QUADRICEPS FORCES DURING VOLITIONAL AND ELECTROSTIMULATED KNEE EXTENSIONS

MONK

QUADRICEPS FORCES DURING VOLITIONAL AND ELECTROSTIMULATED KNEE EXTENSIONS

By

STEPHEN W. MONK, B.Eng.

A Thesis
Submitted to the School of Graduate Studies
in Partial Fulfilment of the Requirements
for the Degree
Master of Engineering
McMaster University
October 1976

Master of Engineering (1976)
(Engineering Physics)

McMaster University
Hamilton, Ontario, Canada

TITLE : Quadriceps Forces During Volitional and Electrostimulated Knee Extensions

AUTHOR : Stephen William Monk, B.Eng.
McMaster University, Hamilton, Ontario, Canada (1974)

SUPERVISOR : Professor M. Milner

NUMBER OF PAGES : xv, 136

TO MY WIFE, BRENDA JEAN

ACKNOWLEDGEMENTS

I wish to thank the following persons for their guidance and support during the experimental phase of this endeavour:

Ted Iler

Tom Mallis

Steve Naumann

Tony Wallace

Dr. Morris Milner is especially thanked for his expert guidance and continual support throughout the entire phase of this research.

Special thanks are extended to Mrs. Sarah Fick for her valuable time in the typing of this thesis.

Extensive use of the resources of the Biomedical Engineering Department, Chedoke Hospitals, served to facilitate this work.

PREFACE

Muscular function is one of the major areas of research involving human locomotion. During the past few decades, an emphasis has been placed on quantitative analysis of such functions, specifically pertaining to parameters such as force, EMG, muscle length ratios, etc.

This research centres on the forces developed by the Quadriceps Muscle Group during knee extension against gravity. Electrogoniometric data was acquired during volitional extension and extension resulting from electrostimulation of the Quadriceps. Subsequent computer analysis on these data provided instantaneous force values at specific extension angles from the vertical, from which graphical representations were produced.

Correlation studies were performed between the volitional and stimulated extension forces, to determine whether normal volitional extension forces could be duplicated by electrostimulation of the Quadriceps Muscle Group. A remarkably close correlation was discovered between two such forces, and is discussed in detail in the text.

Apart from obtaining quantitative data on extension forces and correlation studies, a device evolved from this research which now provides the researcher with immediate values for the mass and the location of the centre of mass of either an arm or a leg. These measurements can readily be made in under fifteen minutes and are accurate to approximately 4%.

In the closing remarks, clinical applications of the author's technique as an evaluation tool are discussed. Such fields which could realize a benefit include physical therapy and pre- and post-operative studies involving knee-joint replacements.

TABLE OF CONTENTS

		Page
APPENDIX 2 -	ELECTROGONIOMETER	128
	Electrogoniometer	129
APPENDIX 3 -	COMPUTER PROGRAM	130
	Program Listing	131
REFERENCES		134
BIBLIOGRAPHY		136

LIST OF ILLUSTRATIONS

Figure		Page

LIST OF TABLES

CHAPTER 1

INTRODUCTION

1.1 Introduction

The investigative nature of this research focused on the forces developed by the Quadriceps muscle group of the anterior thigh, during extension of the knee. These forces were calculated during volitional extension subject to metronome frequencies of 1.0 Hz and 1.5 Hz, and electrically stimulated extension subject to duty cycles of *.667sec/.667sec, 1.0/1.0, and 1.5/1.0, all having a pulse train frequency of 60 Hz. Force studies were also performed subject to different pulse train frequencies of 50, 60, 70 and 80 Hz, all utilizing a duty cycle of 1.0 sec/1.0 sec.

A brief summary of the results will be presented in section 1.7, with the detailed observations and a full discussion to be found in chapter 6 of the text.

1.2 Background Physiology of the Knee Joint and Extensors

This chapter, as the title implies, is an overview of the knee joint as a whole and the muscles affecting it during extension.

* These figures describe on and off times of stimulation

The rotational axis of the knee joint for purposes of this research is defined, as is the actual motion of the instantaneous centre of rotation during extension as described by Otto Fischer.

The function of the patella during extension is discussed and an analytical expression for the increase in force required in the absence of the patella is presented in figure 2.3. Extensor muscles are discussed separately in detail and a comparison of uni-joint and bi-joint muscles is presented, since the rectus femoris muscle is a bi-joint muscle extending over the hip-joint and knee-joint.

To illustrate the actions of the various muscles affecting the knee, a brief discussion of levers appears together with the muscle grouped according to their leverage classification.

Motor units, the fundamental entity of communication between the central nervous system and a muscle, are defined and discussed, and the location of the various motor points of the knee extensors and flexors, are presented in figure 2.10

1.3 Electrostimulation

This chapter opens with a review of the physiological principles involved in contractions of skeletal muscle.

Voluntary contractions elicited by action potentials from the central nervous system are discussed and are compared to contractions elicited via external electrostimulation.

Two techniques of superficial electrostimulation are compared, unipolar and bipolar, and the disadvantages of bipolar stimulation relative to unipolar stimulation are discussed. The rationale behind the selection of unipolar stimulation for this research is given, together with diagrams illustrating the principles of both types of stimulation.

The parameters of the stimulating current pulses which elicited strong contractions of the vastus lateralis muscle of the quadriceps muscle group are listed. These parameters were measured in a pilot experiment in order to obtain a point of origin for the force studies. Once these parameters were determined, individual parameters could be varied and any change in the force-angle curve noted.

Chapter 2 concludes with a description of the stimulator utilized in these efforts. Adjustments on the front panel are explained with accompanying diagrams, to serve as a condensed reference.

1.4 Apparatus

Chapter 4 deals with the devices and apparatus necessary for the collection of data relevant to the force studies.

Two parameters which are necessary in any rotational motion study are the mass and location of the centre of mass, relative to the rotational axis, of the object under scrutiny. The device used to obtain these two parameters, which was conceived, designed and constructed by the author, is discussed in the opening pages of this chapter. The discussion centers on a general description of the device, pertinent mathematical equations, the associated electronic circuitry, calibration and principles of operation. All relevant technical drawings of this device may be found in Appendix 1.

The two types of electrodes used during electrostimulation, the hand-held probe and the stainless steel mesh, are described in detail as to dimensions and construction. One method of application to the skin is described, which is the method used in this study.

The chapter concludes with a discussion of the electrogoniometer constructed by the author and a brief description of the Honeywell Visicorder, from which all permanent records of the angular motion of the lower limb during experiments were obtained.

The electrogoniometer is discussed with respect to accompanying circuitry and calibration, with all calibration data and graphs being presented. Technical drawings of the electrogoniometer may be found in Appendix 2.

1.5 Experimental Procedures

Chapter five begins with a discussion of the experimental objectives together with the fundamental equation of torque, which is comprised of two components, an inertial component, proportional to the angular acceleration of the lower limb and a gravitational component which is proportional to the sine of the angle of the lower limb measured from the vertical. Three physiologic assumptions pertaining to the knee and lower leg are listed and were adopted in order to simplify the force studies and the collection of data during the experiments.

Experimental procedures during volitional extension and the ensuing raw data (deflections on the visicorder printout) appear next, together with a discussion of the computer program developed to obtain the eventual muscle forces involved. A listing of this computer program can be found in Appendix 3.

The electrostimulation experiments follow,accompanied by the respective raw data, as in the volitional phase. In these experiments, the duty cycle (on/off cycle) of the stimulus current was one parameter which was varied. The four cycles examined were .667/.667, 1.0/1.0, 1.5/1.0, 2.0/1.0, each value listed in seconds. The other parameter varied in a separate set of experiments was the pulse train frequency of the stimulus current. This was varied from 50 Hz to 80 Hz inclusive, at 10 Hz intervals.

This chapter ends with the measurements taken on the lower leg, pertaining to the mass and location of the centre of mass. The measured values of the lower leg mass and location of its mass centre are accurate to within 4.45% and 4% respectively.

1.6 Experimental Observations

Chapter six opens with a discussion of the electrogoniometer which was constructed for this study. Its performance during the experiments is evaluated with respect to its features. i.e., hinges, and elastic straps, in that the goniometer does not interfere with knee extension.

During the experiments, a number of interesting observations could be made. These include qualitative muscle and leg response during determination of the motor point combinations for optimal knee extension, qualitative muscle response in determining the optimal pulse width of the stimulating current, and a qualitative comparison of gross muscle contraction during electrostimulated and volitional knee extension.

A necessary parameter in calculating muscle forces during knee extension, is the moment arm of the patellar tendon about the instantaneous centre of rotation of the knee. An excellent paper is presented in the Journal of Biomechanics by G.L. Smidt. His technique for obtaining this data is briefly summarized and relevant data from his studies is listed. Since his data is given at 15^{o} intervals of knee extension, a defining equation needed to be determined, since knee angle data in the author's study was not consistent with Smidt. The coefficients of the resulting second order equation are listed together with a plot of the patellar tendon moment arm vs knee angle.

The latter part of chapter six consists of in-depth discussions of muscle forces during all experiments. Graphs of knee extension angle vs time, inertial and gravitational components of force, and total force plotted against knee extension angle are presented for all three series of experiments; volitional extension, stimulated extension with varying cyclic frequencies and stimulated extension with varying pulse train frequencies.

Chapter six concludes with the correlation study in which there is an extremely close correlation between the volitional extension at a metronome frequency of 1 Hz and the stimulated extension with a pulse train frequency of 50 Hz and a cyclic frequency of .5Hz.

These two curves are compared and are plotted on a separate graph to facilitate examination by the reader.

1.7 Experimental Conclusions and Concluding Discussions

This chapter begins by listing the findings of this study of muscle forces developed during knee extension. The major findings are:

1) the angular variation of force approaches linearity with an overall stimulation cycle length of 1 sec. ON and 1 sec. OFF.
2) there is no significant force variation with changing pulse train frequencies, over the range of 50Hz to 80 Hz.
3) an excellent correlation exists between the forces developed during volitional extension at .5 Hz and stimulated extension with a stimulating pulse train frequency of 50 Hz and an overall cycle time of 2 sec., (1.0 sec/1.0 sec).

Other less significant conclusions are discussed in this chapter.

Sources of significant experimental errors are discussed next. By far the most important source results from mass and centre of mass measurements on the lower limb. These are observed to be approximately 4% in each case, (section 5.3.5).

Other sources include the manual measurement of deflections of the visicorder signal, the assumption of a stationary centre of rotation of the knee, and computer analysis of the raw data obtained from the visicorder.

The chapter concludes with some possible applications of the techniques used in this study. Such areas as knee joint function and evaluation of physiotherapy sessions are possible. Also discussed is an accurate means of measuring the mass and location of the centre of mass of any limb or limb section, by using the device discussed in section 4.1.

CHAPTER 2

THE KNEE JOINT

2.1.1 Discussion

At first glance, the knee joint appears to be a simple hinge-type structure, rotating in one plane only. However, with more detailed observation, it becomes apparent that this is only the major component of a combination of motions.

To support this fact, consider the following observations, which have been clinically documented:

1) During the flexion-extension cycle, the rotational axis of the joint does not remain stationary, but follows approximately a circular arc, with a radius of a few centimeters.
2) The rotational motion is partially due to a combination of rocking and sliding of the cartilages, which cover the joint surfaces, i.e. the heads of the femur and tibia.
3) Although flexion and extension of the lower leg are the primary motions of the knee joint, there also exists a transverse, or side-to-side motion. This motion is best observed when the knee is in flexion. (1)

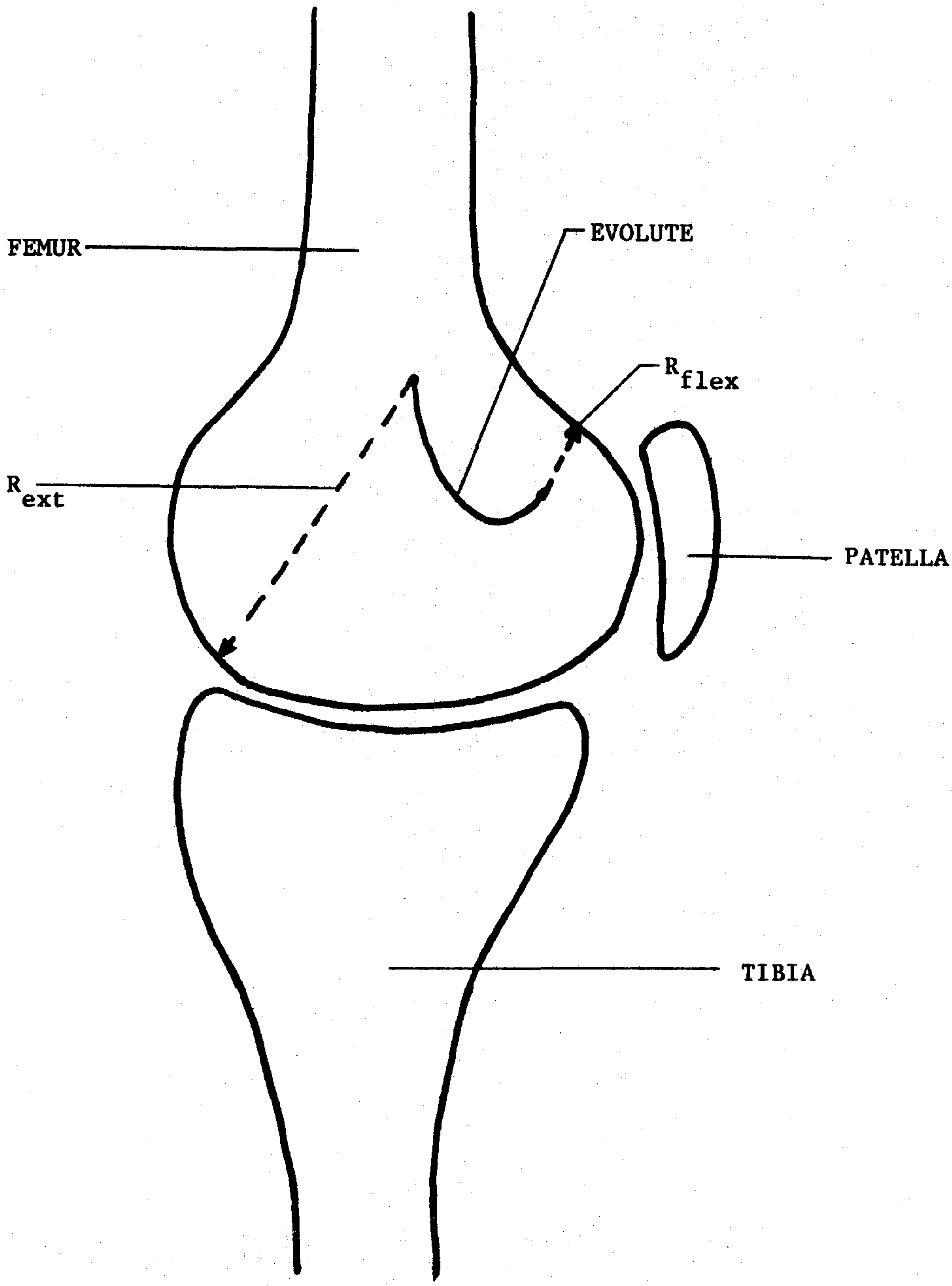


Fig. 2.1

MOTION OF INSTANTANEOUS KNEE-JOINT CENTRE DURING FLEXION-EXTENSION CYCLE

Thus rather than being a true hinge joint of a simple nature, i.e. having one rotational plane, the knee joint has facilities for motions in two planes, the saggital and transverse. Consequently, the knee joint is said to have two degrees of freedom.

For purposes of this research, the knee joint is assumed to have motion in only one plane, the sagittal plane. This is formed by the axis which bisects the leg in the posterior or anterior view. This assumption is based on the fact that other motions are extraneous and are considered negligible.

2.1.2 Rotational Axis of Knee Joint

The rotational axis of the knee for the flexion-extension cycle is an imaginary line piercing the medial and lateral femoral condyles. These are located above the actual contacting surfaces of the femur and tibia, and can be palpated on either side of the patella as bony protrusions through the skin.

Although the rotational axis is known to be through the condyles, it does not remain stationary upon rotation, but follows an approximately circular arc. Otto Fischer discovered this path, and called it the EVOLUTE. (2) Figure 2.1

He took X-ray images of a knee at different angles in the rotational cycle, and noted the points of contact between the heads of the femur and tibia. By measuring the centres of rotation in each image, he could finally trace out the evolute. One of Fischer's observations was that the centre of rotation was minimum at full flexion and maximum at full extension, (see figure 2.1).

Due to the nature of the electrogoniometer used in this research, this axis motion was not accounted for, but was assumed to remain fixed. There does exist, however, a floating polycentre electrogoniometer which accounts for this motion, but was not available at the time of this research.

2.1.3 Patellar Function in Flexion and Extension of Knee

The patella is a triangular bone which is imbedded in the Quadriceps tendon, located in front of and mid-way between the femur and tibia. A ridge located on the posterior portion of the patella allows it to move in the groove of the femoral head, between the two condyles. (3)

The patella is attached to the tibia by means of the patellar tendon. This is actually the central part of the Quadriceps Femoris which originates at the apex of the patella and inserts at the tuberosity of the tibia. Over the anterior region of the patella, this tendon is continuous with the Quadriceps tendon. (The Quadriceps tendon is also located laterally and medially to the patella, and inserts at the tibial tuberosity). An important feature of the patellar tendon is that it does not undergo a change in length when a force is exerted by the Quadriceps, across the joint. The implication is this; when the knee undergoes extension, the patella travels upward along the groove in the femoral head (4)

The moment arm of the transmitted force is greatly enhanced by this patellar motion, as the patella increases the distance of the force from the rotational axis. Studies have supported this fact, as there were significant increases in the muscle force required to extend the knee where patellectomies were performed. This can be shown analytically, as below:

<u>Extension</u>

$R_2 > R_3$

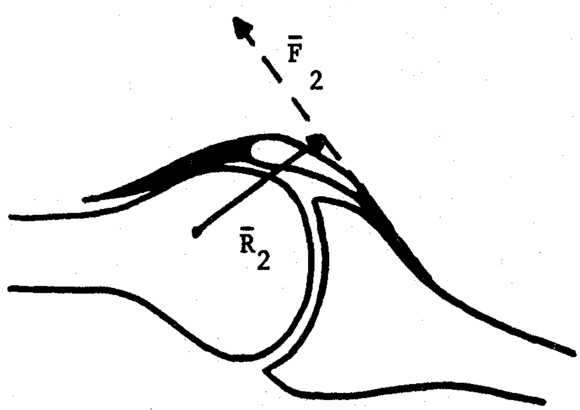


Figure 2.2 (a)

<u>Patellectomy</u>

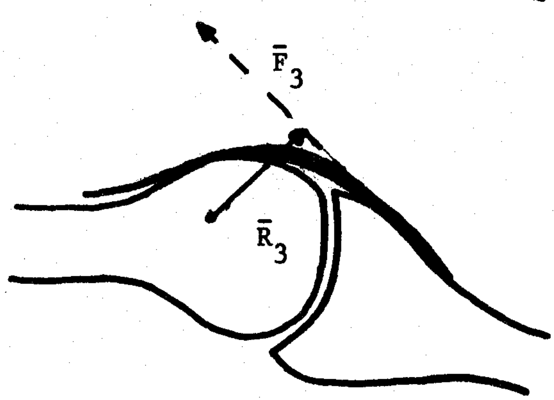


Figure 2.2 (b)

Moment of a force = Force X moment Arm

$$\bar{M}_2 = \bar{F}_2 \times \bar{R}_2 \qquad \bar{M}_3 = \bar{F}_3 \times \bar{R}_3$$

$$= F_2R_2 \sin\theta_2 \qquad = F_3R_3 \sin\theta ,$$

if $\bar{M}_2 = \bar{M}_3$ i.e. consider equal moments

and $\theta_2 = \theta_3$

then, $\bar{F}_2\bar{R}_2 = \bar{F}_3\bar{R}_3 \rightarrow \boxed{F_3 = F_2 \frac{R_2}{R_3}}$ - 1

Fig. 2.3

Thus, the extra force required to obtain the same moment at the same angle of extension is in proportion to the ratio of the two lever arms, as shown in figures 2.2 (a) and 2.2 (b).

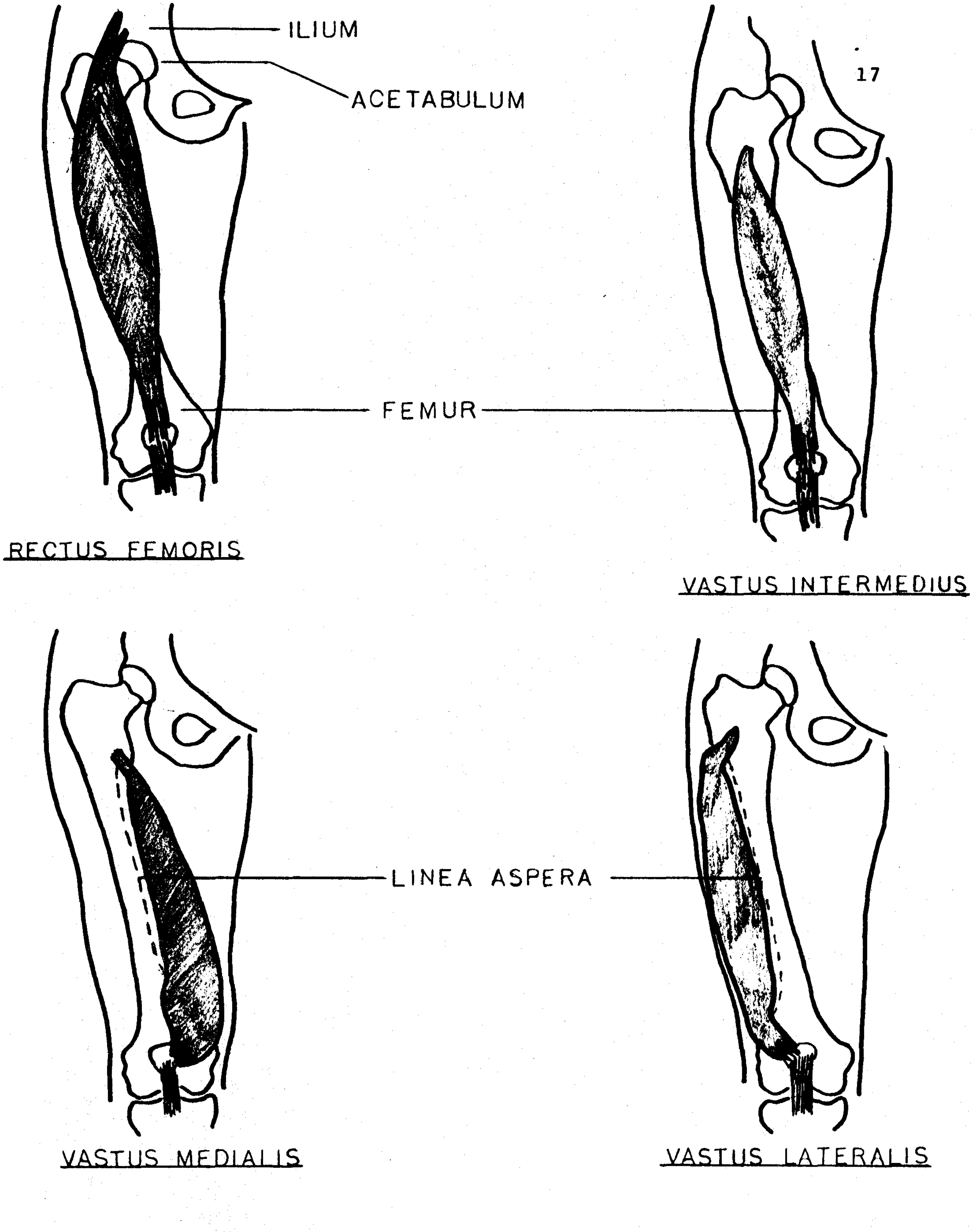
2.2.1 Major Muscles Affecting the Knee (Extensors) (5)

The muscle group known as the quadriceps is chiefly responsible for extension of the knee joint. The quadriceps are rather bulky and are located on the front portion of the thigh. This group is composed of four individual muscles; rectus femoris, vastus lateralis, vastus medialis and vastus intermedius. (see figure 2.4)

The rectus femoris is the central muscle of the quadriceps. It is a superficial muscle, in that it is easily observable directly under the skin and runs down the very front of the thigh. Its origin at the hip is by means of two tendons, the anterior tendon, from the anterior inferior spine of the iluim, and the posterior tendon, from just above the acetalubum. Its insertion is by means of a relatively broad tendon attached to the upper rim of the patella. This tendon also rides over the outer side of the patella and inserts into the tibial tuberosity via the patella tendon. It may be easily palpated by locking the knee in full extension and flexing the thigh muscles.

The vastus lateralis, as its name implies, is the muscle located on the outer front portion of the thigh, next to the rectus femoris. It is also a superficial muscle being readily observable. This muscle originates (via its tendon) in the upper lateral portion of the femur, up to the greater trochanter and posteriorly to the linea aspera. It inserts into the patella, on its upper and lateral sides, and by forming part of the patellar tendon inserts into the tibial tuberosity. Two additional points of insertion are the lateral tibial condyle and the fascia of the lower leg. The head of the vastus lateralis is easily palpated just above and laterally to the patella when the knee is extended and the thigh flexed.

The vastus medialis is located on the opposite side of the rectus femoris, being the innermost superficial muscle of the quadriceps. It originates, by tendons, on the upper medial portion



MUSCLES OF THE QUADRICEPS GROUP

Figure 2.4

of the femur, reaching the intertrochanteric line and posteriorly to the linea aspera. Insertion is on the upper and medial tibial condyle, and the fascia of the lower leg. The head of the vastus medialis can be palpated just above and medially to the patella when the knee is in full extension.

The third muscle in the vasti series is the vastus intermedius, which lies underneath the rectus femoris. The upper portions of this muscle and the other two vasti are indistinguishable, as they blend into one common muscle. The vastus intermedialis originates from the upper anterior and lateral parts of the femur, up to the lesser trochanter and posterior to the linea aspera. It inserts into the upper edge of the patella by a common tendon with the other vasti muscles, and into the knee joint capsule. Again, by means of the patellar tendon, insertion also occurs at the tibial tuberosity.

2.2.2 Uni-Joint and Bi-Joint Muscles

As the names imply, these muscles extend over one and two joints respectively, the knee joint alone, or either the knee and ankle or knee and hip. Uni-joint muscles require no other explanation other than they affect the joint over which they are located. Bi-joint muscles, on the other hand, will be discussed briefly, since the Rectus Femoris is a bi-joint muscle.

