A DEVICE FOR OBJECTIVE MEASUREMENT OF SPASTICITY

by

EDDY BARAK

B.S. WAYNE STATE UNIVERSITY

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Signature of Author ____
Department of Aeronautics and Astronautics
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Certified by ____
Thesis Supervisor

Accepted by ____
Chairman, Departmental Graduate Committee

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ABSTRACT

Spasticity is a neuromuscular disorder, due to lesions of the central nervous system, and is characterized by hypertonus, hyperflexia and clonus. Clinical assessment of spasticity is important to evaluate the efficiency of chemo- and physical therapeutic treatment of spastic patients. At present, evaluation is made based on ill-defined and highly subjective methods, which lead to differences of opinion among clinicians. It is difficult, if not impossible, to obtain a quantitative estimate of the changes effected in the patient's condition for comparison with subsequent examinations. Therefore, a reliable and objective means for clinical measurement of spasticity and other neuromuscular disorders has been developed.

A foot manipulator, that rotates the foot about the ankle joint in a programmed manner, was designed and built. The device is controlled and monitored by a digital computer that records four parameters of interest: position, torque, tibialis EMG, and soleus EMG.

The system is capable of performing three basic types of movement: ramps, sinusoids and triangular displacements.

A preliminary evaluation of the new design has been done on a pool of normal subjects and on two patients having clinical evidence of "spasticity". Different velocity ramps, sinusoids and triangular movements were introduced with a variety of instructions given to the subjects. The results were found to be consistent, highly reproducible and in agreement with previously reported physiological phenomena. Performance of the device attests to the satisfaction of the design objectives and demonstrates the system's usefulness for application in various physiological investigations performed at the ankle.

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CHAPTER I

INTRODUCTION

1.1 Background

Spasticity is a neuromuscular disorder of muscle tone due to lesions of the central nervous system. It afflicts victims of upper motor unit damage such as spinal cord trauma, stroke and a number of congenital diseases referred to as cerebral palsy.

Clinically, the term is used to describe a combination of the following symptoms: increased muscle tone, i.e. increased resistance to lengthening (hypertonus), exaggeration of the passive stretch reflex, increased deep tendon reflexes (hyperreflexia), repetitive contractions in response to suddenly applied, but sustained, stretch of muscle (clonus) and the "clasp knife" phenomenon with sudden "melting" of resistance.

A considerable amount of effort, time and money is being devoted to chemo- and physical therapeutic treatment of spastic patients. Highly touted drugs surface perenially and are being used. New physical therapy programs are continuously being proposed and implemented. Neurosurgical and orthopedic procedures to relieve spasticity have been applied. Three common operations used to date are cutting nerves to prevent impulses from reaching muscles, cutting tendons of the spastic muscle and elongating them, and shortening the opposing muscles. In spite of all this, there is considerable difference of opinion among clinicians as to whether or not these efforts have any real beneficial effects on the patients. One fundamental reason for this situation has been the lack of an objective measurement, as well as ill-defined methods for evaluation.

Usual clinical assessments are based on a variety of wholly or partially subjective evaluations by physicians, therapists, nurses, social workers, and, of course, the patients themselves. Bias and unreliability of such evaluations is inevitable. For example, currently, to evaluate spasticity, an examiner holds one limb segment in a fixed position and moves the attached segment passively, noting the muscular resistance he encounters. It is impossible for the examiner to standardize the speed , force or arc of the movement, or to record and quantify the various variables such as the position, resistance encountered and the electormuscular activity and timing of the events. This may make it difficult, if not impossible, to retain the impression for comparison with subsequent examinations. The present clinical methods are therefore very subjective and highly variable since they depend not only on the state of the patient but also on the state of the examiner. Considering the energy involved, both physical and mental, a more reliable and objective means of clinical evaluation is sorely needed.

1.2 Clinical Features of Spasticity - A Brief Review

In general, gross spasticity is manifested by the state of the arm and leg of a patient with upper motor lesions (corticospinal tract). The leg is stiffly extended and resists passive flexion. The arm is held flexed and resists passive extension. Tendon reflexes are hyperactive and clonus may be evoked if the stretch is maintained. In the upper limb, spasticity effects mostly the flexors of the fingers, wrist and elbow, and to a lesser extent the extensors of the elbow and adductors of the shoulder. In the lower limb, it effects primarily extensors of hip and knee and plantar flexors of the ankle (triceps surae).

The terms "spastic" and "spasticity" have become such a habitual part of the neurological jargon that no one is expected to define them. But for the engineer, a proper definition is essential to be able to establish the design criteria. Dorland's Medical Dictionary defines spastic:

> "1. Of the nature or characterized by spasms. 2. Hypertonic, so that the muscles are stiff and movement awkward."

and spasticity:

> "A state of increase over the normal tension of muscle, resulting in continuous increase of resistance to stretching."

In the neurological literature (Landau, 1974), there are at least <u>six</u> varieties of spasticity that can be discriminated in this context:

1. Spasticity$_{PRR}$ (Proprioceptive Reflex Release)

Quiescent unstimulated muscle with increased proprioceptive reflexes, including, in various degrees, increased phasic tendon jerk reflexes, tonic stretch reflexes, and the tonic clasp-knife reaction.

2. Spasticity$_{GRR}$ (Generalized Reflex Release)

Reflex release including not only proprioceptive but also and especially polysynaptic flexion reflexes, e.g. the flexor spasms of chronic paraplegia.

3. Spasticity$_{UMNS}$ (Upper Motor Neuron Syndrome)

The entire upper motor neuron complex syndrome, including motor performance disability as well as reflex release, e.g. spastic paraparesis.

4. Spasticity$_{DR}$ (Dystonic - Rigid State)

A spectrum of ill-defined dystonic and rigid states of many origins and pathological features, all characterized by some degree of involuntary continuous muscle contraction, e.g. athetoid cerebral palsy.

5. Spasticity$_{M}$ (Mixed)

Combination of the above definitions, especially 3 and 4, usually associated with hemispheral or brain stem lesions.

6. Spasticity$_{U}$ (Undefined)

No basis for determining narrower contextual meaning.

As can be seen, the lack of clinical ability to define spasticity complicates even more the clinical assessment and evaluation and indicates the ambiguity surrounding spasticity. This is expected in light of the fact that very little is known about the mechanisms involved in spasticity.

Basically, there are two "types" of spasticity, which are observed in isolated preparations, cerebral and spinal. Cerebral spasticity might occur as the result of severe injury or destruction of some portion of the frontal lobe of the cerebral hemisphere or due to interruption of the pyramidal motor projection of the lobe at the level of the internal capsule. It can be produced by a variety of causes such as massive hemmorhage, tumor, and, in children, anoxia or metabolic toxins causing cerebral palsy which occurs during gestation, parturition or the neonatal period. Cerebral spasticity can be compared, to some extent, to decerebrate rigidity in animals after midbrain section, primarily affecting the extensors. Spinal spasticity is produced by lesions (complete or incomplete) of the spinal cord at or below the level of entry of the eighth pair of cranial nerves. It is often more severe and irregular than cerebral spasticity. Incomplete spinal lesions may give rise to more severe spasticity than a complete transverse lesion.

Mild spasticity is manifested in response to rapid passive stretching and it is only towards the end of the movement that some resistance is noticed. It is believed that the sensitivity of the stretch receptors is augmented so that tendon reflex can be elicited by a subnormal stimulus, but only the muscles which reflexively contract in the normal subject are involved. In moderately severe spasticity, resistance to the passive movement is noticed earlier in the stretching, and the brake action may be so pronounced that the movement is completely stopped. If the stretch is continued, the resistance suddenly yields. This is autogenic inhibition or the so-called clasp knife phenomenon. Tendon reflex is not only augmented, but also radiates to other groups of muscles, including the antagonists. This might lead to clonus, a state characterized by alternate contraction of agonists and antagonists.

On the basis of the present knowledge of spasticity, only a few hypotheses attempt to explain the hypothesis [Magoun, 1974; Jansen, 1972; Pederson; 1976]. Current theories agree that spasticity is related to the removal of inhibitory influences acting on the spinal stretch reflex and also to the maintained activity of supraspinal influences that facilitate these reflexes.

Animal experiments have shown that lesions in the corticospinal tract lead to increased facilitation of the gamma motoneurons. This in turn causes increased sensitivity of the muscle spindles to static and dynamic stretch that leads to hyperactivity of the alpha motoneruon. Increase in firing rate can lead to contraction of the muscle, i.e. increased muscle stiffness, known as decrebrate rigidity. This occurs in animals after midbrain section. It is reasonable to compare it with cerebral spasticity in man since both are characterized by extensor rigidity of the gamma type. In the case of spinal spasticity, the mechanisms involved are more complex since both the alpha and gamma motoneurons may be influenced and there might be effects of sprouting.

An attempt to go into more detailed explanations of the present knowledge of spasticity would be far beyond the scope of this thesis and it is left to the interested reader to search the vast amount of literature dealing with this topic, including the mechanism of the stretch reflex.

1.3 Related Research

Various attempts have been made at objective quantitative measurement of spasticity. In 1930, after the epidemic of Spanish influenza had left innumerable cases of hypertonia as a sequel of the Von Economo lethargic encephalitis, Dr. Lewis J. Doshay started the study and evaluation of drugs in the treatment of hypertonicity [Doshay, 1964].

He devised an electrical recording device with a simple telegraph key. Each time the patient closed the key, a mark was made on a piece of smoked paper on a revolving kymogram drum. The number of strokes in a five second interval was counted, before and after use of the drug. This gadget afforded a fairly extensive "quantitative" measure of speed of rapid movement and changes produced by medication, yet it was open to a variety of artifacts and biasing factors such as tremor and the fact that active (voluntary) limb motion rather than passive motion was used (contrary to the basic definition of spasticity). This method was crude and the data highly variable.

In the late 1940's came a flood of synthetic drugs demanding evaluation and with it a greater need for speedy and more precise quantitative devices. In spite of this, there were relatively few attempts to quantitatively measure spasticity.

Since spasticity is characterized by hypertonus, exaggeration of the stretch reflex, hyperreflexia and clonus, measurements have been concentrated on the increased reflex response. This has been done by measuring the parameters of the tendon jerk and/or of the passive stretching of the muscle in some well defined manner.

Tendon jerks can be elicited by means of a reflex hammer, which under standardized conditions gives a certain stimulus by striking the tendon. The reflex response can be measured by a force transducer, goniometer or electromyography. Erdman and Heather [1964] and Heather et al [1965] described a device to record force developed at the foot in response to a standardized tendon tap using a spring loaded reflex hammer. Patients with paraplegia or quadruplegia were tested in relation to diazepam therapy. The only conclusion was that the average force, obtained after the drug was administered, was reduced. No sample of characteristic data obtained from patients was presented.

A special form of passive stretch can be obtained by allowing the lower leg to swing freely about the knee joint. The characteristic motion obtained would be of the damped pendulum type. Mumenthaler [1965] used a light bulb attached to the leg, and a camera to obtain a record of the number of pendulations. This method was considered to be too complicated and costly as claimed by Van der Laarse and Oosterveld [1971], who simplified the so-called "pendulum test" by using a goniometer (at the knee) and a pen recorder.

Burry [1967] reported using a spring loaded foot plate to induce clonus and recording the resultant soleus EMG. In his conclusion, he noted the fact that not all patients with spasticity exhibit clonus, which of course limits the usefulness of the method.

Leavitt and Beasley [1964] described studies on one subject with spasticity, in which the patient's leg was moved about the knee manually by the investigator. Two recording tensiometers, one in each hand of the investigator, were used to record the forces required to move the limb. An electonic goniometer measures joint position, while EMGs were recorded from the flexor and extensor muscles of the leg about the knee joint. This system lacked the ability to provide reproducible stimuli to the subject and ignored the dependency of the muscle resistance on velocity of stretch.

Agate [1956] came up with an electronic apparatus known as the "Rigidometer" for the quantificatin of hypertonic changes in the arms under the impact of medication. It was considered a "precise measuring instrument... but required highly complicated calculation including calculus for an exact determination of the change in hypertonicity..." Basically, the device measured the torque position relationship during arm movement about the elbow joint. No attention was given to the velocity of movementand no EMG was recorded. Changes in rigidity were determined from changes in the torque versus position traces. Dr. Agate made successive modifications in the machine, including an abortive effort to adapt it to leg measurements, and no further studies were reported.

In comparison to the Agate device, the gravity driven ergograph described by Timerlade [1964] was somewhat simplistic and unsophisticated: passive movement at the elbow joint was produced by dropping weights which provided the power to move the limb via cables and pulleys. The amount of movement in response to the weights applied was assumed to reflect inversely the amount of spasticity. No conclusive results were obtained since the marked variability in the data did not allow for any meaningful analysis.

Long et al [1964] reported a method of recording the stiffness of a finger. The stiffness was recorded in terms of torque resisting a sinusoidal position change, imposed by a constant speed motor though a double parallelogram linkage. The parameters measured were: peak torque, phase angle, area of force versus position plot (hysteresis loop), and the time constant of stress relaxation. A later study by Long [1968] described the use of the device for quantification of spasticity in the hand, but proved inconclusive and no further studies were reported.

Webster [1964] described an advanced version similar to the one used by Agate. The unit could alternatively flex and extend the limb in a horizontal plane and rotational velocities up to 40°/sec. Torque and position were recorded from a load cell and a potentiometer, respectively. The signals were integrated to obtain an average net work required to complete a cycle of movements. The total work value, averaged over slow, medium, and fast velocities of stretch, was termed the "index of spasticity". The results were very promising, and some drug studies were conducted using this device. However, the limiting range of available velocities and difficulties in positioning of the subjects proved to be a major drawback of the system. A similar device, the "Rotational Joint Apparatus" (JRA) was reported by Herman et al [1967]. It was designed to measure tension length relationships of human muscle about the ankle joint. The same author [1970] applied the device to study the myotatic reflex in hemiplegic subjects. Although the unit was used primarily in physiological investigations investigations, and the results were ambiguous, it indirectly contributed to the advancement of the state-of-the-art in clinical measurement of spasticity.

Bomze [1972, 1973] developed an electrical and hydraulic servomechanism system to move the subject's arm in a programmed manner. A digital computer was used as a controller and data processing unit. The parameters used were limb position, velocity, force and EMG. The system was very promising, though suggesting that patient data are highly variable. The system required more refinement and modifications, and investigations are believed to be still underway at this time.

As was mentioned earlier, some of the investigators tried to use EMG as a parameter. Since the exact knowledge of the relationship between EMG and muscle force was unknown, it was not useful, except for timing of muscular activity. This can be seen from attempts that have been made, utilizing EMG only, to study the mechanisms involved in the stretch reflex in man [Dmitrijevic and Nathan, 1967a, 1967b; Herman, 1970; Burry, 1967; Burke et al, 1971; Matthews, 1965; and many others]. Some investigators have been utilizing the electrically elicited Hoffman reflex (known as the H-reflex) which shows considerable overlapping of responses in normal and spastic subjects. The H-reflex is influenced by many factors and is difficult to handle. Among other things, it fails to measure changes in spindle properties. It is scarcely suitable for routine clinical use in the measurement of spasticity, though some advances have been made in this respect [Angel and Hoffman, 1963; Paillard, 1959; Olsen and Diamantopoulos, 1967; Hugon, 1973; and others].

Other investigators, like Neilson [1972a,b] and Burke and Ashby [1972], as well as Hagbarth and Eklud [1968], used vibration techniques where sinusoidal and stochastic loads were applied to a limb sustaining various levels of voluntary contraction. Frequency response characteristics and cross-correlation techniques of analysis were used. These methods were investigative in character and were not clinically oriented.

In summary, various attempts have been made in the past forty-seven years to objectively measure spasticity. All the devices have been useful to some extent and contributed to the understanding of the different aspects and characteristics of the mechanisms of spasticity. Nevertheless, most of the devices, if not all, have several operational deficiencies, partly due to the state of available technology at that time and partly due to the lack of knowledge about the mechanisms of spasticity. Most of these difficulties seem to be technical, for example, difficulty in the proper positioning of the subject and inability to provide reproducible stimuli. In addition, inability to control the physical and emotional condition of the patients and the maintenance of the same experimental conditions, i.e. temperature, noise, medication, daily activity of the patient, cutaneous stimulation, etc. tend to contaminate the data. All investigators reported a very high variability in the data to the extent that no conclusive findings could be made. However, clinical findings stress the fact that the amount of resistance encountered in response to passive stretching of the patient's limb is primarily velocity dependent [Vodovnik, 1970; Bomze, 1973] and secondarily dependent on muscle length, and the level of neural activity. Nevertheless, thus far, no final device nor definitive method has been suggested for the objective measurement of spasticity.

1.4 Previous Research done at the MVL

This current investigation is a continuation of previous work done in the Man-Vehicle Laboratory (MVL). The prior investigator used an existing modified arm manipulator, originally designed to investigate the dynamic response of the human neuromuscular system for internal-external rotation of the humerus [Allum, 1974]. The above equipment was modified to accommodate the pronation and supination of the forearm.

A few normal subjects and cerebral palsy patients were tested using this device. During the test, the subject was seated upright in a dental chair and grasped a spade like handle, his elbow resting on an elbow cup. This worked fairly well for the normal subjects since they were able to grasp the handle and align (more or less) their arm with the axis of rotation of the handle shaft. It did not work as well with the patients since their hands did not have the same degrees of freedom and maneuverability as the normal subjects due to their disease, corrective surgery and/or fusion of some of the carpal joints (a procedure used in severe cases). The results were highly variable and no meaningful and significant evidence could be extracted, except that in the patients, conscious relaxation took considerably longer than in normal subjects (see Chao, 1976, for more details).

As noted herein some means were required to drive the patient's arm passively (no active grip on the handle). An attempt was made to design such a device without much success. Different methods were used, but they had to be rejected primarily due to induced pain and discomfort at the wrist. There was also a large error in position of the arm due to skin rotation. During the course of evaluation of different designs of wrist cuffs, it was also found that EMG signals that were presumably measured from the pronator teres, were heavily contaminated by signals from other superficial and overlapping muscles. Measurement of the torque output and dynamic response of the torque motor also revealed poor performance to the extent that the data acquired was questionable.

Based on the above evidence, a decision was made to perform the measurement of the stretch reflex at a different joint with a more defined muscle pair (agonist-antagonist) and with the capability to accept torque without discomfort or pain. The ankle joint ad the major muscles associated with it, the gastrocnemius, the soleus and the tibialis anterior, were found to be appropriate. The mode of plantarflexion and dorsiflexion of the foot was considered over the previous pronation/supination mode. (Note: Patients with fused ankle joints will not be tested.)

1.5 Objectives of this Thesis

The long term goal of this project is to develop a clinical device and method which will assist the physician and physical therapist to objectively assess spasticity. The device will be useful in the evaluation of the effectiveness of therapy, and will assist in the early detection and diagnosis of spasticity. This system should be simple to operate and the results reproducible and easy to interpret. It should be as versatile as possible for use in the investigation of spasticity in a variety of ways and must be acceptable by the patient and the clinician using it.

The immediate objectives of this research are as follows:

1. Construction of the prototype experimental device.
2. Preliminary evaluation of the new design on a pool of normal and spastic subjects and suggestion of the proper refinements of the device and test protocol.
3. Provide a guideline for the automatic data analysis procedure, and derivation of the most suitable parameters for clinical assessment.
4. Installation of the equipment in the hospital for the actual evaluation of patients and correlation of the new objective measures with conventional clinical evaluation.

It should be stressed here that the emphasis of this thesis is on the design and development of the clinical instrument, rather than on the physiological investigation of spasticity and its mechanisms.

1.6 Outline of the Thesis

Chapter II describes the device which was developed for the measurement of spasticity. It covers, in some detail, the different components of the apparatus, instrumentation used in gathering the data, safety precautions and the interface between the subjects and the system.

Chapter III deals with the computer control of the system and data acquisition, display and analysis and gives a general description of the equipment and programs used.

Chapter IV describes the experimental protocol and presents the results obtained from experiments with normal subjects. The effects of fast, slow and sinusoidal displacements, with different velocities and initial conditions are presented and the physiological implications of these results are discussed.

Chapter V presents some data obtained from two spastic patients and shows the usefulness of objective assessment of spasticity.

Chapter VI summarizes this work and gives a few conclusions and recommendations for improving the system. Finally, further research is suggested.

CHAPTER II

TWISTER - AN ELECTROMECHANICAL FOOT MANIPULATOR

2.1 Introduction

For the assessment of spasticity, it would be useful to have equipment with which it is possible to expose a muscle to a well defined dynamic and static stretching. Since the exact parameters influencing spasticity are not known at present, the apparatus should be general and flexible enough to allow a variety of interactions with the neuromuscular system under investigation. Basically, the device should be capable of moving the tested limb in a similar manner as is presently done by the clinician. Since spasticity depends on the rate of stretch and amplitude of stretch, it is important that dynamic interactions be obtainable over a wide pertinent physiological range of velocity, position and torque of the joint, in order to be able to extract the most dominant parameters. As mentioned earlier, a variety of mechanical disturbances introduced to the musculature were reported in the literature. Constant velocity ramps, sudden steps in position, and sinusoidal functions are most common. The majority of instruments were too cumbersome and complex to be used in the clinic by a clinician. Thus, a system that would be able to accurately move the patients foot about the ankle joint with capabilities including and exceeding these features had to be constructed.

2.1.1 Design Criteria

To establish the necessary design criteria for the manipulator, parameters of the mechanical limits of the human foot about the ankle joint were obtained from the literature [Hansen, 1958; Kinkade, 1972]. The neutral position for the ankle is with the lateral border of the foot at 90° with the axis of the leg and in midposition as regards to inversion and eversion. The average limit of plantarflexiion ofthe ankle is 35° from the neutral position with the average limit of dorsiflexion at 20° from the neutral position. Thus, the average total travel angle of the foot about the ankle joint is 55°. The actual physiological range of movement during level walking is 20° to -25°. Other investigators [Herman, 1970; Kearney, 1976; Agarwal, 1977] have conducted their studies within smaller ranges: -30° to 10°, ±14° and ±4° respectively. Thus, taking into consideration the variability in the data, a total maximum range of 60° was chosen.

The average, voluntary induced, maximum ankle torque varies between 520 in-lb (58.8 n-m) at 30° plantarflexion, and 1600 in-lb (181 n-m) at 20° dorsiflexion (subjects were aviation cadets). These are fairly high values as expected, since the triceps surae are among the strongest of the human muscles. However, somewhat lower torque levels would be expected during the use of Twister since the patients would not voluntarily introduce maximal forces. As reported in related work, maximums of about 700 in-lb (79.2 n-m) were recorded in moderately severe spastic patients. Most normals and spastic patients did not exceed 200 to 240 in-lb (27 n-m). Therefore, to be conservative, a maximum anticipated torque of 1000 in-lb (113 n-m) was assumed. With respect to the maximum velocities required, it has been reported that velocities up to 150°/s were generally used to test spastic patients [Herman, 1970; Bomze, 1973]. The bulk of the experiments were conducted at speeds between 30°/sec and 100°/sec for ramp function inputs and up to 0.9 Hz for sinsoidal. The physiological range frequency response of the ankle joint for both voluntary oscillations and level walking shows a corner frequency of about 6.0 to 6.5 Hz [Agarwal, 1977]. It would be reasonable to assume that patients would have to be tested at lower frequencies than 6.0 Hz since their muscles' sensitivity to velocity is higher than in most normals. Thus, slewing rate of the loaded system should be about 400°/sec.

In addition to these basic design criteria, the system should have the following features. Simplicity - to avoid complicated manufacturing problems, since most of the components (mechanical and electronic) had to be "do it yourself". Easy to operate in the clinical environment without intimidating the willing clinician. Accurate enough to provide repeatable stimulus and data. Aesthetic and compact - to be accepted by both the patient and clinician, blending into the clinical environment. Rugged and moveable so that it may be wheeled around the clinic easily and safely. It must meet not only the hospital safety standards and regulations, but must be acceptable as safe by the patient and clinician alike.

Based on the above preliminary criteria, two different systems were considered: hydraulic and electrical. An hydraulic system would have been the best solution in terms of dynamic perfomrance and torque, but it was ruled out for several resons: (a) cost - it is about three times as expensive as an electrical system; (b) it would be cumbersome and bulky, especially the power unit (pump, accumulator, actuator, valves, hydraulic lines, etc.); (c) noise, from the pump and valves; (d) messy - requires constant maintenance, oil leaks; (e) not easily transferable - disconnecting hydraulic lines, etc.

An electromechanical position servo system was chosen as the main drive of the foot manipulator, since it is less expensive, less cumbersome, compact, quieter and easy to handle and maintain.

2.1.2 General Description

Figure 2.1 is a photograph of "Twister", the foot manipulator. The patient sits on the dental chair (in a comfortable position), his foot strapped to the adjustable pedal. The pedal is connected on one side to the drive shaft and on the other side to an idle shaft, which in turn is connected to a position potentiometer. A strain gauge bridge is mounted on the drive shaft to monitor the torque. The shaft is driven by a quadrant pulley, cable driven by a permanent magnet DC torque motor. Enclosed in the instrument box are the servo control, safety logic unit and a rechargeable battery. The system is fixed to an adjustable plate, allowing an angle to be set from 0° to 90° relative to the supporting cart. The power unit is placed in the large box contained within the cart's frame. The entire system is mounted on rollers and can be moved about.

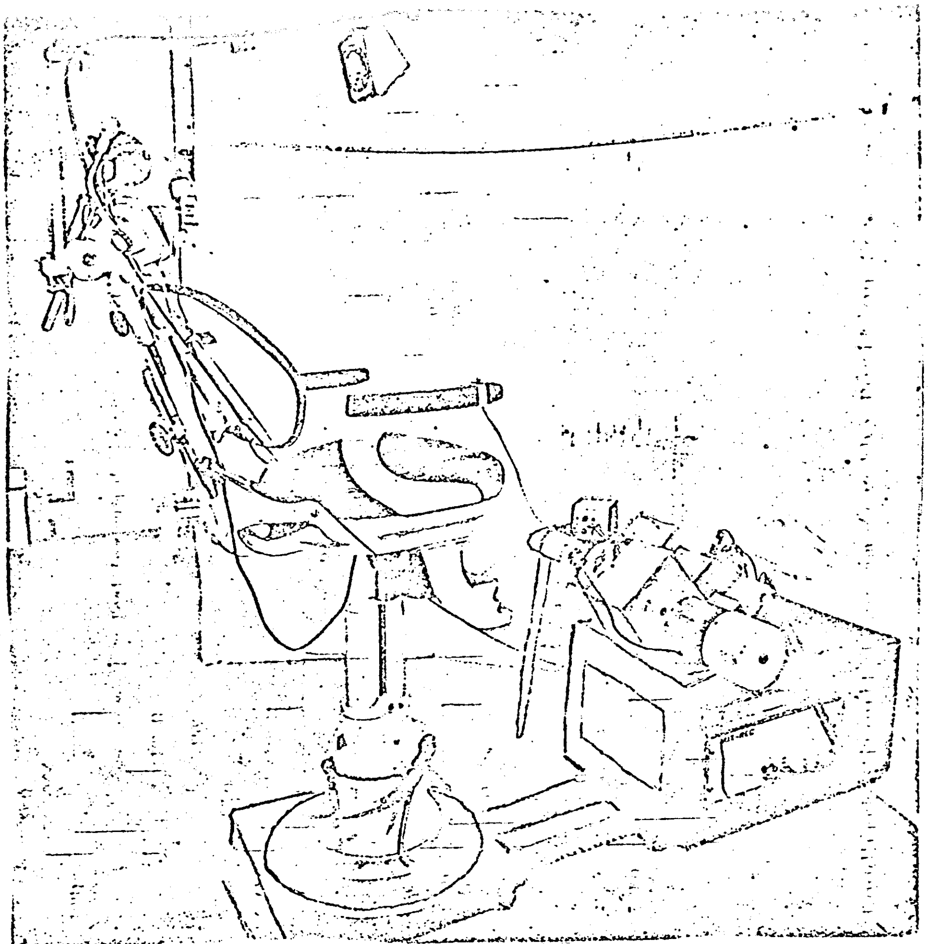
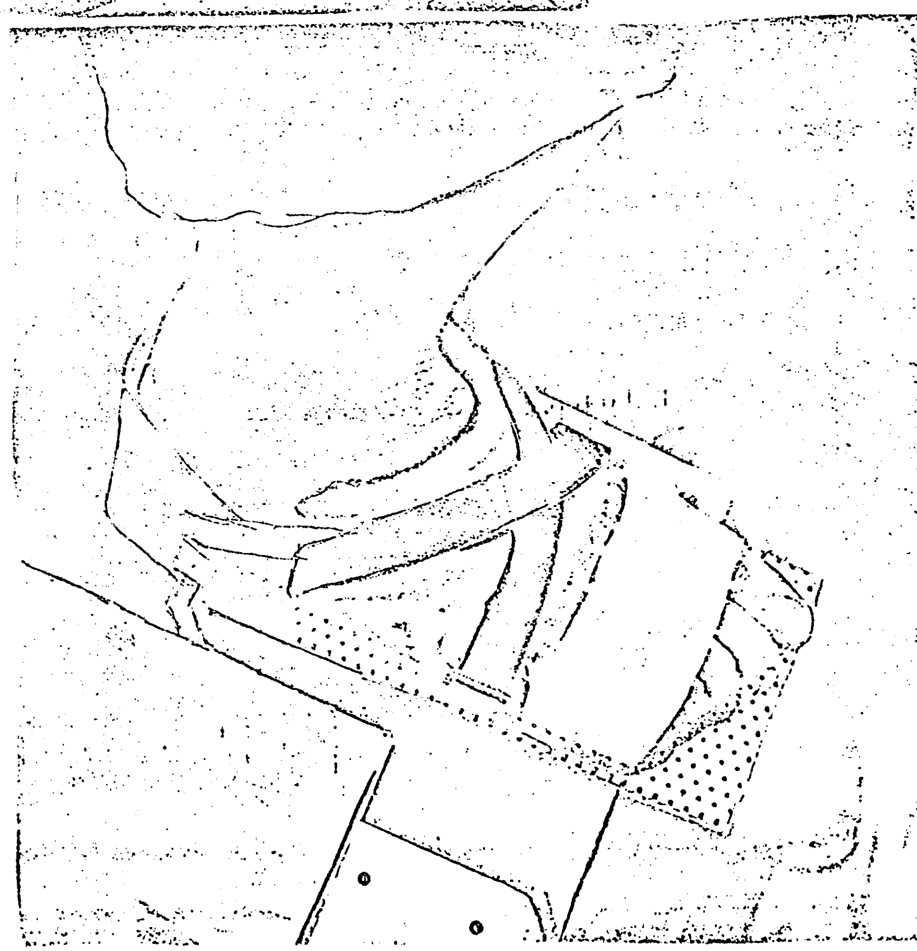


Figure 2.1:
General view of
"Twister", the
foot manipulator

Figure 2.2a:
Close-up of the pedal
showing a subject's
foot strapped in
place



2.2 Mechanical Structures

2.2.1 Supporting Frame

The supporting frame can be divided into two structures: the upper structure consists of a large base, a flange-mount for the torque motor and two bearing supports to accommodate the shafts with the pedal. The entire structure can pivot to allow adjustments in the position of the leg and the foot (for testing at any position between 90° flexed leg about the knee joint to fully extended leg). All parts are made of aluminum alloy, pinned and bolted so that they may be taken apart and put back together without disturbing the alignment. The cart is made out of "Dexion", modular perforated L-strips, bolted together, and carries the upper structure and the power supply unit, on wheels. There are two perforated angles extending from underneath the cart to allow adjustments of the distance between the dental chair and the pedal. It also provides a closed-frame structure between the cart and the chair that takes up the reaction forces and prevents relative motion. (See drawings in Appendix A.)

2.2.2 The Pedal Assembly

Special attention was given to the design of the pedal. It is strong, yet lightweight for low inertia. It accommodates different size feet and holds them in place with minimum relative motion between the foot and the pedal. It is adjustable in both vertical and horizontal directions to enable proper adjustment and alignment of rotation. This is only an approximation since there is no one dimensional perfect axis for the ankle joint [Isman and Inman, 1969]. Of great importance is the maintenance of heel contact throughout the experiment, since plantarflexion of the foot tends to raise the heel. The restraining device for the foot is quick and easy to apply and covers a wide range of different size adult subjects, male and female, with minimal discomfort or pain.

The pedal assembly consists of two segments, the pedal and the foot fixation elements. The pedal is made of perforated aluminum sheath 1/16" thick, bent and reinforced for strength. It rides within a U shaped aluminum frame via elongated slots which permit 2.5" of vertical travel. The pedal length can accommodate up to size 11 feet.

The usual methods to fixate a limb reported by several investigators is to use a molded cast, individually fitted to each subject. This method is very time consuming, requires some degree of expertise, accommodates only one particualr subject and does not allow for variations in the size of the limb due to swelling of body tissues. Others use velcro straps or inflatable cuffs which do not eliminate all relative movement.

Thus, a different type of foot holder was designed. Figure 2.2 shows the foot holder. It consists of a symmetrically moulded heel cup that fits (with or without wedges) over the calcaneum at a level under the malleolus. This holds the heel down firmly and comfortably. To "lock" in place the talus, navicular, cuboid and the three cuneiform bones, two nylon crossed straps with velcro are used. A piece of sheepskin is used under the strap for cushioning. Holding down the rest of the foot (the metatarsals and phalanges) is a 2" wide velcro strap.

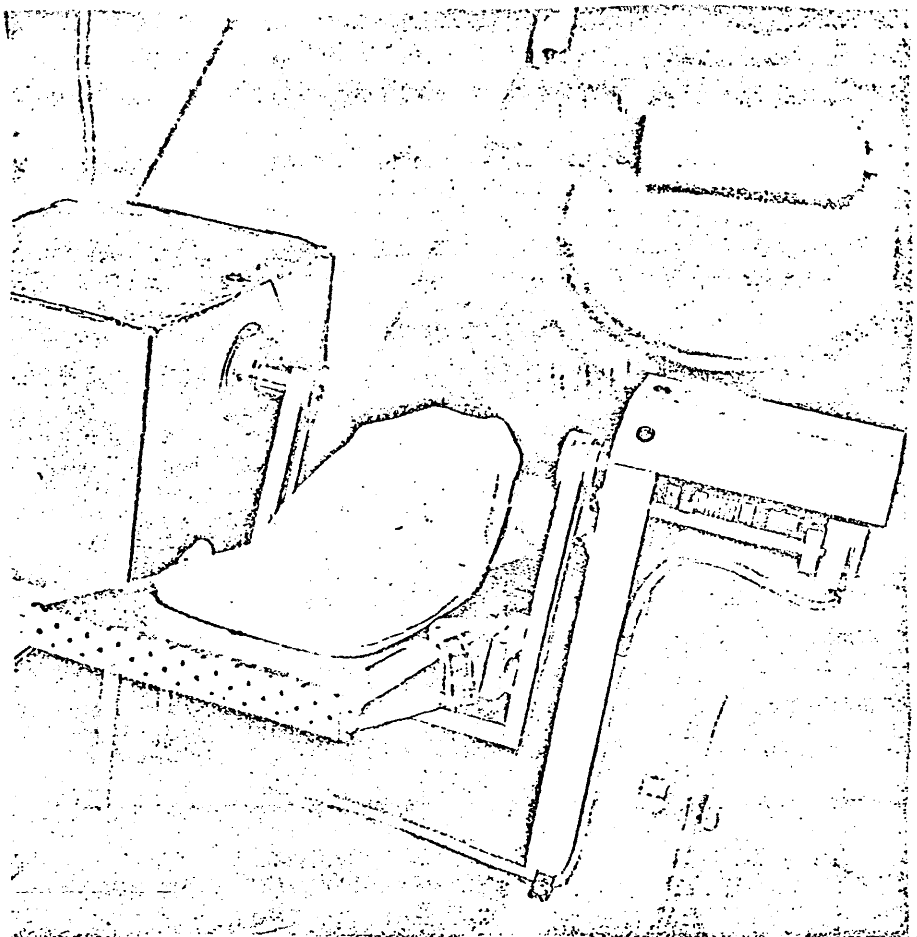


Figure 2.2b
Close-up of the pedal showing heel cup and foot rest

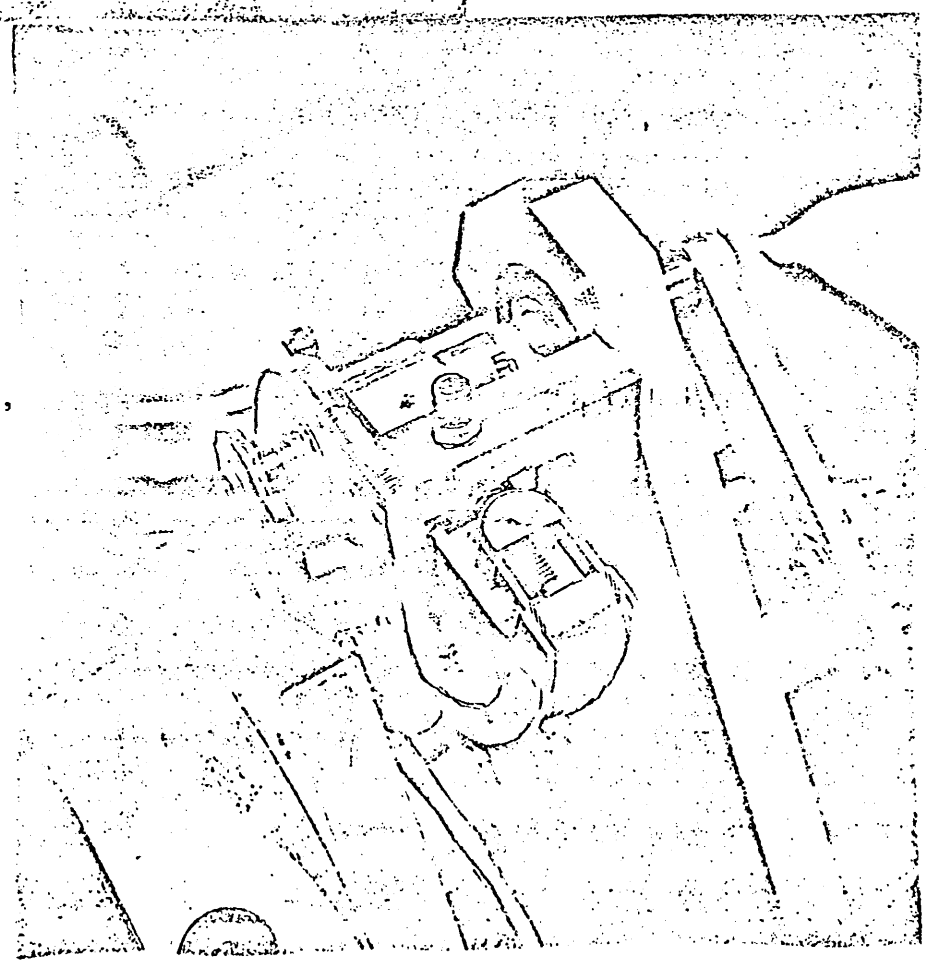


Figure 2.3:
A view of the drive train, cable tensioning device, mechanical stops, cam assembly with limit switches and torque transducer

The heel cup has a flat extension to which two screws are attached. This allows the unit to slide on top of the pedal forward and backward in 1" long slots for horizontal adjustment of the ankle center of rotation. Tests have shown that the foot can be strapped in the pedal for about an hour without the subject feeling any discomfort, pain or edema.

2.2.3 The Drive Train

Since the motor's maximum torque was less than the maximum anticipated, the motor could not be directly coupled to the foot manipulator drive shaft. A 1:7 torque amplification was required. At first a gear system was considered, including worm gears, however, it was ruled out for several reasons: (a) susceptible to backlash, (b) large in size and weight (for the torques transmitted), (c) requires good alignment (accurate machining), (d) introduces increased torsional elasticity in the system, and (e) requires lubrication to reduce friction and wear. A second consideration was gear belts and pulleys. Although it is slightly better than the gears, the size and weight of the components is very large due to the transmitted torque and velocity required, that effects the size of the belt and its allowable radius of curvature. Therefore, a third solution was adopted: a cable pulley combination. Since the foot motion is limited to 60° total range, a section pulley is sufficient.

Figure 2.3 shows the pulleys that were designed to yield practically backlash free motion. It consists of two, pretensioned, stainless steel aircraft cables, 1/6" in diameter, stretched around the surface of two pulleys. The system allows slight misalignment in both the vertical and horizontal planes and is small in size with a high torque to inertia ratio. The torque is transmitted from the motor via the cable pulleys through a semihollow stainless steel shaft (which also serves as the torque transducer) to the pedal assembly. Adjustments of the tension of the cables can be made via the cable tensioning bolts (see Appendix A).

