MODELING THE POSTURAL CONTROL SYSTEM
Modeling the Postural Control System
of the
Exoskeletonally Restrained Human

by
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ABSTRACT

In this work, the postural control system of exoskeletonally restrained human subjects is studied experimentally and theoretically. A mathematical model of the system was developed using information available from the literature. Experimental data describing the responses of exoskeletonally restrained subjects to perturbing pulses of force are presented. The validity of the theoretical model was assessed by simulating the model on a hybrid computer and comparing the simulated responses to those observed experimentally. The model structure and parameters were altered to improve the matching between the simulated and observed responses. An improved model of the postural control system based on both theoretical and experimental data is presented.
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CHAPTER 1
INTRODUCTION

1.1 SUBJECT OF THESIS

In general, the subject of this thesis is the Postural Control System (PCS) of man. The system is of interest in itself because it is responsible for maintaining man's upright stance and it also provides a means of studying integration in the Central Nervous System (CNS) because of the many receptors involved in postural control.

In particular, the thesis describes an experimental and theoretical investigation resulting in the development of a mathematical model of the Postural Control System of exoskeletonally restrained human subjects.

1.2 THESIS OUTLINE

In this section an outline of the thesis is given as an introduction to the work.

In Chapter II the literature related to posture and postural control is reviewed from a historical viewpoint. The many different approaches to the study of posture which have been tried in the past are discussed. In addition, the neurophysiology relevant to postural control is briefly described.

In Chapter III the currently accepted mathematical models of the sensory receptors thought to be involved in postural control are first surveyed. An information flow diagram for the PCS is developed using these models as building blocks. Then a number of assumptions and restraints are made to convert the information flow diagram to a linear model of the PCS which is then used to plan the experimental program and also as a starting point in simulating the PCS.

In Chapter IV the experimental apparatus is described. This consists of:

- instrumentation to measure the state variables of the system;
- an exoskeleton to enforce experimentally the restraints assumed in the analysis of Chapter III; and
- a stimulator to supply a standard stimulus to the system.
In addition, the interfacing of the apparatus with a PDP-12 computer to allow digital data acquisition and computer control of the experiments is discussed.

In Chapter V the experimental protocol used in the investigation is described in detail.

In Chapter VI the data processing techniques applied to the experimental data are first described and then the experimental results are presented and discussed.

In Chapter VII the modeling technique and hardware used in the identification procedure are described. Two proposed models of the PCS are described and the results of simulating them are presented.

In Chapter VIII pertinent conclusions are drawn and recommendations are made for further work.
CHAPTER II
HISTORICAL REVIEW

2.1 INTRODUCTION

Literature dealing with human posture is found throughout the medical and physiological journals. In this chapter a brief survey of the more important reports will be given as a background to the research reported in the rest of the thesis.

2.2 STATIC POSTURE

Many investigators (especially those who were clinically oriented) have treated posture as a static situation and developed measures of posture based on the spatial orientation of body segments. The aim of the research was often to gather data from subjects with "good" posture for use in assessing pathological or "poor" postures. "Poor" posture, other than that due to skeletal or neuromuscular pathologies, has at times been held responsible for such varied disorders as varicose veins, constipation, arthritis, neuritis and hemorrhoids. Little or no proof of these assertions has ever been presented(24).

Early attempts to measure the geometric parameters of static posture found that the variations between "good" and "poor" postures were so slight that precision measuring instruments were required to detect them. Consequently many measuring devices were developed, often using some type of photography. Biplane stereoscopic pictures or mirrors were used to obtain a three-dimensional record of body position. In addition, mechanical devices were developed to measure the depth of the anterior-posterior curvature of the spine. Records obtained with these devices were usually evaluated by some type of visual inspection, based on arbitrary criteria which were more aesthetic than functional in nature(24).

Molhave (48) used a mechanical instrument - the statometre - to define the spatial relationships of the bony landmarks of normal subjects. Hansen (20) later used the same device to study patients before and after operations for
slipped discs (lumbar discectomy). Distinct differences from the normal data of Molhave were noted and used as a basis for suggesting post-operative physical therapy. The work is unique in that large numbers of patients and normals were examined and statistical analysis was used to evaluate the results.

None of the devices developed to "measure" posture is currently in common usage. The only area where static measures appear to be useful is in such diseases as scoliosis where a measure of the progress of the disease is needed. Aside from this, the subjective judgement of the physician appears to be still the best means of assessing static posture.

Nubar and Contini have suggested a theoretical criteria which the body may try to minimize in selecting a posture. Apparently no attempt has been made to verify this theory experimentally.

2.3 SWAY

The sway which always accompanies the act of standing has interested researchers since Romburg first observed that patients with tabes dorsalis tended to sway much more with their eyes shut then they did with them open. Behaviour of this type has come to be called the Romburg sign and is a significant clinical neurological sign. Numerous methods have since been developed to measure postural sway quantitatively as a means of assessing psycho-motor function.

Sway has been measured by many workers simply by attaching some type of position transducer to a body landmark, and recording the variation of position with time. The position of the landmarks has been measured in a variety of ways with differing degrees of sophistication. Some of the early work suffers from the fact that results were presented in arbitrary units and thus cannot be easily compared to other workers' results.

A second and less direct method of assessing sway used by many investigators has been to observe variations in the position of the effective centre of pressure of the subject in the horizontal plane. In the static case this corresponds exactly to the position of the centre of gravity of the subject. However
when motion is present inertial effects can easily be of the same magnitude as gravitational effects. This fact has not been recognized by all investigators (22,23,70). Some type of force plate capable of measuring ground reactions is necessary to observe the variations in centre of pressure position. Numerous such devices have been built and used to study sway (5,21,38,49,68,71).

The data obtained with these two techniques usually consisted of an X-Y plot of either centre of pressure or landmark position in the horizontal plane for a period of time. In some cases the data was presented simply as a plot of position against time. Analysis of this data, in most cases, was either by direct inspection or by use of some vague type of quantitative measure such as total displacement or the limits of sway. Such methods are difficult to compare since they were often presented in arbitrary units or were defined over different periods of time (46,62).

Hellebrandt (22) related the limits of centre of pressure motion to the base of support and showed that the centre of pressure never falls behind the ankle axis during quiet standing. Whitney (72) made similar observations while subjects performed different activities and noted that the centre of pressure could fall behind the ankle axis for short periods during active sway. Thomas and Whitney (70) appear to be the only investigators to attempt to compute RMS values for the sway. Furthermore, although many workers have reported the approximate frequencies of sway, there have only been a few attempts to calculate the power spectrum of sway (36,61). The work of Skogland (62) is the best example of an investigation of the sway of patients with different disorders in which statistical analysis was applied to the results.

There is a qualitative agreement in the literature about the following points:

(i) some sway is always present during quiet standing. The largest component of this sway is at a frequency of about 0.2 Hz (38,65,70), with a small tremor at about 10 Hz superimposed on top of it.

(ii) sway in anterior-posterior (a-p) plane tends to be larger than that in the medial-lateral plane.
(iii) sway tends to be larger with the feet together than with the feet apart.

(iv) sway is significantly increased by closing the eyes (62), while blind people sway about as much as sighted people with their eyes closed (27). Baron and Fillizot (5, 6) showed that the contents of the field of vision can effect the pattern of sway.

(v) the sway of an individual varies considerably from time to time.

(vi) dizziness, alcohol, nicotine, sleeplessness, fatigue, many drugs, and almost all types of neurological diseases cause an increase in postural sway (49).

(vi) certain types of schizophrenic patients display a sway of very small amplitude (68).

Although many factors affect the degree of sway there do not appear to be any patterns of sway which are specific to individual disorders. It would appear that if any useful clinical information is to be obtained from sway information, more sophisticated analysis techniques must be used.

2.4 BIOMECHANICS

Another more theoretical group of studies has attempted to gain a better understanding of the human posture through the application of the principles of mechanics. Works dealing with posture from this viewpoint often deal with locomotion as well.

An early investigation using this approach was that of Braune and Fischer (8). Photographic records of the position of body segments of subjects during standing and walking were obtained. Data from these photos were then combined with figures for segment mass and centre of gravity location, obtained from cadavers, to calculate moments about body joints. The results showed there was instability about the ankle joint unless muscle forces were acting. Braune and Fischer were the first to prove that quiet standing was not statically stable.
Steindler\textsuperscript{(67)} clearly describes the anatomical structures which allow the knee, hip and spine to remain stable with little or no muscular activity. His work was much in advance of its time in attempting to explain clinical observations by the principles of structural mechanics.

Hellebrandt\textsuperscript{(25)} used an elementary force plate and photographic analysis to determine the effect of carrying an army pack on postural stability. She noted a forward shift of the centre of gravity and a forward lean of the upper part of the body and postulated that this was a compensatory device aimed at improving postural stability but did no calculations to prove it. More recently Thomas\textsuperscript{(69)} did a similar set of experiments and reached essentially the same conclusions.

During the late 1940's and early 50's there was considerable work done using biomechanical principles to analyze gait\textsuperscript{(9, 10)}, with the aim of improving the design of lower-limb prostheses. A force plate, glass walkway, and cine-photography were used to do a detailed force and moment analysis of locomotion. Surprisingly this equipment does not appear to have been applied to the study of standing.

Smith\textsuperscript{(64, 65)} determined the forces developed at the ankle and knee by the passive tissue structure. Treating the body as a rigid rod and making other assumptions, he then calculated the forces developed by the calf muscles during standing. However, several of the assumptions made appear to be unreasonable and this part of the analysis is probably not valid.

Thomas and Whitney\textsuperscript{(70)} used a force plate and an abdominally mounted linear accelerometer to demonstrate that the observed force patterns could not be explained by a rigid oscillation about the ankle.

Recent work at the University of Strathclyde has used a force plate, cine-photography, EMG's and detailed mechanical analysis to determine the stresses in ankle, knee and hip joints during walking. The aim of this work is to improve the understanding of joint diseases and aid in the design of internal joint prostheses\textsuperscript{(57, 66)}.

Recently several mathematical models have been developed aimed at
understanding both the statics and kinematics of posture. These models are mathematically very complex, but nevertheless are still a long way from representing a very realistic physical situation and hence are of only limited use (13, 26).

2.5 ELECTROMYOGRAPHY

The role played by muscular activity in postural control has been the subject of considerable investigation over the years. Early workers tried to assess a muscle’s activity by palpating the muscle belly and tendon and through a study of functional anatomy. Palpation failed to detect any activity during quiet standing and hence the theory that postural stability was maintained by passive tendon and ligament forces arose.

The advent of modern electronic amplifiers made possible a new method of assessing muscle function – electromyography. Given enough gain, it is possible to detect with surface electrodes the potentials accompanying the contraction of a single motor unit. Thus, when correctly used, the technique is an extremely sensitive indicator of muscular activity.

Electromyography was first used to study posture in the late 1940’s and since then the activity of almost every muscle in the body has been studied during quiet standing. Among early workers, there was some disagreement about the existence of muscular activity in the muscles of the leg during quiet standing. It now appears that this was due to different techniques and electronics, for today there is general agreement about the nature of the muscular activity controlling posture (38).

Two recent reviews of electromyographic studies of postural muscular activity summarize current viewpoints (7, 37). The most important postural muscular activity occurs in the muscles of the lower leg. This activity consists of a low level phasic activity of the soleus, gastrocnemius and tibialis anterior muscles. Activity in the soleus occurs most frequently while the other two muscles are sometimes completely silent. These contractions are of such a low level that it is not surprising that palpation failed to detect them.

The muscles of the foot, knee and almost all the hip muscles are completely
silent during quiet standing but come into use during forced sway and walking. The iliopsoas at the hip appears to be in constant activity even during quiet standing. Functionally it appears to act as a ligament preventing hyperextension of the hip.

During standing erect, only a very slight, intermittent reflex activity of the muscles of the back occurs. Stabilization appears to be effected by the ligamentous structures of the back. However, flexion of the back or an increased load causes these muscles to become active.

Similarly only slight activity has been detected in the muscles of the abdomen during relaxed standing.

The muscles of the thorax have not been thoroughly investigated but it is felt that they may be responsible for the postural adjustments which compensate for respiratory movements.

There is also a scattered low level activity in the muscles of the mandible and in some of the muscles of the shoulder girdle. These muscles all appear to act as ligaments to resist the pull of gravity.

In summary, it appears that the muscles of the ankle are primarily responsible for the maintenance of postural stability. Although some other muscles are active, they appear to act only as ligaments resisting the pull of gravity.

2.6 NEUROPHYSIOLOGY OF POSTURAL CONTROL

Current neurophysiological theory holds that postural control is effected by a complex reflex mechanism. The mechanism employs sensory information from receptors of many types and appears to be under some type of supraspinal control.

In this section the present state of knowledge about receptors which could be of importance in postural control will be discussed. The anatomical interconnections of these receptors at a spinal and supraspinal level will be described as well as what is known of CNS effects. Since the literature dealing with these subjects is vast, no attempt at a historical survey will be made, and only recent survey articles will be cited.
2.6.1 Muscle Receptors

The principle receptors found in muscle are muscle spindles and Golgi tendon organs (18).

(a) Muscle spindles are the most complex and perhaps the most important of the peripheral receptors used in muscular control. They are found scattered throughout the bellies of most skeletal muscles in parallel with the muscle fibers (41).

The anatomy of a mammalian muscle spindle is quite complex but it essentially consists of a small number of muscle fibers surrounded by a sheath of connective tissue. These intrafusal fibers receive efferent innervation from motor axons of the Gamma (γ) subgroup which are smaller than the Alpha (α) motor axons of ordinary skeletal muscle. Afferent sensory fibers from spindles consist of a single primary ending and several smaller secondary endings (18, 42, 58).

