ELECTROMYOGRAPHIC AND POSITION CONTROLLED FUNCTIONAL ELECTRICAL STIMULATION OF THE MUSCULATURE ABOUT THE HUMAN ANKLE JOINT

BY

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A Thesis
Submitted to the Faculty of Graduate Studies
in Partial Fulfilment of the Requirements
for the Degree
Doctor of Philosophy
McMaster University
(November) 1978

CONTROL OF ANKLE JOINT POSITION BY MEANS OF FES

DOCTOR OF PHILOSOPHY (1978)
(Electrical Engineering)

McMASTER UNIVERSITY
Hamilton, Ontario

TITLE: Electromyographic and Position Controlled Functional Electrical Stimulation of the Musculature about the Human Ankle Joint

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NUMBER OF PAGES: xviii, 193

ABSTRACT

In the last two decades, functional electrical stimulation (FES) has been investigated as a means for replacing lost function of limbs resulting from paralysis. Improvement in the gait of hemiplegic patients when gradually varying stimulation sequences were employed to control footdrop during the swing-phase of gait, led to the work presented in this thesis.

The material described below is original to the field of FES.

Two potential controllers of stimulus intensity and hence ankle joint position on the affected side have been explored. These are: the electromyographic (EMG) activity of the corresponding dorsiflexor and plantarflexor muscles on the contralateral side; and ankle-joint angle variations obtained from the contralateral side. The variance ratio, a statistical descriptor for repeatability, has been invoked to quantify the efficacy of EMG and joint position control. Practical time-constants of averaging have been determined for the processing of control and evoked EMG signals to be used in an FES-based orthosis incorporating feedback. Experiments have indicated that EMG, when used to modulate stimulus strength to effect control of ankle-joint position, is as

efficacious as joint-angle-variation control. These experiments revealed that joint-position information is contained in the EMG records obtained from the prime movers during specific movements of the ankle joint.

This thesis describes an initial attempt to control the affected ankle-joint position of hemiplegics during locomotion. Corresponding signals available from the contralateral side were used to modulate stimulus intensity on the affected side. A computer-controlled interactive program has been used to impose a delay proportional to the period of stepping between recording of the control signals and activation of the stimulators. Preliminary results obtained from a normal and a hemiplegic subject are presented, and their relevance to future thrusts in the field of FES are discussed.

TO PAM, MOSHE, AND TALYA

ACKNOWLEDGEMENTS

This thesis is based upon work performed in the Department of Biomedical Engineering at the Chedoke Rehabilitation Centre, Hamilton, Ontario.

I would like to express my gratitude to:

Dr. M. Milner for his encouragement, assistance, and supervision during the entire period of this thesis.

Dr. M.E. Brandstater and Dr. S. Sarna for their supervision.

The members of the Department of Biomedical Engineering at the Chedoke Rehabilitation Centre and in particular to Dr. H. de Bruin, Dr. C. Hershler, Dr. R. Bloch, Dr. J. Russell, A. Wallace, S. Abdel-Azim, and W. Nemeth for their many stimulating discussions and their assistance provided during the course of this thesis.

Mrs. C. Gowland and Mrs. B. Clarke for providing the clinical link and for their many instructive discussions.

Mrs. S. Fick and Mrs. M. Hickey for typing this manuscript, and for their unfailing support and friendship.

Dr. A. Jutan for his interest and advice.

My mother and parents-in-law for their encouragement and financial support.

Glossary (in order of appearance)

FES	functional electrical stimulation
EMG	electromyography
$\alpha_1(t)$	control signal source
$\alpha_2(t)$	response to stimulation
$c(t)$	conditioned signal
$s(t)$	stimulation source
p.d.	potential difference
A.P.	action potential
D.C.	direct current
CNS	central nervous system
α	extrafusil muscle fibre motor nerve pathway
γ	intrafusil muscle fibre motor nerve pathway
VR	variance ratio
DA	differential amplifier
PI	proportional-plus-integral
f_0	break frequency
T_A	averager time constant
T_D	window length
k	window length ratio
EG	electrogoniometer
LEG1	computer-stored control record
LEG2	stimulated limb

$EMG_1(t)$	control EMG record
$EMG_2(t)$	evoked EMG record
$\theta_1(t)$	volitional (control) joint angle
$\theta_2(t)$	evoked joint angle
m	muscle
$\hat{\theta}$	maximum joint angle
O/L	open loop
F/B	feedback
HS	heel-strike
ST	onset of stance
HT	foot-flat (heel and toe)
TO	toe only
SW	onset of swing
DS	double-support
CAC EGCA	control signal during swing
CBC EGCB	control signal during swing
CAS	evoked signal during swing
CBS	evoked signal during stance
A_C	onset of swing on control side
B_C	onset of stance on control side
A_S	onset of swing on stimulated side
B_S	onset of stance on stimulated side
I3CHAN	signal on controlled side

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CHAPTER 1

ISSUES OF CONCERN

Paralysis of extremities may be due to diseases of the muscles themselves, diseases and injuries to the nerve which delivers stimulating impulses to the muscles, or diseases and injuries to the brain and spinal cord. This study concerns itself with an attempt to replace the absent functional activity of the lower limb when paralysis is due to the destruction of nerve cells in the spinal cord or brain (upper motor neuron lesion). The muscles, in this case, retain their ability to contract. However, the command signals to contract are either absent or contractions occur in an uncontrollable manner. At present, aids used to overcome limitations in independent motor activity include the wheelchair, the orthopedic long-leg brace, and the swivel walker. For the paralyzed person to become independent of others, external aids must be developed whose size, weight and complexity are not unduly prohibitive. Energy consumption must be minimized in devices reliant upon external power sources.

1.1 FES as an Alternative

In the last two decades, functional electrical stimulation (FES) has been investigated as an alternative to other orthotic aids since paralyzed muscles are often still contractile. Unlike the more common supporting or stiffening apparatus, FES attempts to replace the missing muscular stimulation itself in such a way that the paralyzed man again moves his limbs in a functionally useful way. External bracing should be obviated by utilizing the existing musculo-skeletal system. Stimulated muscles amplify the externally provided electrical trigger thus conserving energy.

For FES to become a viable alternative to other orthotic aids, problems due to electrical stimulation such as the rapid onset of muscle fatigue, unpleasant sensory effects, and eliciting functional forces, must be solved. In particular, adequate control of the system must be provided for the user of such an aid without unduly increasing the conscious effort needed to operate the system.

Providing patterns of innervation with adaptability to meet the requirements of a changing environment even during a repetitive cyclic action such as locomotion is complex. This complexity has tended to restrict research in this area to the relatively simpler case of improving hemiplegic gait. The experience gained from such studies will hopefully lead to the successful application of FES in rehabilitating the paraplegic.

1.2 The Need for Orthotic Devices

The incidence rate of strokes per annum in the U.S. is about 170 to 190 per 100,000 of the population. Early mortality may account for 50% of these people (Mossman,1976). A significantly large stroke population thus exists, most of them having a potential need for some form of an orthotic device. One problem prevalent in hemiplegics is that of foot drop. Foot drop is due to the inability of the stroke victim to lift (dorsiflect) his toes during the swing phase of the gait cycle thus impairing his balance.

1.3 Line of Approach

For hemiplegia, it is this author's contention that continuously modulated signals are potentially available from the unaffected side to control the stimulation on the affected side. To be effectively used, a delay proportional to the period of stepping and step interval is needed. This delay should be interposed between the recording of the control signals and the activation of the stimulators applied to the paralyzed limb during walking in a straight line.

1.4 Control Aspects

Two signal sources that theoretically lend themselves to contralateral joint-position control are: the electromyographic (EMG) activity of muscles; and joint-position itself. When EMG signals from corresponding muscles on the unaffected side are used as control signals, an obvious advantage is that the programmatic sequences and levels to be used are derived from a set of muscles that can be regarded to grossly approximate the representation of the functional sequences desired on the paralyzed side. This assumes that the gait has achieved a steady-state with symmetry between the two sides. Using joint-position as the controller simplifies the complexity of a control system since only one signal is needed to control either flexion or extension of a joint.

1.5 Scope of Thesis

To be effective, either control mode should be able to be incorporated into a feedback system thus providing some modicum of adaptive control.

The purposes of this thesis are:

(i) The formulation of a quantitative means for determining the optimal forms of the control and evoked EMG signals for given processors for use in an EMG-controlled closed-loop FES orthosis.

(ii) The quantification of the efficacy of different control modes employing position control of joint position.

(iii) The incorporation of knowledge derived from (i) and (ii) above in the implementation of a program for adaptively controlling the delay between the activity of the uninvolved side and stimulation of the affected side in stroke patients during walking in a straight line, and to study the resulting locomotor responses.

1.6 Chapter Description

Achievement of these purposes necessitated constructing a stimulator whose design depends upon an understanding of the physiological basis of the mechanisms involved in electrical excitation of skeletal muscle, and EMG generation. Chapter 2 contains an historical review of FES as applied to the musculo-skeletal system, a brief functional description of the human lower leg, and pertinent physiological information. Consideration is given to the choice of stimulating electrodes and stimulus waveforms and their relationship to pain. Emphasis has been placed on the physiological control mechanisms involved in gait and their importance in the formulation of the concept of the musculo-skeletal system as a multilevel control system. The relevance of a hierarchical control system for the design of adequate FES systems is stressed.

Chapter 3 is a literature review of proposed methods for controlling paralyzed limbs using electrostimulation and the development of the unique method of one-to-one contralateral control employing EMG or joint-position feedback. A criterion is presented whereby quantitative comparisons between measured variables can be made.

Chapter 4 describes the design philosophy and realization of a four-channel stimulator. (Only 2 channels have been employed in deriving the results of this study. The stimulator was designed so that aspects beyond those considered in this thesis can be investigated). Included in this chapter are experiments used to determine the parameters of a proportional-integral filter, the recovery of evoked EMG signals during stimulation, and the equivalencing of an RC-averager and a digital boxcar-window processor.

Chapter 5 describes the experimental procedure for determining the optimal control and feedback EMG signals when used in a feedback system to control ankle joint position. An interesting result is the high correlation that exists between joint-angle time-histories and processed EMG signals recorded from the prime movers during specific movements (Naumann and Milner, 1978(a)).

The determination of the efficacy of different modes of position control is described in Chapter 6 (Naumann and Milner, 1978(b)). A comparison between EMG and position control is given.

Chapter 7 presents a limited review of human gait and the implementation of an adaptive program utilizing footswitches to control the delay proportional to stepping period and step interval imposed between recording of the control signals from the contralateral side and activation of the stimulators. Results obtained from a normal and a hemiplegic subject are given. Future possiblities are discussed.

1.7 Major contributions

The author has proposed and demonstrated the feasibility of an FES-based orthotic system utilizing remaining intact function in the hemiplegic subject. The system has been applied to prevent foot drop during locomotion.

Contralateral control as described in this thesis is unique in that it realizes a control system based on the principle of maximum autonomy. Decision making is limited to the selection of a particular action. Fixed programs or repetitive patterns are derived from the contralateral side. Adaptive control is provided for by incorporating feedback from the activators as well as changes in step period and step interval.

In realizing the FES-system, it has been demonstrated that EMG recorded from corresponding muscles on the contralateral side is as efficacious a controller of joint position as joint-position itself. In doing so, it was shown that the EMG signal recorded from a prime mover during a specific movement contains positional information of the joint.

A quantitative means has been presented for determining the forms of the control and evoked EMG signals for a given processor which will produce the greatest repeatability between volitional and evoked joint angles.

CHAPTER 2

NEUROPHYSIOLOGICAL ASPECTS OF FES

2.1 Introduction

This chapter briefly introduces the basic components and requirements of an FES-based orthosis.

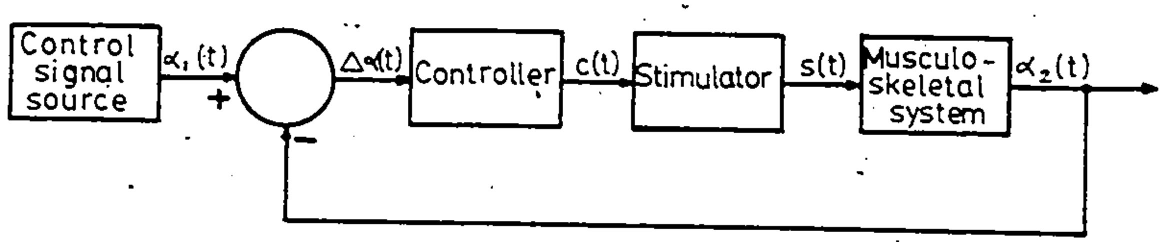


Figure 2.1 Proposed FES scheme

Figure 2.1 is a block diagram of an FES system. The control signal source, $\alpha_1(t)$, is generated by the user of the device through a volitional action and is, in our case, the required ankle-joint position time-history. The response of the musculo-skeletal system to stimulation is $\alpha_2(t)$. Thus information about the actual evoked response is fed back to the controller, allowing any differences between $\alpha_1(t)$ and $\alpha_2(t)$, $\Delta\alpha(t)$, to be accounted for. The controller consists of devices whereby the difference signal, $\Delta\alpha(t)$, is conditioned to provide satisfactory excitation of the muscles to be stimulated. The stimulators amplify the conditioned signals, c(t), and are either constant voltage or constant current sources, s(t). Other feedback paths not shown include vision and possibly reflex pathways.

For an FES-based orthosis to become reality, the following conditions must be met:

(i) The control signal sources should be selected such that the conscious effort required to operate the system does not become prohibitive.

(ii) The control signals must be transduced into an appropriate form. For example, in the case of position control, $\alpha_1(t)$ should be a voltage which varies in proportion to the desired position.

(iii) $\Delta\alpha(t)$ must be transformed into c(t) i.e. the optimal stimulus waveform must be determined and factors such as threshold of the muscle to stimulation and fatigue must be incorporated as will be discussed.

(iv) The form of s(t) must be decided on.

(v) The stimulus waveforms, s(t), must be applied to the muscles by means of some type of electrodes. (surface in our case).

(vi) The evoked responses, $\alpha_2(t)$, must be conditioned into appropriate forms to allow for the incorporation of feedback into the system.

Conditions (iii), (iv) and (v) above are related to sensations due to stimulation, which, if unpleasant, must be minimized.

These conditions will be examined by considering the neurophysiological mechanisms in the intact organism as well as the relevance damage to these mechanisms has to the design of adequate FES systems. The brief historical review that follows will hint at some of the answers being sought after.

2.2 Historical Review of the Use of FES of the Musculo-skeletal System

The source for what follows, up to and including the work of Bordet in 1907, is J.B. Reswick (1973).

The first known report on the application of electricity to muscle to effect contraction was by Kratzenstein in 1745. The previous year he had applied static electricity to the finger of a woman to eliminate contracture.

In 1791 Galvani published the results of experiments on frogs showing a relationship between electricity and muscle contraction. The need for a signal whose amplitude is time-varying for it to be an effective stimulus was observed by Volta in 1799. He applied a continuous electric current noting that stimulation only occurred initially at the time of application and sometimes at the breaking of the circuit. It was Ritter in 1801 who discovered the phenomenon of accomodation of a muscle to a stimulus. Applying a current of varying amplitude he noted that muscle contraction occurred only if the stimulus was applied briskly. Duchenne de Boulogne (1833) laid the foundations of the art of electrostimulation. He discovered that he could stimulate a muscle percutaneously using cloth-covered electrodes, and was one of the first investigators to employ alternating currents. He also realized that stimulation is more effective when applied at specific spots on the surface of the body.

The application of electrical stimulation as a tool in the diagnosis of denervation was first used by Bordet in 1907 when he found that denervated muscle did not accommodate to stimuli.

The use of electrical stimulation as an orthotic aid for paralyzed limbs became reality with the advent of the transistor in the early sixties of this century since small portable stimulators could now be built. Liberson et al (1961) succeeded in effectively preventing drop-foot during the swing phase of walking by stimulating the peroneal nerve. They incorporated a heel-switch into the device as the actuator for onset of stimulation, the heel switch being located in the sole of the shoe on the involved side. To date the basic techniques developed by Liberson are used in the implantable radio frequency controlled, rate dependent electronic peroneal brace (Kralj et al (1971)).

Since the early sixties investigations in electrostimulation of skeletal muscles and their nerves have focused on:

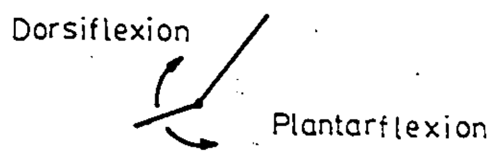
i) neuromuscular mechanisms and their relevance to FES
ii) methods whereby muscles can be stimulated in a controlled manner
iii) assessment of efficacy of FES
iv) design of implantable stimulators and implantation techniques
v) dynamic modelling of stimulated systems and their synthesis.

2.3 Review of Physiology and Anatomy of the Lower Limb

A brief functional description of the human leg is given below. Aspects of the lower leg will be emphasized since this thesis is mainly concerned with the functional control of paralyzed muscles acting about the ankle joint. The principles laid down in this work are sufficiently general that they can in turn be applied to the knee and hip joints. A more detailed description is available in Naumann (1978 (a)) or anatomy texts. The format adopted here is aimed at introducing the system under study to the engineer in terms with which he is familiar.

2.3.1 Muscles of the Lower Leg

(a)



(b)

Figure 2.2 Schematic of lower leg

Figure 2.2(a) is a schematic of the ankle joint and a muscle pair which acts about the joint. Since muscles can develop tension only (Basmajian, 1970), a pair of muscles is needed at a joint to produce rotation in both directions. The bones provide attachments for muscles and give rigidity to the body. They serve as levers in pulley systems whereby movements can be produced by muscles and permitted by joints (Basmajian, 1970).

The muscles of the lower leg can be grouped into posterior (rear or back), lateral (side), and anterior (front) muscles. Only superficial muscles will be considered here. This is because surface stimulation has been employed throughout this work as will be discussed later. The muscles primarily responsible for dorsiflexion and plantarflexion (as defined in figure 2.2(b)) of the ankle are readily accessible to this mode of stimulation.

A single muscle may be called upon to perform more than one function. The classification of a muscle thus depends upon the movement in which it participates. In general, if a muscle is the principal agent in producing a desired movement, it is said to be the prime mover or protagonist. When it opposes a prime mover in order to regulate it, the muscle is termed an antagonist. Finally, a muscle is termed a synergist when it contracts to eliminate some undesired movement that would otherwise be produced by the prime mover (Basmajian, 1970).

Figure 2.3 illustrates some of the pertinent muscles. The gastrocnemius and soleus muscles are both plantarflexors of the ankle joint, each muscle having two heads which unite to form the bulk of the muscles. The gastrocnemius muscle also aids in flexion (or bending) of the knee when the leg is not supporting weight. The gastrocnemius muscle helps maintain extension (i.e. straightening) of the knee during weight-bearing by preventing dorsiflexion at the ankle (Basmajian, 1970); (Hollinshead, 1976).