2.3 The Electromechanical Position servo Mechanism

2.3.1 General Description

The electromechanical position servo mechanism is shown schematically in Figure 2.4. The position command is compared with the feedback from the position potentiometer and tachometer, the error is amplified by the linear power amplifier which drives the DC torque motor to the desired position. To reduce the sensitivity of the system to disturbances and parameter variations, and to have wide system bandwidth, a high gain in the open loop transfer function is desired. Such a system will appear as a "stiff" system that can follow a position command despite disturbing torques. Consequently, as the gain is increased, the system tends to become unstable. To overcome this difficulty, a tachometer feedback is used in addition to the position potentiometer. The electronic part of the system is hard wired on vector board placed in a 5" x 5" c 4 1/4"

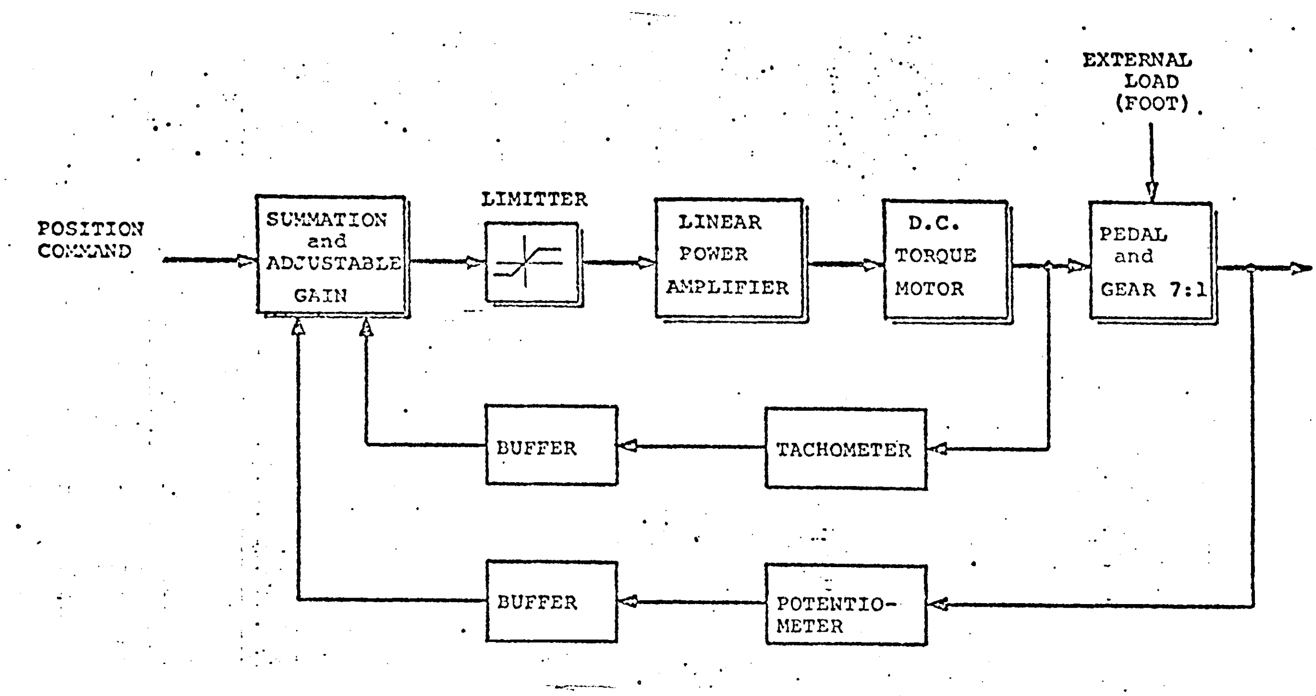
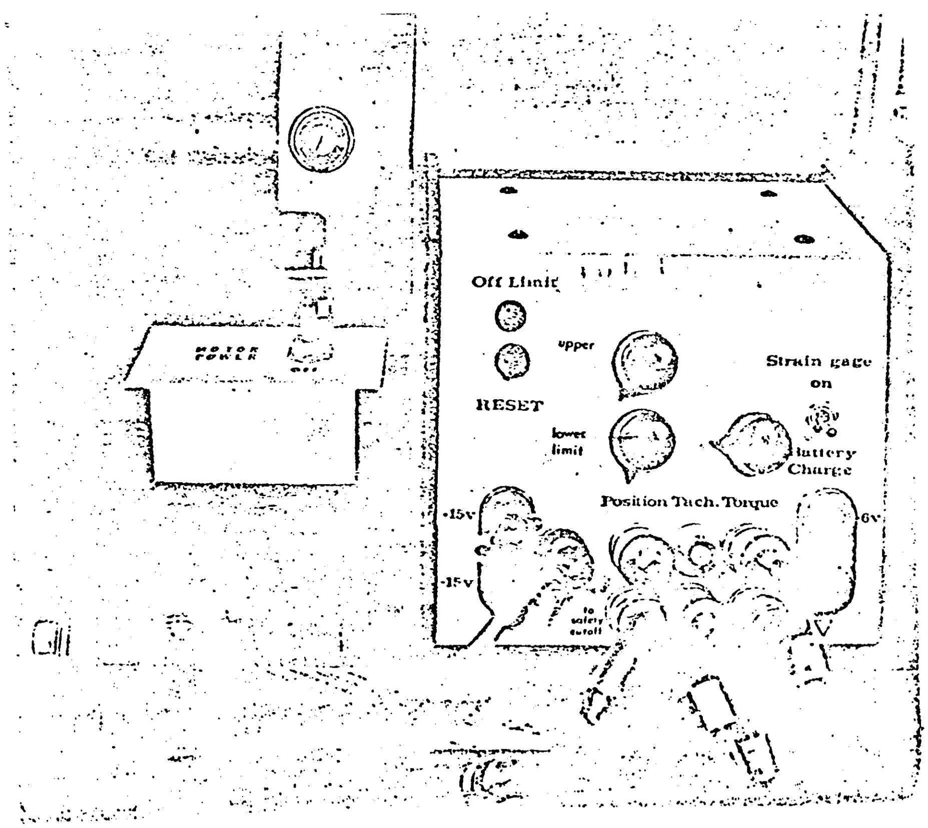


Figure 2.4: Block diagram of the electromechanical position servomechanism

Figure 2.5a:
Front panel of the
hard-wired control
circuitry box



sloped BUD instrument cabinet (along with other circuits) shown in Figure 2.5.

2.3.2 Motor and Power Amplifier

The foot manipulator is driven by a direct current permanent magnet servo motor (Magnetic Technology P/N H5500-750-040) capable of producing 14 ft lbs peak torque and no load speed of 55 rad/sec (see Appendix A). With the 1:7 gear ratio, the required torque and velocity can be achieved. It was determined that the power required to drive the system is about 400 watts. To meet this power demand, a 525 watt linear power amplifier was chosen (Torque Systems, Inc. Model PA-601), along with a matching power supply unit (Torque Systems, Inc., Model DPS-40-25. Figure 2.6 shows the hookup configuration of the amplifier.

Note 1: Both sides of the motor shaft were modified. On the output shaft, a flat surface was machined to secure the driving pulley onto the shaft. the rear end ofthe shaft was machined to accommodate the added tachometer.

Note 2: A pulse width modulated power supply was considered at first, but it produced excessive audible noise with changing pitch when the motor was running. This was a disturbing factor, particularly in the context of measurement of spasticity, which is known to be affected by ambient conditions, such as auditory noise.

2.3.3 Feedback Circuitry

The feedback circuitry (Figure 2.7) consists of three stages. The first stage is a set of pre-amplifiers that allow individual adjustments of the position feedback open loop gain and velocity feedback loop gain. The adjusted feedback signals are summed up with the command signal in the second stage. In addition to the summation at this junction, there is an adjustable gain and bias. The output of this stage is the is the error signal which goes through a simple diode bridge limiter that can be adjusted to limit the maximum velocity of the pedal. (Note: This is optional. At present, with the existing power amplifier, this limiter is not being used.) The circuit is built on a 1 1/2" x 2" vector board. The velocity transducer is a permanent magnet DC tachometer (Servo-Tek SA7247A-2, Size 11) with a sensitivity of 7V/1000 rpm (0.67V/rad/sec). The tachometer is coupled to the motor shaft via a bellows coupling to allow slight misalignments between the tach shaft and the motor shaft. (The position transducer is discussed later in this chapter.

2.4 Electromechanical Transducers

The kinesiological parameters of interest for possible assessment of spasticity are required as output information from the system. The parameters of interest in this study are (1) the angular position of the foot (and its first derivative, i.e. angular velocity); (2) the resistance encountered during the stretch of the muscle, i.e. the torque produced at the ankle

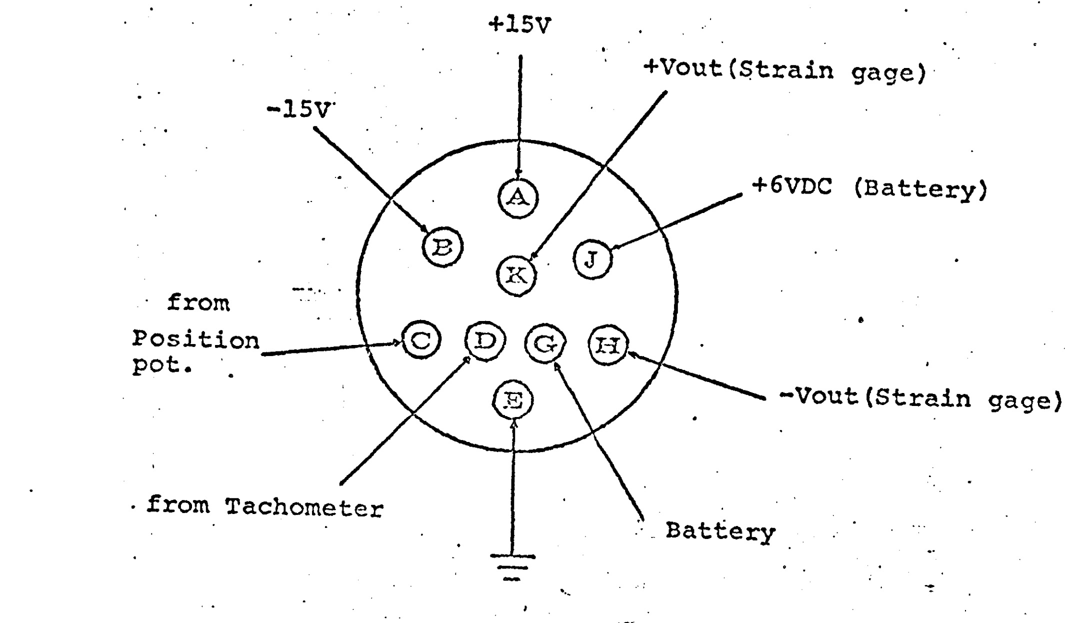


Figure 2.5b: Layout schematic of the control box connector

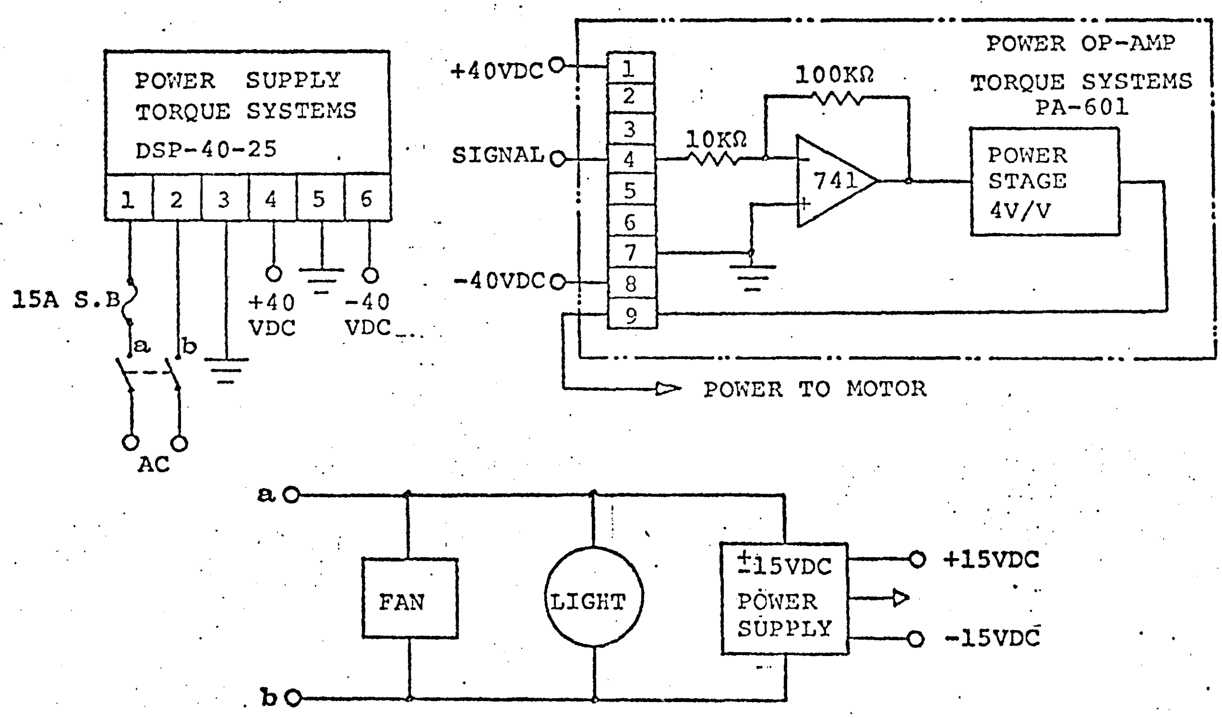


Figure 2.6: Linear power amplifier and associated components; schematic circuitry

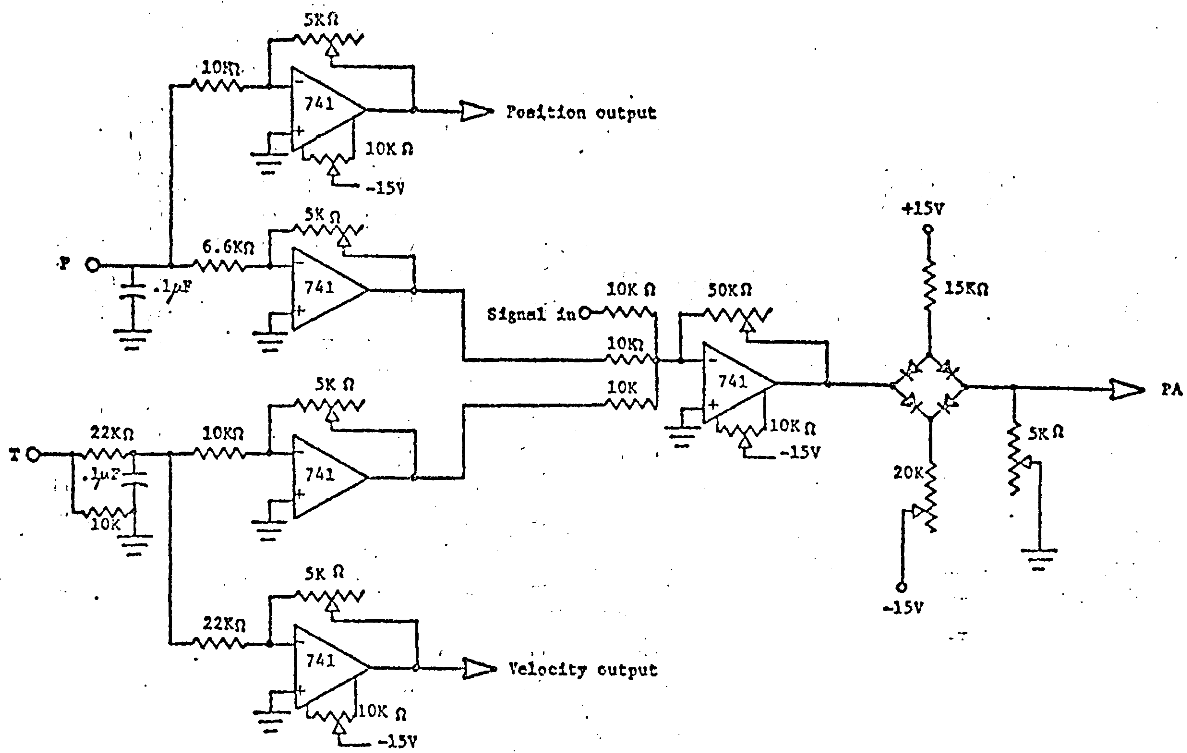


Figure 2.7: Feedback circuit diagram for the position servomechanism

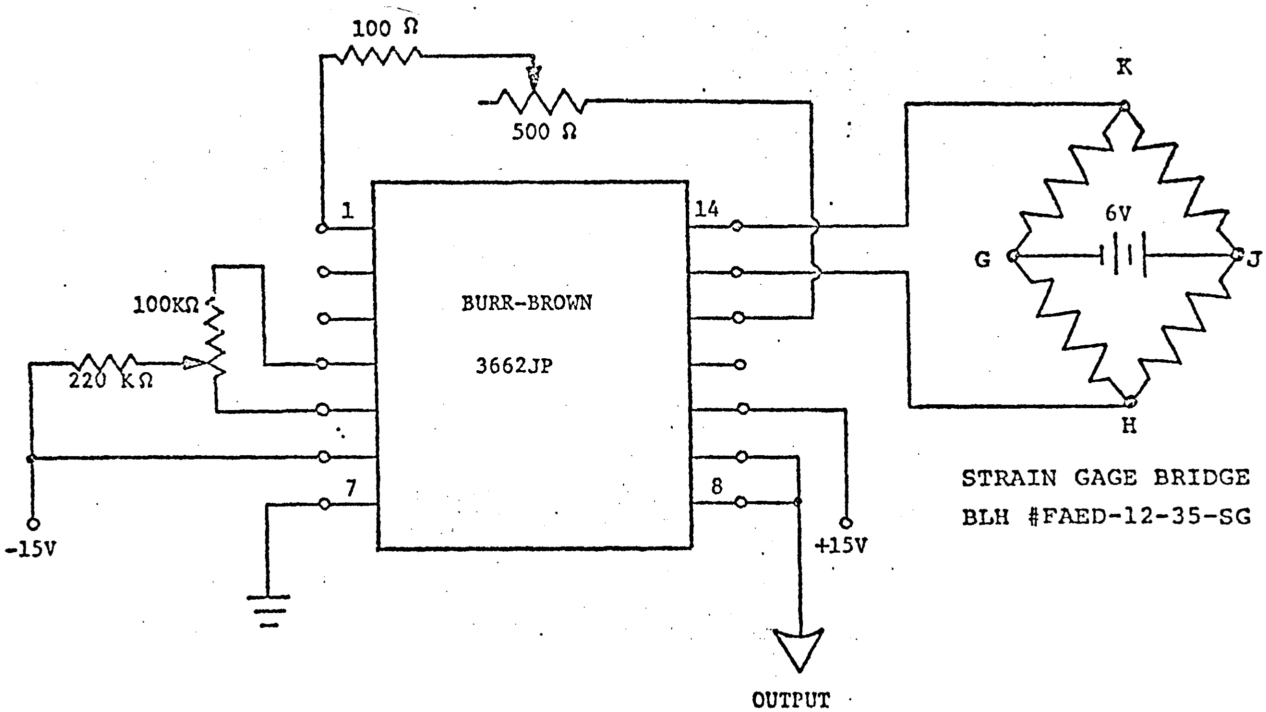


Figure 2.8: Strain gauge bridge and amplifier

joint; and (3) the state of activity of the agonist-antagonist muscle pair, i.e. the EMG of the triceps surae and tibialis anterior. The transducers and circuits used to measure these variables are described in the following sections.

2.4.1 Measurement of Angular Position

For the position feedback and measurement of angular position, an infinite resolution plastic film potentiometer is used (Bourns Model 6674, Custom Precision) with resistance of 5 Kohms and linearity of 0.25% of full schale. Voltages of ±15 volts, symmetric to ground, were applied to either end of the resistive elements so that a full turn produced a 30 V change on the slider. The 5K ohm was chosen to compromise between the amount of current drawn from the power supply and the undesirable variation of output impedance with position. The potentiometer has an actual mechanical travel of 300° ± 5°. Since it is directly coupled to the idle shaft of the pedal via a bellows coupling, its travel is limited to a maximum of ± 30°. In order to minimize non-linearities, the potentiometer signal is fed through a high input impedance buffer. The gain between the potentiometer output to the computer and the angular postiion of the pedal was measured as 0.0235 V/degree. Total travel will, thus, produce ± 0.705 volts at the computer A/D input. This is desirable since the maximum input to the present computer is ± 1 volt.

2.4.2 Measurement of Torque

To measure the torque produced between the motor and the foot pedal, a strain gauge bridge torque transducer was designed. The transducer consists of a hollow round stainless steel shaft, two pairs of 45° rosettes, a voltage supply, and a high gain CMR amplifier.

The two rosettes (BLH Model FAED-12-35-SG) are affixed to the shaft with clear cement (BLH #EPY-150). They are placed in diametrically opposing potentions on the shaft with high precision. This symmetrical configuration virtually nulls all effects of bending strain, tensile or compressive, and measures only pure torque. Temperature effects are also eliminated by this configuration as well as other sources of electrical noise. Sensitivity of the torque transducer is a function of the strain in the shaft section. It is given by

$$\varepsilon_{45°} = M_t r_0/\pi G(r_0^4 - r_i^4) \qquad (2.1)$$

where M_t is the torque moment, G is the shear modulus and r_0 and r_i are the outer and inner radii of the shaft. The gauge voltage is

$$e = V_B \Delta R/R \qquad (2.2)$$

where R is the resistance of each strain gauge, V_B is the battery voltage applied across the bridge and ΔR is the change in resistance when the gauge is stressed.

$$\Delta R/R = GF\varepsilon_{45°} \qquad (2.3)$$

where GF is the gauge factor. Thus, since there are four gauges, the total transducer voltage output is

$$V_T = 4\ V_B\ GF\varepsilon_{45^\circ} \tag{2.4}$$

With the particular shaft diameters, a given GF of 2.1 and an excitation voltage of 6.34 V, the transducer output is 6.71×10^{-6} volts/in-lb. This requires amplification of about 750 to achieve about 0.005 volts/in-lb. For this, a Burr-Brown low drift instrumentation amplifier (3662 JP) with high CMR has been used. Figure 2.8 shows the schematic diagram of the amplifier and strain gauges. (See Appendix A for the transducer shaft dimensions.)

The excitation voltage of 6.34 V is supplied to the strain gauge bridge by a rechargeable battery (GEL/CELL #GC 610-1B). The battery is placed inside the BUD instrument box and can be periodically charged by a battery charger (GEL/CELL #GRC-6150-CDE) via the "battery charge" socket. Usually the limit on bridge current is 30 mA. Since the gauge resistance is 350 ohms, 6.34 volts is a reasonable value for the bridge voltage (yields 18 mA). It is very important to keep the battery charged on a regular basis, in particular before experimental runs. The calibration curve for the torque is shown in Figure 2.9. The torques measured were symmetrical in loading-unloading and highly linear. A linear equation which relates the torque T to bridge output V, is:

$$T = 193.75(V - 0.039)\ \text{in-lb} \tag{2.5}$$

Accuracy is within 1% of full range. The torque measured by the transducer is comprised of a significant inertial component in addition to the torque is given by

$$T_{GAUGE} = T_{ANKLE} + (J_{FOOT} + J_{PEDAL})\alpha \tag{2.6}$$

where T_{GAUGE} = output of the torque transducer

T_{ANKLE} = torque produced by the ankle, i.e. viscous damping, stiffness, reflex and voluntary contractions.

$J_{FOOT} + J_{PEDAL}$ = moments of inertia of the foot and pedal,

α = angular acceleration

Usually, the inertial components are subtracted from the transducer's output. This is done by oscillating the subject's foot passively while the subject is completely relaxed, and measuring the angular acceleration and torque. However, in this apparatus, there are no provisions to measure accelerations nor to compensate for inertial loading. The inability of the patient to relax and the general characteristics of spasticity as mentioned in Chapter 1 make it impractical to measure the inertia of the foot. As a result, a brief transient, in the early portion of the torque response to fast movement is expected.

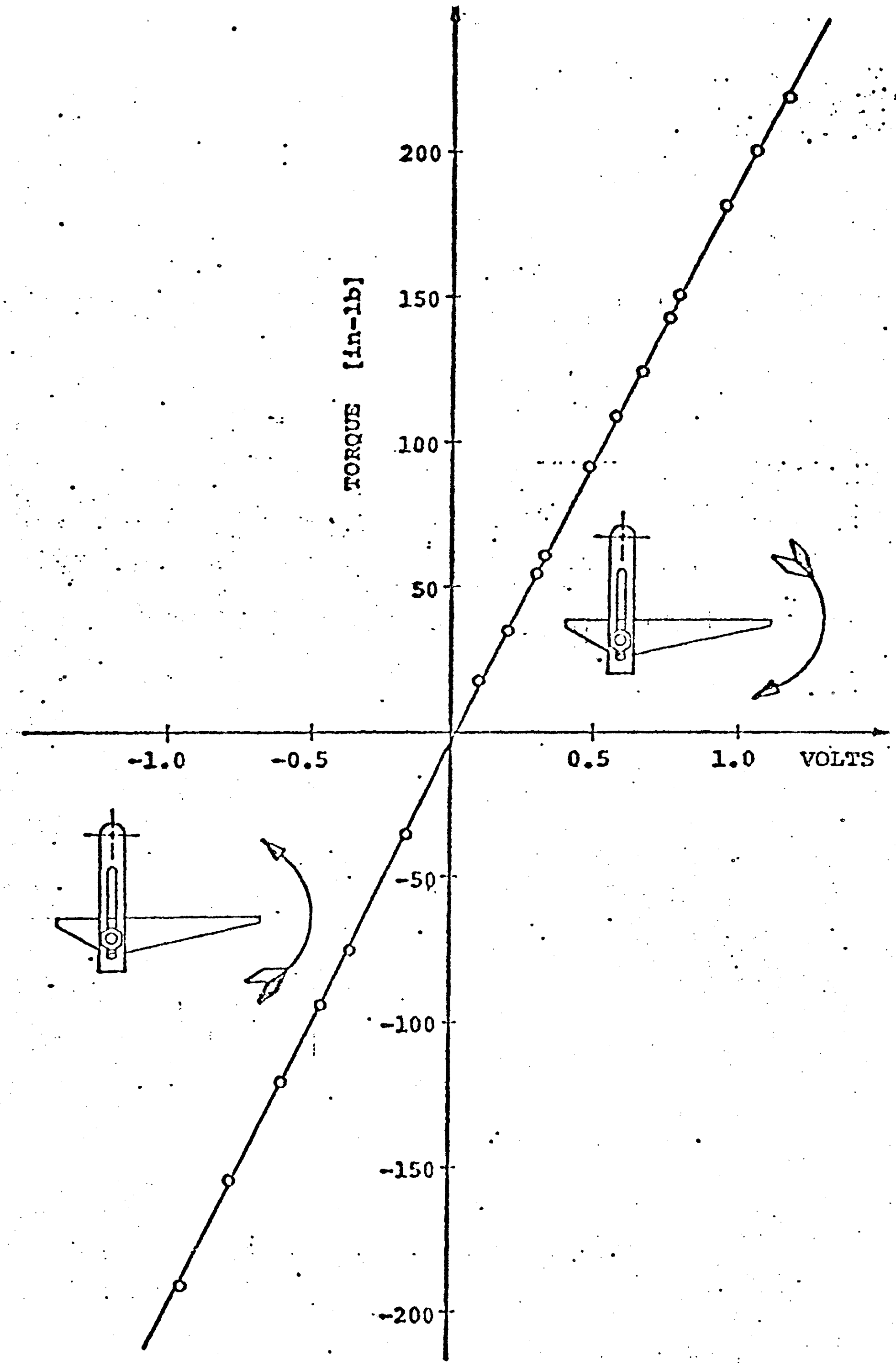


Figure 2.9: Torque Transducer Calibration Curve: Torque Applied to the Pedal Versus Output Voltage.

2.4.3 MES Recording

Assessment of the electrical activity of the muscle can be done through the measurement of the Myoelectrical signals (MES) produced during muscular contraction. Measurement of the MES is valuable for determining the levels of muscular activity during the experiment, the timing of these activities and the relationship of these parameters between the agonist and antogonist muscles. In particular, the MES records can reveal the different components of the muscle response to stretch, such as the onset of the stretch reflex and the timing of a voluntary response.

The MES can be measured either by intramuscular electrodes or surface electrodes. Intramuscular electrodes were not considered because of the complexity and risks involved in their use. These electrodes are invasive and cause pain, tissue damage, infections and require trained personnel to administer them. They also detect activity of a rather small number of muscle fibers which do not represent the overall activity of the muscle as a whole. Surface electrodes were selected for they are simple to apply, give good representation of overall muscle activity, are non-invasive and are readily accepted by the subjects. They cause pratically no discomfort and are safe to use.

The MES surface electrodes used are Jacobson [1973] PDP-3 (Project and Design Laboratory, University of Utah) (currently sold by MCI - Motion Control Inc.). They consist of an at-the-site differential preamplifier with unique features: No conductive electrode jelly is needed due to the high input impedance ($10^{11}\Omega$), high common mode rejection ration (CMRR $\sim$ 91db at 500 Hz) and low input bias current (25 picoamps), and insensitive to ambient electrical and mechanical disturbances. The two electrodes are physically connected and maintain constant relative distance from each other.

The small size of the preamplifier allows it to be placed on the skin directly above the muscle being monitored. It has a gain of about 310 and a bandwidth of 5 Hz to 17,000 Hz. The preamplifier is battery powered (Mallory Duracell #TR286, 12.6 volts) with 6.8 volts supply. This avoids the hazards associated with AC power and provides good electrical safety.

The raw MES is full wave rectified and then low pass filtered with a second order Butterworth filter which reduces the signal by 3 db at around 90 Hz. This provides a short time constant and allows fast changes in the envelope to be followed without introducing much phase shift. Figure 2.16 shows the MES processor and electrodes and Figure 2.10 shows the schematic diagram of the processor electronics. Note: The batteries of the EMG processor should be regularly checked for proper voltage otherwise performance might be impaired.

2.5 Safety Mechanisms

Twister is capable of applying forceful movements to the foot about the ankle joint. A whole hierarchy of automatic and manual limiters has been provided to ensure stringent safety precautions. Special measures have been taken to protect the subject from exceeding his maximum comfort range of travel (i.e. position) as described below.

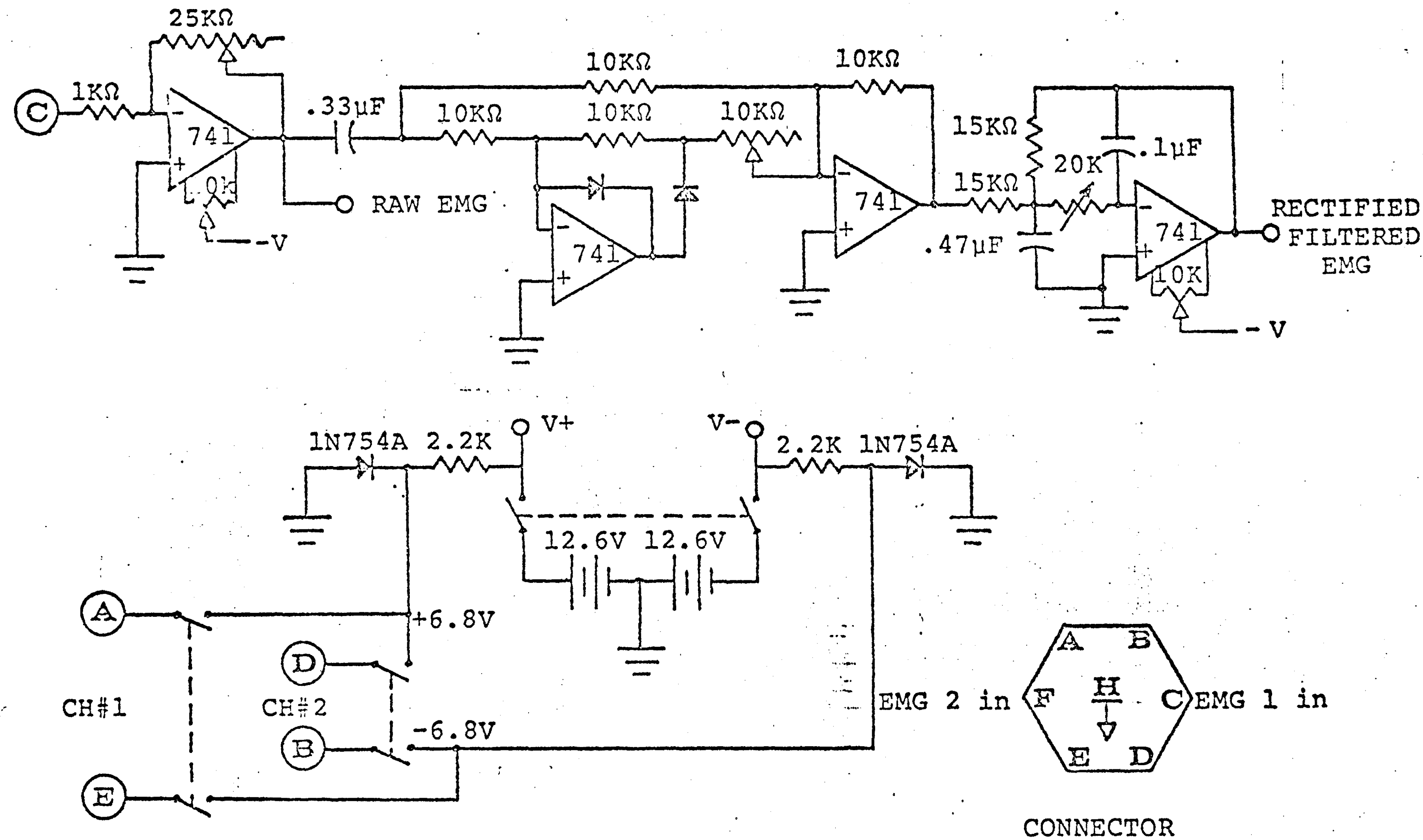


Figure 2.10: Myoelectrical Signal Processor and Electrodes Power Supply Circuit Diagram

The cable-pulling arrangement does not allow movements beyond 33 degrees either in plantarflexion or dorsiflexion. Any movement exceeding this limit would cause the cable to release from the tensioning device and the pedal will be physically disengaged from the motor. This is a very extreme situation. Usually, the pedal will stop, if anything else fails, via the large pulley section, against the mechanical stops (set for $\pm$ 30 degrees). This range can be adjusted to about $\pm$ 15 degrees by the two adjustment screws mounted on the mechanical stops' arms. Before the mechanical stop is reached, a switch will be triggered by an adjustable cam assembly (pick #PL-3), which will cut off the power to the motor. Currently, this trigger is set to operate at $\pm$ 29 degrees.

Ideally, these three safety levels would never be tripped. Generally, the electronic stop safety mechanism would take care of stopping the device in cases where the subject's limits are exceeded. A schematic of the electronic stop circuit is shown in Figure 2.11.

Two potentiometers allow the subject's upper and lower position limits to be set. The position of the pedal is compared by the voltage comparators (LM 311) to the preset limits. As long as the pedal is within the limits, the safety relay coil (Figure 2.12) is energized and power is supplied to the motor. Once the pedal exceeds one of the limits, the logical "OR" gate output goes to logical "HIGH" causing the set-reset flip-flop to reset which is indicated by the activated light emitting diode. Subsequently, the relay is deenergized and the power to the motor is shut-off. Once tripped, the system remains inoperable until a reset button is depressed. (The action is done manually and the operator must make sure that the pedal is at approximately its bias position, otherwise the pedal will move abruptly to that position.)

In addition to these position limits, a torque limiter is provided and can be set via the torque limiter knob. The torque should be set usually by asking the subject to push on the pedal at maximal force and adjusting the system to trigger at that level. Usually it is enough to set the system at about 250 to 300 in-lb. The torque limiter works in the same manner as the position limits.

Another safety feature is a manual "discomfort button" readily accessible by the subject which disables the system when triggered. This switch is mounted on the subject chair's arm and accidental triggering when the pedal is not in motion or loaded at a certain level, will not disable the system.

In addition to the above mechanical and electronic safety "hardware", safety provisions are included in the computer control "software". Subroutine "LIMITS" (see Appendix C) does not allow the program to continue unless the subject's lower and upper position limits are actually set. These limits are sampled by the computer and stored in memory. Then, the computer will not accept any keyboard commands that require positions outside the preset limits.

Usually, the initial position of the pedal at the beginning of the protocol is at zero degrees. If the bias level is at any other position, a built-in

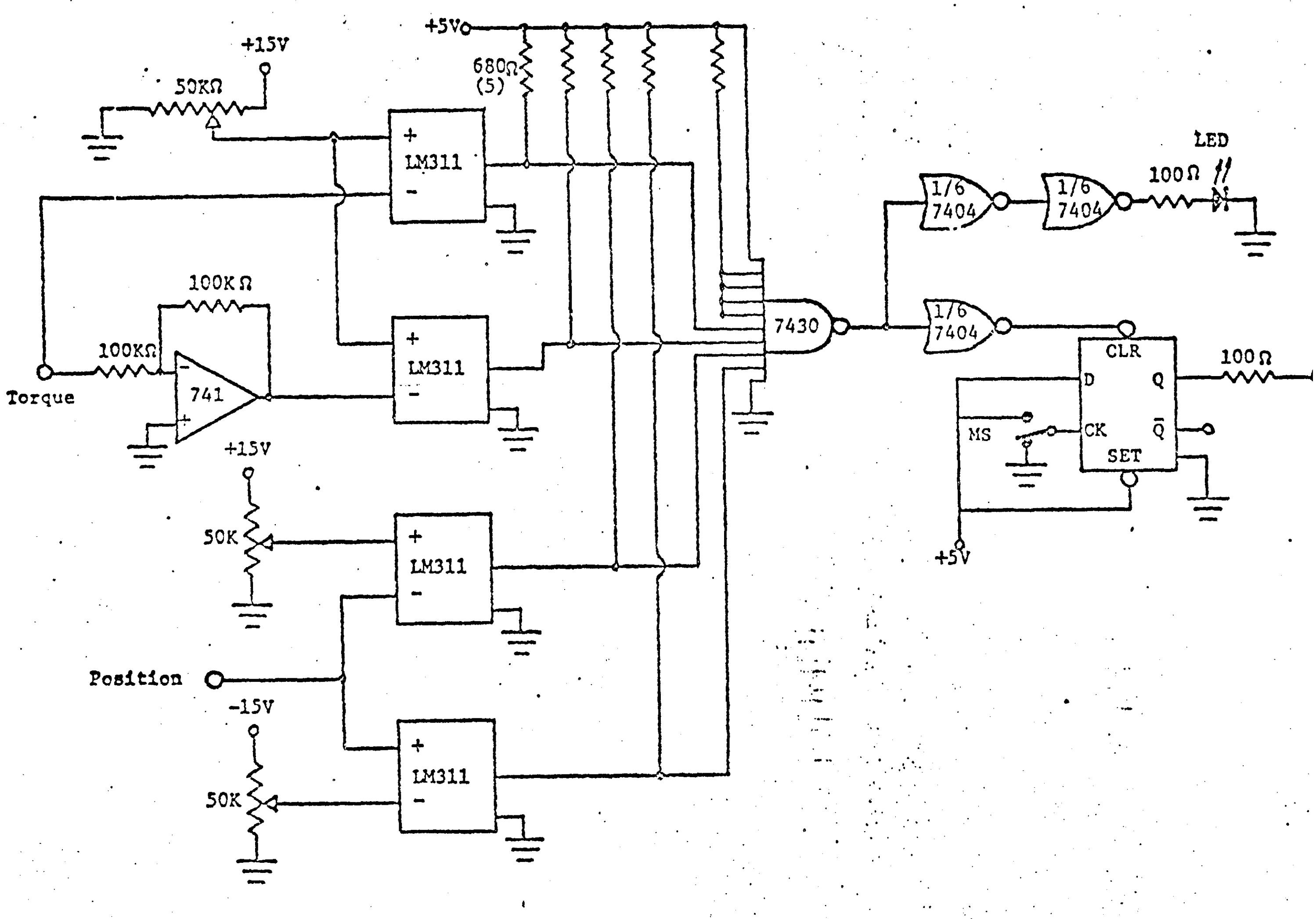


Figure 2.11: Electronic Safety Stop Circuit.

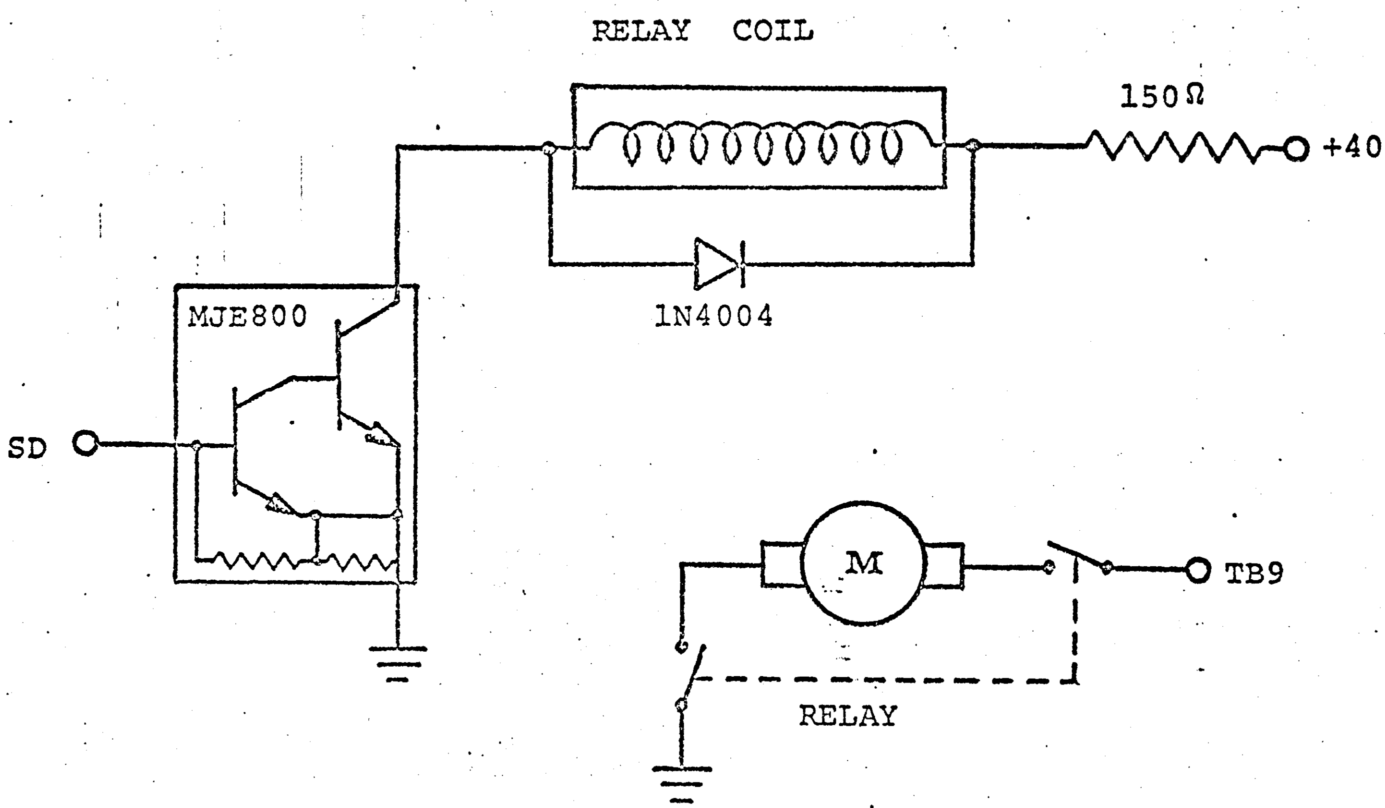


Figure 2.12: Motor Power Cut-off Relay Schematic.

subroutine "DAMPER" (see Appendix C) will prevent an abrupt motion of the pedal to the bias level. "DAMPER" will move the pedal slowly at a rate of one degree per second to the starting position.