The response of both primary and secondary endings is a function of both muscle length and rate of change of length. The velocity dependent (or "dynamic") portion of the response is associated with a short time constant (less than 1 second) while the length dependent (or "static") portion has a much longer time constant.

Stimulation of the "γ" efferents can alter the response of the muscle spindles. Some of the "γ" fibers affect only the "static" portion of the response while others alter only the "dynamic" component of the response. Recent attempts at modelling muscle spindles mathematically will be discussed in the next chapter (15, 18, 40, 42, 58).

Sensory afferents from spindles enter the spinal cord and divide in ascending and descending portions. Descending fibers synapse with motor neurons of the spindle's muscle and its agonists in an excitatory fashion. Other descending fibers form inhibitory synapses with motor neurons of the antagonists to the spindle's muscle. Ascending fibers rise to the level of the cerebellum and thalamus but apparently do not reach the conscious level. These anatomical connections are shown schematically in Fig. 2.1 (15, 18, 42, 58).
Figure 2.1: Neural connections of muscle spindle afferents.
(b) **Golgi tendon organs** are found in the tendons of muscles in series with fibers from several motor units. These receptors respond to the tension in the muscle. Although they display a high threshold to passively applied tension they have a much lower threshold to active contraction of the motor units to which they are attached. A detailed mathematical model for the tendon organ response has recently become available and will be described in the next chapter \((30, 31, 33, 34, 41)\).

There do not appear to be any efferent connections to the tendon organs. The afferent fibers enter the spinal chord where they split in the same way as do the spindle fibers. The descending branches form inhibitory connections with motor neurons of the tendon organ's muscle and with synergistic muscles \((12)\). Collateral fibers form facilitatory connections with antagonist motorneurons. The ascending pathways rise to the cerebellum and thalamus and perhaps to the sensorimotor cortex. These connections are shown schematically in Fig. 2.2.

(c) **Other Receptors.** Other small nerve endings are known to be present in muscle although little is known about them. Their stimulation probably leads to general flexor withdrawals and pain responses \((18)\).

### 2.6.2 Joint Receptors

Receptors in and around joints include tendon organs, Ruffini endings, Pacinian corpuscles and a group of simpler endings.

(a) **Pacinian Bodies** are onion-shaped receptors found in the periosteum, tendon sheaths and deeper layers of sensitive skin. The receptor responds to changes in pressure but has no steady state output with all output dying away after five seconds. It is not known whether or not the Pacinian Bodies contribute to muscle reflexes \((14, 18)\).

(b) **Ruffini Endings** are found in the connective tissue of joint capsules. Changes in joint position give rise to a transient followed by a new steady state firing level which is dependent only on the joint position and not on the past history \((63)\).

Although a single Ruffini ending responds over only a small part of the
Figure 2.2: Neural connections of tendon organ afferents.
joint range, there are a number of receptors throughout the joint with different ranges. Furthermore, cells, higher in the nervous system, which respond to a large range of motion have been observed. These probably integrate the data from many Ruffini Endings (14, 18, 58).

(c) Tendon Organs are similar to Golgi tendon organs but are located in some ligaments (cruciate, patellar, and collateral) in such a manner as to be unaffected by muscle tension. These receptors have a steady state output depending on joint position and hence could serve as position indicators (18).

(d) Afferent Connections from joint receptors enter the spinal chord where they split. Descending fibers enter the local interneuronal pool where their effects are not well known. Ascending pathways take a direct route which may lead to cortical levels with only two synapses. These pathways are very likely important sources of "body image" or conscious proprioception (12).

(e) Other Peripheral Receptors. These include touch, pain, heat, cold, all of which may have some slight bearing on postural control in that they add to the conscious body picture (18).

2.6.3 Vestibular System

Two other receptors which may be of importance in postural control are the semi-circular canals and otoliths which are found in the vestibular system.

(a) Semi-circular canals consist of three mutually perpendicular fluid filled circular canals. Two sets are present, one on either side of the head. Recent work has shown that the canals may be thought of as responding to the angular velocity of the head. However, since the canals have a threshold (about 2 deg/sec) above that found in quiet standing, it is not very likely that the semi-circular canals are of much importance in quiet standing (43, 55, 76).

(b) Otoliths are the other important sensory receptors present in the vestibular system. They consist of essentially a mass supported on a viscoelastic bed. It is felt that they may be thought of as responding to the net acceleration vector. Since the otolith threshold is about 0.005g (or about 1/4 degree of lean)
they could conceivably be of great importance in postural control \(^{(43, 55, 76)}\).

(c) Central Connections of Vestibular System. Vestibular inflow is known to have connections with postural control centres and to have distinct effects on postural reflexes \(^{(18, 58)}\). However, the importance of vestibular input in postural control should not be emphasized since it is possible for humans with no vestibular system to maintain themselves erect with little or no trouble \(^{(35, 52)}\). Of course, humans are highly adaptive and hence may not be operating in the same mode in these two cases.

2.6.4 Vision

Visual clues are without doubt very important in normal control of posture. However, the effect is difficult to quantify since it depends upon the contents of the field of vision \(^{(5, 6)}\).

2.6.5 Higher Centres

The neuroanatomy of higher centres concerned with postural control is too complex to be of much use in the analysis of the postural control system. It is of interest to note that there is evidence that the anatomical pathways used by the motor control system are different from the pathways used by the PCS. The reticulospinal and vestibulospinal pathways are the ones usually thought to be used in the control of posture \(^{(16, 18)}\).

Much is known about the effects of lesions at various levels of the CNS. The details are not of interest but it is worth noting that some input from higher levels is necessary for maintenance of posture despite the many peripheral reflex mechanisms \(^{(16, 18, 35, 58)}\).

2.7 MATHEMATICAL MODELING OF THE PCS

There has been some attempt at mathematical modeling of the PCS. The work of Houk \(^{(28, 47)}\) and Stark \(^{(1)}\) has dealt with the postural reflex occurring in wrist rotation. Mathematical models for each component were integrated into a reflex control model and parameter values were obtained by matching model responses to the responses of human subjects. Agarwal \(^{(2)}\), working with Stark, investigated the tendon jerk response of the ankle muscles and concluded that
the PCS of the ankle is non-linear and cannot be modeled as a simple second order system.

Brookhart's group is carrying out a large scale investigation of postural control in dogs. Although the work appears to be primarily physiological in nature, some attempts at modeling and simulation are apparently planned. At the present time, very little of the published work has dealt with modeling (11, 50, 51).

Nashner (52) appears to be the only worker to have modeled the PCS of humans. The model integrates models of the various receptors involved in postural control into an overall PCS model. Parameters were obtained by matching model responses to human subject responses for various experimental conditions. Gain control is postulated as a central mechanism and several different modes of control are proposed. This work appears to be the only attempt to deal with PCS in operation and it will be of interest to compare the results of this investigation to it.

2.8 SUMMARY

The present state of knowledge concerning the PCS may be summarized as follows:

1. Upright stance is a dynamic situation which is always accompanied by some small degree of sway, which depends upon many different influences.

2. All joints of the body are statically stable during normal quiet standing except the ankle, about which the gravity forces exert a net moment tending to rotate the body forward.

3. Electromyography has shown that the main muscular activity occurring during quiet standing is a low level phasic activity of the ankle muscles.

4. A very complex reflex mechanism employing various types of receptors is available to the body for use in postural control.

5. Signals from the supraspinal levels of the CNS are essential to normal postural control.

6. Present understanding of the PCS is essentially qualitative with very few workers presenting models of the PCS as a whole.

7. The best available model of the PCS is that presented in Nashner's thesis.
CHAPTER III

MODELLING THE POSTURAL CONTROL SYSTEM

3.1 INTRODUCTION

In this chapter the elements of the Postural Control System will be discussed in more detail than in Chapter II. Then an information flow diagram consistent with current ideas of postural control will be constructed using these elements. Next this information flow diagram will be simplified by making certain assumptions and imposing a number of restraints on the system. Finally, the resulting simplified model will be used as an aid in analysing the system and in planning an experimental investigation of it.

3.2 ELEMENTS OF THE POSTURAL CONTROL SYSTEM

3.2.1 Introduction

The elements of the PCS will be discussed in some detail in this section, emphasizing the mathematical description of them rather than qualitative physiology. No attempt will be made to incorporate the extremely complex neuromuscular connections identified in recent years (16, 18).

3.2.2 Nerve

Information is transferred throughout the CNS in the form of discrete pulses called action potentials. These travel along nerve fibers at speeds of more than 100 meters/sec. (75). The information is usually assumed to be coded by some type of pulse frequency modulation (39).

3.2.3 Skeletal Muscle

Skeletal muscles are the actuators of the PCS. They are made up of many functional subunits called motor units. Each motor unit consists of a motor neuron cell body located in the spinal cord, a motor axon leading from the cell body to the muscle and a number of muscle fibers innervated by the terminal branches of the motor axon. The muscle fibers of a given motor unit are dispersed throughout the entire muscle so that the action of a single motor unit does not have only a local effect. The tension exerted by a muscle can therefore be controlled by altering the pulse frequency in the motor axons to either a single motor unit or to
Muscle tension can thus be controlled over a large dynamic range with much sensitivity \((7,45)\).

The layout of a motor unit is shown schematically in Figure 3.1.

Quantitative studies of the behaviour of whole muscles have usually ignored this functional arrangement and treated the muscle as an actuator described by a set of empirical force-length and force-velocity curves. These two curves are non-linear and interdependent. The tension developed by a muscle is found to depend also on the mean input motor neuron pulse frequency and on the length of time for which the muscle has been active \((74)\).

Several investigators have fitted equations to the empirical curves in order to derive a linearized model for muscle \((4,45)\). The equation describing such a model is:

\[
T = T_o + \frac{K_1(f-f_o)}{1+T_3s} - K_m(L-L_o) - B_m s L \quad (3.1)
\]

where:

- \(T\) = muscle tension
- \(T_o\) = operating point tension
- \(f\) = mean input pulse frequency
- \(f_o\) = operating point pulse frequency
- \(L\) = muscle length
- \(L_o\) = operating point muscle length
- \(K_1\) = constant which depends on operating point
- \(K_m\) = "
- \(B_m\) = "
- \(T_3\) = "

This model is only valid for small variations in muscle length about a given operating point, and the parameter values depend upon this point. It should be noted that this formulation ignores the effects of fatigue which become important in very strong or prolonged contractions.
Figure 3.1: Schematic diagram of a motor unit.
3.2.4 Muscle Spindle

The muscle spindle is a complex sensory transducer whose response to changes in muscle length is controlled by two "γ" efferent signals. Furthermore, the two spindle afferents have similar but not identical responses.

The primary ending response is the only one to have been studied extensively. Recently McRuer et al (45) have presented a model of the primary ending response which takes into account the static "γ" fibers, but unfortunately the dynamic "γ" effects are not at present well understood. Their small signal transfer function for the muscle spindle is:

\[
\frac{f_{sp}(s)}{\gamma_s} = \frac{K_s(T_1s + 1)(L - \gamma_s)}{(T_2s + 1)} + \frac{C_1 \gamma_d}{aT_1s + 1}
\]  

(3.2)

where:
- \( f_{sp} \) = primary ending spindle output
- \( \gamma_s \) = static "γ" firing frequency
- \( \gamma_d \) = dynamic "γ" firing frequency
- \( L \) = muscle length
- \( a \)
- \( T_1 \)
- \( T_2 \) = constants which may be altered by \( \gamma_s \) and \( \gamma_d \).
- \( C_1 \)
- \( K_s \)

Other models have recently been presented which indicate that the spindle dynamics are somewhat more complex than indicated by the McRuer model (29, 60). However these models were based on work on deafferented preparations and so do not account for "γ" efferent effects.

3.2.5 Golgi Tendon Organs

Recent work by Houk (30, 31, 33, 34) has presented a detailed model describing the Golgi tendon organ response. The model consists of a linear system preceded by a saturation non-linearity and followed by a threshold device (see Fig. 3.2). Although the exact nature of the threshold and saturation non-linearities is not known, the transfer function of the linear system has been modelled in the form:
Figure 3.2: Schematic diagram of the Golgi Tendon Organ model.

\[ T(t) = \text{Muscle Tension} \]
\[ F(t) = \text{Tendon Organ Output Pulse Frequency} \]
\[ L(s) = \frac{H(s)}{V(s)} = A - \frac{B}{s+\alpha} - \frac{C}{s+\beta} \]  

where  
\( H(s) = \) output of linear system  
\( = \) input to threshold element  
\( V(s) = \) input to linear system  
\( = \) output of saturation element  
\( A, B, C \)  
\( \alpha, \beta \) = constants

For muscle tensions above threshold and for small changes in tension, the tendon organ response can be represented fairly accurately by the linear system response alone.

### 3.2.6 Ruffinï Endings

The only joint receptor whose response has been described analytically is the Ruffini ending. The response \((14, 58, 63)\) is highly non-linear and of the form

\[ f_r(t) = \psi_{st} + \sum_{i=1}^{3} \phi_i(t) \]  

where  
\( f_r(t) = \) Ruffini ending firing frequency  
\( \psi_{st} = \) static component of response  
\( \phi_i(t) = \) dynamic component of response.

In this representation the static component of the response depends directly on the joint position in a non-linear manner. A single receptor responds only over a small portion of the range of motion of the joint (Fig. 3.3). However the range of joint angle over which the ending responds is different for each ending.

The dynamic component of the response is dependent on both the joint velocity and the velocity history. It has been modelled as the sum of three integrals each of the form of Eq. 3.5.
Figure 3.3: Steady state discharge plots of five joint receptor units (redrawn from Eldred).
\[ \phi_i(t) = \int_{-\infty}^{+t} K_i V(T) e^{-T/T_i} dT \]  
(3.5)

where

- \( K_i \) = constants
- \( T_i \)
- \( V(T) = \) joint velocity at time \( T \).