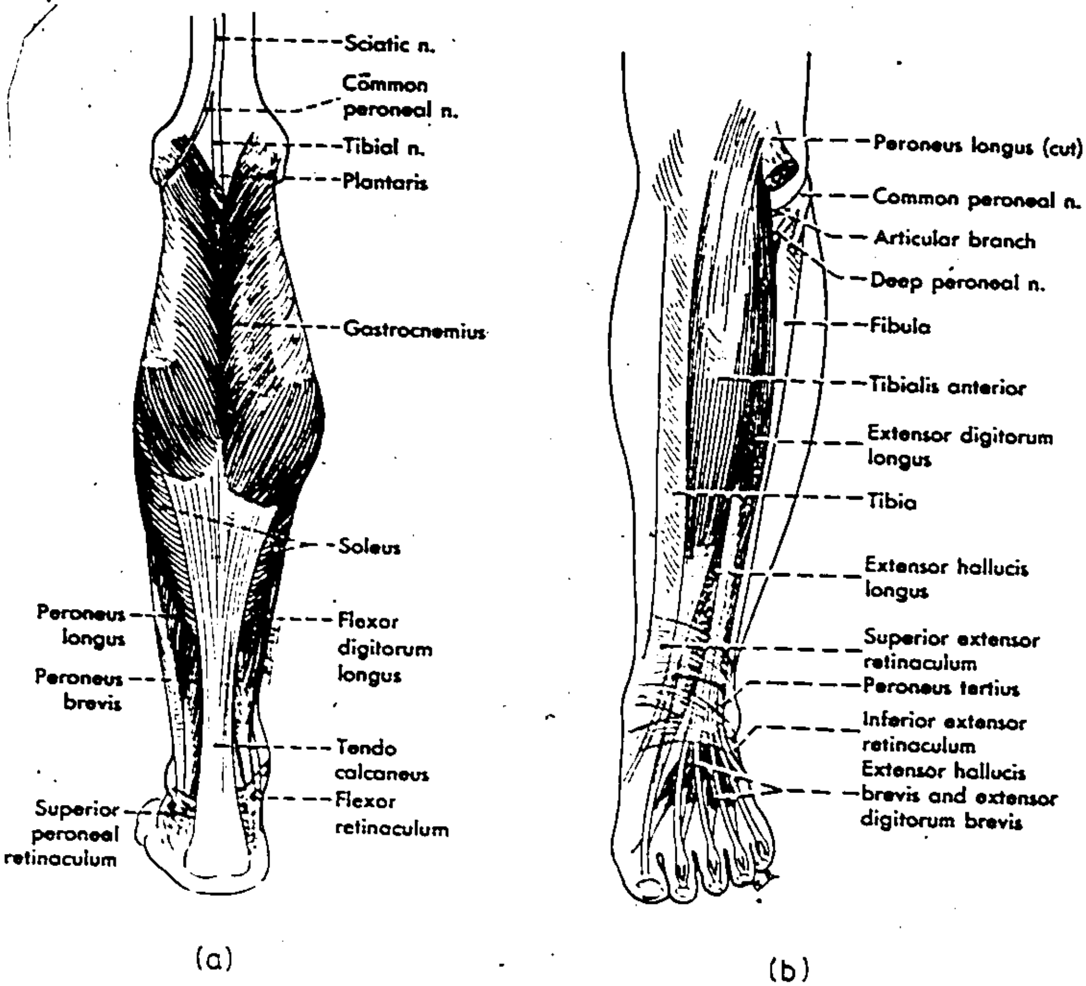


Figure 2.3 Muscles of the lower leg:

(a) Posterior view

(b) Anterior view (taken from Hollinshead (1976)).

The most powerful dorsiflexor of the ankle is the tibialis anterior muscle (see figure 2.3 (b)). Other dorsiflexors of the ankle include the extensor digitorum longus and the extensor hallucis longus muscles.

The group of muscles that effect eversion or turning outward of the foot are the peronei. The tibialis anterior muscle is also a strong invertor (turning in) of the foot.

From the above description of the muscles of the lower leg, it becomes apparent that when a particular movement is effected by means of stimulation, one should rather refer to the movement being evoked than to the particular muscles being stimulated. When surface electrodes are used to effect stimulation, more than one muscle will be stimulated due to current spread. Thus in our terminology, the motor point of a muscle is defined as that point which when stimulated will produce the desired movement with minimum stimulus amplitude.

From figure 2.2 (a) it can be seen that to control joint position, a control signal is needed which can be varied about a preset reference angle resulting in stimulation of either the dorsiflexors or plantarflexors, and is the basis of a system first proposed by Vodovnik et al (1967).

2.3.2 Neural Innervation of the Leg

Each muscle is composed of many muscle fibres attached to the bones via tendons and organized into motor units. A motor unit consists of a motor neuron or single nerve cell situated in a ventral (front) horn of the spinal cord, an attached nerve fibre or axon whose terminal end branches, and several muscle fibres to which the terminal branches attach (see figure 2.4). The terminal branches of the axon end on each muscle fibre in a region called the myoneural junction. The muscle cell membrane in this region is called the motor end plate. An impulse from the motor neuron will cause all muscle fibres in that motor unit to respond. The overall steady contraction of a muscle is achieved by the

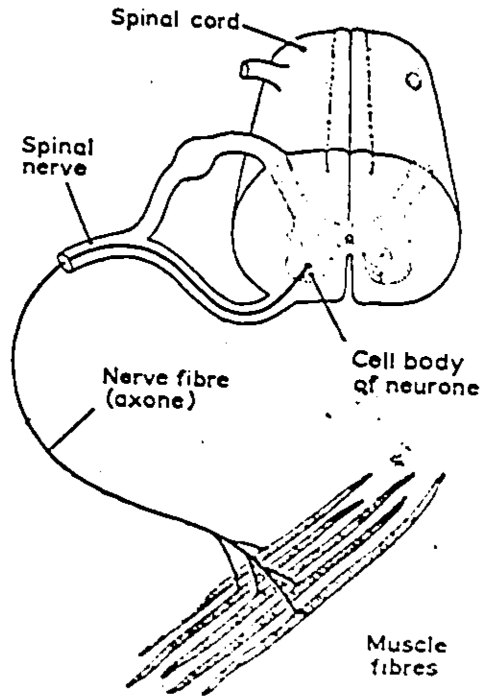


Figure 2.4 Schematic of a motor unit (Basmajian, 1974)

many scattered motor units in a muscle contracting repeatedly and asynchronously (Carlson and Wilkie, 1974); (Basmajian, 1974); (Katz, 1966); (Ruch and Fulton, 1961).

In addition to efferent or motor pathways, afferent or sensory pathways exist. They provide a continuous and automatic feedback of information used to regulate motor activity (Katz, 1966), as will be discussed in section 2.6.

The muscles of the leg and foot derive their neural innervation from the common peroneal nerve and the tibial nerve. Both nerves are segments of the sciatic nerve in the thigh (Hollinshead,1976).

2.4 Morphology and Physiology of Skeletal Muscle and Nerve

A more detailed examination of nerve and muscle structure and function and their relationships to transcutaneous electrical excitation follows. The material presented here is by no means exhaustive and the interested reader is referred to Naumann (1978a) and the references cited here.

2.4.1 The Membrane Theorem

FROG MUSCLE

EXT.	INT.
Na^+ 120	Na^+ 9.2
K^+ 2.5	K^+ 140
Cl^- 120	Cl^- 3-4
	-90mV

SQUID AXON

EXT.	INT.
Na^+ 460	Na^+ 50
K^+ 10	K^+ 400
Cl^- 540	Cl^- 40-100
	-60mV

Figure 2.5 Some electrolyte concentrations (mMl) and potential differences across cell membrane (Katz, (1966) p. 43).

The electrolyte content of a nerve or muscle cell differs greatly from that of the extracellular fluid as shown in Figure 2.5. At rest, the potential difference (p.d.) of the inside of the cell with respect to the outside is 60 to 90 mV negative. A 30 to 40 mV stimulus applied to the fibre will depolarize the surface membrane of the cell reducing its p.d. to an unstable level at which an "all-or-none" response occurs. The level above which a stimulus will cause an action potential (A.P.) response is called the threshold. Subthreshold stimuli produce responses which are quickly attenuated by the cable properties of a fibre. These are capacitive and resisitive leakages through the surface membrane.

When an A.P. is initiated, there is an initial transient increase in Na^+ conductance of the membrane which is regenerative since the increased flow of Na^+ into the cell further depolarizes the membrane thus further increasing Na^+ conductance. Initially conductance of K^+ and Cl^- tends to restore the potential to its resting value. However, if the membrane is further depolarized past the threshold level, the restoring effect of the K^+ and Cl^- is negated by the increased Na^+ permeability and an A.P. results. The opening of the Na^+ gate is a brief transient event and is rapidly followed by an increase in K^+ conductance which operates to return the system to its resting potential. Propagation of the A.P. along the whole length of the fibre is achieved since the potential of the intracellular fluid with respect to the external fluid is sufficient to depolarize the adjacent membrane sections past threshold. The A.P. is thus propagated in both directions from the stimulated region. It does this at a constant velocity dependent on the longitudinal conductance of the fibre core so that velocity is related to fibre size. The influx of Na^+ requires no energy expenditure since the concentration gradient is downhill. However, efflux of Na^+ requires metabolic energy and is the basis of the concept of a "sodium pump". This ionic pump is also necessary for steady state maintenance of the p.d. and is used to expel Na^+ which has leaked into the cell and also help accumulate K^+ in the interior. The result is the large ionic concentration gradients across the surface membrane.

2.4.2 Refractory period and accommodation

The flows of Na^+ and K^+ are not simultaneous, K^+ flow starting after Na^+ flow and continuing past its termination. The refractory period is that period during which the Na^+ gates are shut and the K^+ gates are wide open. This period corresponds to the time during which, if a stimulus occured, no response would be forthcoming. Thus an axon with an absolute refractory period of 1mS can only be driven at rates less than 1KHz.

Accomodation manifests itself in two ways: during passage of a constant current through the membrane K^+ conductance rises and the refractory period increases thus raising the threshold. Similarly, a slowly rising cathodal current will not evoke a regenerative response from the membrane and excitation will not occur due to the outflow of K^+.

Three important restrictions on stimulating waveforms thus exist:

(i) A threshold level exists below which an action potential is not generated. (Related to the threshold level is the duration of the stimulus. This will be discussed in section 2.4.5).

(ii) The frequency of stimulation is limited by the refractory period.

(iii) The rate of change of a stimulating signal must be sufficient to prevent accommodation and produce excitation.

A fourth restriction is due to the fact that since the internal polarization of a cell at rest is negative, depolarization occurs when the potential change in the inner core with respect to the external medium is positive. This condition exists in the vicinity of the negative electrode. Excitation at the anode or positive electrode depends on the "anode-break" mechanism. Here, negative polarization of the fibre is increased by the stimulus resulting in a fall in K^+ current flowing out through the membrane. When the stimulus ends, K^+ flow does not immediately recommence. During the delay, Na^+ flows inward through the membrane for a sufficient time to initiate excitation. The stimulus duration must be sufficiently long (2mS) to cause K^+ flow to fall.

2.4.3 The neuromuscular junction

When a nerve A.P. arrives at each terminal branch on a muscle fibre, a transmitter substance is released which in turn causes the muscle membrane to depolarize locally. The transmitter is acetylcholine (ACh) and is released in individual packets or quanta. The ACh causes a drastic change in the ionic permeability of the membrane resulting in increased Na^+ and K^+ conductance. The end plate potential (epp) which spreads along the muscle membrane is attenuated by the cable properties of the muscle fibre. An A.P. will be initiated by an e.p.p. only if sufficient quanta of ACh are released within a certain time. The A.P. then propagates along the muscle fibre rapidly and without attenuation causing the muscle to contract and tension to be developed. ACh is removed from the end plate region by diffusion and by hydrolysis catalyzed by the enzyme cholinesterase. (Carlson and Wilkie,1974);

(Katz, 1966); (Ruch and Fulton, 1961); (Lale, 1966); (Cooke and Lipkin, 1972).

2.4.4 Muscle Contraction: The Length-Tension Curve

Striated muscle is composed of repeating units called sarcomeres. Within each sarcomere are muscle proteins which allow it to function as a contractile device.

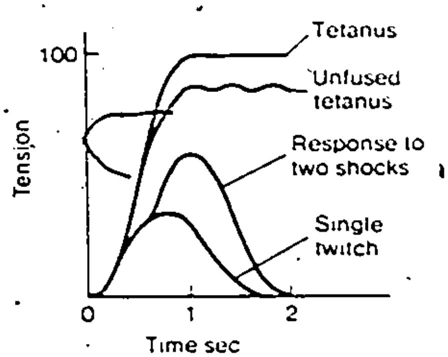


Figure 2.6
Summation of responses following repeated stimulation (taken from Carlson and and Wilkie (1974))

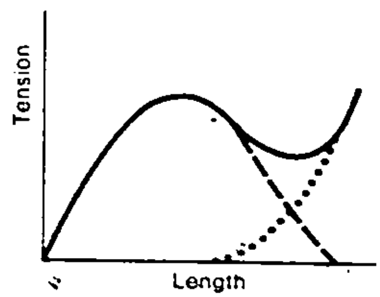


Figure 2.7
TENSION-LENGTH curve
___ total tension
--- active tension
... passive tension
(taken from Carlson and Wilkie (1974))

Consider the case of a whole muscle. If an above-threshold pulse is applied to the muscle, it responds by giving a twitch the time course of which depends on the particular type of muscle being stimulated. By "twitch" is meant a brief period of contraction followed by relaxation. The size of the twitch depends upon the amplitude of the stimulus with no response occurring below the threshold value. Beyond a certain amplitude of stimulus the tension reaches a maximal value. The dependence of tension on current density occurs since, as the stimulus strength is increased, more and more muscle fibres are stimulated until the maximal value is reached when all fibres are being stimulated. The type of muscle can be either slow type 1 or fast type 11 or an intermediary type. (Naumann, 1978a). If a second stimulus pulse is applied to the muscle before the response has died away, summation occurs. A smooth tetanus will result if the stimuli are repeated at a rapid enough frequency as shown in Figure 2.6. However, the magnitude of the tension developed depends upon the length of the muscle at the time the stimulus is applied. The curves in Figure 2.7 were obtained by fixing the length of the muscle and then stimulating and recording the total evoked tension. The passive tension curve was obtained by stretching. This curve is largely determined by the connective tissue which is mechanically in parallel with the contractile fibres. Thus the difference between the two curves is due to the tension developed by the contratcile fibres alone. No tension is produced at the extremes of length since these points are fixed.

Two principles are involved here:

(i) The frequency of stimulation should be sufficient to produce a sustained contraction. Crochetiere et al (1967) found that at a frequency of stimulation of 50 Hz the evoked torque reached a maximum and that the rise time of the response reached a minimum value. Since the present work is concerned largely with a feasibility study, fatigue is not a consideration (see Naumann, 1978a) so that 50 Hz was chosen as the frequency of stimulation.

(ii) Evoked tension depends upon stimulus intensity.

A third controller of tension developed due to stimulation is the pulse duration and is the subject of discussion in the next section.

2.4.5 The Strength-Duration Curve

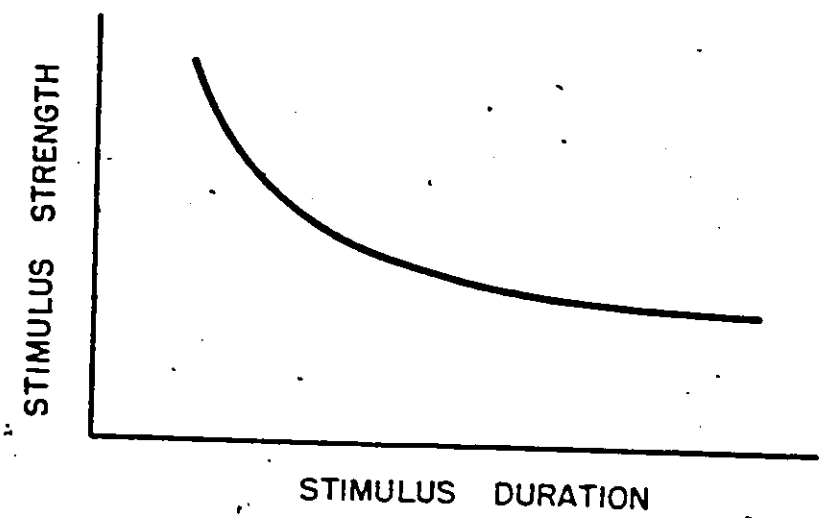


Figure 2.8
STRENGTH-DURATION CURVE

As discussed before, an electrical stimulus produces its effect by depolarizing the surface membrane of the cell to the level where excitation occurs. The current strength which just produces this effect is called the threshold of excitation. The threshold depends not only on the current strength but also on the duration of the stimulus. Since the cell membrane behaves like a leaky condensor (Katz,1966), a minimum amount of charge must pass through it to change its potential to the unstable level (Lale, 1966). The relationship between threshold strength and pulse duration is called the strength-duration (S-D) curve. (Figure 2.8).

The charge required to excite the fibre is proportional to the membrane capacitance (see Appendix A) and is the basis for the clinical application of S-D curves in determining whether a muscle is normally or partially innervated or lacks innervation altogether (Bauman; Shaffer, 1957). The change in internal potential needed to initiate an A.P. is of the same order in muscle and nerve (Katz, 1966). However, the membrane capacitance in muscle is about ten times that in nerve so that the total charge required to depolarize a muscle fibre is one order of magnitude larger than that needed to depolarize a nerve (Ruch and Fulton, 1961). In practice, a pulse of width less than 1mS will excite nerve fibres only.

The author has adopted the classical method of controlling muscle tension by varying stimulus intensity and maintaining a constant pulse width (see also section 2.5.4).

2.5 Electrodes, Pain, and Stimulus Waveform

2.5.1 Comparison of Implant and Surface Stimulation

Jeglic et al (1971) list the advantages and disadvantages of implantable stimulators over those using surface stimulation as follows:

Advantages:

- fixed electrode positon
- constant electrode resistance
- lower energy consumption at output of stimulator
- improved cosmetic appearance
- simplified patient operation of the system

Disadvantages:
- surgical procedure
- danger of infection
- electrode corrosion
- increased reliability required of the implant
- foreign body reaction.

While all the above factors are important when considering the implementation of developed stimulators, the advantages of using surface stimulation for developmental work far outweighs its disadvantages. These advantages are:

(i) Different types and sizes of electrodes can be readily used.

(ii) Electrode positioning can be easily varied.

(iii) Freedom exists in choosing the muscles to be stimulated.

To effectively use surface stimulation, electrode position with respect to the motor point and pain must be considered. A further imposition is that only those muscles whose motor points or innervating nerves are close to the skin surface can be stimulated. Possibilities exist using multiple surface electrodes to manoeuvre the effective point of stimulation by applying stimulation levels of different intensities to the various active electrodes. A second method is to input two high frequency signals via two sets of electrodes. The effective frequency of stimulation will be the difference between the frequencies of the two input signals. A device which operates on this principle is the Nmectrodyne (Wyss,1975). However, the author has decided to use the classical method of applying stimulation since the possiblities described above are research projects within themselves. Their realizations will not invalidate the control methods developed in this thesis.

2.5.2 Pain

Sensory end-organs for pain are spread throughout most of the tissues of the body. Pain receptors are stimulated when a threat of damage to the tissues occurs. Somatic pain is classified as superficial or cutaneous pain, and deep pain from muscles, tendons, joints and fascia. The latter type of pain is common to both surface and intramuscular stimulation and manifests itself as a dull ache that can become intolerable. It is dependent on tension developed and its time course (Ruch and Fulton, 1961).

To prevent atrophy of denervated muscle, electrical stimulation was found to be therapeutic only when vigorous muscle contractions were obtained. (Iddings et al, 1951). High levels of stimulus current are needed when using surface stimulation thus eliciting cutaneous pain. This led to studies of methods whereby a patient's tolerance to higher levels of current could be increased. No relief of pain was achieved when using infra-red heating or electrodes having different resistivities (Iddings et al, 1951). The utilization of novocain to effect a sensory block was found to increase tolerance to higher current levels. However, tension developed either decreased or remained unchanged (Gersten et al, 1954).

Other studies (see below) have examined the relationship between electrode size and stimulus waveform and pain. The difficulty in evaluating such studies is that pain is a subjective phenomenon and therefore difficult to measure quantitatively.

2.5.3 Electrodes

At least two electrodes are needed for stimulation: an active electrode placed close to the desired region of stimulation, and an indifferent electrode at whose site stimulation should be minimized. Making the active electrode smaller in area than that of the indifferent electrode ensures a higher current density at the region of excitation. Limiting factors in size of the active electrode are: current density should be sufficiently low to prevent burning, pain must be minimized, and spread of stimulus to other muscle groups must be prevented. Electrodes should also fulfill the following requirements:

(i) They should be simple, inexpensive and easy to make in different sizes.

(ii) Their application and removal should be easy and quick.

(iii) They should be flexible enough to conform to body contours and thus be capable of providing good electrical contact with the skin.

(iv) They should be nonirritating over a period of a few hours.

Milner et al (1970) describe such an electrode. They used a fine stainless steel mesh which held electrode paste (Redux*) well. The electrode was held in place by masking tape. They found that pain elicited depended on the total peak current but not on electrode size and hence current density. (If the current density is sufficiently high, burning accompanied with pain will result). Maximum force generated

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*Redux paste: Hewlett Packard, Part No. 651-1008, Waltham, Mass.,U.S.A.

was dependent on electrode size. For the tibialis anterior muscle, maximum values were obtained with 13 to 26 cm^2 electrodes.

It has been the author's experience that trauma due to electrolyte irritation can be avoided by thoroughly cleansing the stimulus site on removal of electrodes. Masking tape was also found to be an irritant and is now placed over a gauze bandage wrapped around the leg and electrode. It was also found that placement of the electrode over the motor point of the muscle minimizes noxious sensations. (The motor point of a muscle is that site at which minimum stimulus amplitude is required to effect a desired movement about a joint). Increase in pain due to poor placement of electrodes was also found to occur by Trnkoczy and Gracanin (1971).