For electrical safety, the subject is virtually insulated from the electrical power source. The foot rest and chair are insulated and the EMG electrodes are powered by a low 6.8 volts battery source. The power amplifier and $\pm$ 15 volts supply are isolated from the main 110 volt power line via transformers and the unit is connected to the wall outlet via a hospital electrical safety standards cable and plug.

2.6 Man Machine Interface

2.6.1 Positioning and Fixation of the Subjects

Figure 2.13 shows the general set up of the device with the subject seated in a typical test configuration. In order to assure proper fixation of the subject and proper alignment of the ankle axis of rotation with the pedal's axis, the following adjustments have to be make:

a. The chair's height adjusted such that the subject's thigh rests completely on the seat cushion and the heel of the tested foot comes in contact with the pedal.

b. The distance between the chair and the pedal should accommodate the leg length of the subject and provide an approximate 115 degrees angle between the leg and the thigh.

c. The supporting plate should be adjusted with the pedal at zero degrees to assure a 90 degree angle between the axis of the tibia and the pedal (the plate is set currently at about 40 degrees).

d. Adjustment of the back rest and head rest to the individual subject's comfort to allow maximum relaxation of the subject. (This should set the angle between the trunk and the thigh about 130 degrees).

e. Alignment of the talocrural joint axis with the pedal's axis. This is done by eyeballing the pedal's axis with the points 5 mm posteriorly and distally from the distal tip of tibial melleolus and fibular malleolus. This is an approximation of the anatomical axis as reported by Isman and Inman (1969). (see Figure 2.14)

f. Final adjustment is done by moving the pedal passively by hand and have the subject judge the "feel" of the alignment.

Slight lateral corrections of the ankle axis can be made by pivoting the chair in the horizontal plane. In some cases, strapping down of the thigh may be necessary. This can be done by a cushioned 2" wide seatbalt wrapped over the thigh and the lower structure of the chair's foot rest.

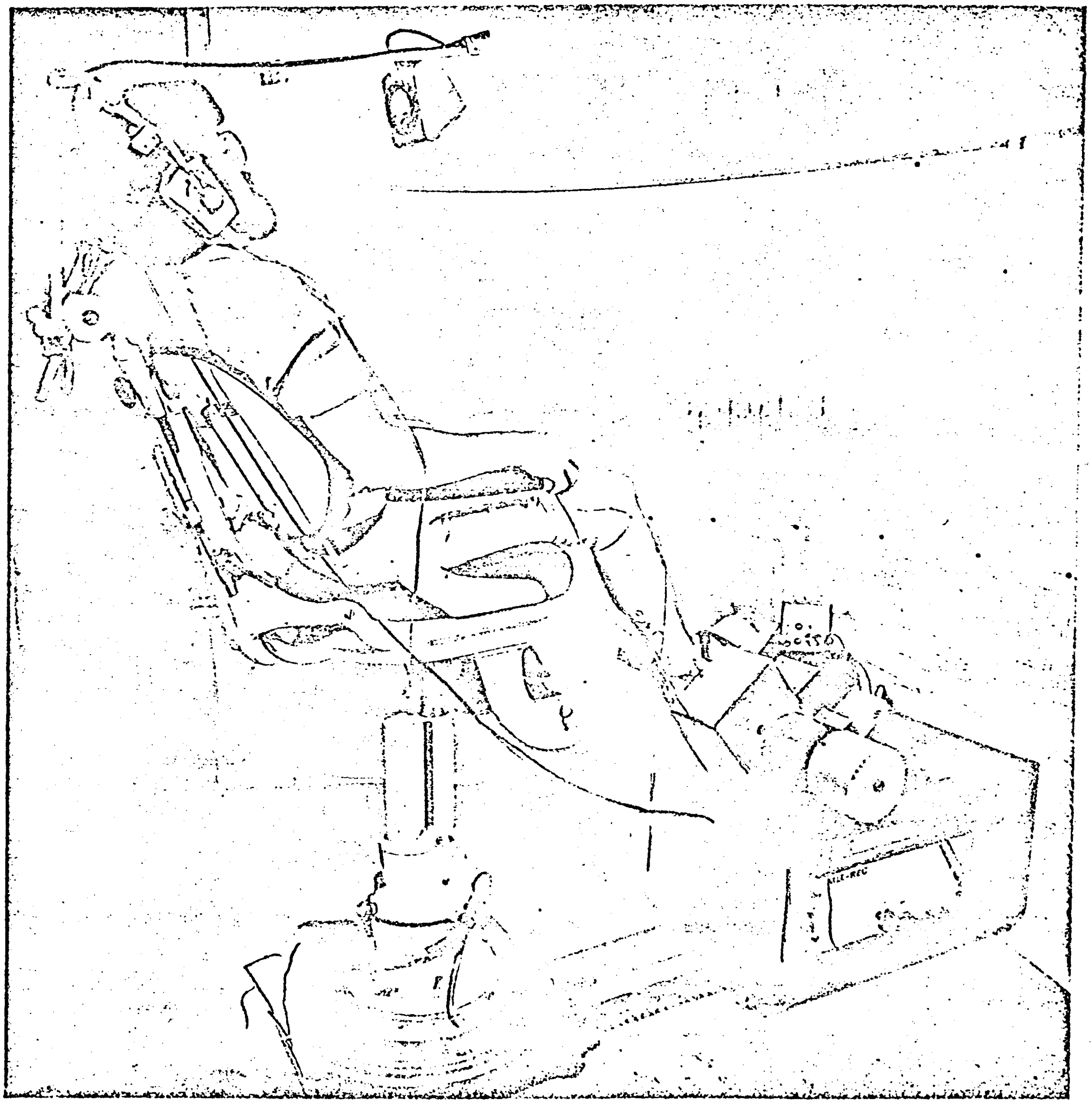
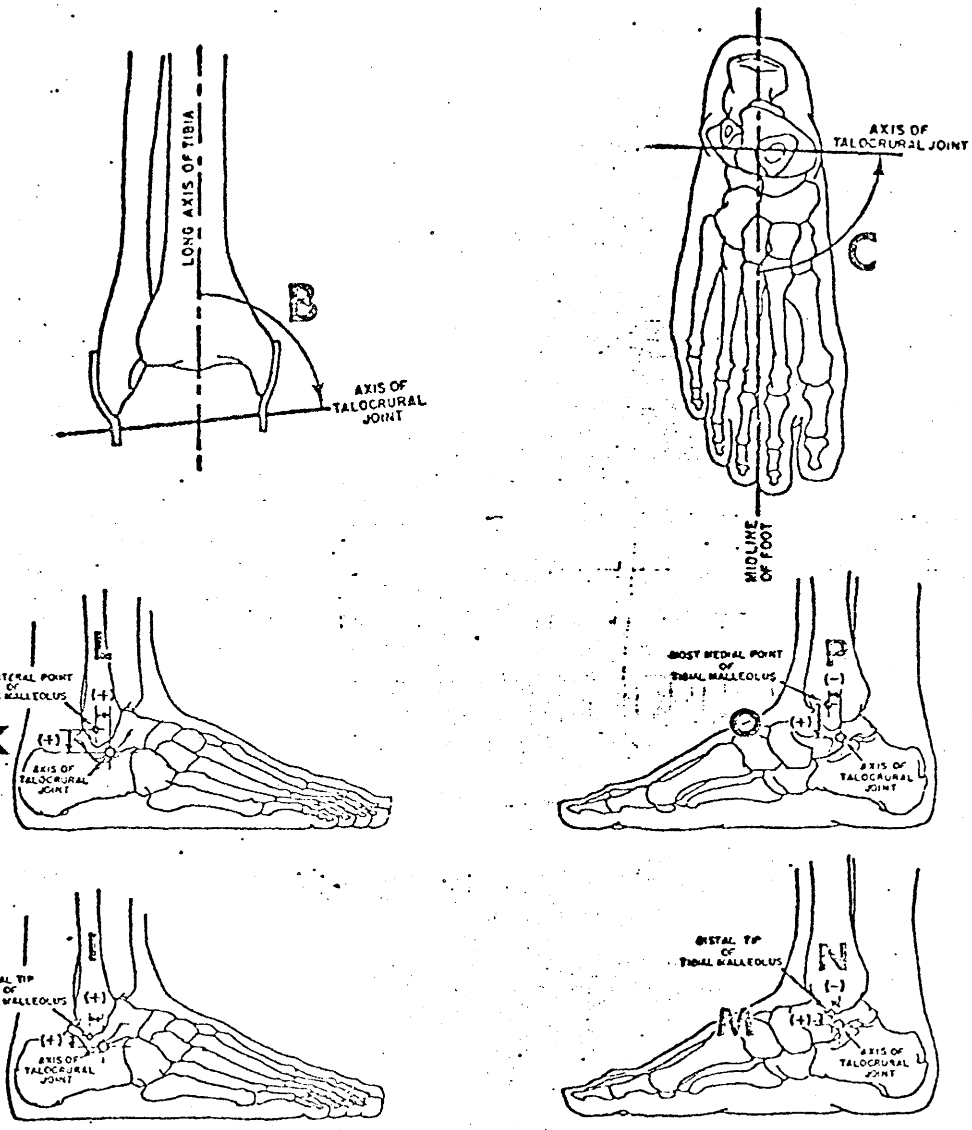


Figure 2.13: General View of the Experimental Set-up with a Subject Seated in a Typical Test Configuration.



$\overline{B} = 80^{o}$; $\overline{I}$ = 8mm; $\overline{K}$ = 12mm; $\overline{M}$ = 5mm; $\overline{O}$ = 16mm

$\overline{C} = 84^{o}$; $\overline{J}$ = 3mm; $\overline{L}$ = 11mm; $\overline{N}$ = 1mm; $\overline{P}$ = 1mm

Figure 2.14: Axis Location of the Talocrural Joint Relative to Anatomical Land Marks. (From Isman and Inman, 1969).

As mentioned in Section 2.3.2, the foot is strapped to the pedal via the special foot holder. When strapping the foot, care should be taken to avoid pinching or over-tightening of the straps. If the foot is seated too low, so that the heel cup does not grip on to the calcaneum, a special flat platform can be added under the foot. Wedges added between the heel and the cup at a level slightly under the malleolus will prevent slippage of the heel out of the cup. This assures good fixation of the heel with minimal discomfort or pain. In general, strapping of the foot is done just before the experimental runs begin, to minimize the time during which the subject is strapped to the pedal. Runs as long as an hour did not cause any discomfort to the subjects, though as good practice, runs should not exceed 30 or 45 minutes.

2.6.2. MES Pickup

Functional anatomy of the lower leg shows that the primary effectors of ankle joint extension (plantarflexion) are the gastrocnemius and soleus muscles. Their joint of insertion on the calcaneous bone is within the plane of the foot dorsiflexion - plantarflexion, and therefore produce pure extions of the foot. The gastrocnemius usually provides propelling force in walking and consists of high-force slow-reaction muscles.

The soleus, on the other hand, is known to exhibit continuous activity and is concerned primarily with steadying the leg in standing. Its postural function is emphasized more than its value as a prime mover, thus it can be related more to the reflex activity of the lower leg. The soleus is also close to the surface and easily distinguished. As a result, the soleus muscle was chosen as the site for measuring MES activity of the extensors.

The electrode is placed on the posterior aspect of the leg lateral to the tendo calcaneus, mid-way between the knee joint and the medial malleolus. Positive identification of the site is done by palpating the leg while the subject keeps his foot extended.

The function of the flexor muscles of the foot is somewhat more comples. The tibialis anterior inserts at a point medial to the foot. Therefore, it functions both as a dorsiflexor of the talocrural joint and inverter of the foot. Pure dorsiflexion can be reached only by coordinated activation of the tibialis anterior and peroneus tertius which is responsible for eversion of the foot. The tibialis anterior is closer to the skin and easy to locate. Thus, the second electrode is placed on the anterior aspect of the leg, lateral to the bitia 10-15 cm below the knee joint. Figure 2.15 shows the general relationship between the MES electrodes and the underlying structures.

A single reference electrode is placed either on the medial malleolus or the lateral malleolus which are located remotely enough from the other two electrodes. The ground electrode is a Beckman 2 mm diameter silver-silver choloride electrode. Prior to attachment it is filled with electrode jelly and adhered with an adhesive collar. The tibialis and soleus electrodes, although they may be used dry, are coated with jelly and attached to the skin with surgical tape #1525. The skin at the electrode site is washed thoroughly with alcohol preceeding application of the electrodes. Figure 2.16 shows a typical electrode placement and attachment.

2.6.3 Visual Display of Torque

In some cases, the experiment requires some pre-biased initial torque level. To accomplish this requirement, the subject is provided with visual feedback of ankle torque in the form of a microamperemeter dial. The gain on the dial can be adjusted, but is set so thateach division corresponds to about 10 in-lbs. The dial and its supporting frame can be removed when not in use.

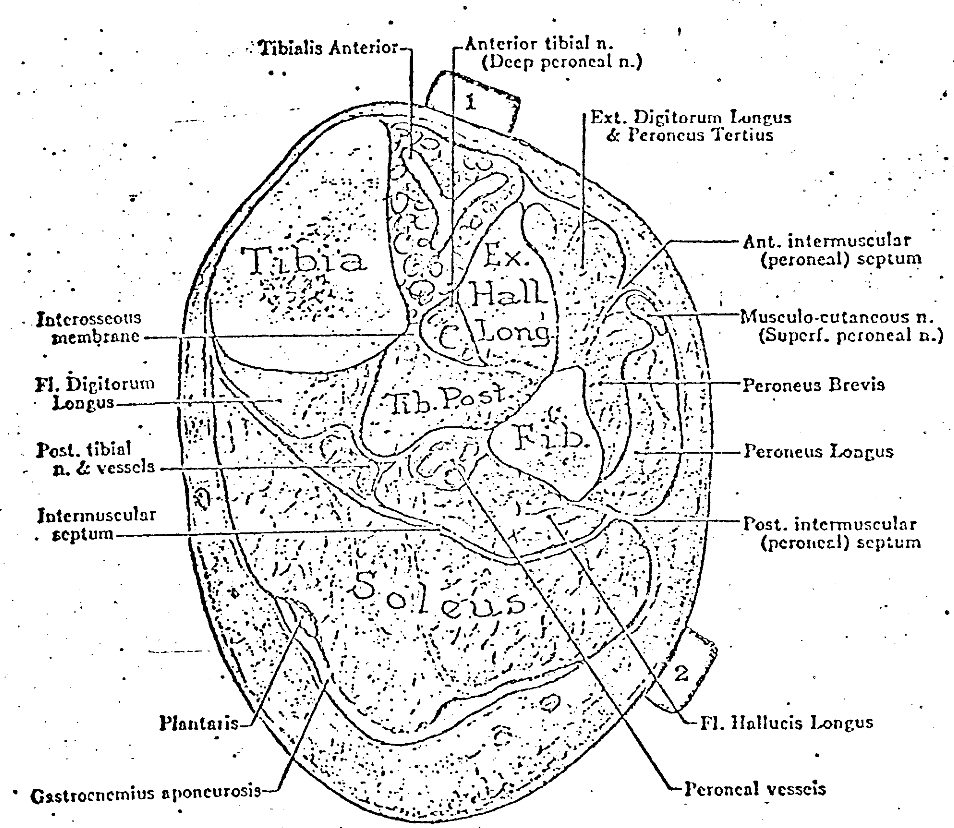


Figure 2.15: The location of the MES electrodes in relation to the underlying structure of the leg (from Grant). Electrode 1 picks up tibialis anterior MES and electrode 2 picks up soleus MES

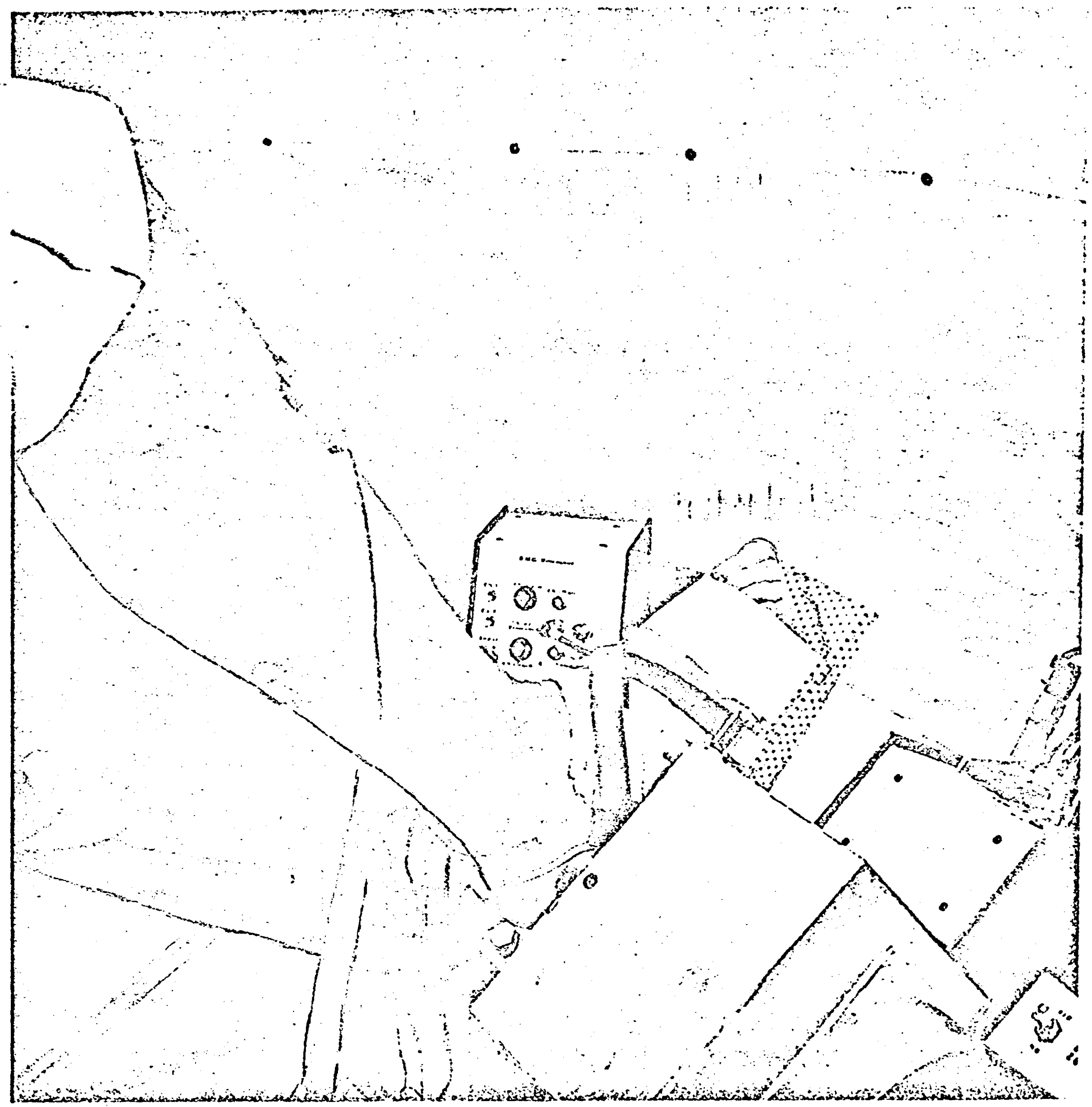


Figure 2.16: A Typical View of the MES Electrodes Placement. Note the Location of the Reference Electrode on the Fibular Malleolus. The Processor can be seen in the Background.

CHAPTER III

COMPUTER CONTROL AND DATA ACQUISITION

3.1 On-line Computer Control and Data Acquisition

The current research has made use of the facilities at the Man Vehicle Laboratory (MVL) located in the Center for Space Research. Figure 3.1 is a photograph of the computer and the peripheral equipment for data acquisition and analysis. Figure 3.2 shows the schematic block diagram of the interactions among the various components of the experimental apparatus.

The core of the system is a PDP-11/34 digital computer (Digital Equipment Corp.) which was used to control experiments, obtain data and store it on disks. The computer has 28K of memory plus two disk drives and a Lab Peripheral System unit with 16 channels of 12 bit, ± 1.0 volt analog to digital converters, 6 channels of 12 bit, ± 10.0 volt digital to analog converters, relays, and a programmable real time clock. This facility is compatible with the equipment used at the Boston Children's Hospital Medical Center (CHMC) Rehabilitation Engineering Center. Twister was designed to operate in the hospital's clinic and since the facilities were the same, it was more convenient to work on the programming and preliminary testing of normals and patients at the MVL. The computer interfaced with the other elements via a very convenient trunk line system. Quick and easy connections between different devices and the experimental room could be done via a patch board (see center of Figure 3.1). The experimenter controls the system through the LA30 Decwriter Data Terminal, a keyboard device. Output information is either displayed on the CRT monitor or printed on paper at the terminal.

The computer is programmed to select the mode of operation of Twister, provide the analog voltage command to Twister and sample the four channels of analog voltage experimental variables. The program "TWIST" is used to control the experiments. The data is sampled at a preselected rate, usually at 250 Hz. Up to 512 samples of each of the four channels are saved in the memory buffer and optionally, if so desired, written onto a file on disk. In conjunction with digital data storage on disk, the analog data can be stored four channels of FM tape for later processing. If desired, digital data can be dumped onto a digital tape for mass storage of data, leaving the disk free for additional files.

The data can be viewed on the CRT display via the program "SEE", or plotted on paper by the program "PLOT". Note: this was done because the MVL does not have a hard copier. The REC does have a Tektronics hard copier which is much faster than the x-y plotter, thus the program "PLOT" will not be used once the system is set up at CHMC.

3.2 General Organization of "TWIST" Program and Its Capabilities

As was previously mentioned, the program TWIST is the experimenal control program. Figure 3.3 shows the general flow chart for this program. The program is written in the FORTRAN language with a few subroutines written in

Figure 3.1
General view of the computer and the peripheral equipment

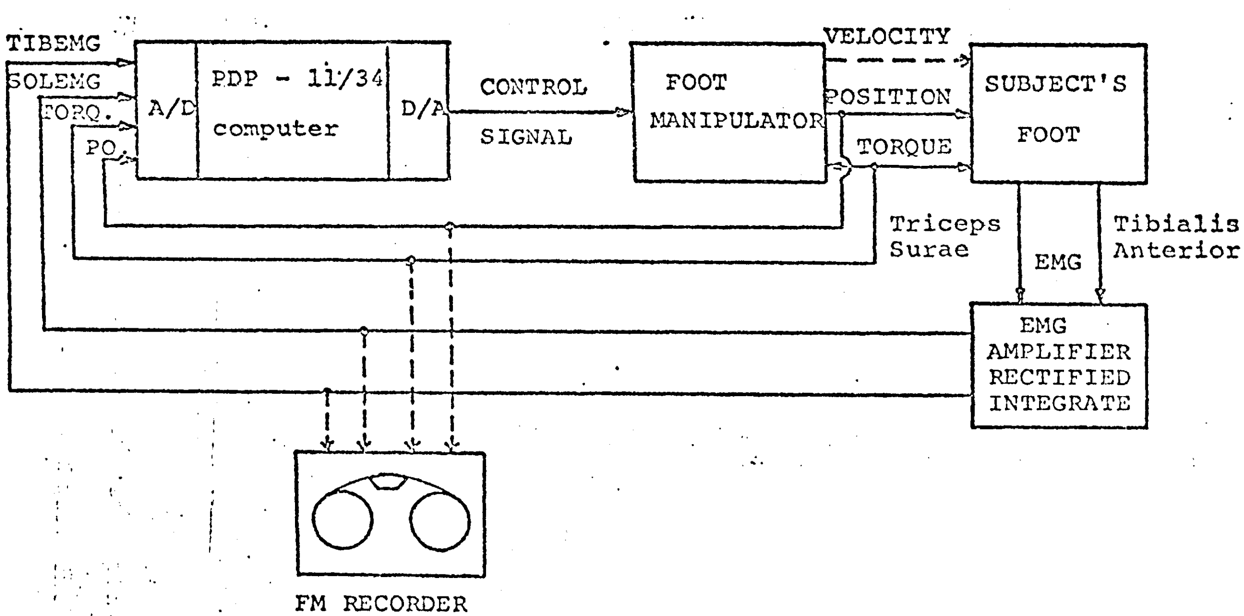


Figure 3.2: Block diagram showing, schematically, the interactions among the various components of the experimental apparatus.

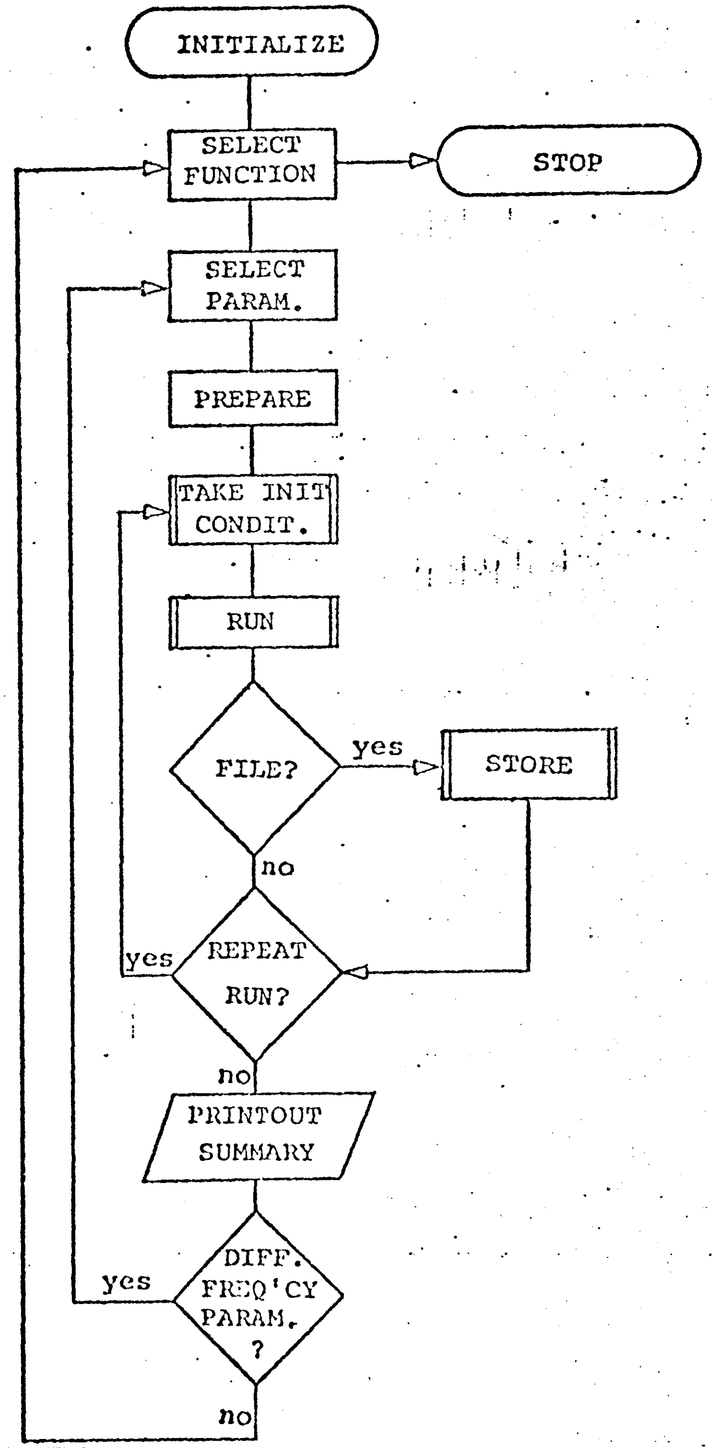


Figure 3.3: Flow Chart for Program "TWIST"

the ASSEMBLER language (to handle the Lab Peripheral System). The program is listed in Appendix B. Since the parameters of interest and the appropriate disturbances to be introduced at the ankle were unknown, the program has been designed to be flexible and easy to modify. It consists of several subroutines which are the building blocks and can be easily modified, changed, or discarded. The program supplies the analog command signal to the pedal and obtains the data from the transducers. It is not responsible for the dynamic feedback control of the servo system. The dynamic reference position signal and simultaneous sampling of the data is done via the "RUN" subroutine.

The main program is called by typing "R TWIST" and the system is initialized (pedal at 0°). After the heading is displayed and the patient's initials and age are typed in, the lower and upper foot motion limits are set via the "LIMITS" subroutine. One of the three forcing functions can then be selected: RAMP, SINE and TRIANGULAR - all performed in the single cycle mode. After selecting the fucntion, the desired parameters are typed in: starting position, maximum travel, velocity, sampling frequency, etc. Once preliminary computations are performed and the pedal is moved to the starting position, then the initial conditions at rest (i.e. torque and position) are sampled and stored for later subtraction from the actual data. After the run is completed, the torque data is digitally smoothed and, if desired, the four data channels are stored on the disk. At the end of the run, a summary of the inputs is displayed or printed out for future reference (see Figure 3.4). The procedure is repeated if any parameters are to be changed. Otherwise, the particular function is terminated and a different function may be selected.

Figure 3.5 shows the typical functions that can be selected by the TWIST program. All test movements (except ramps) are done in complete single cycles. The foot is moved from the initial position (bias) through the maximum position (plantarflexion or dorsiflexion) and back to the initial position. Each run is usually repeated four times, either in sequential order or in a random fashion.

The subroutines responsible for these functions are listed in Appendix C. Subroutine RAMP is used to drive the pedal at a constant velocity. The conversion factor for velocity is 0.04625 volts/degree. Data acquisition starts 50 milliseconds before the beginning of the ramp and ends at a preselected number of milliseconds after the end of the ramp. This is done to provide ample time for establishing the initial starting conditions and determining the nature of the response at the end of the ramp. The total number of sampled points would thus vary with the ramp velocity and can be no greater than 512. Thus, the minimum velocity is range dependent. For instance, this would limit the slow ramp to a minimum of about 5.6° per second for a range of 10°. The above feature is optional and could be changed at any time to accommodate the specific requirements set by the investigator. Following completion of the run, the pedal returns slowly to the initial starting position via "DAMPER". Maximum velocity of theramp is limited to the slewing rate of the mechanism, about 400°/sec.

The sinusoidal forcing function is used to introduce a single cycle sinusoidal disturbance of the form:

$$P = B \pm A((\sin\omega t - \pi/2) + 1) \qquad (3.1)$$

```
PATIENT: DB   AGE:24

SUMMARY OF THE INPUT PARAMETERS FOR TEST#  1 RUN# 1

(RAMP  FUNCTION)

VELOCITY :                          -400.00  DEGREES/SEC.
STARTING POINT IS AT :               -10.00  DEGREES
MAXIMUM TRAVEL IS AT :                -5.00  DEGREES
SAMPLING DATA EVERY  :                    4  MILISECONDS

SAMPLING STARTS 50 MSEC. BEFORE RAMP STARTS.



PATIENT: DB   AGE:24

SUMMARY OF THE INPUT PARAMETERS FOR TEST# 10 RUN# 1

(SINUSOIDAL  FUNCTION)

FREQUENCY :                            2.00  CYCLES/SEC.
STARTING POINT IS AT :                 0.00  DEGREES
MAXIMUM TRAVEL IS AT :                10.00  DEGREES
AMPLITUDE :                           10.00  DEGREES
SAMPLING DATA EVERY  :                    4  MILISECONDS
```

Figure 3.4: A Sample Computer Print-out of Summary of Input Parameters.

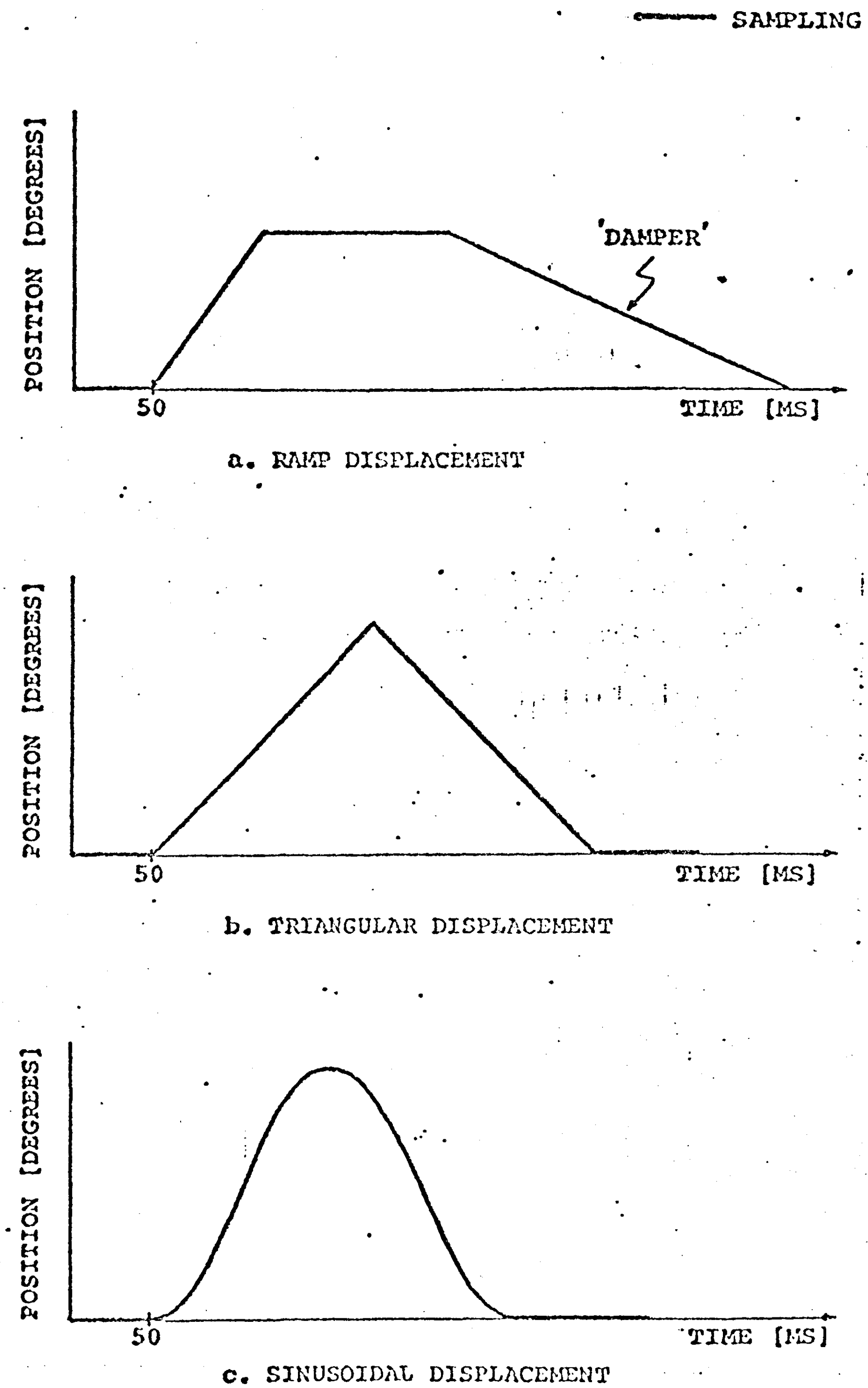


Figure 3.5: Schematic Representation of the Three Typical Displacements Produced by Program "Twist".

where

P is the position
B is the starting position
A is the amplitude
ω is the angular velocity in rad/sec
t is the time in seconds and depends on the sampling frequency
(+) is chosen for the dorsiflexion mode and (-) for the plantarflexion mode, relative to the bias.

The sampling starts 50 msec before the sine function to establish the initial conditions of the run, and ends at a preselected number of milliseconds after the end of the disturbance to measure muscle relaxation events. The maximum number of samples is limited to 512. The maximum frequency should not exceed 6 Hz due to the limitations of the servo-mechanism. Typically the system was run up to 2.0 Hz with satisfactory results. (As will be discussed later, the stretch reflex reaches a constant level at frequencies between 1.0 Hz and 1.8 Hz.)

The triangular function is a double ramp one. It too begins sampling 50 msec prior to the ramp and ends at a preselected number of milliseconds after the end of the ramp. This implies that for a given range, the minimum triangular function velocity will be twice that of a single ramp function velocity.

Data obtained from the run may be stored in an unformatted form in a file via the "FILE" subroutine. Usually the name of the file would be composed of some letter and numeric code combination (maximum of six characters), followed by a dot and the subjects initials.

The "DAMPER" subroutine is a safety feature to avoid abrupt movements of the pedal when the bias position is changed. It provides a slow motion of one degree per second from the present position of the pedal to the bias position. Note: since the analog to digital conversion is presently set for $\pm$ 1.0 volt maximum, the conversion factor for the position is 0.0235 volt/degree. This gives a $\pm$ 0.705 volt maximum for $\pm$ 30°. Also, one must use decimal points for numbers unless the units are msec or no units are used, indicating integers.

3.3 Data Processing

Part of the data, i.e. the torque, is processed (smoothed) on line before being written in the file. Otherwise, the data processing is done off-line, after the runs are completed. Since the data is stored on the disk, it is readily available for analysis. The techniques used for processing and the programs to implement them are described herein.

3.3.1 Smoothing

The raw data is usually contaminated with components which do not contribute to a better understanding of the record. This "noise" would presumably diminish when averaging the records, but since single records are also of interest, it is desirable that the signal to noise ratio be improved. This is particularly important were differentiation of the data is necessary. Analog filtering of the data could have reduced the noise, however, it would have introduced phase shifts, hence timing information could have been lost. To avoid this, the data is smoothed off line with a symmetrical low pass filter of the form:

$$x'_n = \frac{1}{4} x_{n-1} + \frac{1}{2} x_n + \frac{1}{4} x_{n+1} \qquad (3.2)$$

x_n is the nth sample of the record and x' is the new nth point after smoothing. This algorithm approximates a 12 dB octave zero phase shift low pass filter, as shown by Wilcock and Kirsner [1969] with a corner frequency of

$$f_0 = 0.183/T \qquad (3.3)$$

where T is the sampling time interval. Since the sampling time interval is usually 4 msec, the corresponding corner frequency is $f_0 = 45.7$ Hz. To eliminate all high frequency noise, including the 60 Hz power line frequency from the position and torque record, a cut-off frequency of 16 Hz was chosen. Implementation was by using multiple passes (total of 10) through the filter of algorithm (3.2).

Note: One should be aware of the fact that changing the sampling freqeuncy will change the filtering characteristics and, if changed, it should be appropriately compensated for.

3.3.2 Velocity Derivation

Although the velocity can be measured directly form the tachometer, this is not done since the tachometer is connected directly to the motor and does not clearly represent the velocity of the pedal itself. It would also require another analog to digital channel and more space allocated on file. Instead, the velocity, if needed, can be obtained via the "PLOT" program, and is derived by differentiation of the position record. A simple trapezoidal differentiation routine is used. The velocity was of secondary importance and only used occasionally to verify the speeds at which the pedal was supposed to move. It was also used for the computation of the relative equivalent damping, which will be discussed elsewhere in this work.

3.3.3 Averaging

As mentioned before, the data recorded during the run was stored in a file on a disk. Since the system under study is a physiological one, a considerable amount of variation between individual records, given the same stimulus, is expected. When analyzing data, this variation must be taken into consideration and conclusions cannot be made on the basis of only one record. For meaningful interpretation of the data, this variation can be reduced by averaging the individual records assuming that the variation arises from "random noise". If the system is time varying, then coherent averagingis invalid and should not be used. Preliminary testing of normals and one spastic patient showed the system to be stationary in time. Thus, although an averaging program was written, averaging was not done automatically on all records.

The records were first scrutinized visually and only if slight variability over long spans of time were seen were the records averaged. Otherwise, the records were analyzed on an individual basis. Since the investigation is still at a very premature stage, all records, including those which were averaged, have been kept intact. In the future, individual records can be discarded and analysis done on those which were averaged.

The program "AVERAG" is used for coherent averaging. The averaging algorithm is of the form:

$$\overline{X}_N = ((N-1)\overline{X}_{N-1} + X_N)/N \tag{3.4}$$

where

$\overline{X}_N$ is the "new" average

$\overline{X}_N$ is the "old" average

X_N is the data point to be averaged with the previous points

N is the numer of data points averaged

This algorithm is simple, efficient and does not require large storage space. Since the data is in files on disk, the averaging is done first on two files and then a file at a time is added and averaged with the previous ones.