Clearly this type of formulation is too complex to be of much use in neuro-muscular modelling. A simpler, linear model of Ruffini response is needed instead.

### 3.2.7 Semi-Circular Canals

The response of the semi-circular canals to angular rotation has been studied in some detail and linear transfer functions for their response are available \(^{(43, 55, 76)}\). The response of the semi-circular canals of man can be studied either through the subject's subjective feeling of rotation or by observing the visual nystagmus accompanying the rotation. It is presently agreed that the main dynamic characteristics of the canal can be modelled by

\[
\frac{f_{ci}(s)}{w_i(s)} = \frac{K_j s}{(s+d)(s+e)}
\]

(3.6)

where

- \( f_{ci} \) = output pulse frequency of canal in direction \( i \)
- \( w_i \) = angular velocity about \( i \) axis
- \( K_j, d, e = \) constants.

The semicircular canals display a threshold behaviour which for the horizontal canals is about two degrees per second. The majority of experimental work has been done on the horizontal canals, so little is known about the other two sets.

Some neural processing of the canal signal is known to occur before it is used by the CNS. The assumption of different neural processing provides a convenient explanation of why subjective and nystagmus methods of investigation give different results. Presumably some neural processing occurs before vestibular information is used by the PCS, but nothing is known about this as yet.

### 3.2.8 The Otoliths

The otoliths have received much less study than the semi-circular canals.
Otolith responses in man have been studied either through perceived linear acceleration (or degree of tilt) or through the counter-rotation of the eyes which accompanies linear accelerations \((43,76)\). Only the response to horizontal accelerations has received much study.

Currently it is thought that the otolith response may be described by a transfer function of the form

\[
\frac{f_{o_l}(t)}{F_i} = \frac{K_2(s + a)}{(s-b)(s-c)}
\]

where \(f_{o_l}(t) = i\) direction otolith response

\(F_i = \) net acceleration in direction \(i\).

\(K_2, a, b, c = \) constants.

The otoliths appear to have a threshold of about 0.005g which corresponds to an angle of tilt of about 1/4 of a degree.

3.2.9 Parameter Values

Values of the parameters of the models discussed were not given because the different receptors were not studied in the same species of animals. Some parameter values are available for all receptors but not for those of man.

Although it may be reasonable to assume that receptors will behave similarly in all mammals, it certainly is not plausible to assume that the actual parameter values remain the same. Therefore, present models may be used to improve the understanding of the system's function but parameter values must be obtained by fitting to experimental data.

3.2.10 Lumping of Elements

The peripheral receptors are quite numerous; for example, in the cat gastrocnemius there are more than 30 tendon organs and 80 muscle spindles \((18)\). Each of the receptors maintains a typical response but the actual parameter values for a given receptor vary quite a bit. Furthermore, each of these receptors has its own separate nervous pathways to the spinal cord and higher levels of the CNS.

This multiple unit-multiple pathway construction is typical of the CNS. However, any attempt to deal theoretically with this multiplicity is impossible in
all but the simplest of cases. The simplification usually made in modelling the neuromuscular system is to assume that the response of the receptor ensemble may be represented by the response of a single characteristic element. Similarly, it is assumed that the multiple pathways may be represented by a single characteristic pathway. The validity of these assumptions depends upon selecting receptor characteristics and pathways which are truly characteristic of the ensembles.

The lumping of multiple elements and pathways undoubtedly results in a loss of detail in any simulation of the system. However, it should not alter the basic functioning of the system and it makes a functional analysis of the system a great deal simpler. In fact, without making this assumption, it would be impossible to analyse the system at all. Note that the multiplicity of paths does control a single "lump", i.e. one limb. This seems to make the simplification even more reasonable.

3.3 INFORMATION FLOW IN THE POSTURAL CONTROL SYSTEM

3.3.1 Introduction

In this section the elements discussed above will be combined to construct an information flow diagram describing the functioning of the PCS. The system will then be seen to consist of a series of control levels, one nesting inside the other. This appears to be characteristic of most neuromuscular control systems. In constructing this information flow diagram, two assumptions will be made:

(i) Inhibition at a synapse will be represented as a subtraction and excitation by addition. Thus an inhibitory feedback is represented as negative feedback and an excitatory feedback by positive feedback.

(ii) The agonist-antagonist grouping of muscles about a joint will be considered as a single equivalent muscle. Nashner (52) has shown that this is a reasonable assumption for small signal models. Similarly the spindles and tendon organs will be collapsed into equivalent spindles and equivalent tendon organs. This is justified by the complementary manner in which the agonist and antagonist receptors and muscles are mutually innervated. This second assumption is commonly made in dealing with the neuromuscular system since it collapses two sets of unidirectional actuators and receptors into a single bidirectional set.
3.3.2 The Muscle Servo

Skeletal muscle is not a "good" actuator in the engineering sense since its tension depends on muscle length, contraction velocity and active time as well as upon the input motor neuron pulse frequency. Some type of tension feedback is obviously desirable to stabilize the output tension against these influences. The Golgi Tendon Organ appears to be ideally suited to provide this stabilization by providing a negative feedback loop as seen in the block diagram shown in Fig. 3.4. When drawn in this way the stabilizing effect of the Golgi Tendon Organ on muscle tension becomes clearly apparent. The delays in the system arise from the finite time required for an action potential to travel along a nerve fibre and are quite short.

If a load is added to the block diagram there is again an open loop (see Fig. 3.5). Note the load may consist of only the limb and the muscle itself. The muscle spindle apparently serves to close this loop and stabilize the joint position against disturbance forces. This can be clearly seen in Fig. 3.6.

At first sight it appears that the spindles provide positive feedback, but the loop gain is, in fact, negative since an increase in muscle tension produces muscle contraction and a decrease in load position. This negative feedback acts as a stabilizing influence which is one function of the muscle spindle.

The muscle spindle also provides a means of altering muscle position. This is made possible by the "γ" efferent signals which can alter both the static and dynamic parameters of the spindle. A change in the static spindle parameters provides a method of altering a joint position without changing the "α" input pulse frequency. Secondly, changing the dynamic spindle response provides a means of changing the effective damping of the system and hence the system dynamics. The "γ" control route is consequently considerably more versatile and useful than "α" control but due to a larger number of delays (see Fig. 3.6) the "γ" control system is slower than the more direct "α" control route. The longer delays arise because the "γ" control route uses longer and slower nervous tracts and has more synapses than the "α" route.

There is one other feedback loop at the peripheral level which may be involved in neuromuscular control - that involving the Renshaw Cell collateral...
Figure 3.4: Block diagram of the tension control loop. The muscle tension is stabilized by feedback from the Tendon Organ.
Figure 3.5: Block diagram of the tension control loop with load mechanics added.

Note that the load position is not stabilized by the tendon organ feedback.
Figure 3.6: Block diagram of the joint control servo. The load position is stabilized by the feedback from the muscle spindle.
path. The Renshaw Cell provides an inhibitory feedback from the "a" motor axon to the "a" motor neuron; this action can be altered by commands from higher centres in the CNS.

An overall block diagram describing all the important known feedback loops is shown in Fig. 3.7.

This set of three servo loops will be referred to from now on as a joint position servo and may be visualized essentially as in Fig. 3.8. There are three possible control inputs:
- Renshaw Cell control
- "a" motor control
- "γ" motor control.

The disturbance input is due to external forces exerted around the joints controlled by the servo. The three servo outputs consist of:
- joint position
- muscle spindle afferents to higher levels of the CNS
- tendon organ afferents to higher levels of the CNS.

3.3.3 Cutaneous Receptors

The joint receptors appear to have little or no function at the spinal level. However they are very important in conscious proprioception and are apparently responsible for conscious "body image" and so supply feedback for voluntary motor activities. This is shown in block diagram form in Fig. 3.9. Note that the peripheral receptor afferents are also fed to the CNS but do not reach as high a level as the joint receptor afferents. (That is, the spindle and Golgi tendon organ afferents do not reach consciousness.)

3.3.4 Vestibular System

The receptors of the vestibular system provide direct information about the angular velocity and the linear accelerations to which the head is subjected. These two inputs result from the combined action of the joint servos involved in posture and any disturbances so the vestibular receptors give an assessment of overall system performance. Postural reflexes due to vestibular inputs are known to exist for fairly large angular velocity or linear acceleration inputs. Some of these
Figure 3.7: Block diagram of the complete joint control servo including the Renshaw cell.
Figure 3.8: Diagramatic representation of the joint control servo.
Figure 3.9: Information flow between the peripheral receptors and the CNS.
reflexes appear to be of importance in maintaining the head as a stable platform for the eyes. The function of vestibular receptors in the maintenance of a quiet standing posture is not known.

The role played by the vestibular receptors is shown in Fig. 3.10.

3.3.5 Overall Information Flow Diagram

The elements and their interconnections described above may be synthesized into an overall information flow diagram for postural control such as shown in Fig. 3.11.

In this case it has been necessary to combine the joint servos and body dynamics, because the body dynamics are such that a change in state of one joint servo alters the load seen by the other joint servos. The complex feedbacks resulting from this are therefore represented by combining the servo and dynamic elements into a single unit.

The block diagram of Fig. 3.11 is deceptively simple since the complexities arising from the many joints involved are not shown. Furthermore, the extremely complex body dynamics are not shown either. These complexities make any attempt to deal with the whole system analytically all but impossible, at least at present.

3.4 A SIMPLE MODEL FOR THE POSTURAL CONTROL SYSTEM

3.4.1 Introduction

The postural control system, described in some detail so far in this chapter, is very complex despite the many simplifying assumptions already made. There are several restraints which may be imposed upon the system which result in further simplifications. The development of methods to impose these restraints on subjects will form an important part of the experimental program and will be described later. After the restraints are imposed, the form of the block diagram will be modified through standard block diagram transforms to further simplify the system. This simplified block diagram will then be used as an aid in planning the experimental program.
Figure 3.10: Information flow between the peripheral receptors, the vestibular system and the CNS;
Figure 3.11: Information flow in the Postural Control System.
3.4.2 Body Dynamics

The first simplification that can be made to the body dynamics is to assume that all postural motions are in the anterior-posterior plane. Support for this assumption is found in the literature where the a-p motions are consistently reported as being much larger than the medial-lateral motions. Further support for the two dimensional assumption is found from a study of the anatomy which shows that the human posture is inherently much less stable in the a-p plane than in the medial-lateral plane.

Further simplifications can be made to the body dynamics if some of the joints can be restrained from moving. It is known from the literature that most of the postural muscular activity occurs around the ankle (7, 37). Consequently the motions about other joints are probably almost entirely passive. Therefore if motion about these passive joints is prevented, the basic mode of postural control during quiet standing should not be altered. Of course these restrictions will change the control methods used in emergency situations. However it will be assumed that it is possible to remove motion about all joints but the ankle. A free body diagram for this case is shown in Fig. 3.12.

The equation of motion for this situation is

\[ 1 \ddot{\theta} - Rmg \sin \theta = T_A \]  

which can be simplified for small angles by putting

\[ \sin \theta = \theta \]  

(3.9)

giving

\[ 1 \ddot{\theta} - Rmg \theta = T_A \]  

(3.10)

which may be expressed as a transfer function of the form

\[ \frac{\theta(s)}{T_A(s)} = \frac{1}{s^2 - Rmg} \]  

(3.11)

The torque at the ankle \((T_A(s))\) arises from both passive joint forces and active muscle contractions.

The model of body dynamics seen in Equation 3.11 is a great deal simpler than the model of an unrestrained body (which consists of at least 7 simultaneous second order differential equations even if the spine and head are assumed to be
$A = \text{center of rotation of the ankle.}$

$m = \text{mass of the subject.}$

$R = \text{distance from the center of rotation of the ankle to the center of gravity of the subject.}$

$T_a = \text{torque about the ankle.}$

$\theta = \text{angle of lean of subject.}$

$I = \text{moment of inertia of subject about the ankle axis.}$

Figure 3.12: Free body diagram of an exoskeletally restrained subject.
one lump). This degree of simplification makes it worthwhile to spend a good deal of effort in experimentally imposing restraints so that Eq. 3.11 can be considered to be a realistic model.

3.4.3 Further Assumptions

In addition to the simplifications to body dynamics the following further simplifications will be made in attempting to obtain a tractable model of the postural control system.

(i) The three muscle agonist-antagonist group acting around the ankle will be represented by a single equivalent muscle capable of producing bi-directional torques. Furthermore, the receptors of these muscles will be modelled by a single bidirectional equivalent receptor of each type.

This assumption not only reduces the number of elements to be considered but removes a number of non-linearities as well.

(ii) Only the outputs of the receptors about the ankle will be considered. Since motions and muscular activities around joints other than the ankle are assumed to be very small, their receptor outputs must also be small.

(iii) Information transferred along nerve fibres will be assumed to be coded as the instantaneous pulse frequency.

(iv) The CNS control will be assumed to consist solely of the "α" and "γ" efferent signals to the ankle muscles. Renshaw Cell control will be ignored.

(v) The joint receptors will not be considered since postural control is believed to be purely reflexive during quiet standing and the joint receptors are primarily used for conscious sensation.

(vi) Only the response of the semicircular canals to rotation about an axis through the ankles and the response of the otoliths to a linear acceleration in the a-p direction will be considered. All other motions will be assumed to be too small to be of interest.

(vii) The "dynamic" spindle efferents will be ignored.

3.4.4. The Simplified Postural Control Model

Using the simplifications described in the foregoing sections, the information flow diagram of Fig. 3.11 can be redrawn as a block diagram (Fig. 3.13),
which in turn can be redrawn as the much simplified diagram of Fig. 3.14.

This diagram may be transformed using standard block diagram methods into the simpler form of Fig. 3.15. Obviously C(s) is theoretically a very complex controller since it includes not only receptor dynamics but the muscle dynamics and CNS influences as well.