During movement, the muscle and hence the motor point shifts with respect to the skin. Methods have been devised for tracking the motor point in the case of the upper limbs (Crochetiere, 1967); (Beresford et al,1975). However, in the lower limb where muscles are much longer and movements have less finesse, this should not be a problem if the electrodes are large enough.

2.5.4 Stimulus Waveform

Vodovnik et al (1965a) investigated the relationship between pain and different tetanizing stimulation currents in normal subjects. They found that AC currents were comfortable at frequencies above 500 Hz. AC waveforms are rejected for two reasons: the power consumption of an AC stimulator is much more than that for a DC stimulator, and as is discussed in Naumann (1978a) muscles stimulated at high frequencies fatigue faster than those stimulated at lower frequencies. Using D.C.

rectangular pulses, Vodovnik et al (1965) found that short pulses with long rest periods are more comfortable than long pulses with short rest periods. The limits of pulse duration were found to be between 0.1 and 0.3 mS at a frequency between 20 and 40 Hz. Crochetiere (1967) fitted their experimental data to the following curve:

$$R < 25 + 45.5 \log D$$

where R = rest period or pulse interval and D = pulse duration in mS.

Crago et al (1974) compared the minimum energy or charge transfer when using rectangular current waveforms and exponential current waveforms as the stimulus. They found that the latter waveform required higher peak currents than the former but dissipated less energy. Since pain is determined by peak current (Milner et al, 1970) and rectangular pulses are simple to generate electronically, this waveform is most widely used in FES orthoses.

Two types of stimulators are in general use: the constant voltage stimulator ~~and the~~ constant current stimulator. The impedance of the load (equivalent electrode-subject-electrode circuit) contains capacitive elements. Using a constant current stimulator, the voltage waveform is characterized by an exponential rise and fall i.e. the voltage rises slowly and falls slowly and is not constant at any time. The voltage waveform from a constant voltage stimulator has a fast rise and fall time and is almost constant during the time it appears. Burton and Maurer (1974) note the disadvantages of constant voltage stimulators compared to constant current stimulators when used to stimulate transcutaneously:

(i) The low frequency components (flat portion) of the constant voltage square wave increase stimulation of cutaneous pain receptors.

(ii) The high frequency components (fast rise and fall times) contribute to depolarization of deep sensory nerve pathways and hence the "tingling" sensation felt distal to the electrodes.

(iii) The high pass characteristic of skin reduces the net charge transfer to the peripheral nerve.

Crochetiere (1967) also notes that changes in electrode impedance do not affect the current through the electrodes when using constant current stimulation.

Trnkoczy and Gracanin (1971) found that charge flowing through a muscle for each stimulus is the critical parameter for pain sensation. This is in seeming contradiction to Milner et al (1970) who found that maximum instantaneous current governs pain sensation. However, Milner et al (1970) used stimuli of constant pulse width. In contradiction to Burton and Maure (1974), Trnkoczy and Gracanin (1971) found a preference to constant voltage stimulation over that of constant current. Because of the capacitive load presented to the stimulator, using a constant voltage stimulator will cause distortion of the current pulse so that the net charge is less than that obtained when using a constant current source. Trnkoczy and Gracanin (1971 also found that in the ten subjects they tested, no difference in effect of waveform (either rectangular or exponential) on pain was noted.

2.5.5 Stimulus Parameters

Based on the preceding discussions, selected stimulus parameters are:

(i) Stimulus amplitude control of muscle tension at a constant pulse width of 0.2 mS (sections 2.4.5 and 2.5.4)

(ii) 50Hz frequency of stimulation (section 2.4.4)

(iii) Rectangular pulses since these are easy to generate electronically (section 2.5.4)

(iv) Constant current stimulation to ensure that the applied stimuli are independent of impedance changes in the electrode-skin-electrode interface. (section 2.5.4)

(v) Stimulation applied at the cathodal electrode (section 2.4.2)

2.6 Neuromuscular Organization and its Relevance to FES

An understanding of the central nervous system (CNS) and its organization and the effects of injury to that system, are essential if useful rehabilitation aids are to be designed to replace lost function. In particular, when FES is employed, muscle as well as a partially intact nervous system are being stimulated. Thus the adaptive nature of the neuromuscular system must be considered. The following section provides a review of the overall neuromuscular system as well as detailed descriptions of some of its mechanisms. The results of injury to the system and the implications such injury has for the design of rehabilitation aids will be considered.

Except where otherwise referenced, the source of much that follows is Ruch and Patton (1966).

2.6.1 Servomechanisms of Muscle Control

Two of the most important spinal reflexes are the flexion reflex and the stretch reflex. The flexion reflex is responsible for contraction of the ipsilateral (same side) flexor muscles at the ankle, knee and hip in response to a noxious stimulus. Simultaneously, extensor muscles relax, and contraction of the extensor muscles and relaxation of the flexor muscles occurs on the contralateral extremity. The stretch reflex is the result of stretching a muscle and manifests itself as a smooth, sustained contraction of the stretched muscle. Concomitant relaxation of antagonist muscles occurs. The stretch reflex is mediated by monosynaptic arcs while the flexion reflex is mediated by multisynaptic arcs. A third important reflex is the clasp-knife reflex which occurs when a muscle is excessively or rapidly stretched. It causes the stretch reflex to disappear, the antagonists to contract, with occasional concomitant contraction of the extensors on the contralateral side.

An examination of the mechanisms responsible for these reflexes will lead to some understanding of the organization and function of the CNS.

2.6.2 Receptors in Muscle and Tendon

(i) Muscle spindle (see figure 2.9)

Skeletal muscle comprises of extrafusal (outside) fibres which act as prime movers, and intrafusal fibres interspersed throughout the muscle. Each muscle spindle consists of bundles of 2 to 10 intrafusal fibres. The intrafusal fibres are striated and contractile. The central region of the spindle is a nuclear bag having three types of nerve fibres going to it: large myelinated afferent (sensory) fibres which end in unmyelinated helical terminals (annulospiral); smaller, myelinated fibres ending in coils at one or both sides of the nuclear bag endings (flower spray); and small, myelinated efferent (motor) fibres terminating in end-plates on the striated poles of the intrafusal fibres (fusimotor fibres or γ efferents). The nuclear bag is non-contractile.

Stretch stimulation of the receptors can occur in one of two ways: the muscle can stretch, or the efferent neurones can cause the intrafusal fibres to contract. Either method causes tension of the nuclear bag and hence distortion of the afferent nerve endings. The muscle spindle will thus discharge during stretch. Because the muscle spindle is in parallel with the extrafusal fibres, when the muscle shortens, tension on the intrafusal fibres is relieved and the spindle goes on slack. Contraction of the fusimotor fibres via efferent innervation in the absence of, or during external stretch, increases the sensitivity of the spindle so that afferent discharge is markedly increased in frequency. The fusimotor system thus serves as a biasing mechanism regulating the sensitivity of the receptor, while the annulospiral afferents (Ia primary) register exact information on muscle length.

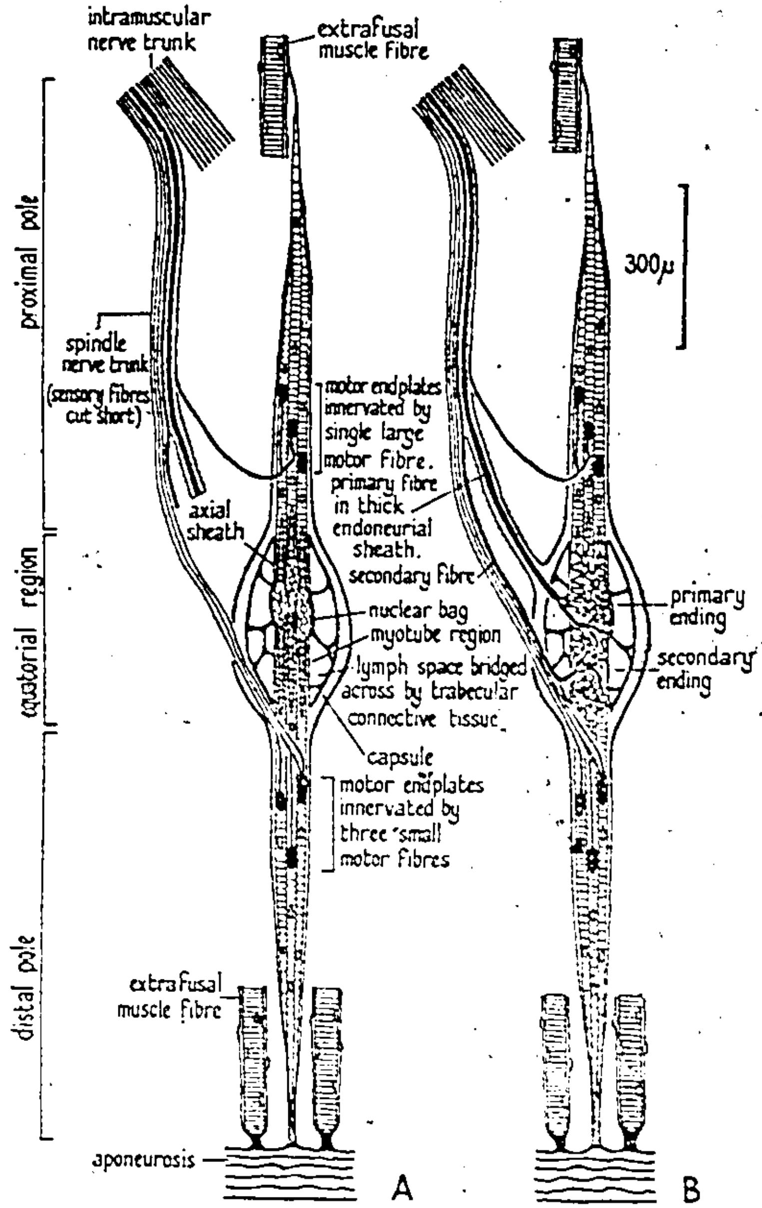
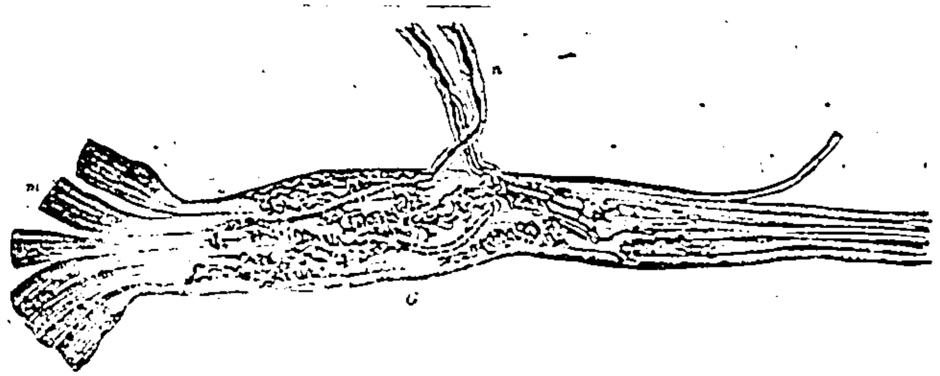


Figure 2.9
The muscle
spindle and its
nerves
(taken from Ruch
and Fulton, 1961)

Figure 2.10
Golgi Tendon Organ

m - muscle fibres
t - tendon
n - nerve fibres
G - end organ
(taken from Ruch
and Fulton, 1961)



The muscle stretch receptors have a low threshold to muscle stretch whereas the Golgi tendon organ stretch receptors have a high threshold.

(ii) Golgi tendon organ (see figure 2.10)

The Golgi tendon organ is found in the tendons of muscles close to their origins. The tendon organ is thus in series with the muscle. One or two myelinated nerve fibres penetrate the enclosing fibrous capsule and then break up into smaller branches, lose their myelin sheaths, and terminate in the tendon bundle.

Tension on the tendon distorting or displacing these endings will result in receptor discharge. This occurs when the muscle contracts. Discharge frequency is proportional to the applied muscle tension. (However, discharge frequency is more responsive to tension due to muscle contraction than stretch.)

2.6.3 Synaptic Function

The action potential (A.P.) has been previously described (see section 2.4) and is the only mode of expression available to the nervous system. The simplest experiences and actions derive from the conduction of A.P.'s over chains of neurons linked by synapses. It is the activity that occurs at the synapses that determines whether an axon impulse will be generated or not. This activity can either be excitatory or inhibitory in form depending upon the source of the afferent volley. The generation of an axon impulse thus depends upon the sum of the excitatory and inhibitory impulses arriving at the neuron at any instant.

There are two types of reflex arcs: the monosynaptic arc where afferent fibres enter the dorsal root of the spinal column and synapse with the motor neuron in the anterior horn (single synapse), and the multisynaptic arc where afferent fibres terminate on neurons in the dorsal root and reach the motor neuron only after transfer through one or more interneurons (internuncial neurons). Thus the motor neuron constitutes the final common path upon which many presynaptic fibres converge.

A movement can occur when a command signal is generated in the motor cortex and transmitted down the spinal cord via an upper motor neuron. The signal is transmitted to the muscle from the spinal cord via an α or γ lower motor neuron. As previously discussed, innervation causes the intrafusal fibres of the muscle spindle to contract and is the process whereby initiation of a movement may occur. The biased spindle will produce an afferent discharge which flows back to the spinal cord along group II (length change) and Ia (rate of change of length) afferent fibres. The signal reaches the motor neurons via the interneurons resulting in the contraction of the extrafusal fibres. Afferent discharge ceases when the muscle has contracted sufficiently to relieve tension on the muscle spindle (Granit, 1973). (Figure 2.11).

The Golgi tendon organ will only fire when the extrafusal fibres contract. The effect of this afferent discharge is inhibition. If stretch of the muscle is excessive or rapid, inhibition is sufficient to prevent firing of the motor neuron (autogenic inhibition). The role of the Golgi tendon organ is thus a protective one.

2.6.4 The Internuncial Pool

The muscle spindle reflex arc is monosynaptic while that of the Golgi tendon organ is multisynaptic. The monosynaptic reflex arc thus originates in a particular muscle and can only discharge the motor neurons supplying the muscle from which the afferent volley originates. However, it does alter the excitability of motor neurons supplying other muscles (heteronymous motor neurons) since each dorsal root fibre breaks into many branches which establish contact with many postsynaptic cells.

The way in which the excitability of a heteronymous motor neuron is affected depends upon the relation of its target muscle to the muscle from which the afferent volley originates. Consider a motor neuron pool. Depending upon whether a particular motor neuron receives many knobs from an activated afferent source or not, that motor neuron will either be liminally excited or subliminally excited. A second afferent volley from a different source may combine to further raise the excitability of the subthreshold motor neurons to threshold. This effect is known as facilitation and is responsible for raising the excitability of motor neurons innervating muscles synergistic to the muscle of afferent discharge origin.

Inhibition, due to an excitatory volley in one pathway, renders neurons less responsive to a subsequent excitatory volley arriving via a second pathway. Inhibition results in muscles antagonistic to the prime mover relaxing. This process is known as reciprocal innervation and ensures that reflexly induced muscular contractions occur without opposition.

Multisynaptic reflex arcs differ from monosynaptic ones in the following ways:

(i) The time course of facilitation and inhibition is more complex since impulses must traverse chains of interneurons thus subjecting the motor neuron to a variable and asynchronous barrage of impulses.

(ii) Irrespective of origin, multisynaptic afferents are probably excitatory to motor neurons supplying ipsilateral flexor muscles and inhibitory to motor neurons innervating ipsilateral extensor muscles.

The multisynaptic reflex may thus originate from widely dispersed afferent fibres, and is distributed diffusely to motor neurons supplying muscles acting at different joints.

The flexion reflex which is multisynaptic in form, has been previously described. The flexion reflex is also responsible for contraction of extensor muscles and relaxation of flexor muscles on the contralateral extremity. This is due to afferent fibres subserving the flexion reflex sending collateral branches to the opposite side of the spinal cord. Thus as the ipsilateral limb is reflexly withdrawn in response to a noxious stimulus, the contralateral limb supports the weight of the body.

2.6.5 The Nature of Higher Control

Reflexes can be influenced by brain centres in the following ways:

(i) Facilitation or inhibition of α motor neurons which innervate the majority of muscle fibres.

(ii) Facilitation or inhibition of γ motor neurons causing contraction of the intrafusal fibres of muscle spindles thereby increasing the rate of spindle firing which in turn influences the amount of α motor neuron firing.

The brain stem is the origin of descending pathways, vestibular and reticular (ear and lower brain stem), that facilitate myototic reflexes of extensor muscles and inhibit flexor muscles.

The cerebral and cerebellar cortex excites the extensor inhibitory reticular system. The reticular extensor facilitatory system receives impulses from ascending afferent systems including those originating in the muscles. Vestibulospinal pathways facilitating extensors are activated through the labyrinth.

Most muscular activity occurs as motor patterns developed during childhood. These activity patterns become more refined and active with age and practice, and appear to be developed as engrams in the extrapyramidal system and are not under direct volitonal perception or control. Voluntary control of precise muscular activity can be achieved only if attention is directed to the activity of one muscle at a time, rapid performance of the activity occurring only after prolonged practice has developed engrams in the extrapyramidal system (Kottke, 1976).

2.6.6 Clinical Significance of Reflexes

When the spinal cord is severed, all muscles innervated from segments below the transection become paralyzed. The cord sectioning produces spinal shock which results in the suppression of all reflexes below the transection up to the first two weeks following injury. Both inhibition and lessening of facilitation may contribute towards spinal shock as follows:

(i) Descending pathways which normally subliminally excite motor neurons, keeping many near the point of discharge, enable a local afferent volley to discharge many motor neurons lying in its subliminal fringe. Thus terminating descending impulses will result in the withdrawal of facilitation from anterior horn cells.

(ii) If descending impulses are interrupted, the inhibitory influence acting upon the interneurons of an antagonistic reflex arc are removed. Thus an afferent volley from a flexion reflex would travel unreduced through the interneurons and inhibit motor neurons of the antagonistic extensor reflex.

With the passage of time, some reflexes return and hyperactivity may occur. The results of hyperreflexia are spasticity and clonus. Spasticity manifests itself by increased briskness and amplitude of deep reflexes and by increased resistance to passive flexion at the joints and indicates the destruction of descending tracts inhibitory to the segmental stretch reflex mechanism. Clonus, where the motor neurons discharge in periodic, synchronous bursts, is due to the stretch receptor being in parallel with the muscle. A tap on the tendon initiates a synchronous volley of afferent impulses resulting in a jerk contraction which relieves the spindles of tension imposed by sustained stretch.

The spindles thus cease firing with the result that the afferent drive to the motor neurons ceases so that the muscle relaxes. This puts the spindles under tension again so that an afferent volley is again initiated and the process repeats itself.

2.6.7 Recovery From Spinal Shock

Recovery from cerebral vascular disease may be partially due to the release of pressure enabling areas neighbouring on the destroyed area to refunction. One of the main mechanisms of recovery may be axon sprouting. It is known that subsequent to spinal transection, the local posterior root fibres sprout new collaterals and produce more synaptic connections with motor neurons and interneurons. This would increase the magnitude of a reflex response to the same peripheral stimulation. There is evidence that sprouting takes place not only in the spinal cord, but also in the brain (Lynch et al, 1976), resulting in the formation of new, functional synapses. The mechanism regulating post-lesion sprouting is unknown, but sprouting may be responsible for both partial recovery of normal function and the development of spasticity.

Recovery is further complicated by the fact that accident induced lesions are usually incomplete so that not all descending pathways are interrupted.

2.6.8 Relevance to FES

Knowledge about information processing in the nervous system is at present incomplete and thus limits the use of conventional control theory in the design of rehabilitation aids. However, control of the skeletal system can be divided into different levels as follows (Tomovic and Bellman, 1970; Tomovic, 1971):

(i) Fixed programs: These consist of invariant, repetitive activities whose timing patterns are acquired through learning. Such activities include manipulation and locomotion.

(ii) Adaptive response: Even if the same pattern of activity is repeated, different speed, load and environmental conditions can exist. Adaptive control is provided by the feedback loops which influence the peripheral neuromuscular system so that the necessary tension and length is automatically maintained.