The successive addition of data points to the average only improves the signal to noise ratio. As the number of samples (N) increases, the signal to noise ratio improves by a factor of the square root of N. In the preliminary runs, due to the limited time (one hour) and the large number of different experimental runs, only four records of each disturbance function were taken in random order. Later, when more specific disturbance functions are established, the number of records per ensemble could be increased and with it an increase in signal to noise ratio. In any case, all records should be screened for anomalous responses.

The listing of the averaging program is in Appendix B. Note that the standard deviations are not computed. This option can be added if desired. At present, the averaging routine is used sparingly and only for displaying the data graphically. Usually "graphical averaging" is used simply by superimposing successive traces of the run data on the x-y plotter. The superimposed trace gives a good impression of the repeatability of the response, although the observer tends to look at the envelope of the superimposed traces rather than their average. Note: the averaged data is written onto a file whose name is entered via the teletype. Commonly, the letters AV precede the four alphanumeric characters allotted for labelling the run, followed by a dot and the subject's initials (no more than three).

3.4 Plotting and Displaying the Data

Since the MVL is not supplied with a hard-copy device, the data had to be plotted by means of an x-y plotter for which the program "PLOT" was written. This option is solely for the purpose of the preliminary tests done at the MVL. The generation of the plots is slow and time consuming. It will not be used in the hospital. At the REC, there is a hard copy device over which the data will be displayed and copied via the Tectronix 4014 terminal.

Prior to plotting, the data is previewed on the CRT via the program "SEE". This program displays position versus time, torque versus time, and torque versus position. This routine is also written for temporary use in the MVL for convenient quick review of the data.

The program "PLOT" is comprised of several subroutines which are listed in the "MVL Lab Report No. 771" [1977]. The program is written in FORTRAN and is listed in Appendix C. A few options are available in this program. The data may be printed out if desired. Position versus time, torque versus time, tibialis anterior EMG versus time, soleus EMG versus time, and torque versus position, and speed versus time may be plotted within a 6" by 9 1/2" rectangle. Axes and tick marks, scales and origin location are all optional but very handy. Initially the plots were done using different color felt tip pens to get well contrasted superimposed plots, but later, for the purpose of xerox copy clarity, the plots were separated. Typical plots generated by the x-y plotter are shown in Figures 4.5 and 4.11.

3.5 Calibration

In order to obtain reproduceable results, it is extremely important to have the system checked and calibrated regularly. This applies primarily to the torque transducer which is battery operated and its output which is directly proportional to the battery drainage. The battery of the torque transducer is rechargeable and should always be fully charged prior to test runs. To check it, the program called "CALIBR" may be used. By loading the pedal with either weights or a force scale at a known moment arm about the pedal's axis of rotation, and sampling the loaded system with "CALIBR", the actual moments can be compared to the measured ones. The angle of the pedal can also be measured by setting the pedal at different angles, measuring the angles with a protractor and comparing them to the ones read off the computer. Usually the ± 15 volt supply to the pedal is very constant and it is not likely that the readout would change. Keeping the strain gauge battery charged will also maintain the proper calibration.

The listing of the calibration program is given in Appendix B.

CHAPTER IV

NORMAL SUBJECTS EXPERIMENTS

4.1 Introduction

Before installing the prototype foot manipulator in the clinic, preliminary evaluation of the equipment was necessary. The major points of interest were:

(a) Performance of the loaded manipulator
(b) Possible problems with man-machine interface, i.e. acceptability of the test procedure by the subject, irritability, discomfort, duration of the test, alignment, etc.
(c) Are there any meaningful physiological results, and how do they compare with other investigations in this area?
(d) Reproducibility of the inputs and the results.

To resolve the above issues, a pool of five subjects, three males and two females, between the ages of 24 and 41, without a known history of neurological or muscular disorders were tested. All are assumed to be "normal" subjects and could be used for comparison with patients. Since the parameters of interest were not known, a large variety of stimuli were introduced. Parallels were drawn between the individual subject's data and it was also compared to the results reported by others. Some of the physiological aspects of interest were: short and long latencies of EMG, the response of relaxed and precontracted muscle, effects of velocity, initial position and range of motion. Relationships between measured parameters (position, torque, EMG) were also studied.

4.2 Experimental Protocol

The experimental set and subject positioning were done as described in Section 2.6.1. The subject was asked to sit comfortably and completely relax. In some cases, earphones with soft music or just earmuffs were provided to minimize auditory cues to the subject. The subject was usually asked to close his eyes and the lights were turned off, although, some of the experiments were done without the earphones and with the lights on. While sitting in the chair, subjects were unable to see their foot. After a short relaxation period of about five minutes, the experimental runs started. There were different instructions and different modes of stimulus function combinations which will be discussed later on in this chapter. Generally, there were four types of instructions.

(a) "Relax as much as you can during the entire experiment. Do not try to resist the movement nor enhance it voluntarily. Just sit back and completely relax."

(b) "Relax as much as you can. As soon as you feel that your foot is moving, try to resist the movement slightly. If the foot is going up, press down slightly on the pedal. If the foot is going down, pull slightly up on the pedal. Relax again when the motion stops."

(c) "Look at the meter in front of you and by pressing lightly down on the pedal, set the indicator needle on the second division to the left of the zero. Maintain a steady force on the pedal and keep the needle in place. As soon as you feel that your foot is moving, push slightly down on the pedal as quickly as you can."

(d) The same as (c), except that the instruction is to pull the foot away from the pedal and bring the meter indication back to zero.

The operation of the system was demonstrated to the naive subjects to familiarize them with it, and to assure them that the system "does not bite" and to make them feel at ease.

Instructions (a) and (b) were used when relaxed muscle studies were performed. Instructions (c) and (d) were used when precontracted muscle responses were under investigation. These instructions were given to help in differentiating between the reflex and the voluntary components of the stretch reflex.

The stimuli consisted primarily of four different modes:

(a) Fast ramps (200 and 400°/s)
(b) Slow ramps (12 to 160°/s)
(c) Sinusoidal (0.6 to 2.0 Hz)
(d) Triangular (12 to 40°/s)

Usually the modes were introduced one at a time at different velocities, in a pseudo-random order (at the same position and range). They were unicyclic, and were introduced with latencies of 30 seconds to 2 minutes between runs, to avoid prediction by the subject.

More details about the runs are given in the following subsections.

4.3 Experimental Results

4.3.1 Introduction

The preliminary tests revealed that the system was highly acceptable by the subjects. There were relatively few incidents of irritability or discomfort, which in all cases were rectified immediately. Runs were trimmed to finish an experiment within 45 minutes to avoid boring the subjects. Performance of the equipment was satisfactory and reproducibility of the disturbances was found to be extremely good, as can be seen from the figures in the following sections. The results were found to be reproducible within the limits of physiological system variations.

The MES records were found to be consistent in their timing, although not always in amplitude, which was expected to to the physiological aspects of muscle contraction.

Careful analysis should be made of the torque parameters. The torque represents the resistance developed by the foot when the foot is being moved. One should be aware of the fact that this resistance is due to four different factors, as mentioned in Section 2.4.2:

(1) The inertial component due to the mass of the foot and the pedal.
(2) Rheological components (visco-elastic stiffness) due to the physical properties of connective tissue, muscle fibers, ligaments, joints, etc.
(3) Reflex components due to the reflex contraction of the passively stretched muscle.
(4) Voluntary components due to conscious activation of the muscle.

The inertial component is directly proportional to the acceleration and is most pronounced at the beginning and end of the ramp type displacements. However, it is absent during the period when the velocity is constant. To a lesser extent, it is a function of theposition of the foot in the gravity field and could be considered negligible for small changes in position (see Equation 4.3).

The rheological components are predominantly dependent on the velocity of stretch and the change in length of the muscle. In addition, these "visco-elastic" components vary with the level of quiescent muscle activity. In general, the above components are muscle "passive mechanical components" and no MES is associated with them. In contrast to these, the reflex components and the voluntary components are associated with MES. Commonly, it is reported in the literature (allum, 1973; Kearney, 1976; Kwee, 1971; Adam, 1976 and others) that there are two bursts of MES associated with a rapid stretch of a muscle: a short latency burst, attributed to the monosynaptic stretch reflex (MSR) and a long latency one known as the functional stretch reflex (FSR). Thus when analyzing the torque records, it is important to consider all the above factors.

To assist in understanding the torque record, some preliminary testing was done to see how the "passive" components influenced the data. For that purpose, the passive components of the foot were simulated by using a two pound weight placed on the pedal about four inches from the axis of rotationa and a spring pulling down on the pedal. The system was run in ramp mode at 200°/s. The result is presented in Figure 4.1. Note the influence of inertia on the start and end of the movement. Steady state is reached when the pedal is at 10° and the torque level is constant due to the stretched spring. Similar components are expected to be seen in the runs' records, and they are actually present in the records as can be seen in the following figures.

4.3.2 Fast Ramp Experiments

A series of fast ramp displacements, typically with velocities of 200°/s and 400°/s, were performed on three subjects, two females and one male. Initially all disturbances were 0° to 10° dorsiflexion. As previously mentioned, different instructions were given to the subjects. The first series of experiments were performed using instruction (a) from Section 4.2. A typical response of each of the three subjects is shown in Figure 4.2. The first two subjects from the top are females. The "position" trace represents the angular position of the ankle with positive being dorsiflexion, and ramp velocity of 200°/s. "Torque" represents the resistance developed by the foot due to the disturbance. The third trace is the EMG of the soleus muscle and the fourth is the EMG of the tibialis anterior muscle. Note that torque is positive in plantarflexion, meaning increased force exerted on the pedal downward gives higher positive torque. In other words, contraction of the soleus will produce an increase in torque, while contraction of the tibialis anterior will produce a decrease in torqe. This is valid for all the records presented herein.

The marker on each record represents the point at which the pedal started moving (in this particular run, sampling started 60 ms before the ramp). As was expected, the soleus EMG displays a burst of activity within the 40 to 60 ms from the initiation of movement. A latency of 44 ms is seen in the top trace. Each burst lasts for approximately 35 ms. The second record shows the same pattern, with the timing being 45 ms nad 33 ms respectively. The third record shows a latency of 55 ms with a duration of 40 ms. This is consistent with what is

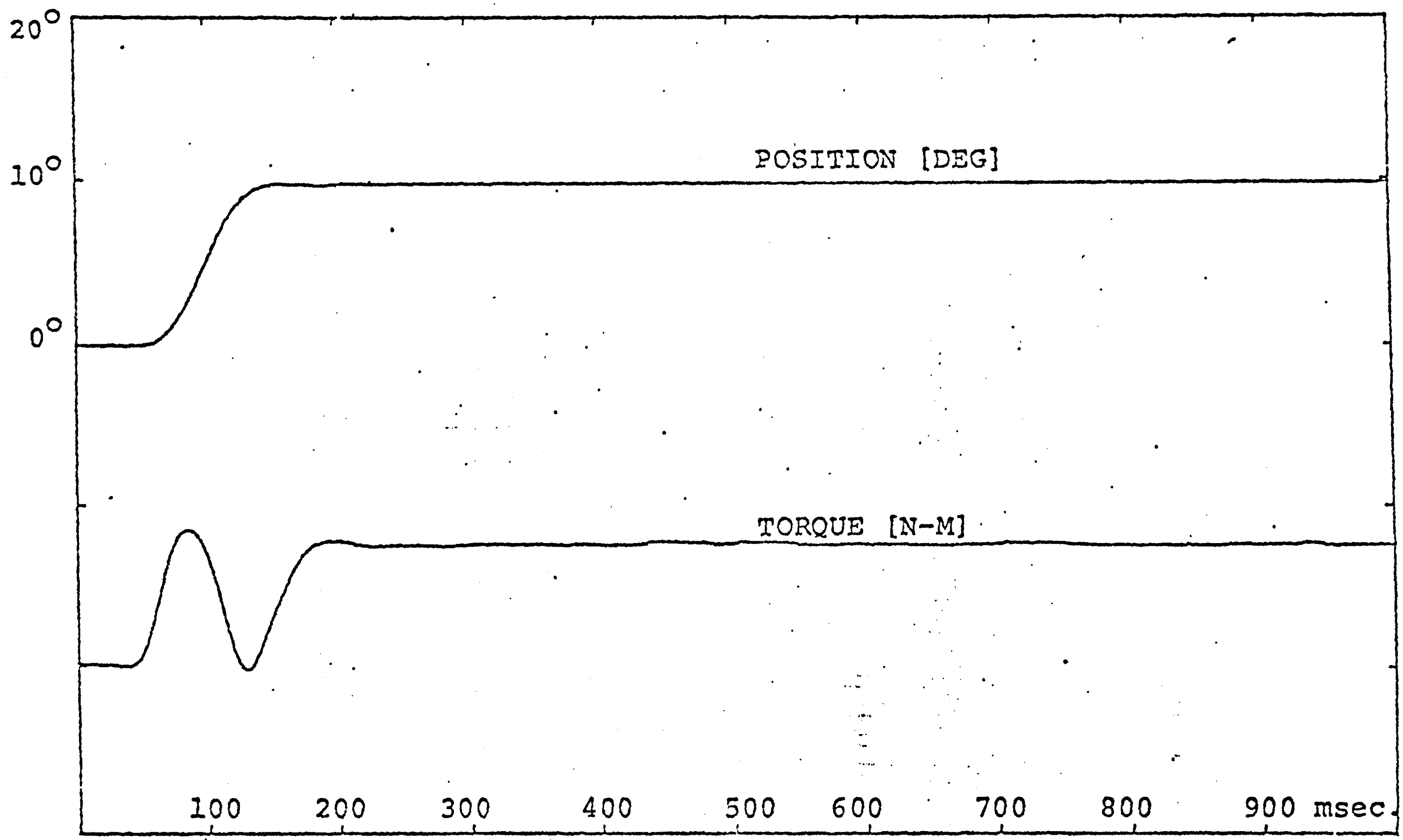


Figure 4.1: Position and Torque Records Showing the Effect of a 200°/sec Ramp on Torque Response for the Foot Pedal Loaded with a 2 lb Weight Placed 4 in. From Axis of Rotation, and a Spring, Simulating Two of the Passive Mechanical Components of the Foot, i.e. Elastic Stiffness and Inertia.

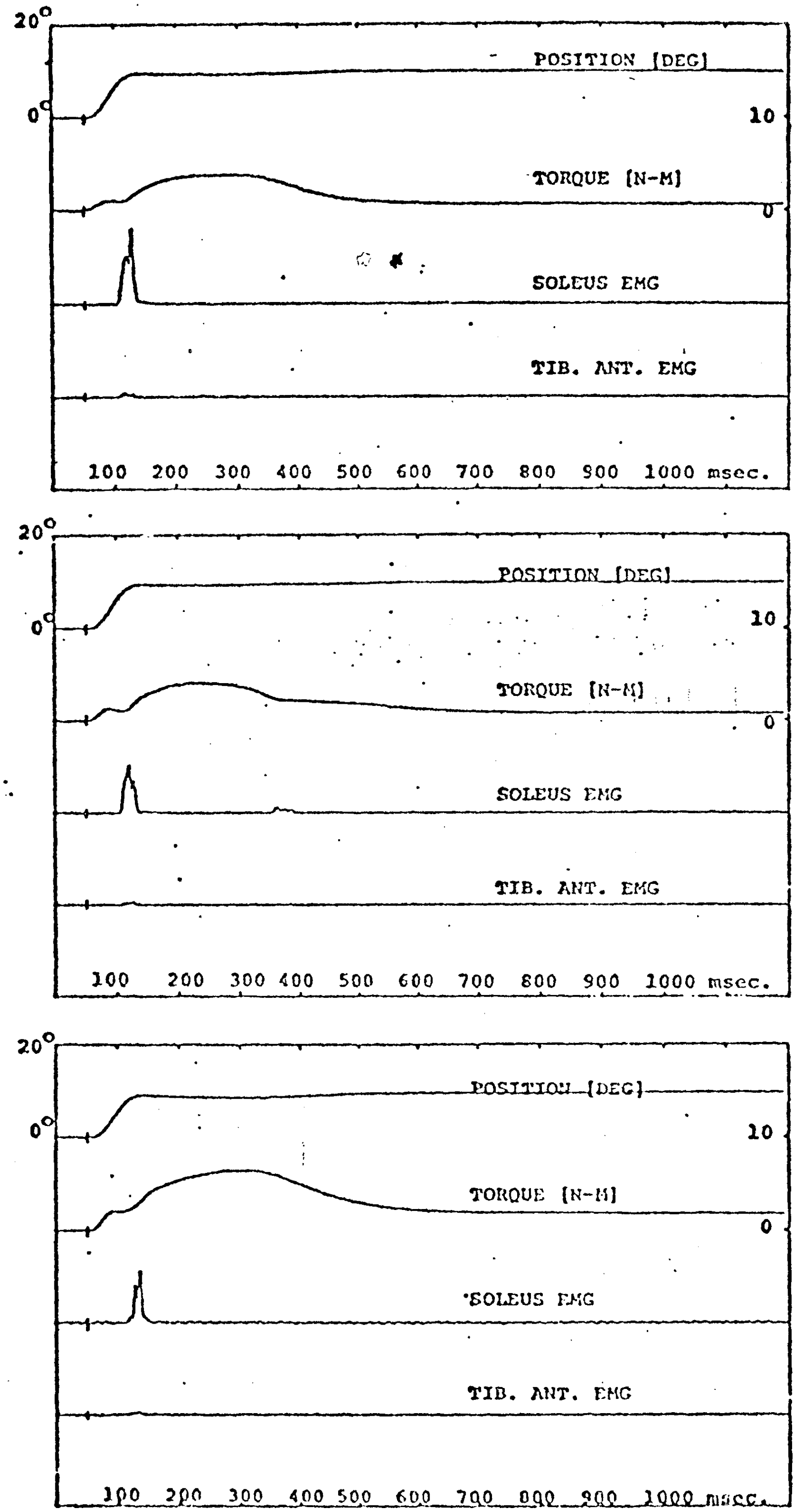


Figure 4.2: Response of Three Subjects to Fast Ramp (200°/sec) Displacement (0° to 10°). The Instruction Given to The Subjects Was to Relax Throughout the Experiment.

reported in the literature (Hammond, 1960; Melvill Jones and Watt, 1971; Marsden and Merton, 1976; Agarwal, 1970). The first burst of EMG is reported to occur with a latency of 40 to 60 ms (for the ankle) and has been referred to as the monosynaptic stretch reflex.

As can be seen from the records, the two females have very similar latency timing, while the male's latency is about 10 ms longer. A possible explanation could be on the basis of nerve conduction time which is proportional to the distance of conduction. Both females are of the same height (5'2"), while the male is 6'2" with longer legs. The second soleus EMG burst seen in the second record is most likely a voluntary contraction, as will be shown later in this section. The "ripple" in the third record of soleus EMG is 60 cycle "noise". Tibialis anterior EMG reveals a low level response timed with the soleus EMG, which is seen in all the records. This might imply that there is a co-contraction of both muscles (?).

The torque record is interesting, particularly in the context of measurement of spasticity. Torque response is extremely consistent and its first part, the small "hump" is constant for a given subject, velocity and range. It is most likely due to the passive components of muscle as noted in Section 4.3.1. Important points to consider are the increase in torque coupled with the first burst of the soleus EMG and the relaxation phase thereafter.

Increase in torque results due to the MSR (?) when the triceps surae muscles are rapidly stretched and the relaxation corresponds with similar results produced by an Achilles tendon tap (Agarwal, 1970).

To further investigate the capability of the apparatus in revealing physiological aspects of the stretch reflex, different sets of instructions were given to the subject. A second series of experiments were performed using instruction (b) from Section 4.2. Figure 4.3 shows three frames representing three consecutive runs of subject DB. The disturbance starts at the mark on the trace with a ramp velocity of 400°/s. In these runs, the second burst of soleus EMG can be seen with a latency of about 265 ms and varies slightly from run to run, while the first burst always appears at exactly 46 ms. This record compares with the top record in Figure 4.2 which was taken from the same subject two weeks earlier. Long latency voluntary contraction seen in the record is also consistent with previous investigations (Gottlieb, 1975). The torque record shows an increase in torque time coupled with the EMG burst revealing that the subject followed the instruction given to her to "push on the pedal slightly as quickly as possible".

A third sequence of experiments were performed using instruction (c). In the experiment, the triceps surae are precontracted, maintaining a constant torque and show constant level of EMG activity above the base line. Results of this experiment are illustrated in Figure 4.4. The dashed lines on the torque record represent the expected response. The actual record shows a flat region due to saturation of the analog to digital converter (± 1 volt). Soleus EMG shows the same characteristic as previously seen in the non-prebiased muscle experiments and timing of the first burst remains constant at 46 ms. However, the shape of the MSR EMG is considerably different, and, also, the latency of the second burst is shorter (245 ms). These, again, are physiologically consistent, since the muscle initial condition is an active state with, probably, increased sensitivity of the muscle spindles and some other neural connections already made. Note the

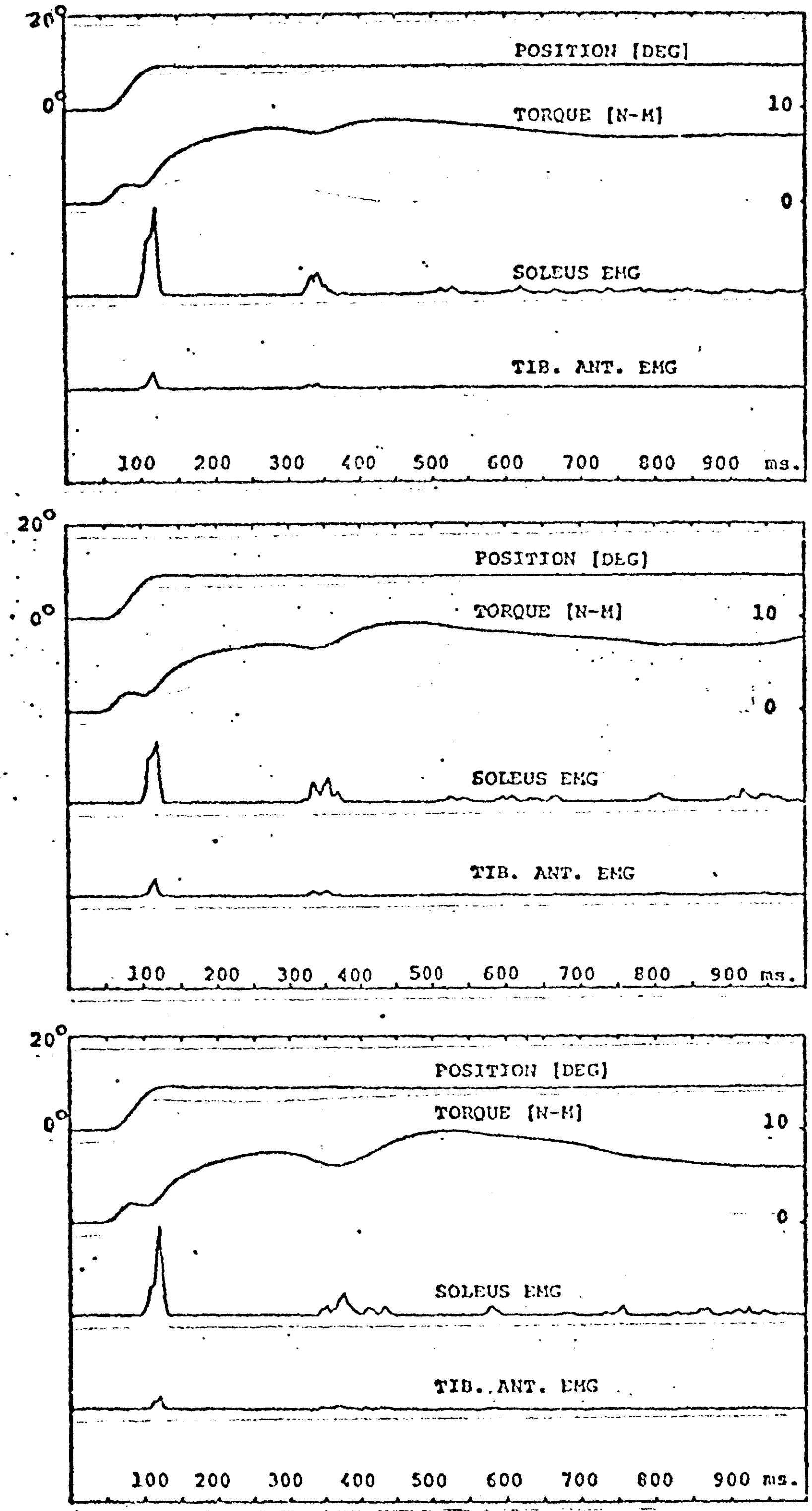


Figure 4.3: Three consecutive Runs of Same Subject. The instruction to the Subject was to Push on the Pedal Slightly as soon as She Feels the Pedal Moving.

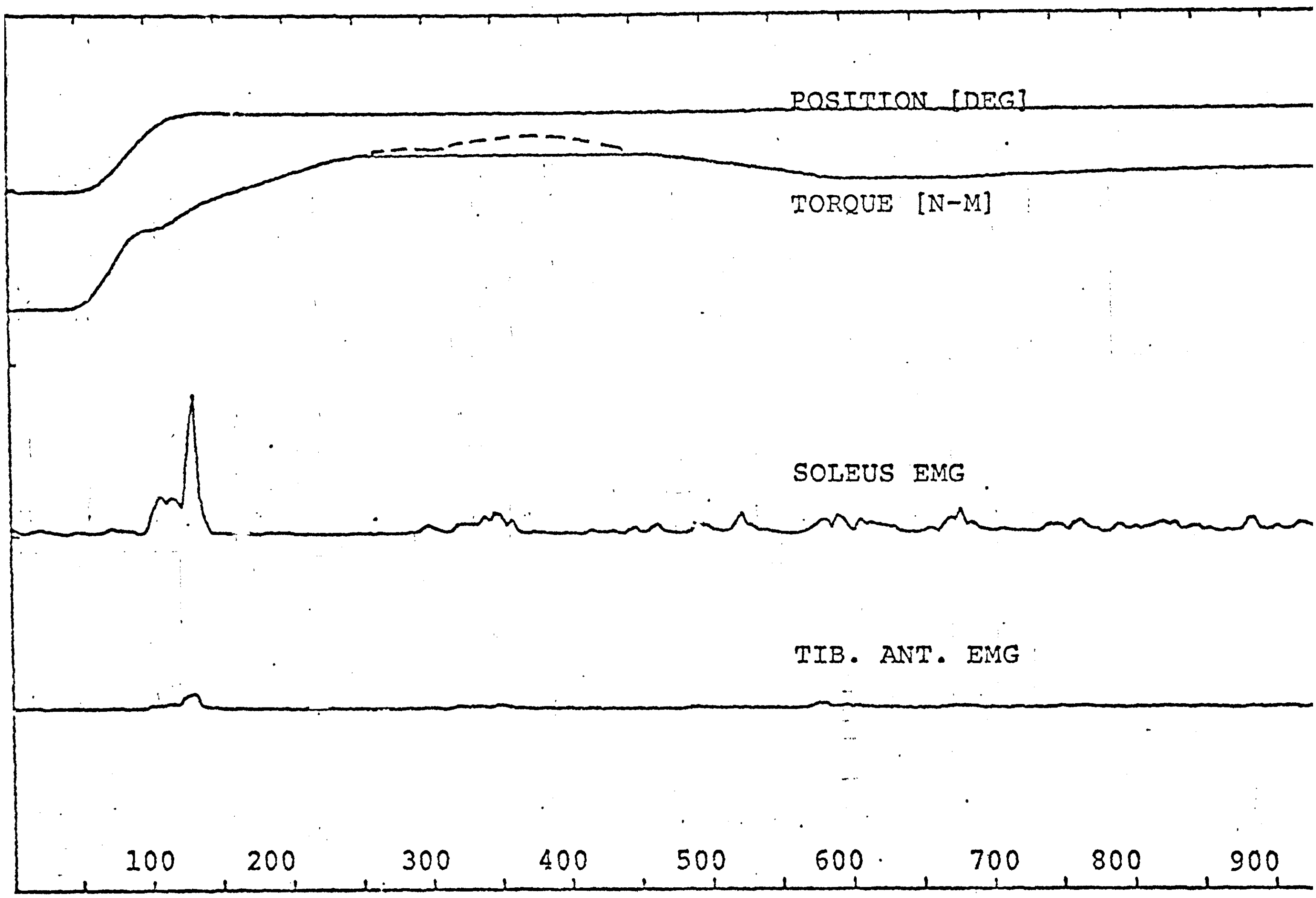


Figure 4.4: Pre-stretch Contracted Muscle Response to a Fast Ramp (200°/sec) Disturbance (0° - 10°) with same Instruction as in Figure 4.3.

amplitudes of the soleus EMG relative to the pre-stretch bias level activity. Also the higher torque, in the records presented so far, is associated with the late burst of EMG, although amplitude-wise, the first burst of EMG is now much higher than the second. The fact that no FSR could be observed within the 100 to 200 ms latency is quite puzzling. This is in contrast to what is reported in the literature, although it is claimed that the FSR could be modified depending on the instructions given to the subjects.

Dr. S. Simon of the Gait Analysis Laboratory at CHMC suggested attempting to detect a dependency of the MSR on the position of the foot and the velocity of stretch. The affect of position was tried initially with the fast ramps and later with the slow ramps. Two 400°/s ramps with the same amplitude of 5°, but with different initial positions were introduced to the foot of a completely relaxed subject. Figure 4.5 presents the data of these runs. The top frame shows the results of the ramp with initial position at -10°. The latency of the soleus EMG first burst is 55 ms. The bottom frame shows the results of the ramp with an initial position at -5°. The latency of the soleus EMG is 46 ms. Note the differences in amplitude of the torque and soleus EMG records. The latency of the soleus EMG in the bottom frame is exactly hte same as seen in previous runs with velocity of 400°/s, and an initial position of 0°. This raises the question why this longer latency in the top frame and why the difference in amplitudes? One hypothesis could be that the muscle is slack at -10° and it takes a few degrees of ankle motion before the muscle and the muscle spindles reach a length at which they are sensitive enough to fire. However, this is questionable, since the muscle spindle is known to adjust its sensitivity within the physiological range of muscle length. Another hypothesis could be that the reflex is dependent not only on the velocity of stretch, but also on the position of the foot.

Thus far the fast ramp experiments show that the apparatus is sensitive enough to obtain meaningful physiological data. The results are highly repeatable and are in accord with the known aspects of the stretch reflex of the ankle as reported by others (Agarwal, 1970; Marsden, 1976; Kearney, 1976).

4.3.3 Slow Ramp Experiments

The ramp velocities ranged from 12°/s to 160°/s. Figure 4.6 shows the data obtained from runs at different speeds from initial position 0° to 10° dorsiflexion. As can be seen, there is a strong dependency of the MSR amplitude on the velocity of stretch. This phenomenon was reported previously by Kearney (1976). Not only is the amplitude dependent on the veloity, but also the latency of the MSR. As the speed of stretching increases, the latency becomes shorter. At 12°/s, no EMG activity is detected, while at 140°/s and 160°/s, it is pronounced and its effect can be seen in the torque record.

This observation is not known to be reported elsewhere. Investigators were applying rapid stretches only, and elicited the "monosynaptic" activity with constant latency. Agarwal (1977) does not use any inputs below 3 Hz and Kearney (1976) does not use any below 112°/s for his investigation about the ankle joint.

Since it was found that the latency of the MSR reaches a constant time of about 46 ms at velocities higher than 150°/s, it has been assumed that this is the minimum latency of the MSR. Subtracting this minimum latency time from the latencies measured for the different slow velocity ramps, and finding the

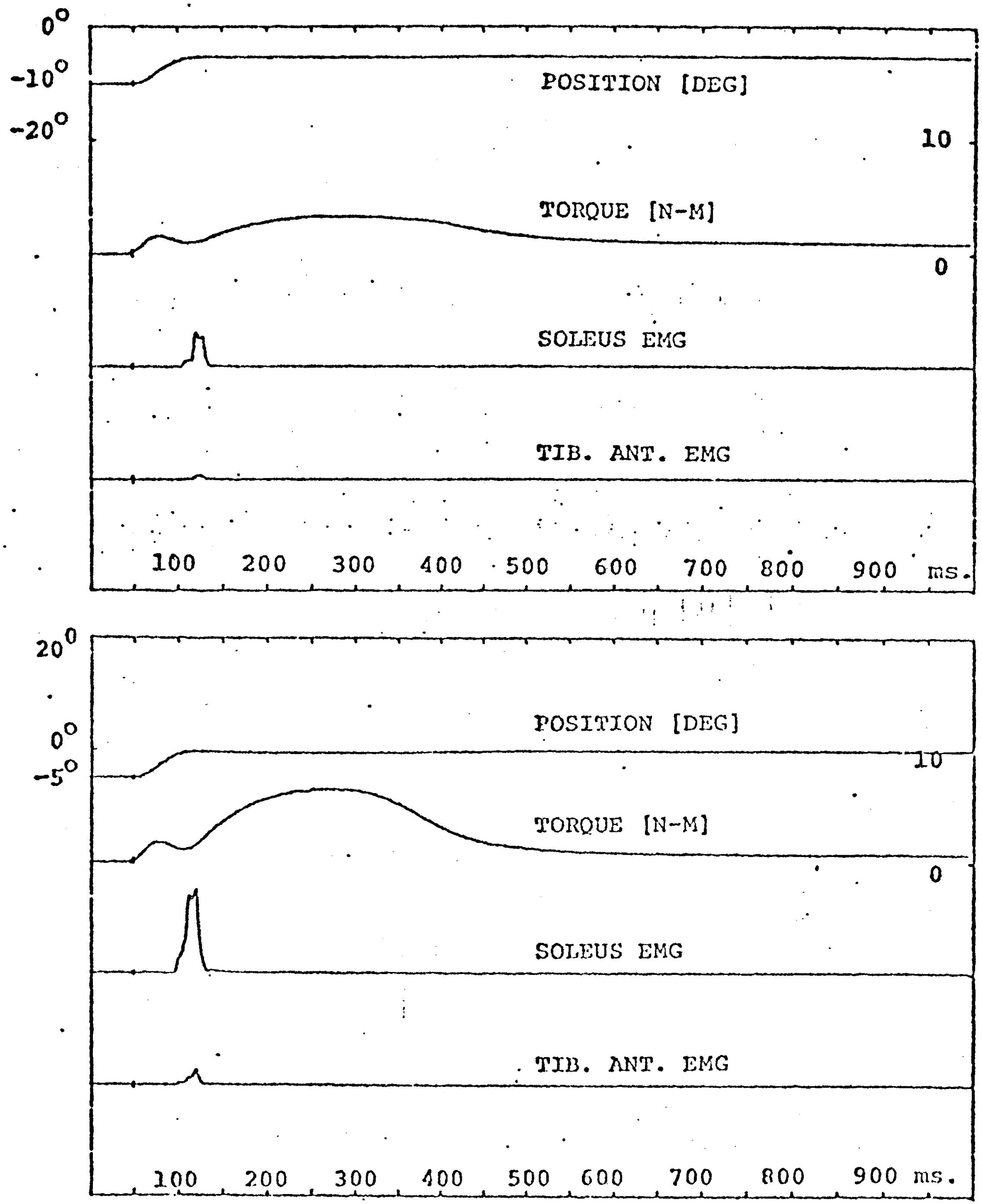


Figure 4.5: Effect of Initial Position on the MSR. 400°/sec. Ramps Were Introduced at Different Initial Position. Top Frame: Disturbance From -10° to 5°. Botton Frame: Disturbance from -5° to 0°. (Compare these with Results in Figure 4.2).

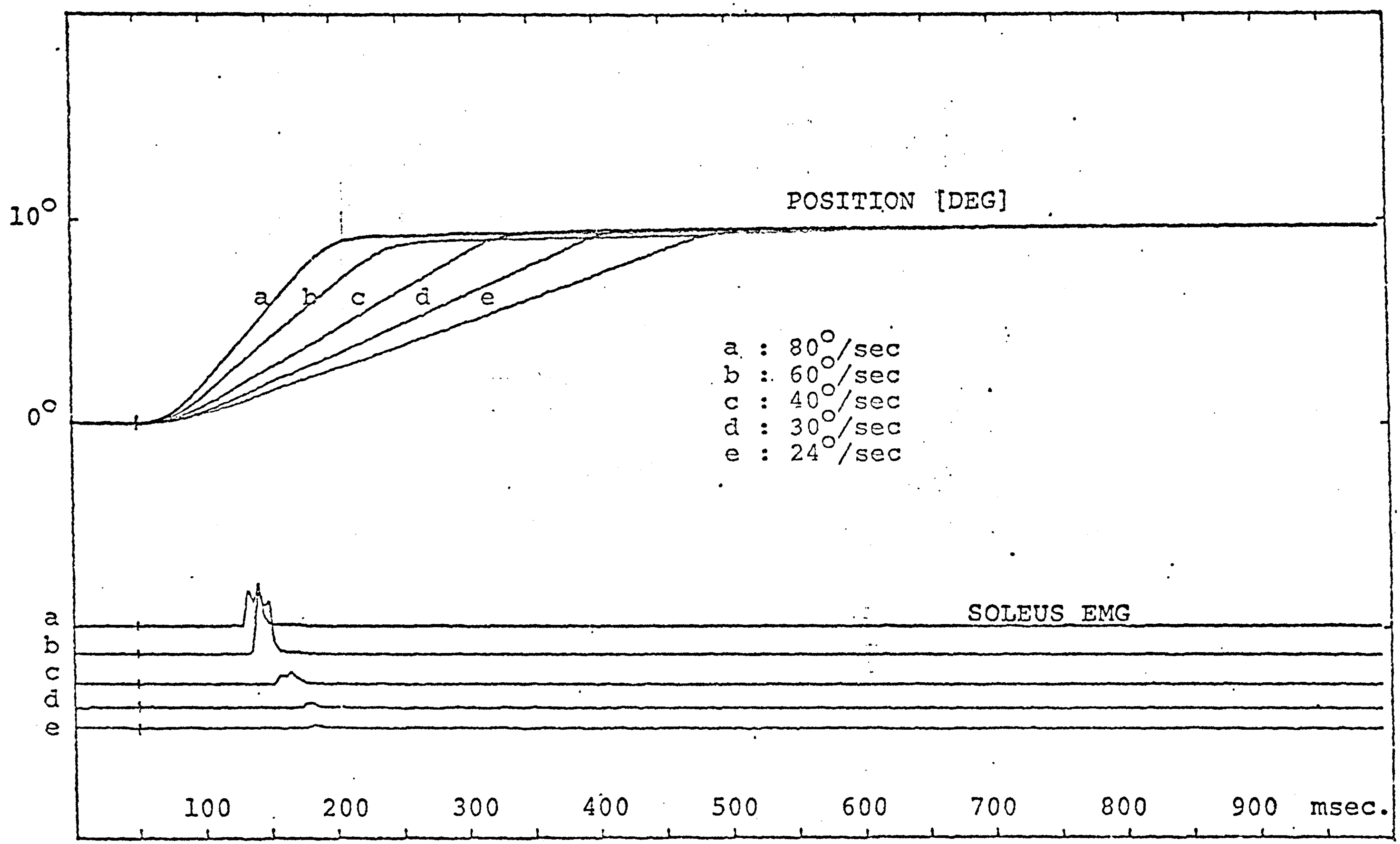


Figure 4.6: Position and Soleus EMG Traces for Different Ramps Illustrating the Dependency of the MSR Amplitude and Latency on the Velocity of Displacement.

corresponding velocity for the new latency, it was found that the MSR is triggered as soon as the foot reaches a threshold velocity of about 28°/s. This implies that the MSR latency is a function of acceleration. To illustrate this, Figure 4.7 shows the velocity versus time corresponding to the different ramps.

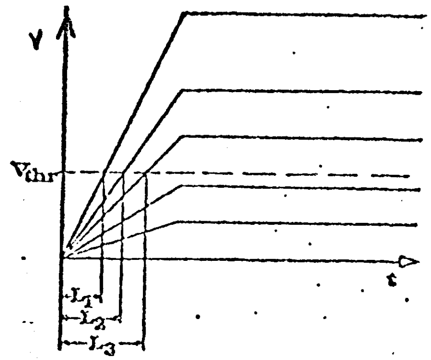


Figure 4.7 Velocity versus time plots for different ramps showing acceleration phases and constant velocity phases, with corresponding latencies.

These latencies can be represented by

$$L_i = (V_{(thr)}/a_i) + L_0 \tag{4.1}$$

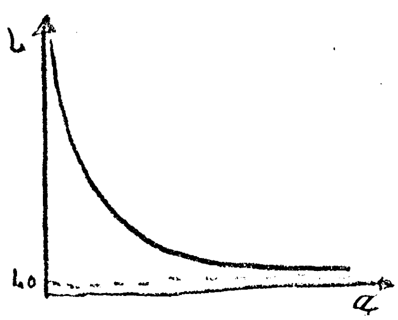
where L_0 = minimum latency of MSR (approximately 46 ms for subject DB)

L_i = latency of MSR due to different slow ramp displacements.

If we let $L = L_i - L_0$, then

$$a_i L = V_{(thr)} \tag{4.2}$$

which represents an hyperbola:



This hypothesis, although speculative, could be used to explain why slow stretch of the muscle does not elicit MSR. Thus, it can be hypothesized that a tendon tap elicits an MSR not by merely stretching the muscle, but rather by introducing a high acceleration impulse.

It is worthwhile to note that the same phenomena was observed from the sinusoidal displacements as shown in Figure 4.8.