3.5 DISCUSSION

The discussion and analysis pursued in this chapter forms the basis for the experimental program to be discussed in Chapters IV, V and VI. In addition, the simple linearized model provides a starting point in seeking to identify the PCS. It is to be expected that the PCS will be considerably more complex than the simple model would imply. The increased complexity may be due to the involvement of more receptors (i.e., S.C.C. and otoliths), the existence of nonlinearities (i.e., thresholds, saturation) or something else (such as the control of the spindle gain by higher centres as suggested by Houk\textsuperscript{(28)} and Nashner\textsuperscript{(52)}. Despite these possibilities, the simple model is invaluable as a basis on which to build.
Fig. 3.13 Linear model of the PCS in an exoskeletonally restrained subject
Fig. 3.14  Simplified model of the PCS in an exoskeletonally restrained subject
Figure 3.15: Simplified model of the PCS structure.
CHAPTER IV
EXPERIMENTAL APPARATUS

4.1 INTRODUCTION

The simplified block diagram describing the PCS implicitly defines the experimental apparatus required to investigate the system. This apparatus will be described in this chapter and consists of:

(i) a force plate to measure the torque developed by the muscles about the ankle ($T_A$),
(ii) a linear displacement transducer for use in defining the subject's body position ($\theta$),
(iii) an orthotic exoskeleton to restrict motion about all joints but the ankle joint,
(iv) electromyographic apparatus for use in assessing muscular activity about the ankle,
(v) a method of applying a standard input torque to the subject, and
(vi) a device to control the experiment and record the results.

4.2 FORCE PLATE

4.2.1 Introduction

A general purpose force analysis platform was designed and built by the author in collaboration with Dr. D. L. Burke. This force plate was envisioned as a general purpose biomechanical research tool and is therefore somewhat more sophisticated than necessary for the investigation described here.

4.2.2 Design and Construction

The force plate (see Fig. 4.1) consists of a three-foot square plate supported by six load cells which are in turn mounted on another plate. Both plates are made of 3/4" thick magnesium to which a triangular frame of aluminium I beams has been bolted to provide added stiffness. Magnesium and aluminium were used in order to minimize the weight of the upper plate and hence keep the plate natural frequency as high as possible.
Figure 4.1: The Force Plate.
The load cells are arranged in a triangular array with the three vertical cells at the vertices and the three horizontal cells forming the sides of an equilateral triangle. The cells are all mounted on flexure rods, which are stiff axially but very flexible in bending. The flexures serve to eliminate cross coupling between the vertical and horizontal cells due to bending moments. The flexures were designed so that less than 1% of the force perpendicular to the flexure axis is taken up by bending of the flexures. Moments of this size will not alter the output from the cells used. Since the flexures are linear, the forces that they do absorb can easily be accounted for during calibration. BLH load cells were selected: 1000 lb. capacity for the vertical cells and 500 lb. capacity for the horizontal cells. The high capacity load cells were selected for the vertical measurement as a protection against shock overloads and to obtain a stiff support for the plate. The stiffness of the plate mounting combined with the mass of the upper plate determines its natural frequency. Note that while semi-conductor load cells are much stiffer, they were not used due to their non-linear output and temperature dependence.

More sensitive cells were used for the horizontal measurements because the size of these forces was expected to be much smaller than that of the vertical forces.

4.2.3 Load Cell Electronics

The load cells are of the foil strain gauge type* and have a low output so that amplification is needed to give outputs suitable for further analog processing.

The S.E. Labs 4000 carrier amplifier system was selected for use. This system energizes the load cells with a 3 KHz carrier signal. The amplitude modulated load cell outputs are amplified by six A.C. amplifiers and this signal is then demodulated in a phase sensitive manner to give a D.C. output in the range \( \pm 1.4 \) volts. The amplifiers are sensitive enough to give a full scale output for a 0.01% bridge unbalance and have infinite gain variability up to this level. The amplifiers also contain circuitry to allow both resistance and inductive unbalances to be removed.

* in a full bridge circuit
The S.E. system also contains calibration and monitoring circuitry which allows the amplifiers to be balanced and their gains set easily.

The amplifiers have an optimum frequency response (flat from 0-1750 Hz) when feeding a 10K load. The amplifier outputs were therefore buffered from the variable impedance analog computation circuits by means of six operational amplifier followers with a fixed 10K input impedance.

The amplifiers in combination with the least sensitive load cells are capable of producing a full scale (1.4 volt) output for a load of less than 5 pounds and a noise level of less than 3 mv. This sensitivity is far in excess of anything required for these studies but could be of use in other studies - for example, balistocardio-graphic studies.

4.2.4 Analog Preprocessing

The amplified load cell outputs are not directly useful until they have been related to the plate geometry. The reference axis selected is shown in Fig. 4.2 where the origin (0) is near the centre of the plate. The variables of interest which must be calculated from the load cell outputs are:

(i) \( F_x \) = total force in \( x \) direction,
(ii) \( F_y \) = total force in \( y \) direction,
(iii) \( F_z \) = total force in \( z \) direction, (total vertical force),
(iv) \( M_x \) = net moment about \( x \) axis,
(v) \( M_y \) = net moment about \( y \) axis,
(vi) \( M_z \) = net moment about \( z \) axis.

At times it may be useful to express \( M_y \) and \( M_x \) respectively in the slightly different form of

(vii) \( X_p \) = effective \( x \) coordinate of \( F_z \),
(viii) \( Y_p \) = effective \( y \) coordinate of \( F_z \),

where \( X_p \) and \( Y_p \) are defined by the relations

\[
X_p = \frac{M_y}{F_z} \quad (4.1)
\]
\[
Y_p = \frac{M_x}{F_z} \quad (4.2)
\]
Figure 4.2: Force plate reference axes.
The detailed derivation of the equation relating these variables to the load cell outputs is found in Appendix A. These algebraic relations have been implemented by means of simple operational amplifier circuits which are patched together on Philbrick-Nexus operational manifolds, using Philbrick P85AU operational amplifiers. The divisions required to calculate $X_p$ and $Y_p$ are done by two Philbrick SPM1A multiplier-divider units.

4.2.5 Calibration

Static calibration of the system was necessary to account for any forces absorbed by flexures and to ensure that the system is both linear and repeatable. This was done by first setting the amplifier gains at about the correct level, calibrating the plate and then resetting the amplifier gains to give a convenient output scale.

Calibration loads were first applied in gradual steps and then removed in the same steps to provide a check on both the linearity and the repeatability of the system.

The vertical calibration loads were easily applied by using dead weights since only positive loads were of interest. The horizontal loads were applied in both directions through a lever arrangement.

While the plate was being calibrated in one direction, the other two force outputs were monitored to check for cross coupling effects. In all cases these effects were completely negligible.

Calibration curves for $F_x$, $F_y$, $F_z$ are shown in Figures 4.3, 4.4, and 4.5.

No attempt was made to calibrate the other outputs since they are derived from the same variables which were calibrated through $F_x$, $F_y$, $F_z$, and so do not need direct calibration.

The moment outputs were of course checked in an approximate manner to ensure that there were no patching errors.

Dynamic calibration of the plate was attempted by applying an impulsive load and observing the response. Typical response curves are shown in Fig. 4.6, 4.7, 4.8 for shocks applied in the $x$, $y$, $z$ directions. These complex responses
Figure 4.3: $F_x$ calibration curve.
Figure 4.4: $F_y$ calibration curve.
Figure 4.5: $F_z$ calibration curve.
Figure 4.6: $F_x$ response of the force plate to a shock load applied in the $X$ direction.

Time (milliseconds)

Figure 4.7: $F_y$ response of the force plate to a shock load applied in the $Y$ direction.

Time (milliseconds)
Figure 4.8: $F_z$ response of the force plate to a shock load applied in the Z direction. (milliseconds)
were treated as damped second order responses to provide approximate values for natural frequency and damping ratios. The values obtained were:

<table>
<thead>
<tr>
<th>Direction</th>
<th>Natural Frequency ($W_n$)</th>
<th>Damping Ratio ($\xi$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>x</td>
<td>168 Hz</td>
<td>.20</td>
</tr>
<tr>
<td>y</td>
<td>105 Hz</td>
<td>.60</td>
</tr>
<tr>
<td>z</td>
<td>200 Hz</td>
<td>.19</td>
</tr>
</tbody>
</table>

These figures apply for the plate supporting a 200 lb. man but the differences between the loaded and unloaded plates were small probably as a result of the loose coupling between the man and plate. These natural frequencies are well above any expected postural frequencies required for accurate measurements. However they do imply that some caution is necessary in studying impulsive type of loading as in fast walking or running.

The force plate electronics were set up to have the following output scales for work described in the thesis.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Scale</th>
<th>Noise Level</th>
<th>Accuracy</th>
</tr>
</thead>
<tbody>
<tr>
<td>$F_x$</td>
<td>10mv/N</td>
<td>5mv</td>
<td>±1/2N</td>
</tr>
<tr>
<td>$F_y$</td>
<td>10mv/N</td>
<td>5mv</td>
<td>±1/2N</td>
</tr>
<tr>
<td>$F_z$</td>
<td>10mv/N</td>
<td>5mv</td>
<td>±1/2N</td>
</tr>
<tr>
<td>$X_p$</td>
<td>1v/cm</td>
<td>50mv</td>
<td>±.5mm</td>
</tr>
<tr>
<td>$Y_p$</td>
<td>1v/cm</td>
<td>50mv</td>
<td>±.5mm</td>
</tr>
<tr>
<td>$M_z$</td>
<td>10v/N-m</td>
<td>50mv</td>
<td>.5N-cm</td>
</tr>
</tbody>
</table>

The noise levels quoted were observed experimentally.

4.2.6 Calculation of $T_A$

The output of the force plate may be easily related to the torque about a subject's ankle joint if the subject's ankle axis is aligned parallel to one of the plate axes. From Fig. 4.9 we see that

$$T_A = F_z(X_p - X_A)$$

(4.3)

and that if the ankle axis is along the $Y$ axis of the plate, then

$$T_A = F_z X_p$$

(4.4)
Figure 4.9: Free body diagram of ankle mechanics.
or more simply

\[ T_A = -M'y \]  

(4.5)

4.3 POSITION TRANSDUCER

A Transducer Controls Corporation Type TCC-PT-101-10A linear displacement transducer with a 10 inch travel was used to measure the position of the subject. The transducer consists of a wire wound onto a constant force spring motor. Extension or retraction of the wire turns a rotary potentiometer attached to the spring motor resulting in a wiper position which is proportional to the displacement of the wire. The retraction force of the unit due to the spring motor is quite small (approximately 6 oz) and remains almost constant throughout the range of motion of the unit.

The potentiometer when excited from a +15 volt power supply had an output of 1.35 volts/inch with a variable output impedance. This output was fed into an operational amplifier circuit which allowed steady state offsets to be subtracted out and also provided for additional gain if required. This circuit also converted the variable output impedance potentiometer output to the small constant output impedance of the operational amplifier. A circuit diagram for this is shown in Fig. 4.10.

This linear displacement transducer can be used to measure the angular displacement of a subject who is only free to move about the ankles. This upright pendulum situation is shown in Fig. 4.11.

For a linear displacement transducer attached to this pendulum at a height \( h_0 \) above its center of rotation the displacement \( dL \) resulting from a rotation of \( \theta \) radians is approximately

\[ dL = L_0 - L_1 \]

\[ \approx h_0 \sin \theta \]

or taking \( \sin \theta \approx \theta \) we get

\[ dL \approx h_0 \theta \]  

(4.7)

or

\[ \theta \approx \frac{dL}{h_0} \]  

(4.8)
Fig. 4.10: Circuit Diagram of Position Transducer Electronics

Fig. 4.11: Position Transducer Geometry
which gives the angle of lean as a multiple of linear displacement. The amplifier gain \( \frac{R_f}{R_a} \) was set to give an output scale of 0.04 rad/volt or 25 volt/rad. This scale was found to give convenient output levels for the experiments conducted.

4.4 ELECTROMYOGRAPHY

4.4.1 Electrodes

The electrodes used in the electromyographic studies were Beckman surface electrodes which were used instead of wire electrodes for three reasons:

(i) the muscles to be studied (soleus, gastrocnemius and tibialis anterior) are superficial muscles upon which surface electrodes can easily be attached,

(ii) surface electrodes pick up signals from a larger volume of muscle than do wire electrodes and so give a more representative idea of overall muscle activity, and

(iii) wire electrodes are less comfortable and more difficult to apply than surface electrodes.

The skin was shaved and then washed with alcohol before the electrodes were applied to insure a good electrical contact. The electrodes were attached to the skin with adhesive collars also supplied by Beckman. Good contact between skin and electrode was further ensured by filling the electrodes with a conducting electrode jelly.

4.4.2 Amplifiers

The low level signals from the surface electrodes were brought through shielded cables to Tektronix type 122 differential preamplifiers. Their gain (1000) was adequate to raise the EMG signals to a level suitable for further analog processing.

The amplifiers also contained a set of variable first order high pass and low pass filters for use in reducing noise levels. Best results were obtained with the high pass filter break point set to 8 Hz and the low pass filter break point set to 1 kHz.
4.4.3 Rectification and Filtering

The amplified EMG's were then passed through an analog network to obtain the envelope of the EMG. This type of processing is quite common in electromyographic work and gives a much better idea of muscular activity than does the raw EMG.

The analog circuit used was developed at the BioMedical Engineering Unit by H.H. Kwee and makes use of two integrated circuit operational amplifiers. The entire circuit is mounted on a small card which is interchangeable with the operational amplifiers in the Philbrick operational manifolds. This set-up provided a cheap and simple method of processing EMG data.