(iii) Decision-making: (i) and (ii) above can be considered as decision implementation. Decision-making is conscious control and is concerned with the selection of a course of actions to be followed and provides the trigger signal which initiates or terminates implementation of these actions.

The skeletal system can thus be considered as a multilevel control system whose subsytems are autonomous and which follows the principle of minimization of data processing. This approach greatly simplifies the design requirements for rehabilitation devices. Fixed programs can be used to implement different actions. Dynamical control can be provided for by inbuilt servomechanisms. Conscious control is then limited to the choice of a certain course of action and to its initiation and termination. The conscious effort exerted by the user of such aids will be minimized thus increasing the acceptance of such aids by patients.

Efferent FES is aimed at the stimulation of paralyzed muscle to evoke functionally useful movements. Because the threshold of excitation for motor nerves is higher than that for sensory nerves, efferent FES will cause activation of the afferent limb of the mono- and multisynaptic reflex arc resulting in augmentation of motor activity. Secondly, because efferent FES bypasses the protective mechanisms such as the Golgi tendon organ, injury may result when sudden environmental changes dictate immediate cessation of the movement (Dimitrijevic et al, 1968).

The first disadvantage of efferent stimulation noted above can be overcome by employing feedback in the system so that the intensity of stimulation can be regulated by the electromyographic response elicited from the stimulated muscle. While still theoretical, the solution to the problem of sudden environmental changes will entail monitoring and transducing these changes so that they can be accounted for in the control system. A second solution to this problem is the use of afferent stimulation. Due to their lower threshold, it is possible to selectively stimulate afferent fibres. Afferent stimulation has the added advantage of evoking facilitation and inhibition. Dimitrijevic et al (1968) have used afferent stimulation which introduces a facilitatory effect to enable a patient to voluntarily extend his wrist. They have also demonstrated the ability of afferent stimulation to inhibit clonus due to reciprocal inhibition. Present limitation of afferent FES are:

(i) The inability to selectively stimulate the muscle spindles.

(ii) Habituation viz. a progressively decreasing response to stimulation which occurs at the receptors and the interneuronal system of the spinal and brain stem levels.

A therapeutic effect of FES that remains largely unexplained is the prolonged post-stimulation increase in voluntary power in foot dorsiflexion that has been reported by many researchers (Liberson et al, 1961; Vodovnik and Rebersek, 1973; Carnstam et al, 1977). Vodovnik (1971) postulated that stimulation affects redundant cells so that they can replace destroyed cells. His model incorporates learning and forgetting capabilities since the carry-over effect of stimulation is transitory.

2.7 Conclusions

In describing the basic components of an FES-based orthosis, the relevancy of the underlying neurophysiological mechanisms in man has been stressed. An understanding of these mechanisms has led to:

(i) The determination of stimulus parameters

(ii) A description of the hierarchical control organization of the skeletal system and its implications in the design of orthotic devices.

CHAPTER 3

AN ASSESSMENT OF CONTROL BY ELECTROSTIMULATION OF PARALYZED LIMBS

The first half of this chapter is a literature review of proposed methods for the control of orthotic devices employing FES. The concept of 'one-to-one' contralateral control follows as a logical extension of previously proposed control systems when considered together with the concept of a hierachical control organization.

The second part of this chapter proposes a method, based on that of Hershler and Milner (1976), whereby EMG control of FES can be assessed. This method is also applied to the quantification of other variables as will be elucidated in chapters 4,5, and 6.

3.1 Generation of Control Signals

Various approaches to the generation and processing of control signals exist and will be considered. The control approaches described here can equally be applied to the control of orthotic and prosthetic devices. However, their application to FES orthoses will be stressed. It must be kept in mind that the availability of control signals depends on the severity of the paralysis or the extent of amputation in each individual. In general, the greater the extent of paralysis, the fewer the number of control sources available and the greater the number of muscles that have to be controlled.

3.1.1 On-Off Control

The simplest mode of control is the binary On-Off switch which has been extensively used in conjunction with the peroneal nerve stimulator for correction of drop-foot during the swing phase of gait in hemiplegics (Liberson et al, 1961; Kralj et al, 1971a). The trigger-switch is attached to the shoe so that on toe-off of the affected side, stimulation is effected and dorsiflexion of the foot occurs. Switches of the following forms have been used:

i) A contact switch placed in the heel of the shoe (Liberson et al, 1961).

ii) A contact tape switch inserted in a shoe insole (Jeglic et al, 1971a).

iii) A pneumatic control switch worn inside the shoe (Van Leeuwen and Vredenbregt, 1969).

The possibility of using more than one foot-switch exists for the control of more complex systems. The main advantage of the foot-switch when compared with certain other control modes used in FES of the lower limbs is that no conscious effort is required by the user to initiate or terminate stimulation. Its use to date has been limited to applications where the sequence of stimulation is preprogrammed.

The foot-switch has also been used as a feedback transducer in that it provides information on stepping frequency. Thus duration and onset of stimulation can be adjusted according to step rate (Kralj et al, 1971a).

3.1.2 Myoelectric Control

The electrical activity of a muscle due to the spatial and temporal summation of A.P.'s during the contraction of the muscle can be recorded using surface or intramuscular electrodes. The use of the electromyographic activity of contracting muscles as a control source for FES was proposed by Vodovnik et al (1965b). Since then, EMG has been used in various forms and modes as a control source:

i) Control of a stimulator due to contraction of an auxiliary muscle: Vodovnik et al (1965b) used the EMG activity recorded from the left trapezius to modulate the output of a stimulator applied to the right extensor digitorum muscle of a quadraplegic patient. This enabled the patient to achieve eight distinct levels of opening or closing his hand by contracting his shoulder muscle.

ii) Synergistic control: Synergists are muscles that contract together with a protagonist muscle when it executes a particular movement. Their function is to aid in the achievement of precise coordination. In certain cases it is possible to record EMG from muscles of the paralyzed limb still exhibiting signs of voluntary movement synergistic with the paralyzed muscles (Pennacchietti et al, 1966).

iii) Contralateral control: In hemiplegia where essentially one side is incapacitated, control signals are available from the contralateral side. This promises to become useful in controlling walking in a straight line. It may be possible to record EMG signals from a particular muscle on the contralateral side, and after appropriately delaying them, use them to control a stimulator applied to the corresponding muscle on the affected side. (Naumann and Milner, 1978a).

The merits and demerits of EMG control will be discussed in section 3.1.8

3.1.3 Position Control

Proposed by Long and Masciarelli (1961), position control has been used for upper extremity control. This method consists of transducing elevation-depression and protraction-retraction movements of the non-involved shoulder of a hemiplegic via linear potentiometers. Thus two channels of information are available for control purposes. Such a system was implemented by Merletti et al (1975) for controlling opening of the hand and elbow extension. Evaluation of the orthosis consisted of moving an object from one location to another. A high degree of mental concentration was required by the subjects to successfully perform the task.

A similar system incorporating a hold switch for tonic activity was implemented by Peckham et al (1974). It enabled the extremity to remain in a fixed position independent of shoulder movement when the hold switch was activated.

3.1.4 Other Control Sources

Control sources other than those mentioned above have also been investigated. Vodovnik and Rebersek (1974) found both EMG and pressure control to be inferior to that of position control when used as a discriminator between signal levels. Crochetiere et al (1969) describe a transducer whose output is proportional to deflection on the surface of the skin.

The best type of control source to use will depend upon the fixed programs (see section 2.6) viz. the courses of actions that the signals have to control, and upon whether adaptive control is provided for. These in turn depend upon the severity of the injury sustained and whether upper or lower limb control is being considered. These points will be elucidated in the succeeding sections.

3.1.5 Fixed Routines

The simplest form of a preprogrammed fixed routine is the peroneal nerve stimulator (Liberson et al, (1961) where stimulation will occur for a predetermined time on activation of a heel switch.

Milner and Quanbury (1969) demonstrated the ability of FES to control movements about two joints simultaneously. They used a 4-channel programmable stimulator to control the quadriceps, tibialis anterior and gastrocnemius muscles in a normal subject. This method depends upon a knowledge of the prime movers involved in a specific movement, their sequence of activation, and the period each muscle is in an active state. This method of gait evaluation and stimulation sequence determination has been extensively employed in the control of gait of hemiplegics. Kralj et al (1974) describe a rate-dependent 3-channel programmed stimulator, and Strojnik et al (1977), a 6-channel programmable stimulator for use in locomotion improvement in hemiplegics. Recently Stanic et al (1977) reported on a programmable stimulator using a gradually varying stimulation sequence. Significant improvement in the gait of two hemiplegic patients led them to suggest that multichannel continuously modulated sequences of stimulation would further improve hemiplegic gait.

In the systems described above as well as those using EMG or pressure control in upper-limb orthoses, the most serious drawback to them is the inability to incorporate dynamic control. In gait, the only variable conducive to some form of feedback control in the above systems is the rate of stepping. Two control modes that theoretically enable the incorporation of dynamic control using the principles of servo-mechanisms are position control in both upper and lower limb systems, and EMG in lower limb systems.

3.1.6 Position Control of a Joint by Stimulation of Antagonists

Vodovnik et al (1967) reported on a system whereby the elbow-joint angle could be controlled by stimulation of the biceps and triceps muscles. A signal proportional to the desired position was fed into the control system. An electrogoniometer (see section 4.3) was used to measure the elbow-joint angle. The elbow angle was set and assigned zero reference so that in this position with zero input no stimulation occurred. Varying the input signal about this reference angle resulted in stimulation of either the flexor or extensor muscles. Feedback was incorporated by comparing the actual position of the joint to that of the desired position. The error signal due to any difference between the two angles was then used to modulate stimulation to the appropriate muscle to minimize the error. No quantitative assessment of the device was given. A similar system incorporating a proportional-integral filter to reduce static errors is described by Stanic and Trnkoczy (1974) for positioning of the ankle joint. The purpose of their work was to determine the model for the ankle joint to be incorporated into a synthesized multilevel controller.

5.1.7 EMG Control of Joint Position Using Feedback

The use of processed EMG recorded from an auxiliary muscle as the control signal in an FES orthosis which employs processed evoked EMG as the feedback signal was first reported in 1969 by Bunimovitch and Aleyev. Their system incorporated a commutator for eliminating stimulus artifact, an averager, and a modulator. Previous to this, other sytems reported on which utilize EMG either as the controller or ~~as~~ the feedback signal, include those of Amosov (1967) and Aleyev and Bunimovitch (1968). In the former, a preprogrammed stimulation sequence was recorded on magnetic tape. The playback head of the tape-recorder was then connected to a pulse generator by which stimulation was effected. Recording electrodes placed on the stimulated muscle were used to pick up evoked EMG which was used as the feedback signal. The latter system again incorporates a tape-recorder, the control signal being processed EMG. No feedback was provided for. The above systems were reported on without any detailed description nor any quantitative results being given.

Chandler and Sedgwick (1973, 1974) demonstrated that the integral of the rectified EMG response to stimulation is linearly proportional to stimulus amplitude in the region between threshold and saturation. They thus constructed a closed-loop system employing processed evoked EMG as the feedback signal. The control signal used was derived from a variable voltage source. It was proposed that the EMG signal from some other muscle be used as the control signal, with the input to the stimulator being reduced proportional to the level of EMG activity in the antagonist muscle.

3.1.8 Comparison of EMG and Position Control Systems

The system to which the controller is applied largely determines its efficacy. For example, the merits of a controller of upper limb prostheses or orthoses where intricate movements of several degrees of freedom are required differ greatly from those where a gross repetitive movement is needed such as in locomotion. The limiting factor in the first case is the number of discrete levels to which the control signal can be adjusted when compared with other control modes.

In particular, Radonjic and Long (1970) describe the underlying reasons for the difficulties encountered when using myoelectric control. It must be stressed that they are mainly considering the use of auxiliary muscles as the control sites for the manipulation of artificial joints. The reasons they give are:

i) Every voluntary act is a combination of the cooperative active action of a number of muscles. These muscles behave according to a pattern of innervation underlying each motor act which varies depending upon initial posture and resisting forces. The basic pattern of innervation is established through a learning process. In myoelectric control where the control muscle is often no longer expected to contract according to its normal pattern of innervation, it becomes difficult to use the muscle to obtain a new function.

ii) Voluntary control of movements is the control of position and not of muscle tension to which EMG is related. Muscles in the normal case adjust their lengths automatically so that the only conscious feedback control available to the user of a myoelectrically controlled device is visual. For externally powered assistive devices, EMG has thus far not been shown to contain information on position.

iii) Because of the complexity of the system to be controlled, such a system absorbs all the manipulative capacities of a patient.

3.1.9 Proposed System

An entirely different approach is proposed when dealing with FES of the lower limbs. (Naumann and Milner, 1978a); (Naumann and Milner, 1978b). As mentioned in section 3.1.2, a one-to-one control possibility exists in the case of a hemiplegic walking in a straight line. Other actions such as sitting and standing will require elimination of the delay imposed between recording of the control signals and activation of the stimulators. Turning might be accomplished by recognizing a change in the EMG patterns on the contralateral side and switching in a preprogrammed routine to execute the function. In paraplegics, insufficient auxiliary control sites are available to control the multitude of simultaneous functions required. The conscious effort required would appear to be prohibitive since integration of all components into meaningful actions requires a continuous mental effort and visual feedback (Radonjic and Long, 1970). The concept of a hierarchical control system,(section 2.6.8) allows for the establishment of general programmed routines that can be switched in when required for the implementation of a specific action such as walking, standing, sitting, turning and stopping. The programmed routines can be derived from EMG records obtained from normal subjects performing these actions and, in its final form, be made accessible from a microprocessor. The microprocessor will actively interlink with the subject to facilitate dynamic adjustments.

These two systems described employ EMG control in a manner that retains the natural patterns of innervation of muscles since the control source muscles are performing the same functions they would in the normal subject. Also, paralyzed muscles are being stimulated in a manner that mimics in a gross fashion their natural innervation with respect to when each muscle is activated during a certain action as well as the amplitude variations of the applied innervating stimuli. The number of degrees of freedom provided for is limited only by the number of muscles to be stimulated.

A definite advantage of EMG control over that of position control in this context exists during isometric contractions such as in posture maintainance. A position controller provides no information about tension developed unless a change in position occurs. The control system will then have to be sensitive to small deviations from the reference position in order to prevent instability of posture. Such information is readily available from EMG records.

For EMG control to be feasible, the EMG signals produced by the two antagonists about a joint must contain positional information of the joint. If they do not, it may become necessary to combine EMG and position control into a single system that will be stable and deterministic for all control situations.

Position control of two antagonists about a joint simplifies the complexity of a control system since only one control signal is needed to control both muscles. For upper limb orthoses this becomes important since it decreases the number of control sites needed and hence the mental effort involved in performing a certain task.

One of the aims of this work is to investigate in a quantitative manner the efficacy of EMG and position control in FES of the lower limbs during locomotion.

3.2 Repeatability Criterion

The feasibility of using processed volitional EMG signals as the controller of an FES orthosis depends in part on the ability to ascertain for a given processor:

(i) The form of the volitional EMG when used as a control signal, and the form of the evoked EMG when used as the feedback signal, that will produce the greatest repeatability between volitional and evoked joint-angle time-histories.

(ii) A quantitative means of assessing the efficacy of open-loop and feedback control.

(iii) Whether EMG recorded from the prime mover during a particular motion contains information on the angular time-history of the joint about which the prime mover acts during that motion.

Using position control of joint angle, a quantitative method is needed whereby the inclusion of the feedback or the effects of other control strategies can be assessed.

3.2.1 Variance Ratio Criterion

Hershler and Milner (1976) proposed an optimality criterion whereby the optimal form of processed EMG signals to be used in the modelling of gait could be found. They also used the criterion to investigate how electrode position and orientation on a muscle affect the repeatability of EMG signals recorded during repeated motions.

The criterion proposed by them is given by the following variance ratio:

$$VR = \frac{\sum_{i=1}^{k} \sum_{j=1}^{n} (X_{ij} - \overline{X}_i)^2 / n(k-1)}{\sum_{i=1}^{k} \sum_{j=1}^{n} (X_{ij} - \overline{X})^2 / (nk-1)}$$

where k = number of time points in each record

n = number of signals to be compared

X_{ij} = value of j^{th} signal at time point i

$\overline{X}_i$ = average of values at time point i averaged over j realizations of the signal

$\overline{X}$ = grand mean of the average signal

$$= \frac{1}{k} \sum_{i=1}^{k} \overline{X}_i$$

For a signal to be completely irreproducible over j realizations, $\overline{X}_i \longrightarrow \overline{X}$ since both will represent the mean value of a white noise signal. In this case, $VR \rightarrow 1$. Similarly, for a signal to be completely reproducible over j realizations, the numerator will tend to zero since $X_{ij} \longrightarrow \overline{X}_i$. The variance ratio is thus an indicator of the repeatability of a set of signals and is sensitive to both differences in amplitude and waveshape. An analogy to this variance ratio can be found in statistical texts and is more fully discussed by Hershler and Milner (1978).

3.2.2 Applications of the Variance Ratio

To readily enable repeated experiments, a PDP11/10 computer is used to store the volitional EMG and joint angle time-histories during the execution of specific movements. Either the angle or EMG data or both is then available to be used as the control signal. This method permits comparisons to be made between the efficacy of the control of the same EMG signal when different forms of a processor are applied to it. Experiments for each form of processed EMG can be repeated. The variance ratio is then used to compare the repeatability between the evoked joint-angle time-history due to stimulation, and that of the volitional joint-angle. Similarly, the variance ratio can be used to quantify the efficacy of feedback control and joint-position control.

Two types of EMG processors are used: The volitional EMG signal is digitally rectified and then smoothed by shifting a boxcar window along its length (see Chapter 4); an analogue rectifier-averager is used to process the evoked EMG signal (due to stimulation) when feedback is incorporated into the system. Changing the length of the boxcar window simulates to some degree changing the time-constant of the analogue averager. The analogue rectifier-averager is also used to process the volitional EMG signals during real-time contralateral control of locomotion. The variance ratio has been used to equivalence the two types of filters (see Chapter 4).

3.3 Conclusions

The concept of 'one-to-one' contralateral control has been introduced. A method whereby the efficacy of different control modes can be quantified has been presented.

CHAPTER 4

DESIGN OF A COMPUTER-CONTROLLED STIMULATOR

4.1 Introduction

Stimulus parameters were determined in Chapter 2. The first part of this chapter describes the construction of a prototype 2-channel stimulator incorporating these parameters. The stimulator permits the use of either position control of a joint by stimulating the protagonist-antagonist muscle groups acting about the joint, or EMG control of joint position by stimulating the prime movers acting about the joint. The facility to use the stimulator as a closed-loop controller is provided for.

Experience obtained with this stimulator led to the incorporation of a proportional-integral filter (PIF) into the control circuit and a method whereby evoked EMG could be extracted from a composite signal containing stimulus artifact. In determining EMG-processor time-constants to be used, the need arose to record EMG from both heads of the gastrocnemius muscle.

These additions were incorporated into a four-channel stimulator. The stimulator permits position control of two joint angles or EMG control of the protagonist-antagonist muscle groups acting about the joint. These latter capabilities were not utilized in the work presented in this thesis.

The final sections of this chapter deal with the equivalencing of a digital boxcar-window averager and an analogue RC-averager. This permits replacing the boxcar-window averager with the analogue averager when on-line processing of volitional EMG is required during locomotion (see Chapter 7).

4.2 The Control System

Figure 4.1 is a block diagram of the control system. The calibration control consists of a simple potentiometer and is used to ensure that maximum input signal amplitudes:

(i) are equal

(ii) lie within the linear range of the control circuit (40 mV).

A level shift (Figure 4.2(a)) is provided to account for any D.C. level differences between the two input channels. When position control of joint angle is used, the level shift enables a 'zero' reference angle to be set between the two channels. Control signal variations above or below the zero reference will result in stimulation of the protagonist and antagonist muscle groups respectively (dorsiflexion and plantarflexion for the ankle).