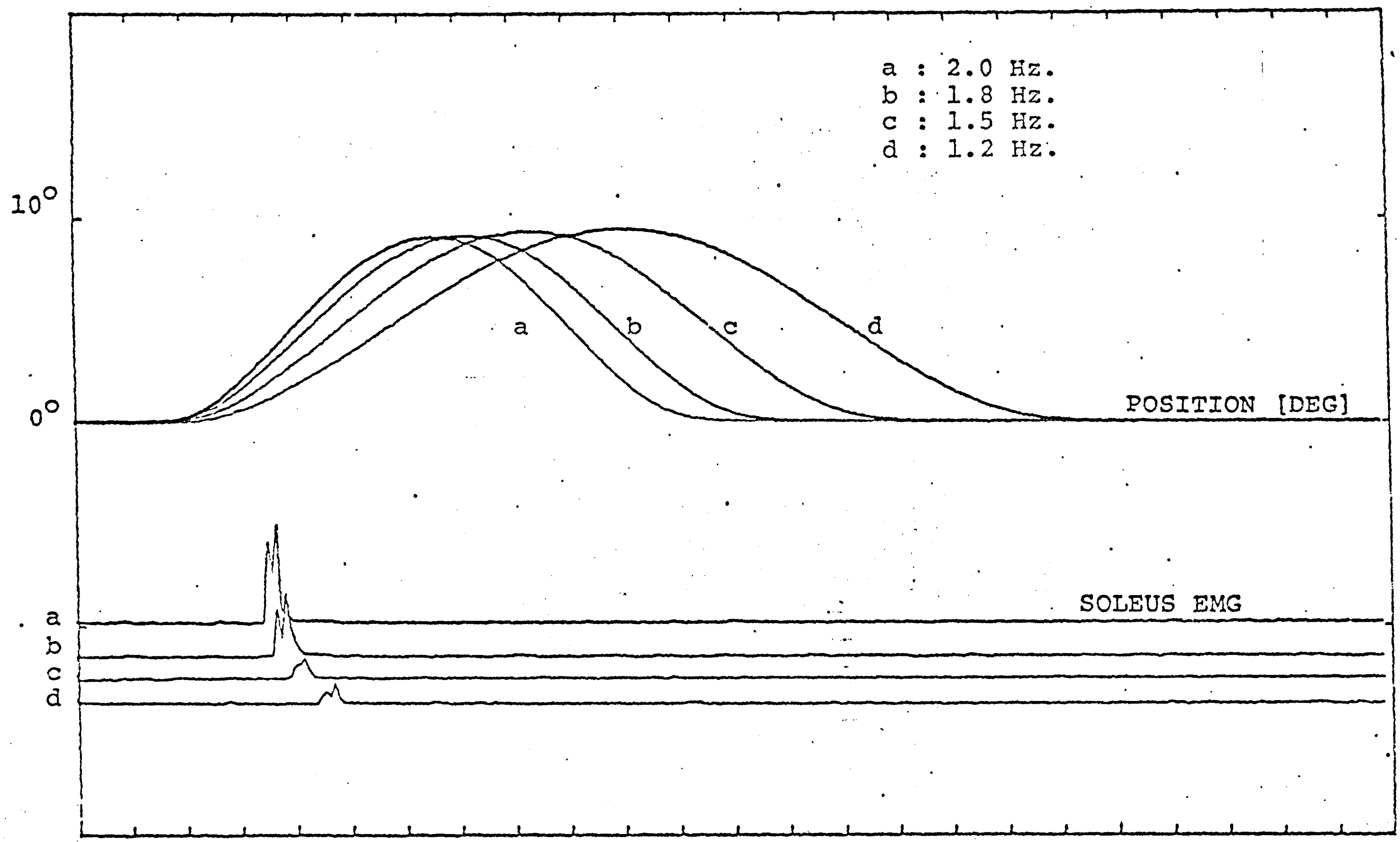


Figure 4.8: Position and Soleus EMG Traces for Different Frequencies, Illustrating the Dependency of the MSR Amplitude and Latency on the Frequency of Displacement.

Since the ankle plays an important role in postural control and there is direct influence from the vestibular system on the stretch reflex, the mechanisms involved are probably different from those corresponding to other joints. Further investigations are needed to explain the above phenomena, but this would be beyond the scope of this thesis.

4.3.4 Sinusoidal Manipulation and RED

Sinusoidal movements of a limb have been most frequently used to determine the energy absorbed per cycle of the imposed movement as a measure of spasticity (Bomze, 1973; Webster, 1964; Long, 1964; and others). It has been reported that spasticity is velocity dependent and some investigators refer to it as "viscous rigidity". Dr. H.H. Kwee, on a visit to our lab, suggested using the apparatus to evaluate his method of measuring the Relative Equivalent Damping (RED) of a joint (Kwee, 1976). The assumption is that the energy required to move the limb passively is a good measure for hypertonia. The linear equation considered for the torque produced at the ankle is:

$$J\ddot{\theta} + b\dot{\theta} + k\theta + mg\ell\sin\theta = T \qquad (4.3)$$

where

- T = torque at the ankle
- θ = ankle position
- $b\dot{\theta}$ = damping torque due to rheological components of muscle and joint and hypertonicity
- $k\theta$ = torque due to elastic rheologic components
- $mg\ell\sin\theta$ = torque caused by gravitation
- J = moment of inertia of foot and pedal

To obtain the equation for the energy balance, all terms in (4.3) are multiplied by the angular velocity and integrated over one cycle of the movement:

$$J\int_0^\tau \ddot{\theta}\dot{\theta}\,dt + b\int_0^\tau \dot{\theta}\,\dot{\theta}\,dt + k\int_0^\tau \theta\,\dot{\theta}\,dt + mg\ell\int_0^\tau \dot{\theta}\sin\theta\,dt = \int_0^\tau T\dot{\theta}\,dt \qquad (4.4)$$

where τ is the period time.

For purely periodical movement, such as a sinusoidal one, the first, third, and fourth terms of equation (4.4) become zero (see Appendix C) and the net energy absorption is caused by the second term, thus:

$$b = \frac{\int_0^\tau T\dot{\theta}\,dt}{\int_0^\tau (\dot{\theta})^2\,dt} \qquad (4.5)$$

The numerator represents the energy absorbed per cycle and the denominator is a characteristic measurement of the imposed movement. "Equivalent damping" is defined as the amount of work done per cycle divided by the characteristic measure for the movement. Since the torque T is a function of maximal muscular force, it will vary from person to person. To normalize the equivalent damping, Kwee (1976) proposed dividing b by the weight of the subject (which represents, roughly, the muscular force capability); thus

$$RED_W = b/W \tag{4.6}$$

where W is the weight of the subject.

This seems to be a very crude approximation and it has been decided to improve it by defining the RED relative to the lower leg volume. This will give a better approximation to the volume of the muscle and, hence, the maximal force capability of the joint under test. Thus, the new definition of the RED would be:

$$RED_{LV} = b/V_L \tag{4.7}$$

and

$$V_L = L\rho^2/4\pi \tag{4.6}$$

where

V_L = lower leg volume (from knee to ankle)
L = distance from knee to ankle
ρ = maximum circumference of the lower leg.

(The computer program which was written to calculate RED_{LV} is listed in Appendix C.)

Test runs of the above method used a series of single cycle sinusoidal movements at different frequencies. These frequencies ranged from 0.6 cps to 2.0 cps with constant amplitude of 10° dorsiflexion and initial position at zero degrees. The subjects were asked to follow instruction (a) of Section 4.2. Usually, four runs at each frequency were performed. Figure 4.9 represents a typical record of the four data channels versus time. Since the parameter of interest is the net energy absorbed, a more useful visualization of the data would be in the form of a torque versus position plot.

The amount of work done on the muscle will be represented by the area enclosed by the torque-position curve. Figure 4.10 shows such a plot resulting from a stretch at 1.8 cps. Note the four distinct components of the trace: a fast increase in tension (a), a moderate slope (b), a second fast increase in tension (c), and a fast decrease in slope (d). Component (a) is most likely due to the inertia of the foot and pedal; component (b) with moderate slope is probably due to the passive properties of muscle, both viscous and elastic. Phase (c) corresponds to the first burst of EMG of the soleus (see Figure 4.9) and indicates an increase in soleus tension. An explanation for this could be that the previously relaxed muscle contracted reflexly. In phase (d), the soleus is being relaxed while the slack tibialis anterior is being stretched slightly, corresponding to the decrease in tension with decrease in angular position.

The average slope of phase (c) might be a good approximation to the increased stiffness produced by the reflex response. In Figure 4.11, the plots of runs at different frequencies were superimposed. The average slope of phase (c) increases with increasing frequency up to 1.8 Hz, beyond which it stays constant.

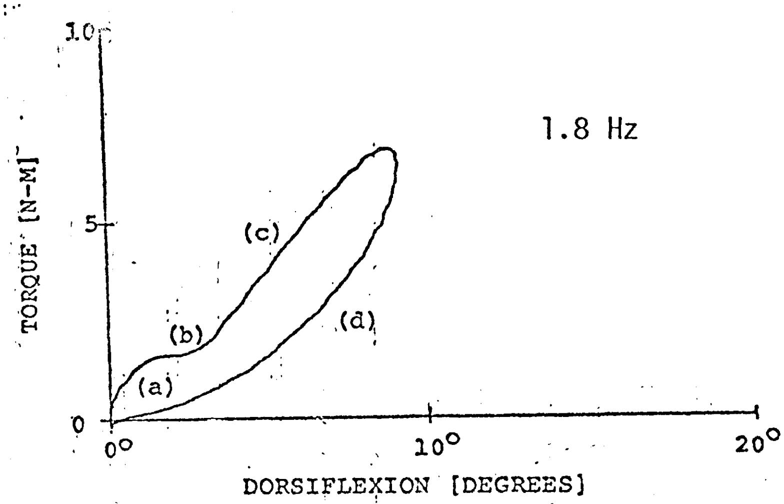


Figure 4.10: Torque versus dorsiflexion plot describing an hysteresis loop. whose area represents the amount of energy absorbed by the muscles. Note the four phases of the loop.

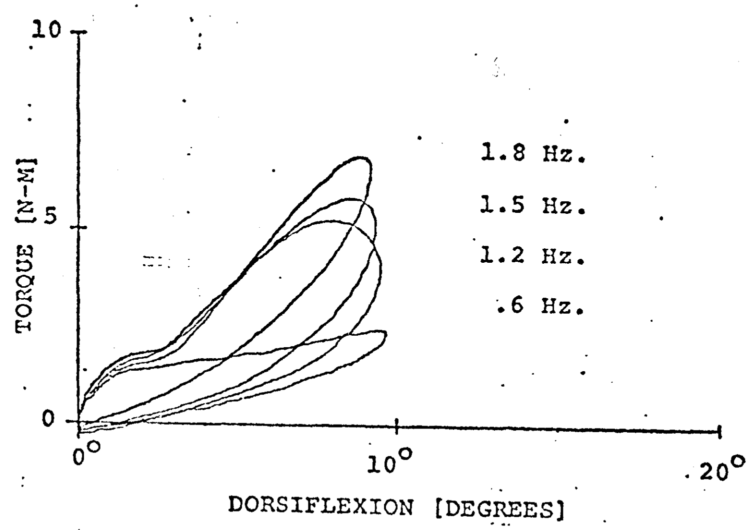


Figure 4.11: Torque versus dorsiflexion plots for the different frequencies, describing hysteresis loops. Note the absence of phase (c) for the 0.6 Hz loop

Comparison between data on different days (with periods up to three weeks between runs) shows repeatability and consistency. This may imply that the changes in the response to sinusoidal stretching at a given frequency could indicate the changes in the state of spasticity. If the above is true, then the computed RED_{LV} should be constant for a normal subject at a given frequency at any time. It would also be expected that normal subjects with similar physical dimensions would have the same RED_{LV}. The computed RED_{LV} for different subjects at different frequencies is tabulated in Table 4.1. As can be seen, there is a great variability between subjects, and comparison between subjects does not seem to be very useful. It should be borne in mind that the ED lumps together three different components of the stretch, as mentioned previously, and this fact complicates the interpretation of the data. The assumption that spasticity is of a viscous nature has a weak basis and has to be further investigated. One of the main features of hypertonicity is the stiffness about a joint, or the co-contracture of the muscles, and ignoring this fact (the constant k in Equation 4.3) might lead to gross misinterpretation of the data.

A comparison of the REDs computed for data taken on different days (up to three weeks between tests) from the same subject is given in Table 4.2.

Note that the lowest variability in the RED is for the data taken at the frequency of 0.6 Hz. This was expected, since the EMG records do not show any reflex activity at this frequency and therefore the measurement represents the passive mechanical muscle response to stretch. As the frequency is increased, the stretch reflex comes into effect and "contaminates" the equivalent damping measurement. This reflex activates the muscle and changes its mechanical characteristics, hence influencing the energy absorption of the muscle.

In general, for a given frequency and a particular subject, the equivalent damping is fairly constant and independent of time. This might be beneficial in assessing changes in the state of the subject due to medication, therapy, exercise, etc. These should be investigated further on spastic patients, and, if possible, compare the effects of medication on normal subjects in order to establish a norm.

4.4 Summary

Performance of the loaded manipulator was good. There were no problems with interfacing the apparatus with the subjects, and the procedure was readily accepted by the subjects. Disturbances introduced to the foot by the device were highly reproducible, yielding consistent results.

Different physiological aspects were tested by a variety of displacements, instructions and initial conditions. Parts of the results are in agreement with previously reported findings, while some parts were not found to be reported in the literature. The main points of interest are:

- For fast ramps (200-400°/s), the first burst of EMG in the soleus comes with a latency between 44 and 55 ms after the initiation of the stretch. This burst could be the monosynaptic reflex (MSR).

TABLE 4.1 COMPUTED RED FOR DIFFERENT SUBJECTS AT DIFFERENT FREQUENCIES

SUBJECT	FREQUENCY Hz	b (N-m-sec)	$RED_{(W)}$ (m-sec)	$RED_{(LV)}$ (N-m-sec)
DB	0.6	1.746	0.332×10^{-2}	488
W = 118 lb	1.2	1.973	0.376×10^{-2}	551
L = 16.25 in	1.5	1.714	0.326×10^{-2}	479
ρ = 13 in	1.8	1.307	0.249×10^{-2}	365
LH	0.6	0.842	0.169×10^{-2}	243
W = 112 lb	1.2	1.459	0.293×10^{-2}	420
L = 15.75 in	1.5	1.536	0.308×10^{-2}	442
ρ = 13 in	1.8	1.061	0.213×10^{-2}	306
WAM	0.6	1.636	0.210×10^{-2}	310
W = 175 lb	1.2	1.074	0.138×10^{-2}	203
L = 18 in	1.5	0.977	0.125×10^{-2}	185
ρ = 15 in	1.8	1.103	0.142×10^{-2}	209

TABLE 4.2 RED FOR SAME SUBJECT (DB) FROM DATA TAKEN ON DIFFERENT DAYS

FREQUENCY Hz	DATE	b (N-m-sec)	$RED_{(W)}$ (m-sec)	$RED_{(LV)}$ (n-m-sec)
	7-12-77	1.850	0.353×10^{-2}	518
0.6	7-31-77	1.746	0.332×10^{-2}	488
	8- 2-77	1.822	0.347×10^{-2}	509
	7-12-77	2.260	0.431×10^{-2}	632
1.2	7-31-77	1.973	0.376×10^{-2}	551
	8- 2-77	1.576	0.300×10^{-2}	440
	7-12-77	1.701	0.324×10^{-2}	475
1.5	7-31-77	1.714	0.326×10^{-2}	479
	8- 2-77	1.887	0.359×10^{-2}	527
	7-12-77	1.701	0.324×10^{-2}	294
1.8	7-31-77	1.307	0.249×10^{-2}	365
	8- 2-77	1.355	0.258×10^{-2}	378

- The latency and amplitude of this first burst seems to be dependent on the position of the foot. It is absent or very small in the range of -10° to -5° (plantarflexion) and is maximal within the range of -5° to +5°.

- Latency and amplitude of the above EMG burst is dependent on the angular velocity of the foot. The latency increases and the amplitude decreases with a decrease in velocity. This might infer a dependency of the above latency on angular acceleration of the foot. (The above was found for both ramps and sinusoidal displacements.)

- There is a significant increase in torque corresponding to the first burst of EMG in the soleus.

- Although a second burst of EMG was expected within a latency of 100 to 200 ms, none was observed in the data records.

- A voluntary reaction could be seen with latencies between 240 and 265 ms, depending on the instructions to the subject.

The above findings are interesting and point to the fact that the stretch reflex at the ankle might involve some special mechanisms. Pursuit of further understanding of the above phenomena is strongly recommended.

CHAPTER V

SPASTIC SUBJECT EXPERIMENTS

5.1 Introduction

As already mentioned in Section 4.1, the main goal of this thesis was to build the apparatus and demonstrate its capabilities for physiological and clinical applicability. As such, the data presented herein, although not statistically representative, does illustrate the usefulness of the appratus, and its potential for objective measurement of spasticity.

Acquiring patients to be tested was a problem due to insufficient time and the fact that the system was not located in the hospital. The system remained in the MVL for several reasons. Since the system is in the evaluation stage, it requires a great deal of time for runs, data presentation and analysis, modifications, etc. This time was not available at the CHMC Gait Analysis Laboratory due to their busy schedule. Also, approval had to be obtained from the Hospital's Committee for the Use of Humans as Experimental Subjects, as well as approval for transferring the equipment from MIT to CHMC.

The first patient examined (JF) was a 28 year old male, classefied as congenital athetotic-spastic. He is not on any medication nor does he smoke. His right foot tends to be inverted and he walks on the lateral aspect of the foot.

The second patient (JD) is a male, aged 66, classified as amytrophic lateral sclerosis (ALS), which developed about three years prior to the examination. His walk is extremely impared due to limited ability to move his feet. He is on daily medication and physical therapy.

The test protocol was the same as for the normal subjects, and instruction (a) of Section 4.2 was given to the patients. Each examination lasted about 45 minutes and the patient's acceptance of the device and test was very good. Only once, during the fast ramp experiment, one patient complained that his heel was pinched. The problem was immediately rectified and no further complaints were reported. The patients were extremely cooperative and easy to communicate with.

The runs were composed of fast ramps, sinusoidal and triangular functions. No slow ramps were used, primarily because of the limited time.

Both feet of patient JD were tested, while only the right foot of patient JF was examined.

5.2 Experimental Results

5.2a Patient JF

Athetosis has been described as a disorder of the central nervous system which is manifested clinically by slow, worm-like, involuntary, uncontrollable, unpredictable and purposeless movements (Koven and Lamm, 1954; Noback and Demarest, 1972). It was reported that the peculiar character of athetoid movements is caused by lack of reciprocal inhibition and the inability to relax the opposing muscles (Hoefer and Putnam, 1940; Putnam, 1939). Also, oscillations of the limb, caused by a series of clonic contractions, and grouping of the impulses in the EMG were observed (Lindsley, 1936; Hefer and Putnam, 1940; Neilson, 1974). The above characteristics were expected to be reflected in the data obtained from the new device.

At first, a series of fast ramps of 200°/s with 10° amplitude and initial position at 0° were performed. These runs were done in two modes: one following instruction (a) and another with the instruction to maintain a bias force and relax as soon as he feels the pedal moving. A typical record of data obtained from runs in the first mode is shown in Figure 5.1. It can be seen that the tibialis anterior muscle is in a continuous state of contraction prior to the stretch, in spite of the fact that the patient was supposed to relax. The soleus shows low level activity in comparison with the activity of the tibialis, therefore, there is some small initial dorsiflexion torque (i.e. the foot is pulled up by the contracted tibialis anterior). As the fast ramp dorsiflexes the foot, the soleus EMG shows a burst of EMG, probably an MSR, similar to the one seen in the normals. The latency of this burst is about 50 ms which might indicate an intact spinal loop. Immediately following this burst of EMG, there is an exaggerated increase in the activity of the tibialis anterior, which continues for a long period with spacially distinguishable groups of EMG pulses. The amplitude of the EMG bursts subsides slowly as the time passes. At the same time, the soleus also shows activity, but of a lesser potency. Consequently, there appear large fluctuations of the torque, resembling clonic oscillations.

The same experiment was then repeated with a modified version of instruction (c). Instead of pushing down on the pedal, the subject was asked to relax as soon as he feels the pedal moving. A data record is presented in Figure 5.2. The data is very similar to the previous run described above. In both cases, the EMG latencies are the same. Worth noting is the quenching, or decrease, in the EMG activity of the tibialis during the first large burst of EMG in the soleus which occurs at a latency of about 50 ms. The level of tibialis EMG drops below the pre-stretch level, i.e. increased inhibition. It is different from the normals, where a small burst of tibialis EMG accompanies the large soleus EMG burst. It would be of great interest to further study the phenomenon and its mechanisms. After this silent period, a large burst of EMG activity is seen in the tibialis at a latency of about 99 ms. From this point on, both muscles are co-contracted and the imbalance in the level of activation of the muscles causes oscillations in torque. The dominant muscle is clearly the tibialis anterior and this is in accord with the state of the patient as described in Section 5.1.

These findings point out the fact that for this athetoid-spastic patient, the muscle response to stretch is reproducible and at least the first part of the response is consistent and the timing of the EMG activity in both soleus and tibialis anterior is consistent.

A second series of tests involving sinusoidal and triangular functions were also performed. Figures 5.3 and 5.4 show responses to a slow sinusoidal displacement at 0.6 Hz and a triangular (constant velocity) displacement at 24°/s, respectively. Again, in both cases, the torque record shows the same oscillatory cahracteristics with a frequency of about 3.5 Hz. This frequency is within the range of 1.5 to 4 Hz at which both agonist and antagonist muscle groups contract vigorously but asynchronously (Neilson, 1974). The tibialis

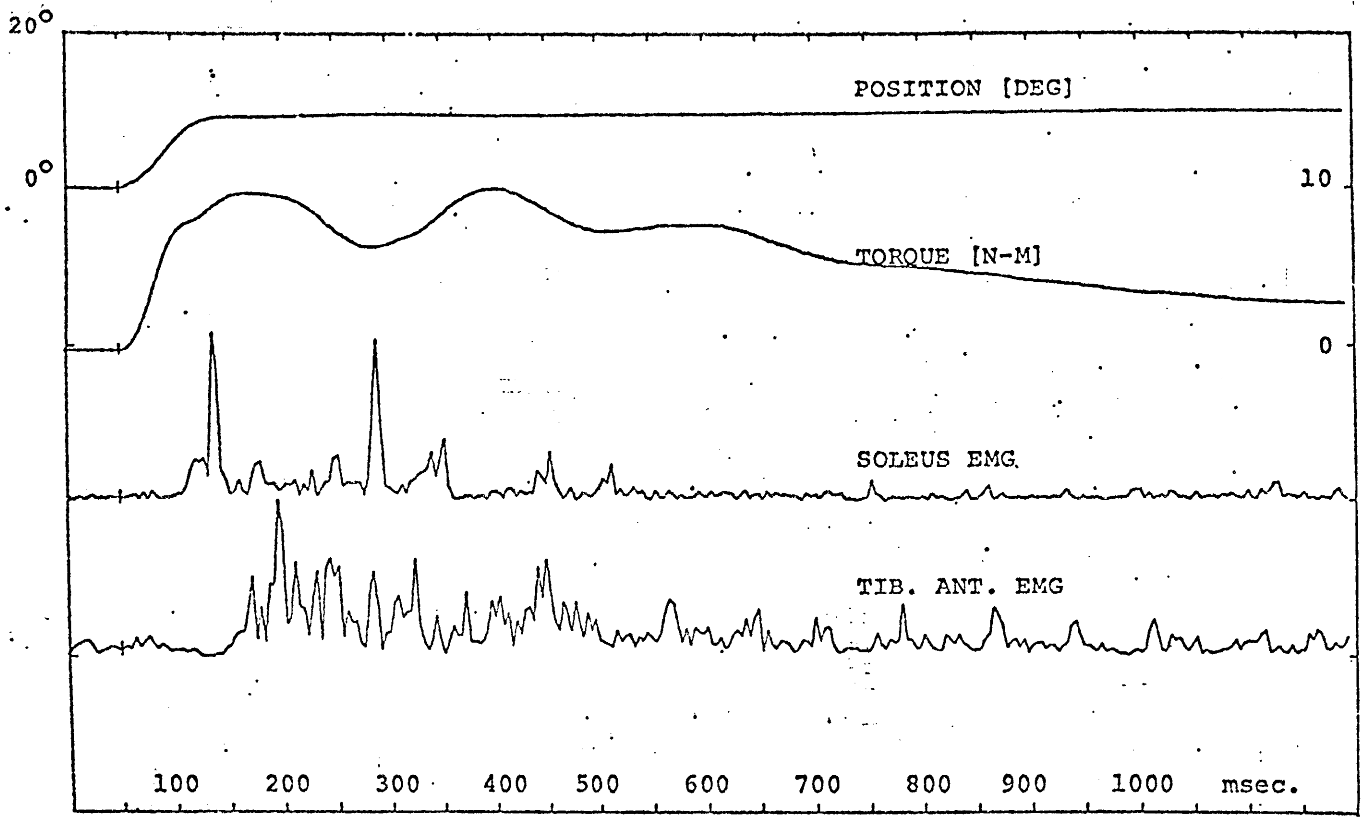


Figure 5.1 Results of a 200°/s dorsiflexion of the right foot of patient (JF). The instruction was to reflex throughout the experiment.

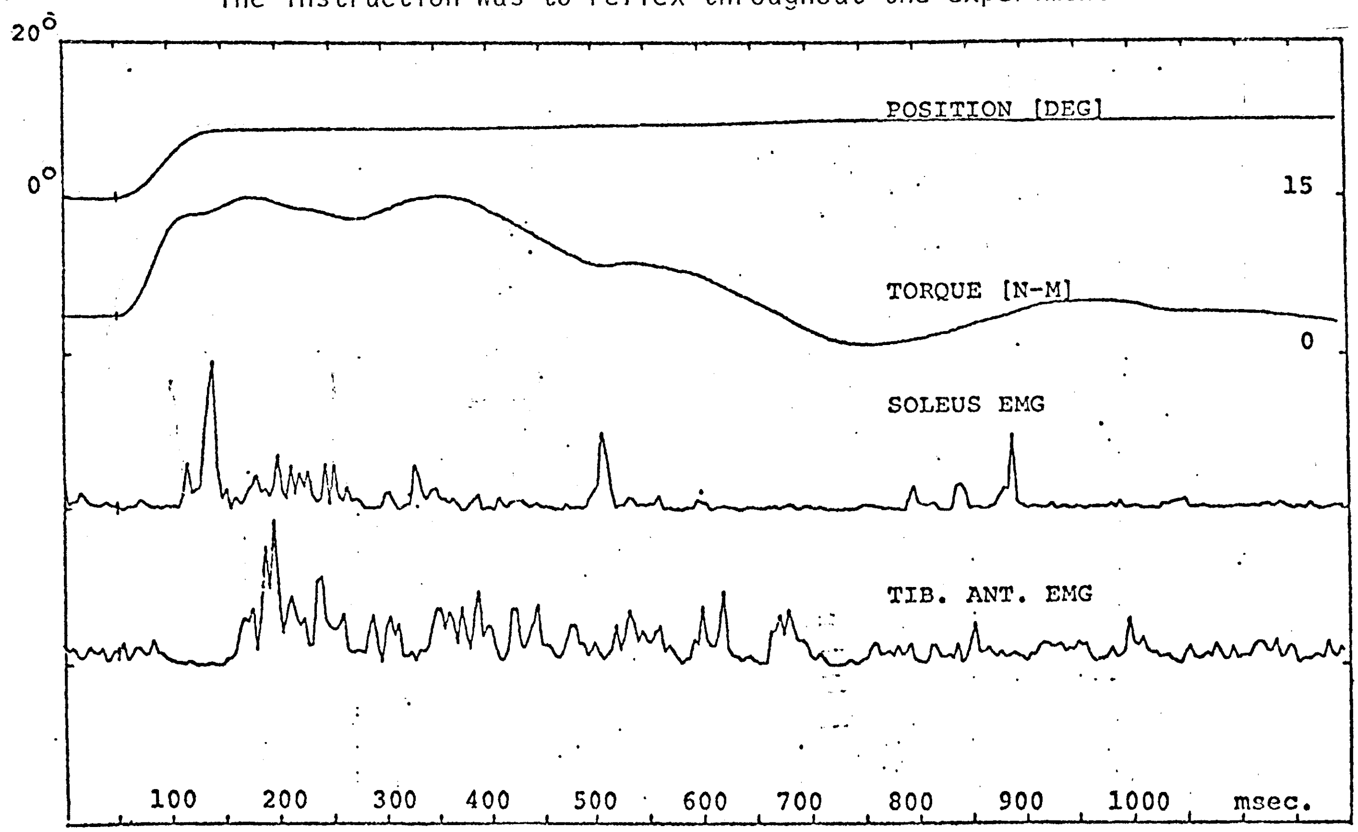


Figure 5.2: Results of a 200°/s dorsiflexion of the right foot of patient (JF), with prestretch bias force. The instruction was to relax as soon as the pedal started moving.

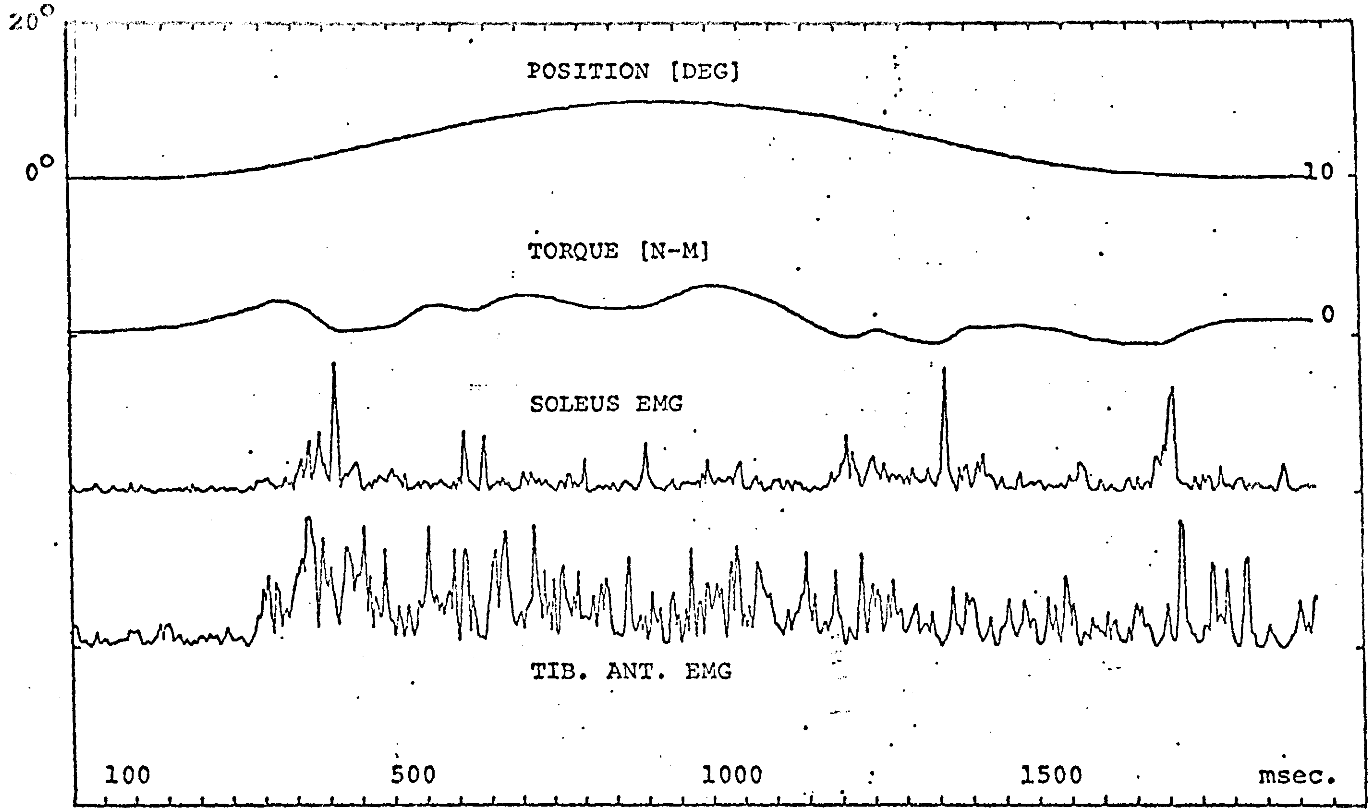


Figure 5.3: Results of a 0.6 Hz dorsiflexion of the right foot of patient JF. The instruction was to relax throughout the experiment.

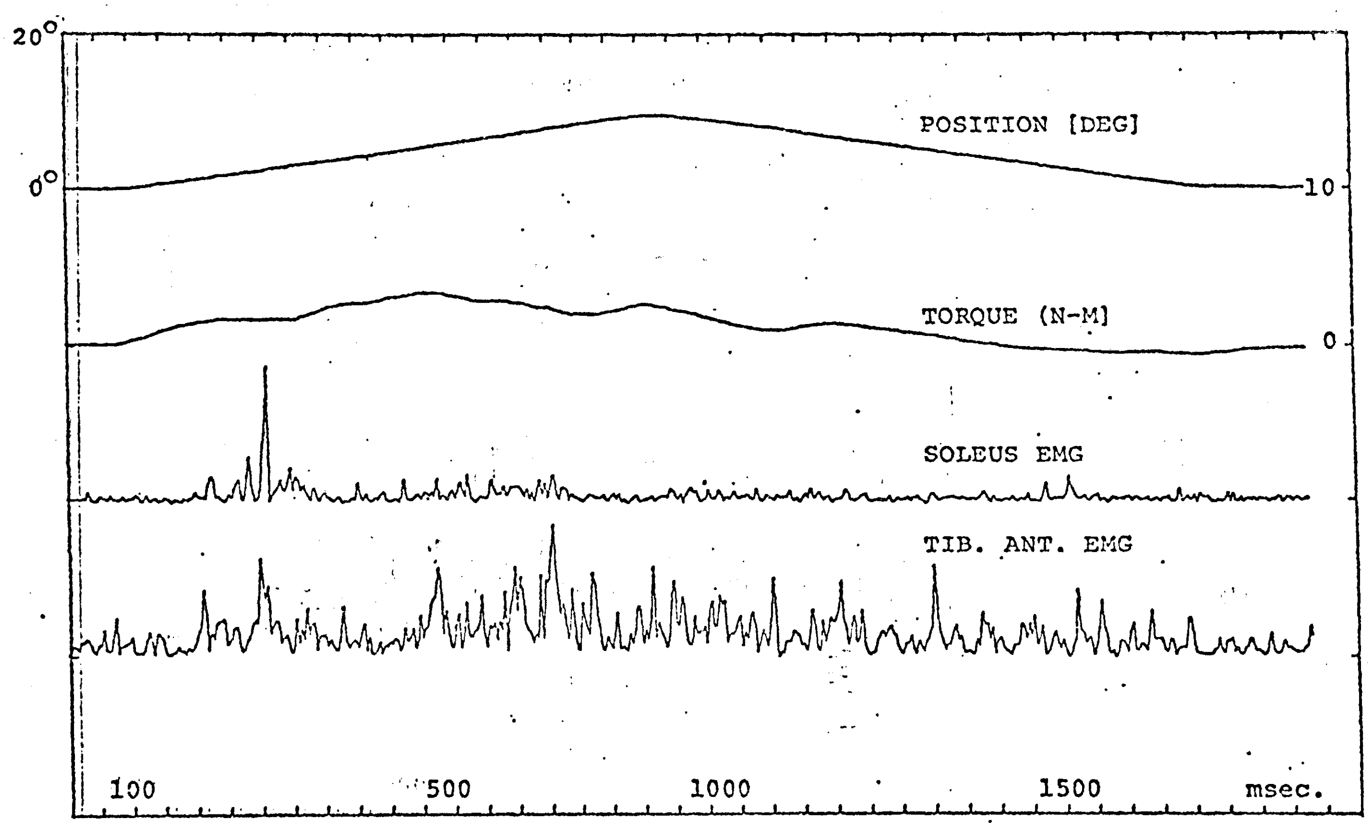


Figure 5.4: Results of a 24°/s triangular dorsiflexion of the right foot of patient JF. The instruction was to relax throughout the experiment

anterior EMG is pronounced and shows low frequency fluctuations of the mean, corresponding to the fluctuations in torque. It was found that, in general, the triangular displacements are more consistent than the sinusoidal ones and are less irregular. This is shown in Figure 5.5 in the form of plots of torque versus position. The loops obtained are markedly different from those obtained from normals. Also, the faster the frequency of the sinusoidal displacements, the more regular is the shape of the loops (Figure 5.6). This might point out the low frequency characteristic of athetosis and its sensitivity to slow movement. The RED was computed for this patient and is shown in Table 5.1

Table 5.1 RED of patient (JF) (Right foot)

Frequency Hz	b (n-m-s)	RED_W (m-s)	RED_{LV} (n-m-s)
0.6	2.868	0.586×10^{-2}	964
1.2	1.474	0.301×10^{-2}	496
1.5	1.725	0.352×10^{-2}	581

The RED is highest for the slow movement and there is a great variability within each frequency, but in general the values are greater than for normals, in particular for the 0.6 Hz frequency. This again points out the same hypothesis mentioned earlier, that the patient is sensitive to slow movements. It is believed that the RED would not serve as a good measure of athetoid-spasticity, but to substantiate that, further studies would have to be performed.

5.2b Patient JD

Amytrophic lateral sclerosis (ALS) is a degenerative disease of the motor system with bilateral involvement of the pyramidal and anterior horn cells. There is a degeneration of both upper and lower motor neurons. Most of the affected muscles show evidence of the degeneration of lower motor neurons, incuding atrophy, fasciculation and weakness. Some muscles exhibit signs of upper motor neuron deficit, with spasticity and hyperreflexia, and, at times, Babinski signs. The lower motor neurons of cranial nerves may also exhibit signs of degeneration (Noback and Demarest, 1972; Clark, 1976).

The same experimental protocol was applied to this patient, except that he was asked to relax throughout the test period. This was necessary, since the patient had reduced voluntary control of his feet. Both feet were tested to see the bilateral effects of the disease.

First, a series of fast ramps, 200°/s, 10° amplitude and initial position at 0°, were performed. Figure 5.7 shows four runs' data superimposed. The results are extremely reproducible and variability is minimal. The latency of the first large burst of EMG in the soleus muscle is 58 ms. Both the soleus and tibialis anterior show continuous low level EMG activity which points out the fact that both muscles are active at rest. This, in turn, causes maintained high level of torque at the end of the displacement. The torque and EMG records

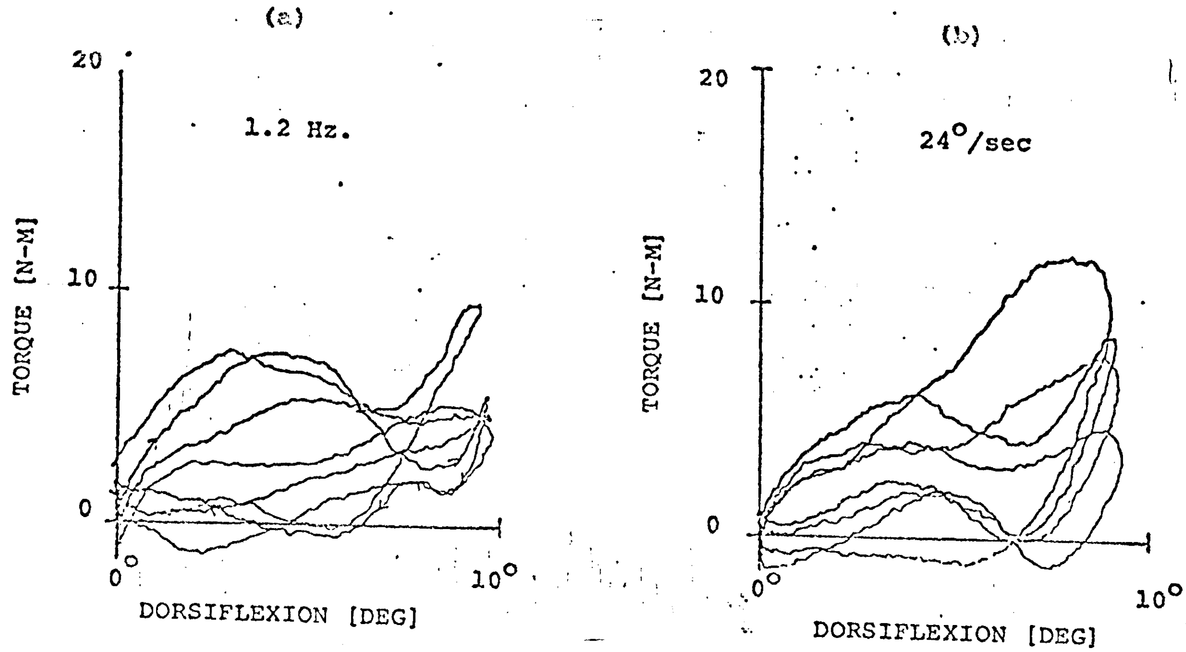


Figure 5.5: Torque versus dorsiflexion plots showing irregularity of hysteresis loops for (a) sinusoidal displacement and (b) triangular displacement with same period. Four runs are superimposed in each frame. Patient JF.

