEM

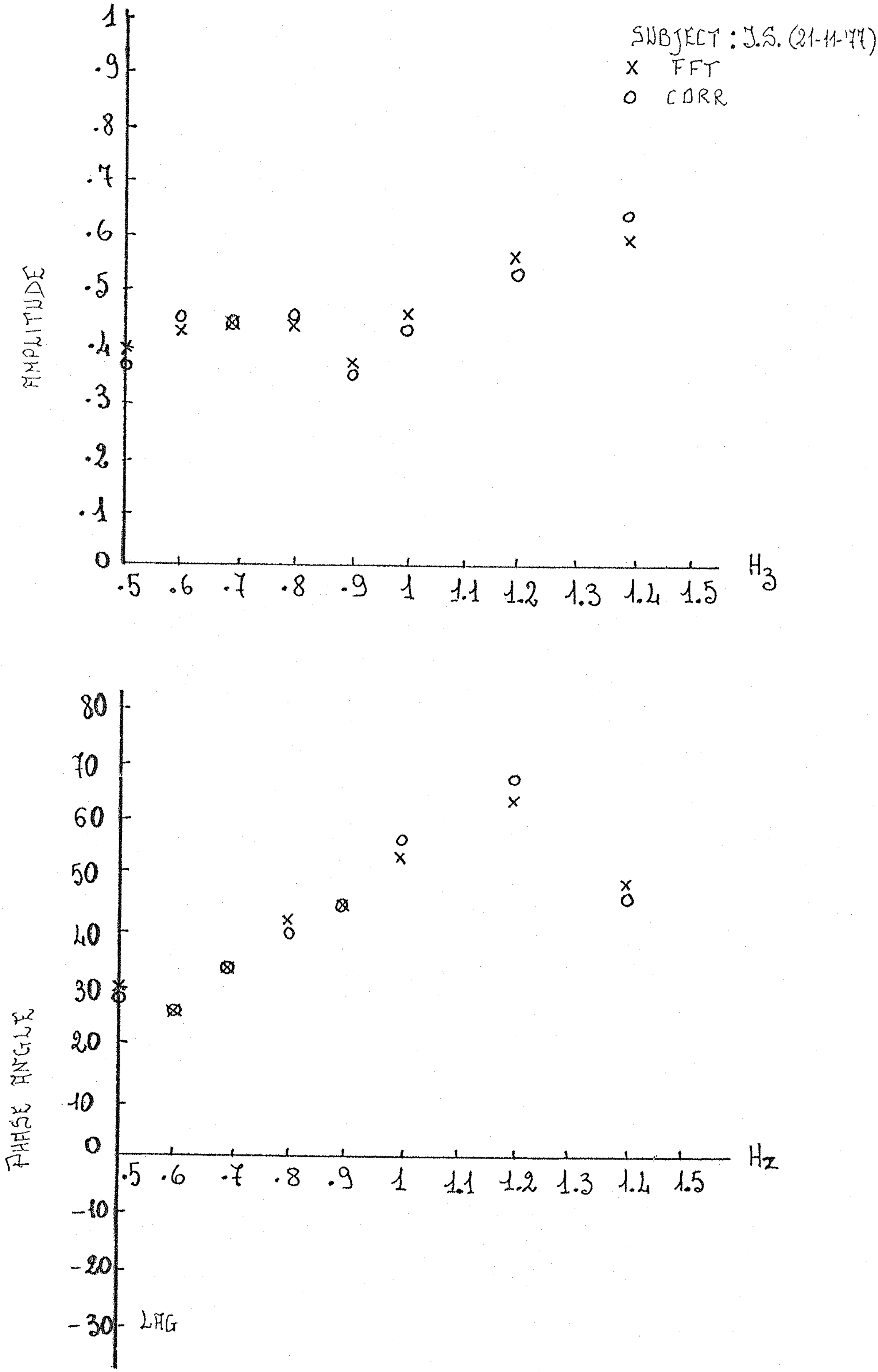


FIG. 10.2 TRACKING EXPERIMENT A (Sine wave input)

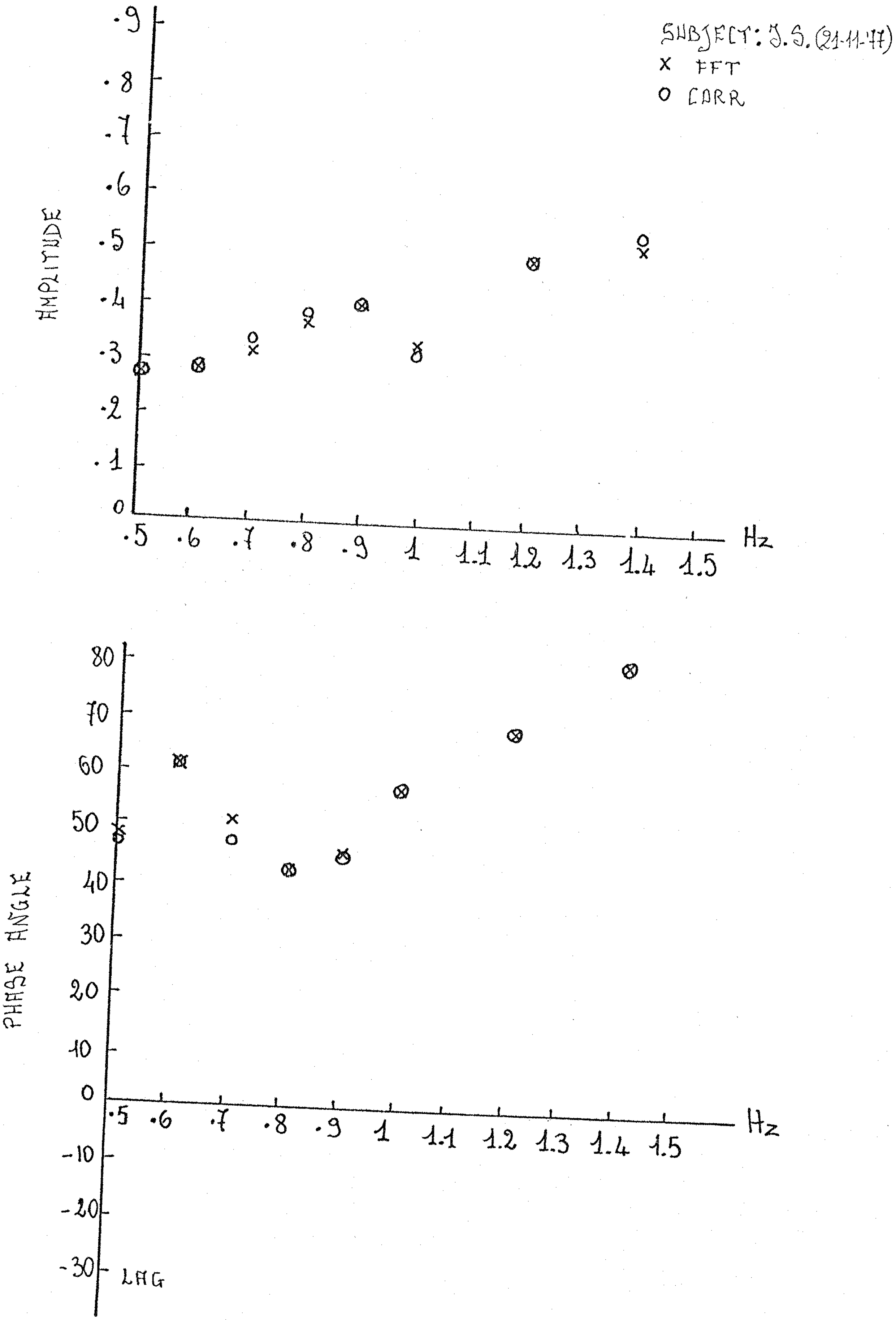


FIG. 10.3 TRACKING EXPERIMENT B (Sine wave input)

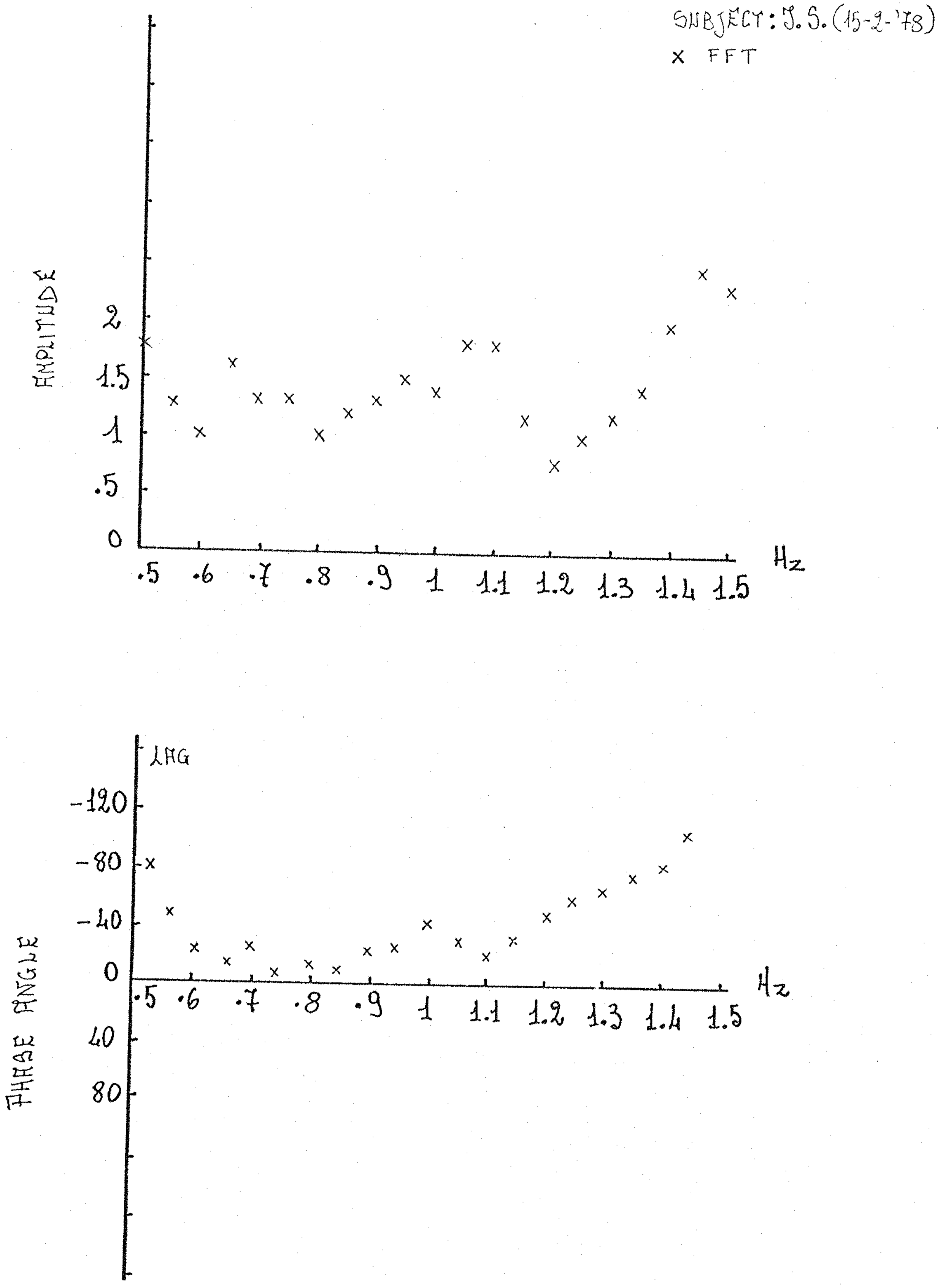


FIG. 10.4 TACTILE EXPERIMENT C (Sine swept input)

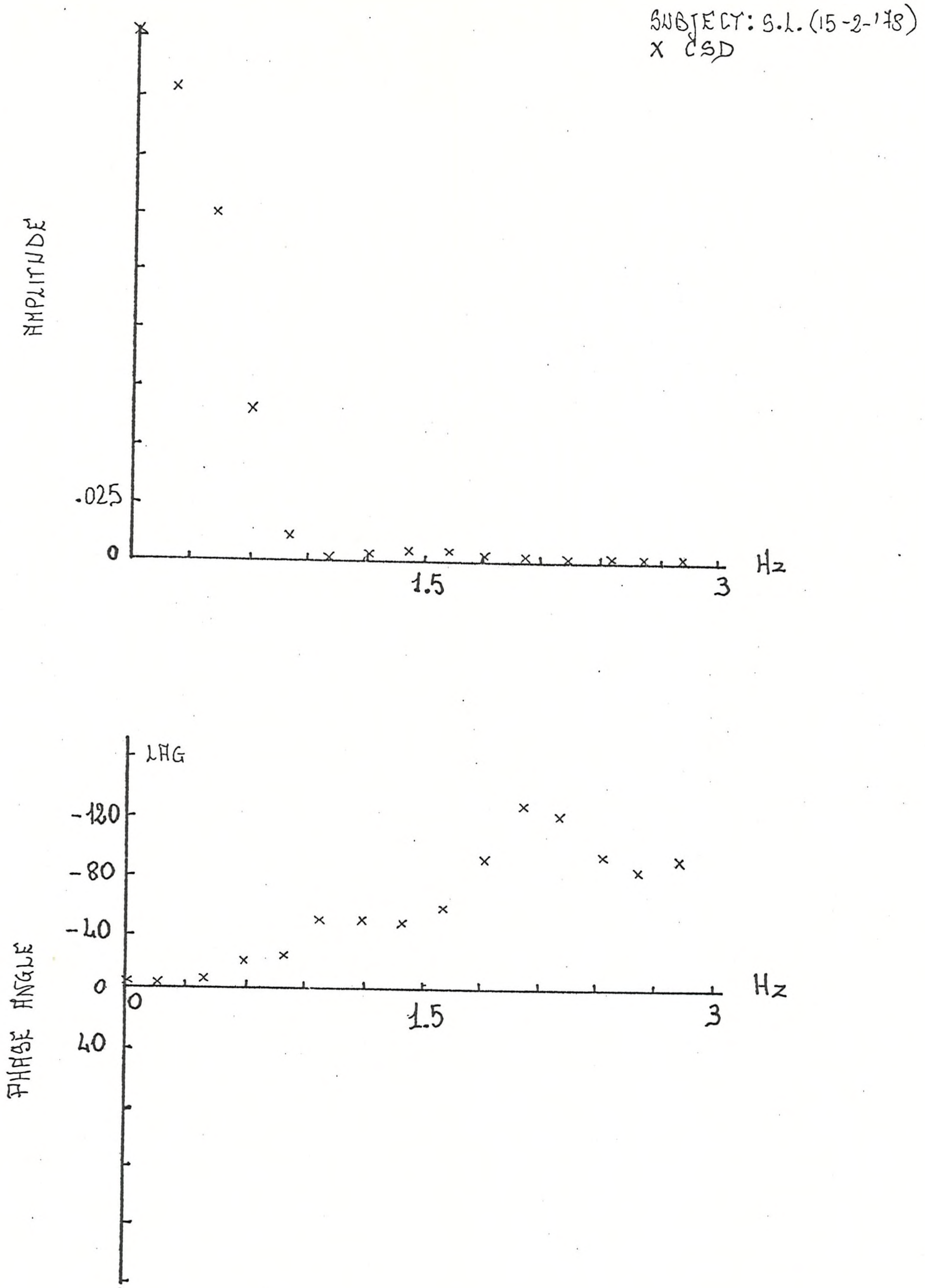


FIG. 10.5 TACTILE EXPERIMENT C (Random input)