Both channels are fed into a differential amplifier (DA), its output being the difference between the two input signals (Figure 4.2 (b)). Using 1% resistors, the DA has a common mode rejection ratio of 80 dB at a gain of 100. The input impedance is of the order of 1M . A meter* at the output of the DA provides a visual display for setting

*ARMACO U157 PANEL METER, ±μ500 A WITH CENTRE NULL (modified for increased sensitivity) R. MACK AND COMPANY LIMITED

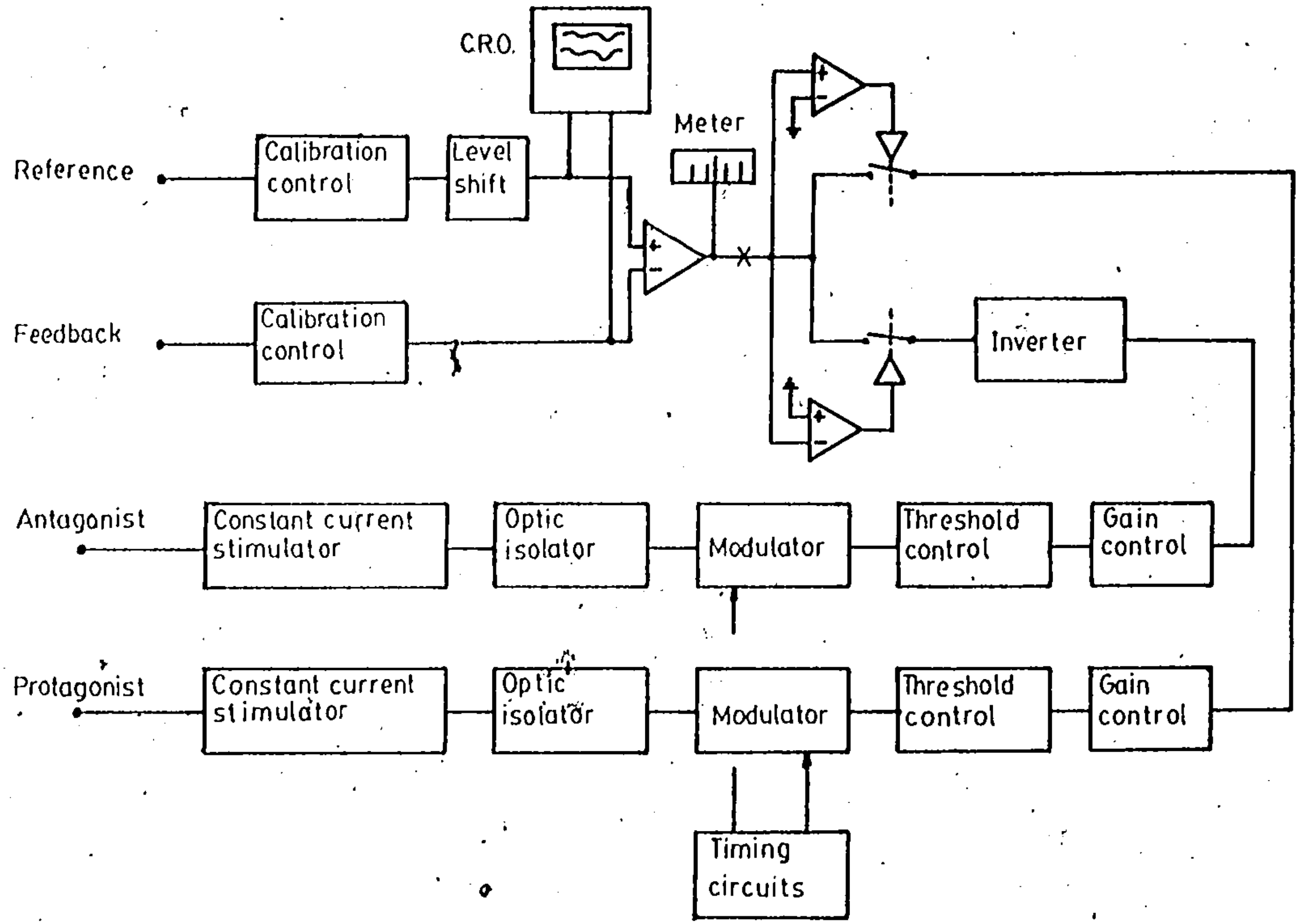


Figure 4.1 Block diagram of feedback control system

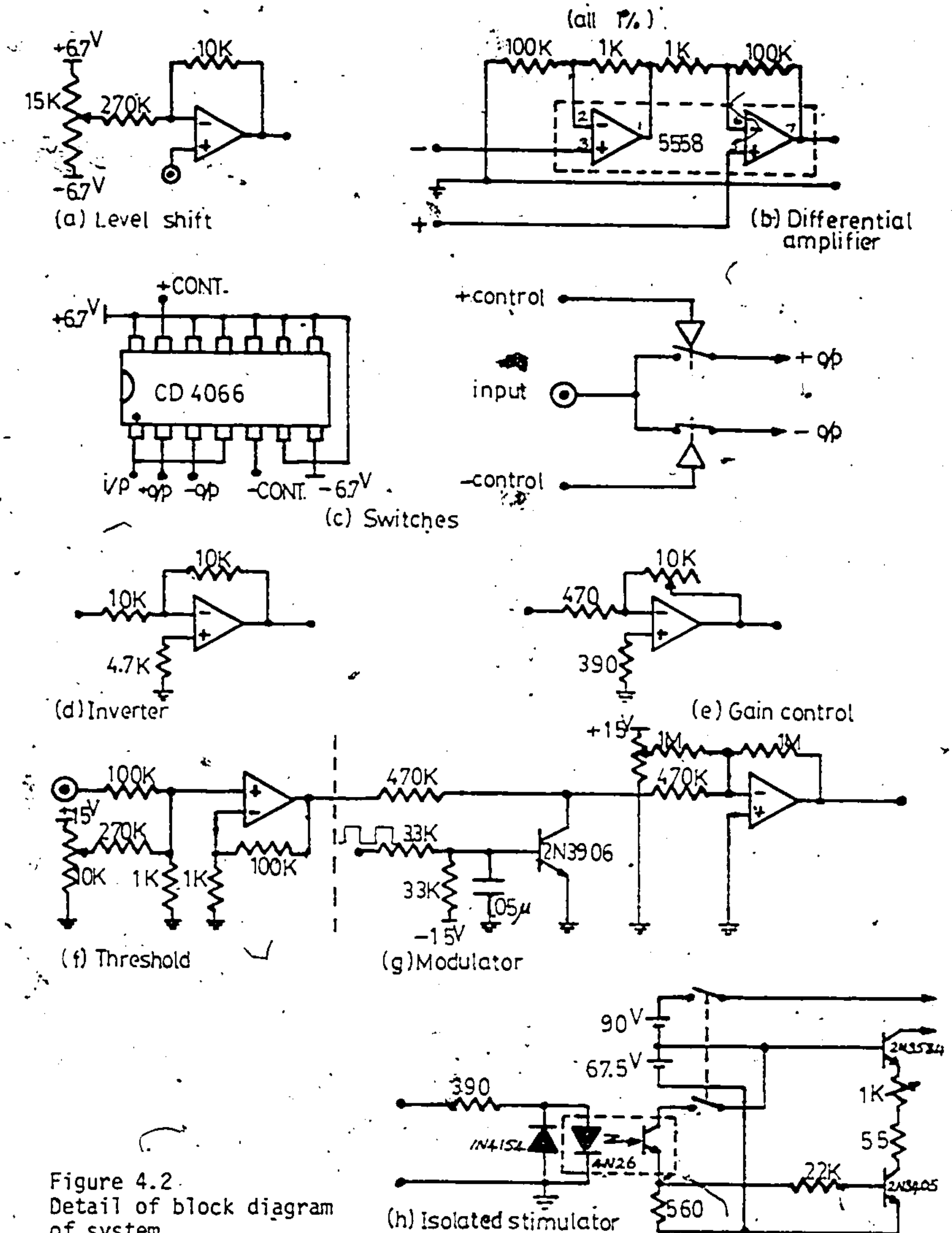


Figure 4.2
Detail of block diagram
of system

the level shift so that at the reference positions (i.e. control and stimulated joint positions set equal, say) the output of the DA is zero. (The proportional-integral filter was later introduced into the circuit at position X and could be switched in if needed.)

Two comparators are used in conjunction with two CMOS switches (figure 4.2(c)) so that an error signal that is positive with respect to the zero reference level results in stimulation of the dorsiflexors and vice versa (as described above). The negative signal is inverted using the circuit shown in figure 4.2(d).

Gain control (figure 4.2(e)) is provided to ensure that the maximum evoked joint angle is equal to the peak of the reference signal in the case of position control, or equal to the peak angle produced by the EMG reference signal in the case of EMG control. Threshold can be set independent of gain setting (figure 4.2(f)), and is used to compensate for voltage drops occurring in the optic isolator and constant current stimulator as well as for applying a threshold stimulus level to the muscle. The modulator (figure 4.2(g)) and timing circuits (figure 4.3(c)) are used to modulate trains of rectangular pulses proportional to the amplitude variations of the error signal from the DA. The modulator has a maximum output of 10^{V} so that the minimum gain of the overall circuit is 250.

An optic isolator is used to isolate the subject from ground. The isolated constant current stimulator is a modified version of that used by Milner et al (1969) (figure 4.2(h)) and includes a variable current limiter.

4.2.1 The Timing Circuit

The timing circuit (figure 4.3(a)) fulfills three functions:

(i) An astable multivibrator generates square pulses. The frequency of the pulses is variable over the range from 15 Hz to 140 Hz.

(ii) A variable switch control is provided for use in the collection of evoked EMG during stimulation. The switch control ensures no recording can occur during the presence of a stimulating pulse so that stimulus artifact due to the stimulus spreading through the body to the recording electrodes is minimized.

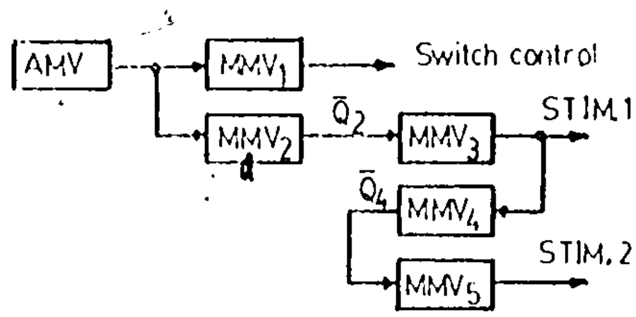
(iii) During the application of a stimulus pulse, current spreads through the tissue surrounding the active electrode and can also spread to the antagonist muscle. In the system used, a threshold was applied to both antagonists about the joint continuously. The author found that unless a just-above-threshold stimulus level was applied to each muscle group, a dead zone existed between the transition from dorsiflexion to plantarflexion and vice versa during which no movement of the joint occurred. However, it became impossible to set the stimulus to threshold on both muscles and to stimulate over the full range of movements without cramping of the muscles occurring due to this current spread. To avoid this problem by decreasing the current strength at any point at a particular time, the stimuli to the two muscles were offset in time with respect to one another (Figure 4.3(b), STIM 1 and STIM 2).

Since the artifact due to stimulation of both muscles had to be blanked out during EMG recording, t_1, t_2 and t_3 as shown in figure 4.3(b) are interrelated. t_1 has to be minimized so that as much of the EMG record as possible can be recovered, while t_2 and t_3 must be set so as to minimize interference between the two channels and still ensure that the stimuli and their artifacts occur during time t_1. Satisfactory values of t_1, t_2 and t_3 to achieve this were found experimentally to be t_1 = 3.5mS; t_2= 60 µS; t_3=1mS. Since the stimulating frequency is 50Hz, 82.5% of the evoked EMG record can be recovered. Figure 4.4 is a record of evoked EMG during maximal stimulation of the right tibialis anterior muscle in a normal subject. The artifact present is due to the switches and not the stimuli.

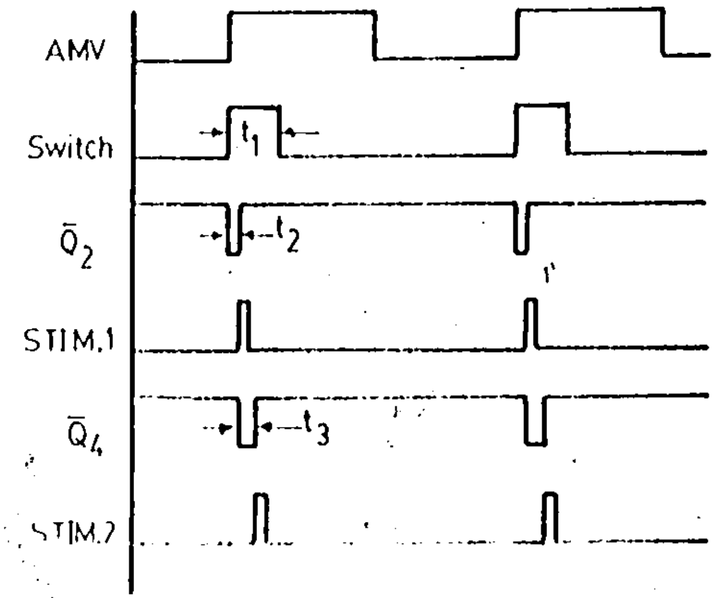
Figure 4.4 was obtained by stimulating the motor point of the muscle, the recording electrodes being distal to the stimulating electrode. Thus there is a relatively short period between the stimulus and the muscle response. The waveshape is due to the summation of the responses of all muscle fibres excited and is biphasic in nature due to first depolarization and then repolarization of the muscle fibres occurring.

4.3 Stimulating Electrodes and Their Placement

Electrodes similar to those described by Milner et al (1970) (Figure 4.5) were used for stimulation. A discussion on these electrodes can be found in section 2.5.3. The inactive electrode used had an area



(a) Block diagram



(b) Timing waveforms

(c) Circuit diagram

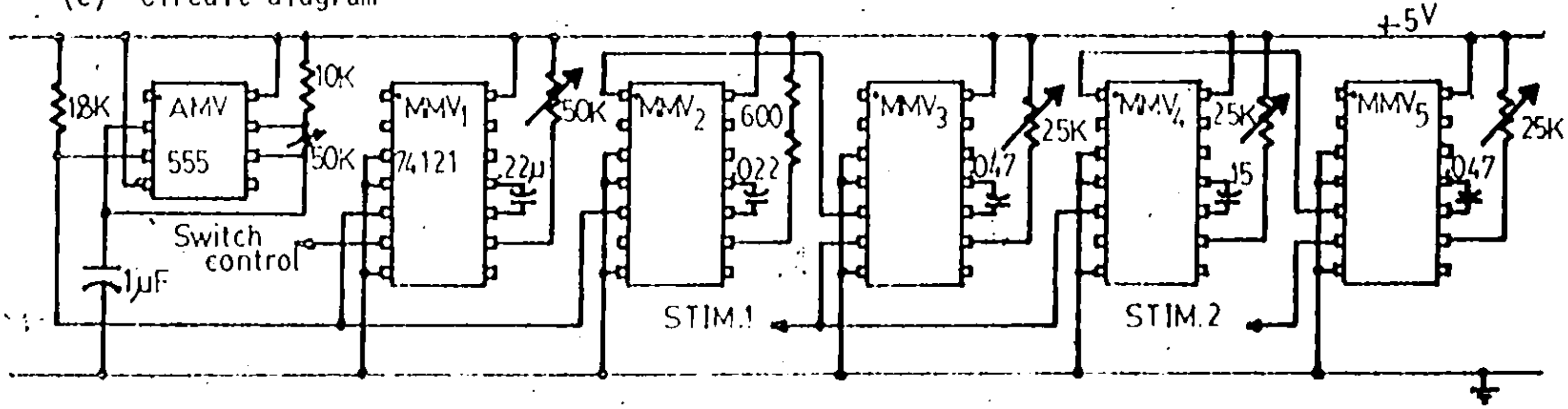


Figure 4.3 Timing circuit

Figure 4.4 Evoked EMG recorded from m. tibialis anterior during stimulation

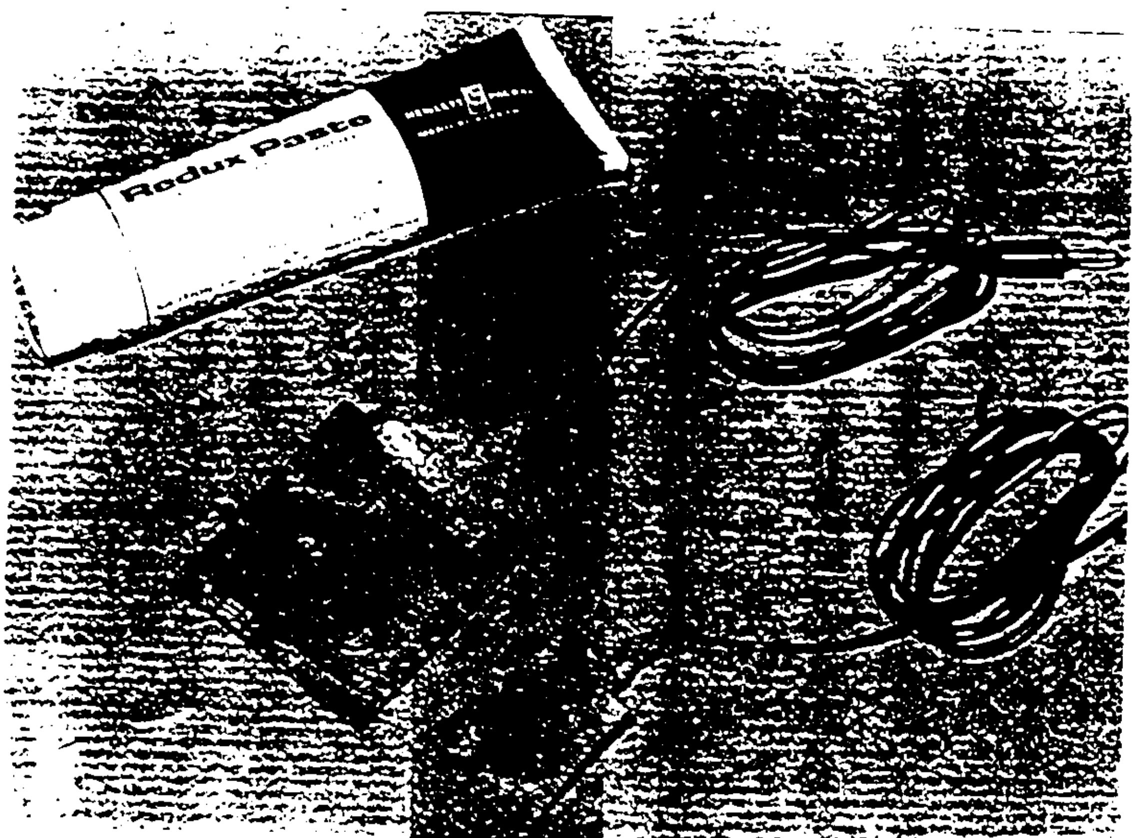


Figure 4.5 Stainless-steel mesh stimulating electrodes

of $40cm^2$. Either one of two active electrodes of area 6.45 cm^2 and 14.5 cm^2 was used. The choice of active electrode size depended on comfort during stimulation which varied for different subjects. In one hemiplegic subject where atrophy had caused a reduction in muscle bulk, the larger active electrode was found to be ineffective.

Using a Digitimer* pulse generator together with an isolated stimulator**, the procedure in placing electrodes was:

(i) Locating an anodal electrode over the extremity of the muscle distal to the motor point.

(ii) Using stimulus parameters of 50Hz, 0.2mS duration, pulse trains on for 1 second and repeated every 3 seconds, a search electrode was used to explore the muscle. The search electrode was made from an insulated brass rod with its bare end covered with gauze which was dampened with tap water prior to use. The motor point was located by finding the area which when stimulated produced the desired motion with the minimum stimulus strength. This point was marked and a permanent electrode placed over it. If two muscles were to be stimulated simultaneously, a common inactive electrode was used. For example, when stimulating the dorsi- and plantarflexors of the ankle joint, the inactive electrode was located at the distal end of the anterior part of the lower leg.

*Digitimer D4030)
**Isolated Stimulator Type 2533)Medical Systems Corp., Great Neck, N.Y.

4.4 Electrogoniometer system

An electrogoniometer was used to measure joint angle either to obtain the control signal or to measure the effects of stimulation. The device measures joint angle variations in the sagittal plane only. It consists of a linear potentiometer attached to special linkages based on a design by Lamoreux (1971). The goniometer is attached via velcro bands, the potentiometer being centred as closely to the joint as possible. For measuring ankle joint angle, an attachment that fits onto and in the shoe is used to secure the distal end. Figure 4.6 is a photograph of the goniometer and figure 4.7(a), the circuit diagram.