The circuit first high pass filters the input to reduce any noise resulting from low frequency body motions (movement artifacts). The signal is then full wave rectified and smoothed by a low pass filter. Both the high pass and low pass filters are simple first order filters with variable break points. A low pass break point of 24 Hz and a high pass break point of 100 Hz were found to give the best results.

A detailed circuit diagram of this network is provided in Fig. 4.12.

4.4.4 Electrode Placement

Two electrodes spaced several centimeters apart were placed near the surface of each of the three muscles studied. Electrode locations were chosen such that there were no other nearby muscles to interfere.

Functional tests were then done to ensure that the electrodes were correctly placed and functioning. This was quite easy to do since the tibialis flexes the foot while the soleus and gastrocnemius extend it. Furthermore, it was found that the soleus was alone active in slight extension with the gastrocnemius becoming active with increased extension. This functional difference made it possible to ensure that the three electrode sets were in fact picking up three separate signals and that there was no cross coupling between them. In all cases it was found to be possible to pick up separate signals from each of the three muscles.

A common ground electrode was used for all three electrode sets and was placed away from them.
Capacitor $C_1$ determines the high pass break point

Capacitor $C_2$ determines the low pass break point

Fig. 4.12: Circuit Diagram of the EMG Processing Card.
4.5 EXOSKELETON

The exoskeleton used to restrict motion at all joints but the ankle was a combination of commercial orthotic devices and individually fitted plaster casts.

The device consists of a cervical brace attached to a set of long leg braces by stainless steel rods. The long leg braces were fixed firmly to the subject by means of plaster casts encasing the subject's legs and the brace. These casts were fitted prior to the experiment, cut in half and removed. During the experiment the casts were held on the subject by means of velcro straps. Holes were cut in the casts at appropriate places to allow the EMG electrodes to be applied.

Additional rigidity was added to the upper portion of the exoskeleton by an aluminium bar running along the front of the brace. The bar was attached to a dental bite plate in the subject's mouth and to the pelvic girdle of the long leg braces.

Although the exoskeleton could by no means remove all motion from the restrained joints, it did greatly restrict it. Further reduction in the motion was possible only at the cost of greatly increased discomfort for the subject and was not therefore attempted.

4.6 STIMULATOR

The device used to apply disturbances to the subjects consists of a four foot high dexion framework to which an aluminium plate can be bolted at various heights. A solenoid and pulley are mounted on this plate. A wire runs from the pelvic girdle of the exoskeleton through a hole in the plate over the pulley and is attached to one end of a spring. The other end of the spring is connected to the core of the solenoid.

In use, the wire is adjusted so that the core of the solenoid hangs about 1 inch out of the solenoid coil. Guide rails have been added to the solenoid to keep the core aligned with the coil. Stimuli are produced by applying current to the solenoid, causing the core to be pulled into the solenoid body thus extending the spring and producing a force. Since linear springs were used,
the force produced will be proportional to the deflection of the spring. If the current to the solenoid is removed before the subject moves appreciably, the force will quickly fall off and the stimulus applied will have the form of a pulse.

The stimulator also contains a logic device which produces a logical high when the solenoid is active and logical low when it is inactive. The anterior (front) stimulator produces a positive logic level while the posterior (rear) stimulator produces a negative logic level.

The stimulators were calibrated by attaching them directly to the force plate and then applying a pulse. Responses were found to be pulse shaped and a typical pulse is shown in Fig. 4.13. Although pulse shaped, the timing of the force is substantially different from that of the logic signal since the force is not fully developed until at least 50 msec after the current is applied to the solenoid. Furthermore, it takes an additional 25 msec for the force to decay after the current is turned off. This must be remembered when considering the timing of the observed responses.

The oscillations occurring at the start and the end of a pulse are of too low a frequency (≈30Hz) to be due to oscillation of the force plate and must therefore be due to the spring in the device.

In the neutral position the stimulator applies a force of 3 Newtons to the subject, due primarily to the spring return of the solenoid. Since the force will be applied to the front and back of the subject by the two stimulators, no net torque will result. When activated, the stimulator applies a force of 18N to the subject which in the experimental situation would represent a net force change of 15N. The actual torque applied to the subject will depend upon the height at which the force is applied and hence will vary from subject to subject.

Although quite crude, the stimulators are certainly capable of applying a standard pulse shaped disturbance of force to the subjects.
Figure 4.13: Stimulator calibration curve.
4.7 PDP-12

The apparatus described above was interfaced with a PDP-12 computer to permit computer control of experiments and digital data acquisition. Interfacing was fairly simple since the PDP-12 is a laboratory-oriented machine and is equipped with the following interface devices:

(i) Six program controllable relays for use in the control of equipment external to the PDP-12.

(ii) A ten bit analog to digital converter operating over the range ± one volt with a conversion time of less than 20 μsec. A multiplexer employed with the A-D converter provides access to 32 separate analogue channels.

(iii) A program controllable real time clock allows events (i.e. sampling from an A-D channel) to be accurately spaced in time.

(iv) A two channel D-A converter and character generator which drives a CRT, allows on-line monitoring of acquired data.

(v) Two Linc tape drives which provide a convenient means of storing acquired data.

The interface therefore consisted simply of:

(i) Eight shielded cables leading from the A-D converter inputs to the instrumentation rack. Eight potentiometers were installed on the rack to provide a means of attenuating the ±10 volt range of instrumentation to the ±1 volt range of the A-D converter.

(ii) The three connections from each of the six computer relays were also led to a panel on the instrumentation rack.

In addition, two heavy duty relays were slaved to two of the PDP-12 relays to carry the large currents required by the stimulator solenoids.

(iii) Finally a program was written to control the experiment and data acquisition. This will be discussed in more detail later in the thesis and in the appendices.

A block diagram of the information flow to and from the PDP-12 is shown in Fig. 4.14.
Figure 4.14: Schematic diagram of the experimental set up.
CHAPTER V

EXPERIMENTAL SET UP AND PROTOCOL

5.1 INTRODUCTION

The experimental program consisted of four similar experiments, three of which were conducted upon the same subject while the fourth used a different subject. The experimental set up and the protocol used in this program is described below.

5.2 EXPERIMENTAL SET UP

Subjects wore only shorts, a tee shirt and had bare feet. The EMG electrodes were applied to the soleus, one head of the gastrocnemius, and the tibialis anterior muscles of the subject's right leg in the manner previously described. A functional check was then made to establish that usable signals from all three muscles were present.

The subject was fitted into the exoskeleton. Considerable care was necessary in order to restrain the subject effectively without causing undue discomfort. Fitting time was reduced as much as possible by tailoring the exoskeleton to the subject prior to the experiment. Once the exoskeleton had been applied, its effectiveness was assessed. In all cases it was evident that the subject's freedom of motion was greatly reduced, especially in the anterior-posterior (a-p) plane. The small freedom of motion which remained could not be eliminated without undue discomfort to the subject.

The exoskeletonally restrained subject was then carefully positioned on the force plate with heels placed symmetrically about the \( X \) axis of the force plate, approximately 20 centimeters apart, and spread at 45 degrees. The axis of rotation of the subject's ankles (which corresponds to a line through the malleoli) was aligned as closely as possible with the \( Y \) axis of the force plate. In this stance, the \( X_p \) output of the force plate multiplied by total vertical force (\( F_z \)) is proportional to the torque acting about the ankle (\( T_A \)) due to ground reaction forces. Since variations in total vertical force are assumed to be small, the \( X_p \) output is
proportional to ankle torque. Furthermore the stance is comfortable and easily reproducible.

The stimulators were positioned in front of and behind the subject so that their line of action coincided with the force plate X axis. The height of each device was adjusted so that when attached to the subject their line of action was horizontal.

The simple design of the stimulating devices made it necessary to adjust the attachments to the pelvic girdle of the exoskeleton in such a manner that both solenoid cores hung the same distance out of the solenoid coils. It was therefore necessary to define a normal angle of lean for the subject; that is, an angle at which the subject was comfortable and stable and to which he could easily return after a perturbation. Once this angle had been determined it was relatively easy to adjust the stimulator attachments.

The position transducer was then attached to the exoskeleton and the circuit adjusted to give a null output for the subject's normal angle of lean. All angle measurements were therefore made relative to this arbitrary zero point.

The subject's arms were strapped to his sides with a velcro belt to prevent involuntary arm motions. Finally the subject was asked to close his eyes and lock his knees in full extension. The experiment was then ready to proceed.

Figure 5.1 shows a subject fully prepared for an experiment to begin.

5.3 EXPERIMENTAL PROTOCOL

The experiment was controlled and data were acquired by means of a PDP-12 digital computer which was interfaced with the experimental set up as described in Chapter IV. Data were first stored in core and then transferred to digital tape for permanent storage.

At the time of these experiments, the PDP-12 was equipped with only 4K of core. Since the controlling program takes up 1K, only 3K was available for data storage.

This relatively limited storage area made it necessary to minimize the number of variables to be recorded. In the experiments discussed in this thesis,
Figure 5.1: Subject prepared for an experiment.
the following six variables were recorded:

(i) Stimulus - a logic level which defines the stimulus applied to the subject. A positive level is present when current is applied to the anterior solenoid and a negative level is present when current is applied to the posterior solenoid. There is no output when neither solenoid is active. This signal is obtained from the stimulator control logic.

(ii) $Y_p$ - the medial-lateral position of the effective centre of pressure of the subject.

(iii) $X_p$ - anterior-posterior position of the subject's effective centre of pressure. As shown above, this is directly proportional to the torque acting at the ankle joint ($T_A$).

(iv) $\theta_A$ - the angular displacement of the subject from the normal angle of lean (assuming the subject sways stiffly), measured by the displacement transducer attached at chest level.

(v) Extensor EMG - the sum of the filtered EMG's of the two ankle extensors (the soleus and gastrocnemius). The two EMG's were summed to reduce the number of variables to be recorded.

(vi) Flexor EMG - the filtered EMG from the tibialis anterior.

The available core allowed 512 samples of each of these six variables to be taken at one time. Consequently the data were acquired in discrete sets, each consisting of 512 samples of six variables. The protocol used was similar for all four experiments and was made up of three sections:

(i) Unperturbed Standing - Ten sets of data were acquired while the subject was standing quietly. A relatively slow sampling rate (100/sec) was used and sampling was only initiated when the subject was observed to be close to the normal angle of lean and not swaying perceptibly. The subject was not informed at what times samples were being taken.

This section was intended to provide baseline data necessary for the correct interpretation of the perturbed standing data.

(ii) Forced Sway - Ten sets of data were then acquired with the subject first standing quietly and then told to sway. Five sets for anterior (forward)
swaying and five sets for posterior (backward) swaying with a sampling rate of 100/sec were taken.

This data was acquired for possible use in identifying the parameters of the body dynamics, but was not subsequently used and so will not be discussed further.

(iii) Perturbed Standing - The final portion of the experiment consisted of gathering data describing the responses of the subject to perturbing pulses of force of two different durations.

For each pulse length ten sets of data were taken, consisting of five responses to anterior pulses and five responses to posterior pulses. Each set of data was initiated when the experimenter noted that the subject was at an angle of lean close to normal and no perceptible sway was present. In the first two experiments a total of twenty sets of data were taken for each stimulus, half with a sampling rate of 100/sec and the other ten sets at 200 samples/sec. The last two experiments used only the 200/sec sampling rate and only ten sets of data were taken.

The simple nature of the stimulator made it very difficult to alter the size of the force pulse once the experimenter had begun and for this reason pulses of different durations were used as the different inputs.

Due to the exoskeleton and the involved nature of setting up, subjects tended to tire quickly. Since fatigue can seriously alter the way in which a subject responds, each experiment was limited to two different pulse lengths (usually 250 msec and 500 msec) and then terminated.
CHAPTER VI

DATA PROCESSING AND EXPERIMENTAL RESULTS

6.1 DATA PROCESSING TECHNIQUES

All data from the four experiments were stored in digital form on Linac tape and hence were easily available to the PDP12. Digital methods were therefore used to prepare the experimental data for analysis and simulation use.

(i) Plotting

A Hewlett Packard Model 7035B X-Y plotter was interfaced with the PDP-12 D-A converter and controlled by the program "PLOTTER". This program allows data stored on Linac tape to be plotted as a function of time and hence provides a means of obtaining hard copies of experimental and processed data. All data presented in this thesis was plotted in this way. Details of the plotting program and interface are given in the appendix.

(ii) Averaging

It was thought desirable to average a number of responses to the same stimulus in order to reduce the importance of random variations in the response. The program "AVERAGE" was used to do this, and at the same time to calculate the mean absolute deviation of the averaged responses to provide a measure of the variability between the responses. Variations due to initial conditions were eliminated by adding an offset to each curve thus making all responses start from a zero initial condition.

The procedure is demonstrated in Fig. 6.1 where part (A) shows five torque responses to a 500 msec pulse superimposed upon one another. Part (B) shows the mean of the five curves and the mean plus and minus the mean absolute deviation as calculated by the program "AVERAGE". Details of the program are given in the appendix.

(iii) Filtering

The angular position ($\theta_A$) signal was obscured by high frequency noise
Figure 6.1: (A) Five superimposed responses to 500 msec posterior stimuli. (B) Solid line – mean value of the five responses in A. Dashed lines – mean value plus and minus one mean absolute deviation.
originating from wiper bounce in the position transducer potentiometer. A three point symmetrical digital filter was applied to the data to remove this noise. This filter has the advantage of attenuating the high frequency noise without introducing any phase lag to the signal. A data processing program "PROCESS" was written to implement this filter and is discussed in more detail later in this thesis.