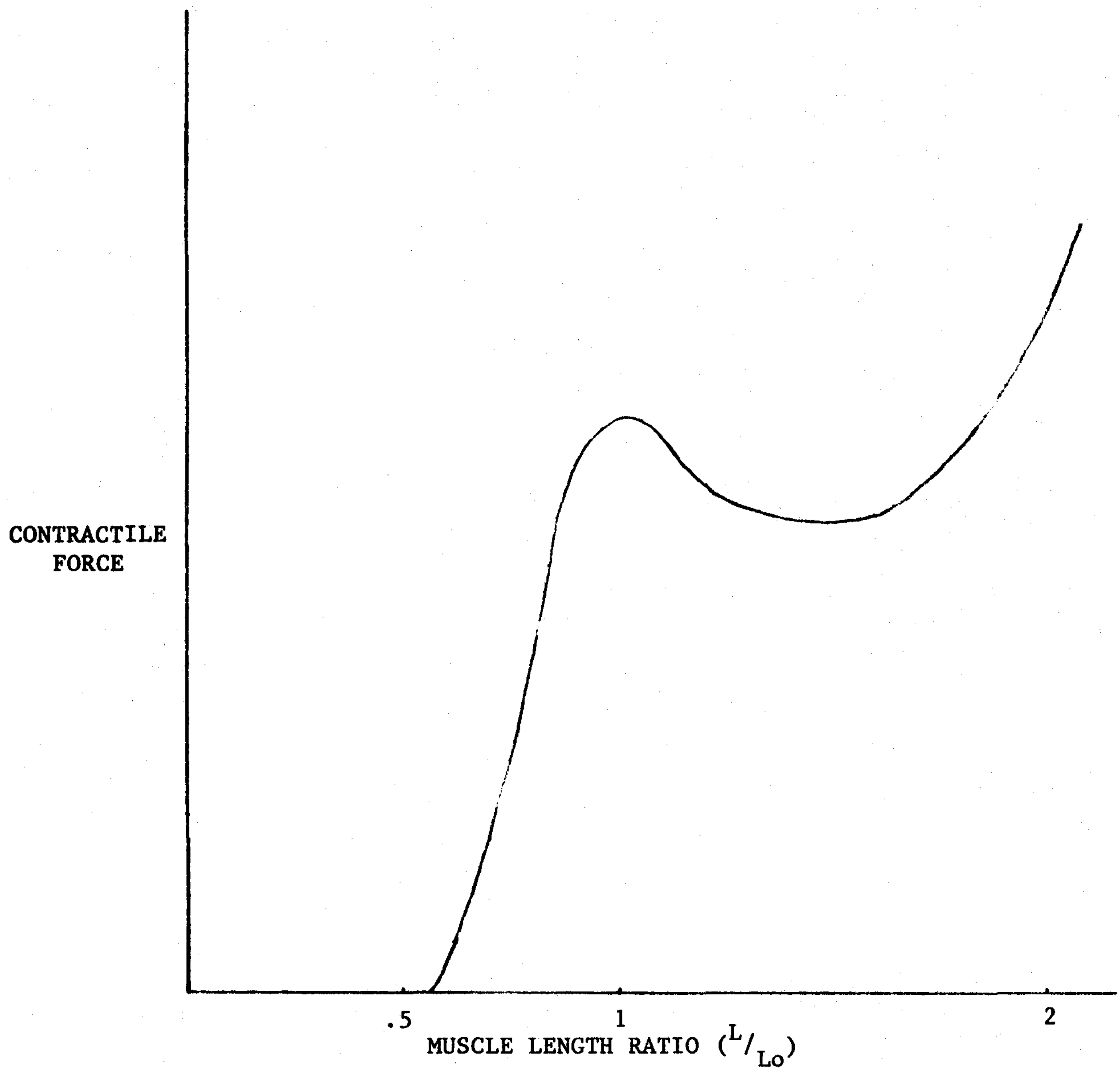


Fig. 2.5

CONTRACTILE FORCE AS A FUNCTION OF MUSCLE LENGTH

(Lo = Resting Length)

A bi-joint muscle will affect both joints simultaneously, but the degree of this movement is dependent on several things. The prime factor is the length-tension relationship. Consider that the two joints derive significant ranges of motion due to contraction of the bi-joint muscle. To accomodate this movement, the muscle would have to shorten considerably. But by observation of a typical length contractile force curve, fig. 2.5 (7) the force of a muscle decreases sharply upon shortening, reaching zero when the muscle has shortened to about 60% of its resting length. Thus, if this motion was permitted, efficiency of movement would decrease markedly. However, other factors enter in to eliminate this loss of efficiency. Gravitational effects can produce a resistance in one of the joints, which will prevent rotation at that joint. Contraction of other muscles can also counteract the effect of the bi-joint muscle at the joint.

Thus, under usual circumstances, a bi-joint muscle will not move both joints simultaneously, but will move the joint with the more favourable conditions of leverage.

Classifying the muscles of the knee previously described, into uni-and bi-joint muscles, one has the following, as in table 2.1;

Uni - Joint	Bi - Joint
Vastus Lateralus	Rectus Femoris
Vastus Medialus	
Vastus Intermedius	

Table 2.1

2.3 Discussion of Classes of Levers Applied to Muscles Of the Knee (8)

Stability and motion of the body is a direct consequence of skeletal muscle exerting a force on the skeleton, usually across one or two joints. How this force is applied depends upon which of the three classes of levers is involved. A brief description of these lever classifications is presented for purposes of illustration.

In a lever of the first class, fig. 2.6, the fulcrum is located between the applied force and the mass to be moved.

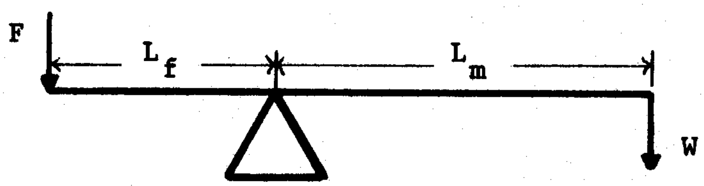


Figure 2.6

For equilibrium in this class, it is required that,

$$F = +W \frac{L_m}{L_f}$$

A lever of the second class, fig. 2.7, has the mass located between the fulcrum and applied force, which is applied at the free end of the lever arm.

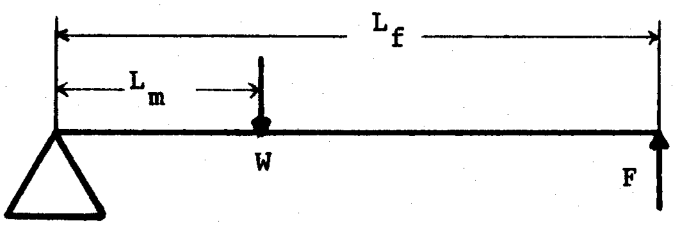


Figure 2.7

Clearly, there is always present in this class, a mechanical advantage, since the lever arm of the applied force is greater than that of the mass to be moved, i.e. L > 1. Thus, for equilibrium,

$$F = +W \frac{L_m}{L_f}$$

A third class lever, fig. 2.8, is the same configuration, except that the locations of the mass and applied force are interchanged. This class operates at a constant mechanical disadvantage, since the lever arm of the force is less that that of the mass, i.e. l < L, and equilibrium requires,

$$F = -W\frac{L_m}{L_f}$$

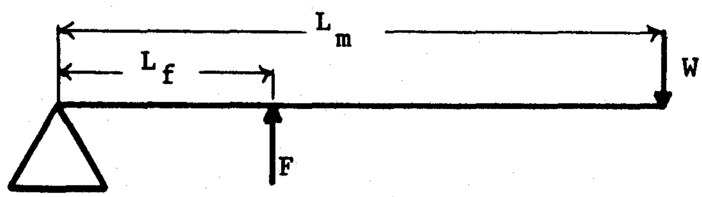


Figure 2.8

Considering the extensors of the knee discussed in section 2.2.1, in relation to the preceeding description of lever classes, one can summarize the results as in table 2.2

FUNCTION	MUSCLE	LEVER CLASS
Knee Extensors	Rectus Femoris	Third
	Vastus Lateralis	Third
	Vastus Intermedius	Third
	Vastus Medialis	Third

Table 2.2

The classification of the knee extensors may not be immediately understood, as they all insert into some aspect of the patella. However, recalling from section 1.1.6, that the patellar tendon inserts into the tibia, and indeed, is a continuation of the rectus femoris tendon in which the patella is imbedded, it becomes clear that these muscles are, in fact, third class levers, (figure 2.9)

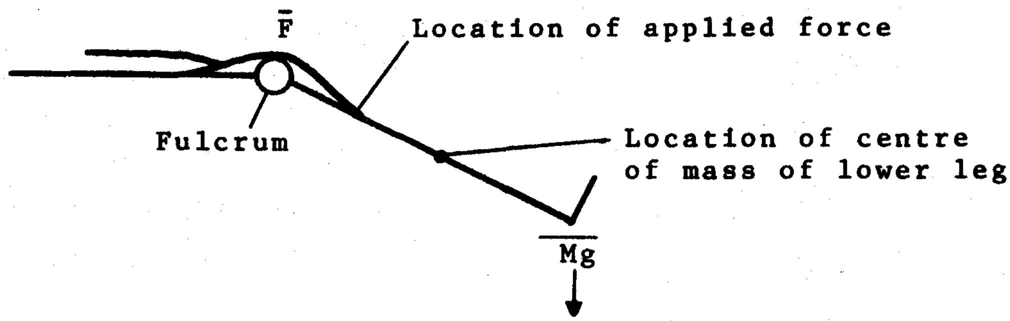


Figure 2.9

2.4.1 Motor Units

Contraction of skeletal muscle is totally dependent on the central nervous system, which exerts its control over the muscle by means of motor units.

A somatic motor nerve from the spinal cord reaches a muscle and enters it, undergoing numerous branchings, each branch terminating on a single muscle fibre in a motor end-plate or myoneural junction; at which the sarcoplasm of the muscle fibre bulges toward the nerve terminus. This physiologic entity, from the spinal cord to the myoneural junctions of the muscle fibres is the motor unit. (9)

A local group of muscle fibres is not controlled by a single branch of the somatic motor nerve. Rather, one branch will terminate on fibres located in different regions of the muscle. This stems from the fact that for efficient muscle contractions, the muscle fibres must contract asynchnonously, i.e. at different times. Thus, the muscle contains a vast network of nerve fibres, intermeshed throughout the muscle.

These are fundamental in the study of electro-physiologic phenomena, as shall be shown in the following section, 2.4.2.

2.4.2 Motor Points

Motor points are known regions of a muscle, which, when electrically stimulated, elicit the strongest muscular contractions.

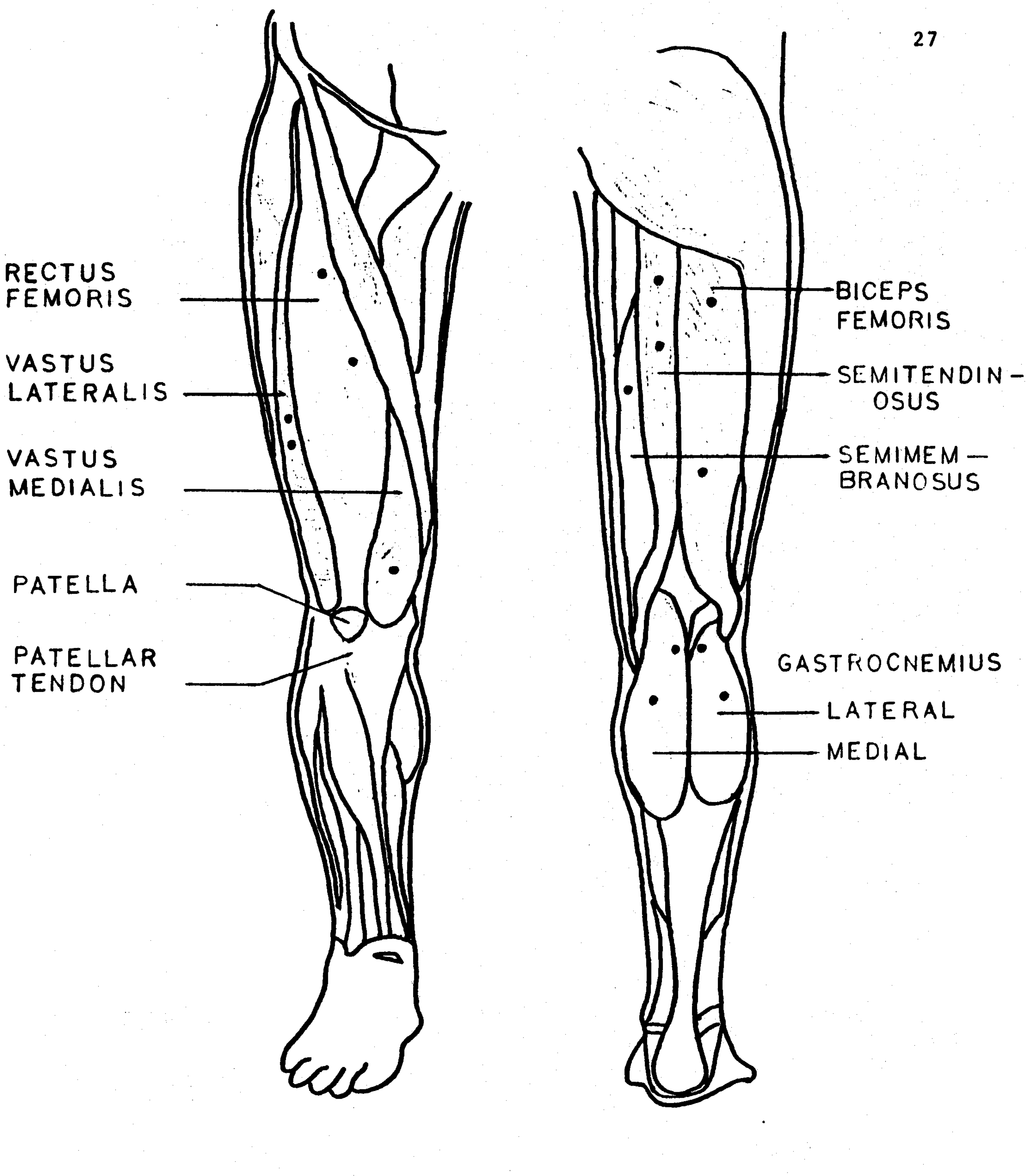
The question of what these areas are and why they should exist were answered by C. Coers, in 1955. (10)

After locating a motor point over a muscle, Coers performed a biopsy on the muscle directly beneath this area of the skin. He observed that muscle fibres from this region constituted a zone of low threshold for excitability, relative to muscle fibres away from these zones, and were oriented perpendicular to the muscle fibres. He noted, too, that this low threshold region contained a high concentration of myoneural junctions.

Thus, he summarized that these motor points, rather than being the entrance of the motor nerve into the muscle, are actually clusters of nerve terminations on the muscle fibres.

These regions of low excitability threshold may be demonstrated in the laboratory.

A large surface electrode is secured to the skin away from the muscle and a probe electrode is grasped by its handle and the current is increased slightly, (discussion of electrodes can be found in section 4.2.1). The probe electrode is moved over the skin above the muscle, and when a motor point is encountered, muscle contractions will be observed. (For pulse parameters of stimulus current, see section 3.3).



LOCATION OF PERTINENT
MOTOR POINTS OF LEG

Figure 2.10

The low current required to elicit a muscle contraction by a motor point will not cause contractions when the probe is moved a few centimeters from this region.

Having attained a sufficient current to cause strong muscular contractions by a motor point, the author has observed a painful sensation when this current is introduced into the muscle in any other location. This can also be shown by an extension of the preceding demonstration.

Locating the probe electrode away from a motor point, increase the current until slight pain is noticed. When the probe is then moved to the motor point, the painful sensation will disappear, and a tingling sensation will be felt along this muscle.

A diagram of motor points of the major muscles affecting the knee is included for convenience in the above demonstrations. (Figure 2.10) (11)

CHAPTER 3

ELECTROSTIMULATION

3.1 Principles of Electrostimulation of Skeletal Muscle

Nerve and muscle fibres are surrounded by a membrane whose resting potential is approximately 90 millivolts (12) with the exterior being positively charged relative to the interior. This potential is maintained by the anions and cations being kept in balance by various processes, concentration gradients, molecular pumps, osmosis, etc. (figure 3.1)

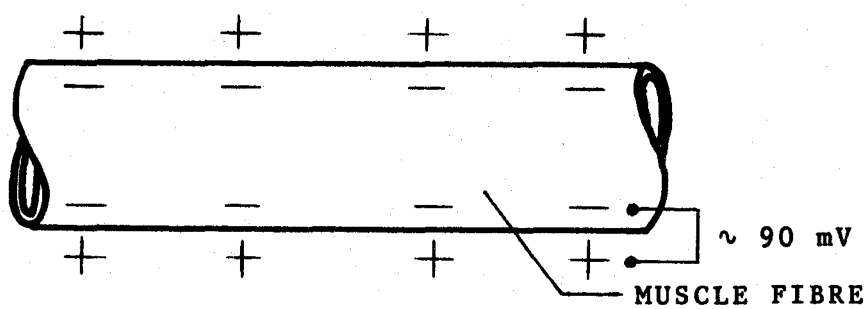


Figure 3.1

Under normal circumstances, i.e. excitation by the central nervous system, an action potential travels along the motor nerve, enters the muscle and terminates on the various muscle fibres in a motor end-plate. Without delving into the molecular processes involved, this action potential depolarizes the muscle fibre, in the local area surrounding the end-plate. (figure 3.2) If the stimulus or action potential attains or exceeds the threshold level of the fibre then an action potential or end-plate potential is initiated in the fibre causing it to contract. Thousands of such processes occurring throughout the muscle cause a gross contraction of the entire muscle. This contraction, in normal subjects, is entirely voluntary, being initiated by an impulse from the central nervous system.

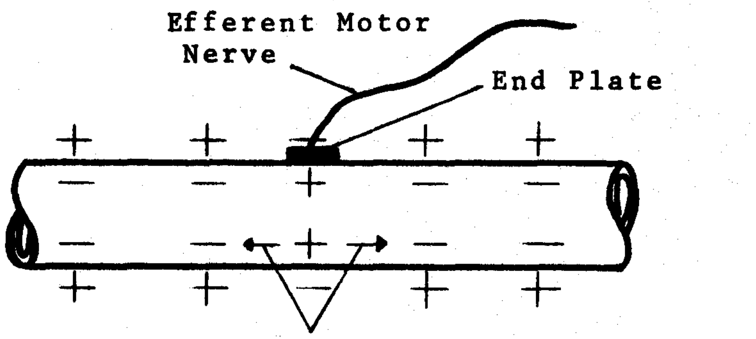


Figure 3.2

External electrical stimulation is clearly involuntary, as it is controlled by a source outside the body. This external stimulus bypasses the central nervous system entirely, initiating the contractile process by direct stimulation of the motor end-plates or efferent motor nerve, supplying the muscle.

In superficial muscle stimulation, a small electrode is placed over a motor point, which is an area of high concentration of motor end-plates (section 2.4.1), and a current transmitted through the muscle from a larger electrode. The high current density near the small (active) electrode, can be sufficient for depolarization of the muscle fibres to occur. The rest of the contractile process is virtually identical to that from the central nervous system.

External stimulation of the efferent nerve is a combination of natural stimulation and that described for superficial stimulation above. In this process, the efferent nerve axon is intercepted between the central nervous system and the muscle, before it undergoes the numerous branchings to the muscle fibres. The electrodes can be either inserted to contact the nerve or applied to the skin directly over the nerve. The current is gradually increased until muscle contraction occurs. By this method, the nerve fibre undergoes depolarization from the external stimulus, which in turn elicits an end-plate potential in the muscle fibres supplied by branches from this motor nerve.

In nerve stimulation, figure 3.3 (a) and figure 3.3 (b), other muscles will be stimulated along with the muscle being studied. But with superficial muscle stimulation, a single muscle may be stimulated alone, since it is only the motor points of that muscle undergoing stimulation.

In view of the above discussion, superficial muscle stimulation was incorporated due to its relative ease of application and selectivity of single muscles to be stimulated.

3.2 Techniques of Superficial Muscle Stimulation

There are two basic methods of direct superficial stimulation of a muscle, differing only in electrode placement and effects.

Bipolar stimulation, (figure 3.4 (a)) utilizes two electrodes of the same size.

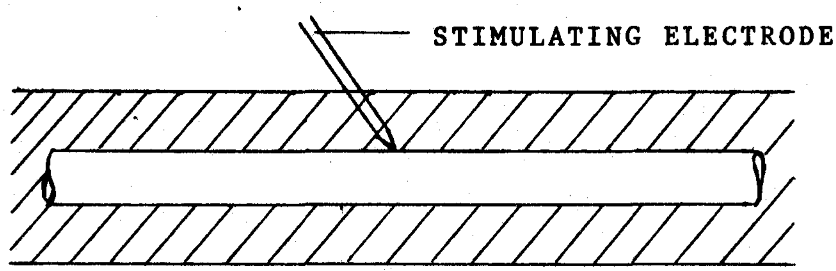


Figure 3.3 (a)

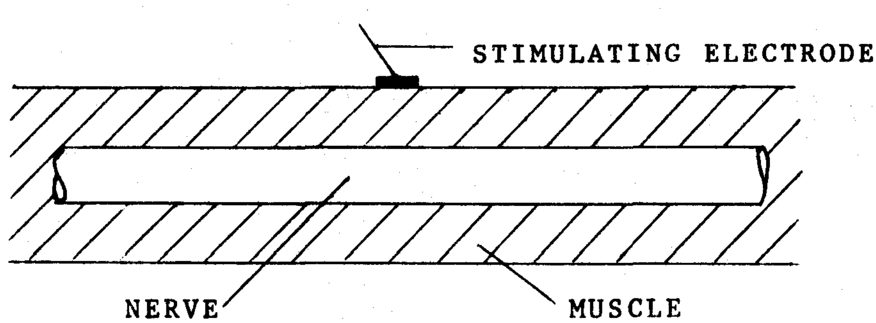


Figure 3.3 (b)

DIRECT AND INDIRECT NERVE STIMULATION

Each electrode is situated over the muscle, on either side of the motor point. Current is applied from an external stimulator, which enters the muscle and stimulates the motor point.

Two drawbacks of this method are as follows. First, because there is no concentration of current at the motor point, there exists a low current density in this region. Thus, a relatively high input current is required to stimulate the muscle and there is a possibility of stimulating muscles other than the muscle being tested. This arises from the relatively high input current required.

Unipolar stimulation (figure 3.4 (a)) also incorporates two electrodes, however they are of different sizes. The smaller active electrode (cathode) is located directly over the motor point, while the larger indifferent electrode (anode) is placed on the skin away from the muscle, or near its point of insertion. Current is again supplied, and at the larger indifferent electrode, there exists a low current density, whereas at the smaller active electrode, there is a higher current density, which is inversely proportional to the ratio of the areas of the two electrodes. Consider:

A_{ind} = area of indifferent electrode (cm^2)

A_{act} = area of active electrode (cm^2)

I = input current (A)

J = current density ($A\text{-}cm^2$)

then, the current density at the indifferent electrode is,

$$J_{ind} = I/A_{ind}$$

and, the current density at the active electrode is,

$$J_{act} = I/A_{act}$$

thus, the ratio of the active to the indifferent electrode current densities is,

$$Jact/Jind = Aind/Aact$$

Thus, to achieve the same contractile strength as in the bipolar arrangement, a smaller input current is required in the unipolar technique. These stimulation techniques are depicted in figure 3.4 (a) and figure 3.4 (b) respectively. (13)

The method of superficial muscle stimulation chosen was the uni-polar electrode technique, since it provides greater selectivity in stimulating single muscles coupled with the lower input current requirements, relative to the bi-polar method.

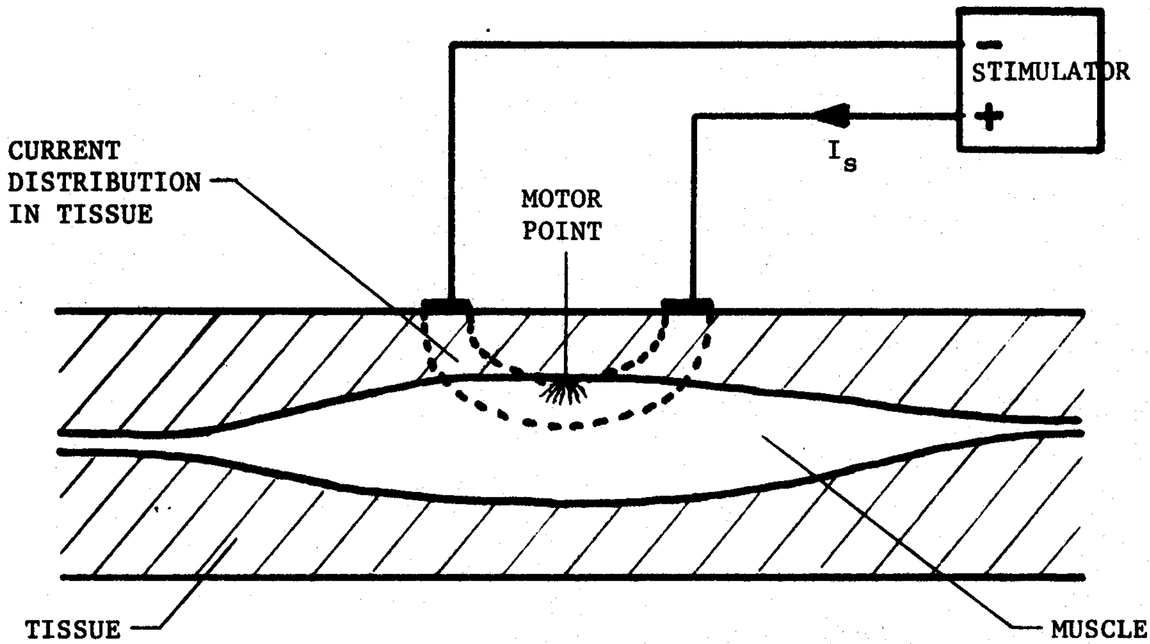


Fig. 3.4(a)

UNIPOLAR MUSCLE STIMULATION

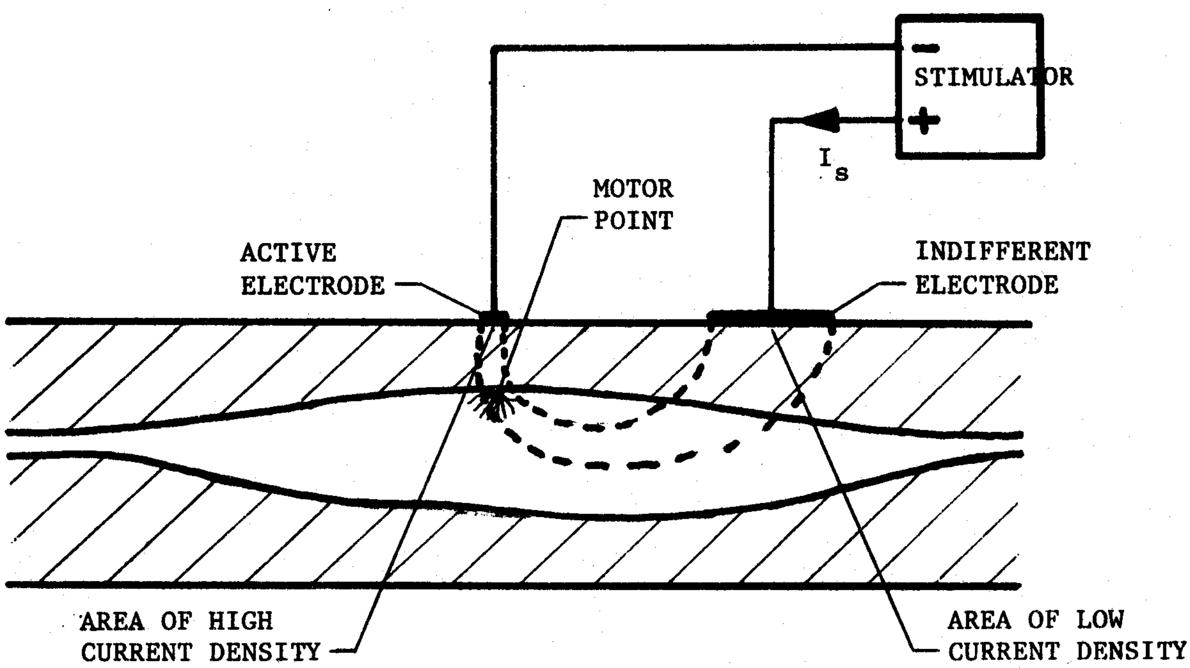


Fig. 3.4(b)

BIPOLAR MUSCLE STIMULATION

3.3 Stimulation Pulse Parameters

A pilot experiment was performed to establish the pulse parameters required to elicit the maximum strength of contraction. The vastus lateralis muscle of the quadriceps group was chosen, since it was easily observable through the skin, and contractions of this muscle was the clearest.

The details of this experiment can be found in section IV, while only the results are presented here, to avoid repetition.

individual pulse frequency 50Hz
width of pulse train .... 1 sec.
number of pulses in train 50
pulse train repitition rate..... 1 sec. ON,
1 sec. OFF
stimulus current .... 36 mA

With these parameters, the strongest contraction of the vastus lateralis muscle was observed. The magnitude of the stimulus current cited was the value just before the onset of painful sensations in the vastus lateralis.

3.4.1 Stimulation

The stimulator used for the purposes of electrical stimulation of muscles in this research was of a constant current variety. This type of stimulator has a very high output impedance, 100 kΩ for this particular model, enabling it to maintain a very stable current between the indifferent and active electrodes, i.e., through the muscle. This is due to the fact that it is largely independent of the variations in skin and tissue impedance, as well as fluctuations of the resistance between the electrode and the skin, because the variations which inevitably result are negligible relative to the high output impedance. Thus, with surface electrodes, as used here, it is advantageous to use a constant current stimulator.