4.5 Proportional-plus-integral control

The error signal which is the difference between the control signal and the feedback signal, is used to control the stimulator output amplitudes. As the feedback signal approaches the control signal in magnitude, the error signal and hence stimulus amplitude decrease. The following effects can result:

(i) A joint position is reached beyond which the decreased error signal cannot drive the limb. The difference between the desired and actual positions is known as the static error.

(ii) If a muscle that acts against gravity is being stimulated, oscillations can occur since, as the error signal decreases the limb will drop. This in turn causes an increase in the error signal and the muscle is again stimulated, the process repeating itself.

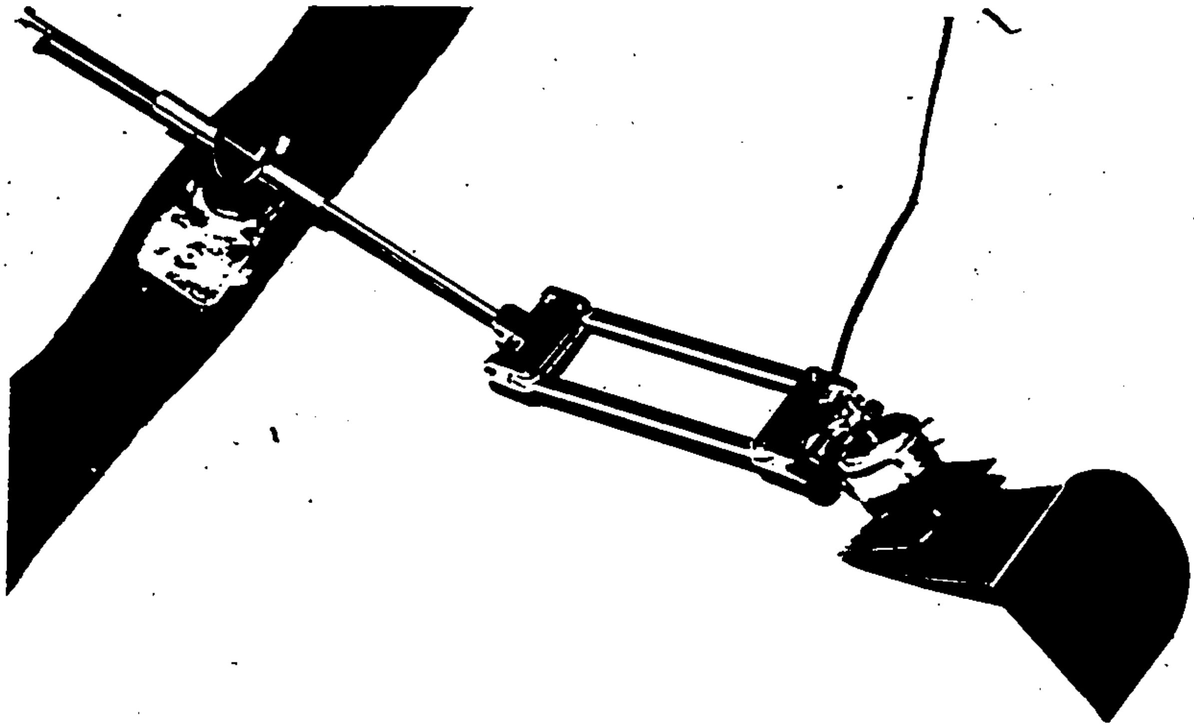


Figure 4.6 Electrogoniometer for measuring ankle-joint movement in the sagittal plane.

Figure 4.8 illustrates the instability of the system due to these effects. A modulated 0.1 Hz sinusoidal was used as the stimulating signal, the joint angle being registered by a goniometer attached to the ankle joint of a normal subject. Dorsiflexion and plantarflexion were effected by stimulating the tibialis anterior and gastrocnemius muscles respectively. Distortions in the output waveform did not always result in oscillations as shown in figure 4.9 obtained during a repeat of the experiment at a later date. Maximum obtainable dorsiflexion and plantarflexion during stimulation were 33^{o} and 13.75^{o} respectively from the resting position. The ankle angle in the resting position was 105^{o} with respect to the vertical. The input signal level was therefore adjusted so that the positive two-thirds of the signal resulted in dorsiflexion of the ankle.

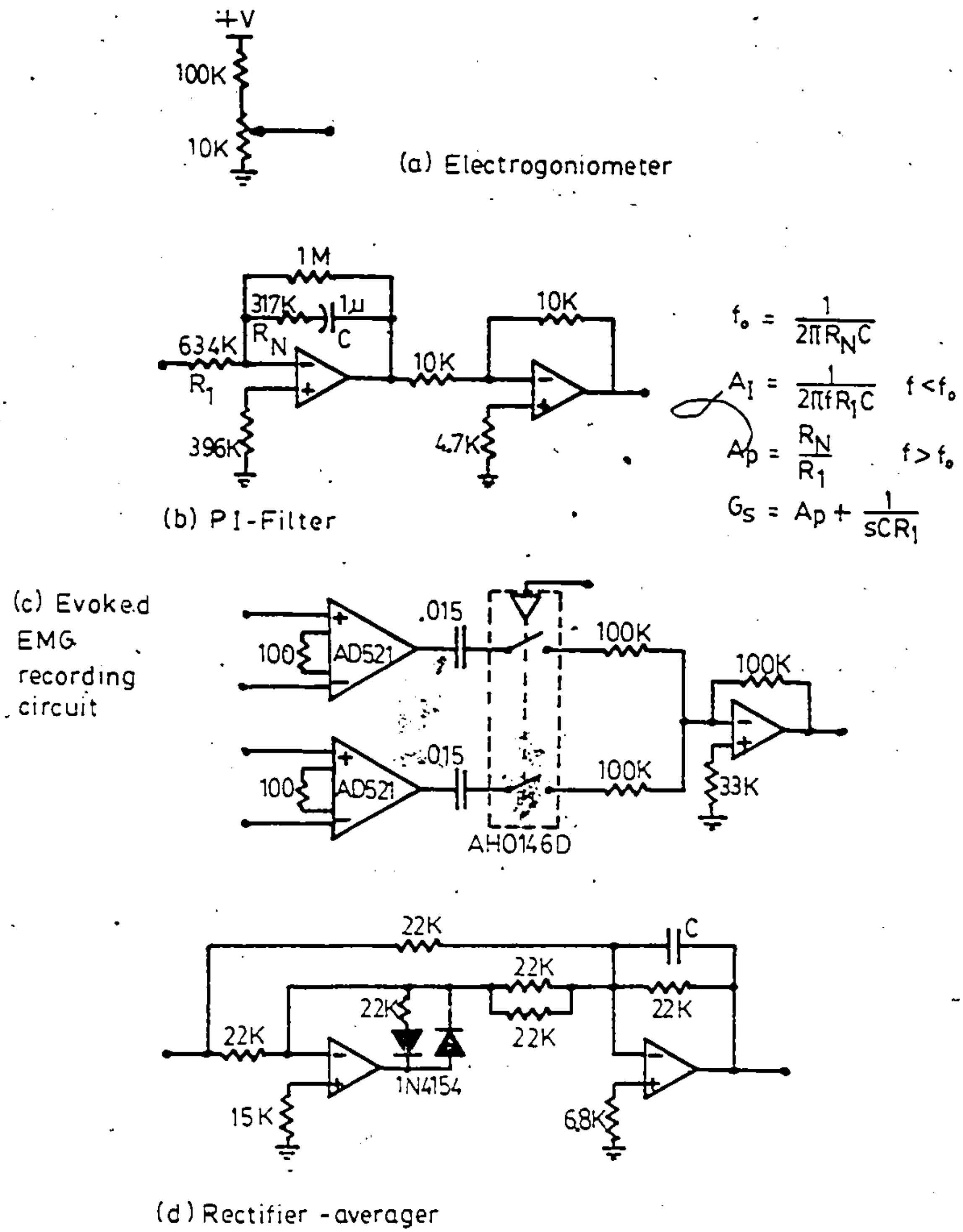


Figure 4.7 Details of block diagram of system

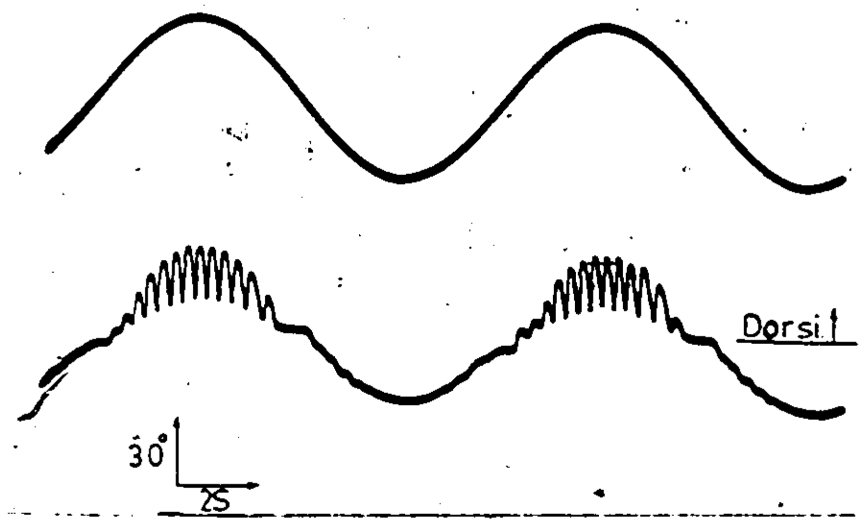


Figure 4.8 Stimulation without PI-filter
Top trace: Reference signal
Bottom trace: Evoked joint angles

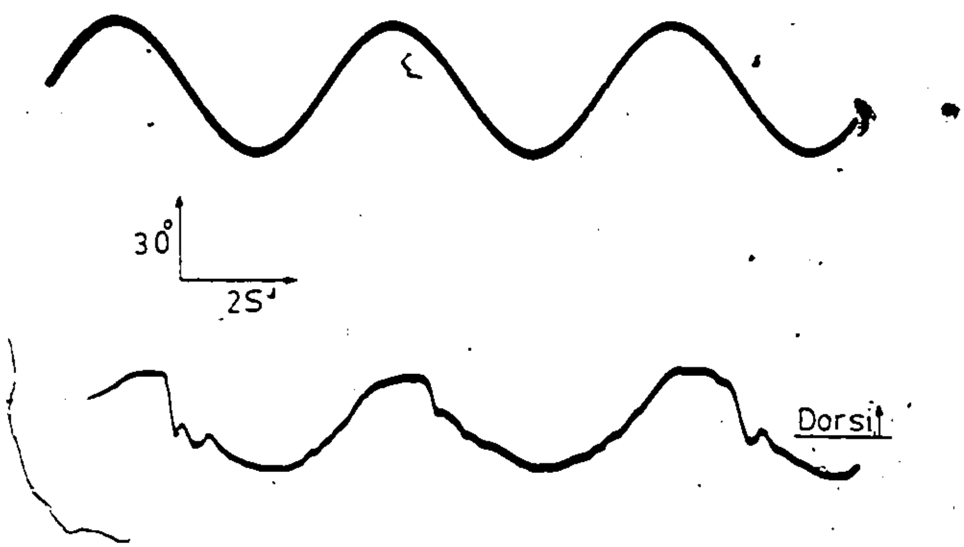


Figure 4.9 Stimulation without PI-filter
Top trace: Reference signal
Bottom trace: Evoked joint angles

It was decided to incorporate a proportional-integral (PI) filter in the system to overcome these problems.

Consider the frequency response of a PI filter shown in figure 4.10. At low frequencies, the integral section of the PI filter will produce an output signal proportional to the summation of all errors. When the error between the control and output signals becomes zero, the PI filter output will decay at a rate determined by the choice of the components of the circuit (R,C in figure 4.7(b)). Thus the integrator will decrease the static error and hence oscillations due to the effect of gravity. A circuit diagram of the active PI filter used (Tietze and Scheck, 1969) is shown in figure 4.7(b). The break frequency f_o must be chosen such that:

(i) The gain of the filter is sufficient at input frequencies greater than f_o and not too large at input frequencies below f_o thus enabling stimulation to be effective over the complete frequency range of interest.

(ii) At higher frequencies of movement, the time response of the system should not be decreased to the point where the system becomes too sluggish to follow the movement.

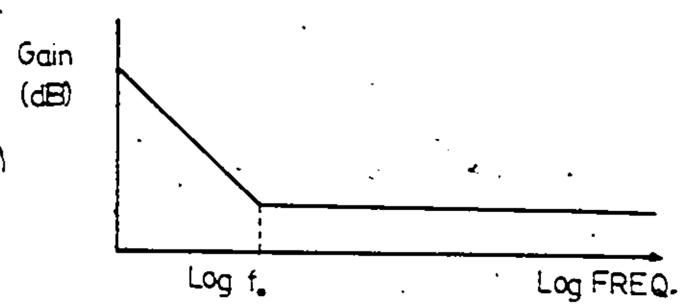


Figure 4.10 Frequency response of PI-filter

The use of a differentiator will increase the response of the system at higher frequencies. However, it was decided not to incorporate a differentiator into the system for the following reasons:

(i) Differential control accentuates the high frequency content of any noise present in the actuating signal. Stanic and Trnkoczy (1974) found that this influence was of the same order of magnitude as that of the stochastic nature of the FES responses.

(ii) The differentiator transforms the error signal into a high initial stimulation voltage which can result in pain from stimulating at this level. Limiting the output decreases the effectiveness of the differential action (Stanic and Trnkoczy, 1974).

Rather than analytically treating the system, it was decided to determine the PI filter parameters experimentally for the following reasons:

(i) An attempt was made to find the transfer function between a sinusoidal input to the system and the resultant joint angle during stimulation of the tibialis anterior muscle in four normal subjects. The repeatability of the measurements between subjects and for each subject was very poor. Problems encountered were the subjects anticipating the stimulus and the onset of fatigue prior to a steady state being reached. Attempts to precondition the subjects to the sensation of stimulation did not improve the repeatability of the results and was thought to be due to subconscious inputs from the subjects into the system.

(ii) Attempts have been made to formulate the relation between stimulating voltage and movement of an extremity due to agonist-antagonist action about the joint. Stimulated muscle contains a dead zone, saturation, delays, static hysteresis, nonlinear dependence on muscle length and joint angle, and is subject to the effects of fatigue and accommodation (Stanic and Trnkoczy, 1974). Simplifications made to obtain a linearized model of the system have resulted in models which are valid only under certain conditions such as isotonic movements or isometric moments (Trnkoczy and Stanic, 1971).

Various PI filters were constructed and tested. In all cases, the proportional gain, A_p, (figure 4.7(b)) was arbitrarily chosen as 0.5. The filter finally selected had a break frequency of 0.5 Hz and a transfer function:

$$G(S) = 0.5 + \frac{1.57}{S}$$

Stanic and Trnkoczy (1974) obtained best results with the following PI filter transfer function:

$$G(S) = 3.22\ (0.5 + 1.69/S)$$

$$f_0 = 0.54 \text{ Hz}$$

As figure 4.11 illustrates, inclusion of the PI filter has resulted in the effective damping of the large oscillations previously present in the evoked joint angle response. Inclusion of the PI filter resulted in an increase in the time lag between input and output. Onset of delay was measured as 90 mS compared to a delay of 60mS when the filter was omitted. The ability of the joint angle to follow a randomly varying input is shown in figure 4.12.

Figure 4.13 shows the goniometer output in response to a randomly varying input signal. High frequency oscillations can be seen to occur. When the experiment was repeated two hours later, no oscillations occurred. Stanic and Trnkoczy (1971) also experienced spurious oscillations which they could not explain. They repeated the experiment and stimulated for 3.5 minutes to see whether fatigue was the cause but the oscillations did not repeat. In an experiment to determine the source of these oscillations, the author found that they occurred only after 12 minutes of continuous stimulation and that fatigue was therefore the cause.

4.6 Recording of EMG

Silver-silver chloride electrodes (figure 4.14) are used for recording evoked EMG during stimulation and EMG due to volitional movements. The skin surface is prepared by vigorously rubbing it with an alcohol swab. In attempting to minimize artifact during stimulation, the electrodes are placed on an approximate equipotential line between the active and inactive electrodes. The electrodes are gelled in order to decrease the impedance at the electrode-skin interface.

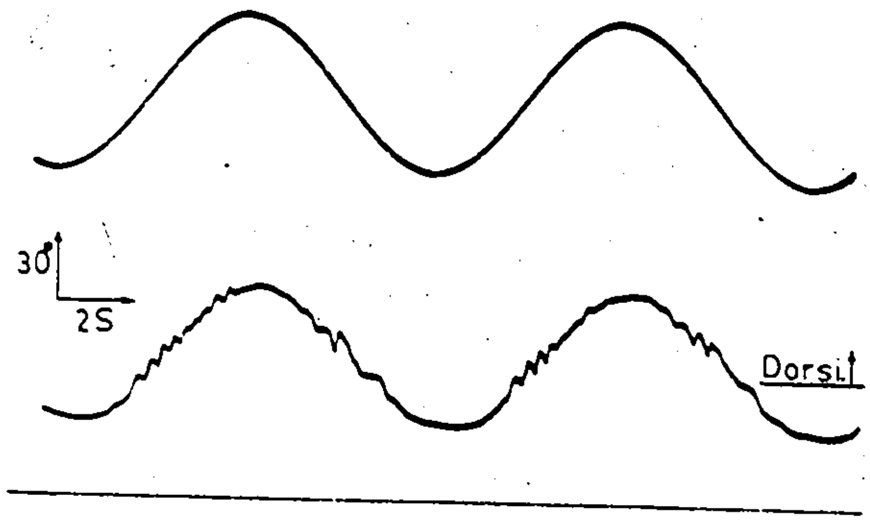


Figure 4.11 Stimulation with PI filter (cf. figure 4.8)
Top trace: Reference signal
Bottom trace: Evoked joint angle

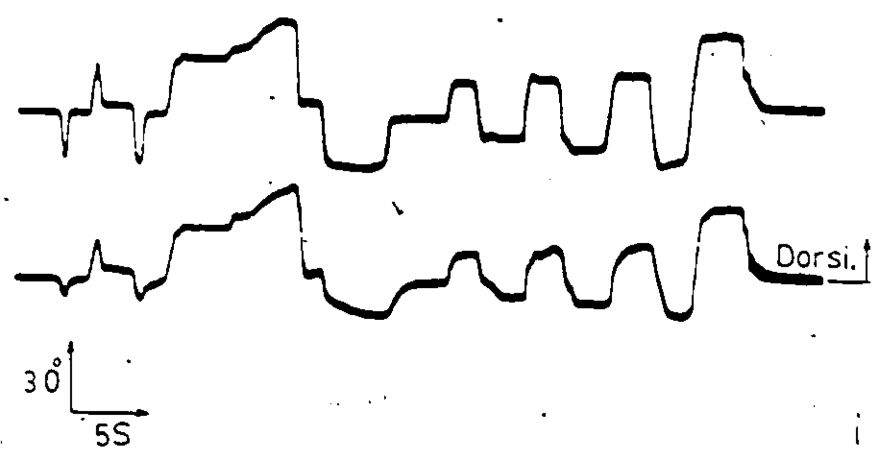


Figure 4.12 Stimulation with PI filter: random input
Top trace: Reference signal
Bottom trace: Evoked joint angle

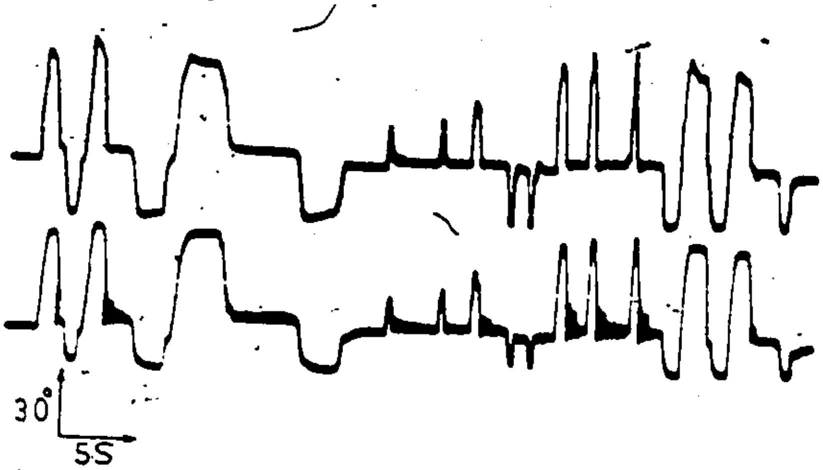


Figure 4.13 Stimulation with PI filter showing occurrence of oscillations
Top trace: Reference signal
Bottom trace: Evoked joint angle