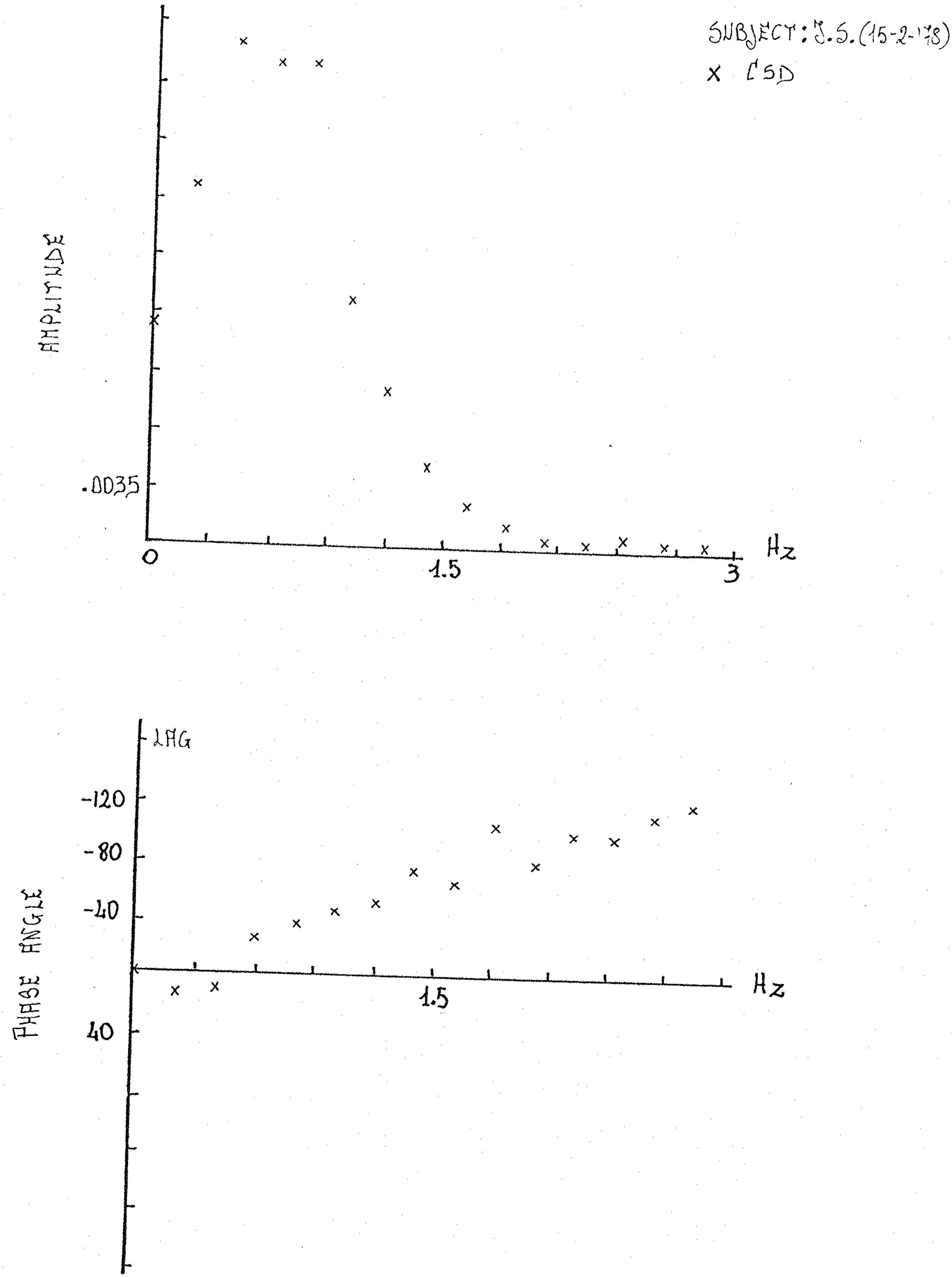


FIG. 10.6 TACTILE EXPERIMENT C (Random input)

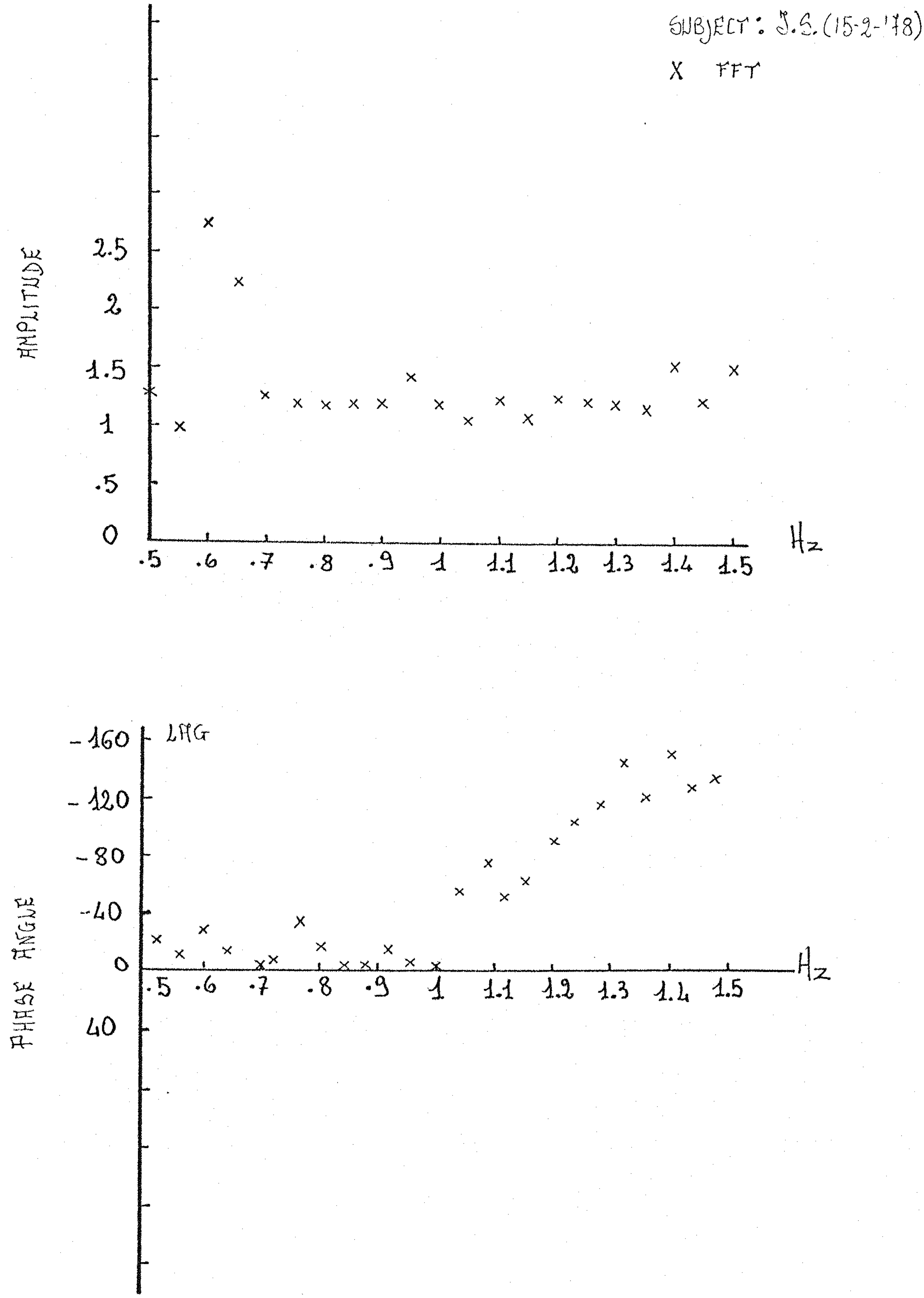


FIG. 10.7 TACTILE EXPERIMENT D (Swept sine input)

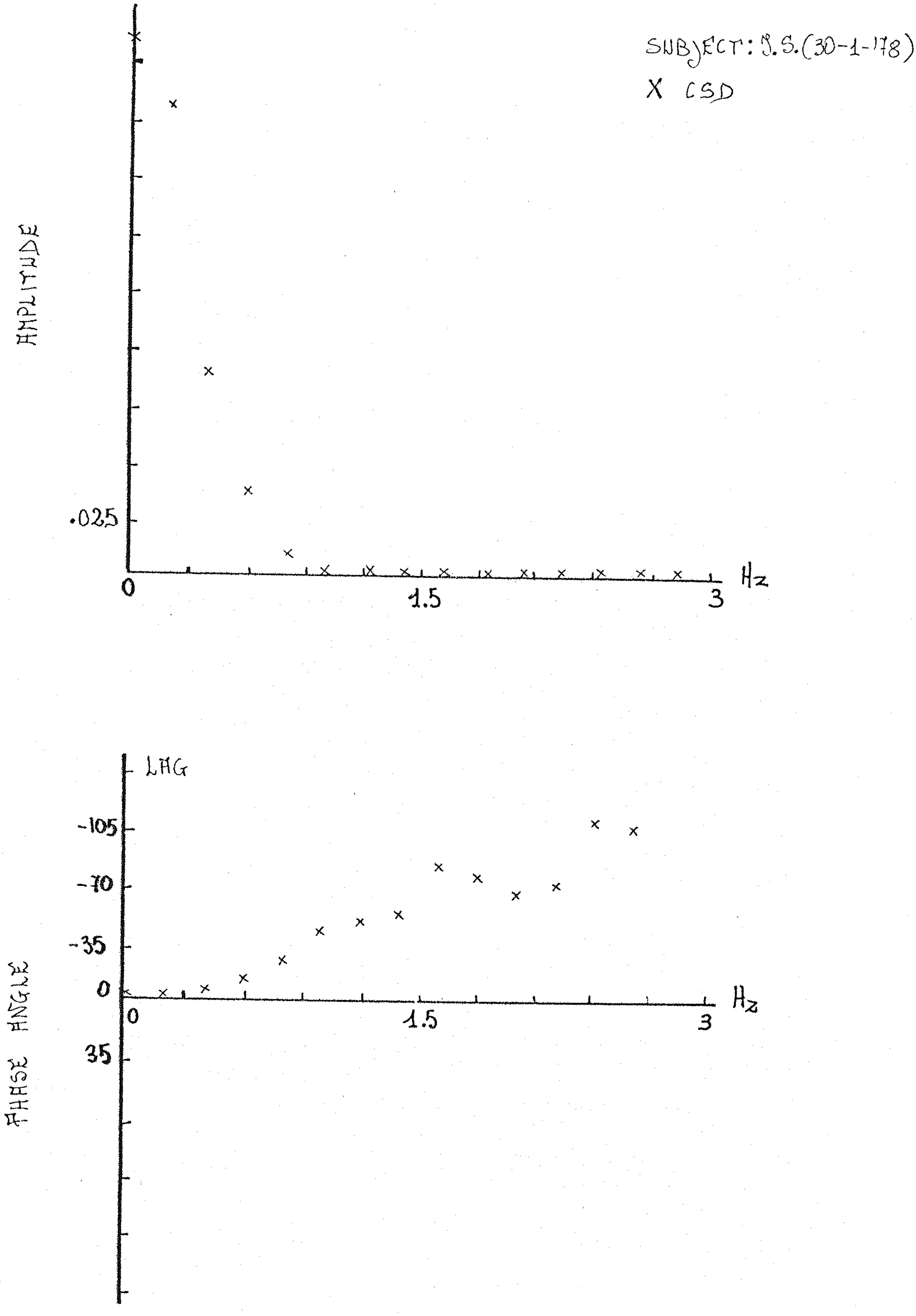


FIG. 10.8 TACTILE EXPERIMENT D (Random input)

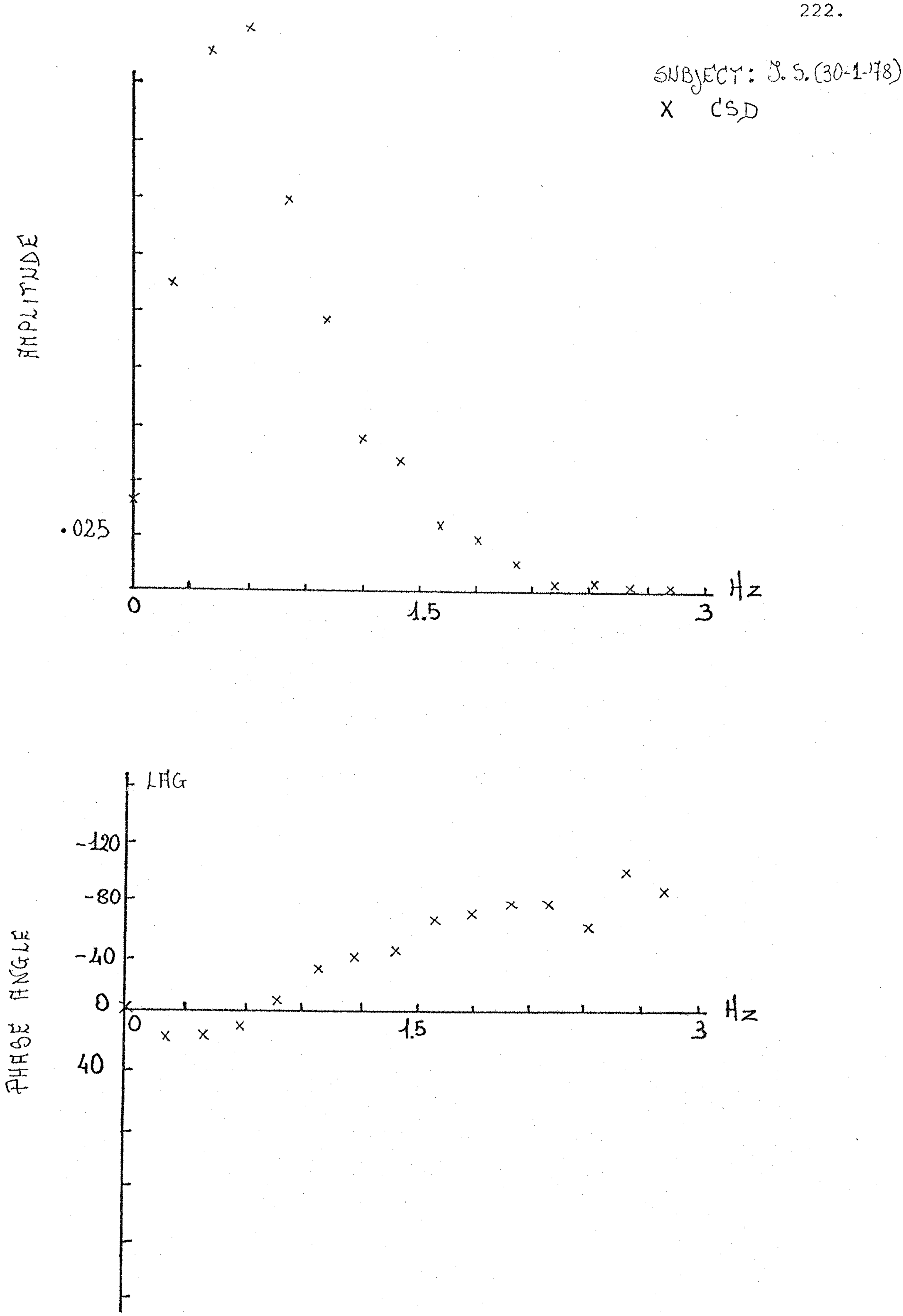


FIG. 10.9 TACTILE EXPERIMENT D (Random input)

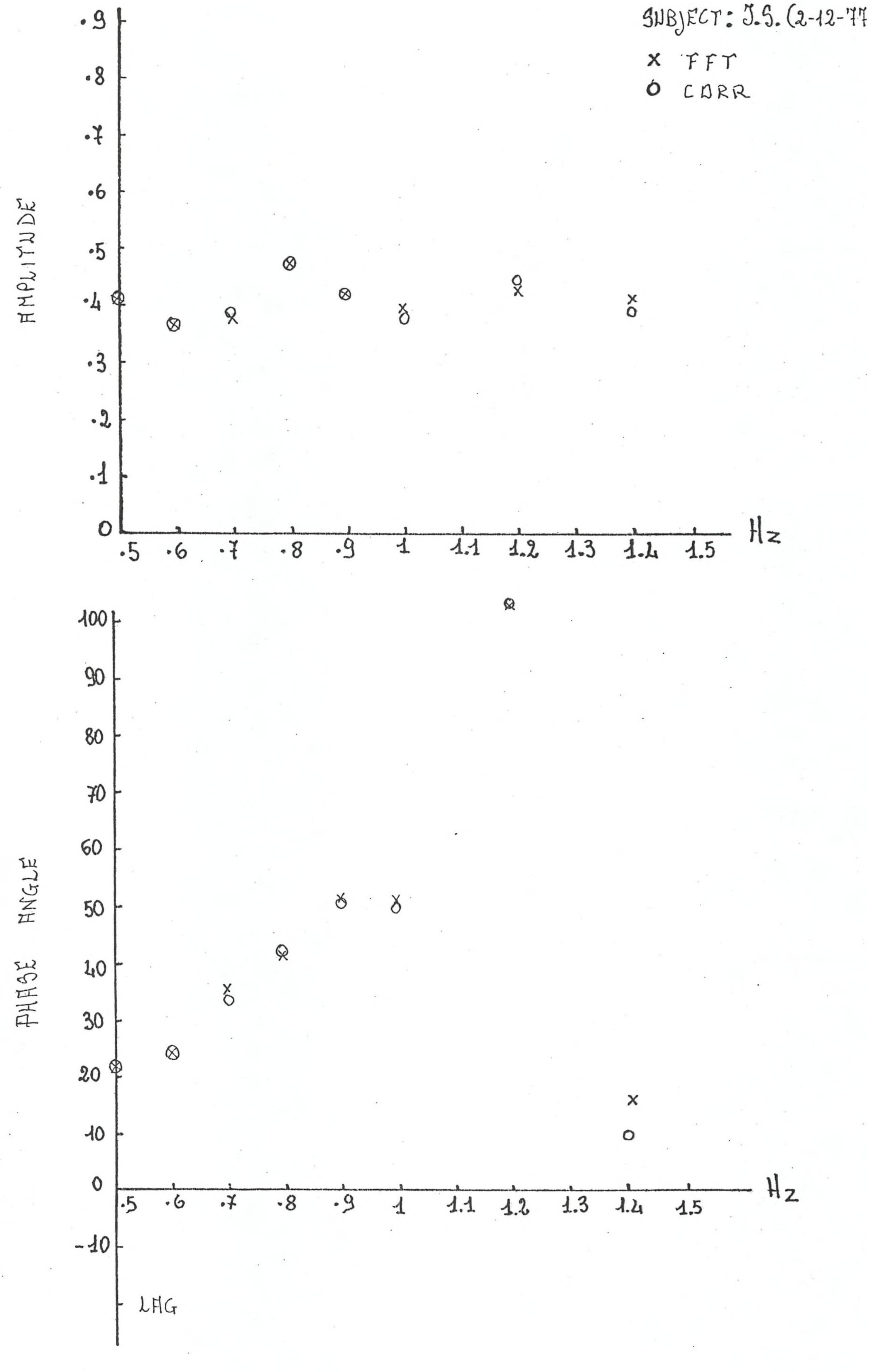


FIG. 10.10 TACTILE EXPERIMENT D (Sine wave input)