3.4.2 Stimulator Adjustments

This stimulator has four separate channels for use in muscle stimulation. The first channel contains the master oscillator and its parameter adjustments, which control all four channels simultaneously.

The master oscillator frequency knob controls the spacing of each individual pulse of the signal. This parameter can be varied continuously between 1.4 milliseconds and 50 milliseconds, and all four channels will have the same frequency, (figure 3.5 (a))

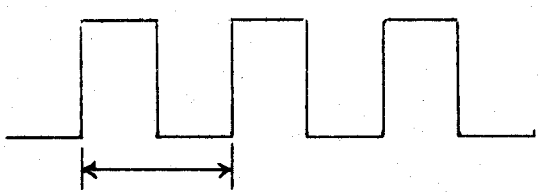


Figure 3.5 a

MASTER OSCILLATOR FREQUENCY

The <u>master oscillator pulse width</u>, as the name implies, controls the width of each individual pulse. This parameter has two ranges for continuous adjustment, from .1 msec. to .8 msec., and from 1.4 msec to 10 msec. This adjustment also controls all four channels, (figure 3.5 (b)).

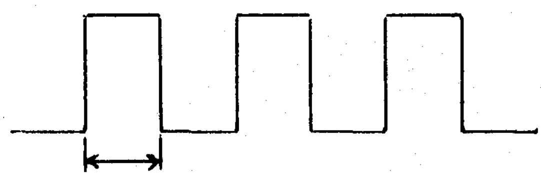


Figure 3.5 (b)
MASTER OSCILLATOR PULSE WIDTH

The train width is the width of a single group of pulses. It also has two ranges for continuous adjustment, from 7 msec. to 80 msec., and from .4 sec. to 3 sec. The effect of increasing the train width is to add pulses to the train at the same frequency as the other master oscillator pulses. Decreasing the train width has the opposite effect, (figure 3.5 (c)).

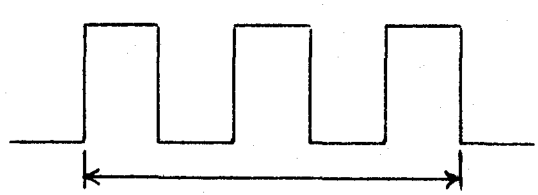
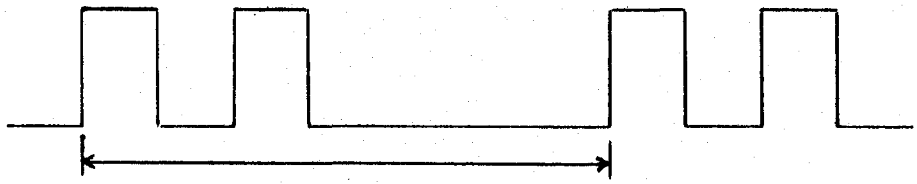


Figure 3.5 (c)
TRAIN WIDTH

The <u>sweep rate</u> controls the frequency of the trains of pulses which are delivered into the muscle. It is analogous to the master oscillator frequency control if one replaces each pulse in the latter with a train of pulses. It has a continuous range from .3 sec. to 50 msec., (figure 3.5 (d)).



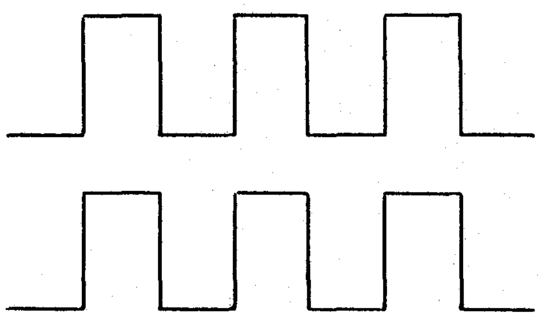
SWEEP RATE

Figure 3.5 (d)

The <u>output level</u> control knob adjusts the amount of current being delivered into the muscle, i.e. the pulse or train height.

For channels, 2,3 and 4, the controls for the train width and output level are identical with those for channel 1. However, these three channels have one unique control feature. This is the <u>train phase</u>. This parameter is the time delay between the start of the pulses from channel 1 and the start of the pulses from either channels, 2,3 or 4. The adjustment has two possibilities. Either the train phase is zero, (figure 3.5 (e)), in which case the start of the pulses from the other three channels coincide with the start of the pulses of channel 1, or the train phase is variable (figure 3.5 (f)), in that the starts of the other three channels are delayed relative to channel 1.

TRAIN PHASE= ZERO

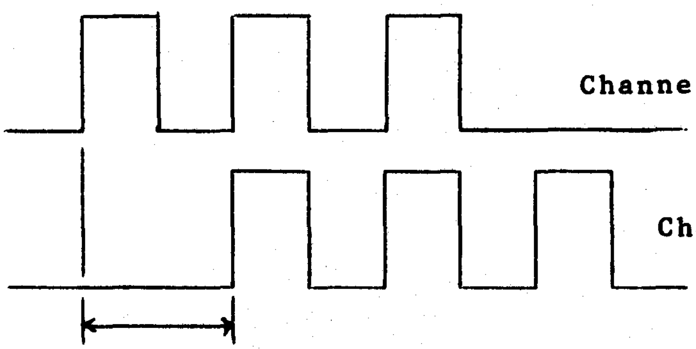


Channel 1

Channel 2

Figure 3.5 (e)

TRAIN PHASE = VARIABLE



Channel 1

Channel 2

Train Phase

Figure 3.5 (f)

CHAPTER 4

APPARATUS AND DEVICES

4.1.1 Location of Centre of Mass

In many physiologic studies, one needs to know where the centre of mass of the object being studied is located. This is especially true if meaningful calculations of torques are to be obtained.

In this study, it was necessary to obtain an accurate measure of this parameter for the lower limb, as well as the lower limb mass. Since a commercially marketed device was not readily available, it was necessary to design and construct such an apparatus.

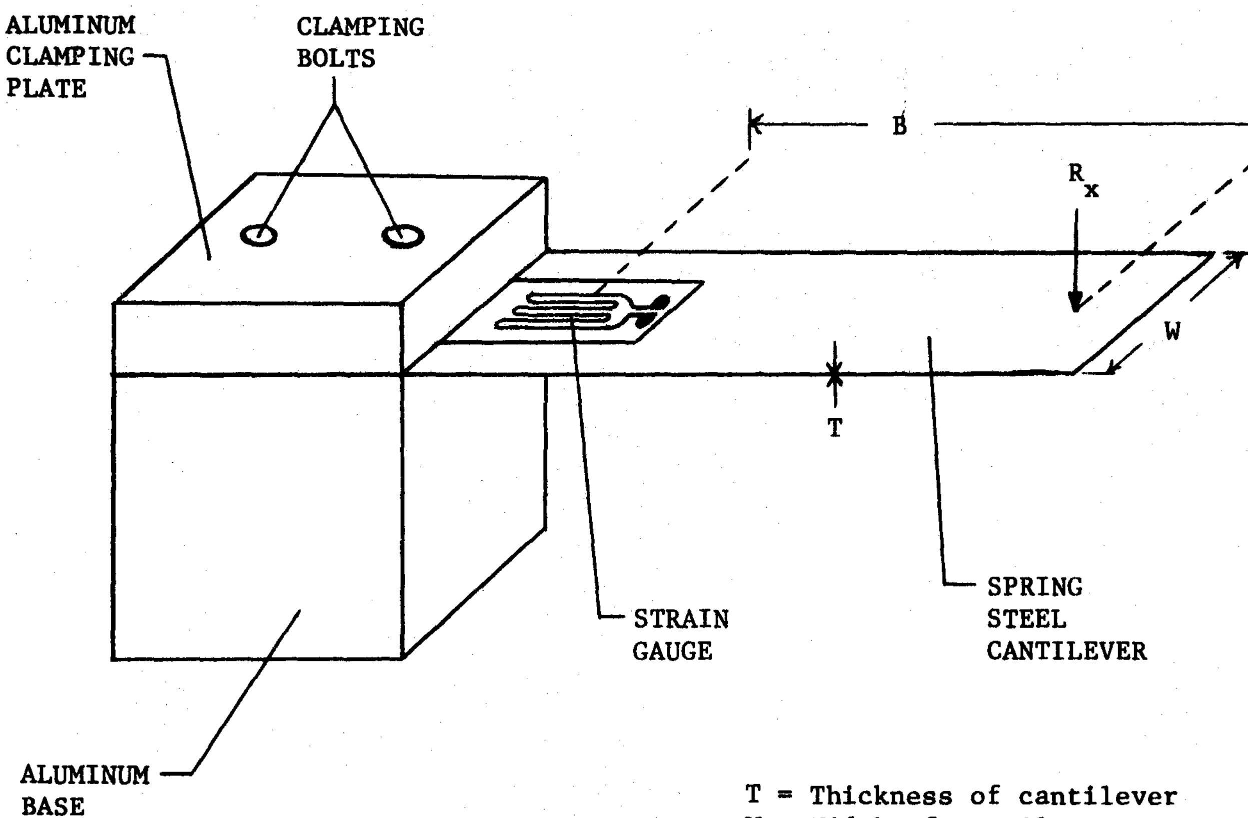
4.1.2 Final Design of Centre of Mass Measuring Device

This model incorporates the simple cantilever idea, i.e., a bar which is clamped at one end and free on the other. Upon this bar, at the clamp, are located two strain gauges, one on the top side and one directly beneath, on the bottom side, figure 4.1. Two gauges were used rather than one for two reasons. One, the sensitivity of the unit is doubled and two, effects of temperature variation of the cantilever are made negligible.

CANTILEVER STRAIN EQUATION —— $\epsilon_x = \frac{6B}{EWT^2} R_x$

ϵ_x = Strain due to loading force

E = Elastic modulus of spring steel cantilever



T = Thickness of cantilever
W = Width of cantilever
B = Distance from gauge centre to point of loading
R_x = Force of loading

Fig. 4.1

CANTILEVER UNIT

Three such cantilever units were used, and were placed at the corners of an isoceles triangle. Each unit was mounted on an aluminum block, one inch thick, to ensure that each unit was at the same relative height. The entire sub-assemblies were mounted on a three-quarter inch thick plywood base by means of ¼" bolts; such that the free ends of the cantilevers were facing towards the middle of the plywood base, at right angles to the edge of the base.

Resting on these three cantilevers, is a three-quarter inch plywood board. The contact between this support board and the three cantilevers is accomplished by means of three metal points protruding through the board in such a manner as to fit into a 1/8" diameter hole drilled through each cantilever along its midline, one quarter inch from the free end.

The two strain gauges on each cantilever form two arms of a balanced bridge circuit. The details of this circuit are discussed in section 4.1.4.

4.1.3 Mathematical Analysis

For a simple cantilever, clamped at one end, the relation between the applied load and the resultant strain is given by,

$$\varepsilon_x = \left[\frac{6b}{E\ wt^2}\right] R_x \qquad (14)$$

(4.a)

in which R_x and ε_x are the applied load and resultant strain respectively, w and t are the width and thickness of the cantilever, respectively, b is the distance from the centre of the gauge to the location of the applied force, and E is the modulus of elasticity of the cantilever material.

By inspection, one can see that this relationship is linear, with a slope given by the term in brackets. This makes the equation easy to utilize since a change in the load is directly proportional to the change in the observed strain. For the particular gauges used, the apparent strain due to temperature effects is zero at room temperature, 25° C, thus the strain observed is the actual strain experienced by the cantilever.

4.1.4 Circuitry for Device

When a resistive strain gauge experiences a strain with the member to which it is attached, it undergoes a change in length, which corresponds to a change in its resistance. A relation for resistance is,

$$R = \frac{p\ l}{A} \qquad (4.b)$$

with R the resistance, p the resistivity, l the length and A the cross-sectional area of the wire of the gauge. Thus, when the gauge experiences a lengthening, the length increases and the cross-sectional area decreases. Both of these contribute to the increased resistance.

The usual method of utilizing strain gauges is to incorporate them as part of a balanced bridge circuit. Three such bridges were used on this device, one for each of the three cantilevers. Initially, the gauges were driven by a D.C. voltage of 15 volts, but a 10 mV drift was observed when calibration of the device was attempted. This corresponded to an error ranging from 5% for high loads (3kg) to 20% for low loads (<1Kg). Thus, it was decided to drive the gauges with an A.C. signal.

To this end, a signal generator was incorporated, with a frequency of 1 KHz being fed into the gauges. This output from the gauges was put through a rectifier and averager circuit (see Appendix 1) and this signal was used to drive the oscilloscope. No drift could be observed on the scope at its lowest sensitivity, (.01 volts/cm), hence the error due to drift was reduced to less than 1% for high loads up to 3% for low loads. Effective calibration could then be achieved.

A block diagram of the circuit appears in figure 4.2.

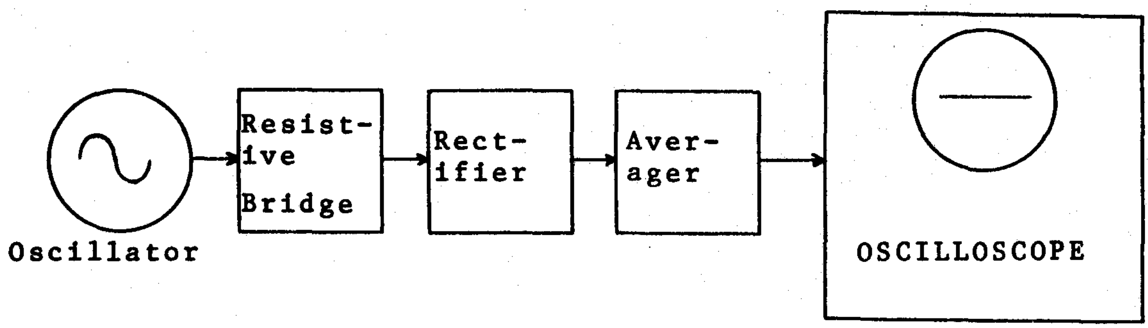


Figure 4.2

All amplifiers used were the 741 TC operational amplifiers, which were driven by +15 volts D.C. The reader is referred to Appendix 1 for a detailed schematic diagram of all circuits used with this device.

4.1.5 Calibration

Calibration of this device was accomplished using several standard weights, whose denominations included, 100 gm, 200 gm, 500 gm. and 1 Kg. Measurements of output voltage on the oscilloscope were taken at 200 gm. intervals.

The sets of weights were suspended from the free end of the cantilever by means of a Hollman clamp, directly beneath the point of contact of the support board and cantilever.

Recall from section 4.1.2, that this distance was ¼ inch from the free end along the longitudinal axis.

The observed signal on the oscilloscope was quite linear as expected by inspection of the equation in section 4.1.3 in the range of .8Kg to 4.0 kg. The slope of this range is, on the average, 3.2 mV/100 gm. (3mV/100 gm. to 4 mV/100 gm.). In the range from 0Kg. to .8 Kg., the deviation from linearity can be attributed to the straightening of the slight curvature of the cantilever resulting from cutting it from a larger sheet of stainless steel. The cantilever was straightened as much as possible without doing damage to it, however, a slight residual curvature remained.

4.1.6 Principles of Operation

The support board, on which the limb rests, contacts each of the three cantilevers at one point, thus forming a triangular base. This presents an excellent basis of stability so long as the limb is initially arranged such that the bulk of the limb is located within the area defined by these contact points. Placing the limb carefully along the longitudinal axis of the support board will achieve this.

A simplified illustration of the device in operation appears in figure 4.3.

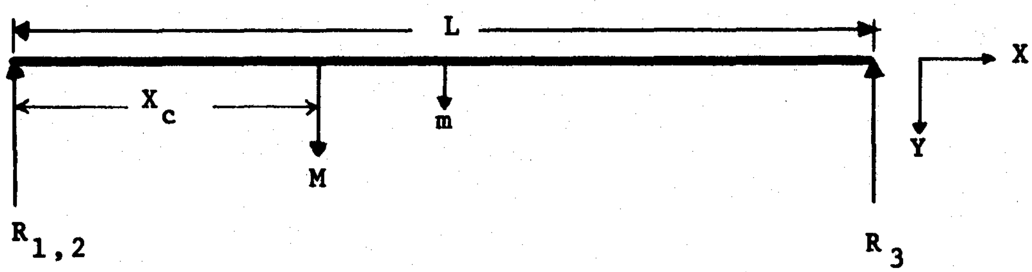


Figure 4.3

$R_{1,2}$ = sum of reactions from cantilevers 1 and 2.

R_3 = reaction from cantilever 3

M = mass of limb section being tested

m = mass of support board.

Xc = distance of centre of mass from rotational axis of joint.

L = length of support board.

Calculation of limb mass, M, is obtained through balance of vertical forces, i.e.;

$$\Sigma F_y = 0 = M + m - R_{1,2} - R_3$$

$$\text{OR} = \boxed{M = R_{1,2} + R_3 - m} \qquad (4.c)$$

Calculation of the location of the centre of mass, X_c, is obtained through balancing of moments about R_1, z_1, i.e:

$$\Sigma M_z = 0 = MX_c + m \frac{L}{2} - R_3 L$$

$$\text{or} \quad X_c = \frac{R_3 L - \frac{mL}{2}}{M} \qquad (4.d)$$

But a relation was obtained in equation 4.c for M, thus this last equation becomes;

$$\boxed{X_c = \frac{(R_3 - \frac{m}{2})\,L}{R_{1,2} + R_3 - m}} \qquad (4.e)$$

Thus, we have two equations, one each for the mass and location of centre of mass for the tested limb. In each case the unknown quantities are in terms of known or measurable quantities. Only one measurement is necessary to obtain these unknown quantities.

Detailed drawings of this device can be found in Appendix 1, and results of actual measurements on an intact limb appear in section 5.3.5.

4.2.1 Electrodes

Probe Electrode

To locate the motor points of the muscles involved, a probe electrode was fabricated. A phonograph jack plug was obtained and the tip was flattened to its widest diameter.

A plastic coated wire was soldered to the jack, and a six inch length of 1/8" O.D. acrylic tubing was slid over this wire, to be fitted into the base of the phonograph jack. The free end of the wire was soldered to a connector plug, which was then able to be connected to the stimulator terminal, (figure 4.4).

When utilizing this probe, a gauze pad was attached to the flattened tip by an elastic band. This gauze was saturated with a conductive EKG gel, and kept moist by immersing the gauze periodically in ordinary tap water. This provides a good electrical contact while avoiding direct contact of the skin with the bare metal of the jack. When probing with this electrode, it is held by the plastic base of the jack, similar to holding a pencil. The researcher is thus isolated from the stimulator current.

Active and Indifferent Electrodes

Both the active and indifferent electrodes were cut from a single sheet of stainless steel wire mesh, having one hundred wires per linear inch (designated "100 mesh").

The active electrode measured 3/4" inches square, thus having an area of 1 1/8" square inches. A square piece of mesh was cut, measuring 1¼ inches on each side. Side flaps of ¼ inch were measured off and the four corners were cut off squarely. The remaining flaps were then folded over and a lead wire was soldered onto one side, under the flap (and a connector was soldered onto the free end of the wire). The folded sides were covered with electrical tape so as not to have sharp ends of the mesh protruding.

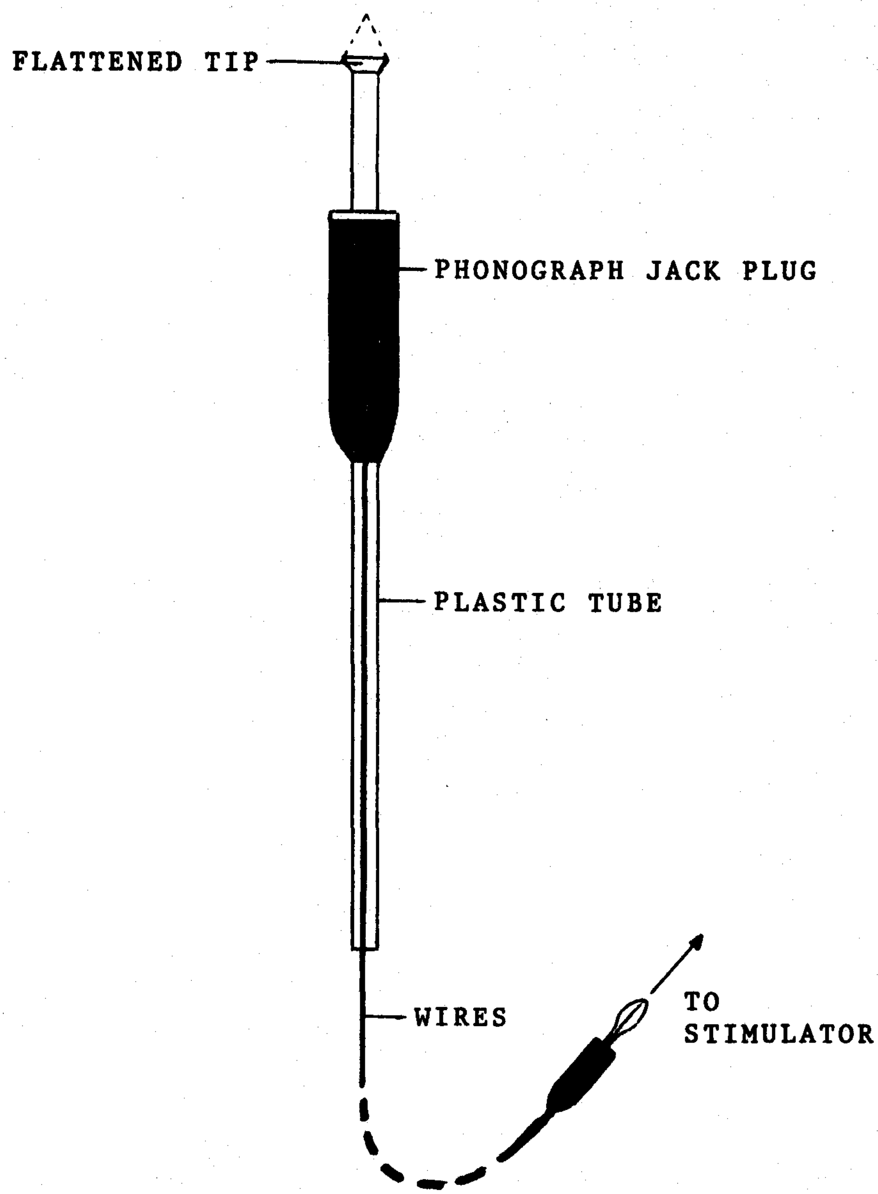
The indifferent electrode was fabricated in an identical manner, except that the sides measured two inches in length, thus having an area of four square inches.

4.2.2 Application of Surface Electrodes

Having located the motor point of the muscle, it is marked with a spot for easy identification later. At the location of the indifferent electrode and the motor point, EKG electrode paste is vigorously rubbed into the surrounding skin. This serves to improve conduction as it enters the pores and reduces the effective skin resistance. The rubbing also erodes the surface layer of the skin.

The conductive paste is rubbed into the interstices of the electrode mesh, in order to ensure that the current is spread to the entire area, rather than being restricted to the fine wire mesh. The electrodes are then secured to the skin, the smaller active electrode being placed directly over the spot which marks the location of the motor point, the larger indifferent electrode being secured over the skin, away from the region of the motor point (figure 4.5). Securing was done with surgical tape of one inch width, but this is not a restriction, as any effective means of securing may be used. Surgical tape proved to be the most convenient.

Once both electrodes have been secured to the skin, they are ready to be plugged into the stimulator. Electrode polarity is a factor with electrodes of different sizes. It was observed that connecting the indifferent electrode to the positive (red) terminal and the active electrode to the negative (black) terminal of the stimulator provided no pain at the active electrode site, whereas pain was observed here when the polarity was reversed. This is because in the latter arrangement, there is a relatively high current density at the active electrode due to the current entering the muscle. In the former case, the current enters the muscle by the indifferent electrode with a correspondingly lower current density. The current density is still the same value as in the reversed polarity, but with the current entering the indifferent electrode, the high value is reached gradually, rather than relatively instantaneously.



PROBE ELECTRODE

Figure 4.4

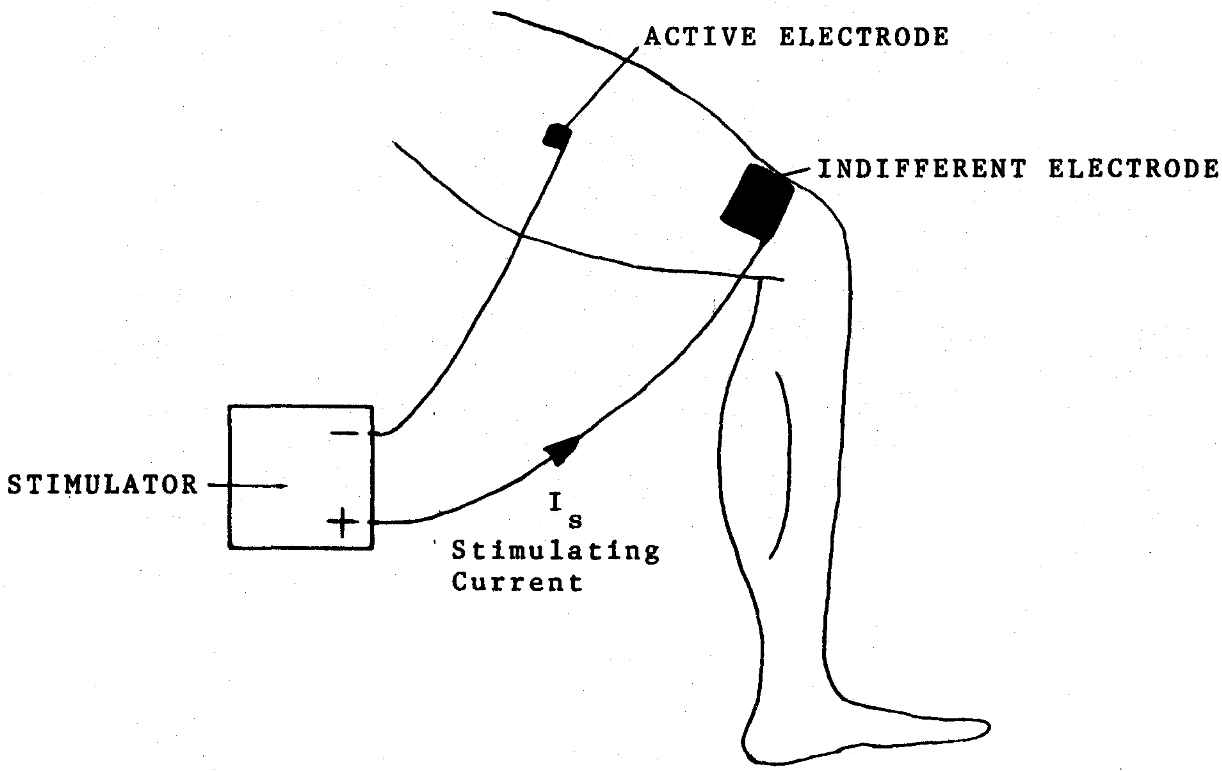

ACTIVE ELECTRODE
INDIFFERENT ELECTRODE
STIMULATOR
−
+
I_s
Stimulating
Current

Figure 4.5

4.3.1 Electro-goniometer

An electro-goniometer is a device by which the relative angular rotation of one limb section to an adjacent section may be measured, i.e. forearm and upper arm or, in this case, the lower leg with respect to the upper leg or thigh.

The heart of this device is a potentiometer, which also serves as the pivot point. The two arms of the goniometer are fitted to the potentiometer such that one arm may rotate relative to the other arm. The potent-iometer is then situated directly over the rotational axis of the joint and the arms are secured to both limb sections, along the longitudinal axis of the limbs. The technical drawing of the goniometer used in this study can be found in Appendix 2.

4.3.2 Circuitry

The circuit involved with an electro-goniometer is a very simple one, being essentially a voltage divider, (figure 4.6).

The input voltage is D.C., being supplied by a 1.5 Volt battery. A D.C. power supply may be used, but a battery is more convenient, as it allows more portability.

The output voltage of this circuit, by inspection, is seen to be;

$$V_o = \left(\frac{R_x}{R_1 + R_x}\right) V_i \qquad (4.f)$$

and may be measured on an oscilloscope, or, for a permanent record, it may be fed into a visicorder, which records the voltage fluctuation on a printout.