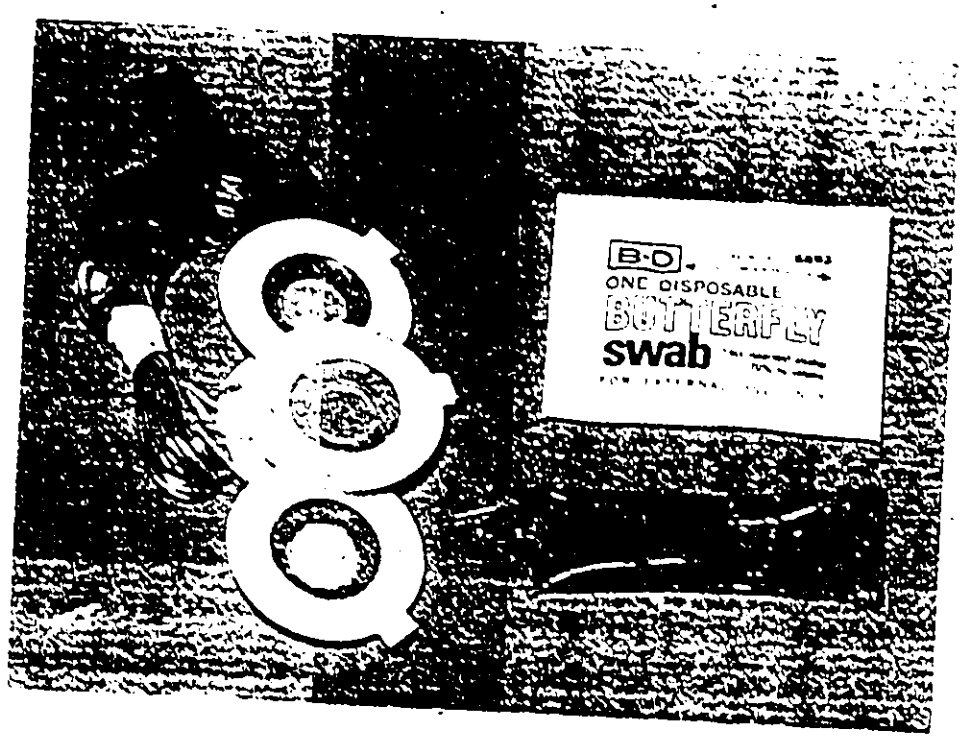


Figure 4.14 Silver-silver chloride recording electrodes

Any common-mode signals are rejected by using a high quality differential amplifier. An instrumentation amplifier with a typical CMRR of 110 dB is used. The high input impedance ($3\times10^9\,\Omega$) of the amplifier relative to the source impedance (3-$30K\,\Omega$) ensures that signal distortion does not occur. A high pass filter (f_o=10Hz) at the output of the amplifier eliminates any DC offset voltages due to electrode-skin-electrode imbalances. The circuit used is shown in figure 4.7(c). High quality analogue switches (8 Ω ON resistance) are used to disconnect the amplifiers from the rest of the recording system during the presence of a stimulus pulse. The switches are gated by a switch-control pulse (see t_1, figure 4.3(b)) obtained from the timing circuit.

A rectifier-averager is used to obtain an envelope of the EMG activity (see section 4.7) which in turn is used to modulate the stimulator output. This is shown in figure 4.7(d) and consists of an active full-wave rectifier and a smoothing circuit whose RC time constant (22KxC) determines the period over which the signal is averaged.

Electrode leads are kept short and all cables used are shielded to prevent 60Hz pickup.

Figure 4.15 shows the evoked muscle response to a train of stimuli recorded from the tibialis anterior muscle. The muscle response has been rectified and averaged with a time constant of 2.2mS.

Figure 4.16 is a simultaneous recording of evoked ankle-angle, and evoked EMG processed with an RC time constant of 72.6mS. Figure 4.17 is a simultaneous recording of volitional ankle-angle and volitional EMG processed with the same time constant of averaging. The

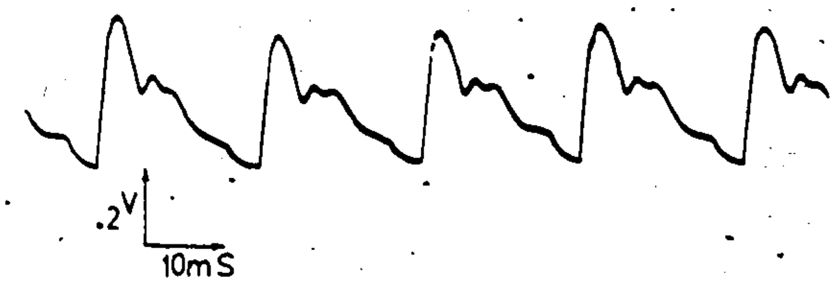


Figure 4.15 Evoked response from m. tibialis anterior during stimulation (rectified and averaged with 2.2 mS time-constant)

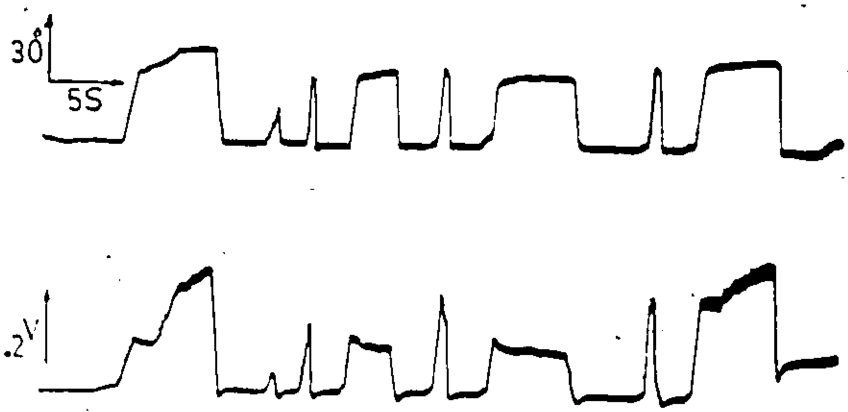


Figure 4.16 Evoked angle (top trace) and EMG from m. tibialis anterior (bottom trace) records. (EMG rectified and averaged with 72.6 mS time-constant)

similarity between the joint angle and EMG waveforms led the author to investigate the feasibility of implementing an EMG-controlled closed-loop system.

4.7 4-Channel Stimulator

A second stimulator was constructed with the following additions:

(i) For EMG control: (a) Two-evoked EMG signals can be recorded from each stimulated muscle group. These signals are summed following rectification and averaging. Since EMG signals are stochastic, rectification prior to addition ensures that no cancellation between portions of the two signals will occur.

(b) Optic couplers isolate the battery-powered recording circuits from the rest of the electronics which is powered from the main-line supply

(c) The facility to employ EMG control of both the protagonist and antagonist muscle groups acting about a joint is provided for.

(ii) For position control, protagonist-antagonist muscle groups acting about two joints can be stimulated independently.

(iii) In general: (a) PI-filters are included in the circuit and can be switched in or out as required.

(b) Provision is made whereby the threshold stimulus can be switched in or out of the circuit. When switched in, the threshold stimulus will be present at all times. This enables the threshold levels to

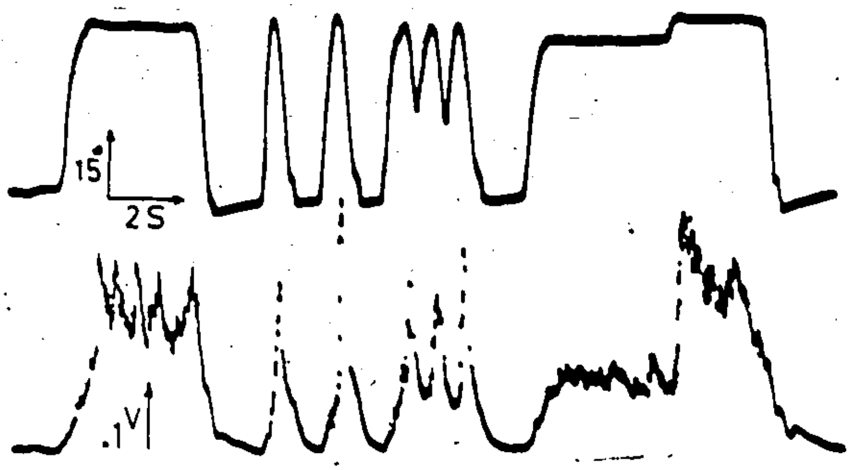


Figure 4.17 Voluntary angle, (top-trace), and EMG records, (bottom-trace). (EMG rectified and averaged with 72.6mS time-constant). Tibialis anterior muscle.

be set during calibration. When switched out, the threshold stimulus will appear at the output of a particular channel only when that channel is activated by the controller. Thus fatigue of the muscle due to constant simulation is minimized. (see section 4.2.1).

Figure 4.18 shows the control panel of the 4-channel stimulator. Appropriate connectors are provided for both on-line and off-line control of stimulation. The stimulator was constructed in a modular form (figure 4.19) for easy access to effect repairs, and consists of three plug-in boards. Figure 4.20 is a block diagram of the controller and figures 4.21, 4.22, and 4.23 are circuit diagrams for each of the modular boards.

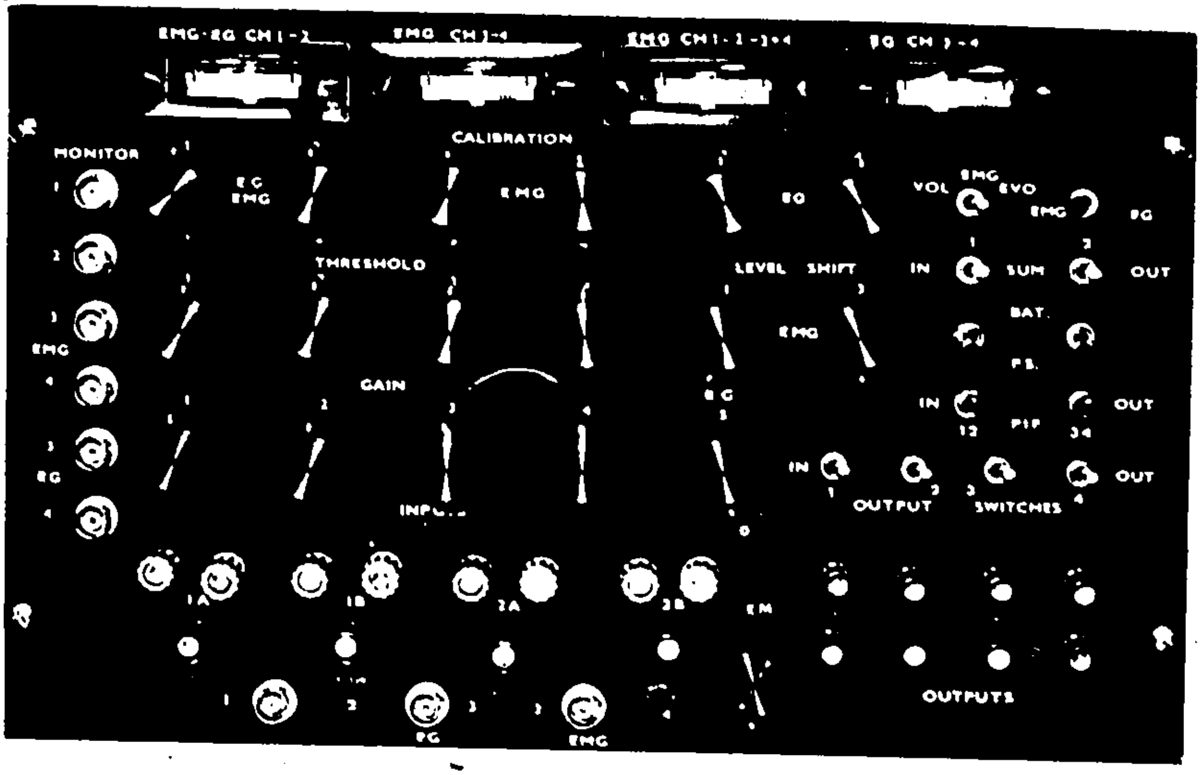


Figure 4.18 Control panel of 4-channel stimulator

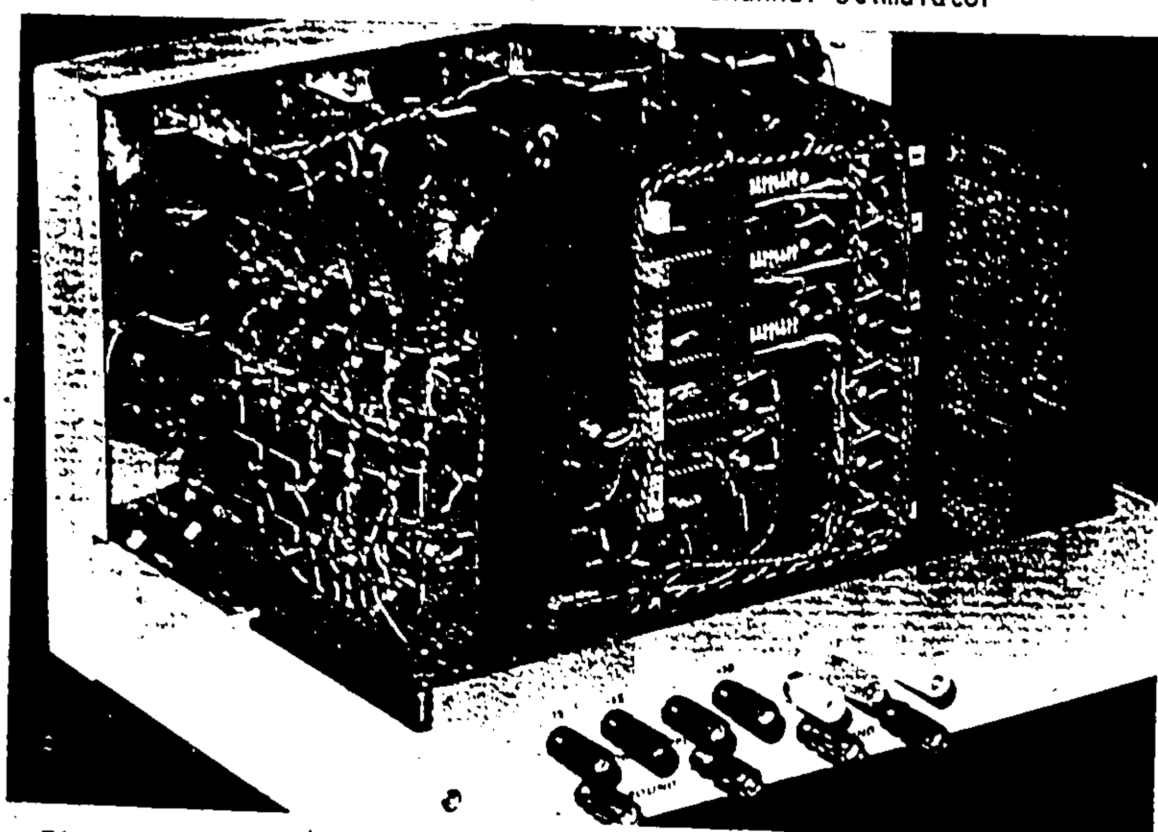


Figure 4.19 Rear view of stimulator illustrating plug-in boards

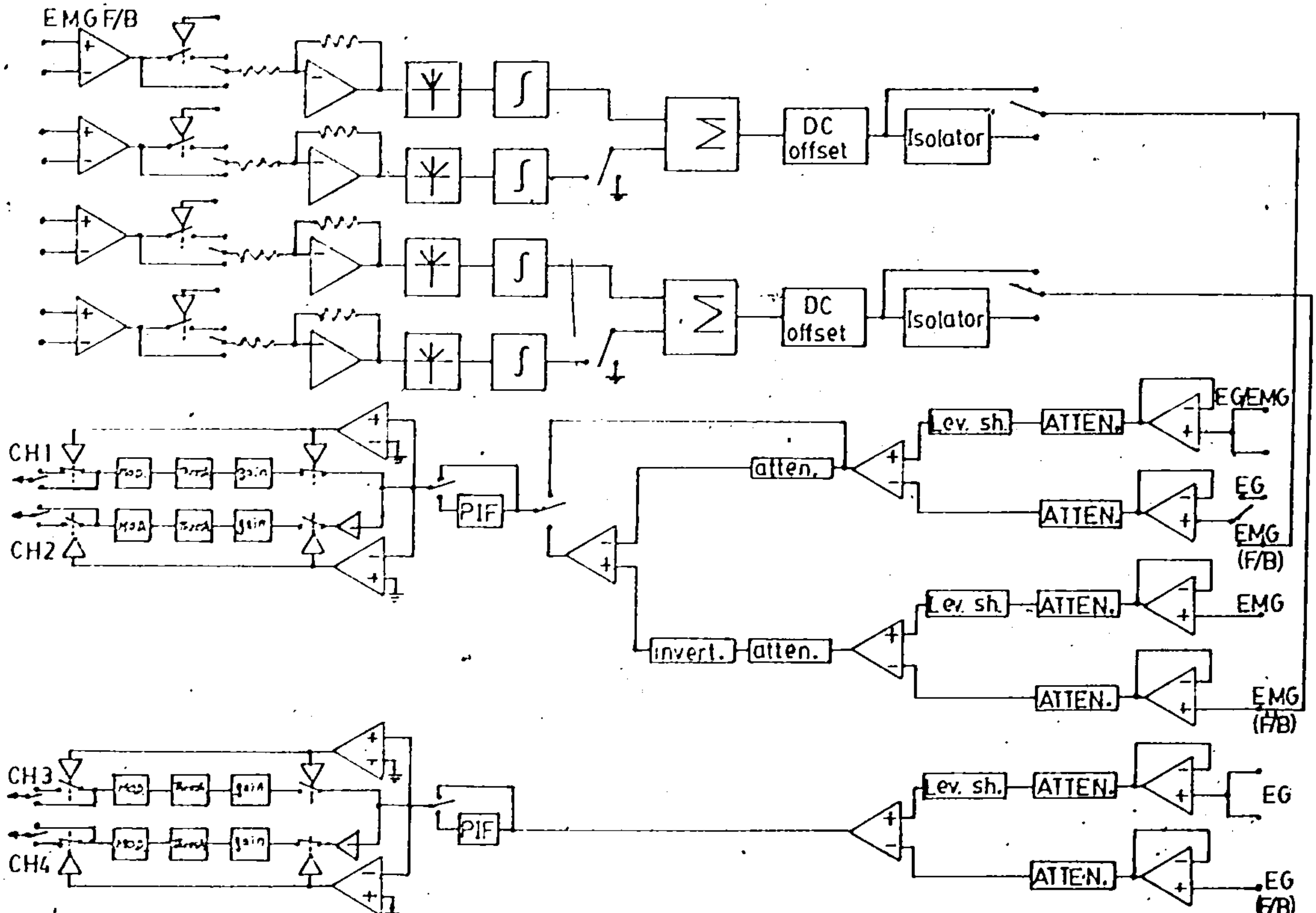


Figure 4.20 Block diagram of 4-channel stimulator

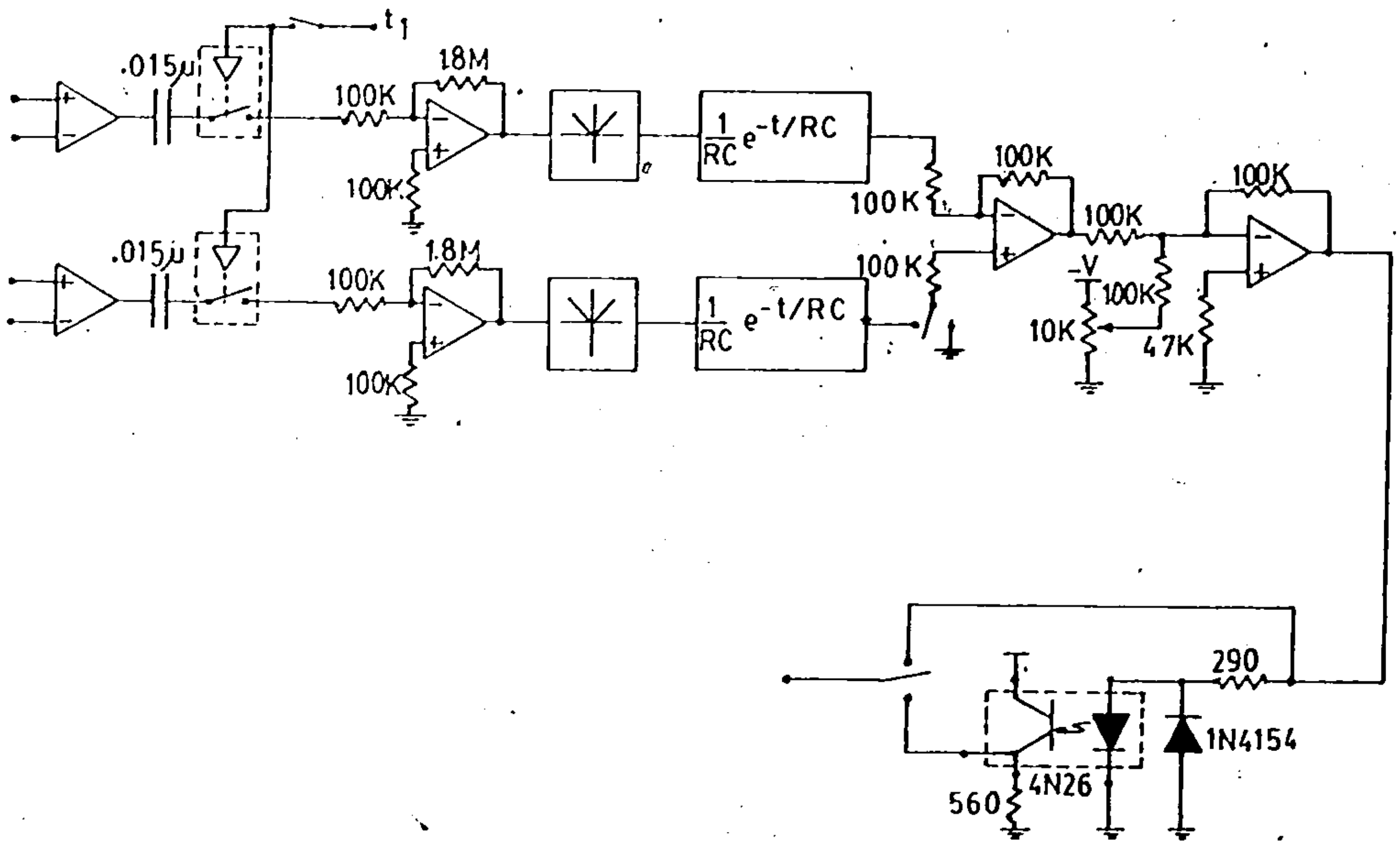


Figure 4.21 Evoked EMG recording and processing unit (one channel only)