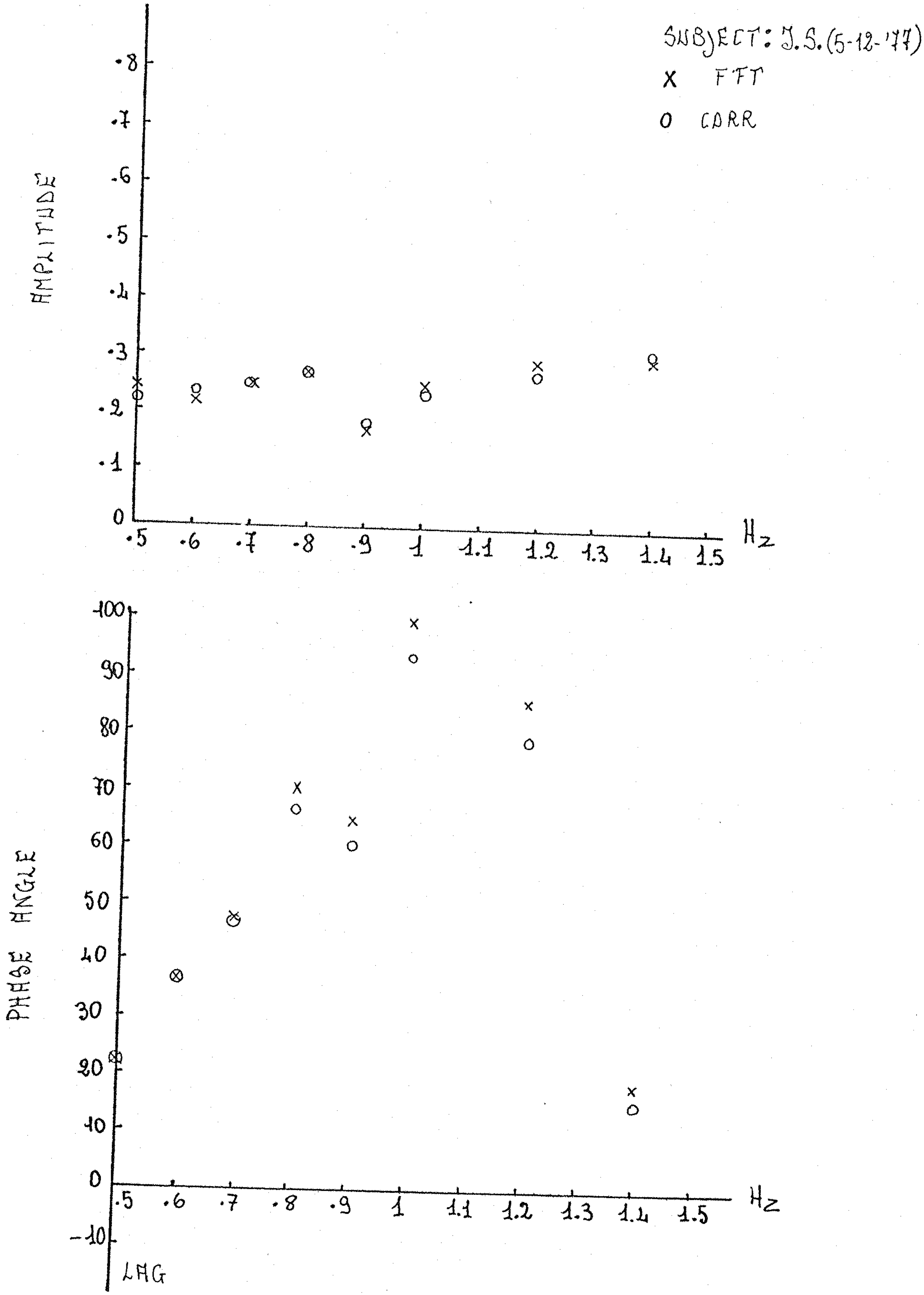


FIG. 10.11 TACTILE EXPERIMENT E (Sine wave input)

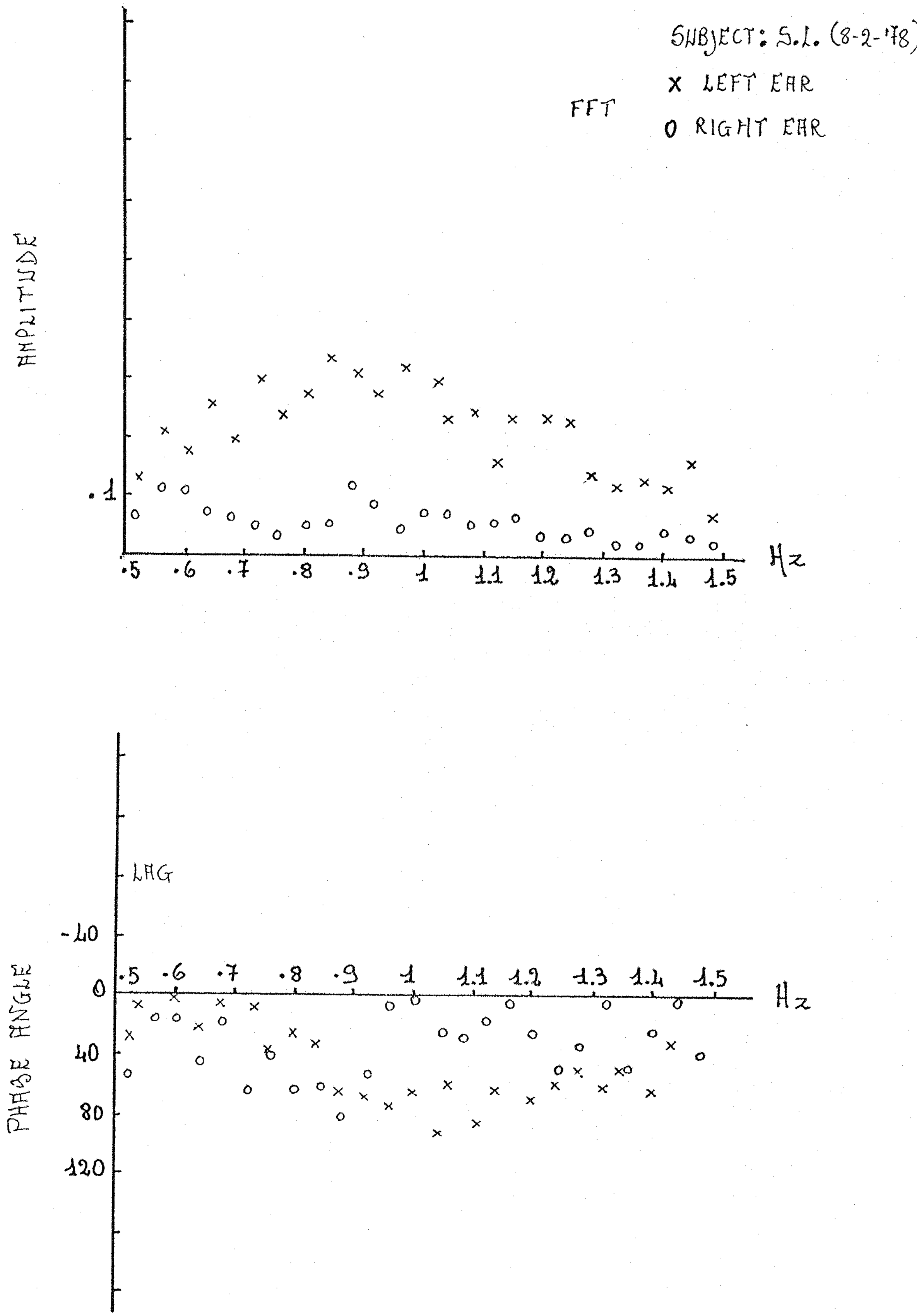
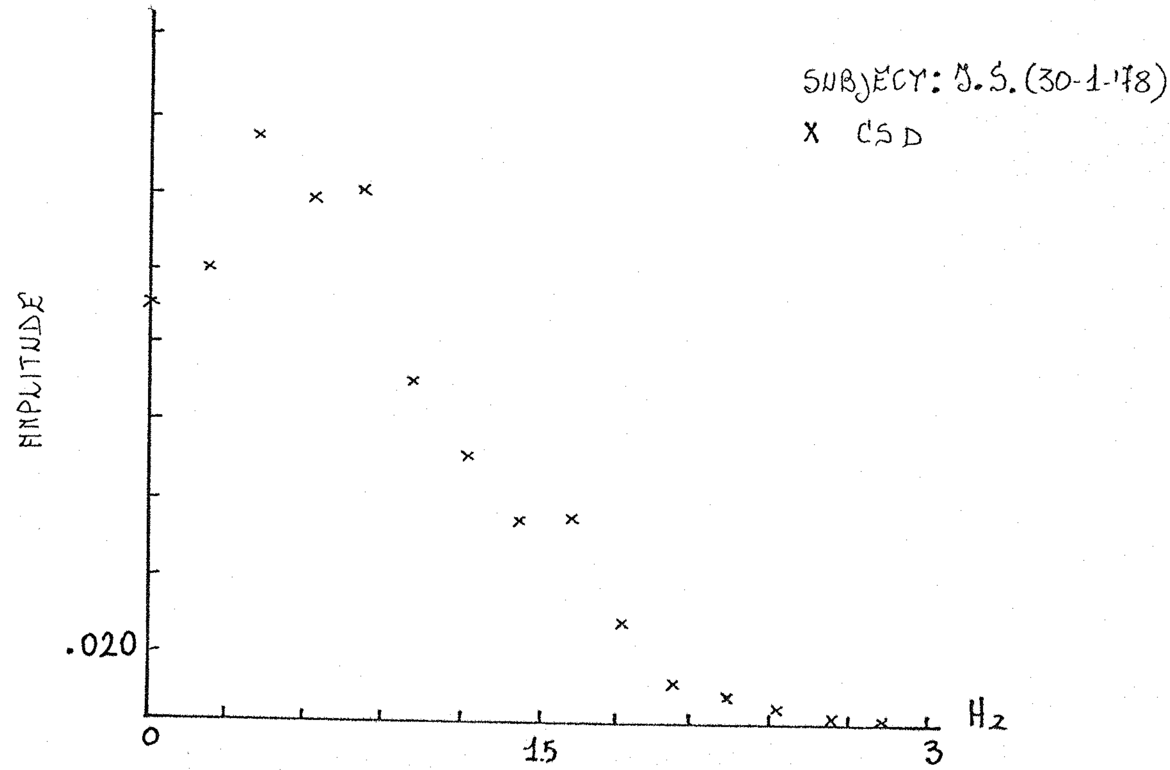


FIG. 10.12 TACTILE EXPERIMENT E (Swept sine input)



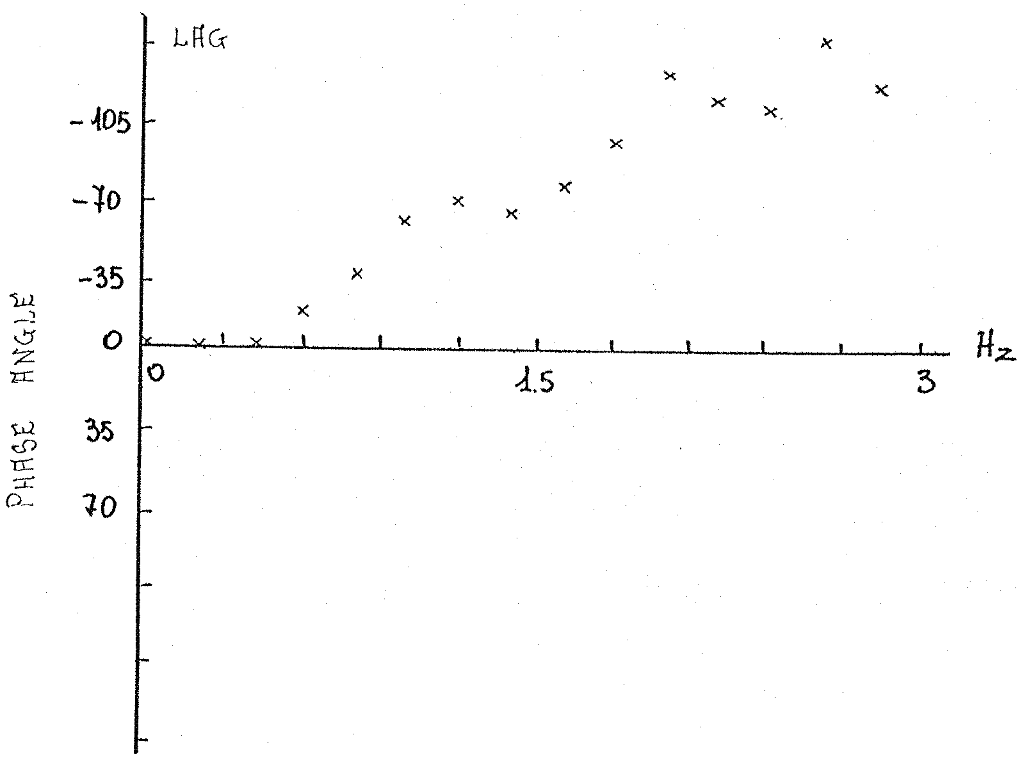


FIG. 10.13 TACTILE EXPERIMENT E (Random input)