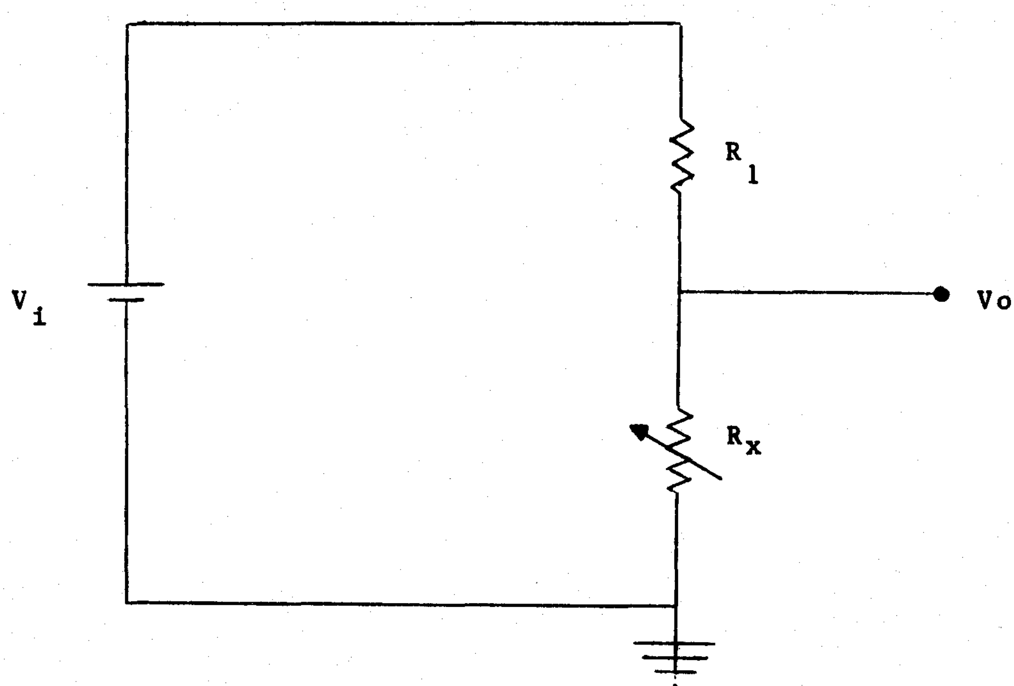


Figure 4.6

4.3.3 Calibration

The apparatus incorporated in the calibration of the electro-goniometer included a Tektronix 564 B storage oscilloscope, a Honeywell visicorder, and a plastic plate on which angles from 0^{o} to 180^{o} were scribed, in 10^{o} intervals. Two holes were drilled into this plate about the angular centre, in order to accomodate the potentiometer, whose maximum resistance was 10 KΩ

From the connector box, a BNC terminal was connected, first to the oscilloscope, to obtain a record of the voltage output and second, to the visicorder, to obtain a permanent record of the relation between angular rotation and visicorder deflection. The arrangement of the apparatus appears in figure 4.7.

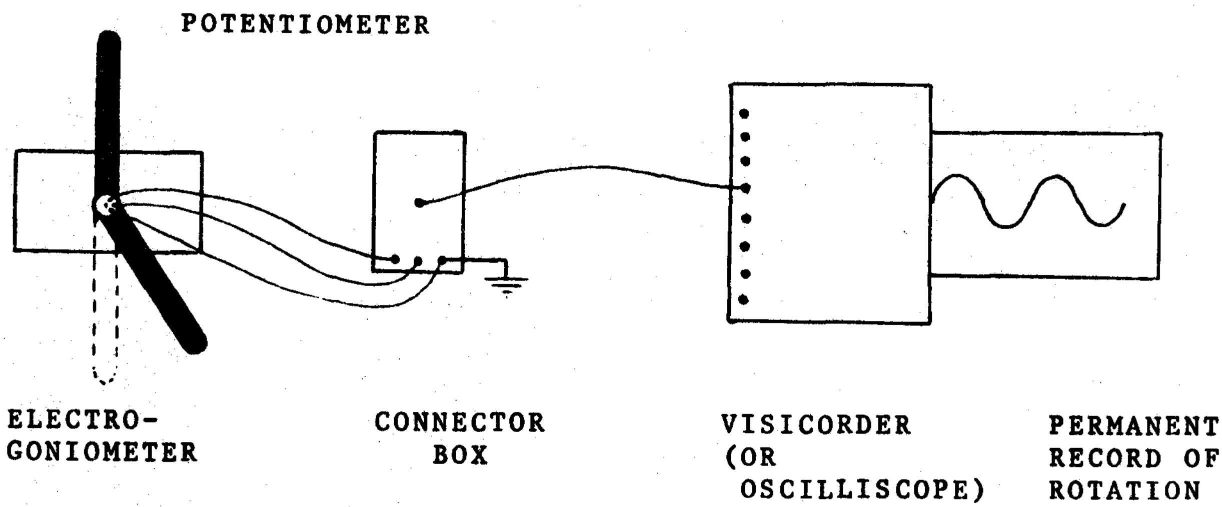


Figure 4.7

The goniometer was rotated from 0° to 180° in 10° intervals. At each interval, the deflection was recorded on the visicorder printout.

The sensitivity of the visicorder channel was set at its lowest value, 50 mV/cm. in order that maximal accuracy could be achieved. Having obtained the permanent record, each interval was measured manually, relative to the 0° line. The resultant data appear in table 4.1.

ANGLE (deg)	DEFLECTION (mm)
0	0
10	8
20	16
30	23
40	32
50	42
60	51
70	60
80	71
90	80
100	89
110	98
120	108
130	118
140	126
150	135
160	142
170	149
180	154

Table 4.1

GONIOMETER CALIBRATION - INITIAL DATA

The graph of these points appears in figure 4.8. Since linearity of this curve was sought, and observing the non-linearity near the end points, it was decided to consider only the linear region, i.e. from 30° to 150°, the range which corresponds to the maximum flexion angle of the lower leg. Hence, it was possible to perform a translation of the axes, such that the point (30°, 23mm) in the original data became (60°, 0 mm) on the translated graph. The operations performed to achieve this were;

$$\text{ANGLE} \rightarrow \text{ANGLE} + 30^{\circ}$$

$$\text{DEFLECTION} \rightarrow \text{DEFLECTION} - 23\text{mm}.$$

Thus, the tabulated data and resultant graph appear in table 4.2 and figure 4.9, respectively.

ANGLE (deg)	DEFLECTION (mm)
60	0
70	9
80	19
90	28
100	37
110	48
120	57
130	66
140	75
150	85
160	95
170	103
180	112

Table 4.2

GONIOMETER CALIBRATION - LINEAR REGION

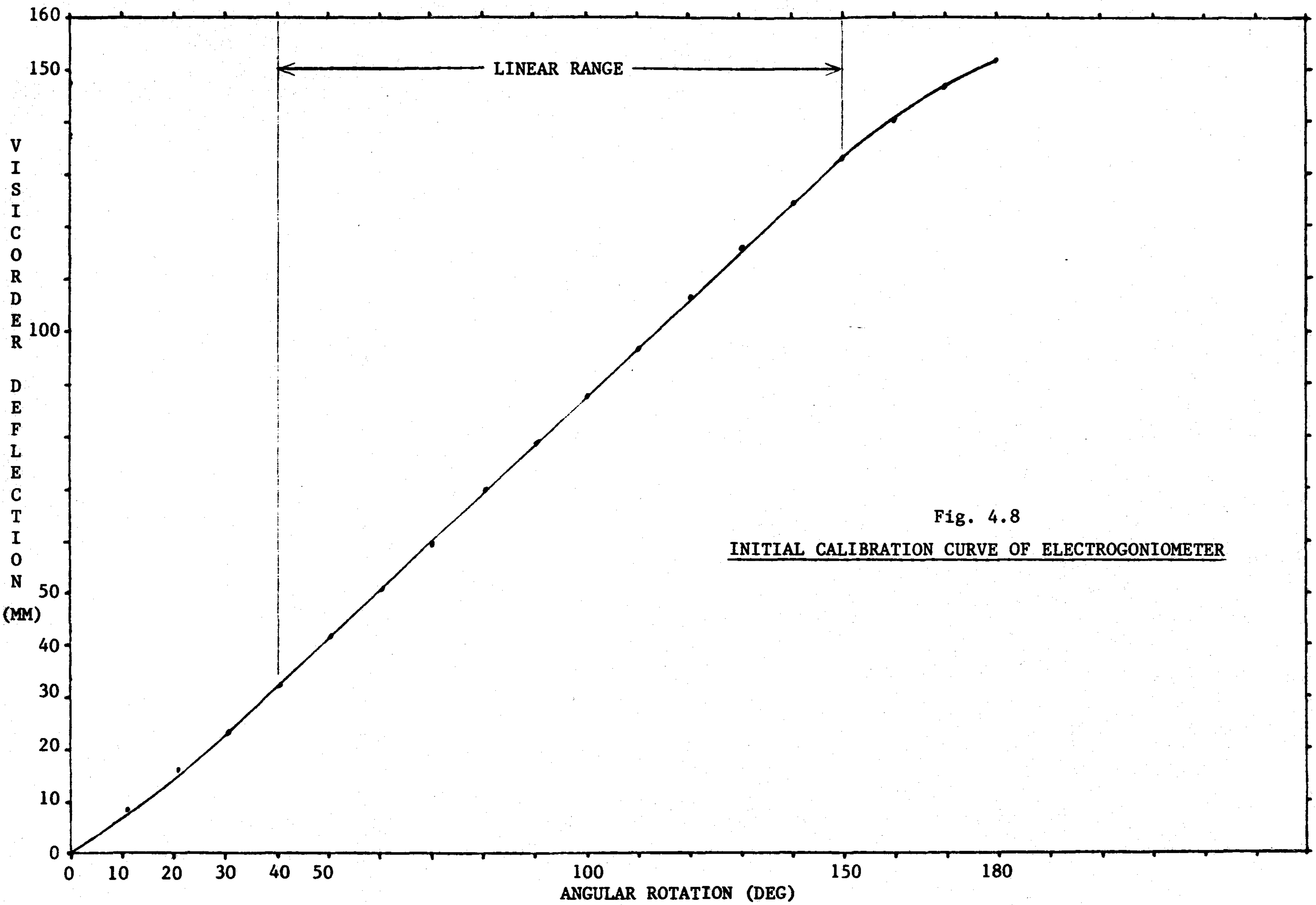


Fig. 4.8

INITIAL CALIBRATION CURVE OF ELECTROGONIOMETER

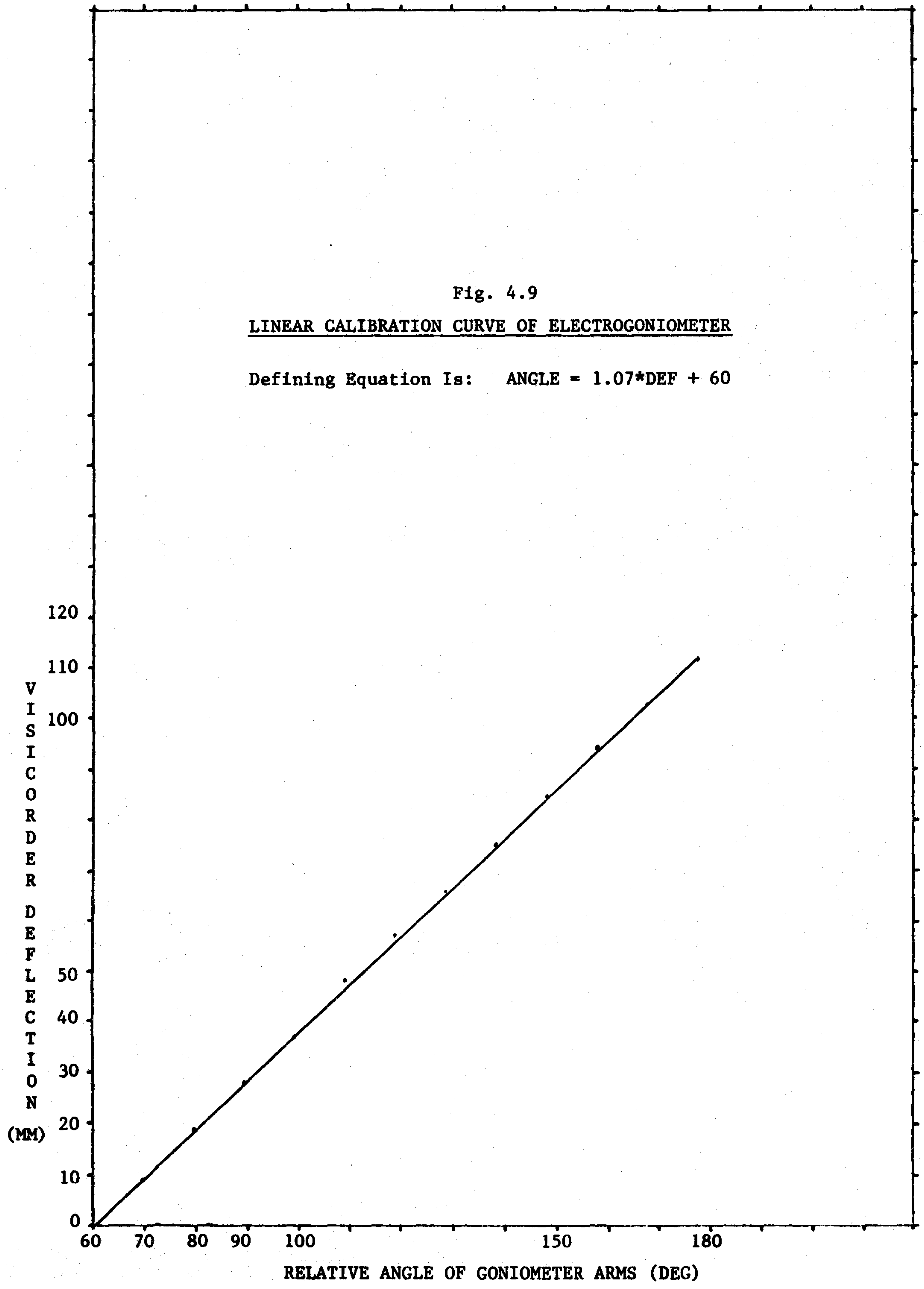


Fig. 4.9

LINEAR CALIBRATION CURVE OF ELECTROGONIOMETER

4.3.4 Attachment to Limb

The pivot point of the electro-goniometer, i.e., the potentiometer, is placed directly over the bony protrusion of the lateral femoral condyle. The arms of the goniometer are extended along the lengths of the lower and upper legs, and secured there by means of one inch wide elastic straps. Strips of velcro on the arms and on the ends of the straps enable the arms to be snugly secured to the leg such that no lateral movement is permitted during flexion and extension of the lower leg.

Hinges located on both arms permit the goniometer to follow the rotational cycle with no dislocation, as the hinge joints will fold during flexion and extend during extension. Thus, there is no extra force applied to the elastic straps, which would be the case if the goniometer arms were were a continuous strip of plastic.

4.4 Visicorder

This device, which was used to obtain a permanent record of the electro-goniometer response during the flexion-extension cycle, is a Honeywell 1858 CRT visicorder. It has eight separate channels to accept up to eight separate signals simultaneously.

It uses photo-sensitive recording paper, onto which an electron beam is deflected when a signal is introduced. To observe the recording, the paper must be exposed to light, whereupon, the recorded signal becomes a dark line relative to a lighter background. This permanent record does not fade appreciably over long time periods.

CHAPTER 5

EXPERIMENTAL PROCEDURES

5.1.1 Discussion of Experimental Objectives

Basically, the objectives of this research endeavour are twofold; to obtain an angular variation of the Quadriceps muscle force about the knee joint during voluntary and electrically induced rotation in the saggital plane against gravity, and to deduce any correlation between the forces during volitional and stimulated knee extension.

To this end, a basic equation of torque has been incorporated,

$$T = I\ddot{\theta} + MgL\sin\theta \qquad (5.1)$$

where,

T = torque exerted about knee joint.

I = inertial moment of lower limb (leg & foot)

M = Mass of lower limb (leg and foot)

g = gravitational constant

θ = angle between lower limb and thigh

$\ddot{\theta}$ = second time derivative of θ

L = location of centre of mass of lower limb as measured from knee joint centre.

Examining the right hand side of equation 5.1, one can see that it is a combination of two components. The first term, $I\ddot{\theta}$, is the inertial torque, and is the product of the inertial moment and angular acceleration. The second term is the gravitational component, and is the product of a constant term, MgL, and the sine of the angle between the longitudinal axis of the lower limb with respect to the vertical.

To obtain the exerted force about the knee, one merely divides the total torque by the moment arm of the patellar tendon about the rotational axis of the knee. This will be presented in section IV.5, incorporating G.L. Smidt's[15] data on the patellar tendon moment arm during knee extension. It will also be shown how each force component (inertial and gravitational) dominates one portion of extension and how they combine to produce the total force curve as a function of knee extension angle.

5.1.2 <u>Assumed Parameters</u>

A number of assumptions have been considered in this research which are necessary for simplification of procedure and analysis.

The assumptions are listed below;

1) The knee joint is considered as a simple hinge, in that it has only one axis of rotation, that being an imaginary line piercing the central medial and lateral protrusions of the femoral condyles.
2) The axis of rotation is considered stationary for all angles of knee rotation, the location of which was given in the first assumption.
3) The centre of mass of the lower leg does not deviate from its value, measured when the leg is horizontal, upon elevation at any angle to the horizontal.

Under these conditions, the following experiments were performed:

5.2 Volitional Extension of Lower Leg

5.2.1 Experimental Procedure

The subject was dressed in a pair of gym shorts, with his shoes and socks removed. In the standing position, the central protrusion of the lateral femoral condyle of the left leg was located, over which the potentiometer of the electro-goniometer was situated. The arms of the goniometer were extended along the longitudinal axes of the upper and lower leg, such that one arm was at an angle of 180° relative to the other, in the standing position.

The subject was then seated on the edge of a table, with his lower leg relaxed and extending towards the floor. Padding was placed under the thigh, near the knee, such that the angle of the lower leg relative to the thigh was 90°.

The lead wires from the goniometer were connected to the terminals of the connector box, from which a BNC cable was connected directly to channel three on the visicorder. The sensitivity of this channel was set at .2 volts/division, which resulted in an easily observed deflection curve when extension of the knee occured. Having the calibration equation of the goniometer, the 90° flexion angle of the knee in the relaxed position could be readily checked and the thigh padding adjusted accordingly.

This being accomplished, the experiments could be initiated.

The goniometer signal was adjusted on the visicorder printout, such that when the knee was in the resting position (90° of flexion), the signal corresponded to one of the channel lines. Thus, initiation of the extension cycle could be easily observed.

The subject was allowed to practice extending his lower limb in preparation for recording on the visicorder. Knee extension was performed subject to metronome frequencies of 1.0 Hz and 1.5 Hz, by which one beat corresponded to full extension and the second beat to the initial rest position of 90° knee flexion. Thus, the extension cycle frequencies were .5Hz and .75Hz. respectively. With the subject alternately extending and relaxing his leg at each metronome frequency, the visicorder was turned on, at a recording speed of 1 in/sec, and the timing lines were set at .1 sec. intervals. Five separate cycles at each frequency were recorded on the visicorder printout under these conditions. The data were then analyzed.

5.2.2. Data Acquisition and Analysis

The electrogoniometer response on the visicorder printout was measured manually, using a ruler. The point of zero deflection was taken to be the point on the record corresponding to an angle of 60° on the goniometer, as this was the onset of linearity of the goniometer response, (see calibration graphs in section 4.3.3. All deflections were measured relative to this line.

The raw data, acquired in this fashion, appear in tables 5.1 and 5.2.

DEFLECTION (MM)

Time (sec)	D_1	D_2	D_3	D_4	D_5	TOTAL(mm)	Average(mm)
0.0	29	30	31	32	30	152	30.4
.1	37	38	38	38	37	188	37.6
.2	51	53	52	52	51	259	51.8
.3	67	70	67	67	68	389	67.9
.4	81	85	81	81	84	412	82.5
.5	92	95	92	92	94	465	93.0
.6	100	101	98	99	102	500	100.0
.7	106	106	105	106	108	531	106.2
.8	110	109	108	109	110	546	109.2

Table 5.1 - CYCLE FREQUENCY = .5Hz

DEFLECTION (MM)

Time(sec)	D_1	D_2	D_3	D_4	D_5	TOTAL	Average(mm)
0.0	31	30	31	32	31	155	31.0
.1	43	39	40	40	45	207	41.4
.2	62	58	55	58	62	295	59.0
.3	82	78	75	76	79	390	78.0
.4	94	94	90	90	93	461	92.2
.5	104	105	102	102	103	516	103.2
.6	111	111	111	108	109	550	110.0

TABLE 5.2 CYCLE FREQUENCY = .75Hz.

A computer program has been developed whereby manually measured deflections on the visicorder printout serve as the input and all pertinent results, angular rotation, angular velocity, angular acceleration, torque and force, are the output. The actual program statements appear in Appendix 2.

From this output listing, many plots are possible which show either time dependence or angular dependence of many parameters. Time dependence of angular rotation of the lower limb, and angular dependence of forces may be found in Chapter 6.

5.2.3 Discussion of Computer Program

The comments throughout the computer program should be self-explanatory, however, this discussion will focus on the details of the computations involved.

Following the DIMENSION and DATA statements, the next group of statements reads the measured deflections into the computer, the variable N corresponding to the number of points measured. This varies with extension frequency and duty cycle length. The first DO loop computes the angles corresponding to each deflection, from the calibration curve of the goniometer, obtained previously. Thus, discreet angles are obtained at equal time intervals of .1 sec. in this case. Since discreet values are of no use in the computations which follow, a polynomial which describes the time variation of the angular rotation must be obtained.

The statement CALL LESQ accomplishes this, with the degree of the polynomial as M. In this case, a polynomial of degree M=4 accurately describes the angular rotation, however it may be necessary to try different values of M in different instances. The output of the LESQ subroutine consists of the polynomial coefficients, beginning with the constant term in C(1), and ending with the fourth power coefficient in C(5).

The next group of statements calculates the angular rotation using the LESQ polynomials, and prints these values together with the actual discreet angles, as a checking procedure, to ensure that the degree M is correct.

The group of statements beginning with G=9.8, is the actual computation of all parameters involved. The parameters XCOM and XMASS are obtained using the centre of mass apparatus discussed in section 4.1, and will necessarily be different for each subject. In the DO loop, the co-efficients from the LESQ subroutine are incorporated together with the rule of derivatives whereby,

$$\text{if} \quad Y = X^n$$

$$\text{then} \quad \frac{dy}{dx} = nx^{n-1}$$

to obtain the first and second derivatives of the angular rotation of the knee. Angles and angular parameters are printed in degrees and radians to facilitate checking of values and/or plotting of results.

The remainder of the program are instructions to the computer pertaining to the spacing and printing of the total output.

Thus, a convenient computer program is available into which raw data (measured deflections) are read, and various parameters such as angular rotation, and derivatives, torques and forces are calculated and printed, for easy inspection.

A program listing can be found in Appendix 3.

5.3 Extension by Electro-stimulation of Quadriceps Group

5.3.1 Experimental Procedure

The acquisition of electrostimulation data was performed with two objectives. The first, was to determine any correlation in the Quadriceps muscle force when extension was confined to the extension cycle frequencies in the volitional portion, (.5Hz and .75Hz). The second was to determine any change in the Quadriceps force when the pulse frequency of the stimulation current was varied, from 50Hz to 80Hz.

5.3.2 Initial Procedure

To ensure significant data, full extension needed to be achieved during electrostimulation of the Quadriceps. This depended strongly on electrode geometry and the level of stimulation current.

The optimal electrode geometry occurred with the indifferent electrode over the rectus femoris on the upper thigh. Two active electrodes were used, one over the motor point of the vastus lateralis and one over the motor point of the vastus medialis. The active-indifferent electrode distances, from centre to centre, were 17 cm. for each, and the active-active electrode distance, from centre to centre was 10 cm.

The resulting knee extension was such that locking of the knee at full extension was achieved, using synchronous currents of different magnitudes, the lateral vasti requiring 36 mA and the medial vasti, 54 mA. Due to the current distribution in the muscle group, the motor points of the rectus femoris muscle were also stimulated, thus all three muscles were effectively responsive using only two active electrodes.

The inter-electrode distances, from centre to centre were 17cm. from the indifferent electrode to each of the active electrodes, and the stimulation performed under this geometry and current levels were very comfortable. No painful sensations were observed at either active electrode site, although the contractions were very strong.

With full extension being achieved with known current levels, further experimentation proceeded.

5.3.3 Variation of Duty Cycle

Duty cycle, as interpreted for the purposes of electrostimulation is the ON/OFF cycle of the stimulating current. In the ON phase, current is passed through the active electrode sites, and in the OFF phase the current is blocked and the muscle is allowed to relax. The electrode geometry and current levels described in section 5.3.2 were utilized.

The positioning of the knee was identical to that described in the volitional phase, in section 5.2.1.

Four separate runs were performed at cycle times of .667/.667, 1.0/1.0, 1.5/1.0, 2.0/1.0, each phase being measured in seconds, and recorded on the visicorder print-out. Note that the first two duty cycles correspond to the two extension frequencies in the volitional portion. The pulse train frequency was chosen to be 50Hz, guided by previous work.

Data acquired in this manner appear in tables 5.3 to 5.5 respectively.

TIME(sec)	AVERAGE * DEFLECTION (mm)
0.0	38.83
.1	35.88
.2	46.88
.3	67.80
.4	85.23
.5	95.96
.6	104.54
.7	111.52

Table 5.3 CYCLE TIME = .667/.667 (sec)

TIME(sec)	AVERAGE* DEFLECTION(mm)
0.0	30.25
.1	30.25
.2	40.17
.3	58.41
.4	75.31
.5	87.65
.6	94.35
.7	100.79
.8	106.15
.9	109.37
1.0	111.25

Table 5.4 CYCLE TIME = 1.0/1.0 (sec)

TIME(sec)	AVERAGE* DEFLECTION(mm)
0.0	30.25
.1	30.25
.2	35.88
.3	46.61
.4	58.95
.5	69.68
.6	78.26
.7	83.36
.8	86.31
.9	90.60
1.0	95.16
1.1	99.72
1.2	102.94
1.3	105.35
1.4	107.23
1.5	109.64

Table 5.5 CYCLE TIME = 1.5/1.0 (sec)

* AVERAGE OF 6 SEPARATE CYCLES

5.3.4 Variation of Stimulation Pulse Train Frequency

Again, utilizing the same electrode geometry and stimulus currents as in the previous section (5.3.3) permanent records of the extension cycle were obtained for pulse train frequencies of 50Hz, 60Hz, 70Hz and 80Hz. The experimental arrangement of the leg was identical with the previous section, (5.3.3)

The overall cycle time for this experiment was arbitrarily chosen to be 1.0/1.0, and the resulting data appear in tables 5.6 to 5.9 respectively.

TIME(sec)	AVERAGE* DEFLECTION(mm)
0.0	32.93
.1	32.93
.2	43.12
.3	58.14
.4	74.50
.5	87.38
.6	95.69
.7	102.13
.8	106.96
.9	109.91
1.0	111.25

Table 5.6 - FREQUENCY = 50HZ

TIME(sec)	AVERAGE* DEFLECTION(mm)
0.0	32.40
.1	32.40
.2	43.66
.3	57.61
.4	70.48
.5	85.77
.6	93.28
.7	99.45
.8	106.42
.9	111.52
1.0	113.40

Table 5.7 - FREQUENCY = 60Hz.

TIME(sec)	AVERAGE* DEFLECTION(mm)
0.0	30.2
.1	30.2
.2	36.75
.3	49.74
.4	67.32
.5	79.96
.6	88.81
.7	96.05
.8	102.26
.9	107.55
1.0	111.8

Table 5.8 - FREQUENCY = 70Hz.

TIME(sec)	AVERAGE* DEFLECTION(mm)
0.0	29.9
.1	29.9
.2	35.37
.3	46.55
.4	64.01
.5	75.66
.6	86.60
.7	93.94
.8	98.59
.9	104.76
1.0	111.75

Table 5.9 - FREQUENCY = 80Hz.