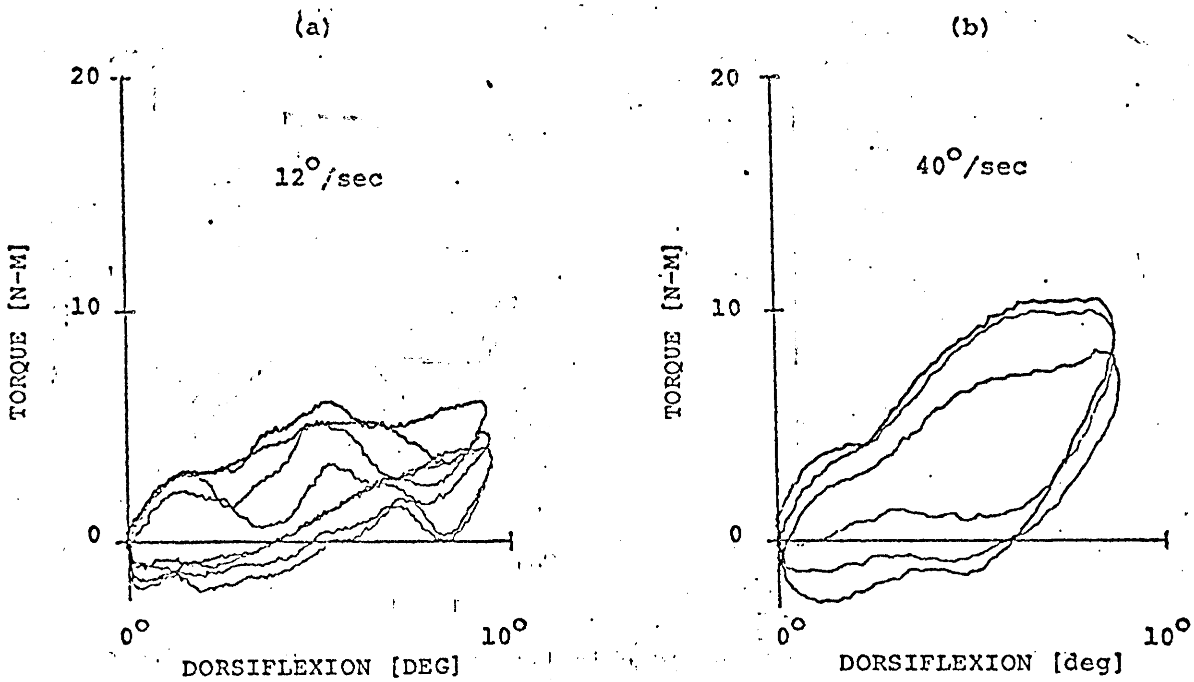


Figure 5.6 Torque versus dorsiflexion plots for triangular displacements at two different velocities, oscillations are more pronounced at the lower velocities. Patient JF.

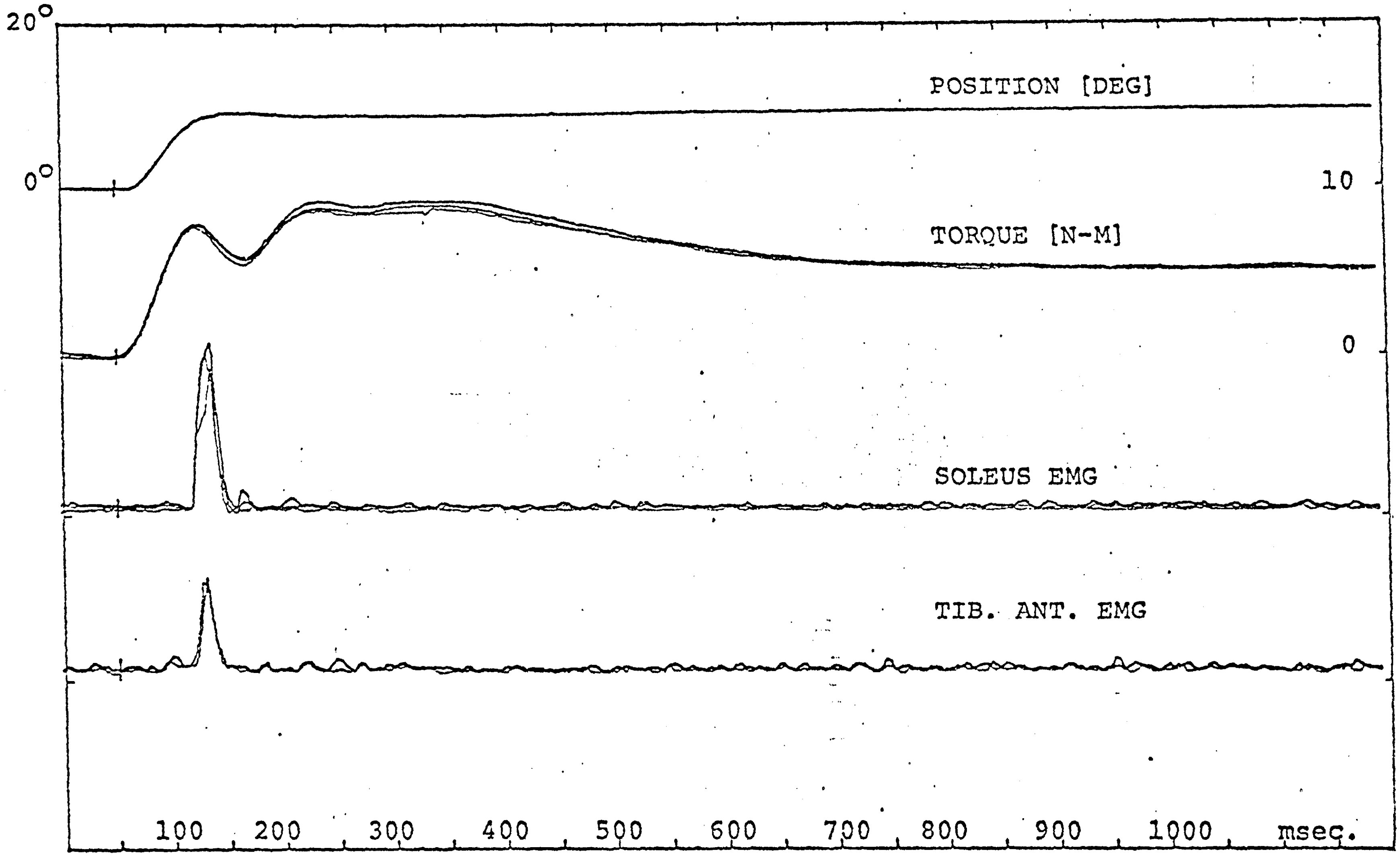


Figure 5.7: Superimposed Results of Four Runs at 200°/sec Dorsiflexion of the Right Foot of Patient (JD).

are very similar to those obtained from normal subjects, except for the co-contraction of the muscles.

Second, sinusoidal displacements were performed, and, again, reproducibility was good. A typical sinusoidal record for the right foot is shown in Figure 5.8 with four runs superimposed. Plots of torque versus position for the same four runs are presented in Figure 5.9. The shape of the hysteresis loop is quite different from that of the normal or athetotic subjects. Note that the slope of the loop, which represents the stiffness, is higher than for normals. (Compare Figures 4.2 and 4.3).

The same protocol was applied to the left foot, except that the range of motion was limited to less than 10° dorsiflexion and testing could only be performed up to 6°. Within the range, the foot was less stiff than the right foot.

This patient was known to have (by clinical criterion) both upper and lower motor neuron damage (but more UMN). From the fast ramp response, it does not seem likely that the lower motor neurons are damaged, since there appears to be a "normal" spinal response to stretch. The problem might lie with the lack of inhibition or increased facilitation from higher centers. The tone of the muscles is increased (hypertonus), but there are no signs of increased deep tendon reflexes (hyperreflexia) nor clonus, unless they cannot be detected due to hypertonicity. The RED values for this patient are listed in Table 5.2.

Table 5.2 RED for both feet of Patient JD

Foot	Frequency Hz	b (n-m-s)	RED_W (m-s)	RED_{LV} (n-m-s)
	0.6	7.211	1.030×10^{-2}	1898
	1.2	3.364	0.482×10^{-2}	885
	1.5	2.891	0.414×10^{-2}	760
	0.6	3.742	0.537×10^{-2}	983
	1.2	2.082	0.298×10^{-2}	548
	1.5	1.717	0.246×10^{-2}	452

These values, especially for the lower frequencies, are much higher relative to corresponding values seen in normals. This finding, and the fact that there is minimal varialbility in these results for a given frequency, could be advantageous in assessing <u>changes</u> in the state of the patient. Of course, further tests have to be performed on patients to verify the results and to further understand the characteristics of his neuromuscular disorder.

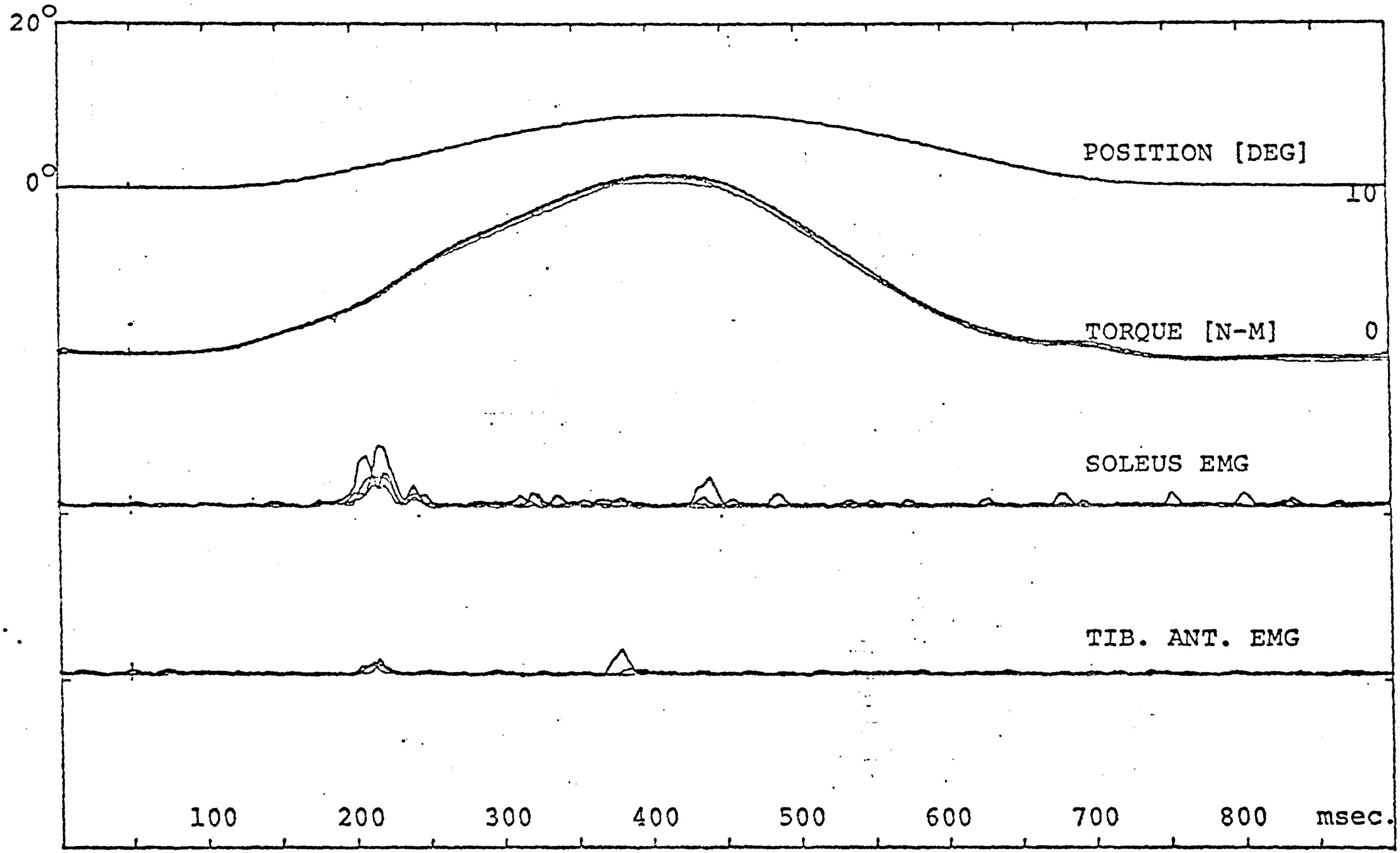


Figure 5.8: Superimposed Results of Four Runs at 1.5 Hz. Dorsiflexion of the Right Foot of Patient (JD).

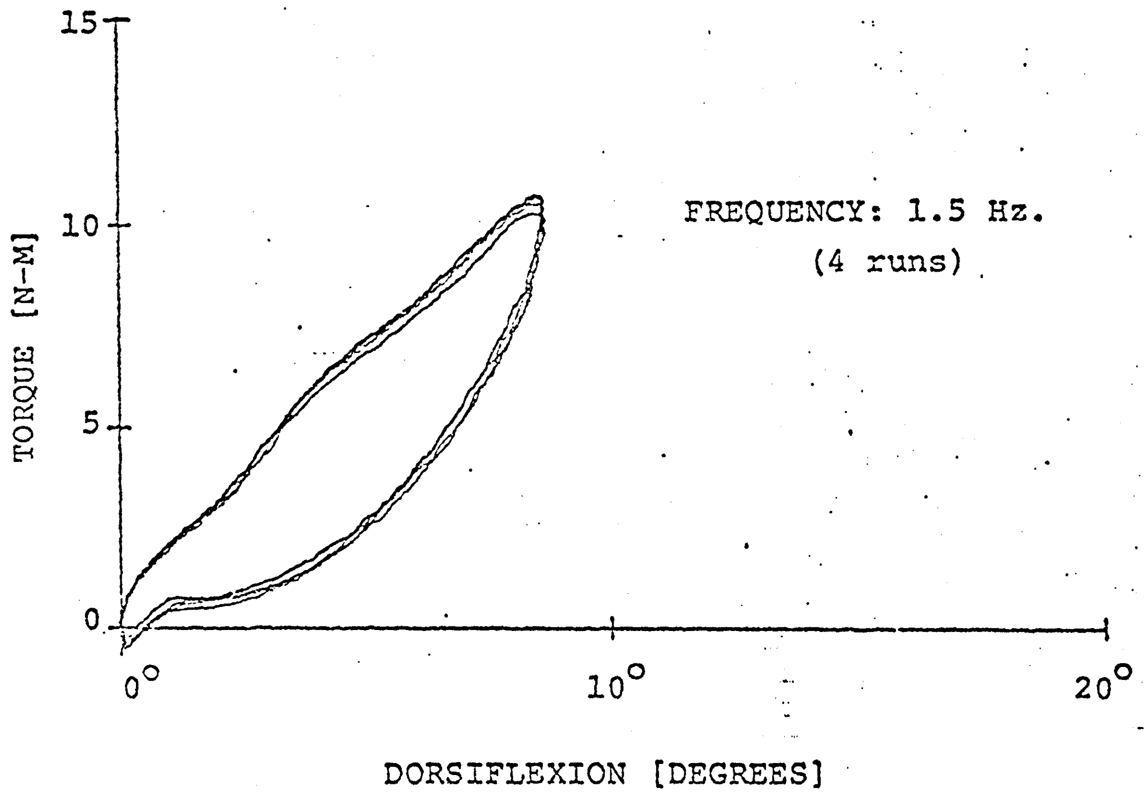


Figure 5.9: Four Superimposed Plots of Torque vs. Dorsiflexion from the Data Presented in Figure 5.8. Compare this with Figures 4.10 and 4.11.

5.3 Summary

Although only two patients were tested, and statistical analysis could not be applied to the data for quantitative studies, semi-quantitative analysis was very useful. Both patients represent different cases of spasticity and their clinical state is of a complicated nature. It would have been preferable to have hemiplegic patients for comparison, but none were available. In spite of this, the data from the two patients points out the characteristic response for different disturbances in position. These were found to be different for each patient, and from those of normal subjects.

Patient JF who is clinically diagnosed as athetotic-spastic showed the typical oscillatory response associated with athetosis. The data pointed out the fact that the tibialis anterior is the predominantly affected muscle and that there is also a lack of reciprocol inhibition and facilitation from supraspinal centers. Of interest is the fact that the latency of the MSR is normal and there is reciprocal inhibition during the MSR. This might point out the fact that the lower motor neuron is intact and the MSR supresses the influence of higher centers. On the other hand, the stretch triggers an exaggerated reaction in both the agonist and antagonist muscles, and consequently an oscillatory clonic reaction of low frequency. The data also shows the sensitivity of the patient to slow stretches is high and causes violent reactions. It seems that the reaction is more irregular and exaggerated when the foot is moved sinusoidally rather than at constant velocity (triangular motion). This might imply sensitivity to velocity, or more likely, changes in velocity, i.e. acceleration.

Patient JD, although clinically defined as LS, is not a clear cut case, since the cause and nature of his disease is unknown. He is suspected of having both upper and lower motor neuron degeneration of unknown severity and location. The data obtained for this patient is extremely repeatable and consistent. EMG recordings show that both agonist-antagonist muscles are active continuously and the patient does not have full voluntary control over his feet. The MSR latency, following a rapid stretch is constant and seems to be normal, but there is an increase in the size of the MSR of the tibialis anterior which indicates increased co-contraction relative to the one seen in normals. "Stiffness of the joint" is the major characteristic and was found to be much higher in normals, due to the fact that both muscles are contracted. The lack of any changes in the level of activity of the muscles following the stretch is quite interesting. This was reported also by Kearney (1976) who testified that ankle displacement failed to elicit anything but a monosynaptic reflex in spastic patients, which was very puzzling, but seems to be indicative of "non-EMG" features in the "stiffness".

Further studies are needed on these patients to find out whether the state of the patient has changed and what the characteristics of the response are. The repeatability of the responses for patient JD could be used as a basis for detecting <u>changes</u> in the progression of the disease or the effect of medication. THe main point is the fact that the increased stiffness in patient JD correlates with the clinical impression and this implies quantification of the "loose" clinical impression. The results presented herein are far from being conclusive and many further studies have to be conducted on different types of spastic patients, in particular hemiplegic ones.

CHAPTER VI

SUMMARY CONCLUSIONS AND RECOMMENDATIONS

6.1 Summary

The main objective of this thesis was to develop a prototype device which may be applied for objective clinical assessment of spasticity. Such an apparatus has been designed and built, and preliminary testing of normal and pathological subjects has been performed.

The device is an electromechanical position servomechanism, capable of moving a foot in a controllable manner. Position, torque and two myoelectrical signals from the triceps surae and tibialis anterior are the parameters of interest. The position reference signal, recording of the data, processing and displaying it, are obtained via a PDP-11/34 digital computer. Several normal subjects and two spastic patients were tested in different modes. Fast ramps, slow ramps, sinusoidal and triangular displacements have been used, with a variety of combinations such as initial position, initial torque, range of motion, different instructions to the subjects, etc.

The primary goals of these preliminary testes were to find out how well the device performs, if the apparatus would be accepted by the subjects, and whether or not the results obtained from normal and patients are in agreement with the known physiological aspects of the neuromuscular system, and if they can be of any use in the assessment of pathological cases.

The number of normals and patients was too small for a sound statistical analysis, due to the limited time available. Nevertheless, the semi-quantitative data analysis revealed a great deal of valuable information.

In the analysis of the data, several approaches have been taken. EMG signals were compared on the basis of shape, size and latency, with the latter being a major parameter. Torque records were plotted vs. time and examined for reproducibility and correlation with EMG activity. Plots of torque vs. position for sinusoidal disturbances have been used to visually present the hysteresis loops and identify components related to the muscle passive and active properties. Relative equivalent damping coefficients were computed for the sinusoidal runs and were correlated to the frequencies and subjects. Most records were individually analysed, although some records, that showed high similarity, were averaged. Commonly runs lasted for no longer than one hour, and in the cases of normal subjects only the right foot was tested.

6.2 Conclusions

The apparatus developed to manipulate the foot about the ankle joint has been found to produce valuable information relating to the different aspects of the leg muscles' response to passive stretch. The device has been working well and the subjects' acceptance of the system has been uniformly good. The system is easy to operate and requires little skill, particularly the computer operation which is self-explanatory.

Normal subjects' data was generally found to be consistent with previously reported related research. Soleus EMG latencies in response to fast ramps in dorsiflexion were found to be in the range of 40 to 60 ms, while voluntary reaction time was about 250 ms. Additionally, the data also revealed new phenomena, which have not been found in the literature, or just vaguely reported in the past. In spite of the fact that subjects were fully relaxed, "monosynaptic" reflex activity was observed. This reflex was found to contribute significantly to the increase in torque produced by the muscle, contrary to what has been reported in the literature with respect to different muscles (Allum, 1974; Vallbo, 1973).

Slow velocity ramps revealed that the latency and amplitude of the MSR changes as a function of change in velocity. It has been found that the latency of the MSR increases with decreasing velocities while its amplitude decreases. The amplitude changes of the MSR with velocity of stretch were reported elsewhere (Kearney, 1976), but no available source has reported changes in the MSR latency with slow movements. Indeed, most investigators have been introducing moderate to fast stretches, and reported that observations of "monosynaptic" activity in the biceps apply only to rapid stretches, with no response to slow stretch (Hammond, 1960; Allum, 1974).

It was found during the course of testing normals, that the MSR latency decreases with increasing ramp velocity up to 160°/s, beyond which the latency stays constant (at about 46 ms). Further investigations showed a dependency of this latency on acceleration, since the MSR was triggered as soon as the velocity of stretch reached a threshold velocity. Considering the important role played by the ankle muscles in postural control and the direct influence of the vestibular system on these muscles, this may reflect a special control system operating at the ankle joint.

In addition, it was found that the MSR is dependent on the position of the foot and beyond a certain angle of plantarflexion (typically less than 5°), there will be no MSR observed. All the findings above apply also in the case of sinusoidal displacement. These findings are quite interesting and require confirmation and further studies which are beyond the scope of this thesis.

Interesting results were obtained from the sinusoidal displacements (which were similar to those obtained from triangular motion). Plots of torque vs. position describe hysteresis loops with four distinct phases. The first phase (a) is attributed to the rheological properties of muscle and the inertia of the foot, phase (b) corresponds to viscous damping (torque almost constant), phase (c) represents an increase in muscle stiffness related to the MSR, and phase (d) return to initial position. These loops depend on frequency (increased stiffness with increase in frequency up to 1.8 Hz where it remains constant), but are quite repeatable for a particular frequency. Corresponding relative equivalent damping values are also constant for a given frequency.

Patients' data were much different than normals' data, in particular in the case of the athetoid patient. In general, patients have very little, or no, voluntary control over their limbs and are unable to relax. In fact, if they are told to consciously relax, their condition might be aggravated. Both patients manifested pre-stretch EMG activity in both muscles (soleus and tibialis), implying a continuous state of co-contraction of the agonist-antagonist muscle, and hence, "stiff" ankle joint.

There was a clear difference of the data between the two patients as well as between the patients and the normals. The athetotic patient demonstrated the typical oscillatory response attributed to athetosis, with a characteristic frequency of about 3.5 Hz. Following the fast ramp stretch, the EMG activity in both muscles intensified with dominance of the tibialis, indicating stronger involvement of this muscle. This is consistent with the fact that the patient's foot is inverted, a state caused by permanent contracture of the tibialis.

The ALS patient showed constant stiffness with no change in EMG activity level except a normal MSR burst accompanied by contraction of the opposite muscle, indicating a lack of reciprocal inhibition, or rather facilitation of the antagonist. Variability in the data for this patient was minimal and this fact might be useful for detecting changes in the state of a patient who undergoes therapy.

The tests for both patients were performed in one session and were not repeated on different days. As such, the data connot be considered representative, and more tests are required.

In recapitulation, it may be said that the results obtained with the new device are very promising and consistent. The system can be applied to the study of a variety of neuromuscular disorders, but the examiner should always be aware of the neuromuscular symptoms and disability of the patient whom he confronts. As Landau (1974) put it "...to use the activity of the ankle joint jerk as a measure of therapeutic success, in some conditions might be like mistaking the moss on a tree for the forest...". Doshay's (1964) original hope to achieve an apparatus that could quickly tell the investigator that a new drug "X" is perhaps 37% more or 20% less effective than Artane, has not, as yet, been reached. However, one should have reason to believe that the foot manipulator and other exploratory efforts towards improvements in the current instrumentation will bring the day when this goal will be realized.

6.3 Recommendations

Since the system developed is a ptorotype, there is a lot of room for improvements. The device was built to be as flexible as possible and to accommodate certain needs. During the course of testing and evaluation, a few points for improvement came about.

Hardware: At the present moment the system is built to accommodate patients with their legs positioned from fully extended to fully flexed. This is one of the reasons why the device is mounted on the high profile cart. It is most likely that the system will be used in a mode where the leg is flexed about 120°. The use of the system on a fully extended leg is not likely, since the patients have problems extending their legs. Thus, the cart structure can be modified to a lower profile. This will be advantageous, for it will eliminate the need to raise the patient with the dental chair.

The attachment of the device to the chair should also be modified to allow easier adjustment of the chair relative to the pedal. Provisions should be made for both medial and lateral adjustments.

A big improvement would be replacement of the old-fashioned dental chair with a more modern one to make the system more esthetically appealing.

To accommodate a larger population of patients, including children, three heel cups and strap combinations should be made for small, medium and large feet.

A foot rest is also needed for the opposite leg. This foot rest can be either an integral part of the frame of the device or a separate stool or sling. If the system is lowered enough, then it might be sufficient for the foot to rest on the floor.

Either a separate cart or vibration insulation should be used for the power supply, if mounted directly on the pedal's cart, since the transformer and fan cause some audible noise and vibration that might influence the patient.

A ratchet wrench whould be used for easier adjustment of the pedal's height. In some cases, wehre the reactions of the muscles to stretch are exaggerated, there might be a need for strapping the thigh or even the uppper body. This woold be required for the safety of the patient and for the prevention of misalignment in the positioning of the subject.

In order to make the system more universal and easy to use in a clinician's office, it is advisable to eliminate the need for the PDP 11/34 computer. A microprocessor based minicomputer, specifically tailored for the foot manipulator, could be used. This computer could not only provide the analog reference signal, but also monitor and process the data and present it on a CRT, a chart recorder or an x-y plotter. (Such a microcomputer can be purchased from the Biomedical Engineering Center at MIT.)

Software: At present, a very general program is being used to control the system. It is written in FORTRAN and is self-explanatory, but it is slow and takes some 200 blocks of storage space. A great deal of improvement could be done in this area. For instance, selecting the most useful modes of operation and rewriting it in assembly language would improve speed of execution and would shorten testing time. The system can be fully automated with the functions and parameters preselected and a complete sequence of tests performed without operator intervention. This could be easily done by using the microprocessor based computer. Pattern recognition routines, event timing, processing of data, comparison of the tests with previous data, and other statistical analyses could also be implemented.

6.4 Further Research

It seems that in the area of physiological investigations, the deeper one gets into the subject, the more questions arise about the mechanisms involved and the nature of the system.

The great importance and role of the ankle joint and its musculature in maintaining an upright posture and locomotion implies that the system might be using special mechanisms, different from those of other joints. Very little work has been done on the ankle joint and the little data obtained thus far strongly indicates that much further investigation remains to be performed, many aspects of which are mentioned below.

6.4.1 Physiological Studies on Normals

The physiological aspects of the stretch reflex of the triceps surae muscle, in conjunction with postural control mechanisms, are wide open to investigation. There are many questions to be answered in this area before applying the system to pathological cases. There are many important questions to be answered such as:

- What are the effects of displacement velocity, inertial position, range of motion and torque on the neuromuscular response?
- What are the parameters to which the system is most sensitive?
- How do instructions influence the response?
- What are the mechanisms involved in eliciting the stretch reflex and modulating it?
- How does the vestibular sytsem interact with the stretch reflexes?

To help answer some of these questions, different approaches could be taken, such as blocking cutaneous joint receptors, measuring EEG activity associated with ankle response to stretch, and applying more complex patterns of displacements as stimuli.

6.4.2 Clinical Investigations

Much further research is needed in the area of pathological cases. Since the system allows for measurements at the ankle joint, it is feasible to cover a a larger spectrum of patients with neuromuscular disorders. It will be possible to test cerebral pathological cases and also patients with spinal cord lesions at any level. As seen in previous chapters, different types of spasticity are quite distinguishable, allowing for categorization of responses and their severity. Detection of changes in the state of the patient would be useful in determining the contribution of drugs or physical therapy to the improvement of the patient.

In addition to passive and active manipulation of the foot, tendon taps may be introduced and the responses measured via the pedal. An instrumented neurological hammer could trigger the computer and the data could be monitored and processed. Also H-reflex data could be compared to the data obtained from a stretch and thus verify EMG latencies.

6.4.3 Gait Stimulation

In conjunction with regular gait analysis of patients, gait patterns, angles, and velocities, measured from actual gait could be simulated on the apparatus and the results compared to those of the actual gait. This will help in the analysis of the major causes for impaired gait.

Is gait impaired due to cutaneous and joint receptors, or is it due to disorders of the lower or upper motoneurons?

How do the EMG's timing correspond in active and passive stretch?

What are the effects of position, velocity and torque on the response of muscle, and how does this response interfere with normal gait pattern?

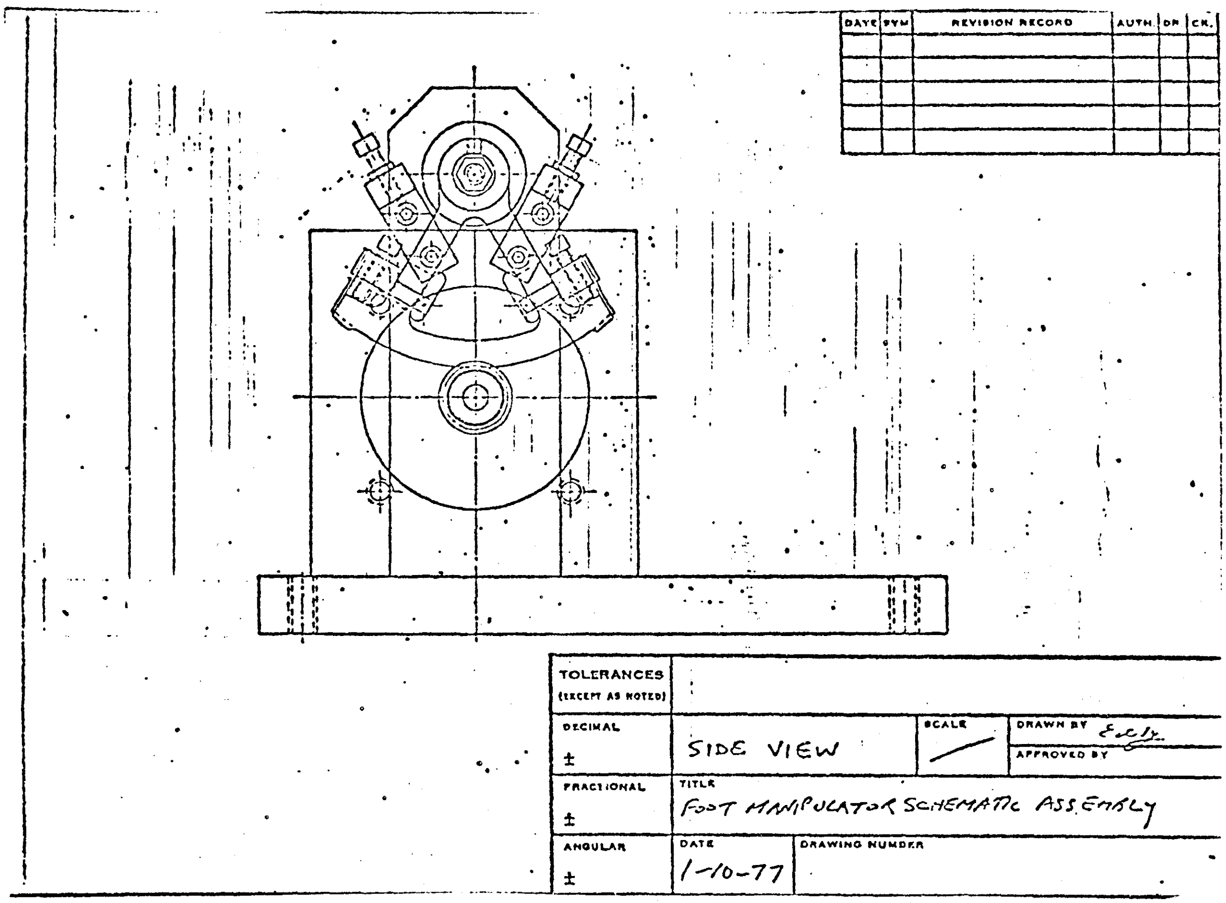
These are only a few of the possibilities in this area.

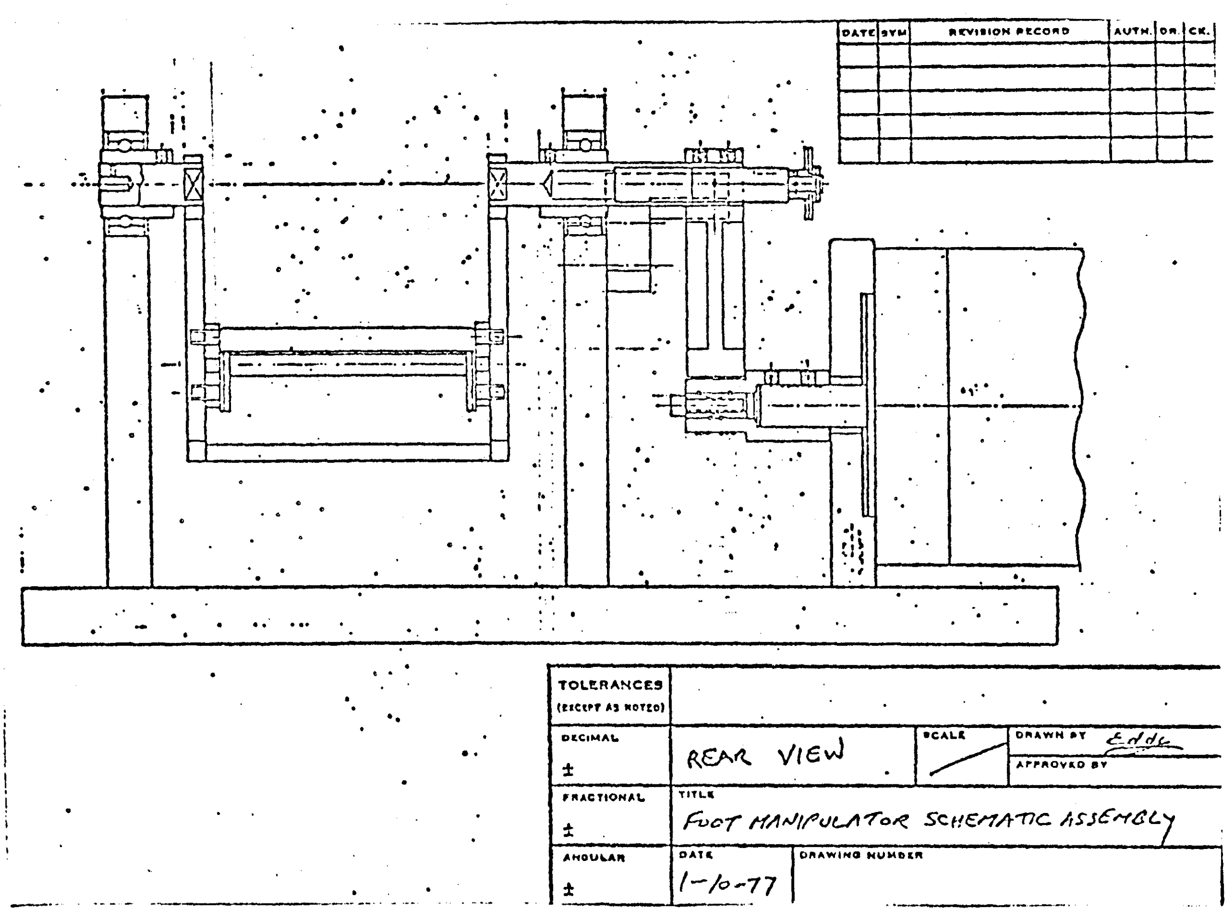
6.4.4 Orthopedic Applications

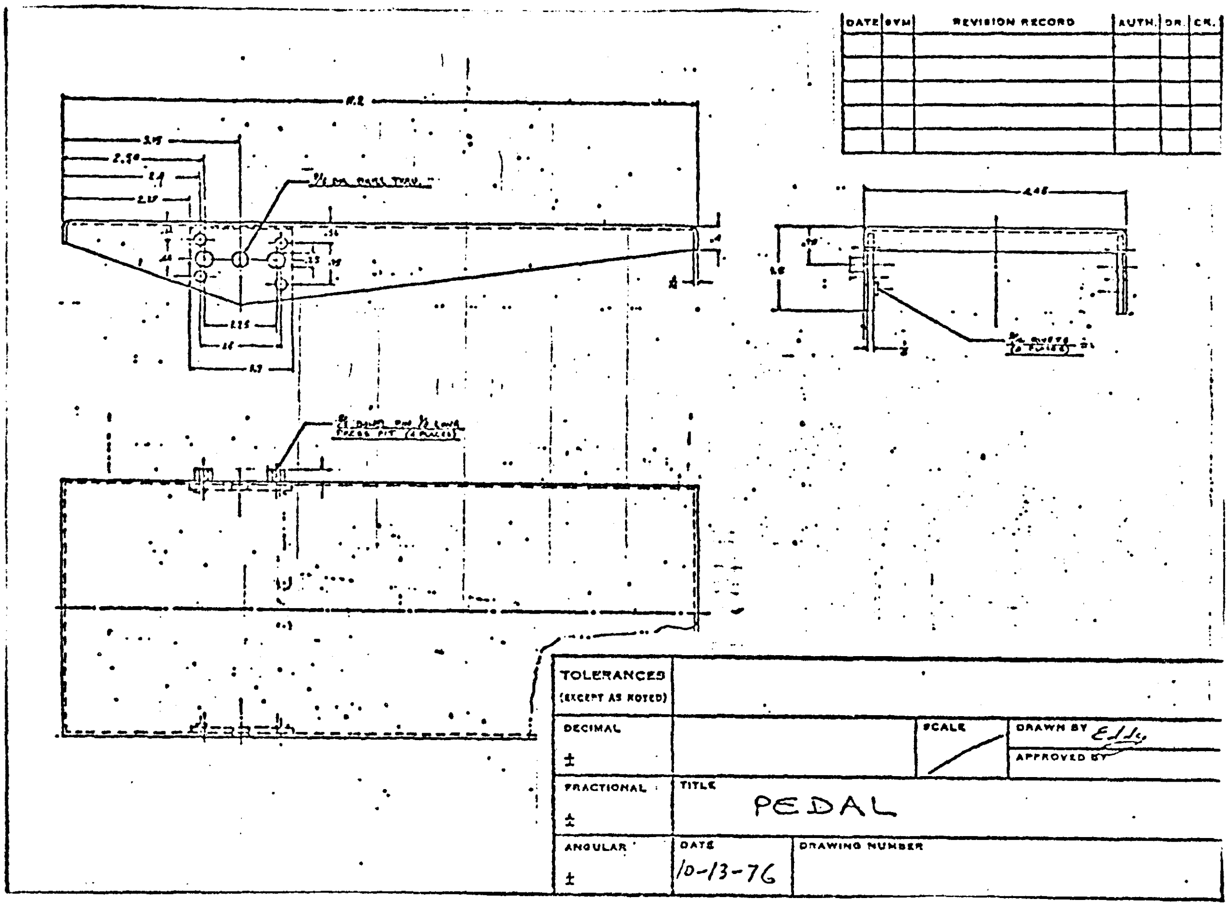
In some cases, permanent contracture of muscles requires surgical relief of tension. This is done by releasing the tendons or elongating them, or by partially cutting the muscle. Also contracted collagenous tissue mass or bound-down joint capsules are being cut. In this context, determination of the forces developed during passive stretch of the ankle muscles and the corresponding relationship between torque and position, could aid the orthopedic surgeon in his decision. Post-operative measurements would indicate the change in muscle tension length characteristics and could be compared with the pre-operative data. This, in turn, could verify the benefits gained by the operation.

The device might be used also for post-operative passive exercises, which are usually performed by a physical therapist.