6.2 QUIET STANDING DATA

The quiet standing data were simply plotted in the same form as obtained experimentally. A typical set of curves describing quiet standing are presented in Figure 6.2. The following observations were made from inspection of the quiet standing data:

(i) The effective centre of pressure, given by \( X_p \) and \( Y_p \), varies continually about a mean position with an amplitude of less than 0.5 cm. At times the mean anterior-posterior position of the centre of pressure (\( X_p \)) drifts slightly but the mean medial-lateral position (\( Y_p \)) remains constant. These variations are without doubt the postural tremor or sway reported in the literature (24, 62, 70).

(ii) The subject's angle of lean (\( \Theta_A \)) remains constant most of the time. Sporadically a slow drift linked to the \( X_p \) drift is observed. This is undoubtedly the low frequency component of postural sway.

(iii) Two different patterns of EMG activity were observed in the four experiments.

In the first two experiments a continuous low level activity was observed to be present in the ankle extensors (soleus, gastrocnemius). Unfortunately, EMG's were not obtained from the flexor muscles and it was not possible to draw conclusions about static flexor activities.

In the final two experiments EMG's were obtained from both the extensor and flexor muscles of the ankle. In these experiments a continuous low level EMG activity was observed in the flexor muscles while no activity was observed in the extensor muscles.

The EMG levels observed in quiet standing were very much smaller than
Figure 6.2: Typical recordings taken during quiet standing.
those observed during active contraction of the muscles. Therefore the failure to detect postural flexor activity in the first two experiments, as opposed to the last two, was most probably due to slight differences in electrode placement and contact impedance. Similarly the differences in the patterns observed between experiments were most likely the result of slight differences in the exoskeleton and stimulator set up.

In general, the behaviour of exoskeletally restrained subjects during quiet standing is not significantly different from that of unrestrained subjects.

6.3 PERTURBED STANDING

The objective of the first perturbed standing experiments was to determine an optimum sampling rate. That is, the fastest possible rate which would still provide a record of the subject's complete response in 512 samples. Consequently, runs were done on each subject using a fairly slow sampling rate to ascertain for what period samples should be taken.

Averaged responses to five anterior and posterior pulses of force lasting 500 msec are presented in Fig. 6.3. These responses were from subject J.C. but the other subject (J.M.) responded in a similar fashion. The subject's response to the stimulus is essentially complete within 2.5 seconds of the application of the stimulus.

Since the stimulus in this case (500 msec) was the longest used in the experimental program, a record length of 2.5 seconds was considered to be adequate. Consequently a sampling rate of 200/sec was used since this gives a sample length of just over 2.5 seconds; this rate was adopted for all subsequent experiments.

In these preliminary runs the vertical force (FZ) output from the force plate was recorded on the stimulus channel by adding it to the logic signal. The top curve in Fig. 6.3 represents this combination with all D.C. components removed and shows clearly that FZ is constant throughout the response. Hence the assumption of constant FZ made earlier, has been verified experimentally and consequently the torque at the ankle (TA) is in fact directly proportional to the anterior-posterior position of the centre of pressure (Xp).

The five responses to a given stimulus in each experiment were averaged and are presented in the appendix. The responses from the three experiments
Figure 6.3 a: Averaged responses to 500 msec anterior stimuli. Time (seconds) for Experiment 2. Subject J.C.
Figure 6.3.b: Averaged responses to 500 msec posterior stimuli. (seconds)

Experiment #2. Subject J.C.
conducted on subject J.C. were also averaged together. It is these data which are presented in Fig. 6.4 and are considered in detail.

In general the responses to the two durations and directions of stimulus are similar. The 500 msec stimulus elicits a larger and more repeatable response than does the 250 msec stimulus. The larger response is to be expected, while the increased repeatability probably results from the larger response being less influenced by minor variations in the initial conditions.

The following observations can be made from an examination of the averaged responses:

(i) The $Y_p$ tremor has a larger amplitude after the application of a stimulus than during quiet standing. There is no obvious relation between the $Y_p$ response and the $X_p$ response and the $Y_p$ response is much smaller than the $X_p$ response. Furthermore, the $Y_p$ response maintains the nature of tremor rather than appearing to be a characteristic response, although it may be a side effect coupled to the major a-p response.

It may therefore be concluded that the subjects were well aligned with the force plate and that the largest part of the response took the form of an angular sway in the anterior-posterior (a-p) plane. This supports the previous assumption of two dimensional body dynamics.

(ii) The angular position response ($\Theta_A$) of the subject was essentially the same for anterior and posterior stimuli. The subject first sways in the direction of application of force starting shortly after it is applied. The sway continues smoothly, reaching a peak some time after the force is removed (250 msec for the 500 msec pulse and 375 msec for the 250 msec pulses). The subject then returns to a new steady state angle of lean with no oscillation larger than normal postural tremor.

The average amplitude of the responses to anterior and posterior force pulses 250 msec long are identical but the anterior response to a 500 msec force pulse is about twenty per cent larger than the response to a posterior force pulse. It is not clear whether this is due to a nonlinearity in the control system, the body dynamics or the stimulator set up.
(iii) The $X_p$ response, which is directly proportional to ankle torque $T_A$, starts to rise 40 to 60 msec after the sway starts and rises steadily for about 175 msec. The rate of increase then decreases for about 125 msec and the curve then climbs sharply again to its maximum value. Finally, it decays smoothly to its final value. As with the angular position response ($\theta_A$) the final value of $X_p$ is not necessarily the same as the initial value. The levelling out or "jog" in the curve is most pronounced in the responses to 250 msec pulses but is present in all cases. It is possibly due to a nonlinearity in the system or to the effects of higher centres of the CNS, since the time at which the "jog" starts, about 250 msec after onset of stimulus, corresponds approximately to the delays expected in higher centre actions.

(iv) A characteristic and coordinated EMG response of the flexor and extensor muscles was elicited by both anterior and posterior force pulses.

The response to an anterior pulse consists first of a fall in the flexor activity (if present), and an accompanying rise in extensor activity causing an increase in extending torque which acts to resist the disturbance. The extensor activity continues for a short period after the subject's displacement reaches a peak and starts to return. At this point the extensor activity falls off and flexor activity returns to a normal static level, preventing overshoot and stabilizing the subject in his new position.

The response to the posterior stimulus is essentially the reverse of that to an anterior pulse. That is, the extensor activity (if present) first falls as flexor activity rises. After the peak in displacement the flexor activity falls and extensor activity returns to normal.

The first EMG changes are noted less than 100 msec after the stimulus is applied. Such a fast response can only be the result of a spinal reflex. Although influences from higher centres may become important after longer time delays, the reflex activity is obviously an important part of the response.
Figure 6.4 a: Averaged responses to 250 msec anterior stimuli. Time (seconds)
Experiments 2, 3, 4. Subject J.C..
Figure 6.4 b: Averaged responses to 250 msec posterior stimuli. Time (seconds)
Experiments #2, #3, #4. Subject J.C..
Figure 6.4 c: Averaged responses to 500 msec anterior stimuli. Time (seconds)

Experiments #2, #3, #4. Subject J.C.
Figure 6.4d: Averaged responses to 500 msec posterior stimuli. (seconds)
Experiments #2, #3, #4. Subject J.C.
CHAPTER VII

SIMULATION OF THE POSTURAL CONTROL SYSTEM

7.1 INTRODUCTION

The analysis performed in Chapter III demonstrated that the PCS can be expected to have the structure shown in Fig. 7.1.

Furthermore, it has been shown experimentally that the assumption of two dimensional body dynamics is reasonable for the exoskeletally restrained subject. The variables of interest for the experimental situation are the torque around the ankle ($T_A$) and the angle of lean ($\theta_A$) rather than the torque and position vectors.

In the experimental program the input (angle of lean $\theta_A(t)$) of the controller and the output (ankle torque $T_A(t)$) were recorded while standard disturbance torques were applied to the subjects. The resulting identification problem is that of identifying the dynamic structure and parameters of a "black box" whose input and output are both known (Fig. 7.2).

The identification procedure adopted involved the simulation of mathematical models of this simplified PCS, based on current physiological knowledge, on a hybrid computer. The experimentally observed position data ($\theta_A(t)$) was supplied as an input to the simulation and the model response recorded. The validity of the model was assessed by comparing the model response to the physiological response. The model parameters and structure were varied to obtain the best matching of responses, based on a mathematical error criterion, and hence the "best" model.

The simulation techniques employed, the details of the models simulated and the results of the simulations are discussed in this chapter.

7.2 SIMULATION TECHNIQUES

7.2.1 Introduction

The hybrid computer used in the simulations consisted of a PDP-12 digital computer and an EAI-380 analog computer. The digital machine was used to supply the experimental inputs to the simulations, to store the simulated response, to calculate the error criterion of the simulation results and to control the mode of the EAI-380. The analog machine was used to simulate the dynamics of the PCS models.
Figure 7.1: Structure of the Postural Control System.

Figure 7.2: The system to be identified.
electronically. The combination of the capabilities of the two types of machines produced a powerful, fast and very flexible simulation tool.

7.2.2 The Hybrid Computer

The PDP-12 and EAI-380 were easily hybridized since the PDP-12 is a laboratory computer and the 380 was designed to be operated in conjunction with a digital computer. The following hardware modifications were necessary to permit the two machines to be operated together.

(i) An instrument rack was installed beside the EAI-380 to serve as a switchboard for all interface connections. All cables leading between the two machines terminate on panels mounted on this interface rack. In addition, all the interface patching is done on these panels. This arrangement makes it possible to remove the PDP-12 from the hybrid system with a minimum of difficulty.

(ii) Logic circuitry, consisting of two flip flops and a device selector card, was installed in the PDP-12 to make two logic signals under PDP-12 program control externally available. The logic signals were connected by cables to a panel on the interface rack and terminated in banana plugs.

(iii) Eight analog trunk lines were led from the interface rack to a connector at the rear of the EAI-380. These trunk lines terminate on the EAI-380 patch panel. Patch panels may therefore be changed on the machine without altering the connections to the interface panel.

Two of these trunks were patched to the two PDP-12 logic signals terminated on the interface rack. These trunk lines were also patched to the mode control inputs on the EAI-380 patch panel. The logic levels of the PDP-12 and the EAI-380 are compatible and so the analog machine was thus placed under the control of the PDP-12.

(iv) Leads from eight of the PDP-12 analog converter inputs were led to the outputs of potentiometers mounted on the interface panel. The inputs of the potentiometers terminate on the front of the panel. The potentiometers were adjusted to attenuate the ±10 volt range of the EAI-380 to the ±1 volt range of the PDP-12 A-D converter.

Signals may be led to the PDP-12 from the EAI-380 simply by patching them
to the analog trunk input on the patch panel and then- patching the analog trunks to the potentiometer inputs on the interface panel.

(v) The two outputs from the PDP-12 digital to analog (D-A) converter were led to the interface rack. Here an operational amplifier circuit provides the D.C. offset and gain necessary to convert the asymmetric 0. to -5.00 volt D-A output to a symmetric ±10 volt signal. The processed signal is then terminated on the front of the panel.

Signals can thus be fed to the 380 patch panel from the PDP-12 simply by patching the processed D-A outputs to the analog trunk inputs on the interface panel.

(vi) All twelve connections to the PDP-12 were made with two multi­connector cables terminating in Blue Ribbon connectors. The PDP-12 can therefore be removed from the system simply by removing these two connectors.

(vii) Additional components may be mounted in the interface rack and used in the simulation by patching into the analog trunks. Two multipliers and a full wave rectifier were used from this position in the simulations discussed in this thesis. The additional versatility provided by this arrangement was extremely useful since at the time of this work the EAI-380 had only been partially expanded.

A schematic diagram of this hybrid computing system is given in Fig. 7.3.

In addition to this hardware a program was written to control the simulations and make use of the facilities of the hybrid machine. The functions that the control program "HYBRID" performs are discussed in some detail later in this chapter and in the appendix.

7.2.3 Simulation Data

The modeling technique adopted for the simplified PCS requires the angular position of the subject to be supplied as an input to the simulation; in addition, any realistic model will also require the angular velocity. The program "HYBRID" reads these two variables from Linc tape into core and then proceeds with the simulation using the D-A to supply them to the model.

The angular position (θ_A(t)) of the subject was available from the experimental data but was obscured by high frequency noise due to wiper bounce in the
Figure 7.3: Schematic diagram of the hybrid computing set up.
position transducer potentiometer. It was necessary to filter out this noise before differentiating to obtain the angular velocity. The filtering and differentiating were carried out digitally using the data processing program "PROCESS".

The digital filtering algorithm used was a symmetric three point filter defined by:

\[ X'(t_0) = \frac{X(t_0 - T) + 2X(t_0) + X(t_0 + T)}{4} \]  

(7.1)

where

- \( X'(t_0) \) = filtered sample at time \( t_0 \)
- \( X(t_0) \) = unfiltered sample at time \( t_0 \)
- \( T \) = sampling interval.

This filter has been described in the literature (73) and approximates a 12dB/octave, zero phase shift, low pass filter for the frequency range of interest. Additional attenuation is obtained by filtering the data repeatedly. After \( n \) filter applications, the filter function is

\[ F_2'(f) = (F_2(f))^n \]  

(7.2)

where \( F_2(f) \) is the filter function for one pass and is given by

\[ F_2(f) = \frac{1}{2} \left( 1 + \cos \frac{1.14}{f_1} \right) \]  

(7.3)

and

\[ f_1 = \frac{0.183}{T} \]  

(7.4)

where

- \( f \) = frequency
- \( F_2(f) \) = attenuation at frequency \( f \)
- \( T \) = sampling interval.

The sampling interval used in all experiments described in this thesis was 5 msec so that ten filter passes will attenuate signals of less than 1 Hz by less than 0.5%. There was no significant component of the position response above 1 Hz, and ten filter passes effectively removed the noise from the signal, so all position data was filtered ten times and then stored for use in simulation. The same filtering could be obtained by one pass through an equivalent higher order filter, but the increased programming required and reduced flexibility made this
approach uneconomical.