5.3.5 Determination of Mass and Location of Mass Centre of Lower Leg

Using the centre of mass measuring device, discussed in section 4.1, the following results were obtained:

$$V_1 = 86\text{mV}$$

$$V_2 = 74\text{mV}$$

$$V_3 = 94\text{mV}$$

Translating these voltages into reaction forces, from the calibration curves,

$$R_1 = {}^{V}1/35.5 = 2.42 \text{ Kg.}$$

$$R_2 = {}^{V}2/33.7 = 2.19 \text{ Kg.}$$

$$\underline{R_3 = {}^{V}3/31.25 = 3.01 \text{ Kg.}}$$

$$\Sigma R_i = 7.62 \text{ Kg.}$$

Thus, the lower limb mass is,

$$M = \Sigma R_i - M_b$$

$$= 7.62\text{Kg.} - 2.0\text{Kg.}$$

$$= 5.62\text{Kg.} \qquad \rightarrow \qquad M = 5.62 \text{ Kg.} \pm .25\text{Kg.}$$

And the location of the lower limb mass centre is,

$$X_c = \frac{(R_3 - {}^{M}b/_2)L}{M}$$

$$= \frac{(3.01 - 1.00)\ 65.4}{5.62}$$

$$= 23.39 \text{ cm.} \qquad X_c = 23.39 \text{ cm.} \pm .93 \text{ cm.}$$

CHAPTER 6

EXPERIMENTAL OBSERVATIONS AND DISCUSSIONS

6.1 Data and Experimental Observations

6.1.1 Performance of Electrogoniometer

The electrogoniometer, discussed in section 4.3 operated as was anticipated. Once the pivot was positioned over the lateral femoral condyle, and the arms secured by elastic straps along the lower leg and thigh, the goniometer reproduced the extension cycle of the lower leg. No movement of the arms, relative to the respective limbs to which they were attached, was observed.

The hinges located on the arms of the goniometer, permitted the arms to bend in flexion, and straighten in extension of the knee. Thus, the arms did not deviate from their original position, nor did they fracture when the knee was in flexion.

The elastic straps which secured the goniometer to the leg in no way inhibited the motion of the knee during any phase of the experiments, and the pivot of the goniometer remained stationary, relative to the assumed rotational axis of the knee joint.

6.1.2 Experimental Observations During Stimulation

ELECTRODE GEOMETRY

An interesting point was observed during electrode geometry trials. In one case, the indifferent electrode was placed over the distal anterior portion of the thigh, with three active electrodes placed over motor points of the lateral and medial vasti and the rectus femoris muscles. The current was increased until an intolerable level was attained at the active sites, and the maximum extension of the knee was approximately 45^{o}.

In another case, the indifferent electrode was placed over the proximal anterior thigh, the distal edge of which was about 5 centimeters above the rectus femoris motor points, while those over the two vasti muscles were left intact. The current was again increased, and strong contractions of the vasti muscles were observed at a comfortable current level, (35 mA for the lateral and 54 mA for the medial vasti). In addition, the rectus femoris was also observed to be undergoing strong contractions, although no active electrode was present over either of its motor points. Current distribution in the rectus femoris must have been such that the threshold of its motor points was exceeded, due to the current being drawn by the two active vasti sites. This simplified the experiments considerably, since only two active electrode sites were required to stimulate the quadriceps muscle group.

At current levels of 36 mA and 54mA, locking of the knee at full extension was achieved, with no painful sensations at any electrode site. A pulse width of .2 msec. was chosen on the basis of the following discussion.

PULSE WIDTH

Pulse widths of .1, .2, and .5msec. were tested to determine an optimum value. The .2msec. width resulted in a strong muscular contraction of the quadriceps group, with no pain and full extension as stated above, using current levels of 36 and 54 mA.

With a pulse width of .5msec., the muscle contractions were noticeably stronger than with .2msec., and full extension was again achieved. However, at the same current levels as was used with .2msec., very painful sensations occurred at the active electrode sites. Although the pain was tolerable, it was extremely uncomfortable. One other sensation occurred during the periods in which current was being drawn. This can best be described as a sensation of "roughness" throughout the muscle, as though one could feel individual pulses of the stimulating current.

Since pulse widths of more than .2 msec. produced discomfort upon stimulation, a pulse width of less than .2 msec. was investigated, specifically, .1 msec.

In this case, no pain was felt whatsoever at the electrode sites, and during stimulation, the sensation of roughness was absent, and in fact, felt smoother than did the .2 msec. width. However, the current levels which produced strong contractions and good extension with .2 msec. pulse widths, resulted in noticeably weaker contractions and a correspondingly lesser extension with .1 msec. To achieve full extension, the current levels were increased, and although not actually measured, were easily 25% greater.

Thus, the decision to use pulse widths of .2 msec. resulted from a compromise between input current levels to the quadriceps and sensations of pain.

Qualitative Aspects of Quadriceps Contractions

It is interesting to compare the quadriceps contractions during the volitional and stimulation phases of the experiments.

During stimulation, the quadriceps was observed to contract much more violently than during volitional extension. The stimulated contractions appeared as though one was performing isometric contractions, in that the individual contracted muscle bulks were quite clearly visible, much more so than during volitional contractions.

One important difference between the volitional and stimulated contractions must be considered. Unlike volitional contractions, which are consciously controlled by somatic motor neurons from the central nervous system, the stimulated contractions are controlled by an external source, which by-passes these motor neurons. This fact has two implications which affect contractions.

During volitional contractions, only a sufficient number of motor units (motor neuron plus muscle cells) are recruited to cause the desired motion. This is generally referred to as the graded strength principle (16) whereby the contractile strength is proportional to the load. Secondly, under normal conditions, the contractile strength is proportional to the number of muscle fibres contracting simultaneously.

During stimulated contractions, however, this inherent feedback loop is not present, as contractions are produced by an external stimulation. Thus, only those muscle fibres whose motor end-plates comprise the motor point, or those excited as a result of the current distribution in the muscle, will contract. Consequently, the muscle contractions are relatively localized, in that the contraction is not due to the entire muscle contracting.

Latent Periods

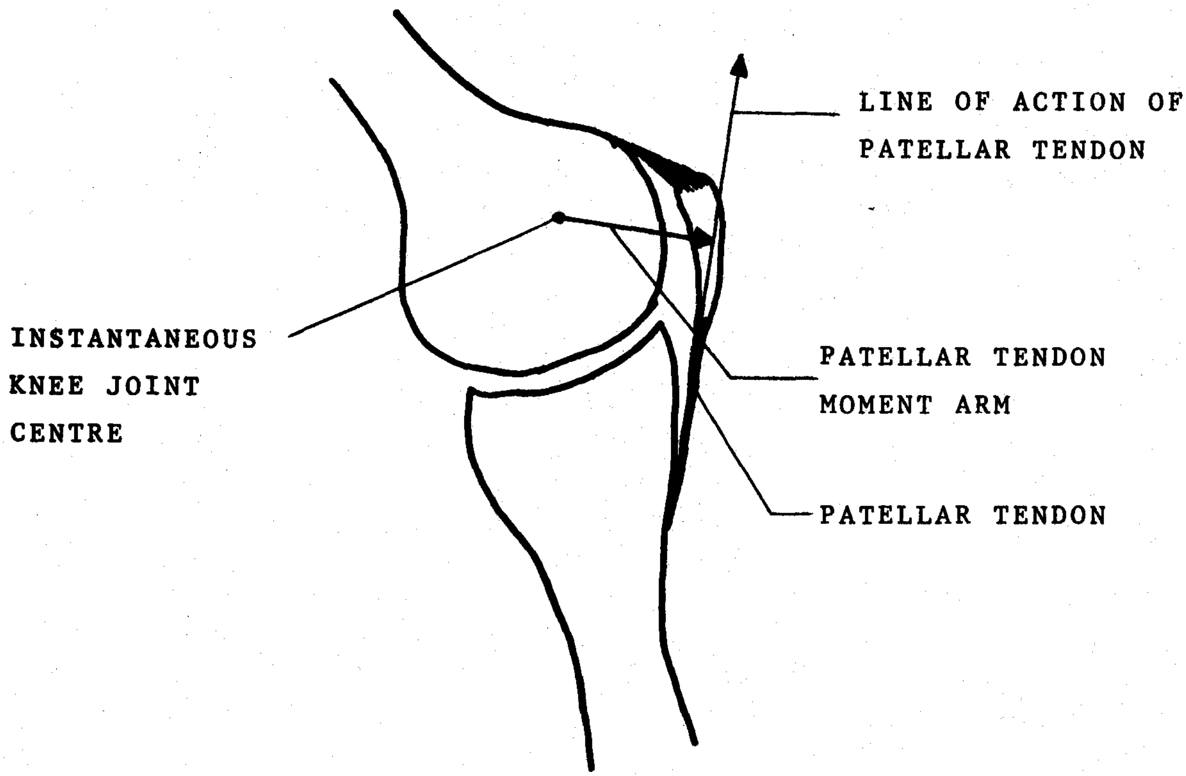
The latent period is defined as the length of time between the onset of stimulation of the muscle fibre and the onset of contraction of the fibre.

During stimulation experiments, it was possible to obtain an approximate numerical value for this parameter on a gross scale, since the stimulation signal was superimposed on the goniometer signal, on the permanent visicorder printout. By recording time markings at equal intervals of .1 sec., one could clearly observe the onset of the stimulation and the onset of gross muscular contraction resulting in extension of the knee.

The latent period of the quadriceps muscle group as a whole, was observed to be approximately 100 msec. This remained constant throughout all phases of the stimulation experiments, in that variation of the stimulus duty cycle or pulse train frequency did not alter this value significantly. In the case of the .667sec/.667sec. duty cycle, it was not possible to measure the gross latent period since the next stimulus duty cycle was initiated before the lower leg could attain the 90° flexion resting position. In all other cases, the latent period is clearly observable.

6.1.3 Patellar Tendon Moment Arm and Extension Angle

To obtain an order of magnitude figure for the force developed by the knee extensors, a value for the lever arm of the force must be obtained. This data has been acquired by G.L. Smidt et al. A brief summary of their techniques and results is presented here.



PATELLAR TENDON MOMENT ARM

Figure 6.1 (17)

They obtained X-ray films of twenty-six knees in the lateral position in order to measure several anatomical aspects, one of which was the length of the moment arms for flexion and extension of the knee. The films were exposed at 15° intervals, beginning with 90° flexion and terminating at full extension.

They considered the rotational axis of the knee to be the centre of the lateral femoral condyle. On each x-ray exposure, they measured the perpendicular distance from this axis to the patellar ligament, along which the extension force acts. Their measurements are presented in Table 6.1;

Knee angle (deg)	Mean Moment Arm (cm)	Stand. Dev.
90°	4.4	.51
105°	4.7	.51
120°	4.9	.53
135°	4.9	.46
150°	4.7	.48
165°	4.3	.44
180°	3.8	.50

Table 6.1
SMIDT'S DATA ON PATELLAR TENDON MOMENT ARM

These data are graphically depicted in figure 6.2.

Having obtained the torque as a function of time and knee angle, the force exerted by the quadriceps muscle group during extension is immediately available by introduction of the data in table 6.1.

As stated in section 2.1.3, the torque about an axis of rotation is the product of the force and the perpendicular distance from the rotational axis to the line of action of the force. This perpendicular distance is the moment arm.

To transpose the measured torque into force, the torque values are divided by the length of the moment arm at that particular knee angle.

Smidt's data were incorporated in a computer program which outputs the coefficients of the best-fit polynomial, by the method of least squares.

The resultant polynomial was a quadratic equation, i.e.:

$$\text{MOMENT ARM} = A_2\Theta^2 + A_1\Theta + A_0, \qquad \Theta = \text{KNEE EXTENSION ANGLE}$$

where

$$A_2 = 3.9153 \times 10^{-6}$$

$$A_1 = 1.1238 \times 10^{-3}$$

$$A_0 = 3.1548 \times 10^{-2}$$

With the knee extension angle measured in degrees, 180° corresponding to full extension, the moment arm is calculated in meters. The comparative data are given in table 6.2.

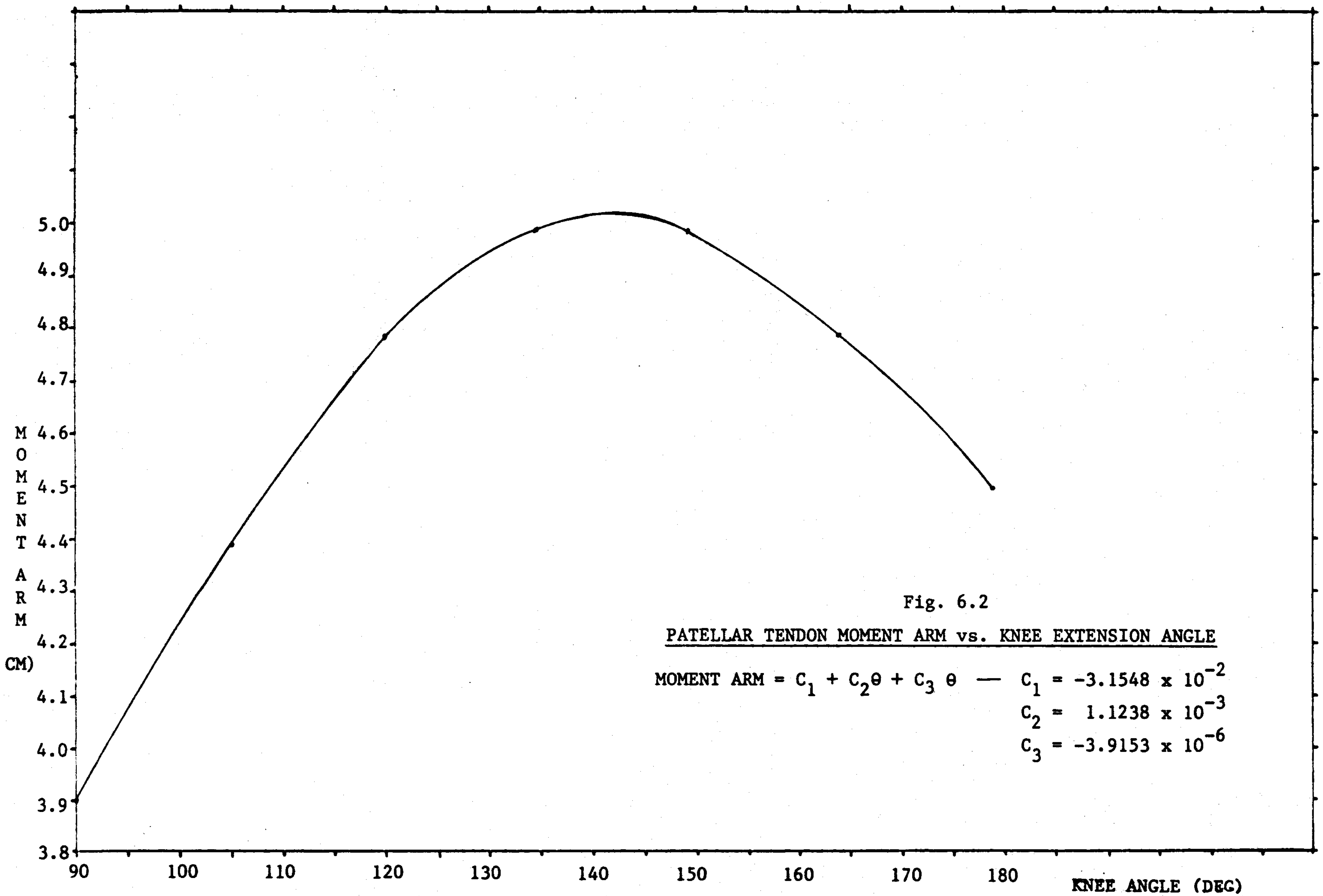


Fig. 6.2

PATELLAR TENDON MOMENT ARM vs. KNEE EXTENSION ANGLE

MOMENT ARM = $C_1 + C_2\theta + C_3\ \theta$ — $C_1 = -3.1548 \times 10^{-2}$

$C_2 = 1.1238 \times 10^{-3}$

$C_3 = -3.9153 \times 10^{-6}$

KNEE ANGLE (deg.)	(Y_d) DATA MOMENT ARM (m)	(Y_p) POLYNOMIAL MOMENT ARM (m)
90	.0380	.03788
105	.0430	.04328
120	.0470	.04693
135	.0490	.04881
150	.0490	.04893
165	.0470	.04728
180	.0440	.04388

Table 6.2

COMPARISON OF SMIDT'S DATA AND LEAST SQUARES ANALYSIS DATA

6.2 Volitional Forces Subject to Metronome Frequencies

The total quadriceps muscle forces are shown as a function of knee extension angle in figure 6.6, with metronome frequency as the varying parameter. In this discussion of the total force-angle variation, references will be made to the graphs of inertial force component (figure 6.5), and the gravitational force component, (figure 6.12).

Initially, the force is totally dominated by the inertial component, since the lower limb is originally at rest, and the gravitational force component approaches zero, since the angle and its sine of the of the lower leg relative to the vertical, both approach zero.

Thus, all of the force generated by the quadriceps, initially, is to overcome the inertia of the lower limb and begin the extension cycle.

The inertial force decreases rapidly at both metronome frequencies, while the gravitational component increases as $\sin\Theta$ with the vertical, such that at approximately 16^{o} into extension, the inertial and gravitational components are equal. From this point, as the gravitational component increases the inertial component continues to decrease, but at an ever-decreasing rate, attaining its least value at about 68^{o} into extension. For the remainder of the cycle, the inertial component increases at approximately the same rate as it decreased during the intial 16^{o}.

To understand why this latter increase occurs, one must consider the angular velocity during extension, (figure 6.4). The angles at which the inertial force becomes zero correspond to a maximum velocity at approximately 123^{o} of extension and a local minimum occurring at roughly 165^{o} extension. From 165^{o} to full extension at 180^{o}, the velocity increases slightly, and an increasing velocity constitutes an acceleration, which from basic physics, requires the application of a force. Hence the increase in the inertial force component during the latter 15^{o} of extension.

Addition of the two components to give the total force during extension results in figure 6.6. The domination of the inertial component in the initial phase of extension is easily observable. The slope of this total force curve begins to decrease as the gravitational component becomes increasingly dominant. The slope of the force curve increases slightly during mid-extension, as a result of a decreasing inertial force and an increasing gravitational force. This increase becomes more extreme during the final 15^{o} of extension, as the inertial force increases due to the increased velocity.

Consider the two volitional force curves, figure 6.6, in terms of actual events during extension.

Initially, the higher frequency extension, (.75Hz.) requires a greater force. Since the inertial component is completely dominant, this is attributable to the greater acceleration needed to initiate extension, and the resultant force is directly proportional to this acceleration. The magnitude of this difference in initial force between the two frequencies is approximately 60 Newtons.

As extension progresses, the higher frequency requires less force to maintain the rotation, since initially the greater force necessarily causes a greater angular velocity and hence a greater momentum. Thus, less force is required to maintain extension at the higher cycle frequencies.

The point at which the two total force curves intersect results from the two inertial forces becoming zero. This point corresponds to the maximum angular velocity in both instances, and is the point at which deceleration of the lower limb begins.

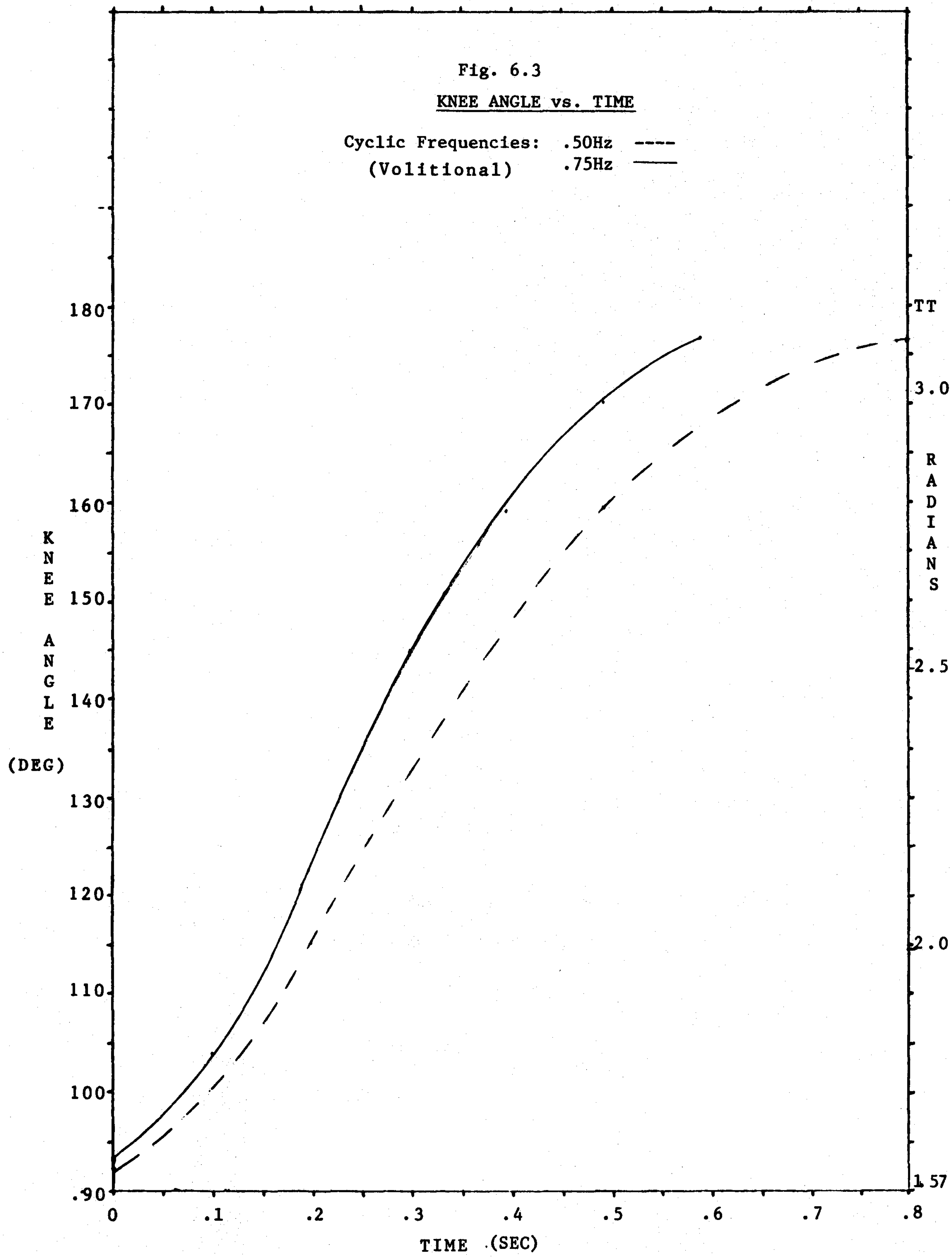
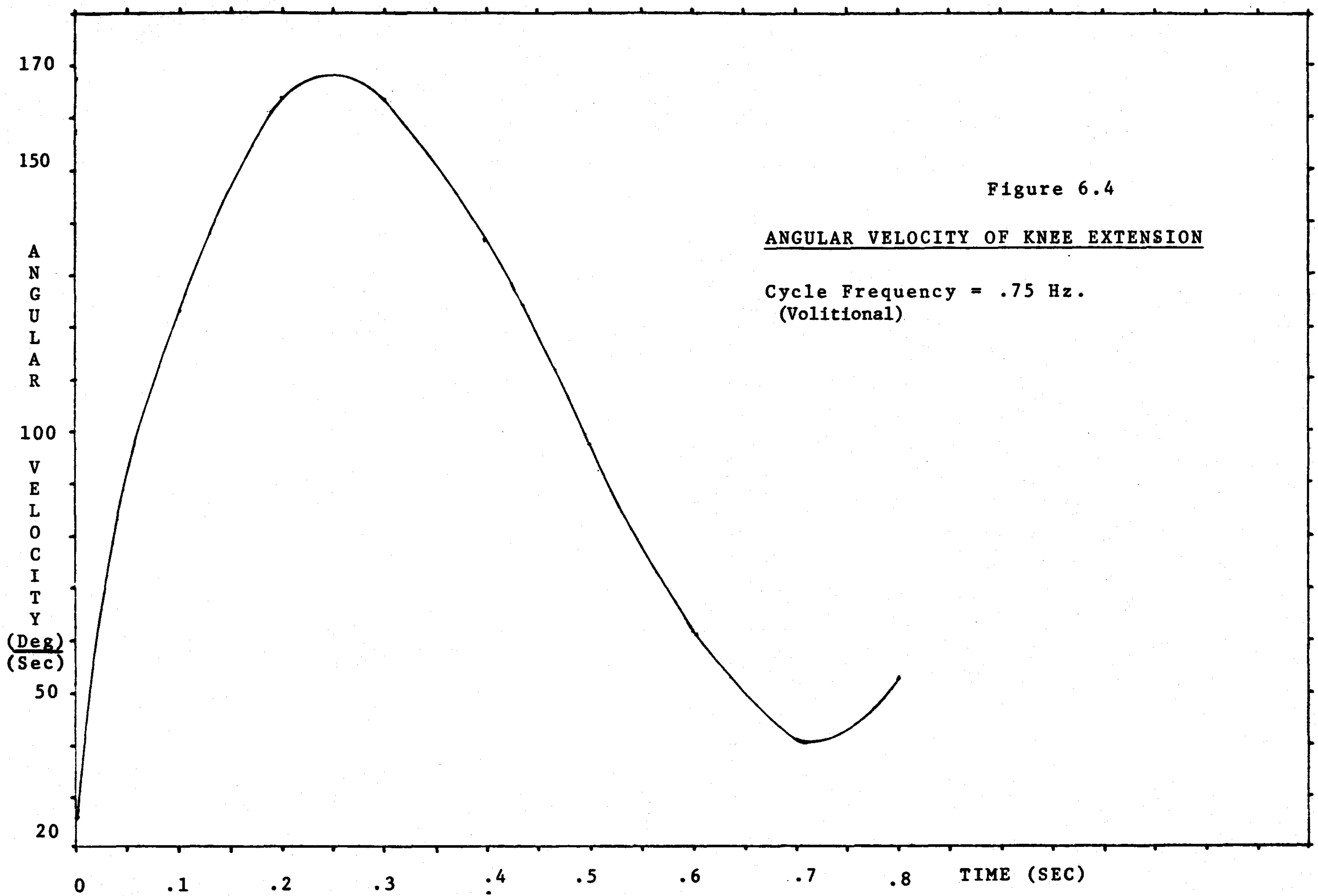


Fig. 6.3
KNEE ANGLE vs. TIME
Cyclic Frequencies: .50Hz ----
(Volitional) .75Hz ——
KNEE ANGLE (DEG)
180
170
160
150
140
130
120
110
100
.90
TT
3.0
2.5
2.0
1.57
RADIANS
0
.1
.2
.3
.4
.5
.6
.7
.8
TIME (SEC)



170
150
100
50
20
ANGULAR VELOCITY (Deg)/(Sec)
0
.1
.2
.3
.4
.5
.6
.7
.8
TIME (SEC)
Figure 6.4
ANGULAR VELOCITY OF KNEE EXTENSION
Cycle Frequency = .75 Hz.
(Volitional)