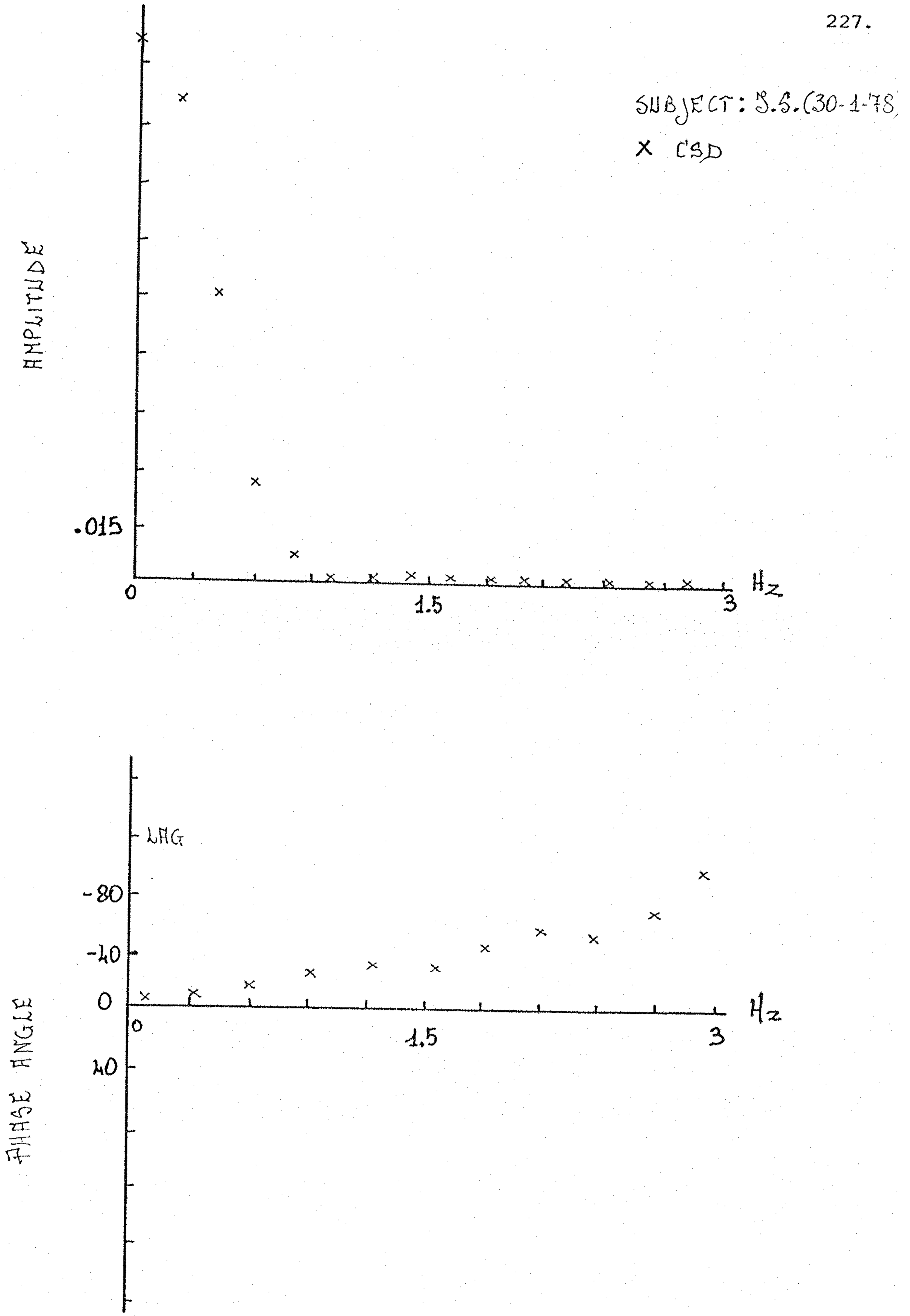
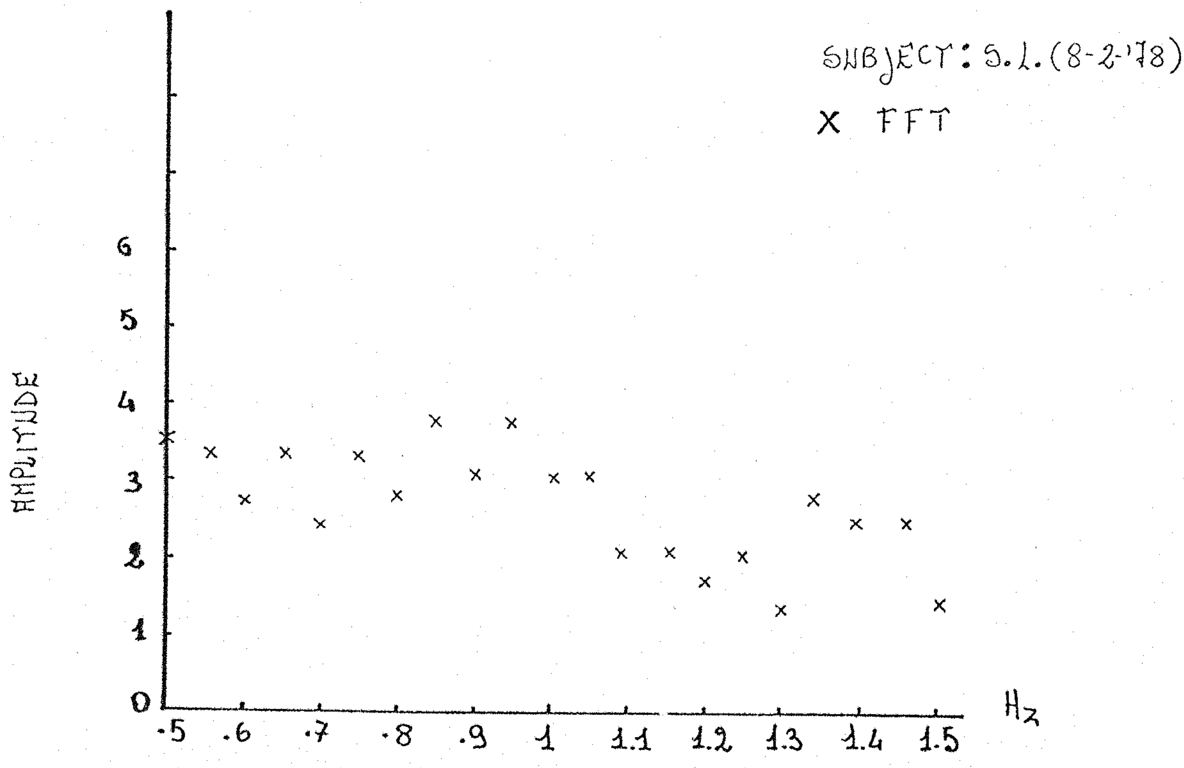


FIG. 10.14 TACTILE EXPERIMENT E (Random input)



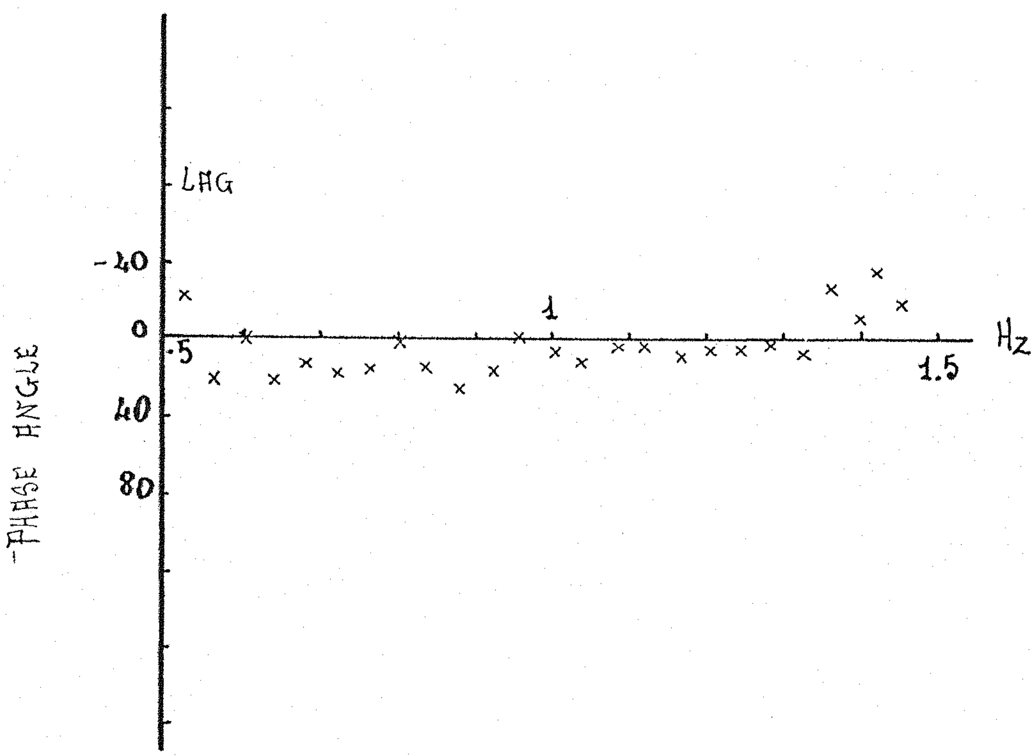


FIG. 10.15 TACTILE EXPERIMENT F (Swept sine input)

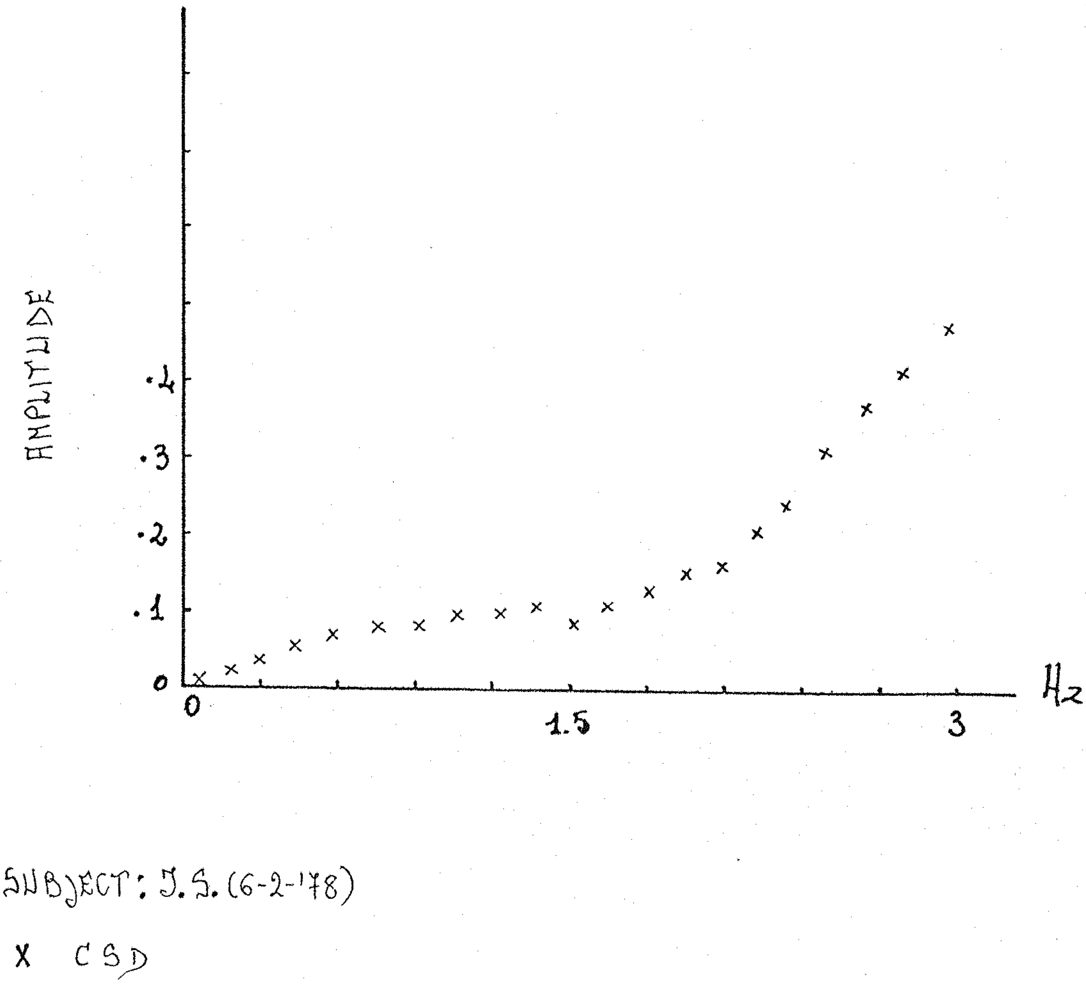


FIG. 10.16 VISUAL EXPERIMENT G (Random input)

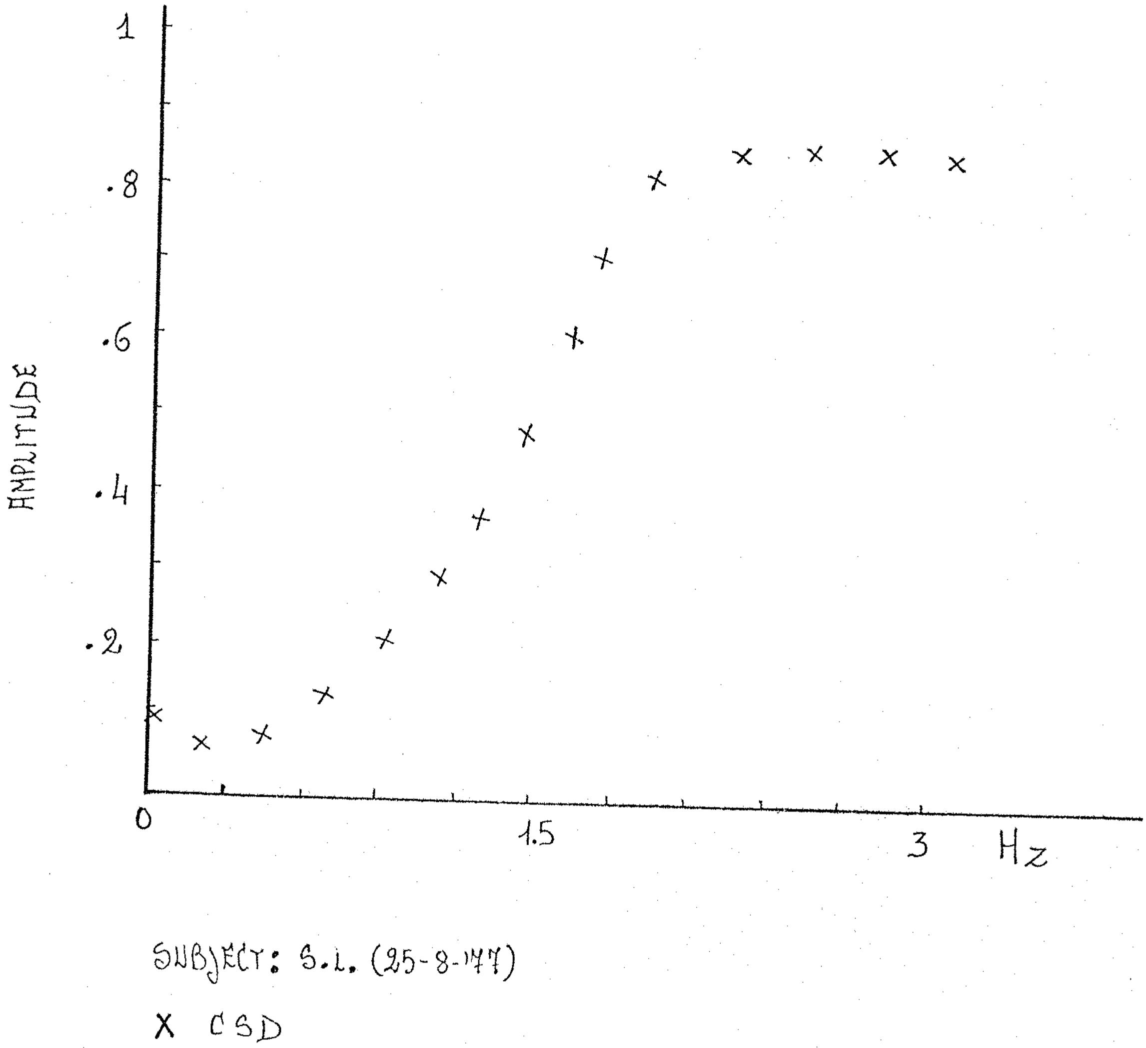
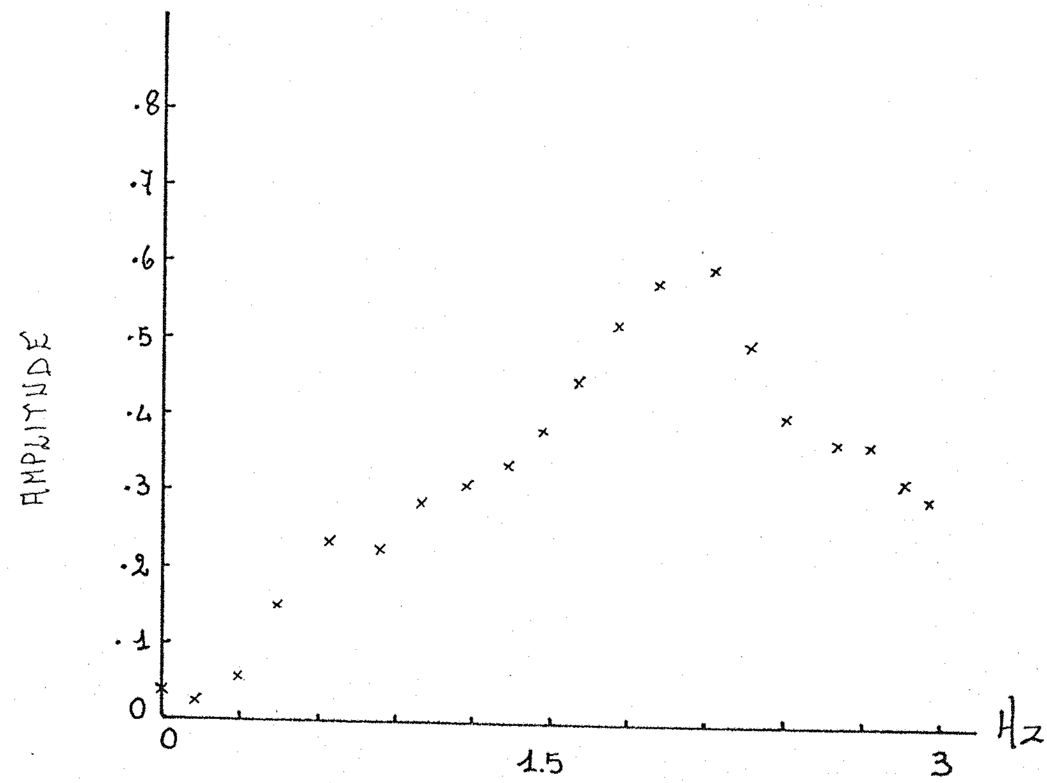


FIG. 10.17 VISUAL EXPERIMENT G (Random input)



SUBJECT: J.S. (9-10-77)
X CSD

FIG. 10.18 BLIND EXPERIMENT H (Random input)