The angular velocity \( \dot{\theta}_A(t) \) was obtained from the filtered position \( \theta_A(t) \) by taking "fifth point differences" and then dividing by the time to get the correct magnitude as defined in Eq. 7.5. Fifth point differences were taken to reduce truncation error.

\[
\dot{\theta}(t) = \frac{1}{4T} (\theta(t + 2T) - \theta(t - 2T)) \tag{7.5}
\]

The differentiated position data was scaled to the correct magnitude for use in the simulation and stored on tape.

7.2.4 Error Criterion

A quantitative measure of the degree of agreement between the simulated and experimental responses was required for the matching procedure to work efficiently. The program "HYBRID" calculates such a criterion defined by

\[
E = \frac{\sum_{i=1}^{n} |X_i - R_i|}{\sum_{i=1}^{n} |R_i|} \times 100 \tag{7.6}
\]

where

- \( E \) = error criterion
- \( n \) = length of simulation
- \( \frac{\text{length of simulation}}{\text{sampling interval}} \)
- \( X_i = i^{th} \) sample of simulated response
- \( R_i = i^{th} \) sample of experimental response.

The error is expressed as a percentage of the area of the response data and consequently does not depend upon the magnitude of the response data.

The program obtains the reference data from tape and this data must therefore be stored prior to the start of a simulation. Once the error criterion has been calculated, its value is displayed on the screen of the PDP-12. Simultaneously the experimental and simulated responses are superimposed on the screen to provide a qualitative as well as quantitative impression of how well the two responses agree.

7.3 SIMPLE REFLEX MODEL

The first model that was simulated was a simple reflex model which considers
only the muscle mechanics and muscle spindle response. A block diagram of this model is shown in Fig. 7.4.

The patch diagram used in the simulation of this model on the hybrid computer is shown in Fig. 7.5 where the potentiometer settings are defined by the relations

\[
\begin{align*}
P_{00} &= K_m \\
P_{01} &= B_m \\
P_{02} &= \frac{1}{T_2} \\
P_{03} &= \frac{T_1}{T_2} \\
P_{04} &= \frac{1}{T_2} \\
P_{05} &= \frac{K_S}{T_3} \\
P_{06} &= \frac{1}{T_3}
\end{align*}
\]

The actual potentiometer values were altered by magnitude and time scaling of the above relations.

The control program ("HYBRID") takes samples from A-D#1 and A-D#2 at every increment of time. At the end of a run the transmission delay of the muscle spindle is accounted for digitally by forming the following sum:

\[
R(N) = S1(N) + S2(N - D)
\]  

(7.6)

where

- \(R(N)\) = \(n\)th value of model output
- \(S1(N)\) = \(n\)th sample from A-D#1
- \(D\) = integral value of \(\frac{d}{T}\)
- \(d\) = delay length
- \(T\) = sampling interval
- \(S2(N - D)\) = 0 for \(N\) less than \(D\)
- \(= (N - D)^{th}\) sample from A-D#2 for \(N\) greater than or equal to \(D\).

It would be very useful to have estimates for all the model parameters prior to starting the simulation. Unfortunately information about these values is very hard to come by.

The spindle delay is the only parameter for which an estimate, based directly on experimental evidence, is available. In an experiment performed by Stark the
Figure 7.4: Block diagram of the simple reflex model of the PCS.

Note: $K_s$ corresponds to $K_s \cdot K_1$ in Fig. 3.13; $T_s$ is a new variable.
Figure 7.5: Analog patching diagram for the simple reflex model of the PCS.
EMG activity of the ankle muscles was observed while a sudden tap was applied to the Achilles tendon of the subject\(^{(2)}\). A delay of 40 msec was observed between the application of the tap and the first changes in the EMG activity. These electrical changes are without doubt due to a stretching of muscle spindles in the ankle muscles and hence the delay is due to the transmission delay of the spindle outputs. An estimate of 40 msec for the spindle transmission delay \((d)\) will therefore be used in the simulations.

Estimates for the values of the time constants \((T_1, T_2, T_3\) in Fig. 7.4) were obtained from the work of Nashner\(^{(52)}\) and Houk\(^{(28)}\). These values were obtained by fitting a model response to experimental data and were

\[
\begin{align*}
T_1 &= 2.0 \\
T_2 &= 0.1 \\
T_3 &= 0.08.
\end{align*}
\]

No reasonable estimates were available for the gains so these were chosen arbitrarily to start with.

Starting with these initial values, the parameters of the simple reflex model were adjusted iteratively to obtain the best possible matching of the simulated and experimental responses. It was found to be necessary to treat the anterior and posterior stimuli cases separately. Parameters were found for which the model responses matched the experimental responses to posterior stimuli (see Fig. 7.6 where the experimental and simulated responses are superimposed).

Different parameters were necessary in order to obtain matched responses for anterior stimuli. Even with the change in parameters, the model response did not match the response to the 500 msec pulse very well, although the matching for for the 250 msec pulse was adequate. The inadequacy of the 500 msec simulation response is shown clearly in Fig. 7.7.

A summary of the parameter values and error criteria for the models for the anterior and posterior responses is given in Table 7.1. It should be noted that the errors calculated for the first half of the response may be of more value than those calculated for the entire response since the latter place excessive importance on steady state error.
Figure 7.6: (a) Simple reflex model response and physiological response to 250 msec posterior stimuli.
(b) Simple reflex model response and physiological response to 500 msec posterior stimuli.
Figure 7.7: (a) Simple reflex model response and physiological response to 250 msec anterior stimuli.
(b) Simple reflex model response and physiological response to 500 msec anterior stimuli.
<table>
<thead>
<tr>
<th>Parameter</th>
<th>Posterior Response</th>
<th>Anterior Response</th>
</tr>
</thead>
<tbody>
<tr>
<td>(K_m)</td>
<td>1.75 m/rad</td>
<td>1.37 m/rad</td>
</tr>
<tr>
<td>(B_m)</td>
<td>0.08 m/rad/sec</td>
<td>0.005 m/rad/sec</td>
</tr>
<tr>
<td>(K_s)</td>
<td>-0.045 m/sec</td>
<td>0.025 m/sec</td>
</tr>
<tr>
<td>(T_1)</td>
<td>2.5 sec</td>
<td>2.5 sec</td>
</tr>
<tr>
<td>(T_2)</td>
<td>0.1 sec</td>
<td>0.1 sec</td>
</tr>
<tr>
<td>(T_3)</td>
<td>0.08 sec</td>
<td>0.08 sec</td>
</tr>
</tbody>
</table>

Error for 50% of 250 msec response: 10% vs. 13%
Error for all of 250 msec response: 37% vs. 26%
Error for 50% of 500 msec response: 9% vs. 20%
Error for all of 500 msec response: 31% vs. 28%

Note: \(K_m, B_m, K_s\) have been normalized with respect to the subject's body weight.

To convert m/rad to n-m/rad multiply by 660.
An examination of Figures 7.6 and 7.7 shows that the simple reflex model provides a response which closely resembles the experimental response for posterior stimuli but not for anterior stimuli. The simulated response differs most from the experimental response to posterior stimuli at the start and the end of the response. At both times the velocity and displacement are small and hence nonlinear effects such as stiction may be of importance. It is also possible that the "jag" at the start of the response is due to a higher centre effect, with a long delay, causing a change in some parameter. However it is not possible to determine the nature of this effect with the present experimental data and simulation capabilities. Apart from these two difficulties the model response is very close to the physiological response to posterior stimuli.

The model response for anterior stimuli does not agree well with the experimental data. The problem is most evident when considering the response to a 500 msec pulse of force, but is present in the response to a 250 msec pulse as well. The "best fit" model responds in a much more sluggish fashion than does the subject. This is because the ankle torque rises and falls much more quickly than can be obtained by altering model parameters. This is a serious flaw in the model and indicates that the simple reflex model is not an adequate explanation of the response of the simple PCS to an anterior stimulus.

7.4 GAIN CONTROL MODEL

In an attempt to obtain a more realistic response to anterior stimuli, the simple reflex model of the PCS was modified to include gain control of the muscle spindle ($K_s$ in Fig. 7.4). The gain of the spindle output was made proportional to the displacement of the subject from the normal angle of lean. This has been suggested by several workers as a possible means of control (28, 52) and may be explained physiologically as a change in motor neuron threshold as a result of vestibular or proprioceptive inflow.

A block diagram for the gain control model of the PCS is shown in Fig. 7.8. This model is essentially the same as the previous one except that a full wave rectifier and multiplier are employed to control the gain of the spindle signal. The patch diagram for this model is shown in Fig. 7.9 and the pot settings are
Figure 7.8: Block diagram of the gain control model of the PCS.
Figure 7.9: Analog patching diagram for the gain control model of the PCS.
defined by the relations

\[
\begin{align*}
P_{00} &= K_m \\
K &= P_{04} = \frac{1}{T_2} \\
K &= P_{01} = B_m \\
K &= P_{05} = \frac{K_s}{T_3} \\
K &= P_{02} = \frac{1}{T_2} \\
K &= P_{06} = \frac{K_R}{T_3} \\
K &= P_{03} = \frac{T_1}{T_2} \\
K &= P_{07} = \frac{1}{T_3}
\end{align*}
\]

The actual pot settings are determined by amplitude and time scaling these relations.

It was possible to find parameters for which the gain control model response corresponded well to the physiological responses to an anterior stimulus. These responses are presented in Fig. 7.10. However, it was not possible to find a set of parameters for which the model responses would improve upon the responses obtained with the simple reflex model. The responses of the model used for anterior stimuli to posterior stimuli are presented in Fig. 7.11 to demonstrate the failure of the gain control model for these stimuli. The gain control model results in error criterion values of more than twice the values obtained with the simple reflex model. The parameter values used in these stimulations and the resulting error criterion values are summarized in Table 7.2.

The gain control model is clearly a great improvement over the simple reflex model in dealing with responses to anterior pulses of force. The model response now rises and falls as quickly as the experimental response and in general closely resembles the experimental response. However the model has the same failing as the simple reflex model in that it fails to predict the initial "jog" in the curve.
Figure 7.10: (a) Gain control model response and physiological response to 250 msec anterior stimuli.
(b) Gain control model response and physiological response to 500 msec anterior stimuli.
Figure 7.11: (a) Gain control model response and physiological response to 250 msec posterior stimuli.
(b) Gain control model response and physiological response to 500 msec posterior stimuli.
TABLE 7.2. GAIN CONTROL MODEL PARAMETERS

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Anterior Response</th>
<th>Posterior Response</th>
</tr>
</thead>
<tbody>
<tr>
<td>$K_m$</td>
<td>1.55 m/rad</td>
<td></td>
</tr>
<tr>
<td>$B_m$</td>
<td>0.0</td>
<td></td>
</tr>
<tr>
<td>$K_s$</td>
<td>-0.01 m/rad</td>
<td></td>
</tr>
<tr>
<td>$K_R$</td>
<td>0.260 m/rad</td>
<td></td>
</tr>
<tr>
<td>$T_1$</td>
<td>2.5 sec</td>
<td></td>
</tr>
<tr>
<td>$T_2$</td>
<td>0.1 sec</td>
<td></td>
</tr>
<tr>
<td>$T_3$</td>
<td>0.08 sec</td>
<td></td>
</tr>
<tr>
<td>Error for 50% of 250 msec response</td>
<td>13%</td>
<td>27%</td>
</tr>
<tr>
<td>Error for all of 250 msec response</td>
<td>23%</td>
<td>51%</td>
</tr>
<tr>
<td>Error for 50% of 500 msec response</td>
<td>10%</td>
<td>20%</td>
</tr>
<tr>
<td>Error for all of 500 msec response</td>
<td>16%</td>
<td>44%</td>
</tr>
</tbody>
</table>

Note: $K_m, B_m, K_s, K_R$ have been normalized with respect to the subject’s body weight. To convert m/rad to n-m/rad multiply by 660.
7.5 DISCUSSION

It is clear that neither of the two simple models discussed is capable of predicting the complete system response. However, the simple reflex model appears to represent adequately the behaviour of the system in response to stimuli applied anteriorly.

Based on these observations, it seems reasonable to postulate that the control of posture is qualitatively as well as quantitatively different in the anterior and posterior directions. This appears reasonable when it is noted that the projection of man's centre of gravity normally falls just in front of the ankle axis. In this position there is a much larger base for support anterior to the ankle than posterior to it. Consequently it would be reasonable to set the gain much lower for anterior disturbances than for posterior disturbances, since larger deviations can be tolerated anteriorly than posteriorly. The advantage of this is that the lower the initial gain is the less the resting energy requirement, since the initial gain is maintained by muscle tone which requires energy. Such an arrangement would be very easy to implement - perhaps through "γ" loop control of agonists and antagonists. The initial gain of the simple reflex model is 30% larger than the initial gain of the gain control model.

Neither of the two models discussed in this chapter is capable of predicting the fine details of the early parts of the system response. Clearly the system is more complex than either of the two models alone or of their combination would imply. As mentioned previously, there are several possible explanations for the early behaviour of the system. Although it is beyond the scope of this thesis to deal with this problem it should certainly be the subject of further experimental and simulation work.
CHAPTER VIII

CONCLUSIONS AND RECOMMENDATIONS

8.1 CONCLUSIONS

The basic goal of this thesis has been to develop a mathematical model of the postural control system in an exoskeletonally restrained man. It was found that two separate models had to be used in order to predict the system response adequately. The simple reflex model (Fig. 7.4) satisfactorily models the system response to stimuli applied posteriorly to the subject; but a more complex gain control model is required (Fig. 7.8) to predict the system response to anterior stimuli. Together these two models predict the overall system response quite well; although it is apparent that a more sophisticated model is required to predict the fine details of the response.