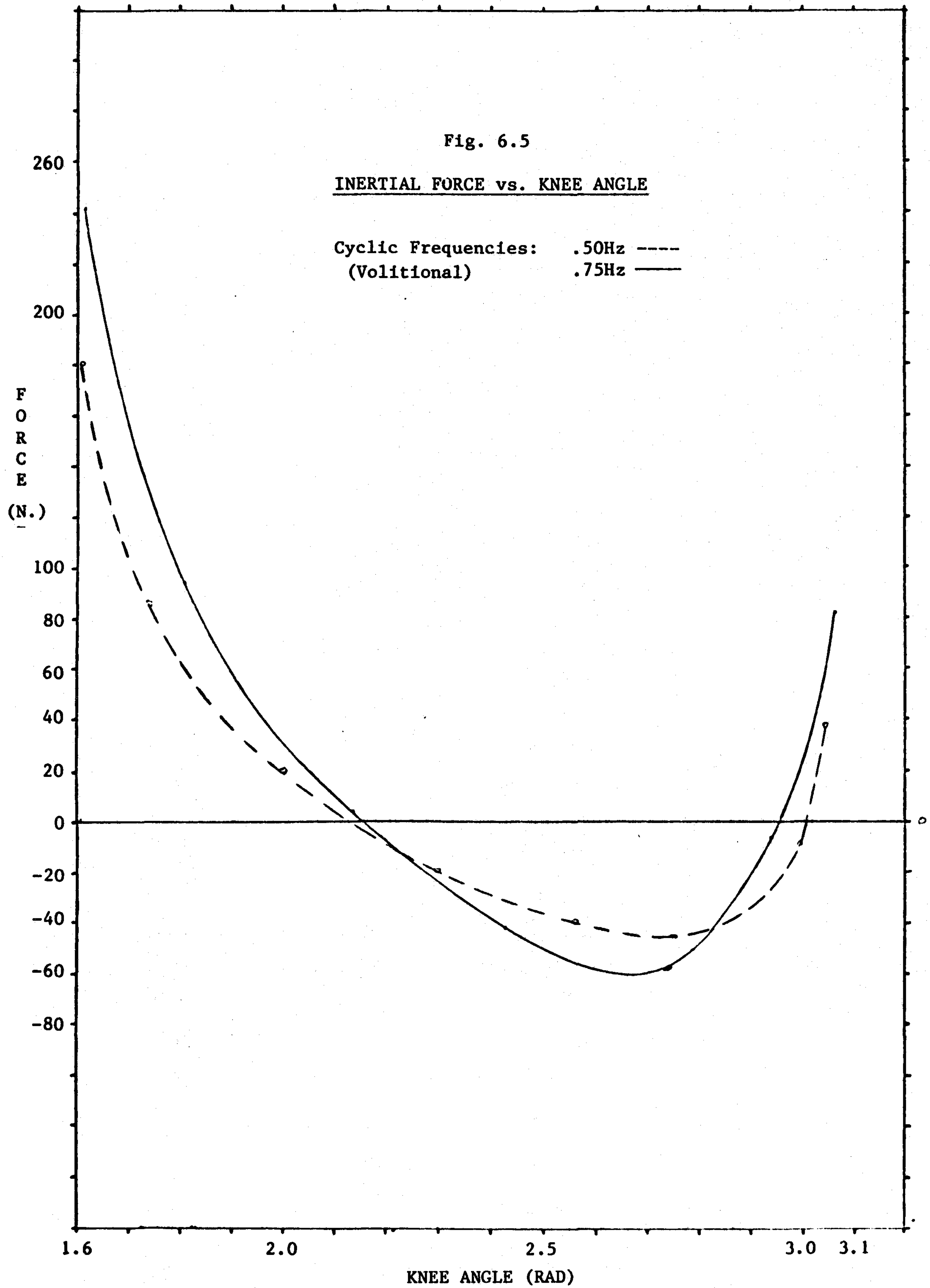


Fig. 6.5
INERTIAL FORCE vs. KNEE ANGLE
Cyclic Frequencies: .50Hz ----
(Volitional) .75Hz ——
FORCE (N.)
260
200
100
80
60
40
20
0
-20
-40
-60
-80
1.6
2.0
2.5
3.0
3.1
KNEE ANGLE (RAD)

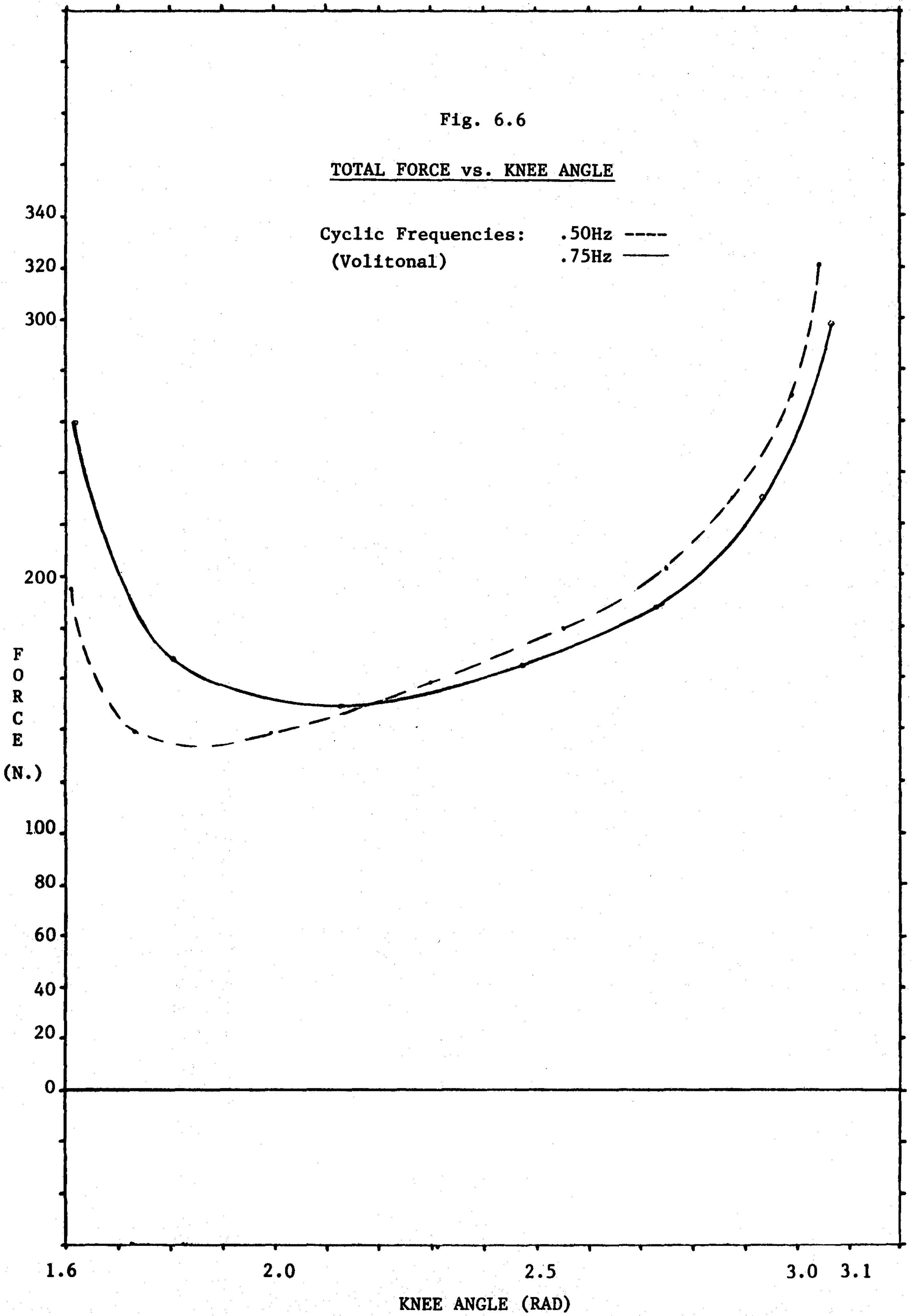


Fig. 6.6
TOTAL FORCE vs. KNEE ANGLE
Cyclic Frequencies: .50Hz ----
(Volitonal) .75Hz ——
340
320
300
200
100
80
60
40
20
0
FORCE
(N.)
1.6
2.0
2.5
3.0
3.1
KNEE ANGLE (RAD)

6.3 Stimulated Muscle Forces

6.3.1 Variation of Cycle Frequencies

The total force developed by the quadriceps is shown in figure 6.9, with cycle frequency being the changing parameter. In discussing these force curves, reference will again be made to inertial and gravitational components, figures 6.8 and 6.12 respectively.

The length of the cycle in this phase is the reciprocal of the frequencies in the volitional phase, so that the cycle length of .667 sec.+.667 sec. corresponds to the .75 Hz volitional and the cycle length 1.0 sec + 1.0 sec. corresponds to the .5 Hz volitional cycle.

The trend of the family of force curves in this case follows much the same pattern as in the volitional case. The force required to initiate knee extension is greatest for the shortest cycle (highest frequency of extension) and the initial force decreases as the duty cycle length increases, each successive force being approximately 100 Newtons less than the previous force.

Referring to the total force curves of figure 6.9, the domination of the inertial forces are apparent in the first phase of extension. Again, as in the volitional case, the mid-range of extension is dominated by the gravitational component, as the curves flatten or begin to increase.

There is a definite pattern developed in the family of total force curves of figure 6.9. As the cycle length increases (decreasing extension frequency), the change in the magnitude of the force becomes less. Comparing the two extreme cyclic curves, the one with the shortest cycle produces a sharp decrease in total force initially, with a flattening out in mid-extension before a final sharp increase. The longest cycle, however has increasing force from beginning to end of extension, at a relatively constant rate, although it is not exactly linear.

This pattern arises from the angular accelerations in all three cases. Referring to figure 6.8, the inertial forces, which are directly proportional to the angular accelerations, undergo less extreme changes in force magnitudes as the cycle length increases.

This is reasonable in physical terms since the lower limb must go from rest to full extension in increasing lengths of time in the three instances, constrained as such by the cycle times set by the stimulator.

The best correlation in the total force curves between volitional extension and stimulated extension with varied cycles, occurs with the .5Hz volitional curve and the 1.0/1.0 cycle time curve. These two curves are plotted together in figure 6.14 to facilitate examination.

These curves produce an extremely close correlation except at the beginning of extension. Initially the force produced as a result of stimulation is somewhat less than that produced voluntarily, by approximately 40 Newtons. This could be due to use of surface electrodes, in that they excite the muscle less efficiently than does the central nervous system. However, this difference quickly disappears, as the two force curves are virtually identical from approximately 20° of extension to full extension.

The volitional force curve of .75 Hz. frequency and the force curve for .667/.667 duty cycle stimulation can not be realistically correlated since the differences in force magnitudes are much greater in this case.

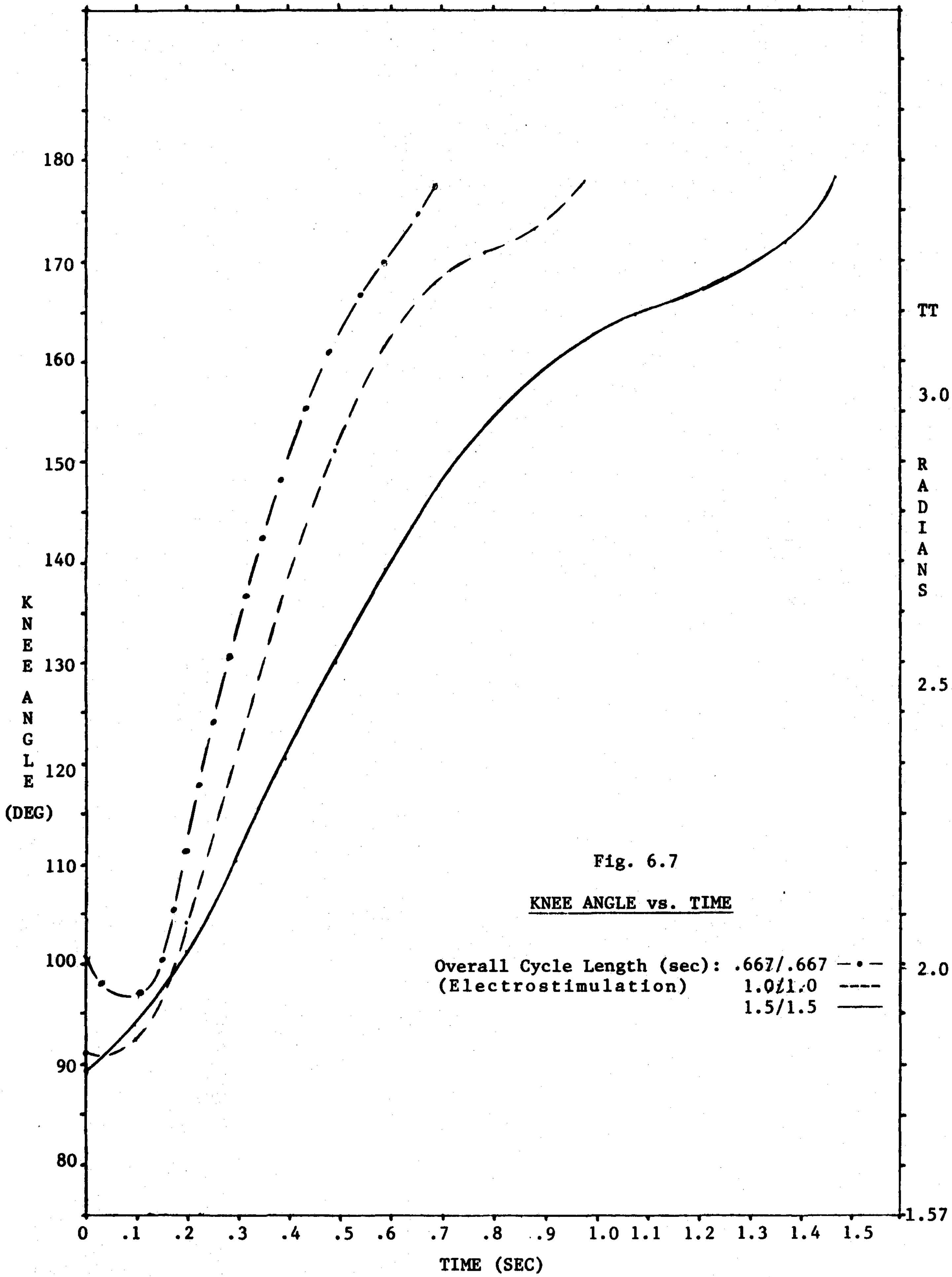


Fig. 6.7

KNEE ANGLE vs. TIME

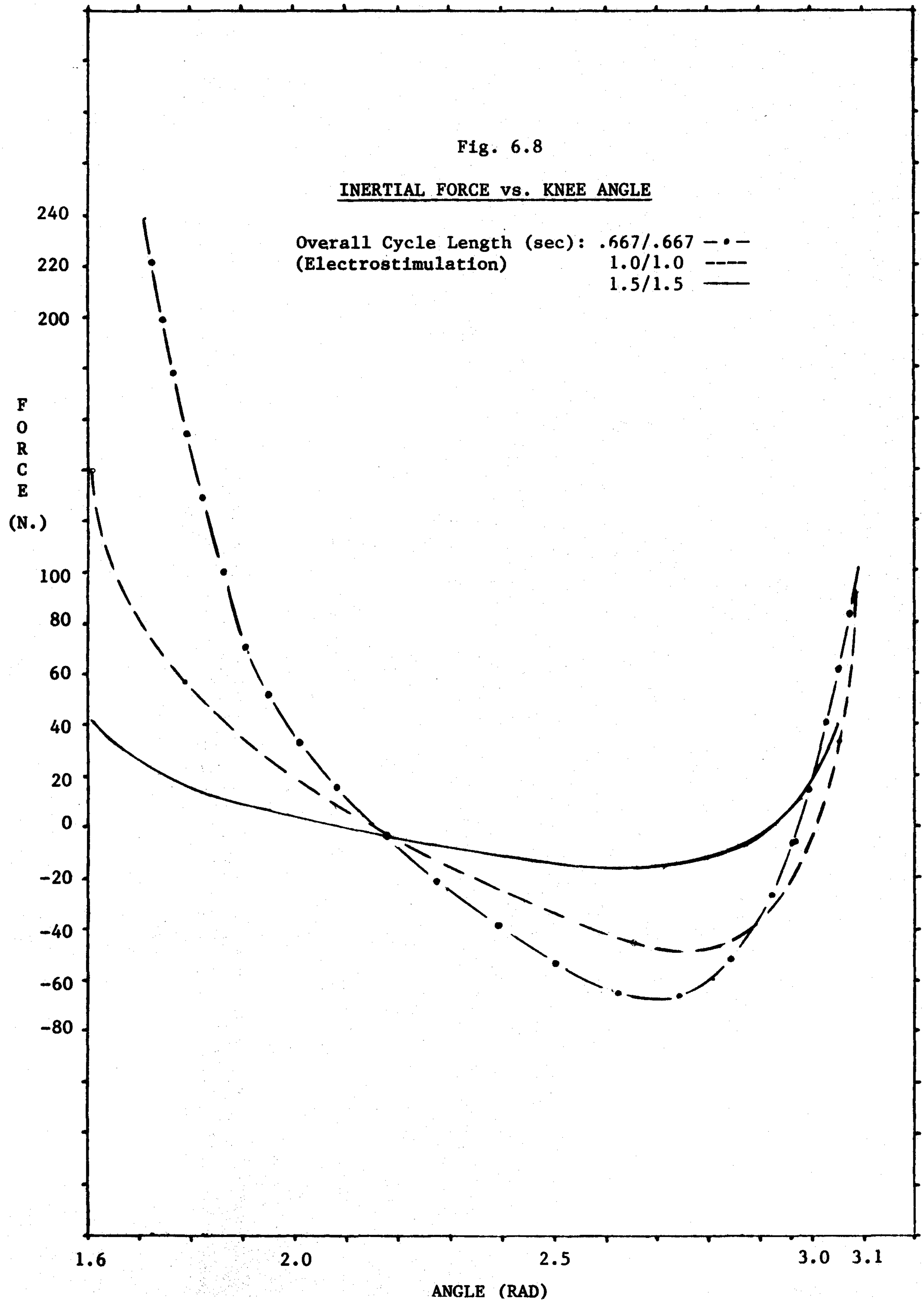


Fig. 6.8
INERTIAL FORCE vs. KNEE ANGLE
Overall Cycle Length (sec): .667/.667
(Electrostimulation)
1.0/1.0
1.5/1.5
240
220
200
100
80
60
40
20
0
-20
-40
-60
-80
FORCE (N.)
1.6
2.0
2.5
3.0
3.1
ANGLE (RAD)

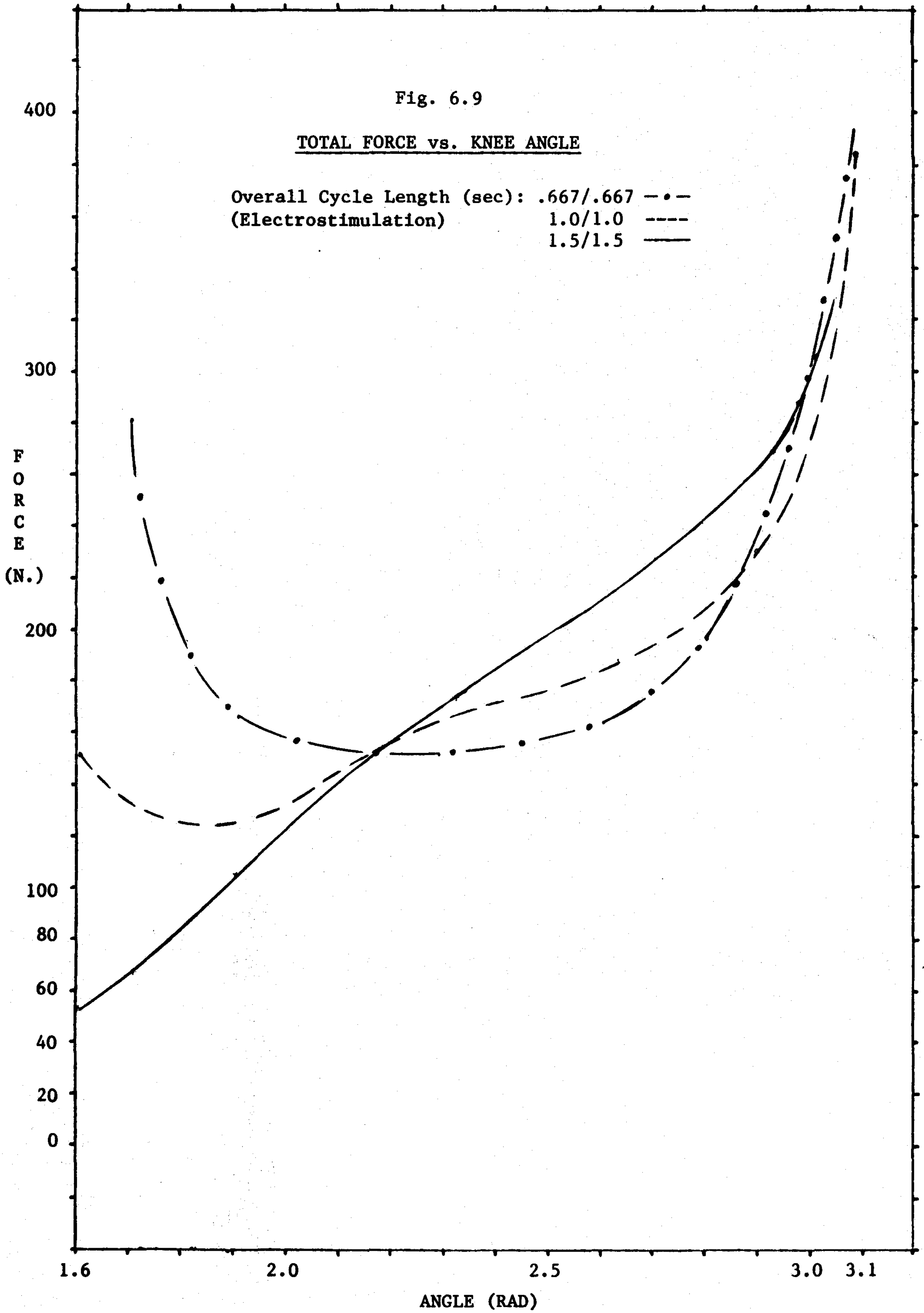


Fig. 6.9
TOTAL FORCE vs. KNEE ANGLE
Overall Cycle Length (sec): .667/.667
(Electrostimulation)
1.0/1.0
1.5/1.5
FORCE (N.)
400
300
200
100
80
60
40
20
0
1.6
2.0
2.5
3.0
3.1
ANGLE (RAD)

Although the two curves are roughly the same shape, they do not approach the degree of correlation of the two previously discussed force curves.

6.3.2 Stimulated Forces Subject to Varying Pulse Train Frequencies

Curves of the inertial force components due to varying of the pulse train frequency of the stimulating current appear in figures 6.11 and 6.13 respectively.

Contrary to the volitional and cycle time cases, there are no marked differences in either the inertial force component curves or the total force curves as they follow virtually the same patterns. There are only two aspects worthy of mention regarding the total force curves in this case, and they both pertain to the pulse train frequency of 60 Hz.

Initially, the force produced by the trains at a frequency of 60 Hz to begin extension is approximately 20 Newtons less than that produced by the other three pulse train frequencies, (fig. 6.13). Secondly, the 60 Hz trains appear to produce a less dramatic rise in the exerted force throughout extension, in that it is the only force curve which is relatively constant over the initial 15° of extension.

However, since one of the experimental objectives was to obtain a correlation between volitional and stimulated forces, the 50 Hz frequency must be given preference. It provides the best correlation between itself and the volitional force curve of .5 Hz cyclic frequency, (fig. 6.14), although the differences in the total forces during extension with different pulse train frequencies are negligible in comparison to those derived from changing the cycle times.

As stated in section 5.3.4, the overall cycle time for all of the above pulse train frequencies was maintained at 1.0/1.0 sec. The stimulating currents and pulse widths were kept constant, at 36mA and 54mA for the lateral and medial vasti respectively, with all pulse widths being .2 Msec. Electrode geometry was common to all phases of the electrostimulation experiments.

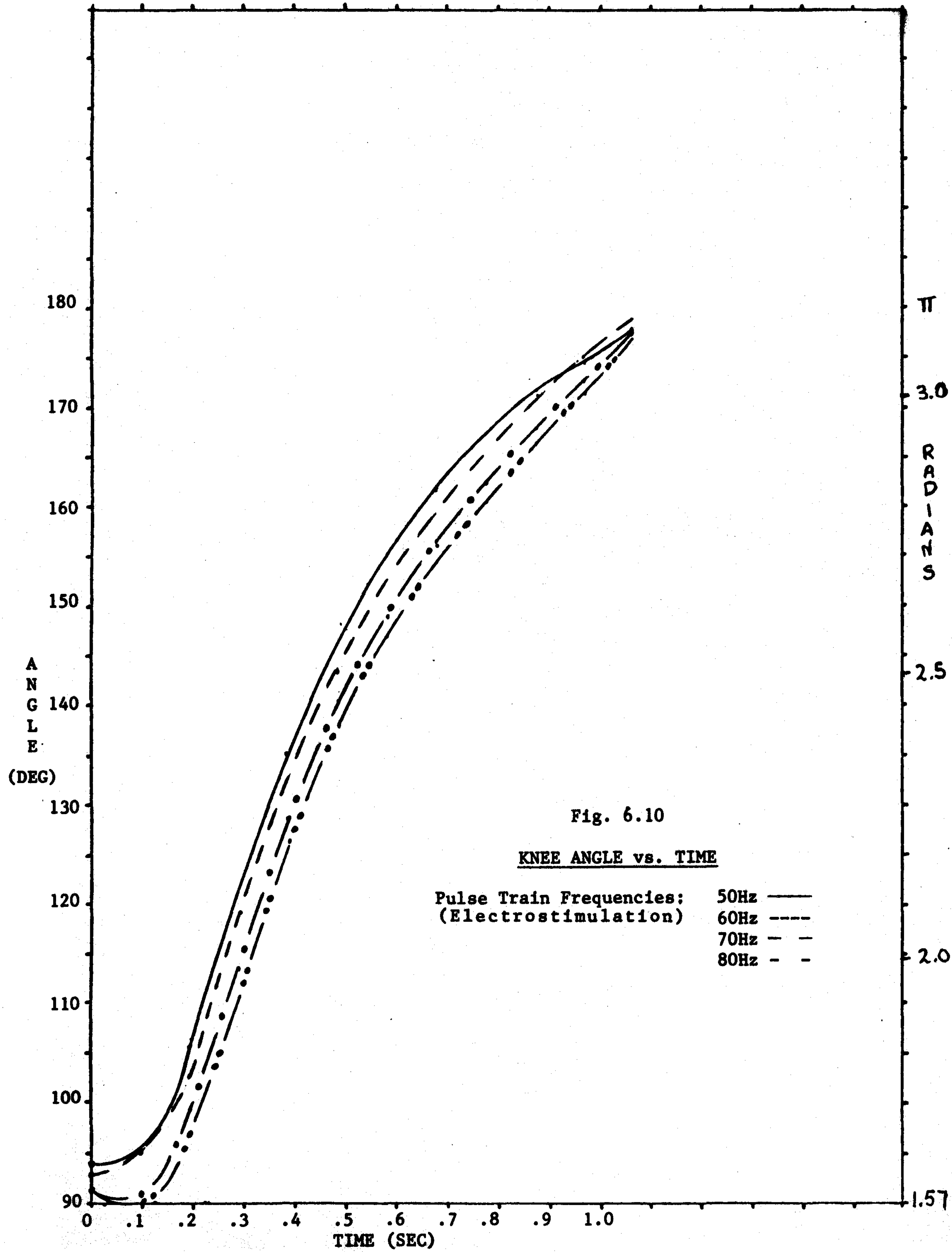


Fig. 6.10

KNEE ANGLE vs. TIME

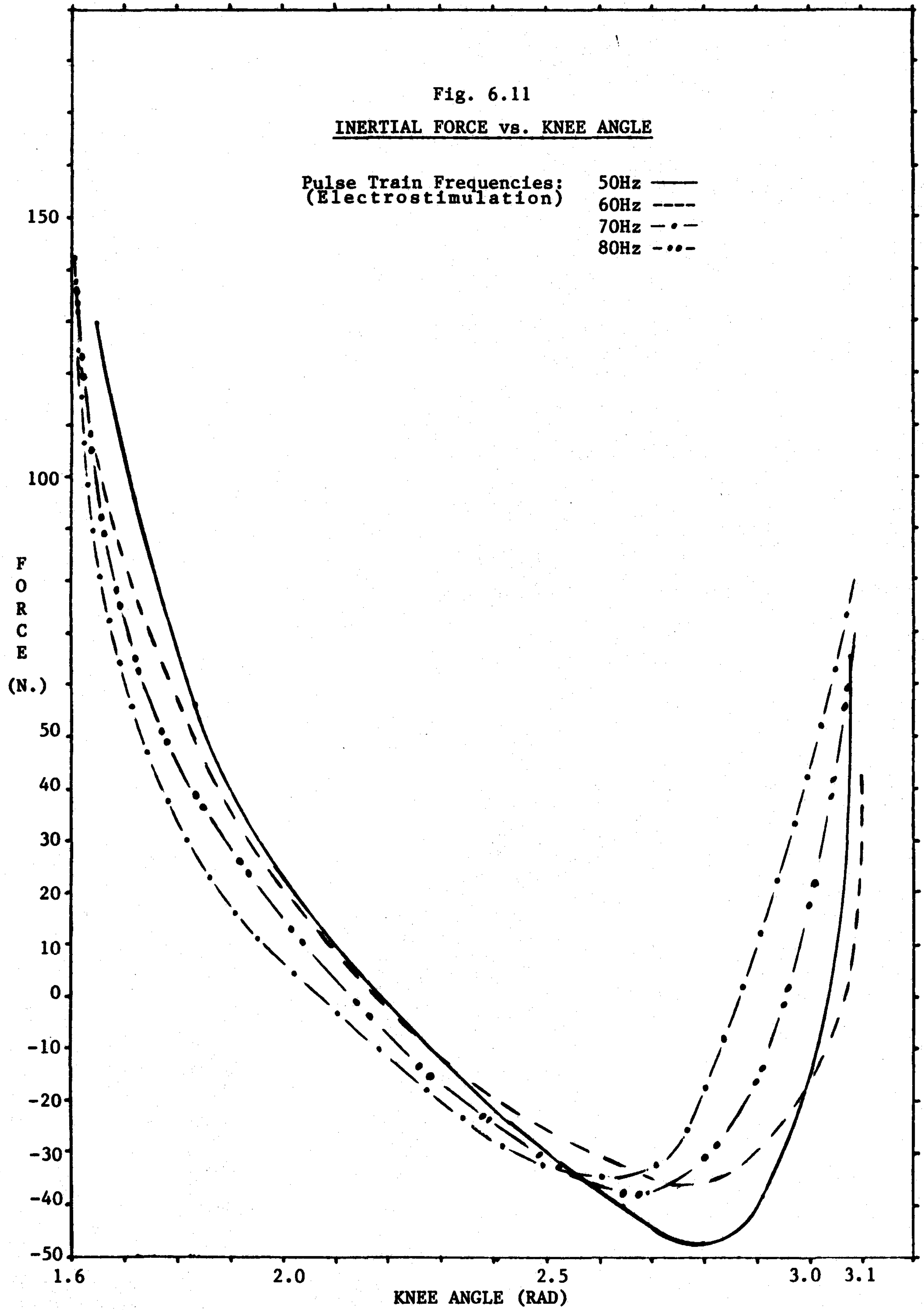


Fig. 6.11
INERTIAL FORCE vs. KNEE ANGLE
Pulse Train Frequencies:
(Electrostimulation)
50Hz
60Hz
70Hz
80Hz
150
100
FORCE
(N.)
50
40
30
20
10
0
-10
-20
-30
-40
-50
1.6
2.0
2.5
3.0
3.1
KNEE ANGLE (RAD)