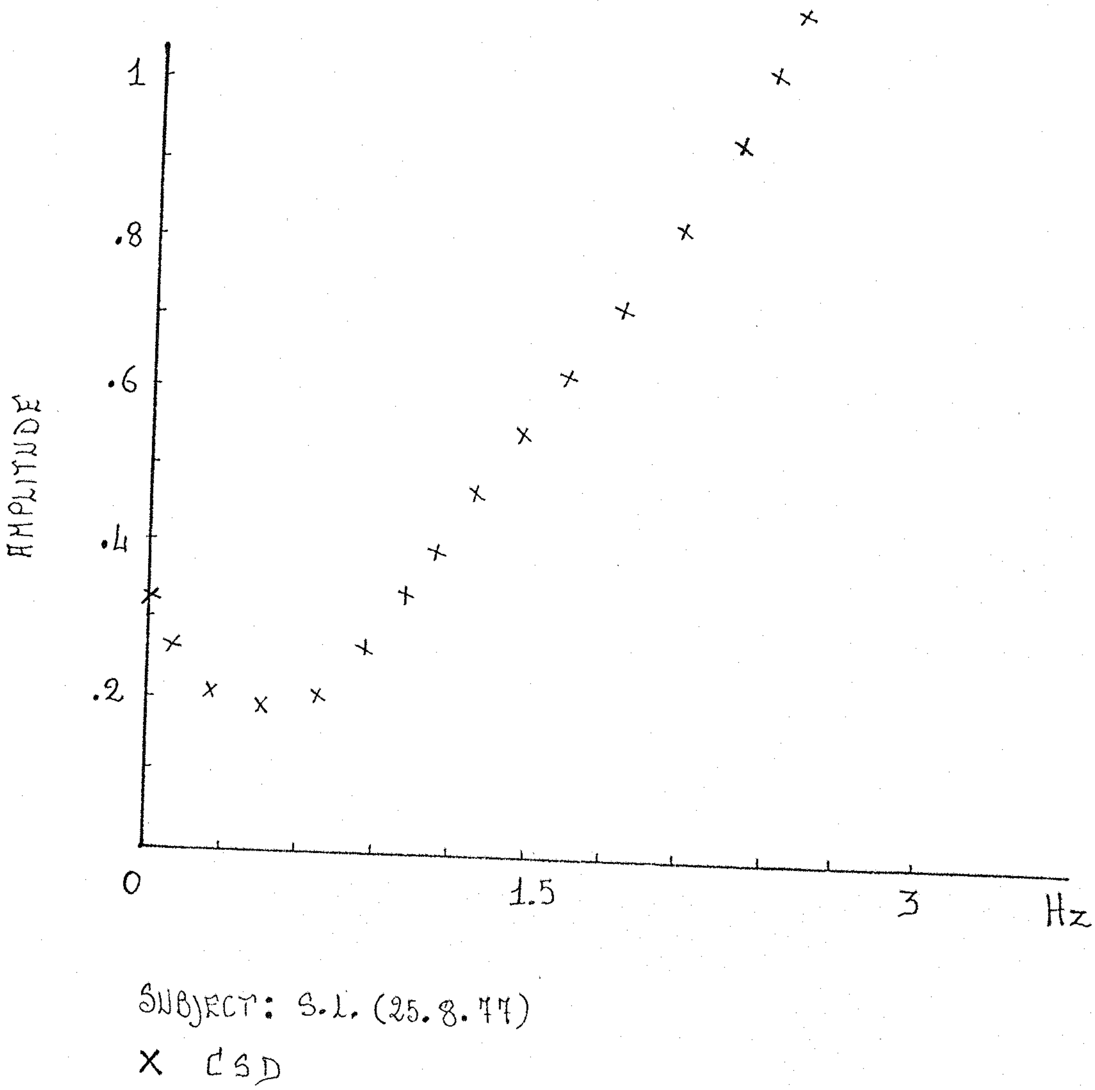


FIG. 10.19 VISUAL EXPERIMENT H (Random input)

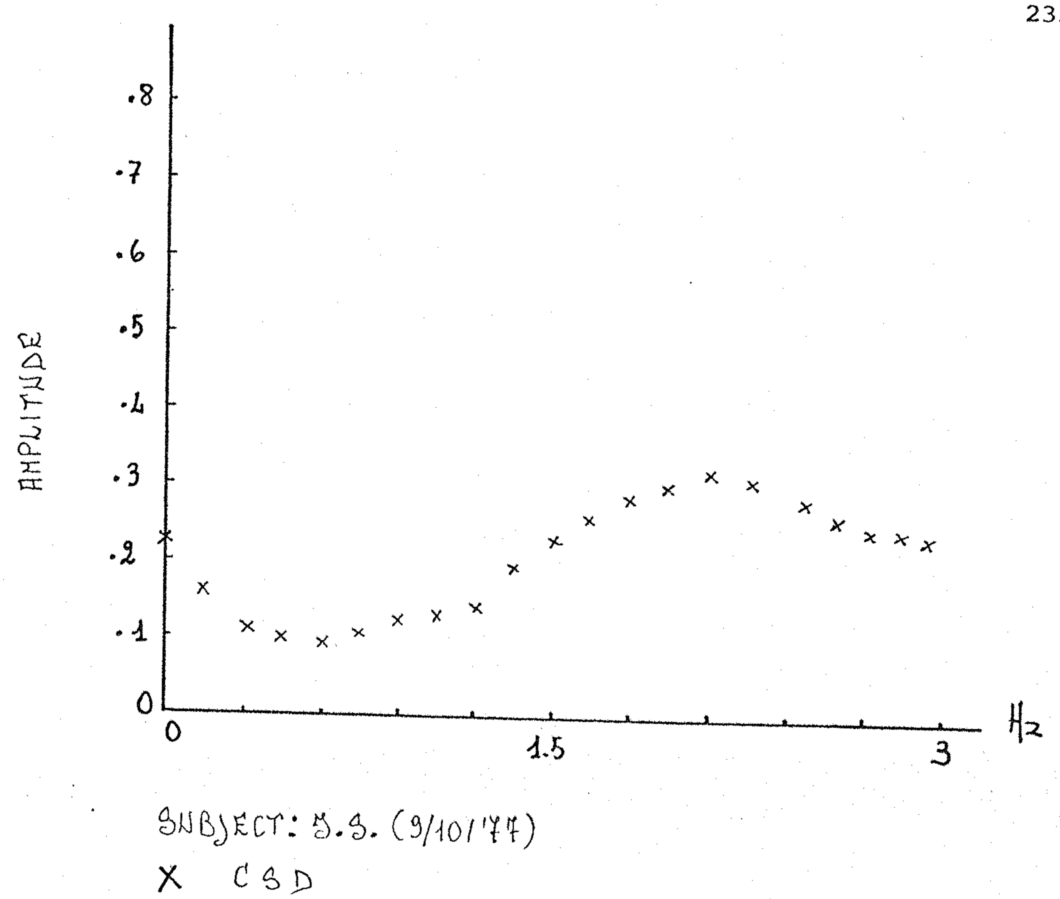


FIG. 10.20 BLIND EXPERIMENT J (Random input)

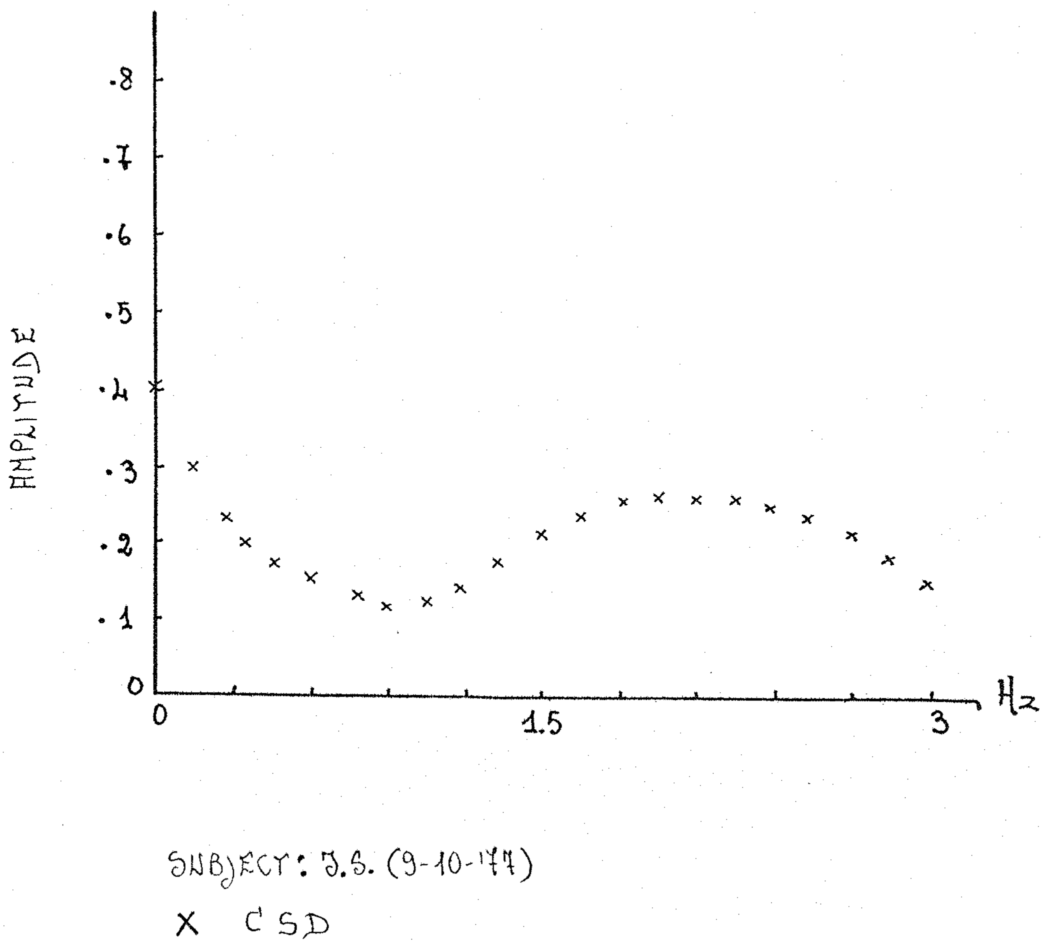


FIG. 10.21 BLIND EXPERIMENT J (Random input)

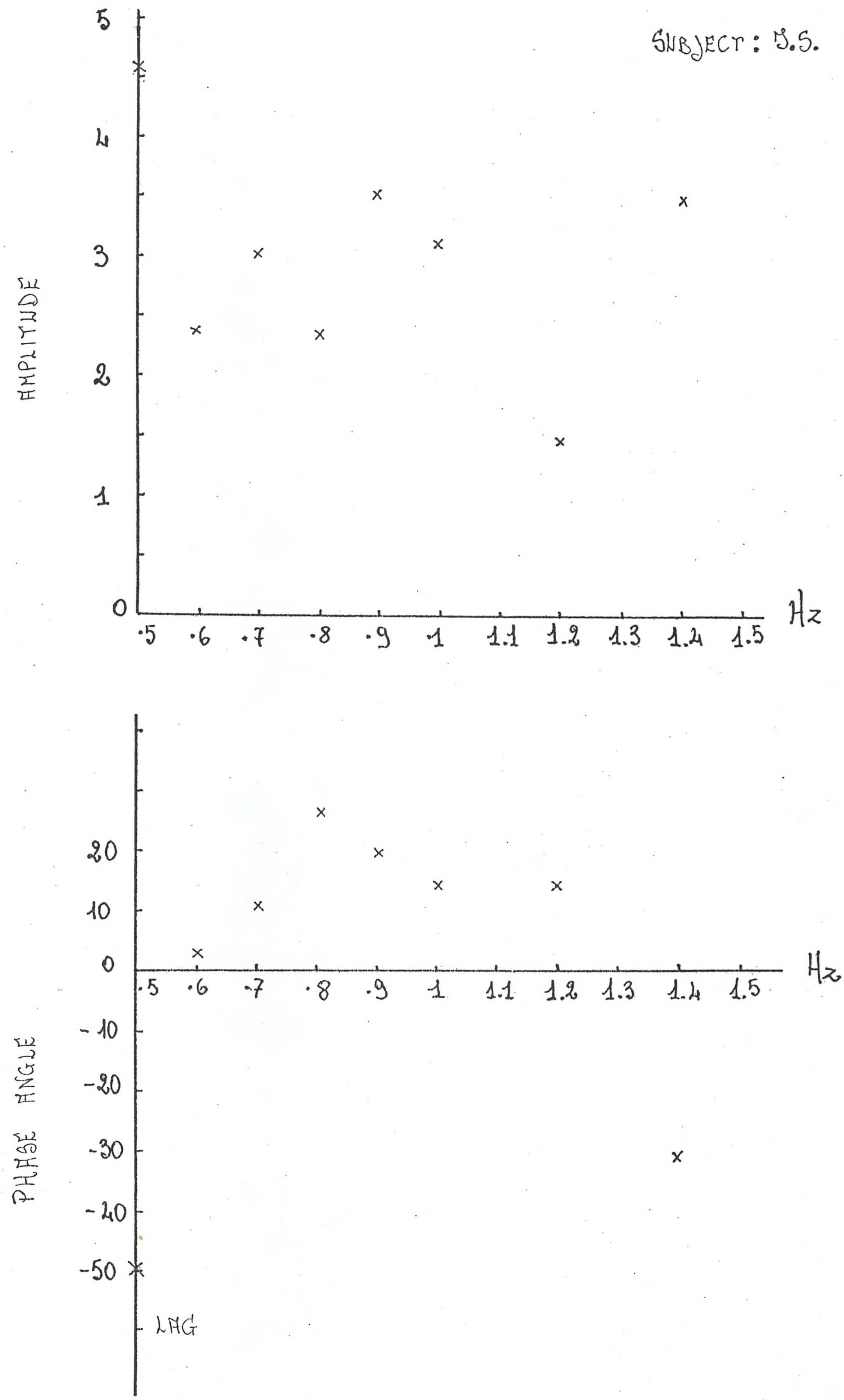


FIG. 10.22 VESTIBULAR SENSITIVITY - V

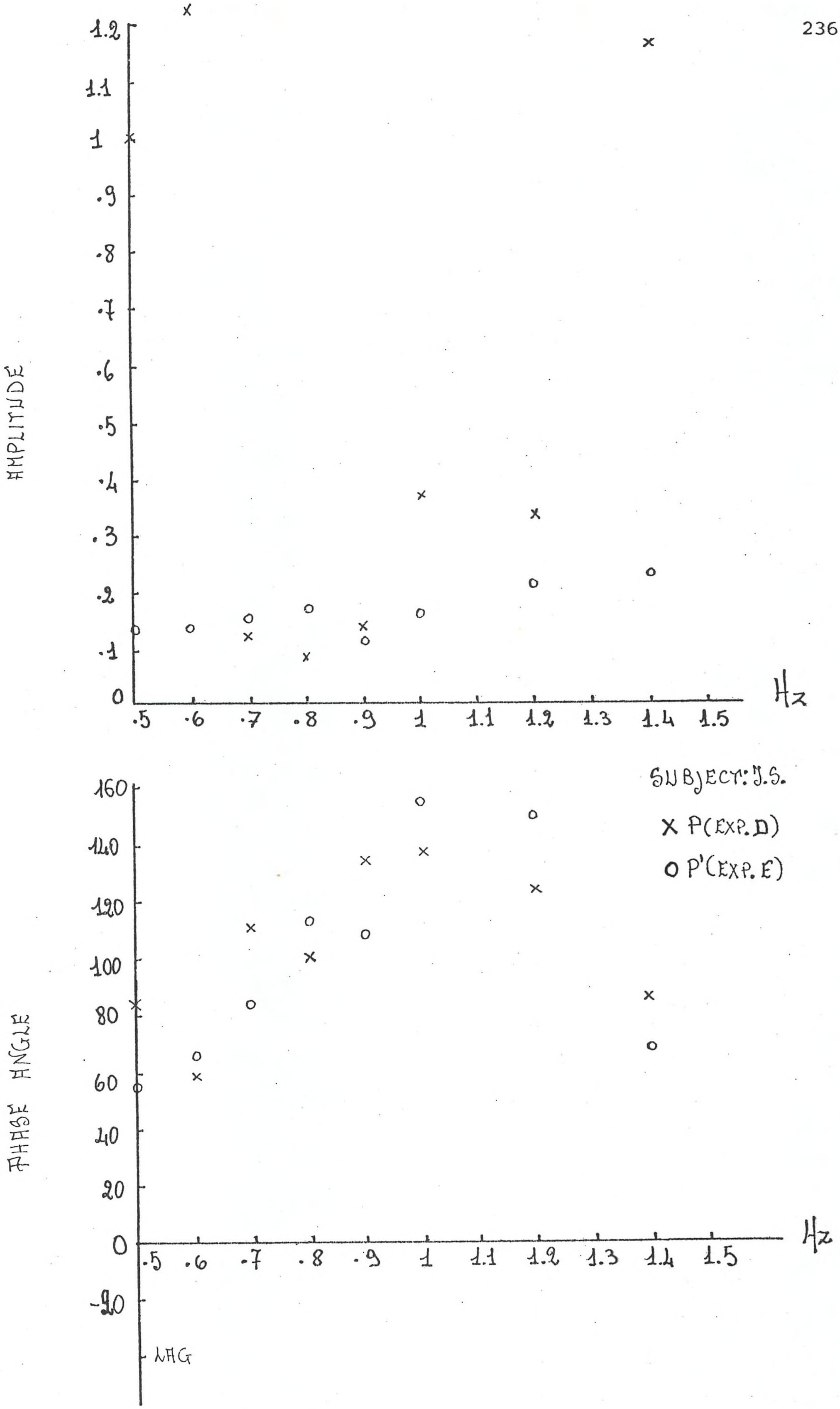


FIG. 10.23 PROPRIOCEPTIVE SENSITIVITY - P and P'