APPENDIX A: DRAWINGS OF COMPONENTS AND PARTS



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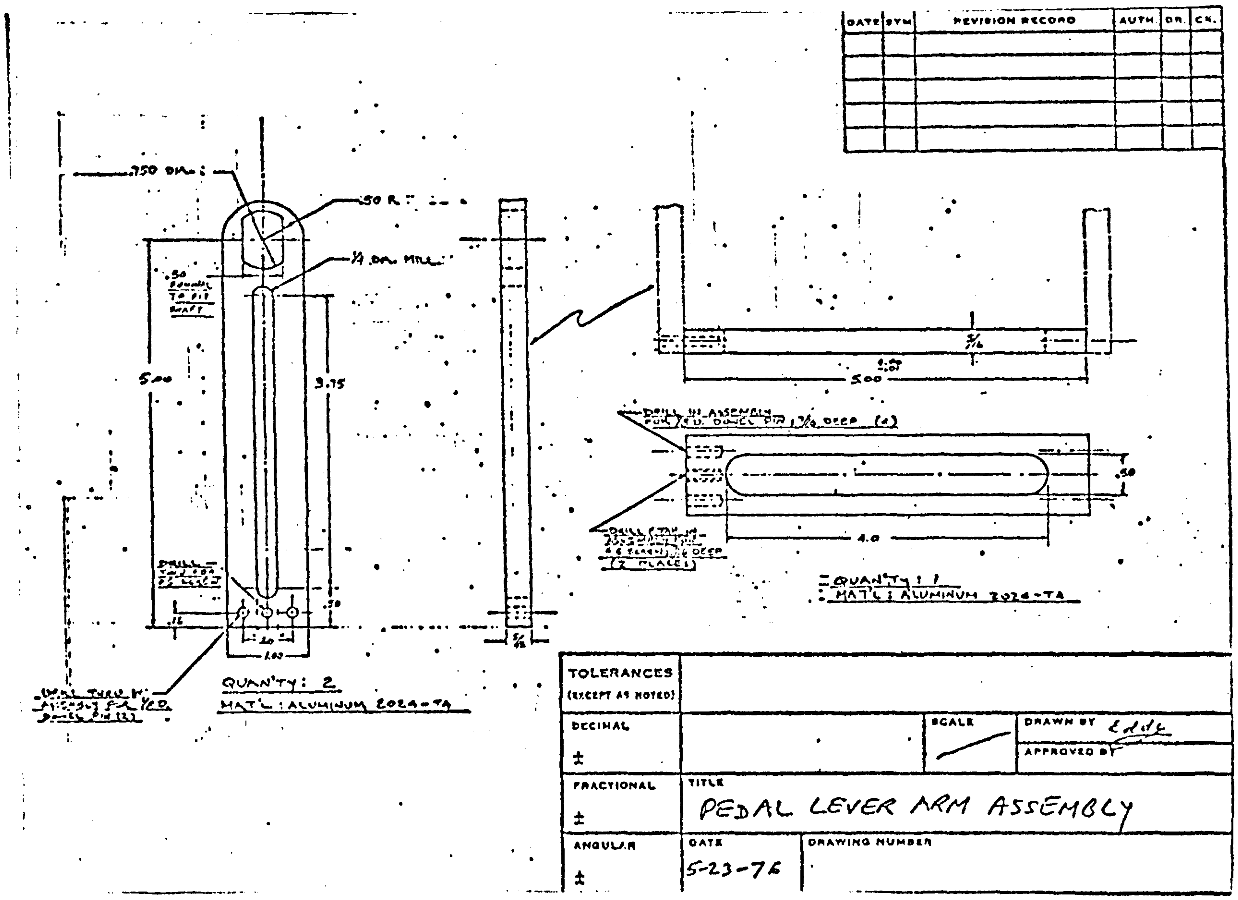
DRAWING NUMBER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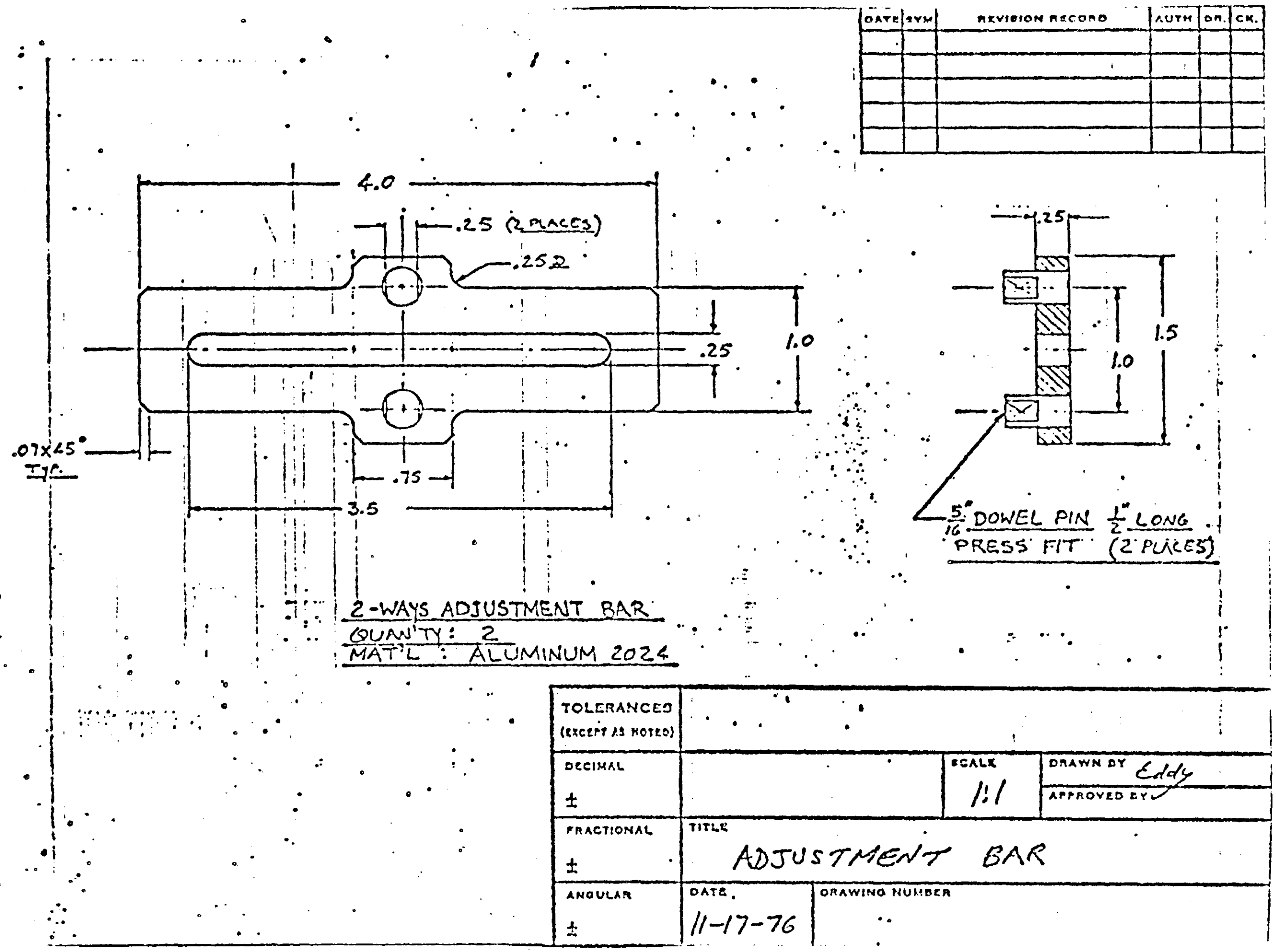
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4.0
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1.0
5/16" DOWEL PIN 1/2" LONG
PRESS FIT (2 PLACES)
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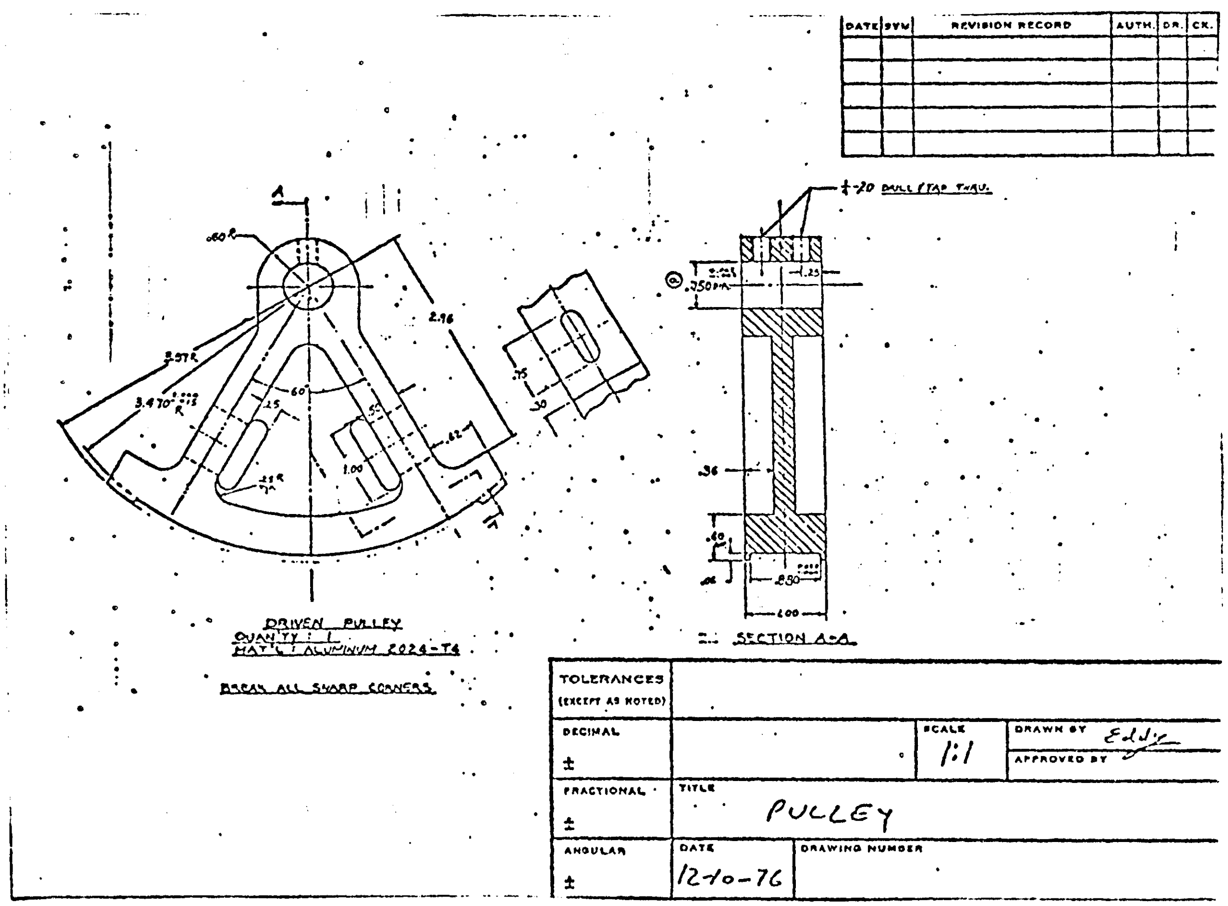
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CK.
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.25 R TYP
.75
.30
1/4-20 DRILL & TAP THRU.
.250 DIA
.25
.36
.40
.06
.250
1.00
SECTION A-A
DRIVEN PULLEY
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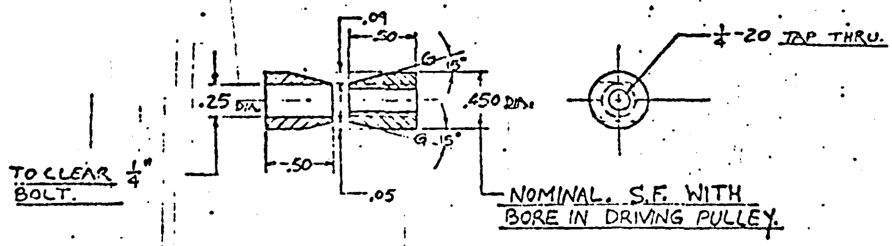

CABLE CLAMP
QUAN'TY: 1 OF EACH
MAT'L: SST 302

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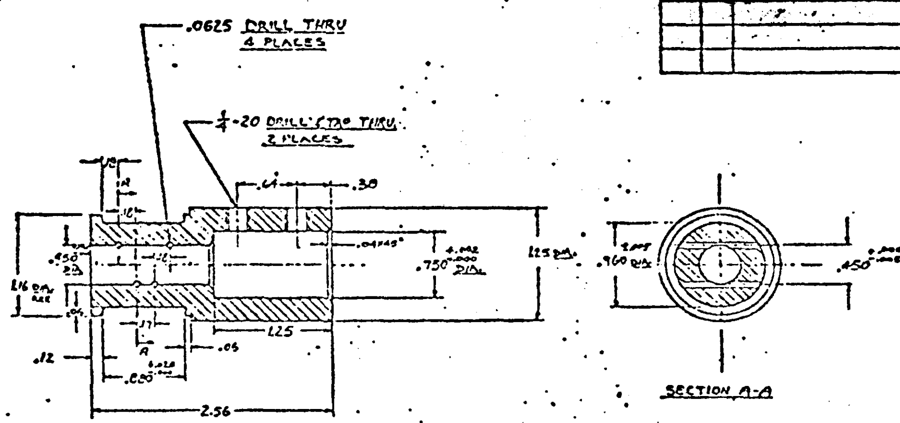

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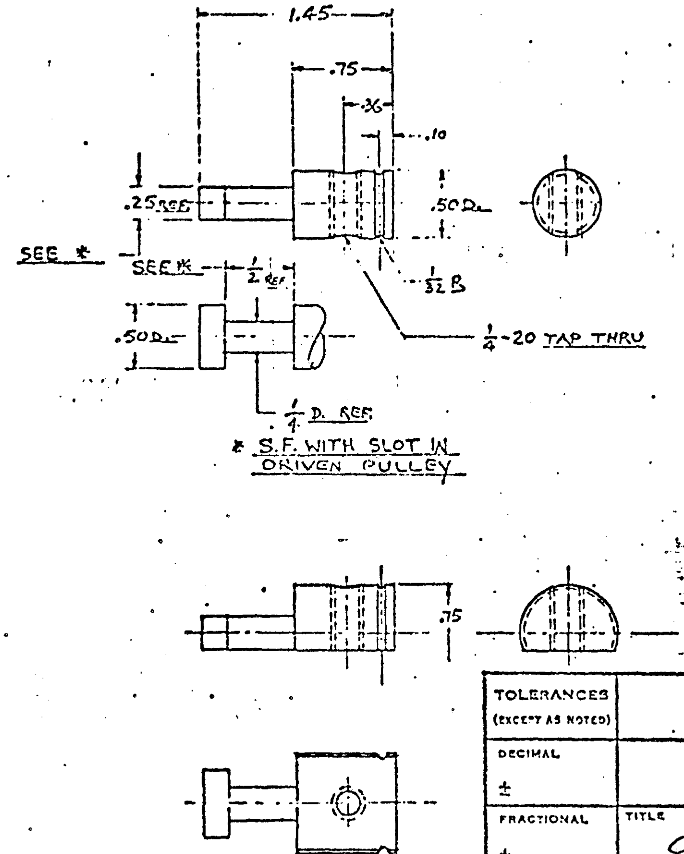

CABLE HOOKS
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MAT'L: STEEL

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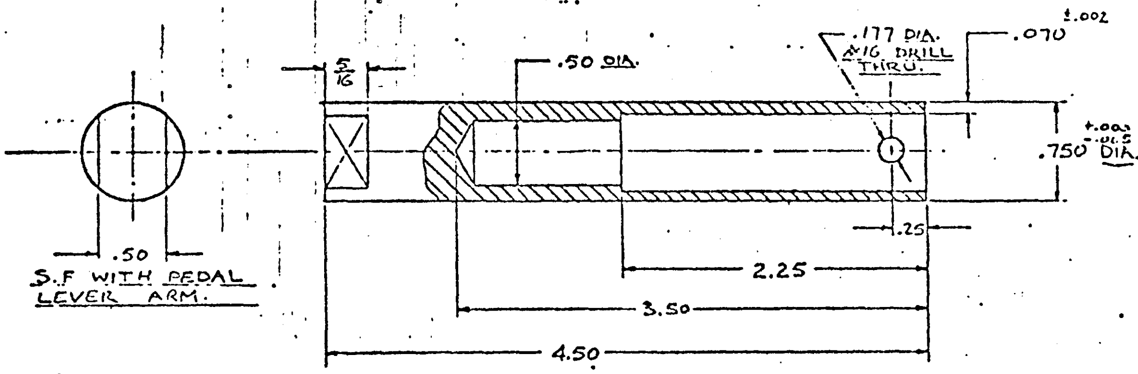

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MAT'L: SST 302

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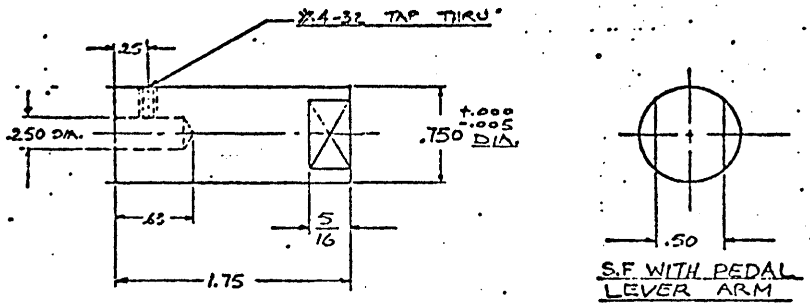

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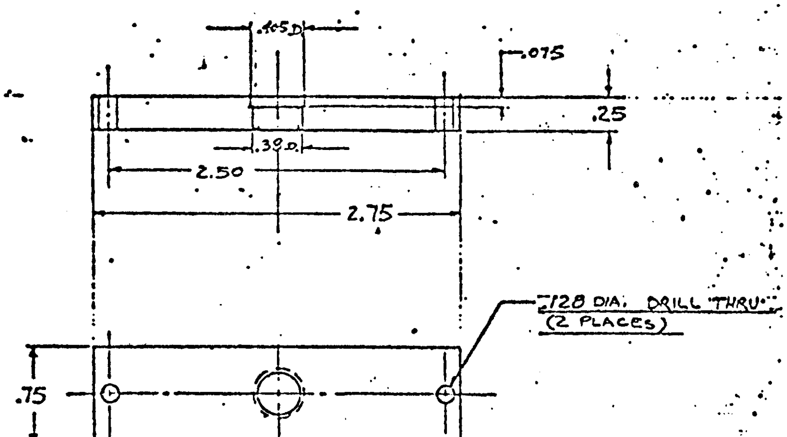

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MAT'L : ALUMINUM

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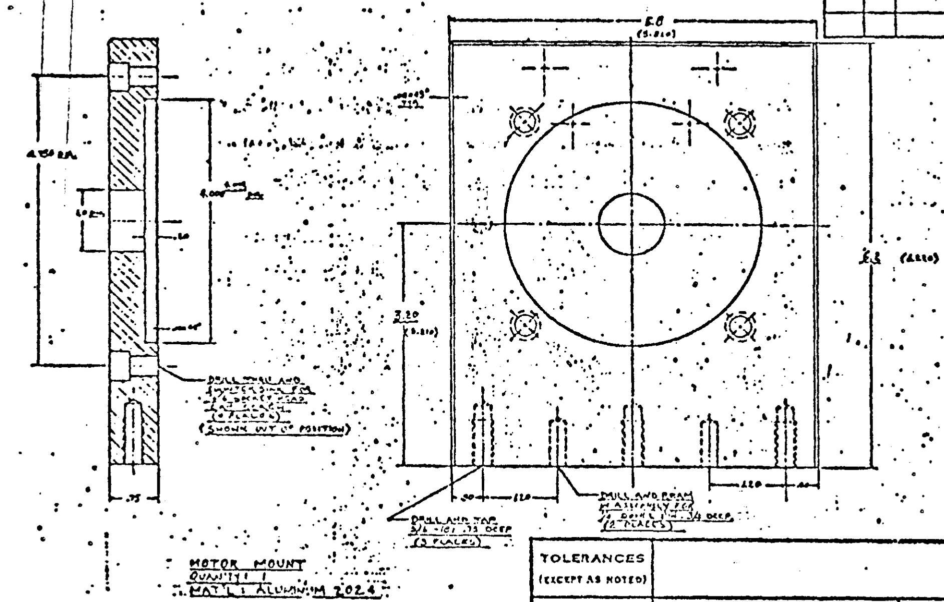

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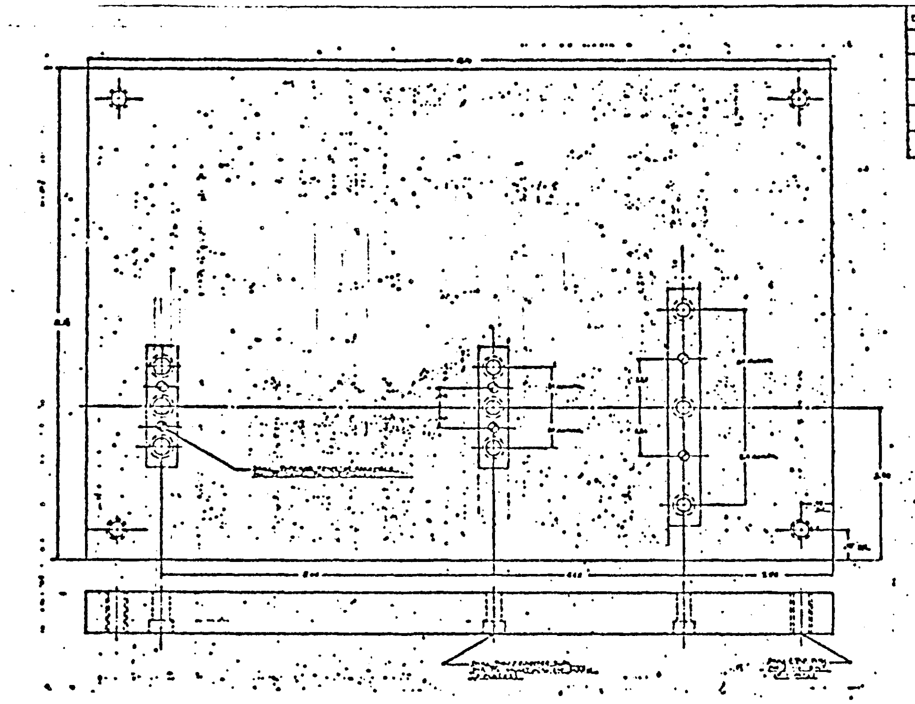

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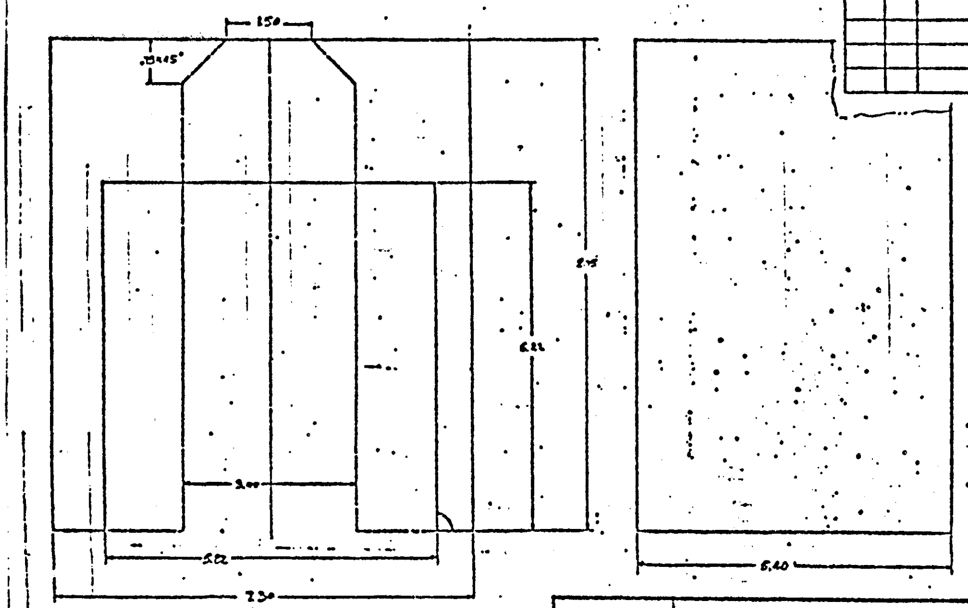

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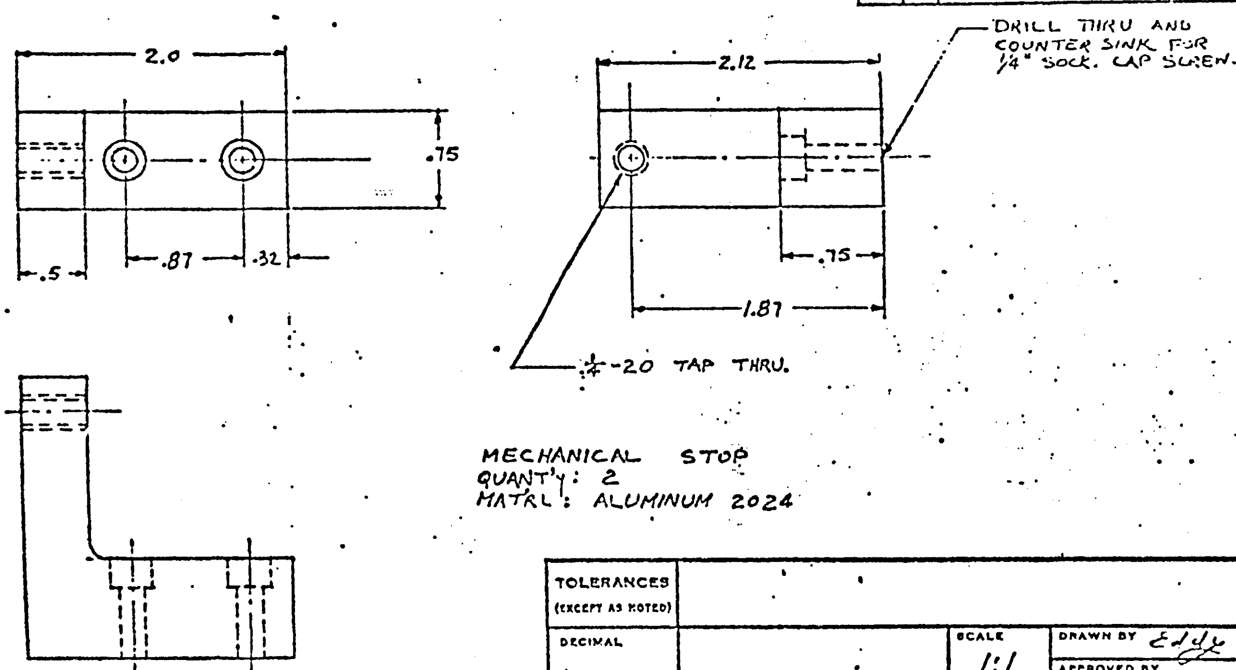

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MATRL: ALUMINUM 2024

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FRACTIONAL ±	TITLE MECHANICAL STOP		
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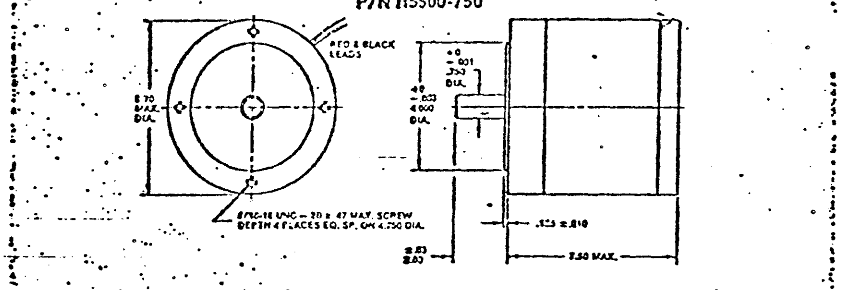
DC SERVO MOTOR

14 LB-FT PEAK TORQUE

P/N H5500-750

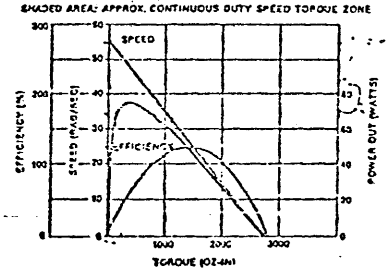


Rotation: Clockwise from shaft end with red lead positive.

Performance Data		
Parameter:	Unit:	Value:
Peak Torque (T_p)	oz-in	2750
Power at T_p* (P_p)	watts	1125
No Load Speed (ω_{NL})	rad/sec	55
Total Breakaway Torque (T_f)	oz-in	16
Ripple Torque (T_r)	% (avg to pk)	3
Ripple Frequency	cycles/rev	25
Temperature Rise	°C/watt	1.0
Max. Allowable Winding Temperature	°C	155
Moment of Inertia (J_m)	oz-in-sec²	0.8
Weight	lb.	25.6
Damping Factor* (F_o)	oz-in/rad/sec	50
Elect. Time Constant* (τ_e)	sec	.006
Mech. Time Constant (τ_m)	sec	.016
Motor Constant* (K_m)	oz-in/$\sqrt{watts}$	83
Max. Theoretical Acceleration (α_m)	rad/sec²	3440
Max. Power Rate (P)	oz-in/sec²	2,500,000

*at 25°C winding temperature

SHADED AREA: APPROX. CONTINUOUS DUTY SPEED TORQUE ZONE



Winding Constants (Values are typical. Other configurations are available.)

Motor

Parameter:	Units:	Tol:	-006	-040	-113
Resistance* (R_m)	ohms	±12.5%	.6	3.7	11.3
Voltage* at T_p (V_p)	volts	ref	25.8	64.5	111
Current at T_p (I_p)	amps	rated	43.7	17.4	9.8
Torque Sensitivity (K_t)	oz-in/amp	±10%	63	160	280
Back E.M.F. (K_b)	volts/rad/sec	±10%	.45	1.13	2.0
Demagnetizing Current	amps	max.	64	22	12.4

APPENDIX B: COMPUTER PROGRAMS

B.1 Program "TWIST" and Subroutines

```
C      PROGRAM TWIST.EDB 4-JUL-77
C
C      USED TO DRIVE THE PEDAL AND OBTAIN DATA FOR THE
C      OBJECTIVE MEASUREMENT OF SPASTICITY, AND OTHER
C      NEUROMUSCULAR SYNDROMES.
C
       PROGRAM TWIST
C
       DIMENSION POSIN(512),ANGLE(512),TORQ(512),TIBEMG(512)
     1  ,SOLEMG(512),TT(512)
C
       INTEGER TESTNU
       REAL LASTV
C
       COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
       COMMON POSIN,ANGLE,TORQ,TIBEMG,SOLEMG
       COMMON PMIN,PMAX,RMAXV,TORQUE,TIBIA,SOLEUS
       COMMON ANGL,MM,K,TT
C
       LASTV=0.
       CALL AOUT(1,LASTV)
       WRITE(5,10)
    10 FORMAT(' 'OBJECTIVE CLINICAL MEASURMENT OF SPASTICITY'',/,
     1  '        REHABILITATION ENGINEERING CENTER         ',/,
     1  '            CEREBRAL PALSY CLINIC                 ',/,
     1  '                CHMC  BOSTON                      ',///,
     1  ' ENTER THE FOLLOWING INFORMATION:',//,
     1  '  PATIENT INITIALS:',$)
       READ(5,11) PINITI
    11 FORMAT(A3)
       WRITE(5,12)
    12 FORMAT('  AGE:',$)
       READ(5,13) IAGE
    13 FORMAT(I3)
C
       CALL LIMITS
C
    20 WRITE(5,21)
    21 FORMAT(' SELECT TYPE OF FORCING FUNCTION:',/,
     1  ' RAMP (CONSTANT VELOCITY) = 1',/,
     1  ' SINUSOIDAL = 2',/,
     1  ' TRIANGULAR = 3')
       READ(5,13) IFUNC
C
       GO TO(100,200,300) IFUNC
C
   100 CALL RAMP
       GO TO 201
C
   200 CALL SINE
       GO TO 201
C
   300 CALL TRIANG
       GO TO 201
   201 TYPE 202
   202 FORMAT(' DO YOU WISH TO CONTINUE THE EXPERIMENTS
     1  (1=YES) ?')
       READ(5,13) ICONT
       IF(ICONT.EQ.1) GO TO 20
C
       WRITE(5,1001)
  1001 FORMAT(///,' END OF TWIST PROGRAM',/)
C
       CALL EXIT
       END
*
```

Subroutine "LIMITS"

```
C      LIMITS.EDB 30-JUN-77
C      PROGRAM TO SET LOWER AND UPPER LIMITS
C      OF SUBJECT'S FOOT. (SAFE ZONE).
C
       SUBROUTINE  LIMITS
C
       DIMENSION POSIN(512),ANGLE(512),TORQ(512)
     1   ,TIBEMG(512),SOLEMG(512)
       INTEGER*2 CH(4)
       INTEGER TESTNU
       REAL LASTV
C
       COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
       COMMON POSIN,ANGLE,TORQ,TIBEMG,SOLEMG
       COMMON PMIN,PMAX,RMAXV,TORQUE,TIBIA,SOLEUS
       COMMON ANGL,MM,K
C
    10 PMIN=0.
       PMAX=0.
       N=0
C
       PAUSE 'SET FOOT AT LOWER LIMIT AND HIT 'RETURN''
C
     5 CALL KWSET(10,'411)
       DO 1,I=1,100
       IF(KWWAIT().NE.0) STOP 'SYNC ERROR'
       CALL ATOD(CH(1))
       ANGLE(I)=CH(1)
       DUMMY=CH(2)
       DUMMY=CH(3)
     1 DUMMY=CH(4)
       CALL KWSET(0,0)
C
       IF(N.EQ.1) GO TO 4
C
       DO 2,I=1,100
     2 PMIN=PMIN+ANGLE(I)
C
       PMIN=((PMIN*.01-2048.)/(2048.*.0235))
C
       N=N+1
C
       PAUSE 'SET FOOT AT UPPER LIMIT AND HIT 'RETURN''
C
       GO TO 5
       CONTINUE
C
     4 DO 3,I=1,100
     3 PMAX=PMAX+ANGLE(I)
C
       PMAX=((PMAX*.01-2048.)/(2048.*.0235))
C
       WRITE(5,6)
     6 FORMAT(' DO YOU WANT TO REPEAT PROCEDURE? (1=YES)')
       READ(5,7) IDO
     7 FORMAT(I1)
       IF(IDO.EQ.1) GO TO 10
       RETURN
       END
```

Subroutine "RAMP"

```
C       PROGRAM RAMP.EDB 4-JUL-77
C
C       USED TO DRIVE THE PEDAL IN A RAMP FUNCTION
C       TO GET CONSTANT VELOCITIES IN DORSIFLEXION OR
C       IN PLANTARFLEXION. SINGLE RAMP MODE.
C
        SUBROUTINE RAMP
C
        DIMENSION POSIN(512),ANGLE(512),TORQ(512),TIBEMG(512)
      1  ,SOLEMG(512)
        INTEGER*2 CH(4)
        INTEGER TESTNU
        REAL LASTV
C
        COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
        COMMON POSIN,ANGLE,TORQ,TIBEMG,SOLEMG
        COMMON PMIN,PMAX,RMAXV,TORQUE,TIBIA,SOLEUS
        COMMON ANGL,MM,K
C
      1 WRITE(5,2)
      2 FORMAT(' ENTER TEST NUMBER: ',$)
        READ(5,6) TESTNU
        IRUN=0
        IF(ISPEED.EQ.1) GO TO 22
C
C       NOW INITIALIZE THE SYSTEM. INPUT ALL PARAMETERS.
C
    222 CALL INPUTS
C
     22 WRITE(5,3)
      3 FORMAT ( ' WHAT WOULD YOU LIKE THE VELOCITY TO BE
      1  (DEG./SEC.) ?')
        READ (5,4) SPEEDD
      4 FORMAT (F7.3)
C
     75 WRITE(5,5)
      5 FORMAT(' SAMPLING FREQUENCY (MSEC.)? (3 MINIMUM)')
        READ(5,6) K
      6 FORMAT(I4)
     76 WRITE(5,77)
     77 FORMAT(' ENTER DURATION OF RUN IN MSEC.',/)
        READ(5,6) KK
C
C       NOW AFTER ALL THE PARAMETERS WERE TYPED IN, THE
C       PRELIMINARY COMPUTATIONS CAN BE PERFORMED.
C
        SLOPE=SPEEDD*.04625          ! SLOPE IN VOLTS/SEC.
C
C        CONVERSION FACTOR IS .04625 VOLTS PER DEGREE.
        DELTAV=ABS(RMAXV-BIASV)
        IF(DELTAV.GT.20*.04625) GO TO 222 ! 20 DEG. MAX.
        DELTAT=K*.001
        DELTAP=DELTAT*SLOPE
        TAUP=DELTAV/SLOPE

        N=(50/K)+1 ! # OF POINTS BEFORE RAMP STARTS
        NN=(TAUP/DELTAT)+N ! # OF POINTS TO TOP OF RAMP
        MM=KK/K
        MMM=MM-NN
        IF(MMM.LT.50) GO TO 76
C
        DO 10,I=1,N
     10 POSIN(I)=BIASV
C
        DO 15,I=NN+1,MM
     15 POSIN(I)=RMAXV
C
        IF(RMAXV-BIASV) 30,222,20
     20 CONTINUE
        DO 25,I=N+1,NN
     25 POSIN(I)=POSIN(I-1)+DELTAP
        GO TO 40
C
     30 CONTINUE
        DO 35,I=N+1,NN
     35 POSIN(I)=POSIN(I-1)-DELTAP
C
     40 PAUSE 'HIT <CR> TO SET THE SYSTEM AT STARTING POINT'
        IF(BIASV.EQ.LASTV) GO TO 45
C
        CALL DAMPER
C
     45 PAUSE 'RELAXED SUBJECT. INITIAL CONDITIONS TO BE MEASURED'
```

```
      C
            CALL IC
      C
         50 IRUN=IRUN+1
      C
            PAUSE 'SYSTEM IS READY TO RUN !!!'
      C
            CALL RUN
      C
            CALL DAMPER
      C
            CALL SHOOT
      C
            CALL FILE
      C
            WRITE(5,60)
         60 FORMAT( ' SHOULD THIS RUN BE REPEATED AGAIN? (1=YES)')
            READ(5,6) IRERUN
            IF(IRERUN.EQ.1) GO TO 45
      C
      C     OBTAIN A SUMMARY OF ALL THE INPUT PARAMETERS.
      C
            BIAS=-BIASV/.04625
            RMAX=-RMAXV/.04625
      C
            WRITE (5,100)
        100 FORMAT ( ' TURN THE TELETYPE ON TO GET A SUMMARY',/,
           1  ' OF ALL THE INPUT PARAMETERS, AND HIT 'RETURN'.')
            PAUSE
            TYPE 110,PINITI,IAGE
        110 FORMAT('  PATIENT: ',A3,'  AGE:',I2,//)
            TYPE 120,TESTNU,IRUN,SPEEDD,BIAS,RMAX,K
        120 FORMAT (' SUMMARY OF THE INPUT PARAMETERS FOR TEST#',I3,
           1  ' RUN#',I2,//,' (RAMP  FUNCTION)',///,
           1  ' VELOCITY :',24X,F7.2,'  DEGREES/SEC.',/,
           1  ' STARTING POINT IS AT :',12X,F7.2,'  DEGREES',/,
           1  ' MAXIMUM TRAVEL IS AT :',12X,F7.2,'  DEGREES',/,
           1  ' SAMPLING DATA EVERY  :',15X,I3,'  MILISECONDS',//,
           1  ' SAMPLING STARTS 50 MSEC. BEFORE RAMP STARTS.',//////)
      C
            WRITE(5,70)
         70 FORMAT(' DIFFERENT VELOCITY? (1=YES)')
            READ(5,6) ISPEED
            IF(ISPEED.EQ.1) GO TO 1
      C
            WRITE(5,80)
         80 FORMAT(' ARE THERE ANY PARAMETERS TO BE CHANGED?
           1  (1=YES)')
            READ(5,6) ICHANG
            IF(ICHANG.EQ.1) GO TO 1
      C
            PAUSE  'END OF RAMP FUNCTION RUNS'
      C
            RETURN
            END
*
```