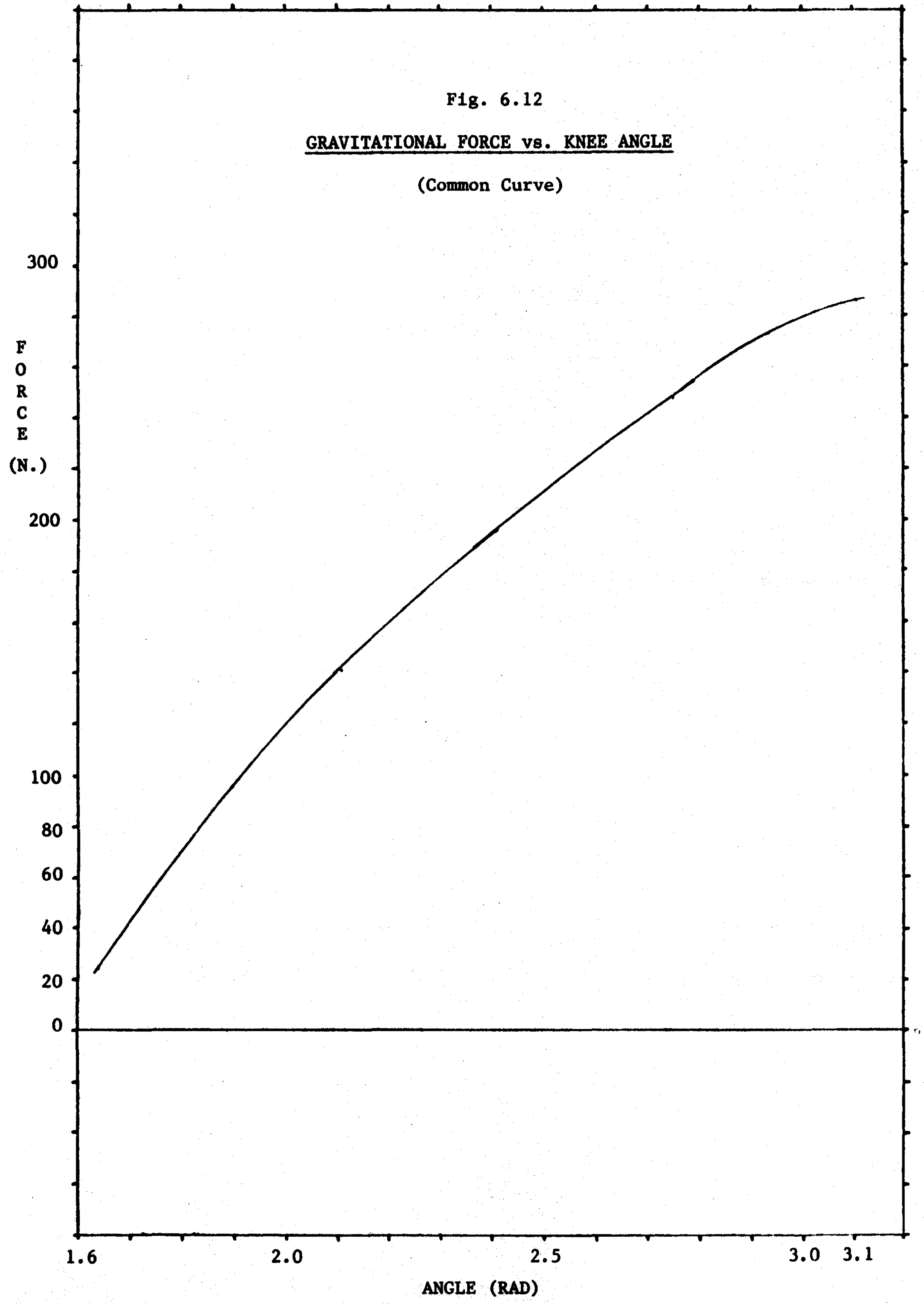


Fig. 6.12
GRAVITATIONAL FORCE vs. KNEE ANGLE
(Common Curve)
FORCE (N.)
300
200
100
80
60
40
20
0
1.6
2.0
2.5
3.0
3.1
ANGLE (RAD)

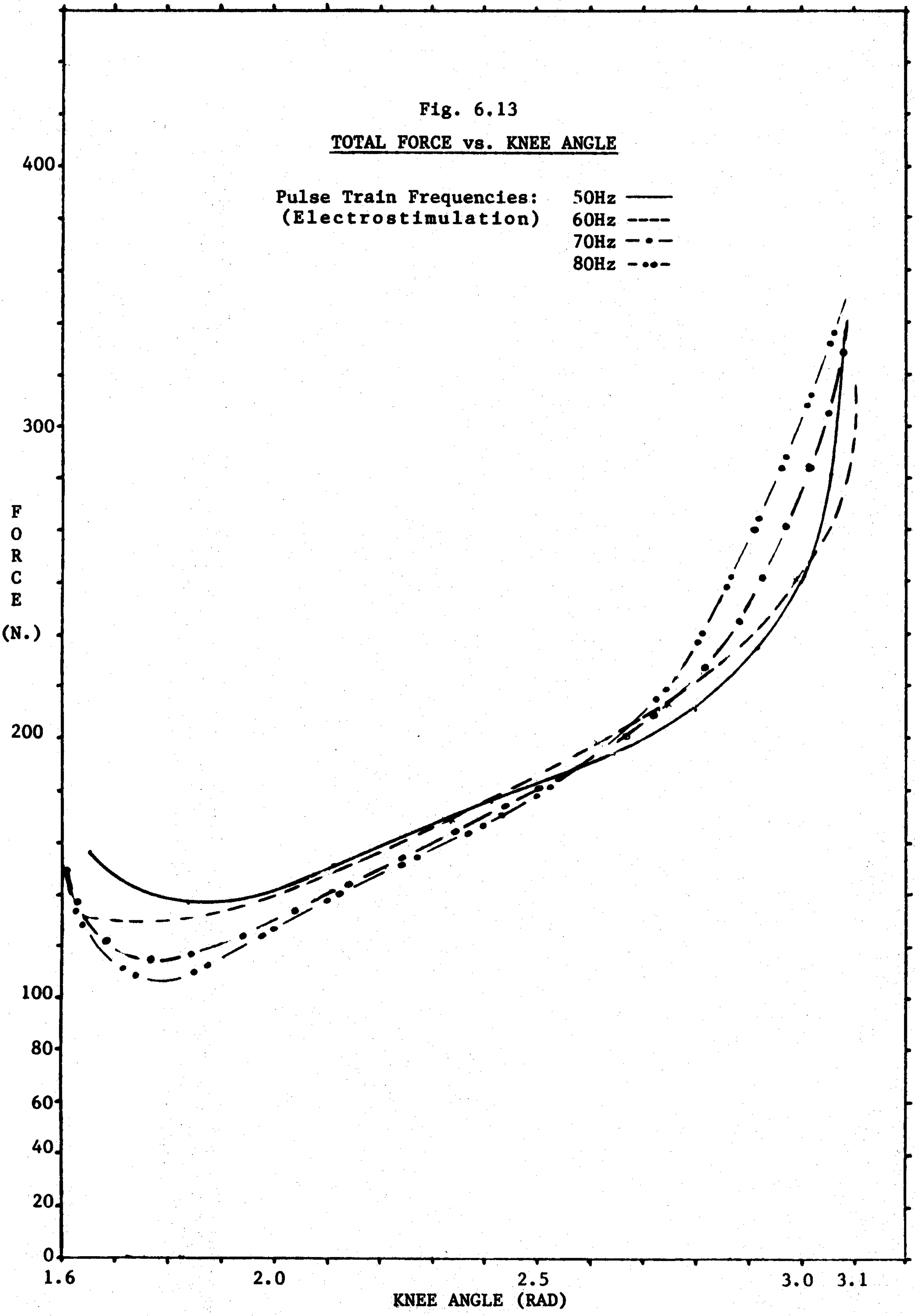


Fig. 6.13
TOTAL FORCE vs. KNEE ANGLE
Pulse Train Frequencies:
(Electrostimulation)
50Hz
60Hz
70Hz
80Hz
400
300
200
100
80
60
40
20
0
FORCE (N.)
1.6
2.0
2.5
3.0
3.1
KNEE ANGLE (RAD)

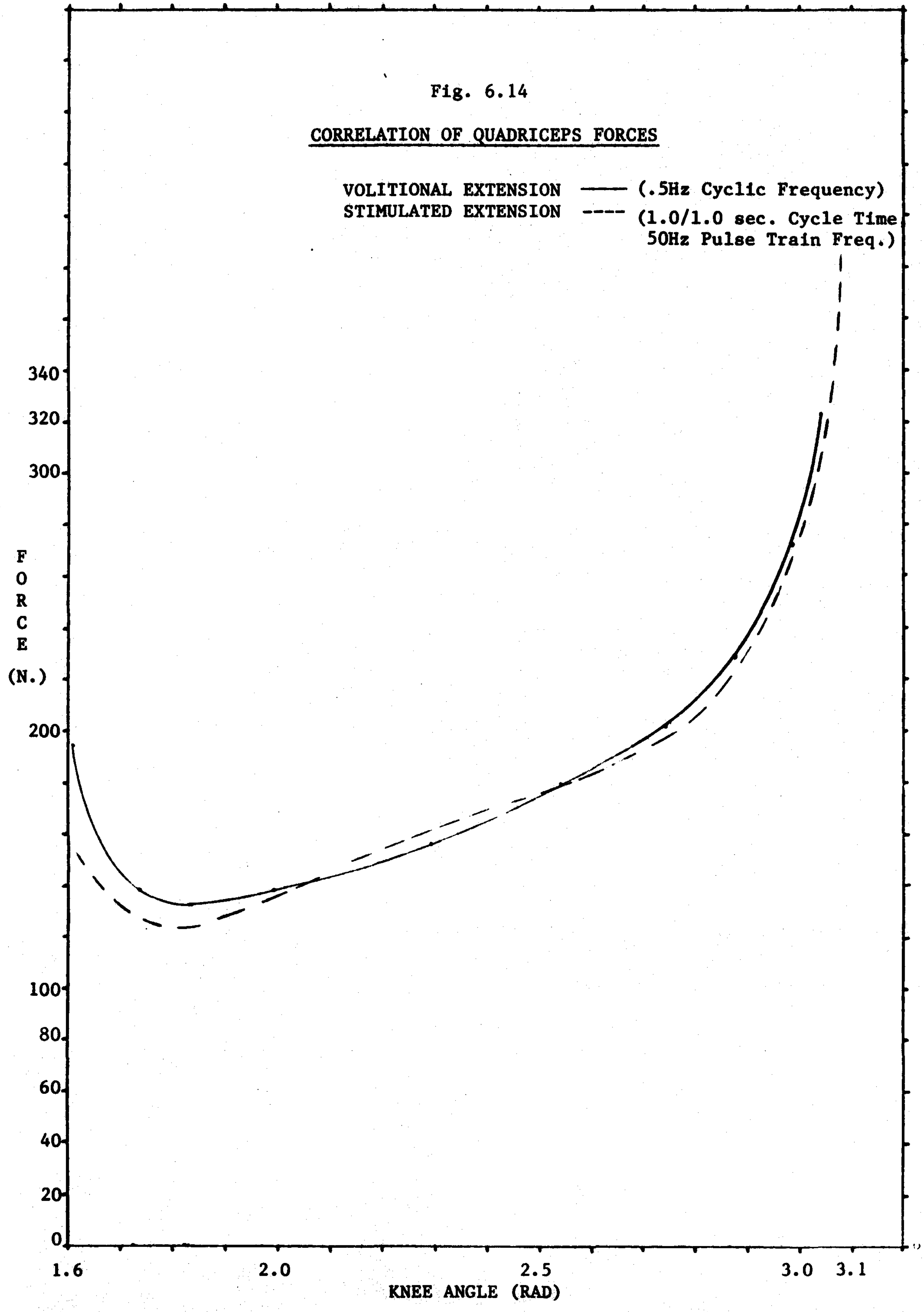


Fig. 6.14
CORRELATION OF QUADRICEPS FORCES
VOLITIONAL EXTENSION —— (.5Hz Cyclic Frequency)
STIMULATED EXTENSION ---- (1.0/1.0 sec. Cycle Time
50Hz Pulse Train Freq.)
340
320
300
FORCE
(N.)
200
100
80
60
40
20
0
1.6
2.0
2.5
3.0
3.1
KNEE ANGLE (RAD)

CHAPTER 7

CONCLUSIONS

7.1 Experimental Conclusions

Considering the force data and ensuing discussions appearing in sections 6.2 and 6.3, several conclusions may be drawn with respect to the muscle forces developed (both volitional and stimulated), during extensions of the knee against gravity.

These are listed below:

1) The strongest contractions of the quadriceps femoris muscle group, with the least painful sensations were elicited with a stimulating current pulse width of .2 msec.
2) The force required to initiate knee extension is proportional to the frequency of extension, whether volitional or stimulated.
3) The angular variation of force tends to be nearly linear at a stimulated cycle time of 1.0 sec/1.0 sec., although the variation is not exactly linear.
4) There is no appreciable difference in the

4) (Cont.)

forces exerted by the quadriceps during stimulated knee extension with changing pulse train frequencies, over the range 50 Hz. to 80 Hz.

5) The magnitudes of forces in all experiments are greatest at or near full extension, the magnitudes being about 350 Newtons.

6) The stimulating current parameters which produce the best force correlation with normal volitional forces during knee extension are:

(a) pulse train frequency of 50 Hz.

(b) overall cycle time of 2 sec. (1.0sec/1.0sec).

the correlation being with the volitional extension at a frequency of .5 Hz.

7.2 Sources of Error

By far the most significant source of error in this study was in the measurements made in order to determine the lower limb mass and the location of the centre of mass. Referring to section 5.3.5, one can see that these errors amount to approximately 4% in both measurements. However, since the design objective of this device was to restrict the error to within 5%, these measurements can be considered quite accurate.

The remainder of the error sources are negligible when compared to the parameter errors in the preceding paragraph. These include;

1) The initial manual measurement on the visicorder printout, of the visicorder deflection resulting from knee extension.
2) The assumption that the centre of rotation remains stationary throughout the extension cycle and the placement of the goniometer over this centre. (In reality, the instantaneous rotational centre traces out a curved path. Recently, a self-aligning goniometer has become available, which follows the path of the instantaneous centre.)
3) Errors encountered as a result of computer analysis. However, these were kept to a minimum by using five significant figures. This tends to reduce round-off or truncation errors.
4) Errors encountered during calibration of the potentiometer used in the goniometer. However, by repeating the calibration sequence several times, the random errors from misalignment with the protractor scale cancels. In any case, the worst error expected would be $\pm .5^{\circ}$.

7.3 Authors Technique as an Assessment Tool

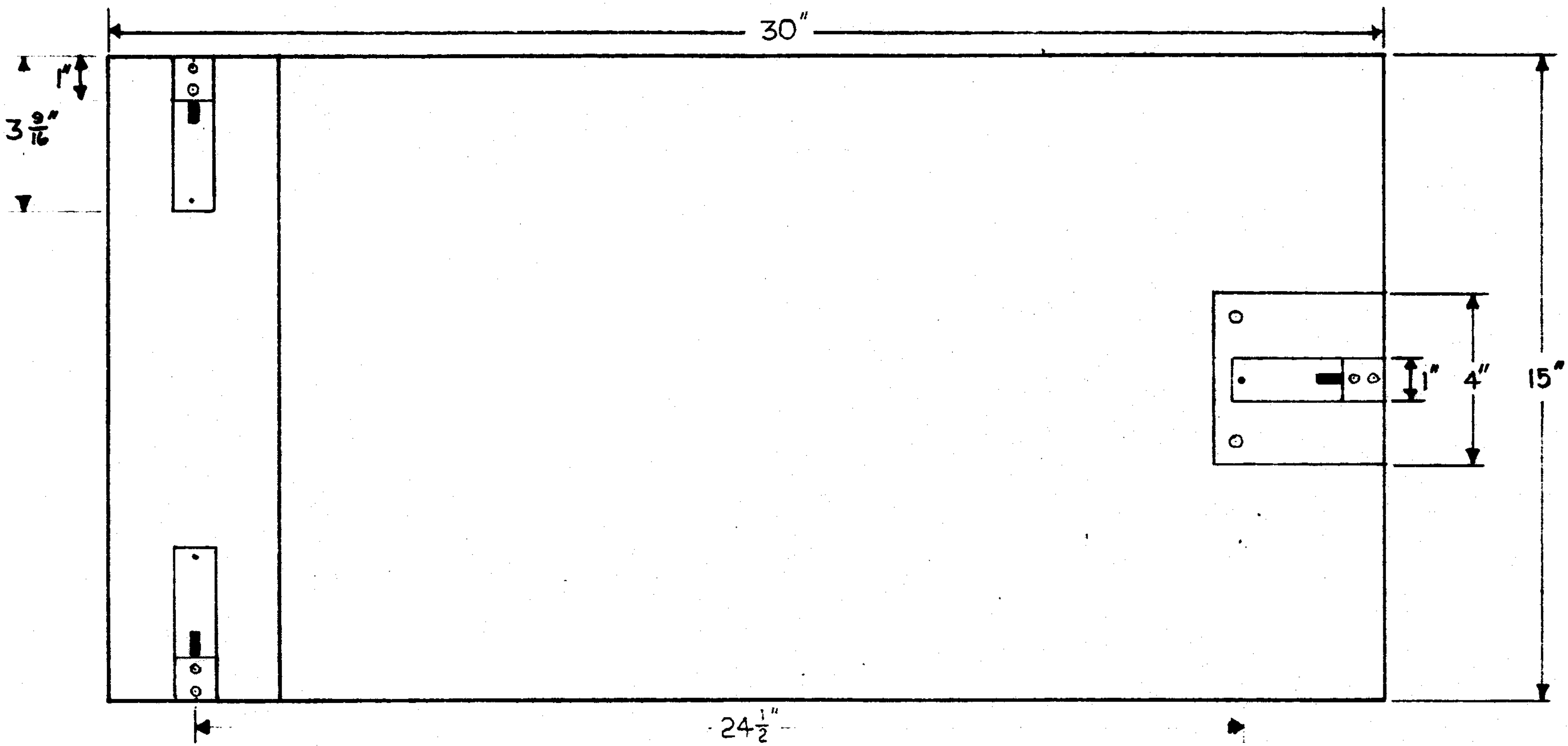
Assessment of knee joint function is possible by an extension of the technique used in this study. For example, it is known that the patella travels in a groove in the femoral head, during extension, and contacts this groove at one point only. Thus, with a sufficiently sensitive goniometer, some irregularities in the femoral groove which are suspected on X-rays might be confirmed by observing the goniometer response. Any sudden motion during the extension cycle would appear as an irregularity on the goniometer signal, which is normally smooth and continuous. The angle at which the discontinuity of the signal occurs could be calculated to within one or two degrees, thus giving the physician a very good indication of where along the femoral groove the irregularity is located. Perhaps the magnitude of the femoral irregularity may be given using this technique, although further studies on this hypothesis would be necessary to give this technique credibility among physicians.

Success or failure of physical therapy could be deduced with the goniometer. Preceding the therapy sessions, the patient would be requested to extend his knee at his maximum rate, with the goniometer properly secured over the knee joint. Following the sessions, he would be requested to repeat this test. Forces derived by the computer program in Appendix 3 would be graphed individually, thus enabling a comparison of the forces exerted before and after therapy to be performed. Successful therapy would be indicated by a general upward excursion of the post-therapy curve, i.e. in the direction of increasing force, whereas failure would be suggested by little or no movement of the post-therapy curve in this direction.

A direct consequence of this study, although not of the technique, is the availablility of a device to accurately determine the mass and location of the centre of mass of any limb or part thereof. Whereas previous calculations of torques and forces relied on approximate formulae from the literature using total body weight and dimensions as the reference, there now exists a device that will measure these parameters directly through the use of two simple equations. Since the output of this apparatus is a voltage, it may be interfaced with a computer, and the necessary calculations unnecessary. Larger or smaller models of this unit can be constructed to suit any situation that may occur. Appropriate technical and circuit diagrams may be found in Appendix 1.

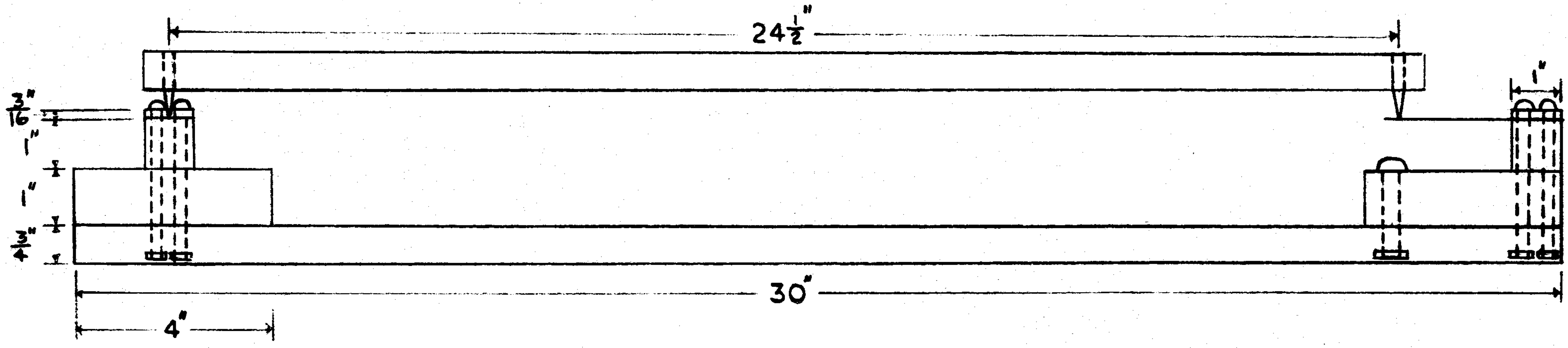
APPENDIX 1

CENTRE OF MASS AND MASS MEASURING DEVICE

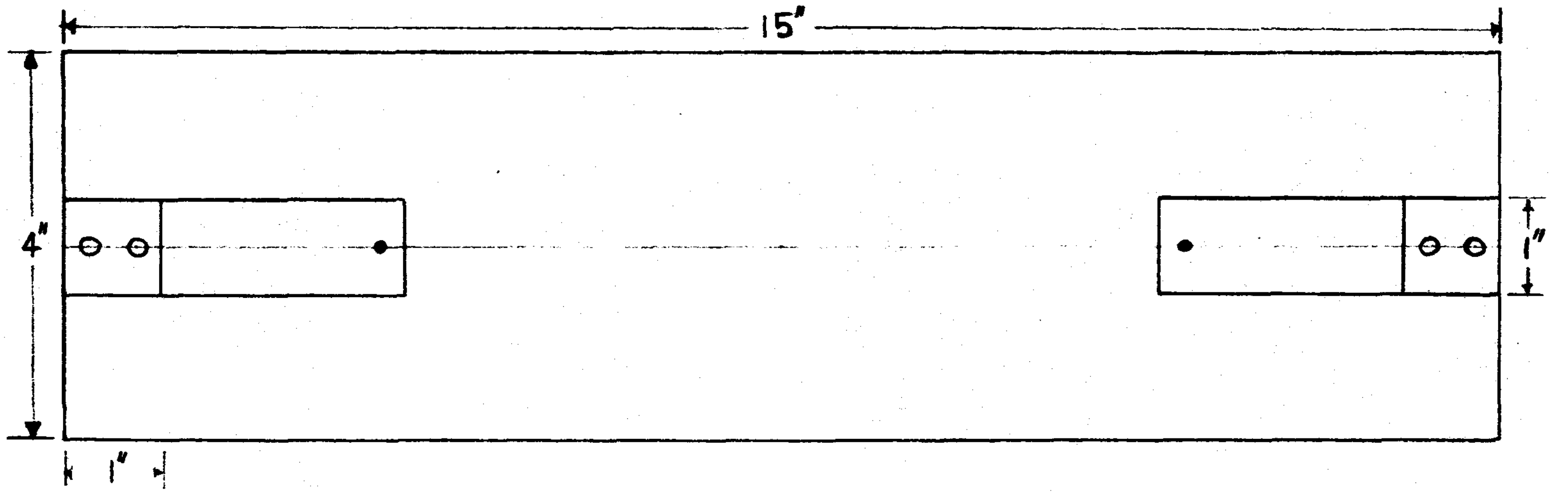


CENTRE OF MASS MEASURING DEVICE (MAIN SECTION)