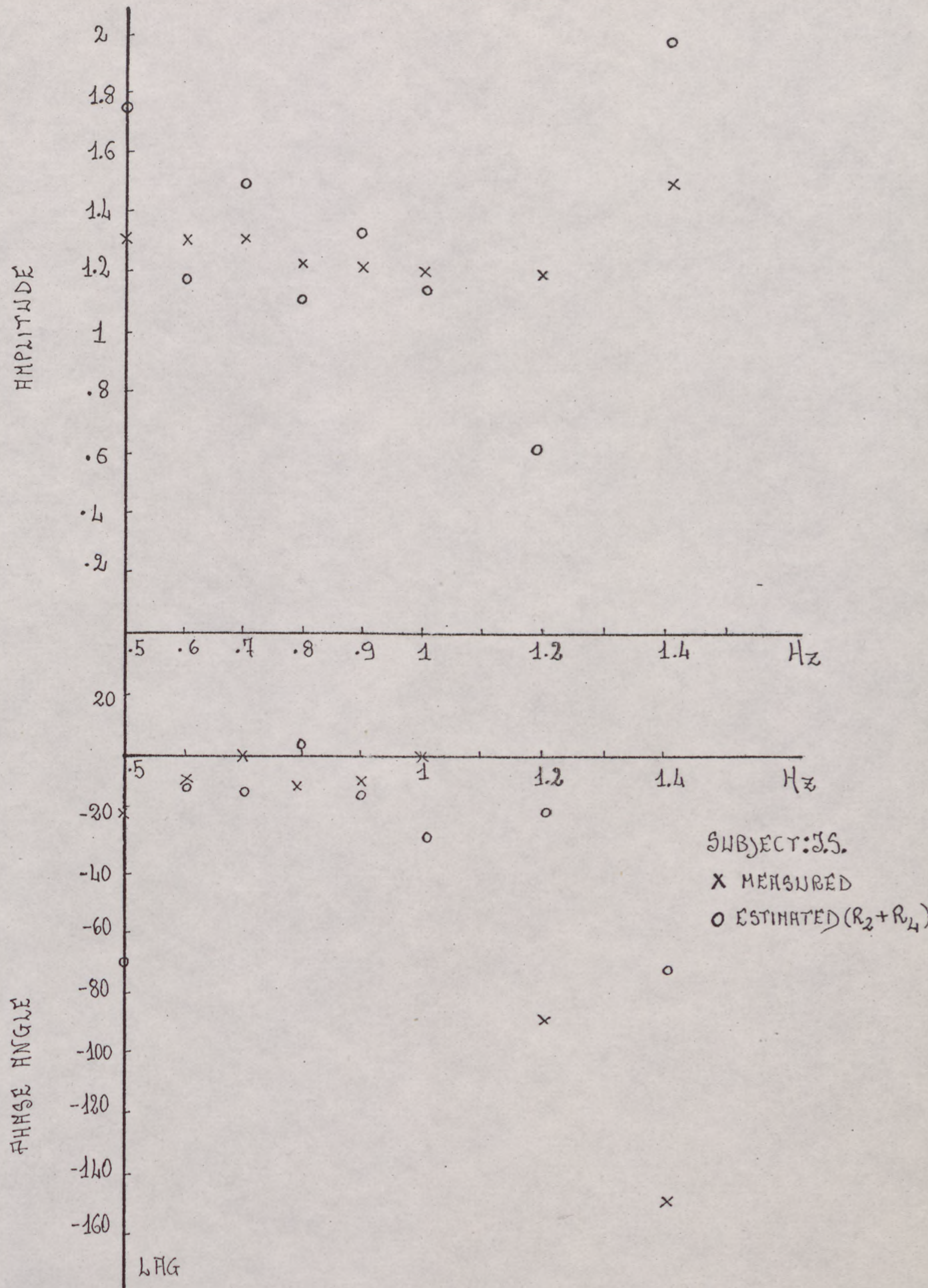


FIG. 10.24 R_3 – MEASURED AND ESTIMATED VALUES

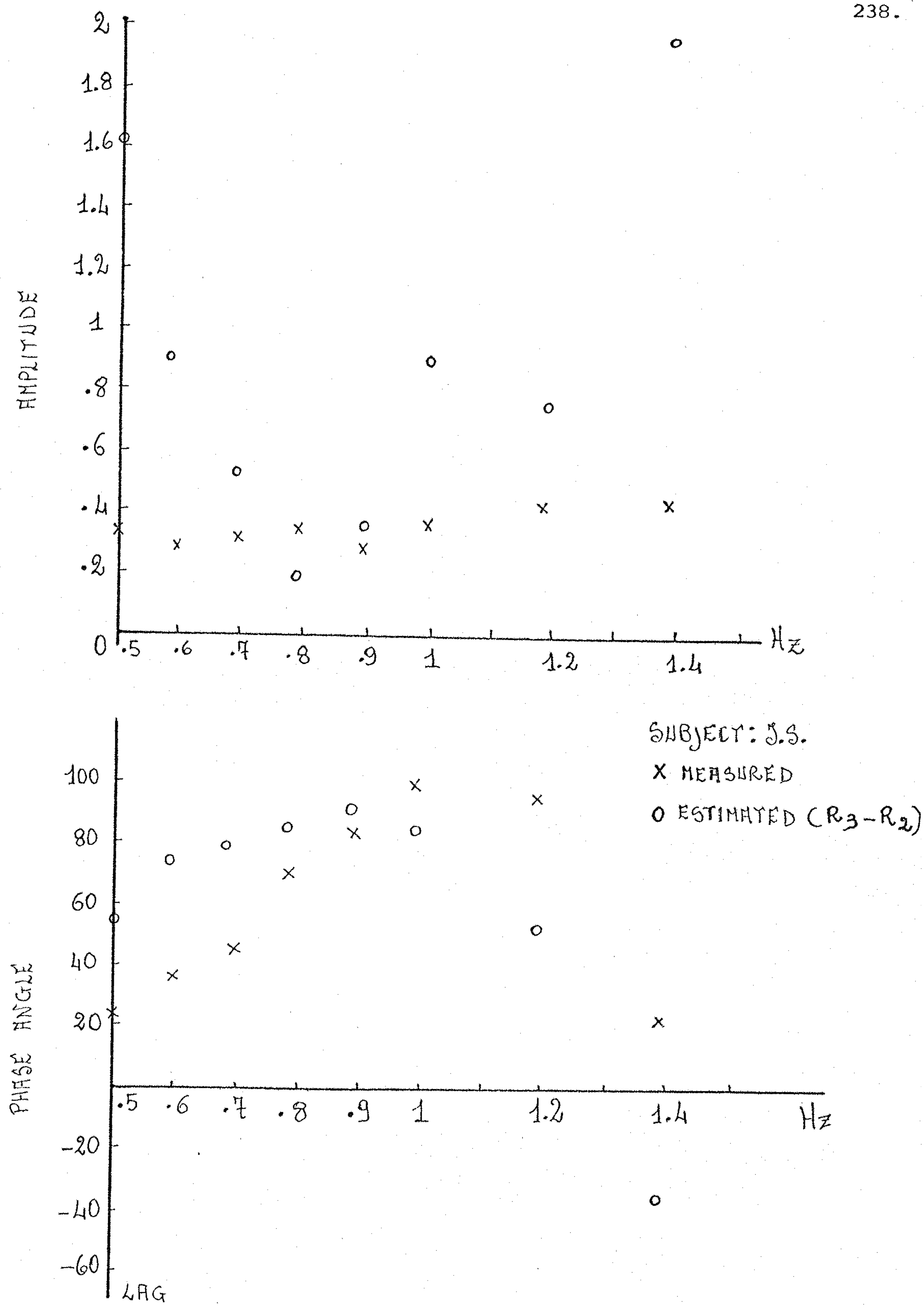


FIG. 10.25 R_4 - MEASURED AND ESTIMATED VALUES

APPENDIX 10.1

COMPUTER PROGRAMS USED

FAST FOURIER TRANSFORM: See Appendix 7.2

CORRELATION (CORR.)

```
∫ KILL (1, ..... 6)

∫ ACQUIR (12, 512, 2, 1)

∫ CONV (1, *, 4)

∫ CONV (2, *, 4)

∫ NORM (*, 3, 3)

∫ NORM (*, 4, 3)

∫ CORREL (1, 3, 4, 5, *, *)

∫ CURV (5, 6, 20, 200)

∫ DISPHY (5, 0, 6, 0)

END
```

Parameters: Cross Correlation

1. Mode = 1
2. Input file name
3. Input file name 2
4. Output file name
5. First lag
6. Last lag

Notes:

(a) Last lag number must be more positive than first lag

(b) Input file must be real

(c) A lag is defined as 1/sampling rate.

CROSS SPECTRAL DENSITY (CSD)

```
∫ KILL (1,2, ........ 10)
∫ ACQUIR (52, 4096, 2, 1)
∫ CONV (1, *, 4)
∫ CONV (2, *, 4)
∫ NORM (*, 5, 1)
∫ NORM (*, 6, 1)
∫ CSD (5, 6, 7, *)
∫ MOPH (7, 8, 0, 1)
∫ DISPLY (8, 0, 8, 1000)
END
```

Parameters:

1. Input file name 1
2. Input file name 2
3. Output file name
4. Resolution
5. Window (1 = Hanning; 2 = Parzen; 3 = Bartlett)
6. Degrees of freedom

Notes:

1. Actual resolution

   This is always less than or equal to the desired resolution because fast Fourier transform (FFT) techniques are used.

2. Degrees of freedom

   Equation = 2 [(2N/L) - 1]

   where L = SR/Actual resolution

   N = Number of samples in file

For the Results shown in Chapter 10:

Resolution = 0.25

Actual resolution = 0.203

Degrees of freedom = 60

Sampling rate = 52

Number of samples = 4096.

CHAPTER ELEVEN

TRANSIENT RESPONSE AND EMG ACTIVITY

11.1. INTRODUCTION

Although these experiments are described last, they were actually the first to be performed. The initial objective was to compare the time behaviour of the EMG response with the transient displacements to see if the arm displacements were actually under neurological control during visual and non-visual alignment tests. This was necessary, to establish if the vestibular or some other mechanism could be used to compute the position of the hand in space. Subsequently a model was proposed for the system and its frequency domain characteristics determined experimentally. The results of the transient tests then offer a simple confirmation of the frequency response results and hence of the model structure, although too much significance should not be attached at this stage.

11.2. EXPERIMENTS

11.2.1. TRANSIENT DISPLACEMENTS

The subject was seated on the platform with his arm outstretched and splinted as described in the previous chapters. Before a step displacement was applied to the platform he was required to align the hand with a fixed horizontal marker.

Step displacements of magnitude 4.5 cm and 2.5 cm were applied at instants unknown to the subject. He was often required to reposition the arm so that the hand again aligned with the marker. The tests were carried under the following conditions:

(a) looking, with and without bite bar as described in Chapters 9 - 10.

(b) Not looking with and without bite bar and with vibratory excitation of the anterior deltoid and other shoulder muscles.

Tests were carried with upward and downward displacements of the platform.

11.2.2. EMG TESTS

Physically the tests were the same as the visual tests (a) described in the previous section. As before the subject first aligned his hand with a horizontal marker and then attempted to realign it after a step displacement of the platform. EMG surface electrodes were applied to the belly of the anterior deltoid and other shoulder muscles.

Since the greater EMG activity was observed in the deltoid muscle, whose major function is to abduct the arm, see Chapter 4, the majority of experiments were performed with measurements taken only from this muscle.

11.3. MEASUREMENT TECHNIQUES

11.3.1. TRANSIENT DISPLACEMENTS

The displacement input to the platform was obtained from a waveform generator. Although this reference signal was a square pulse, of duration longer than the transient response of the subject, the dynamic response of the platform led to an actual input which contains transients, see Figure 5.8. Two accelerometers were placed, one on the shoulder as described in Chapter 7 and the other was fixed rigidly to the splint, see Figure 9.8. proximal to the top of the hand. The accelerometers were calibrated before each test.

The accelerometer outputs were recorded on a 6-channel UV recorder (SE Laboratories - Type 3006/DL) and on an FM Tape recorder (SE Lab - Type T-3000).

11.3.2. EMG TESTS

The platform displacement input was applied as before. It was measured by a linear potentiometer attached to the platform, see Chapter 5. The EMG signals were detected using Ag-AgCl surface electrodes of 8 mm diameter. One electrode was attached over the muscle belly and the other 2.5 cm away on the muscle axis. The electrodes were attached by doubled-sided adhesive tape disks. The cavity beneath each electrode was filled with conducting jelly to reduce skin contact resistance effects. Standard skin cleansing techniques were used. The associated instrumentation is described in Chapter 6 section 6.6.

The displacement and the processed EMG signal were recorded on the 6-channel UV Recorder.

11.4. ANALYSIS TECHNIQUES

11.4.1. TRANSIENT DISPLACEMENT

The data from the tape recorder were analysed off-line on the PDP - 11/50 computer. Since the outputs were accelerations rather than displacements it was necessary to integrate twice with respect to time. Standard integration routines were used. It was seen that the displacement at the shoulder was virtually the same for all subjects and was considered to be the actual input to the system.

11.4.2. EMG TESTS

The processed EMG signals were quantified by hand since visual inspection gave the most reliable means of identifying the characteristic features of the response.

11.5. RESULTS

11.5.1. DISPLACEMENT TESTS

Five subjects were used and their personal data are given in table 11.1.

The experiments were repeated approximately 10 time for each subject.

Recordings of displacements for the various conditons of the displacement tests are shown, for 2 subjects, in Figures 11.1 - 11.4. It can be seen that there is a difference between the responses for positive and negative displacements. Generally, for upward displacements, the responses (hand displacement) show an overshoot followed by an undershoot and then a reasonable equilibrium condition.

For downward displacement the responses are generally monotonic but with one pronounced fluctuation. Best performance is obtained when looking, with no impediment. Here final alignment is good.

When blindfolded the final alignment shows a positive error (hand above the reference mark). With the bite bar both tests show a response which at times moves in the wrong direction. This is particularly surprising in the tests where the subject was looking, and clearly indicates that the vestibular feedback overrides the other sensory feedback. The same trends were observed for downward motion even though the responses generally tended to be monotonic.

The worst performance of all was obtained when both bite bar and shoulder muscle stimulation were used. Here proprioceptive shoulder feedback was impaired and final errors were very large.

Ideally it would be desirable to relate the transient responses with the frequency responses discussed in chapter 9 and 10 and with the proposed models. Only simple correlation was possible with the results obtained although this could be usefully investigated in the future.

The transient response times and accuracy appear consistent with the frequency response. It is felt that the overall control system is so much more complex than the simple model we have proposed that only broad agreement might be expected. From Chapter 10 we have found that the closed loop bandwidth is approximately 2 Hz, while the transient experiment gives a rise time approximately equal to 1.65 seconds and therefore it follows that:

$$\text{Bandwidth} \times \text{Rise time} \approx 3.30$$

For a second order system this product should be 3.142. It may therefore be considered that the experimental result is in good agreement with the expected value.

As can be seen in the next section, the EMG results confirm that during this period the shoulder muscle is being actively driven. Thus the response is a controlled response rather than a passive mechanical reaction. This again offers some confirmation of the model structure.

11.5.2. EMG TESTS

Five subjects were used in the experiments and their personal data are reported in table 11.2.

For each subject the experiments were repeated approximately 10 times. Although the fluctuations for each subject were quite small the values reported here are averages taken of all subject responses. Figure 11.5 shows representations of platform displacement and EMG signals. On this diagram are shown the following response times which thought to be significant.

- τ_1 - time to positive peak overshoot for upward displacement
- τ_2 - time to negative peak overshoot for upward displacement
- T' - total duration of EMG response for upward displacement
- τ_3 - time to maximum overshoot for downward displacement
- T'' - total duration of EMG response for downward displacements

The average values of these parameters are given in Table 11.3.