Subroutine "SINE"

```
C     PROGRAM SINE.EDB 4-JUL-77
C
C     USED TO DRIVE THE PEDAL IN A SINUSOIDAL FUNCTION
C     SINGLE CYCLE MODE FROM A BIAS LEVEL.
C
      SUBROUTINE SINE
C
      DIMENSION POSIN(512),ANGLE(512),TORQ(512),TIBEMG(512)
     1 ,SOLEMG(512)
      INTEGER*2 CH(4)
      INTEGER TESTNU
      REAL LASTV
C
      COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
      COMMON POSIN,ANGLE,TORQ,TIBEMG,SOLEMG
      COMMON PMIN,PMAX,RMAXV,TORQUE,TIBIA,SOLEUS
      COMMON ANGL,MM,K
C
      DATA PI/3.1415926535/
C
    1 WRITE(5,2)
    2 FORMAT(' ENTER TEST NUMBER: ',$)
      READ(5,6) TESTNU
      IRUN=0
      IF(IFREQ.EQ.1) GO TO 22
C
C     NOW INITIALIZE THE SYSTEM. INPUT ALL PARAMETERS.
C
  222 CALL INPUTS
C
   22 WRITE(5,3)
    3 FORMAT ( ' WHAT WOULD YOU LIKE THE FREQUENCY TO BE
     1  (CYCLES/SEC.) ?')
      READ (5,4) FREQCY
    4 FORMAT (F7.3)
C
   75 WRITE(5,5)
    5 FORMAT(' SAMPLING FREQUENCY (MSEC.)? (3 MINIMUM,BUT USE 4)')
      READ(5,6) K
    6 FORMAT(I4)
C
C     NOW AFTER ALL THE PARAMETERS WERE TYPED IN, THE
C     PRELIMINARY COMPUTATIONS CAN BE PERFORMED.
C
C
      DELTAV=ABS(RMAXV-BIASV)
      A=DELTAV/.04625
      DELTAT=K*.001
      TAUP=1./FREQCY
      AV=.5*DELTAV
      OMEGAR=2.*PI*FREQCY
      N=(TAUP/DELTAT)+1  ! # OF POINTS PER CYCLE
      II=50/K
      NN=N+II
   76 WRITE(5,77)
   77 FORMAT(' ENTER DURATION OF RUN IN MSEC.',/)
      READ(5,6) KK
      MM=KK/K
      IF(MM.GE.513) GO TO 222
      MMM=MM-NN
      IF(MMM.LT.50) GO TO 76
C
      DO 7,I=1,II
    7 POSIN(I)=BIASV
C
      DO 8,I=NN+1,MM
    8 POSIN(I)=BIASV
```

```
C
      T=-DELTAT
C
      IF(RMAXV-BIASV) 30,222,10
C
   10 CONTINUE
      DO 20,I=II+1,NN
      T=T+DELTAT
   20 POSIN(I)=BIASV+AV*(SIN(OMEGAR*T-.5*PI)+1.)
      GO TO 40
C
   30 CONTINUE
      DO 35,I=II+1,NN
      T=T+DELTAT
   35 POSIN(I)=BIASV-AV*(SIN(OMEGAR*T-.5*PI)+1.)
C
   40 PAUSE 'HIT <CR> TO SET THE SYSTEM AT STARTING POINT'
      IF(BIASV.EQ.LASTV) GO TO 45
C
      CALL DAMPER
C
   45 PAUSE 'RELAXED SUBJECT. INITIAL CONDITIONS TO BE MEASURED'
C
      CALL IC
C
   50 IRUN=IRUN+1
C
      PAUSE 'SYSTEM IS READY TO RUN !!!'
C
      CALL RUN
C
      CALL SMOOT
C
      CALL FILE
C
      WRITE(5,60)
   60 FORMAT( ' SHOULD THIS RUN BE REPEATED AGAIN? (1=YES)')
      READ(5,6) IRERUN
      IF(IRERUN.EQ.1) GO TO 45
C
C     OBTAIN A SUMMARY OF ALL THE INPUT PARAMETERS.
C
      BIASD=-BIASV/.04625
      RMAX=-RMAXV/.04625
C
      WRITE (5,100)
  100 FORMAT ( ' TURN THE TELETYPE ON TO GET A SUMMARY',/,
     1 ' OF ALL THE INPUT PARAMETERS, AND HIT 'RETURN'.')
      PAUSE
      TYPE 110,PINITI,IAGE
  110 FORMAT('  PATIENT: ',A3,'  AGE:',I2,//)
      TYPE 120,TESTNU,IRUN,FREQCY,BIASD,RMAX,A,K
  120 FORMAT (' SUMMARY OF THE INPUT PARAMETERS FOR TEST#',I3,
     1 ' RUN#',I2,//,' (SINUSOIDAL  FUNCTION)',//,
     1 ' FREQUENCY :',23X,F7.2,'  CYCLES/SEC.',/,
     1 ' STARTING POINT IS AT :',12X,F7.2,'  DEGREES',/,
     1 ' MAXIMUM TRAVEL IS AT :',12X,F7.2,'  DEGREES',/,
     1 ' AMPLITUDE :',23X,F7.2,'  DEGREES',/,
     1 ' SAMPLING DATA EVERY  :',15X,I3,'  MILISECONDS',///)
C
      WRITE(5,70)
   70 FORMAT(' DIFFERENT FREQUENCY? (1=YES)')
      READ(5,6) IFREQ
      IF(IFREQ.EQ.1) GO TO 1
C
      WRITE(5,80)
   80 FORMAT(' ARE THERE ANY PARAMETERS TO BE CHANGED?
     1 (1=YES)')
      READ(5,6) ICHANG
      IF(ICHANG.EQ.1) GO TO 1
C
      PAUSE  'END OF SINUSOIDAL FUNCTION RUNS'
C
      RETURN
      END
*
```

Subroutine "TRIANG"

```
C       PROGRAM TRIANG.EDB 30-JUN-77
C
C       USED TO DRIVE THE PEDAL IN A TRIANGULAR FUNCTION
C       TO GET CONSTANT VELOCITIES IN DORSIFLEXION AND
C       IN PLANTARFLEXION. SINGLE CYCLE MODE.
C
        SUBROUTINE TRIANG
C
        DIMENSION POSIN(512),ANGLE(512),TORQ(512),TIBEMG(512)
      1   ,SOLEMG(512)
        INTEGER*2 CH(4)
        INTEGER TESTNU
        REAL LASTV
C
        COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
        COMMON POSIN,ANGLE,TORQ,TIBEMG,SOLEMG
        COMMON PMIN,PMAX,RMAXV,TORQUE,TIBIA,SOLEUS
        COMMON ANGL,MM,K
C
      1 WRITE(5,2)
      2 FORMAT(' ENTER TEST NUMBER: ',$)
        READ(5,6) TESTNU
        IRUN=0
        IF(ISPEED.EQ.1) GO TO 22
C
C       NOW INITIALIZE THE SYSTEM. INPUT ALL PARAMETERS.
C
    222 CALL INPUTS
C
     22 WRITE(5,3)
      3 FORMAT ( ' WHAT WOULD YOU LIKE THE VELOCITY TO BE
      1   (DEG./SEC.) ?')
        READ (5,4) SPEEDD
      4 FORMAT (F7.3)
C
     75 WRITE(5,5)
      5 FORMAT(' SAMPLING FREQUENCY (MSEC.)? (3 MINIMUM)')
        READ(5,6) K
      6 FORMAT(I4)
     76 WRITE(5,77)
     77 FORMAT(' ENTER DURATION OF RUN IN MSEC.',/)
        READ(5,6) KK
C
C       NOW AFTER ALL THE PARAMETERS WERE TYPED IN, THE
C       PRELIMINARY COMPUTATIONS CAN BE PERFORMED.
C
        SLOPE=SPEEDD*.04625          ! SLOPE IN VOLTS/SEC.
C
C         CONVERSION FACTOR IS .04625 VOLTS PER DEGREE.
        DELTAV=ABS(RMAXV-BIASV)
        DELTAT=K*.001
        DELTAP=DELTAT*SLOPE
        TAUP=DELTAV/SLOPE   ! FOR HALF A CYCLE
        N=(TAUP/DELTAT) ! # OF POINTS PER HALF CYCLE
        II=(50/K)+1 ! 60 MILISECONDS BEFORE RAMP
        NN=N+II
C
        DO 7,I=1,II
      7 POSIN(I)=BIASV
C
        MM=KK/K
        MMM=MM-2*N-II
        IF(MMM.LT.50) GO TO 76
        IF(MM.GE.513) GO TO 222
C
        DO 8,I=(2*N+II+1),MM
      8 POSIN(I)=BIASV
C
        IF(RMAXV-BIASV) 30,222,10
C
     10 CONTINUE
        DO 20,I=II+1,NN
     20 POSIN(I)=POSIN(I-1)+DELTAP
        DO 25,I=1,N
     25 POSIN(NN+I)=POSIN(NN-I)
        GO TO 40
C
     30 CONTINUE
        DO 35,I=II+1,NN
     35 POSIN(I)=POSIN(I-1)-DELTAP
        DO 36,I=1,N
     36 POSIN(NN+I)=POSIN(NN-I)
        GO TO 40
```

```
C
   40 PAUSE 'HIT <CR> TO SET THE SYSTEM AT STARTING POINT'
      IF(BIASV.EQ.LASTV) GO TO 45
C
      CALL DAMPER
C
   45 PAUSE 'RELAXED SUBJECT. INITIAL CONDITIONS.TO BE MEASURED'
C
      CALL IC
C
   50 IRUN=IRUN+1
C
      PAUSE 'SYSTEM IS READY TO RUN !!!'
C
      CALL RUN
C
      CALL SHOOT
C
      CALL FILE
C
      WRITE(5,60)
   60 FORMAT( ' SHOULD THIS RUN BE REPEATED AGAIN? (1=YES)')
      READ(5,6) IRERUN
      IF(IRERUN.EQ.1) GO TO 45
C
C     OBTAIN A SUMMARY OF ALL THE INPUT PARAMETERS.
C
      BIAS=-BIASV/.04625
      RMAX=-RMAXV/.04625
C
      WRITE (5,100)
  100 FORMAT ( ' TURN THE TELETYPE ON TO GET A SUMMARY',/,
     1  ' OF ALL THE INPUT PARAMETERS, AND HIT 'RETURN'.')
      PAUSE
      TYPE 110,PINITI,IAGE
  110 FORMAT('  PATIENT: ',A3,'  AGE:',I2,//)
      TYPE 120,TESTNU,IRUN,SPEEDD,BIAS,RMAX,K
  120 FORMAT (' SUMMARY OF THE INPUT PARAMETERS FOR TEST#',I3,
     1  ' RUN#',I2,//,' (TRIANGULAR  FUNCTION)',//,
     1  ' VELOCITY :',24X,F7.2,'  DEGREES/SEC.',/,
     1  ' STARTING POINT IS AT :',12X,F7.2,'  DEGREES',/,
     1  ' MAXIMUM TRAVEL IS AT :',12X,F7.2,'  DEGREES',/,
     1  ' SAMPLING DATA EVERY  :',15X,I3,'  MILISECONDS',///)
C
      WRITE(5,70)
   70 FORMAT(' DIFFERENT VELOCITY? (1=YES)')
      READ(5,6) ISPEED
      IF(ISPEED.EQ.1) GO TO 1
C
      WRITE(5,80)
   80 FORMAT(' ARE THERE ANY PARAMETERS TO BE CHANGED?
     1  (1=YES)')
      READ(5,6) ICHANG
      IF(ICHANG.EQ.1) GO TO 1
C
      PAUSE  'END OF TRIANGULAR FUNCTION RUNS'
C
      RETURN
      END
*
```

<u>Subroutine "INPUTS"</u>

```
C       PROGRAM TO INPUT PARAMETERS
C       INPUTS.EDB 29-JUN-77
C
        SUBROUTINE  INPUTS
C
        DIMENSION POSIN(512),ANGLE(512),TORQ(512)
     1  ,TIBEMG(512),SOLEMG(512)
        INTEGER TESTNU
        REAL LASTV
C
        COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
        COMMON POSIN,ANGLE,TORQ,TIBEMG,SOLEMG
        COMMON PMIN,PMAX,RMAXV,TORQUE,TIBIA,SOLEUS
        COMMON ANGL,MM,K
C
      5 WRITE(5,10)
     10 FORMAT(' FROM WHAT POSITION SHOULD THE PEDAL START?
     1  (DEGREES)')
        READ(5,20) BIAS
     20 FORMAT(F7.2)
C
        IF(BIAS.GT.PMAX) GO TO 5
        IF(BIAS.LT.PMIN) GO TO 5
C
        BIASV=-BIAS*.04625 ! CONVERSION FROM DEG. TO VOLTS
C
     25 WRITE(5,30)
     30 FORMAT(' TO WHAT POSITION SHOULD THE PEDAL MOVE?
     1  (DEGREES)')
        READ(5,20) RMAX
C
        IF(RMAX.GT.PMAX) GO TO 25
        IF(RMAX.LT.PMIN) GO TO 25
C
        RMAXV=-RMAX*.04625
C
        RETURN
        END
*
```

Subroutine "IC"

```
C       PROGRAM IC.EDB 30-JUN-77
C       USED TO DETERMINE INITIAL CONDITIONS
C       OF SUBJECT'S FOOT AT REST (NO LOAD).
C
        SUBROUTINE  IC
C
        DIMENSION POSIN(512),ANGLE(512),TORQ(512)
      1  ,TIBEMG(512),SOLEMG(512)
        INTEGER*2 CH(4)
        INTEGER TESTNU
        REAL LASTV
C
        COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
        COMMON POSIN,ANGLE,TORQ,TIBEMG,SOLEMG
        COMMON PMIN,PMAX,RMAXV,TORQUE,TIBIA,SOLEUS
        COMMON ANGL,MM,K
C
        ANGL=0.
        TORQUE=0.
        TIBIA=0.
        SOLEUS=0.
C
        CALL KWSET(20,'411)
        DO 1,I=1,100
        IF(KWWAIT().NE.0) STOP 'SYNC ERROR'
        CALL ATOD(CH(1))
        ANGLE(I)=CH(1)
        TORQ(I)=CH(2)
        TIBEMG(I)=CH(3)
      1 SOLEMG(I)=CH(4)
        CALL KWSET(0,0)
C
        DO 2,I=1,100
        ANGL=ANGL+ANGLE(I)
        TORQUE=TORQUE+TORQ(I)
        TIBIA=TIBIA+TIBEMG(I)
      2 SOLEUS=SOLEUS+SOLEMG(I)
C
        BIAS=-BIASV/.04625
C
        ANGL=((ANGL*.01-2048.)/(2048.*.0235))-BIAS ! DEGREES
        TORQUE=(((TORQUE*.01-2048.)/2048.)-.039)*193.75 !IN-LB
        TIBIA=-(TIBIA*.01)/2048. ! VOLTS
        SOLEUS=-(SOLEUS*.01)/2048. ! VOLTS
        RETURN
        END
*
```

Subroutine "RUN"

```
C       PROGRAM RUN.EIB 30-JUN-77
C       USED TO RUN THE SYSTEM AND TAKE DATA
C
        SUBROUTINE RUN
C
        DIMENSION POSIN(512),ANGLE(512),TORQ(512)
     1  ,TIBEMG(512),SOLEMG(512)
        INTEGER*2 CH(4)
        INTEGER TESTNU
        REAL LASTV
C
        COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
        COMMON POSIN,ANGLE,TORQ,TIBEMG,SOLEMG
        COMMON PMIN,PMAX,RMAXV,TORQUE,TIBIA,SOLEUS
        COMMON ANGL,MM,K
C
        MILSEC=2000
C
        CALL DELAY(MILSEC)
C
        CALL KWSET(K,'411)
        DO 1,I=1,MM
        IF(KWWAIT().NE.0) STOP 'SYNC ERROR'
        CALL AOUT(1,POSIN(I)) ! D/A CH 0
        CALL ATOD(CH(1)) ! A/D CH'S 0,1,2,3
        ANGLE(I)=CH(1)
        TORQ(I)=CH(2)
        TIBEMG(I)=CH(3)
      1 SOLEMG(I)=CH(4)
        CALL KWSET(0,0)
C
C
        LASTV=POSIN(MM)
C
        DO 2,I=1,MM
        ANGLE(I)=((ANGLE(I)-2048.)/(2048.*.0235))-ANGL
        TORQ(I)=((((TORQ(I)-2048.)/2048.)-.039)*193.75)-TORQUE
        TIBEMG(I)=-((TIBEMG(I)-2048.)/2048.)
      2 SOLEMG(I)=-((SOLEMG(I)-2048.)/2048.)
C
        RETURN
        END
*
```

Subroutine "DAMPER" and "SMOOT"

```
        SUBROUTINE DAMPER
        INTEGER TESTNU
        REAL LASTV
        COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
        Y=LASTV
        B=BIASV
        IF(Y-B) 9,20,10
      9 CALL KWSET(10,'000411)
     91 Y=Y+.0004625
        IF(Y.GE.(B+.0004625)) GO TO 20
        IF(KWWAIT().NE.0) STOP 'SYNC ERROR'
        CALL AOUT(1,Y)
        GO TO 91
     10 CALL KWSET(10,'000411)
     11 Y=Y-.0004625     !DELTAY=SLOPE(1DEG/SEC)XDELTAT(.01SEC)
        IF(Y.LE.(B-.0004625)) GO TO 20
        IF(KWWAIT().NE.0) STOP 'SYNC ERROR'
        CALL AOUT(1,Y)
        GO TO 11
     20 CALL KWSET(0,0)
        LASTV=Y
        RETURN
        END
*
```

```
C       PROGRAM SMOOT.EDB 11-JUL-77
C       USED TO SMOOTH THE TORQUE DATA
C
        SUBROUTINE SMOOT
C
        DIMENSION POSIN(512),ANGLE(512),TORQ(512),TIBEMG(512)
      1  ,SOLEMG(512),TT(512)
C
        COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
        COMMON POSIN,ANGLE,TORQ,TIBEMG,SOLEMG
        COMMON PMIN,PMAX,RMAXV,TORQUE,TIBIA,SOLEUS
        COMMON ANGL,MM,K,TT
C
        DO 20,J=1,10
        DO 10,I=1,MM
     10 TT(I)=TORQ(I)
        DO 20,I=2,MM-1
     20 TORQ(I)=.25*TT(I-1)+.5*TT(I)+.25*TT(I+1)
        CONTINUE
        RETURN
        END
*
```

Subroutine "FILE"

```
C       PROGRAM FILE.EDB 30-JUN-77
C       USED TO PUT THE DATA ON A FILE
C
        SUBROUTINE  FILE
C
        DIMENSION POSIN(512),ANGLE(512),TORQ(512)
      1  ,TIBEMG(512),SOLEMG(512)
        INTEGER*2 CH(4)
        INTEGER TESTNU
        REAL LASTV
C
        COMMON BIASV,PINITI,IAGE,TESTNU,LASTV
        COMMON POSIN,ANGLE,TORQ,TIBEMG,SOLEMG
        COMMON PMIN,PMAX,RMAXV,TORQUE,TIBIA,SOLEUS
        COMMON ANGL,MM,K
C
        WRITE(5,1)
      1 FORMAT(' DO YOU WANT THIS RUN ON A FILE? (1=YES)')
        READ(5,2) IDO
      2 FORMAT(I3)
C
        IF(IDO.NE.1) GO TO 5
C
        WRITE(5,3)
      3 FORMAT(//,2X,'ENTER FILE NAME :',/)
C
        CALL ASSIGN(1,,-1)
        N=1
        WRITE(1) MM,K,N
        DO 4,I=1,MM
      4 WRITE(1) ANGLE(I),TORQ(I),TIBEMG(I),SOLEMG(I)
        CALL CLOSE (1)
C
      5 RETURN
        END
*
```

Subroutines "KWSET" and "KWWAIT"

```
        .TITLE  KW              ; CLOCK SET INSTEAD OF SETR

LPSKWS=170404

        .GLOBL  KWSET,KWWAIT

KWSET:  INC     %5              ; CALL KWSET(BUFFER,STATUS)
        INC     %5
        MOV     #LPSKWS,%0      ; BUFFER & STATUS ARE INTEGERS
        CLR     (%0)+           ; TO BE LOADED INTO THE L P S
        MOV     @(%5)+,%1       ; REGISTERS
        NEG     %1
        MOV     %1,@%0
        MOV     @(%5)+,-(%0)
        RTS     %7

KWWAIT: MOV     #LPSKWS,%0      ; IF(KWWAIT().NE.0) STOP 'SYNC ERROR'
        TSTB    @%0
        BMI     2$
1$:     TSTB    @%0
        BPL     1$
        BICB    #200,@%0
        CLR     %0
        RTS     %7
2$:     CLR     @%0             ; TURN CLOCK OFF
        MOV     #-1,%0          ; AND RETURN ERROR FLAG
        RTS     %7
        .END
```

Subroutine "A to D"

```
ATOD    ; ANALOG TO DIGITAL CON RT-11 MACRO VM02-12    13-JUN-77 00:03:13
 PAGE 1

                              .TITLE  ATOD     ; ANALOG TO DIGITAL CONVERSION
                                                ; OF FIRST FOUR CHANNELS OF
                                                ; THE LPS (0,1,2,3)

                              .GLOBL  ATOD

          170400              LPSADS=170400    ; A/D CONVERTOR STATUS
          170402              LPSADB=170402    ; A/D CONVERTOR BUFFER

   00000 005725 ATOD:         TST     (%5)+     ; SKIP PAST PARAMETER COUNT
   00002 011500               MOV     (%5),%0   ; GET ADDRESS OF BUFFER
   00004 005037               CLR     @#LPSADS  ; INITIALIZE CONVERTER
         170400

   00010 005237 1$:           INC     @#LPSADS  ; START CONVERSION
         170400

   00014 105737 2$:           TSTB    @#LPSADS  ; WAIT FOR CONVERSION
         170400
   00020 100375               BPL     2$        ; TO FINISH

   00022 105737               TSTB    @#LPSADS+1 ; CHECK FOR CONVERSION
         170401
   00026 100411               BMI     4$        ; ERROR

   00030 013720               MOV     @#LPSADB,(%0)+ ; SAVE DATA IN BUFFER
         170402
```

```
00034 105237 3$:    INCB    @#LPSADS+1 ; POINT TO NEXT CHANNEL
      170401
00040 123727        CMPB    @#LPSADS+1,#4  ; CHECK CHANNEL NUMBER
      170401
      000004
00046 002760        BLT     1$              ; AND CONTINUE THRUGH
                                            ; FIRST FOUR CHANNELS
                                            ; (0,1,2,3)
00050 000207        RTS     %7              ; RETURN


00052 012720 4$:    MOV     #-1,(%0)+ ; FORCE ERRONEOUS DATA TO -1
      177777
00056 042737        BIC     #100000,@#LPSADS ; CLEAR ERROR FLAG
      100000
      170400
00064 000763        BR      3$                 ; CONTINUE IN LOOP


      000001'      .END
```

B.2 Program "CALIBR"

```
      C       PROGRAM CALIBR.EDB 25-JUN-77
      C
      C       THIS PROGRAM ALLOWES CALIBRATION OF THE
      C       TORQUE,AND POSITION.
              DIMENSION ANG(100),TORQ(100),ANGLE(20),TORQUE(20)
      C
              INTEGER*2 CH(4)
           10 WRITE(5,11)
           11 FORMAT(' SAMPLING TIME (MSEC)? [100 SAMPLES/DATA POINT]')
              READ(5,12) K
           12 FORMAT(I2)
              IF(K.LT.3) GO TO 10
              DO 1,I=1,20
              ANGLE(I)=0.
            1 TORQUE(I)=0.
              WRITE(5,2)
            2 FORMAT(' HOW MANY DATA POINTS? (20 MAX.)')
              READ(5,3) N
            3 FORMAT(I2)
              DO 30,J=1,N
              IF(J.GE.2) GO TO 100
              PAUSE 'READY TO TAKE DATA OF UNLOADED SYSTEM'
              GO TO 101
          100 PAUSE 'LOAD THE SYSTEM !!!'
      C
          101 CALL KWSET(K,'411)
              DO 20,I=1,100
              IF(KWWAIT().NE.0) STOP 'SYNC ERROR'
              CALL ATOD(CH(1))
              ANG(I)=CH(1)
              TORQ(I)=CH(2)
              DUMMY=CH(3)
           20 DUMMY=CH(4)
              CALL KWSET(0,0)
      C
              DO 25,I=1,100
              ANGLE(J)=ANGLE(J)+ANG(I)
           25 TORQUE(J)=TORQUE(J)+TORQ(I)
              ANGLE(J)=(ANGLE(J)*.01-2048.)/(2048.*.0235)
              TORQUE(J)=193.75*(((TORQUE(J)*.01-2048.)/2048.)-.039)
      C
           30 CONTINUE
              THETA=ANGLE(1)
              TAU=TORQUE(1)
      C
              DO 35,J=1,N
              ANGLE(J)=ANGLE(J)-THETA
           35 TORQUE(J)=TORQUE(J)-TAU
      C
              PAUSE 'TURN TELETYPE ON TO GET SUMMARY OF DATA'
              WRITE(5,3329) THETA,TAU
         3329 FORMAT(5X,'INITIAL POSITION=',F10.5,/,5X,
            1   'INITIAL TORQUE=',F10.5,///)
              WRITE(5,3330)
```

```
 3330 FORMAT(5X,' #',5X,' ANGLE',5X,' TORQUE'
     1     ,/,12X,'  DEG ',5X,'  IN-LB',//)
      DO 3333,I=1,N
      WRITE(5,3331) I,ANGLE(I),TORQUE(I)
 3331 FORMAT(5X,I2,5X,F7.2,5X,F7.2)
C
 3333 CONTINUE
      WRITE(5,702)
  702 FORMAT(//,2X,'ENTER FILE NAME: ')
      CALL ASSIGN(1,,-1)
      NN=1
      WRITE(1) N,K,NN
      DO 703,I=1,N
  703 WRITE(1) ANGLE(I),TORQUE(I),DUMMY,DUMMY
      CALL CLOSE(1)
      END
*
```

B.3 Program "AVERAG"

```
C      AVERAG.EDN 29-JUN-77
C      AVERAGING PROGRAM OF TWO DATA FILES AT A TIME.
C
       DIMENSION ANGLE(512,2),TORQ(512,2)
      1     ,TIBEMG(512,2),SOLEMG(512,2)
C
     1 WRITE(5,10)
    10 FORMAT('  ENTER FIRST FILE NAME:',/)
       CALL ASSIGN(1,,-1)
       READ (1) MM,K,N
       NI=N
       DO 12,I=1,MM
    12 READ (1) ANGLE(I,1),TORQ(I,1),TIBEMG(I,1),SOLEMG(I,1)
       CALL CLOSE (1)
C
    19 WRITE(5,20)
    20 FORMAT('  ENTER ANOTHER FILE NAME:',/)
       CALL ASSIGN(1,,-1)
       READ (1) MM,K,N
       NII=N
       DO 21,I=1,MM
    21 READ(1) ANGLE(I,2),TORQ(I,2),TIBEMG(I,2),SOLEMG(I,2)
       CALL CLOSE (1)
C
       N=NI+NII
C
       IF(NII-NI) 31,31,32
    31 CONTINUE
C
       DO 30,I=1,MM
       ANGLE(I,1)=((N-1)*ANGLE(I,1)+ANGLE(I,2))/N
       TORQ(I,1)=((N-1)*TORQ(I,1)+TORQ(I,2))/N
       TIBEMG(I,1)=((N-1)*TIBEMG(I,1)+TIBEMG(I,2))/N
    30 SOLEMG(I,1)=((N-1)*SOLEMG(I,1)+SOLEMG(I,2))/N
       CONTINUE
C
       GO TO 33
C
    32 CONTINUE
       DO 33,I=1,MM
       ANGLE(I,1)=(ANGLE(I,1)+(N-1)*ANGLE(I,2))/N
       TORQ(I,1)=(TORQ(I,1)+(N-1)*TORQ(I,2))/N
       TIBEMG(I,1)=(TIBEMG(I,1)+(N-1)*TIBEMG(I,2))/N
       SOLEMG(I,1)=(SOLEMG(I,1)+(N-1)*SOLEMG(I,2))/N
    33 CONTINUE
C
       NI=N
C
       WRITE(5,34)
    34 FORMAT(' ONE MORE FILE TO AVERAGE WITH PREVIOUS FILES?
      1  1=YES')
       READ(5,26) MORE
    26 FORMAT(1I)
       IF(MORE.EQ.1) GO TO 19
```

```
C
      WRITE(5,40)
   40 FORMAT('  ENTER AVERAGED FILE NAME:',/)
      CALL ASSIGN(1,,-1)
      WRITE(1) MM,K,N
      DO 42,I=1,MM
   42 WRITE(1) ANGLE(I,1),TORQ(I,1),TIBEMG(I,1),SOLEMG(I,1)
      CALL CLOSE (1)
C
      PAUSE ' AVERAGING DONE !'
C
      WRITE(5,50)
   50 FORMAT(' DO YOU WANT TO AVERAGE OTHER FILES ? 1=YES')
      READ(5,26) IWANT
      IF(IWANT.EQ.1) GO TO 1
C
      END
*
```

B.4 Program "PLOT"

```
C     PLOTTING PROGRAM PLOT.EDB 19-JUL-77
C
      PROGRAM PLOT
C
      DIMENSION ANGLE(512),TORQ(512),TIBEMG(512)
     1  ,SOLEMG(512),X(2,512),XMAX(2),XMIN(2),
     1  XZ(2),VZ(2),TICK(2),XS(2,512)
C
      INTEGER FLAG(2)
C
    1 WRITE(5,5)
    5 FORMAT(//,2X,'ENTER DATA FILE NAME:',/)
      CALL ASSIGN(1,,-1,)
      READ (1) MM,K,NN
      N=MM
      DO 10,I=1,MM
   10 READ(1) ANGLE(I),TORQ(I),TIBEMG(I),SOLEMG(I)
      CALL CLOSE (1)
C
      WRITE(5,11)
   11 FORMAT(' DO YOU WANT A PRINT OUT OF THE DATA? (Y)')
      READ(5,12) IDO
   12 FORMAT(A1)
      IF(IDO.NE.FLAG(1)) GO TO 31
      WRITE(5,20)
   20 FORMAT(6X,' TIME',5X' ANGLE',5X,'TORQUE',7X,'TIBEMG',5X,
     1     'SOLEMG',/,6X,'(MSEC)',5X,'(DEG)',5X,'(IN-LB)',//,)
      DO 29,I=1,MM
      ITIME=(I-1)*K
      WRITE(5,30) ITIME,ANGLE(I),TORQ(I),TIBEMG(I),SOLEMG(I)
   30 FORMAT(5X,I5,5X,F7.2,5X,F7.2,5X,F7.2,5X,F7.2)
C
   29 CONTINUE
   31 WRITE(5,40)
   40 FORMAT(' WHAT WOULD YOU LIKE TO PLOT?',//,
     1     ' POSITION VS. TIME = 1',/,
     1     ' TORQUE VS. TIME = 2',/,
     1     ' TIBEMG VS. TIME = 3',/,
     1     ' SOLEMG VS. TIME = 4',/,
     1     ' TORQUE VS. POSITION = 5',/,
     1   ' SPEED VS. TIME = 6',/,
     1     ' NONE OF THE ABOVE = 7',//)
      READ(5,41) IPLOT
   41 FORMAT(I1)
      IF(IPLOT.EQ.7) GO TO 7000
C
      GO TO (1000,2000,3000,4000,5000,6000) IPLOT
C
 1000 CONTINUE
      DO 1001,I=1,MM
 1001 X(1,I)=ANGLE(I)
      DO 1004,J=1,10
      DO 1002,I=2,MM-1
 1002 X(2,I)=.25*X(1,I)+.5*X(1,I)+.25*X(1,I+1)
```

```
      DO 1003,I=2,MM-1
 1003 X(1,I)=X(2,I)
 1004 CONTINUE
      X(2,1)=ANGLE(1)
      X(2,MM)=ANGLE(MM)
      GO TO 6006
C
 2000 CONTINUE
      WRITE(5,2014)
 2014 FORMAT(' WANT TO CONVERT TORQUE TO N-M ? Y',/)
      READ(5,12) ICON
      IF(ICON.NE.FLAG(1)) GO TO 2016
      DO 2015,I=1,MM
 2015 TORQ(I)=TORQ(I)*.113076
C
 2016 CONTINUE
      IF(IPLOT.EQ.5) GO TO 5001
      DO 2001,I=1,MM
      X(1,I)=(I-1)*K*1.
 2001 X(2,I)=TORQ(I)
      CONTINUE
      GO TO 50.
C
 3000 CONTINUE
      DO 3001,I=1,MM
      X(1,I)=(I-1)*K*1.
 3001 X(2,I)=-TIBEMG(I)
      CONTINUE
      GO TO 50
C
 4000 CONTINUE
      DO 4001,I=1,MM
      X(1,I)=(I-1)*K*1.
 4001 X(2,I)=-SOLEMG(I)
      CONTINUE
      GO TO 50
C
 5000 CONTINUE
      GO TO 2000
 5001 DO 5002,I=1,MM
      X(1,I)=ANGLE(I)
 5002 X(2,I)=TORQ(I)
      CONTINUE
      GO TO 50
C
 6000 CONTINUE
      DELTAT=K*.001
C
      DO 6001,I=2,MM-1
      TEMP=ANGLE(I)-ANGLE(I-1)
 6001 X(2,I)=TEMP/DELTAT
      X(2,1)=0.
      X(2,MM)=0.
C
      DO 6006,J=1,10 ! 10 PASSES THRU DIGITAL FILTER.
      DO 6003,I=2,MM-1
 6003 X(1,I)=.25*X(2,I-1)+.5*X(2,I)+.25*X(2,I+1)
      DO 6004,I=2,MM-1
 6004 X(2,I)=X(1,I)
 6006 CONTINUE
      DO 50,I=1,MM
      X(1,I)=(I-1)*K*1.
   50 CONTINUE
      WRITE(5,55)
   55 FORMAT(' WANT TO CHANGE SCALE? (Y)',/)
      READ(5,12) ICHANG
      IF(ICHANG.NE.FLAG(1)) GO TO 511
C
      CALL MAXMIN(N,X,XMAX,XMIN)
C
      WRITE(5,500) XMAX(1),XMIN(1),XMAX(2),XMIN(2)
  500 FORMAT(' XMAX=',F7.2,/,' XMIN=',F7.2,/,
     1 ' YMAX=',F7.2,/,' YMIN=',F7.2,//,
     1 ' INPUT DESIRED MAX-MIN VALUES FOR X & Y (F7.2)')
      READ(5,501) XMAX(1),XMIN(1),XMAX(2),XMIN(2)
  501 FORMAT(F7.2)
C
  511 WRITE(5,555)
  555 FORMAT(' WANT TO CHANGE ORIGIN? (Y)',/)
      READ(5,12) IORIGN
      IF(IORIGN.NE.FLAG(1)) GO TO 503
C
      WRITE(5,502)
  502 FORMAT(' INPUT X-Y COORDINATES OF DESIRED ORIGIN
     1 (F7.2)')
```

```
      READ(5,501) XZ(1),XZ(2)
C
      CALL SCALER(1,XZ,XMAX,XMIN,VZ)
C
      CALL ZERO(VZ)
C
  503 WRITE(5,504)
  504 FORMAT(' WANT AXES DRAWN? (Y)',/)
      READ(5,12) IAXES
      IF(IAXES.NE.FLAG(1)) GO TO 53
C
      WRITE(5,51)
   51 FORMAT(' ENTER TICK SIZE (F10.3) X-AXIS <CR> Y-AXIS')
      READ(5,52) TICK
   52 FORMAT(F10.3)
C
      CALL AXIS(XZ,XMAX,XMIN,TICK)
C
   53 CALL SCALER(N,X,XMAX,XMIN,XS)
C
      CALL LINE(N,XS)
C

      GO TO 31
C
 7000 CONTINUE
      WRITE(5,600)
  600 FORMAT(' WANT TO ENTER A NEW FILE? (Y)',/)
      READ(5,12) NEWFIL
      IF(NEWFIL.EQ.FLAG(1)) GO TO 1
C
      DATA FLAG/'Y','N'/
      DATA XZ/0.,0./
C
      END
```

APPENDIX C

RELATIVE EQUIVALENT DAMPING

Herein is an example of how the first term of Equation (4.2) becomes zero when the movement is purely periodic.

The energy balance equation (4.2) is rewritten

$$J \int_0^{\tau} \ddot{\theta}\dot{\theta}\, dt + b \int_0^{\tau} \dot{\theta}\dot{\theta}\, dt + k \int_0^{\tau} \theta\dot{\theta}\, dt + mg\ell \int_0^{\tau} \dot{\theta} \sin\theta\, dt$$

$$= \int_0^{\tau} T\dot{\theta}\, dt \qquad \text{(C.1)}$$

The displacement applied to the foot is purely periodic and of the form

$$\theta = A[\sin(\omega t - \tfrac{\pi}{2}) + 1] \qquad \text{(C.2)}$$

where

A is the amplitude

ω is the angular velocity

t is time

Differentiating equation (C.2) twice gives

$$\dot{\theta} = A \cos(\omega t - \tfrac{\pi}{2}) \qquad \text{(C.3)}$$

$$\ddot{\theta} = -A\omega^2 \sin(\omega t - \tfrac{\pi}{2}) \qquad \text{(C.4)}$$

also

$$\sin(\omega t - \tfrac{\pi}{2})\cos(\omega t - \tfrac{\pi}{2}) = \tfrac{1}{2}\sin(2\omega t - \pi) \qquad \text{(C.5)}$$

and:

$$\omega = 2\pi f \tag{C.6}$$

$$f = 1/\tau \tag{C.7}$$

From (D.6) and (D.7) we obtain

$$\omega = \frac{2\pi}{\tau} \tag{C.8}$$

Thus:

$$J \int_0^{\tau} \ddot{\theta}\dot{\theta}\ dt = J \int_0^{\tau} [-A\omega^2 \sin(\omega t - \frac{\pi}{2})][A\omega\cos(\omega t - \frac{\pi}{2})]dt \tag{C.9}$$

From (D.5)

$$= -JA^2\omega^3 \int_0^{\tau} \frac{1}{2} \sin(2\omega t - \pi)dt = \frac{1}{4}JA^2\omega^2 [\cos(2\omega t - \pi)]_0^{\tau}$$

$$= \frac{1}{4}JA^2\omega^2 [\cos 2\omega t - \pi) - \cos(-\pi)]$$

$$= \frac{1}{4}JA^2\omega^2 [\cos(2\pi - \pi) - \cos(-\pi)] \qquad \text{from C.8}$$

$$= \frac{1}{4}JA^2\omega^2 [\cos(\pi) - \cos(-\pi)] = \underline{\underline{0}}$$

In a similar manner, the other two terms of stiffness and gravity also become zero.

```
C     EDRED.EDB 13-JUL-77
C     A PROGRAM TO COMPUTE THE EQUIVALENT
C     DAMPING AND RELATIV E.D. OF THE ANKLE JOINT.
C
      DIMENSION ANGLE(512),TORQ(512),OMEGA(512),W(512)
C
   44 WRITE(5,45)
   45 FORMAT(' ENTER PERSON'S WEIGHT [LB],LEG LENGTH AND CIRCUM.
     1  [IN] (F6.3)',/)
      READ(5,50) WEIGHT,D,P
   50 FORMAT(F6.3)
C
    1 WRITE(5,10)
   10 FORMAT('  ENTER FILE NAME:',/)
      CALL ASSIGN (1,,-1)
      READ (1) MM,K,N
      DO 20,I=1,MM
   20 READ(1) ANGLE(I),TORQ(I),DUMMY,DUMMY
      CALL CLOSE (1)
C
      WRITE(5,30)
   30 FORMAT(' DO YOU WANT TO COMPUTE E.D. AND R.E.D? 1=Y')
      READ(5,35) IWANT
   35 FORMAT(1I)
      IF(IWANT.NE.1) GO TO 100
C
      DELTAT=K*.001
      OMEGA(1)=0.
      OMEGA(MM)=0.
      DO 22,I=2,MM-1
      TEMP=ANGLE(I)-ANGLE(I-1)
   22 W(I)=TEMP/DELTAT
      W(1)=0.
      W(MM)=0.
      DO 24,J=1,10
      DO 23,I=2,MM-1
      OMEGA(I)=.25*W(I-1)+.5*W(I)+.25*W(I+1)
   23 OMEGA(I)=OMEGA(I)*.017453 ! RADIANS
   24 CONTINUE
C
      SUM1=0.
      SUM2=0.
C
      DO 40,I=1,MM
      TORQ(I)=TORQ(I)*.113076 ! N-M
      TEMP=TORQ(I)*OMEGA(I)
      SUM1=SUM1+TEMP
      TEMP=OMEGA(I)*OMEGA(I)
   40 SUM2=SUM2+TEMP
C
      ED=SUM1/SUM2 ! EQUIVALENT DAMPING.
C
      PI=3.14159265
      REDBW=ED/(WEIGHT*4.4518) ! N
      REDLV=(ED*4*PI)/(P*P*D*1.6387E-5)
C
      WRITE(5,60) ED,REDBW,REDLV
   60 FORMAT(' THE EQUIVALENT DAMPING IS:',F10.7,' N-M-SEC',//,
     1  ' THE RELATIVE (BW) E.D. IS:',E12.3,' M-SEC',//,
     1  ' THE RELATIVE (LV) E.D. IS:',E12.3,' N-SEC/M',///)
C
  100 WRITE(5,101)
  101 FORMAT(' WANT ANOTHER FILE? (SAME SUBJECT) 1=Y',/)
      READ(5,35) IYES
      IF(IYES.EQ.1) GO TO 1
      WRITE(5,102)
  102 FORMAT(' WANT ANOTHER SUBJECT? 1=Y',/)
      READ(5,35) IWANT
      IF(IWANT.EQ.1) GO TO 44
      STOP
      END
*
```

APPENDIX D

START UP AND SHUT OFF PROTOCOL

1. Check strain gauge battery for proper charge (6.34V).
2. Make sure all lines are connected:
   2.1 Position command comes from D/A channel 1
   2.2 Angular position goes into A/D channel 0
   2.3 Torque goes into A/D channel 1
   2.4 Tibialis anterior EMG (electrode #1) goes into A/D channel 2
   2.5 Soleus EMG (electrode #2) goes into A/D channel 3
   2.6 Feedback to power supply
   2.7 Safety cutoff
   2.8 $\pm$ 15 V supply line
   2.9 Motor power line
   2.10 EMG electrodes to processor
3. Turn computer on and run "TWIST" program
4. Turn main power unit "on"
5. Turn strain gauge switch "on".
6. Turn motor power switch "on".
7. Set, manually, pedal at 0° position
8. Press "reset" button
9. Have subject seated in chair and raise it to the proper level.
10. Take subject's shoe off and insert his heel into the heel cup and through the straps
11. Adjust positioning and alignment of the subject. Make sure heel makes light contact with pedal. Align talocrural joint axis with pedal's axis. This is done via the nut on the side of the pedal's frame and the two wing nuts under the pedal.
12. Clean electrode sites with alcohol. Attach an adhesive collar to the reference electrode ad fill it with electrode paste. Apply a thin coat of paste to the electrodes' plates and place them on with tape.
13. Turn all switches on the EMG processor to "on" position.
14. Tighten peal straps
15. Turn "motor power" switch "off".
16. Set upper and lower limits
    16.1 Adjustment of the knobs: Set foot at limit. Turn knob until "off limit" light comes on, then turn knob back slightly until light goes off. The limit is set.
    16.2 Set same limits on computer by pressing "return" on terminal keyboard.
17. Return pedal to 0° position, turn "motor power" switch back to "on" and press "reset" button. The pedal should stay at that position.
18. System is ready to follow commands via the computer terminal. Just follow the instructions and answer the questions that appear on the CRT. Remember to input decimal numbers and integers properly!

To shut system off:

1. Turn "motor power" switch to "off".
2. Turn "strain gauge" switch to "off".
3. Turn power unit switch "off".
4. Turn all switches of the EMG processor to "off" position.
5. Unstrap subject and remove his foot from the pedal.
6. Remove EMG electrodes and clean off electrode paste.
7. Lower the chair.

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