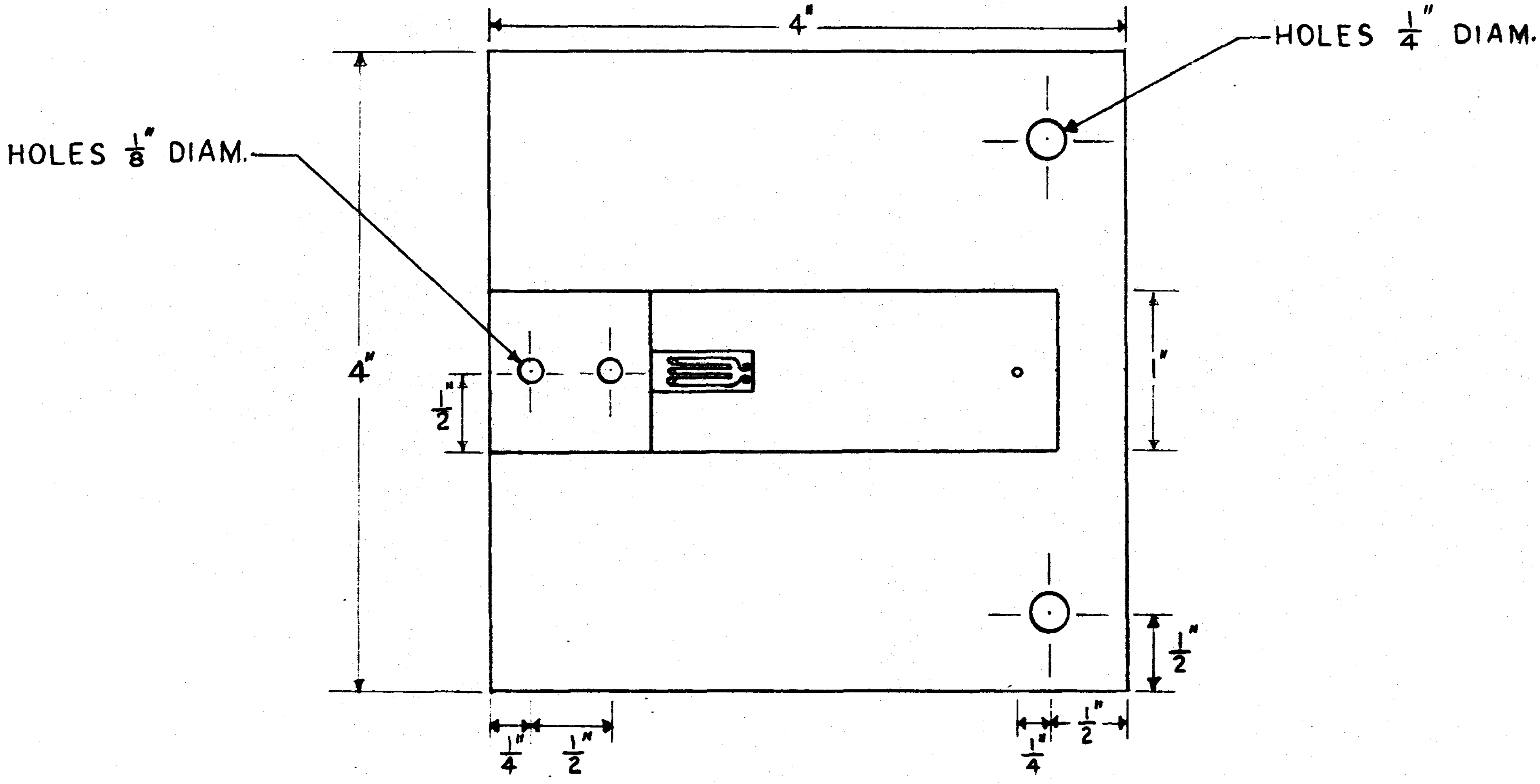
SCALE = 1:4



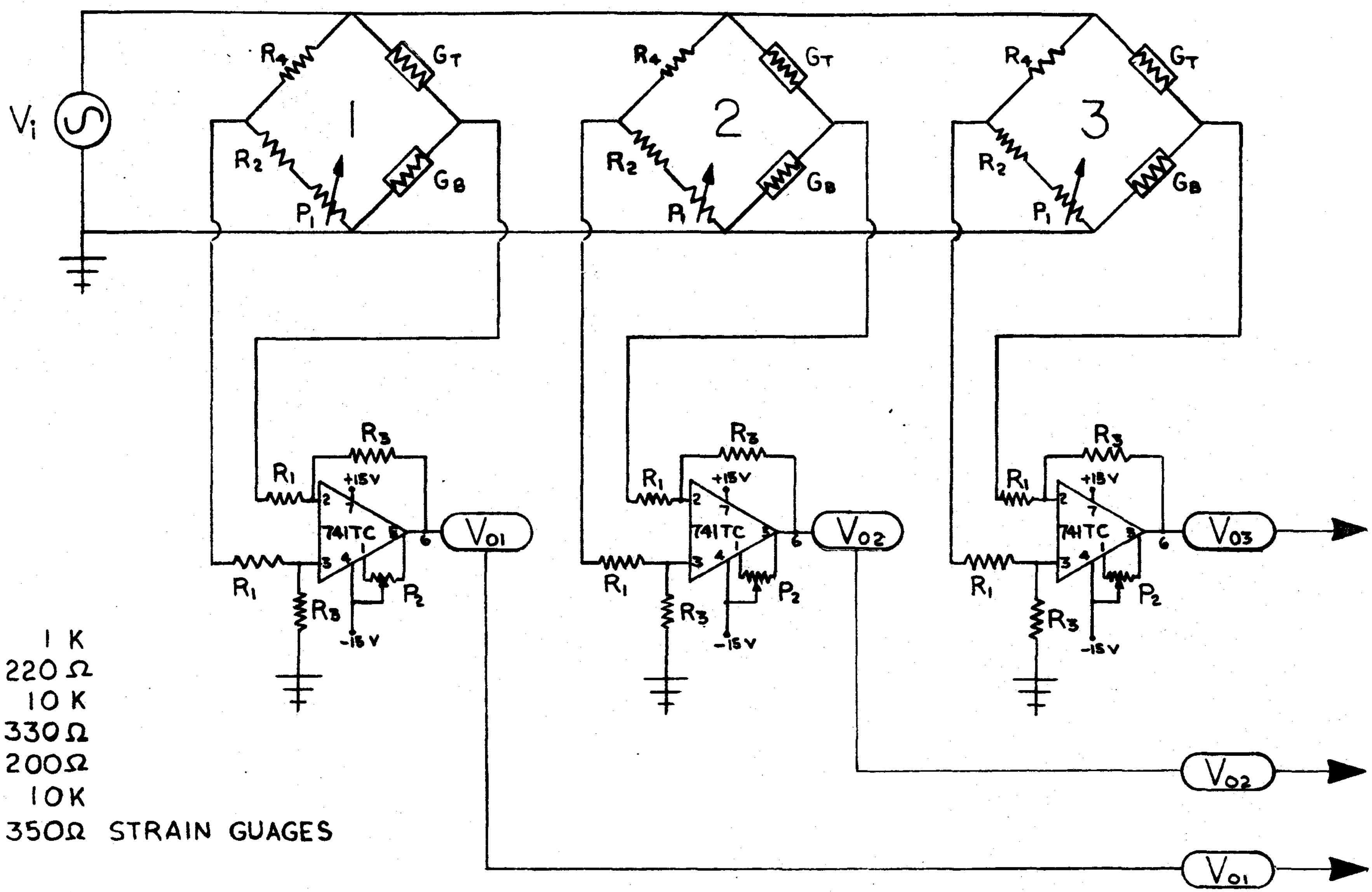
CENTRE OF MASS DEVICE (SIDE VIEW) SCALE = 1:3



CANTILEVERS 1 AND 2 (TOP VIEW) SCALE = 1:2

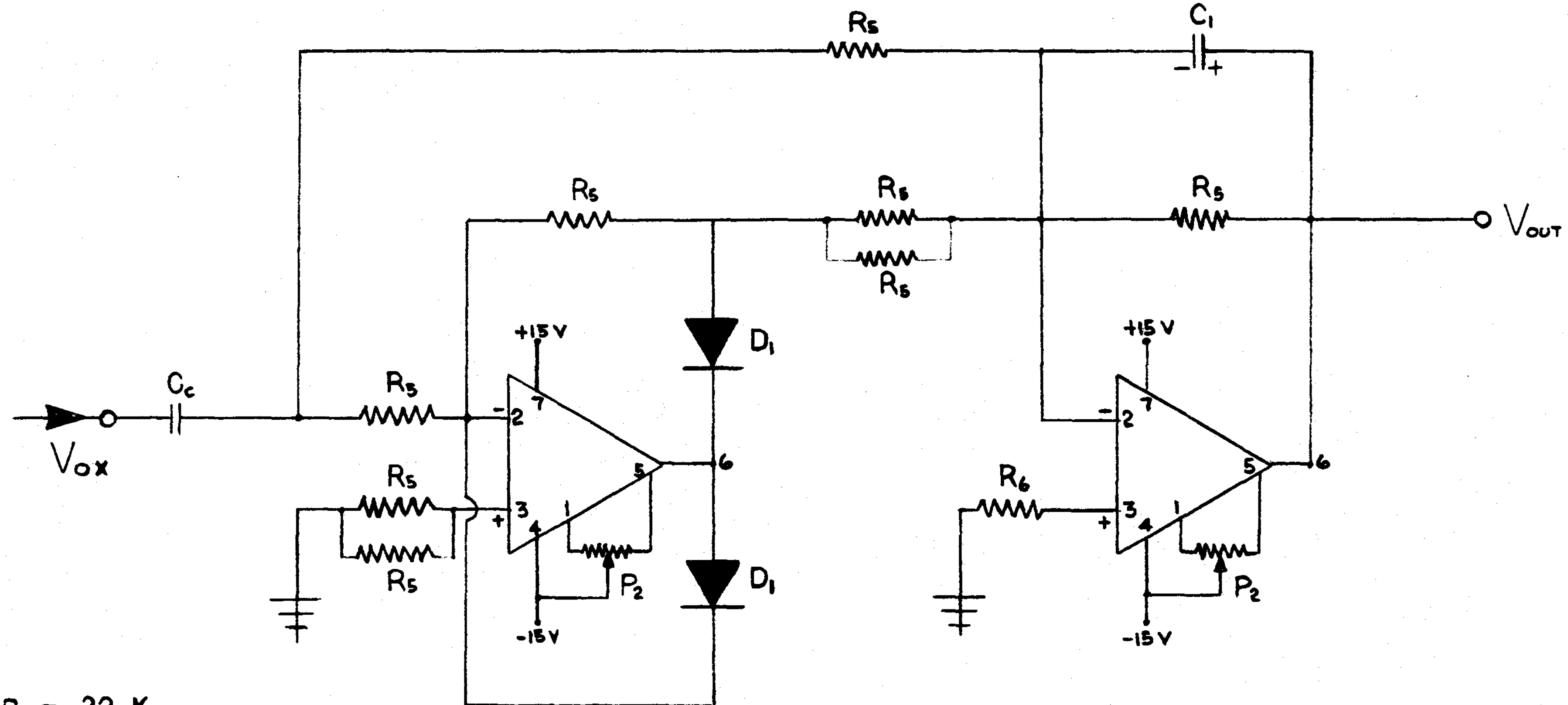


CANTILEVER 3 (TOP VIEW) SCALE = 1:1



R_1 = 1 K
R_2 = 220 Ω
R_3 = 10 K
R_4 = 330 Ω
P_1 = 200 Ω
P_2 = 10 K
G_T, G_B = 350 Ω STRAIN GUAGES

CIRCUIT DIAGRAM FOR CENTER OF MASS MEASURING DEVICE

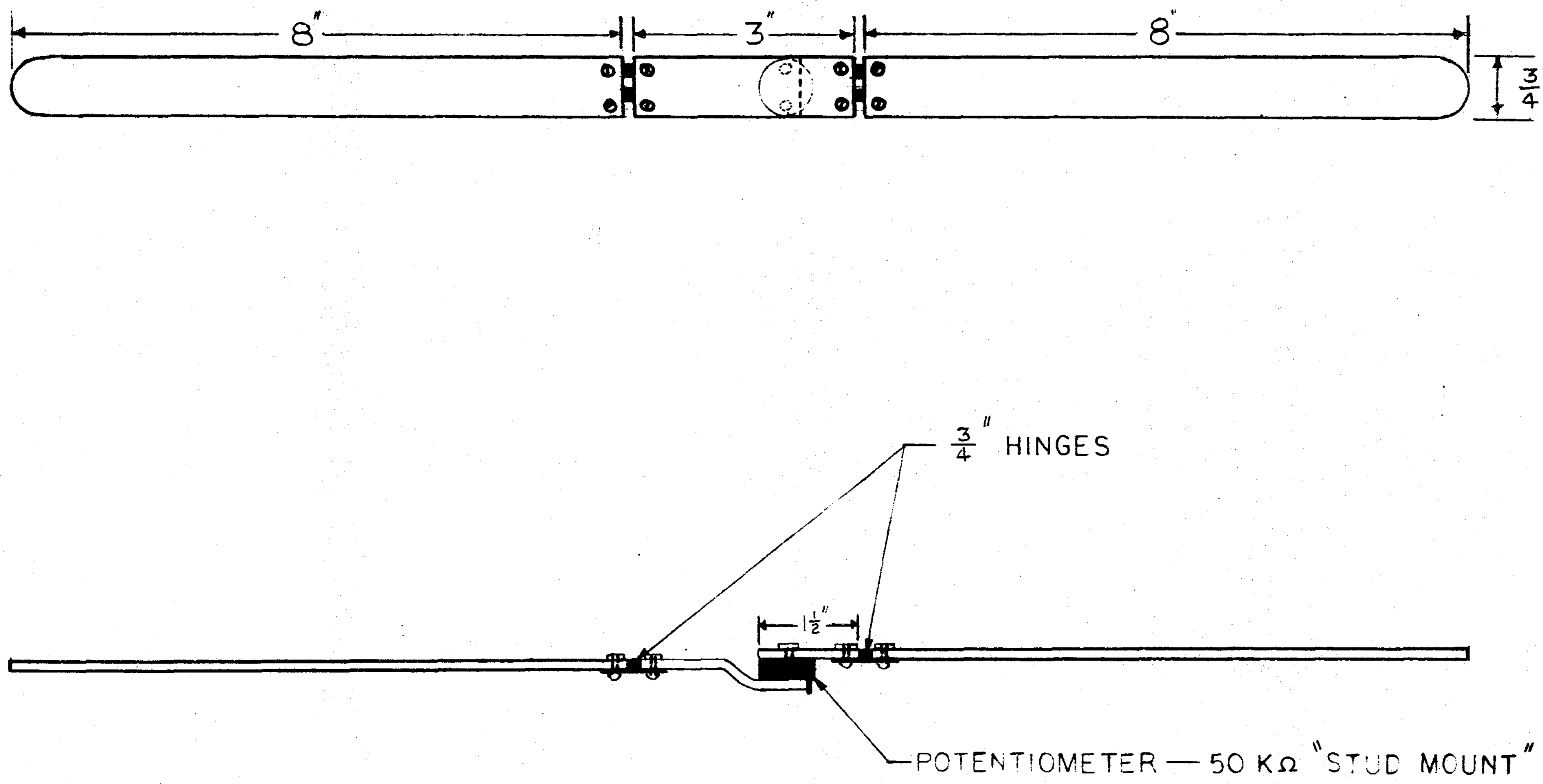


R_5 = 22 K
R_6 = 5.6 K
P_2 = 10 K
C_1 = .22 μf
C_c = 10 μf
D_1 = IN4154 SIGNAL DIODE

FULL WAVE RECTIFIER AND AVERAGER

APPENDIX 2

ELECTROGONIOMETER



ELECTROGONIOMETER SCALE = 1:2

APPENDIX 3

COMPUTER PROGRAM

```
FTN.
ATTACH(MACLIB)
LDSET(LIB=MACLIB)
LGO.
≠        6400 END OF RECORD
      PROGRAM TST (INPUT,OUTPUT,TAPE5=INPUT,TAPE6=OUTPUT)

C     ****************************************************************

C     THIS PROGRAM COMPUTES THE FOLLOWING PARAMETERS, WITH INPUT BEING
     C  THE DEFLECTIONS ON A PERMANENT RECORD, AS THE RESULT OF KNEE
     C  EXTENSION. ANGULAR PARAMETERS ARE GIVEN IN DEGREES AND RADIANS.
C             A) COEFFICIENTS OF THE LEAST SQUARES FIT OF ANGULAR ROTATION
C                AND TIME
C             B) ANGULAR ROTATION USING COEFFICIENTS IN A)
C             C) ANGULAR VELOCITY OF LOWER LEG DURING KNEE EXTENSION
C             D) ANGULAR ACCELERATION OF LOWER LEG DURING KNEE EXTENSION
C             E) TORQUE ABOUT ROTATIONAL AXIS OF KNEE DUE TO ACTION OF
C                QUADRICEPS MUSCLE GROUP
C             F) COEFFICIENTS OF POLYNOMIAL DESCRIBING VARIATION OF
C                PATELLAR TENDON MOMENT ARM DURING KNEE EXTENSION, BETWEEN
C                90 DEGREES FLEXION AND 180 DEGREES FULL EXTENSION
C             G) QUADRICEPS MUSCLE FORCE DURING KNEE EXTENSION
C             H) INERTIAL AND GRAVITATIONAL COMPONENTS OF QUADRICEPS
C                MUSCLE FORCE DURING KNEE EXTENSION

C     ****************************************************************
      DIMENSION FORCE(20),XLEVER(20),DEF(20), C(5), THETAT(20),
     C  THETA1(20), THETA2(20), TIME(20), TORQUE(20), ANGRAD(20), B(5),
     C  THETAL(20), D(5), A(50), THTRAD(20), SA(20), ST(20), TH1RAD(20),
     C  TRMINT(20), TRMGRV(20)
      DIMENSION FRCINT(20), FRCGRV(20)
      DATA(TIME(I), I=1,16)/0.,.1,.2,.3,.4,.5,.6,.7,.8,.9,1.0,1.1,1.2,
     C  1.3,1.4,1.5/
      DATA(XLEVER(I), I=1,11)/.0388,.0388,.04065,.04445,.0483,.0493,
     C  .0488,.0478,.04695,.04575,.04425/

C     READ IN MEASURED DEFLECTIONS FROM PERMANENT RECORD, WITH N BEING
C     THE NUMBER OF DATA POINTS
      N=11
   10 READ(5,20) (DEF(I), I=1,N)
   20 FORMAT(8F8.2)

C     POLYNOMIAL DESCRIBING ANGULAR ROTATION IS FOURTH ORDER IN TIME
C     AND HAS FIVE COEFFICIENTS, C(1) BEING THE CONSTANT TERM

      M=4
      MM=M+1

C     COMPUTE ANGULAR ROTATION FROM MEASURED DEFLECTIONS ON PERMANENT
C     RECORD, USING THE CALIBRATION EQUATION OF THE ELECTROGONIOMETER

      DO 1 I=1,N
      THETAT(I) = 59.956 + 1.0634*DEF(I)
    1 CONTINUE
      WRITE(6,36)
   36 FORMAT(*1           COEFFICIENTS OF LESQ FIT (ANGLE AND TIME)*)
```

```
C      DETERMINE THE LEAST SQARES FIT TO THE DATA OF ANGULAR ROTATION
C      OF LOWER LEG WITH TIME, AND PRINT THE COEFFICIENTS OF THE
C      RESULTING PLOYNOMIAL

       CALL LESQ(A,C,TIME,THETAT,M,N)
       WRITE(6,32) (C(I), I=1,MM)
    32 FORMAT(1H0, 5E12.5)
       WRITE(6,34)
    34 FORMAT(/,*0      TIME               THETAT              THETA(LESQ)*,/)

C      CALCULATE THE ANGULAR ROTATION FROM THE LEAST SQARES COEFFICIENTS
C      AND COMPARE THE RESULTS WITH THE ACTUAL DATA POINTS

       DO 2 I=1,N
       THETAL(I)=C(1)+TIME(I)*(C(2)+TIME(I)*(C(3)+TIME(I)*(C(4)+TIME(I)*
      1C(5))))
       WRITE(6,33) TIME(I), THETAT(I), THETAL(I)
    33 FORMAT(1H ,6X, F3.1,6X, E12.5,6X, E12.5)
     2 CONTINUE
       WRITE(6,60)
    60 FORMAT(1H0)
       WRITE(6,30)
    30 FORMAT(1H , 9H     TIME, 10X, 10HDEFLECTION, 14X, 5HTHETA, 10X,
      C  8HTHETA 1D,10X, 8HTHETA 2D, 9X, 6HTORQUE/)

C      ENTER THE VALUE OF THE CONSTANT OF GRAVITATIONAL ACCELERATION, G,
C      AND THE VALUES OF THE LOWER LIMB MASS AND THE DISTANCE OF THE
C      CENTRE OF MASS OF THE LOWER LIMB , FROM THE ASSUMED ROTATIONAL
C      AXIS OF THE KNEE. ALL OF THESE VALUES ARE GIVEN IN THE MKS SYSTEM.

       G=9.8
       XCOM = .2339
       XMASS = 5.62
       XINERT=XMASS*(XCOM**2)

C      CALCULATE ALL PARAMETERS LISTED IN PARTS C) TO H) INCLUSIVE, AT
C      THE BEGINING OF THIS PROGRAM

       DO 3 I=1,N
       THETA1(I)=C(2)+TIME(I)*(2*C(3)+TIME(I)*(3*C(4)+TIME(I)*4*C(5)))
       THETA2(I)=2*C(3)+TIME(I)*(6*C(4)+TIME(I)*12*C(5))
       THTRAD(I)=(THETAT(I)-90.) * 3.14159/180.
       TH1RAD(I) = THETA1(I)*3.14159/180.
       ANGRAD(I)=THETA2(I)*3.14159/180.
       SA(I)=SIN(ANGRAD(I))
       ST(I) = SIN(THTRAD(I))
       TORQUE(I)=XINERT*ANGRAD(I) + XMASS*G*XCOM*ST(I)
       WRITE(6,37) TIME(I), DEF(I), THETAL(I), THETA1(I),  THETA2(I),
      1TORQUE(I)
    37 FORMAT(1H ,5X,F3.1,13X,F6.2,12X,E12.5,4X,E12.5,6X,E12.5,5X,E12.5)
     3 CONTINUE
       WRITE(6,63)
    63 FORMAT(1H0)
       WRITE(6,43)
    43 FORMAT(*0     THETA2     ANGRAD              THETAT      THTRAD*)
       WRITE(6,45)
    45 FORMAT(1H+,61X,6HTHETA1,10X,6HTH1RAD)
       WRITE(6,61)
    61 FORMAT(1H )
```

```
      DO 57 I=1,N
      WRITE(6,44) THETA2(I),ANGRAD(I), THETAT(I), THTRAD(I)
   44 FORMAT(1H , 2X, E12.5, E12.5, 3X, E12.5, E12.5)
      WRITE(6,36) THETA1(I),TH1RAD(I)
   38 FORMAT(1H+,57X,E12.5,2X,E12.5)
   57 CONTINUE
      WRITE(6,71)
   71 FORMAT(1H )
      WRITE(6,39)
   39 FORMAT(*      LESQ COEFFICIENTS/// ANGLE (RADIANS) AND TIME///*)
      CALL LESQ(A,D,TIME,THTRAD,M,N)
      WRITE(6,70) (D(I), I=1,MM)
   70 FORMAT(1H , 5E12.5)
      WRITE(6,100)
  100 FORMAT(1H1,6X,6HTORQUE, 7X,9HLEVER ARM, 7X,5HFORCE)
      WRITE(6,76)
   76 FORMAT(1H )
      DO 101 I=1,N
      FORCE(I) = TORQUE(I)/XLEVER(I)
      WRITE(6,102) TORQUE(I), XLEVER(I), FORCE(I)
  102 FORMAT(2H ,E12.5,7X,F6.4, 8X, E12.5)
  101 CONTINUE
      WRITE(6,74)
   74 FORMAT(1H )
      WRITE(6,73)
   73 FORMAT(*CINERTIAL TORQUE     GRAVITATIONAL TORQUE INERTIAL FORCE
     C  GRAVITATIONAL FORCE*)
      WRITE(6,75)
   75 FORMAT(1H )
      DO 2 I=1,N
      TRMINT(I) = XINERT*ANGRAD(I)
      TRMGRV(I) = XMASS*G*XCOM*ST(I)
      FRCINT(I) = TRMINT(I)/XLEVER(I)
      FRCGRV(I) = TRMGRV(I)/XLEVER(I)
      WRITE(6,72) TRMINT(I), TRMGRV(I),FRCINT(I), FRCGRV(I)
   72 FORMAT(1H ,2X,E12.5,8X,E12.5,7X,E12.5,5X,E12.5)
    2 CONTINUE
      STOP
      END
```

CDTOT 159

REFERENCES

(1) Brunnstrom, S., Clinical Kinesiology - 2nd Edition, F.A. Davis Co, Philadelphia, 1969, pp. 171-175.

(2) Brunnstrom, S., Clinical Kinesiology - 2nd Edition, F.A. Davis Co., Philadelphia, 1969, page 172.

(3) Warwick, R., and Williams, R.L. (Ed.), Gray's Anatomy- 35th Edition, 1966, page 405.

(4) Warwick, R. and Williams, R.L. (Ed.) Gray's Anatomy- 35th Edition, 1966, page 406.

(5) Anthony, C A., Textbook of Anatomy and Physiology - 7th Edition, The C.V. Mosby Company, St. Louis, 1967, page 142.

(6) Anthony, C A., Textbook of Anatomy and Physiology - 7th Edition, The C.V. Mosby Company, St. Louis, 1967, pp. 156-7.

(7) Guyton, A.C., Basic Human Physiology: Normal Function and Mechanisms of Disease, W.B. Saunders Co., Toronto, 1967, page 76.

(8) Basmajian, J.V., Primary Anatomy - 6th Edition, Williams and Wilkins Co., Baltimore, 1970, pp. 121-3.

(9) Anthony, C.A. Textbook of Anatomy and Physiology - 7th Edition, The C.V. Mosby Company, St. Louis, 1967, page 130.

(10) Licht, S.H. (Ed.), Electrodiagnosis and Electromyography- 3rd Edition, Waverly Press Inc., Baltimore, 1971, pp. 153-7.

(11) Licht, S.H. (Ed.), Therapeutic Electricity and Ultraviolet Radiation - 2nd Edition, Waverly Press Inc., Baltimore, 1967, page 130.

(12) Licht, S.H. (Ed.), Electrodiagnosis and Electromyography - 3rd Edition, Waverly Press Inc., Baltimore, 1971, pp. 353-8

(13) Licht, S.H. (Ed.), Therapeutic Electricity and Ultraviolet Radiation - 2nd Edition, Waverly Press Inc., Baltimore, 1960, pp. 125-6.

(14) Parrish, A. (Ed.), Mechanical Engineering Reference Book - 11th Edition, Butterworth and Company (Canada) Limited, 1973, page 1239.

(15) Anthony, C.A., Textbook of Anatomy and Physiology - 7th Edition, The C.V. Mosby Co., St. Louis, 1967, pp. 130-132.

(16) Smidt, G.L., Biomechanical Analysis of Knee Flexion and Extension, Journal of Biomechanics, Vol. 6, No. 1, Pergammon Press, New York, Jan. 1973, pp. 79-92.

B I B L I O G R A P H Y

Basic Human Physiology: Normal Function and Mechanisms of Disease, Guyton, A.C., W.B. Saunders Co., Toronto, 1967.

Biomechanical Analysis of Knee Flexion and Extension, Smidt, G.L., Journal of Biomechanics, Vol. 6, No. 1, 1973.

Biomechanics of Human Motion, Lissner, H.T. and Williams, M. J. American Phy. Ther. Assn., Vol. 43, 1963

Clinical Kinesiology, Brunnstrom, S., F.A. Davis Co., Philadelphia, 1970

Electrodiagnosis and Electromyography - 3rd Edition, Licht, S.H. (Ed.), Waverly Press Inc., Baltimore, 1971.

Forces and Energy Changes in the Leg During Walking, Elftman, H., Amer. J. Phys., Vol. 125, 1939.

Gray's Anatomy - 35th Edition, Warwick, R. and Williams, R.L. (Ed.), Waverly Press Inc., Baltimore, 1960.

Mechanical Engineer's Reference Book - 11th Edition, Parrish, A. (Ed.), Butterworth and Company (Canada) Ltd., 1973.

Muscles Alive - 3rd Edition, Basmajian, J.V., Williams and Wilkins Co., Baltimore, 1974.

Physiology and Biophysics - 19th Edition, Ruch, T.C. and Patton, H.D., W. B. Saunders Co., Philadelphia, 1965.

Primary Anatomy, Basmajian, J.V., Williams and Williams Co., Baltimore, 1970.

Textbook of Anatomy and Physiology, Anthony, C.A., The C.V. Mosley Co., St. Louis, 1967.

The Influence of Electrical Stimulation of Muscles on the Development of Fatique in Man, Hartman, A.C. and Bouman, H.D., Phys. Ter. Rev., Vol. 30, 1950

The Mechanism of Muscular Contraction, Huxley, H.E., Scientific American, Vol. 213, No. 12, Scientific American Inc., 1965.

Therapeutic Electricity and Ultraviolet Radiation, Licht, S.H. (Ed.), Waverly Press Inc., Baltimore, 1967.