8.2 RECOMMENDATIONS

The approach used to study the PCS in this thesis has proven to be quite successful and it is recommended that the work be extended. In the continuation of the work the following approaches would appear to be worthy of investigation:

(i) The development of a servo controlled stimulating device should be undertaken to replace the rather crude one currently in use. Such a device should be capable of applying stimuli of many different waveforms and amplitudes to the subject. It would allow the use of different identification techniques (such as frequency response, step response, correlation methods, etc.) and hence greatly extend the possible approaches to studying the system.

(ii) A closer investigation of the EMG activity of the ankle muscles would make available a system variable which has not been already smoothed by the muscle mechanics, as has the ankle torque. EMG's from all three ankle muscles of both legs would be required and might prove to be a very sensitive indicator of PCS activity.

(iii) A rotating support platform, as used in Nasher's experiments, could be used to supply stimuli to the PCS. This platform could also be servoed
in such a way as to keep the ankle angle constant and hence eliminate position sensitivity from the PCS. The importance of the position signal in postural control could then be assessed by comparison of these and normal responses.

(iv) Subjects with vestibular or neurological disorders should be tested in order to assess the effects of their disorders on postural control. It would be necessary to have close medical collaboration in such a project to interpret the results of such experiments.

(v) The modelling should be extended to include models of the semi-circular canals, the otoliths, proprioceptors and other receptors. Such an expanded model would hopefully predict the fine details of the system response and help to explain the role of the different receptors in postural control.
Derivation of the Force Plate Equations

The force plate variables defined in Chapter IV were obtained by algebraic manipulation of the load cell outputs. The relations defining these manipulations are derived in this appendix.

Referring to Fig. A.1 which shows the layout of the load cells in the force plate -

sum the forces in the X direction to get

\[ F_x - F_4 + F_5 \cos 60 + F_6 \cos 60 = 0 \]  \hspace{1cm} (A.1)

rearranging and evaluating of constants gives

\[ F_x = F_4 - 0.5 (F_5 + F_6) \]  \hspace{1cm} (A.2)

Similarly summing forces in the Y direction gives

\[ F_y - F_5 \sin 60 + F_6 \sin 60 = 0 \]  \hspace{1cm} (A.3)

and

\[ F_y = 0.866 (F_5 - F_6) \]  \hspace{1cm} (A.4)

The vertical force is simply

\[ F_z = F_1 + F_2 + F_3 \]  \hspace{1cm} (A.5)

Now sum the moments about the X axis to get

\[ M_x + L_1 F_1 - L_2 F_2 + L_1 F_3 = 0 \]  \hspace{1cm} (A.6)

which gives

\[ M_x = -20 F_1 + 45.9 F_2 - 20 F_3 \]  \hspace{1cm} (A.7)

Similarly summing moments about the Y axis gives

\[ M_y - L_3 F_3 + L_3 F_1 = 0 \]  \hspace{1cm} (A.8)

and

\[ M_y = 38.1 (F_3 - F_1) \]  \hspace{1cm} (A.9)

sum moments about the vertical axis

\[ M_z - L_1 F_4 - L_2 F_5 \cos 60 + L_1 F_6 \sin 30 - L_3 F_6 \cos 30 = 0 \]  \hspace{1cm} (A.10)
giving
\[ M_z = 20F_4 + 22.95F_5 + 23F_6 \quad \text{(A.11)} \]

The position of the effective centre of pressure is defined by the relations

\[ F_z X_p = M_y \quad \text{(A.12)} \]
or
\[ X_p = \frac{M_y}{F_z} \quad \text{(A.13)} \]

Using Eq. (A.9) and (A.5), we get

\[ X_p = \frac{38.1(F_3 - F_1)}{F_1 + F_2 + F_3} \quad \text{(A.14)} \]

Similarly,

\[ F_z Y_p = M_x \quad \text{(A.15)} \]

and
\[ Y_p = \frac{M_x}{F_z} \quad \text{(A.16)} \]

and using Eq. (A.7) and (A.5)

\[ Y_p = \frac{-20F_1 + 45.9F_2 - 20F_3}{F_1 + F_2 + F_3} \quad \text{(A.17)} \]

summarizing

\[ F_x = F_4 - 0.5(F_5 + F_6) \quad \text{(A.18)} \]
\[ F_y = 0.866(F_6 - F_5) \quad \text{(A.19)} \]
\[ F_z = F_1 + F_2 + F_3 \quad \text{(A.20)} \]
\[ M_x = -20F_1 + 45.9F_2 - 20F_3 \quad \text{(A.21)} \]
\[ M_y = 38.1(F_3 - F_1) \quad \text{(A.22)} \]
\[ M_z = 20F_4 + 22.95F_5 + 23F_6 \quad \text{(A.23)} \]
\[ X_p = \frac{38.1(F_3 - F_1)}{F_1 + F_2 + F_3} \quad \text{(A.24)} \]
\[ Y_p = \frac{-20F_1 + 45.9F_2 - 20F_3}{F_1 + F_2 + F_3} \quad \text{(A.25)} \]
The three vertical load cells $F_1, F_2, F_3$ form the vertices of an equilateral triangle.

$L_1 = 20 \text{ cm}.; L_2 = 45.9 \text{ cm}.; L_3 = 38.1 \text{ cm}.;

Figure A.1 : Layout of the load cells on the force plate.
APPENDIX B

Description of Computer Programs

B.1 PEXCO - Postural Experiment Control Program

The program starts with a series of messages displayed on the PDP-12 CRT in which the experimenter is asked to enter a number of program variables by way of the teletype. These variables determine the rate at which samples are to be taken, the first block of Lincl tape on which data is to be written, the number of initial condition samples to be taken and the length of stimulus to be applied to the subject.

The program then enters its monitor mode in which the value of a selected A-D channel is displayed graphically on the PDP-12 CRT. The A-D channel is selected by setting the channel number in the left console switches.

The program leaves the monitor mode and enters the data acquisition mode when the experimenter sets either sense switch zero or sense switch one. In this mode the program takes samples from each of six A-D channels at a rate controlled by the PDP-12 real time clock. A number of initial condition samples are taken and then the program closes either relay zero or relay one depending upon whether sense switch zero or one is set. The closing of one of these relays applies current to one of the stimulators and hence initiates the stimulus applied to the subject. Sampling continues until the length of stimulus to be applied is reached; the relay is then opened. The program continues to sample until 512 samples from each of the six channels have been taken.

The program then leaves the data acquisition mode and enters its display mode in which the sampled data is displayed on the CRT, two channels at a time. The experimenter then has the option of either destroying the data or storing the data on Lincl tape as a set of six variables of 512 samples each. The program then returns to the monitor mode and is ready to obtain another set of data.

B.2 AVERAGE - Averaging Program for Data Acquired by PEXCO

The program starts with a message which asks the operator to enter the first tape block to be read, the number of data sets to be averaged, and the first
tape block to be written on by way of the teletype.

The program then calculates the mean value of each of the variables using a non-growing averaging algorithm to avoid any problems with overflow. The algorithm used is

\[ A_{i,n} = A_{i,n-1} + \frac{(S_{i,n} - A_{i,n-1})}{n} \]  

(B.1)

where \( A_{i,n} \) = average value over \( n \) sets of data of the \( i \)th sample.

\( S_{i,n} \) = \( i \)th sample of the \( n \)th data set

\( n \) = number of data set.

The program then calculates the mean absolute deviation (MAD) for each of the variables using the algorithm

\[ M_{i,n} = \frac{\sum_{k=1}^{n} |S_{i,k} - A_{i,n}|}{n} \]  

(B.2)

where \( M_{i,n} \) = mean absolute deviation of the \( i \)th sample calculated over \( n \) data sets.

The program then displays the average value and average value plus and minus one MAD of each variable. The operator then has the option of either destroying the averaged data or storing it on Linc tape.

B.3 PROCESS - Data Processing Program

The program is used to apply various data processing techniques to 512 pairs of data and is capable of doing the following things:

(i) Read 512 points of data from Linc tape into a buffer.

(ii) Write 512 points from the buffer into Linc tape.

(iii) Invert the data in the buffer.

(iv) Low pass filter the data in the buffer using the three-point digital filter discussed in Chapter 7.

(v) Multiply the data in the buffer by a constant.
B.4 HYBRID - Hybrid Computer Control Program

The program starts with a series of messages which ask for the rate at which data is to be displayed, the first block number from which to read data and the first block number onto which data is to be written.

After receiving these codes the program enters the calibration mode in which the outputs of the two D-A's is determined by the console switches. This mode is used to allow the D-A circuitry to be adjusted to the right levels.

The program then reads two variables of data and a reference curve from tape and displays it on the PDP-12 scope. Upon a command from the teletype the program puts the analog computer in the Initial Condition mode and displays the initial value of the two variables on the D-A's. After a delay set by the left switches the EAI-380 is put into the operate mode and the PDP-12 displays the two variables on the D-A's at a rate controlled by the PDP-12 real time clock. At the same time two channels of A-D are sampled and stored in core. This continues until all data has been displayed.

The program then returns to a display mode in which it displays the sum of the two sampled variables shifted with respect to each other (as defined in Chapter 7), and a reference curve. At the same time, the error criterion is calculated and displayed numerically on the scope.

The operator then has the choice of altering the delay between the two sampled curves with a PDP-12 pot, altering one of the EAI-380 parameters and making another run, reading in new data or storing the simulated data and then reading in new data.

B.5 PLOTTER - Plotting Control Program

The program starts with a message asking for codes which set the clock rate. It then enters a calibrate mode in which the D-A outputs are set from the console switches to allow the plotter scales to be set.

The program then asks for the number of the first tape block to be entered and reads in data from Linc tape. The program interprets the data as being two variables and their MAD's and displays the corresponding curves on the scope.
The operator then selects the curve to be plotted using the sense switches and starts the plot. The D-A then displays the point 0, 0 and waits for sense switch two to be set. When this is done, the program closes relay zero to lower the plotter pen and proceeds to plot out the curve at a rate controlled by the PDP-12 real-time clock. If the 4000 code is set in the left console switches, the program plots the curve as a dashed line; otherwise it is plotted as a solid line.

After the curve has been plotted, the operator has the option of either plotting another curve stored in core or of reading a new set of data into core.
APPENDIX C

Experimental Results

The solid curves are the average of five responses recorded in one experiment.

The dashed lines are the average plus and minus one mean absolute deviation.

The variables are:

(i) Stimulus - logic output from stimulators.

   Positive level for anterior stimulus.
   Negative level for posterior stimulus.

(ii) $Y_p$ - medial-lateral position of the centre of pressure of the subject.

(iii) $X_p$ - anterior-posterior position of centre of pressure of the subject.

(iv) $\theta_A$ - angle of lean of subject.

(v) Extensor EMG - sum of filtered EMG's from the soleus and gastrocnemius muscles of the subject's right leg.

(vi) Flexor EMG - filtered EMG from the tibialis anterior muscle of the subject's right leg.
Figure C.1: Averaged responses to 125 msec anterior stimuli. Time (seconds)

Experiment #1, Subject J.M.
Figure C.2: Averaged responses to 125 msec posterior stimuli. (seconds)
Experiment #1. Subject J.M.
Figure C.3: Averaged responses to 220 msec stimuli.
Experiment #1, Subject J.M.
Figure C.4: Averaged responses to 220 msec posterior stimuli. Time (seconds)

Experiment #1. Subject J.M.
Figure C.5: Averaged responses to 500 msec anterior stimuli. Time (seconds)

Experiment #1. Subject J.M.
Figure C.6: Averaged responses to 500 msec posterior stimuli. Time (seconds)
Experiment 1. Subject J.M.
Figure C.7: Averaged responses to 250 msec anterior stimuli.
Experiment #1. Subject J.M.
Figure C.8: Averaged responses to 250 msec posterior stimuli. Time (seconds)
Experiment #1. Subject J.M.
Figure C.9: Averaged responses to 250 msec anterior stimuli. Time (seconds)
Experiment #2. Subject J.C.
Figure C.10: Averaged responses to 250 msec posterior stimuli. Time (seconds)

Experiment #2. Subject J.C..
Figure C.11: Averaged responses to 250 msec anterior stimuli. Time (seconds)
Experiment #2, Subject J.C.
Figure C.12: Averaged responses to 250 msec posterior stimuli. Time (seconds)
Experiment 2. Subject J.C.
Figure C.13: Averaged responses to 500 msec anterior stimuli. Experiment #2. Subject J.C..
Figure C.14: Averaged responses to 500 msec posterior stimuli. Time (seconds)

Experiment #2. Subject J.C.
Figure C.15: Averaged responses to 500 msec anterior stimuli. Experiment 62. Subject J.C.
Figure C.16: Averaged responses to 500 msec posterior stimuli. Time (seconds)
Experiment #2. Subject J.C..
Figure C.17: Averaged responses to 250 msec anterior stimuli. Time (seconds)
Experiment #3. Subject J.C..
Figure C.18: Averaged responses to 250 msec posterior stimuli. Experiment #3. Subject J.C.
Figure C.19: Averaged responses to 500 msec anterior stimuli. Time (seconds)
Experiment #3. Subject J.C.
Figure C.20: Averaged responses to 500 msec posterior stimuli. Time (seconds)

Experiment #3. Subject J.C.
Figure C.21: Averaged responses to 250 msec anterior stimuli. (seconds)
Experiment 4. Subject J.C.
Figure C.22: Averaged responses to 250 msec posterior stimuli. Time (seconds). Experiment #4, Subject J.C.
Figure C.23: Averaged responses to 500 msec anterior stimuli. Time (seconds)
Experiment #4. Subject J.C.
Figure C.24: Averaged responses to 500 msec posterior stimuli. Time (seconds)
Experiment #4. Subject J.C..
REFERENCES


References - cont'd


References - cont'd


References - cont'd


References - cont'd