11.6. DISCUSSION OF RESULTS

There is a notable difference between all subject responses for downward motion compared with that for upward motion, see Figures 11.1 - 11.5. Since visual feedback is unchanged in each case, it must be conjectured that it is the other mechanism which are sensitive to the different motional direction.

For upward motion there is over compensation shown by the displacement records. This is confirmed by the EMG signal which shows successive agonist and antagonist activity.

For downward motion there is a monotonic displacement response which again is confirmed by EMG activity of mono-polarity. There is also some difference in the duration of the processed EMG response but this may be accounted for by variation in the platform response times for upward and downward displacements.

It must also be noted that lags in the EMG processing filter (0.95 sec. - 1st order lag) cause considerable time shift in the EMG recordings.

CHAPTER 11

LIST OF TABLES

LIST OF FIGURES

TABLE 11.1.

PERSONAL DATA OF THE SUBJECTS TESTED

DISPLACEMENT TESTS

SUBJECT	AGE (YEARS)	WEIGHT (Kg)	HEIGHT (cm)	SITTING HEIGHT (cm)	FOREARM LENGHT (cm)
I.S.	22	72	172	87	60
S.L.	33	58	167	74	63
B.F.	23	81	180	93	59
M.A.	31	65	175	85	57
M.T.	23	69	182	90	65
AVERAGE	26.40	69	175.20	85.80	60.80
STANDARD DEV.	5.18	8.51	6.06	7.26	3.19

TABLE 11.2.

PERSONAL DATA OF THE SUBJECTS TESTED

EMG TESTS

SUBJECT	AGE (YEARS)	WEIGHT (Kg)	HEIGHT (cm)	SITTING HEIGHT (cm)	FOREARM LENGTH (cm)
M.T.K.	23	62	168	80	59
S.L.	33	58	167	74	63
T.D.	17	78	192	88	68
A.F.	30	69	172	83	57
M.D.	33	76	170	80	60
AVERAGE	27.20	68.60	173.80	81	61.40
ST.DEV.	7.0.	8.65	10.35	5.10	4.28

TABLE 11.3

AVERAGE VALUES EMG TESTS

SUBJECT	UPWARD DISPLACEMENT τ_1	τ_2	T'	DOWNWARD DISPLACEMENT τ_3	T''
S.L.	.93	1.84	2.79	.98	2.2
T.D.	.91	1.53	2.21	1	2.23
M.K.	1	2.1	3.10	1.09	2.91
A.F.	.99	2.19	3.45	1.14	2.74
M.D.	.96	1.89	2.97	1.17	2.43
AVERAGE	.96	1.91	2.90	1.08	2.50
ST.DEV.	.04	.26	.46	.08	.31

CHAPTER 11

LIST OF FIGURES

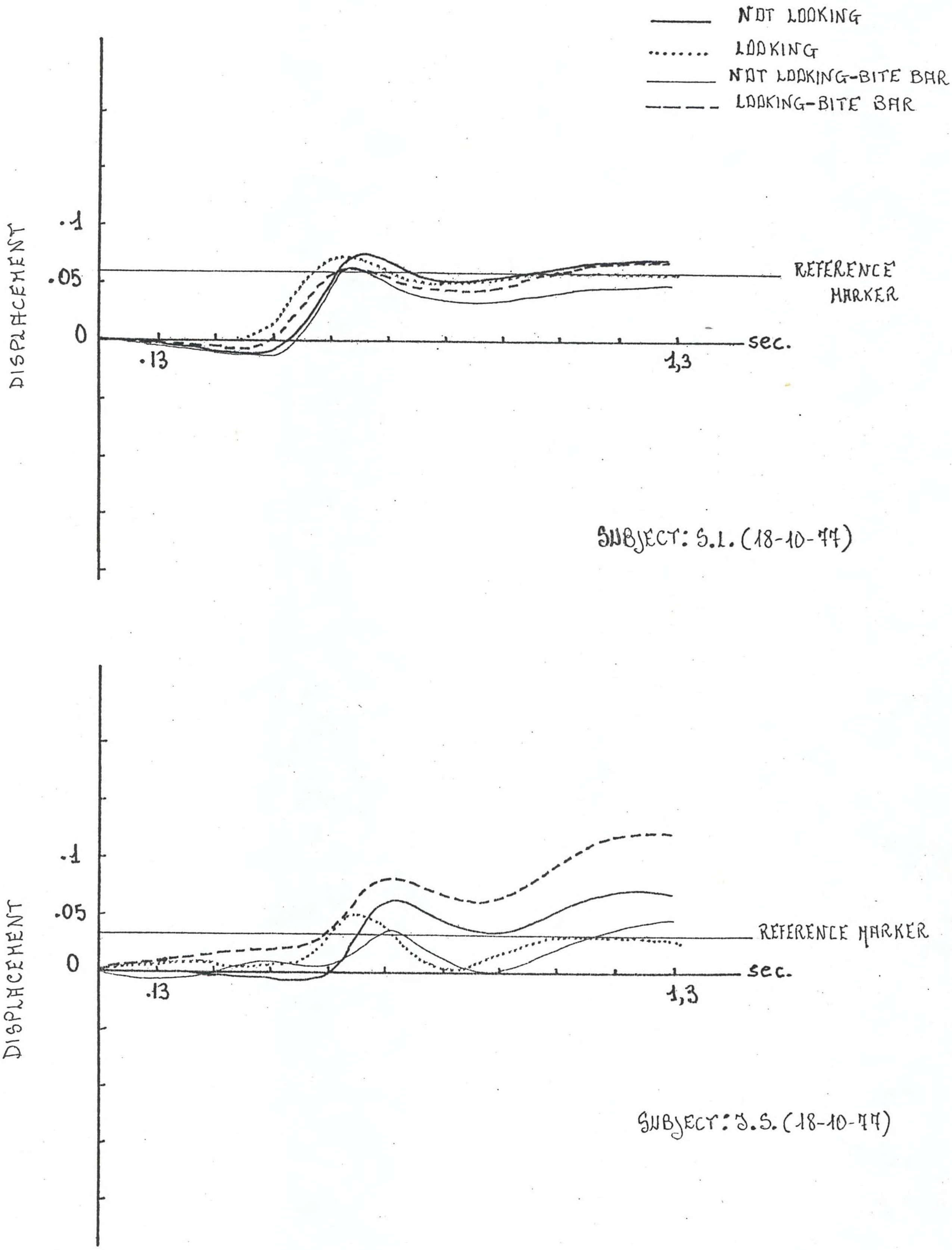


FIG. 11.1 TRANSIENT RESPONSE TEST (Upward motion)

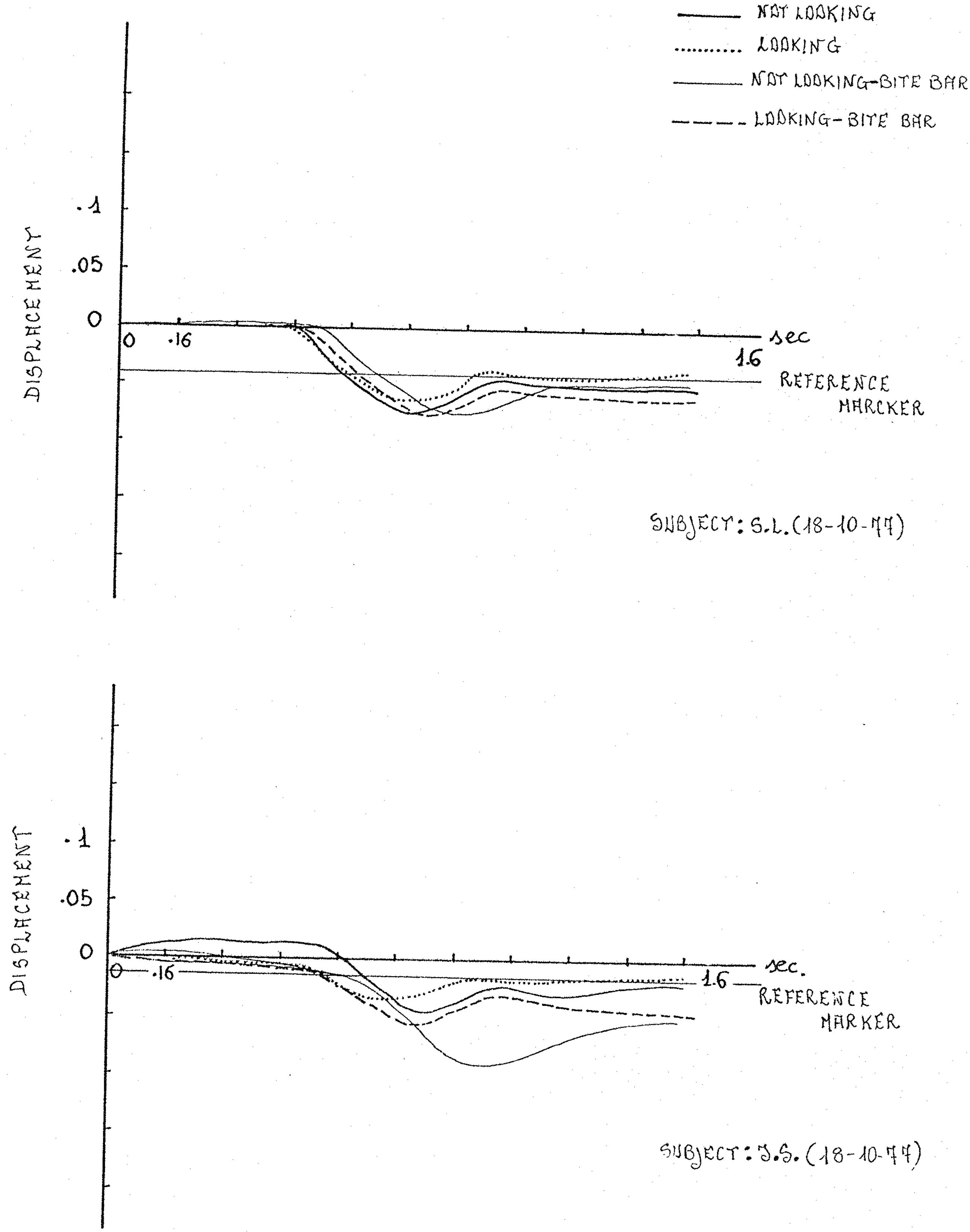


FIG. 11.2 TRANSIENT RESPONSE TEST (Downward motion)

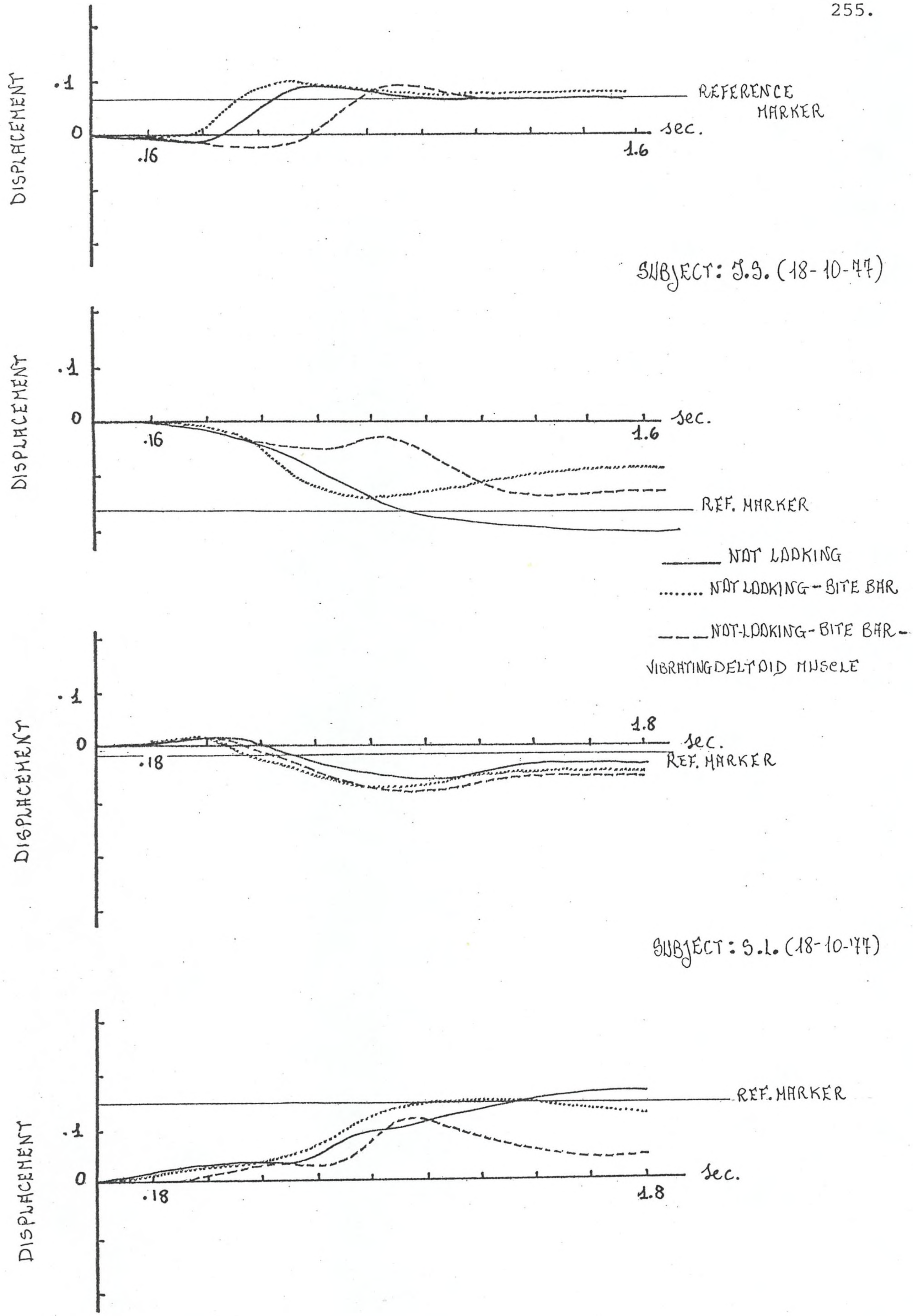


FIG. 11.3 TRANSIENT RESPONSE TEST (Upward and downward motions

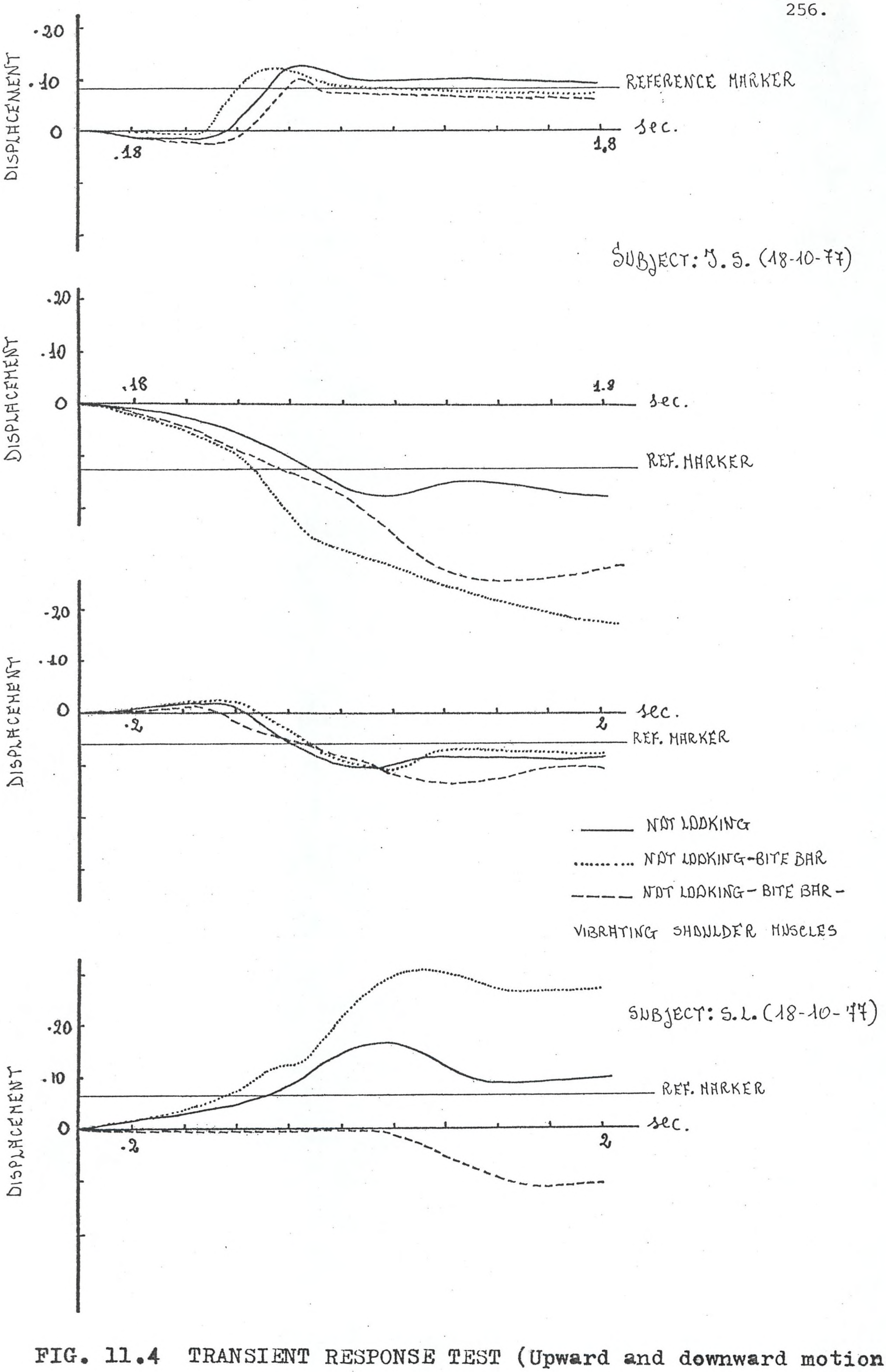
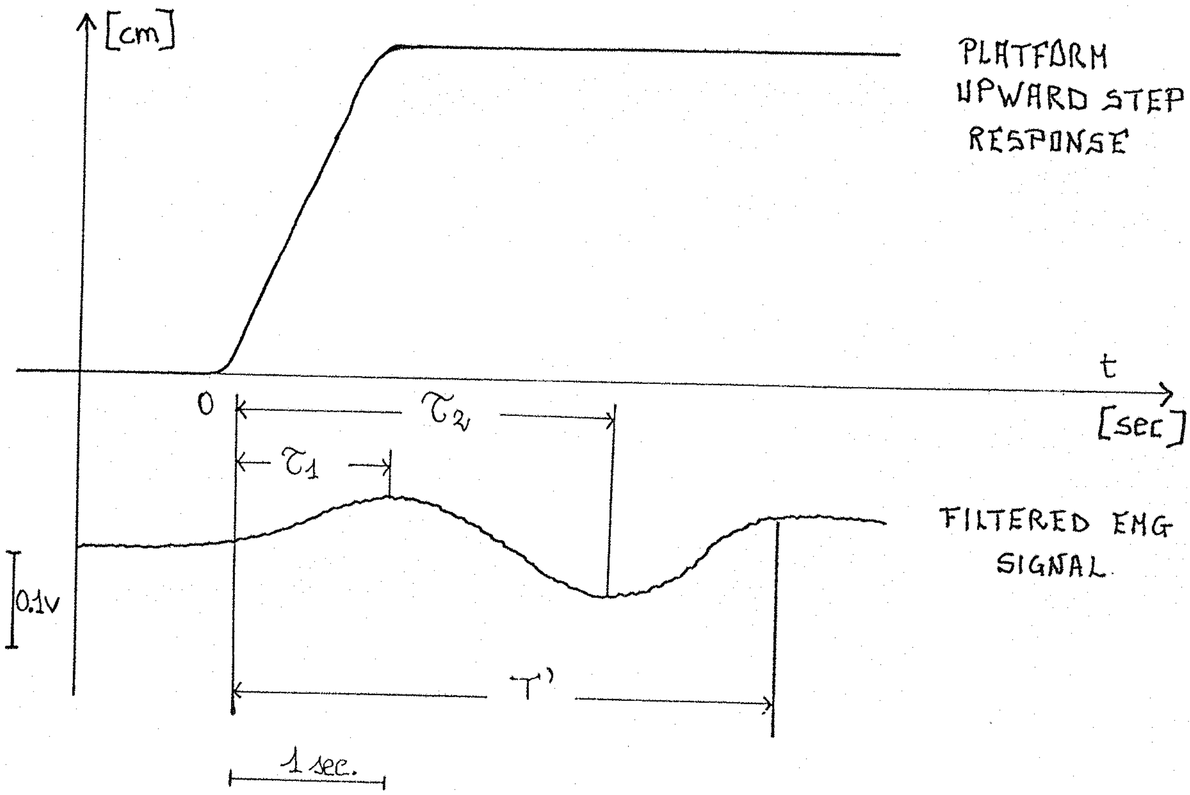


FIG. 11.4 TRANSIENT RESPONSE TEST (Upward and downward motions)



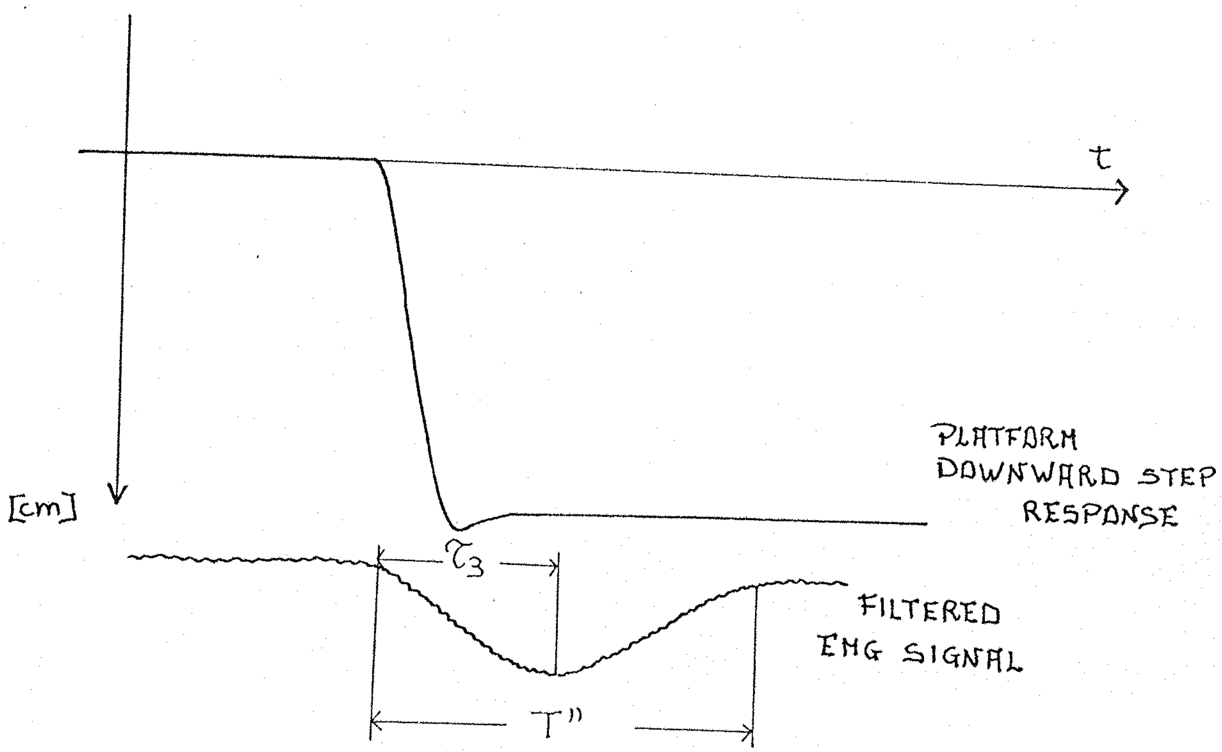


FIG. 11.5 EMG TESTS

CHAPTER TWELVE

CONCLUSIONS AND DISCUSSION

A methodical investigation of a complex system such as human postural control is severely limited because of the indeterminate structure of the CNS and because many internal variables cannot be measured without altering the behaviour of the system. While it is clear that no principles of general applicability can be established by a small study with so many variables under observation, some aspects stand out from the bulk of data which show sufficient distinctive consistency that they appear to merit attention despite the limitations of the investigation. We have attempted, first to devize non-invasive tests to effectively separate certain aspects of motor function in terms of individual elements and to isolate these elements by impairing or disabling other parts of the system

Secondly, the interpretation and validation of the experimental results so achieved is desirable. Many models of the mechanisms involved in human postural control have been formulated but almost without exception they are quantitative and cannot be verified. The model proposed here at least has the advantage of simplicity and the various sensory channels through which postural compensation may be achieved involve only superposition of relatively simple systems. The adaptive behaviour which is known to govern human function in relevant circumstances can thus be explained solely in terms of the selection of appropriate modes of control. This simple selective type of decision process is much more believable in terms of physiological structure of the CNS than the complex adaptive models which rely on a computer-like structure in the brain. No evidence for a programmable brain really exists according to recent views in neurophysiology.

The crude model which has been presented herein therefore represents a necessary compromise between simplicity and completeness.

It contains a minimal number of parameters which, in future work, may be related to the physiological properties of the control system under investigation. At the same time, some phenomena which are presently believed to be of importance have been included. The model is linear but since the true control structure is not known, the inclusion of some arbitrary non lineraties would not contribute much to our understanding of the mechanisms being considered.

Given that only a crude model has been established then we can make suggestions for improving the experimental investigation and the evaluation of the model in terms of numerical parameters. Finally the significance of the results to the design of prostheses may be attempted. The limitations of the work here presented are as follows:

(a) Difficulty or perhaps impossibility of obtaining physical measures of variables without distorting the experiments. This is partly due to the tenuous balance between voluntary and involuntary responses when the human operator performs a given task.

(b) Difficulty in removing completely spurious effects such as audio and other feedbacks which offer alternate subconscious feedback mechanisms to the operator.

(c) Difficulty in eliminating predictive and learning characteristics of operator.

(d) Difficulty to disable particular pathways when trying to establish the superposition of multi-feedback channels.

(e) Extreme variability of the results amongst different subjects.

(f) Only contrived experiments are available to separate, if possible, the individual component in the system. The results must be interpreted only in appropriate situations. But this is also true for any tracking type of experiment.

(g) Low cost apparatus has been used and experimental techniques could certainly be improved by use of more sophisticated low-frequency shakers with particular emphasis on the suppression of noise and vibration associated with the motor drive.

(h) Physiological aspects such as the duration of each run, particularly in swept input excitation, to obtain a more uniform spectrum by using a lower sweep rate; caloric stimulation which shows very variable results, the effect of fatigue and anxiety on the operator performance, etc.

Bearing in mind the above limitations, the following comments can be made from the various tests carried out.

(i) Visual feedback appear to dominates all the other feedbacks available to the human operator when it is available.

(ii) The human operator response indicates that, in absence of visual input, vestibular and proprioceptive feedback provide compensation.

(iii) The human operator response to a periodic input clearly shows a leading phase almost certainly due to his predictivity. When the frequency of the input signal increases or when the input is a random signal, the predictivity is lost and therefore the phase becomes increasingly lagging.

(iv) Proprioceptive feedback, whenever it is functioning, improves the postural compensation of the human operator.

(v) The left and right sides of the vestibular apparatus are not balanced with respect to vertical motion.

(vi) When the head of the human operator is bent in one direction, impulses from the neck proprioceptors contribute to a better postural control.

(vii) The vestibular sensitivity appears to be more significant than the proprioceptive one from the results calculated indirectly from the model.

(viii) Both vestibular and proprioceptive responses show a leading phase characteristic. The proprioceptors give greater leading angles.

(xi) There is compensation within the human control processor which is certainly adaptive in nature. With our model this adaptation could merely be selection of an appropriate feedback channel.

(x) Transmissibility of low frequency vertical vibrations to the shoulder show a resonant frequency around 5 Hz with a lagging phase of 60 degrees.

(xi) Transmissibility of low frequency vertical vibrations to the head show an average resonant frequency of 4.5. Hz with a lagging phase of 47 degrees.

(xii) The human operator response to a sudden displacement, i.e. step input, is a controlled response rather than a passive mechanical reaction.

(xiii) The human operator frequency bandwidth is somewhere near 2 Hz.

(xiv) The human operator rise time to a sudden displacement is around 1.65 seconds.

One test of the feasibility of such a model of a physiological system is whether a mechanical system could be built which functions successfully using the same principles. Thus the prehension control scheme for a fully-flexured artificial hand which has been developed in the Control Group at Southampton University (TODD, 1970; CODD, 1975; STORY, 1977; NIGHTINGALE-SWAIN, 1978; MOORE, 1980;) does give a useful model for the study of the human hand even if it is not a structural analogue.

Recently work in the group has been directed to the control of a fully-articulated mechanical arm with 6-degrees of freedom. Once again a control scheme has been sought in which conscious human participation is no more than for the normal hand (SWAIN-NIGHTINGALE, 1980). This has necessitated the use of sensors to provide much information on the kinematics and kinetics of the limb and its tactile element. A scheme has been proposed, see Figure 12.1. which does make use of artificial proprioceptors, through potentiometers, etc. and inertial sensors (the equivalent of the vestibular system). Some experimental work has already been performed on this system. A central computer facility has been used but it has not fulfilled a numerical-algorithmic role. Rather it has been used to make selective decisions, based on simple criteria, at relative points in the control structure. Only in this way has it been found possible in these complex prostheses to meet the conflicting requirements of multiple decision-making and speed of response. Certainly so far as prosthetic and other bio-engineering systems are concerned, the kind of model adopted here seems very appropriate.

CHAPTER 12

LIST OF FIGURES

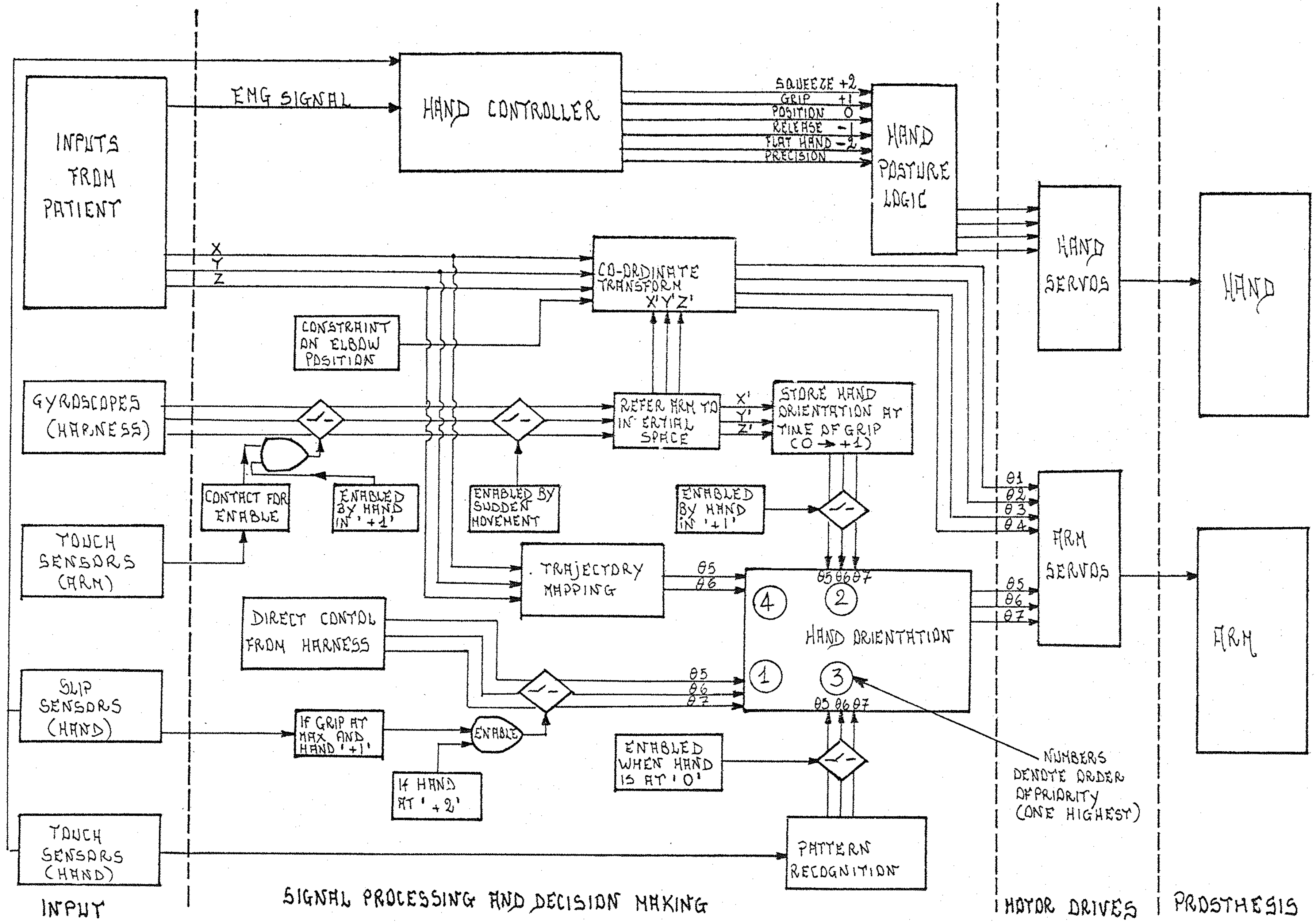


FIG. 12.1 COMPLETE SOUTHAMPTON HAND/ARM CONTROL SYSTEM (Swain - Nightingale, 1980)

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