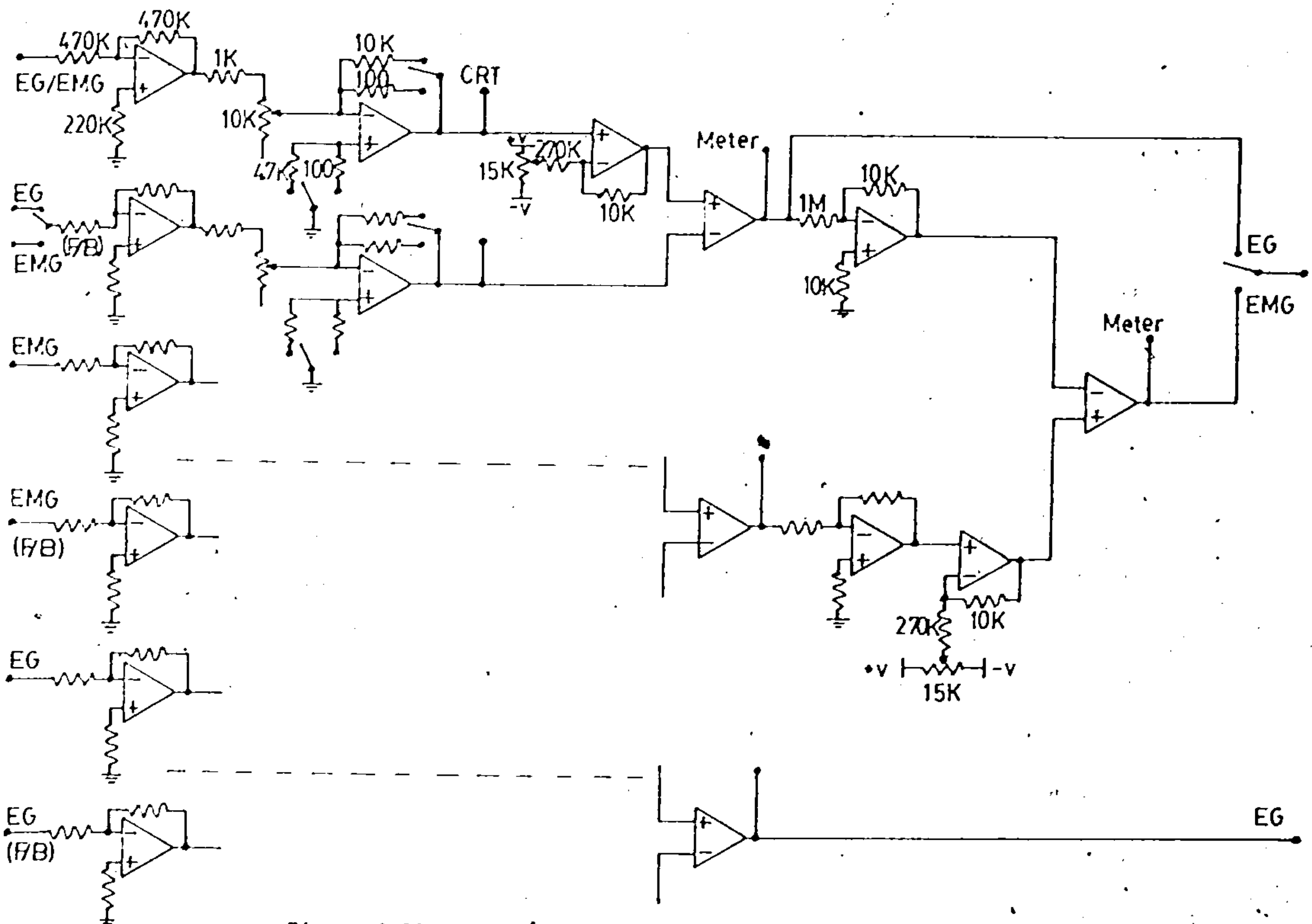


Figure 4.22 Calibration, level shift and differencing circuits

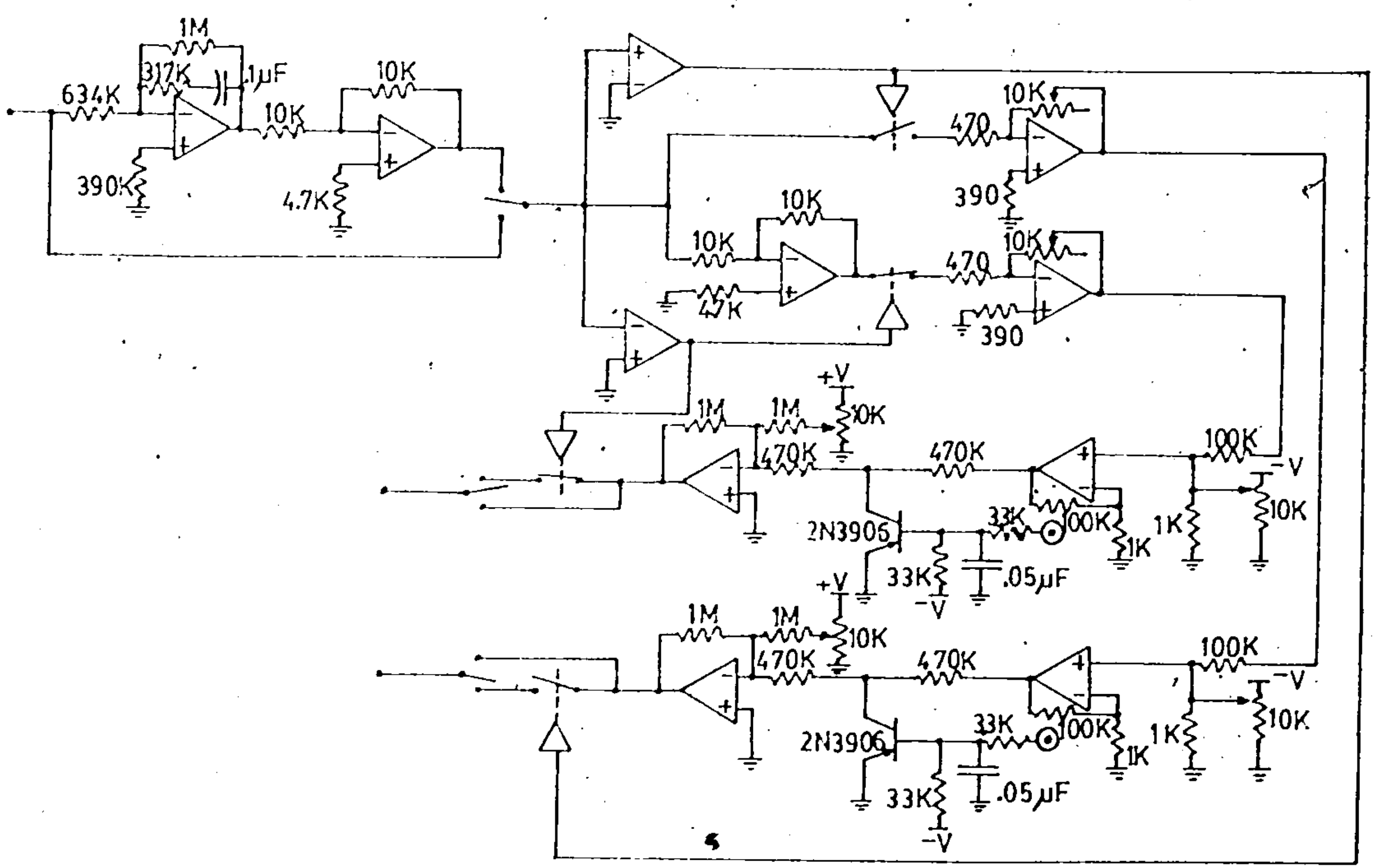


Figure 4.23 PI-filter, control, gain and threshold circuits (one channel only)

4.8 Filter Comparison

One of the objects of this work is to determine the form of the control EMG signal which will produce the greatest repeatability between volitional and evoked joint angles. To readily enable repeated experiments, a means must be found whereby the same control EMG record can be processed in different ways and a comparison be made between the efficacy of each processed EMG waveform when used to control stimulus amplitude.

Various devices are available to store the volitional EMG and joint angle data recorded simultaneously during a particular movement. If a computer is used, the sampled EMG data are then available for repeated processing and outputting to the stimulators. The evoked joint angle time-history (due to stimulation) can then be stored in the computer during stimulation for later comparison with the previously stored volitional joint angle data. Once the desired form of the processor has been determined, an equivalent analogue processor can then be substituted to permit on-line 'contralateral' control. This section presents a method whereby the parameters of the analogue processor are determined so that the two processors are as similar as possible.

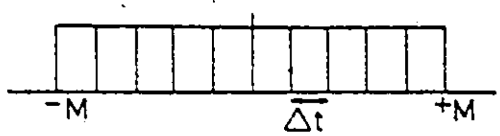
4.8.1 Filter Forms

The analogue rectifier-averager shown in figure 4.7(c) was chosen because of its easy implementation. The digital processor employed is also simple to implement. Rectification is achieved by taking the absolute value of the sampled EMG signal stored in the PDP11/10 computer. Averaging is effected by shifting a boxcar window along the length of the signal. A weight of one is given to each data point in the window. The point corresponding to the middle of the window is replaced by the average value of all points in the window and stored in a different buffer to the one containing the original data. Changing the length of the window is analagous to changing the time constant of the analogue averager.

4.8.2 Method of Equivalencing

An analytic method of equivalencing the two filters was first attempted.

Consider a moving average boxcar window:



The window averages over (2M+1) points. The mean value of the i^{th} point is given by:

$$Y_i = \frac{1}{2M+1} \sum_{k=-M}^{M} X_{i+k}$$

which yields the following transfer function in the frequency domain (Enochson and Otnes, 1968):

$$H_D(f) = \frac{\sin[(2M+1)\pi\Delta tf]}{[(2m+1)\pi\Delta tf]}$$

or

$$H_D(w) = \frac{\sin w\, T_D/2}{w\, T_D/2} \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots (1)$$

where: T_D = window length

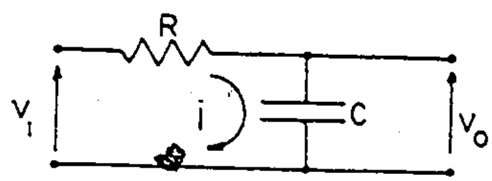
= $(2M+1)\,\Delta t$

Δt = sampling increment

f = frequency of sampling

= $w/2\pi$

Consider the equivalent circuit of the analogue averager:



$$\frac{V_o}{V_i} = \frac{1}{jwRC + 1}$$ so that

$$|H_A(w)| = \frac{1}{(1+w^2C^2R^2)^{\frac{1}{2}}}$$

$$= \frac{1}{(1+w^2T_A{}^2)^{\frac{1}{2}}} \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots (2)$$

To minimize the differences between the two transfer functions, the square of the difference between equations (1) and (2) is minimized as follows:

$$\text{Let } k = T_D/2T_A$$

$$H_D(w) = \frac{\sin(wkT_A)}{wkT_A}$$

The expression to be minimized becomes:

$$E(w) = [|H_A(w)|^2 - H_D(w)^2]^2$$

$$= \left[\frac{1}{1+w^2T_A^2} - \frac{\sin^2(wkT_A)}{(wkT_A)^2}\right]^2$$

By specifying T_A (or T_D) and varying k, $E(w)_{MIN}$, the minimum squared error can be found by summing the coefficients of E(w) over the frequency range of interest. Figure 4.24 shows the squared error for a fixed window of $T_D = 100$ plotted against frequency for a range of RC time constants. Immaterial of the value at which T_A (or T_D) is fixed, the minimum squared error yields a value of $T_D/T_A = 2.33$ for all cases.

Superimposed on figure 4.24 is the frequency spectrum of a rectified EMG signal recorded from the tibialis anterior muscle. The phasic movement during which the EMG was recorded is reflected by the large low-frequency component of the spectrum and is present due to the prior rectification of the EMG signal. This is analogous to the retrieval of a low frequency modulating signal from a high frequency carrier. From figure 4.24 it is apparent that minimizing E(w) is not a sufficient criterion for equivalencing the two filters since both the error difference and the EMG signal are frequency dependent. Thus a method is needed whereby the influence of each filter on a particular EMG signal can be measured. The variance ratio of section 3.2 lends itself to this application. A figure of merit can be given to the repeatability between the two output waveforms resulting from the operation of the two filters on the EMG signal for different filter time constants. The method adopted was:

(i) A specific T_A was chosen

(ii) The two output waveforms are obtained by convolving the input EMG waveform with the transfer function of each filter. In the frequency domain this is achieved by multiplying the frequency spectrum of the EMG signal with the frequency transfer functions of the filters. This process was repeated for a range of different k's (i.e. T_D's).

(iii) For each value of k, the convolved signals were then compared to each other using the variance ratio. The value of k that produced the minimum variance ratio for a particular T_A thus indicates the optimum equivalencing of the two filters for the particular EMG signal being considered.

Figure 4.25 shows the outputs of the filters for such a case. Optimal filter time-constants for various input signals are shown in Table 4.1. Using this method it is now also possible to compare results obtained using either filter.

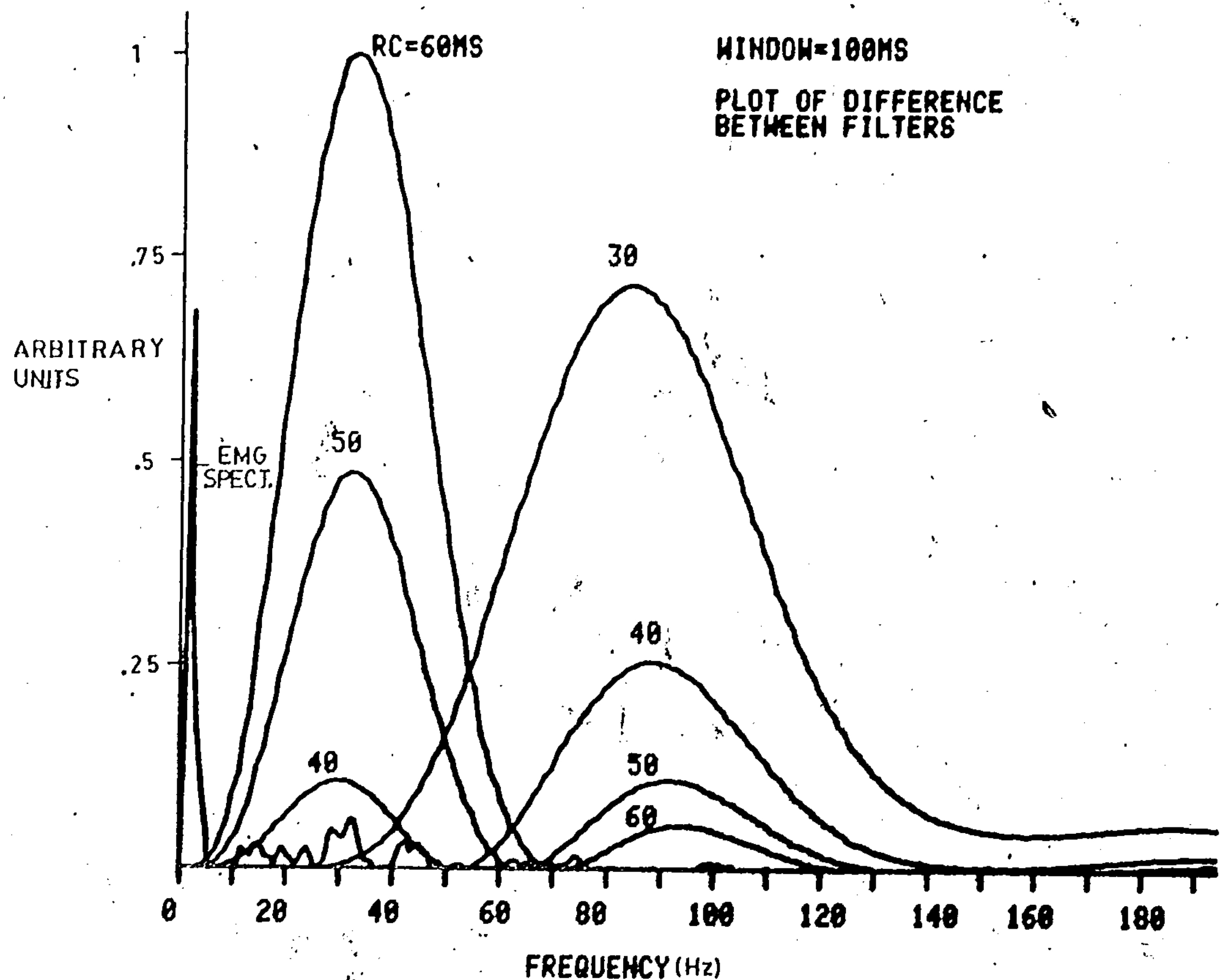


Figure 4.24 Plot of error function and frequency spectrum of rectified EMG signal as a function of frequency

EMG	T_D(MS)	T_A(MS)	T_D/T_A
VOLITIONAL TIB. ANT.	100	37	2.70
	200	65	3.08
	300	106	2.83
VOLITIONAL GASTROC.	200	67	2.99
	300	110	2.73
EVOKED TIB. ANT.	100	46	2.17
	200	106	1.89
	300	170	1.76

Table 4.1: Optimal filter time-constants for various EMG input signals

4.9 Conclusions

The design of a two-channel stimulator for effecting joint-position control has been presented. Also given were the rationale and experimental corroboration for the need to include the following features into a 4-channel stimulator:

(i) An offset between stimulus pulses applied to the protagonist-antagonist muscle groups acting about a joint.

(ii) Inclusion of a PI filter in the forward path of the control loop.

(iii) An analogue switch to eliminate stimulus artifact from recorded evoked EMG.

(iv) Circuitry whereby multiple sources of recorded EMG can be combined.

A method has also been presented whereby a comparison can be made between the effects of a digital and analogue filter operating on the same EMG signals.

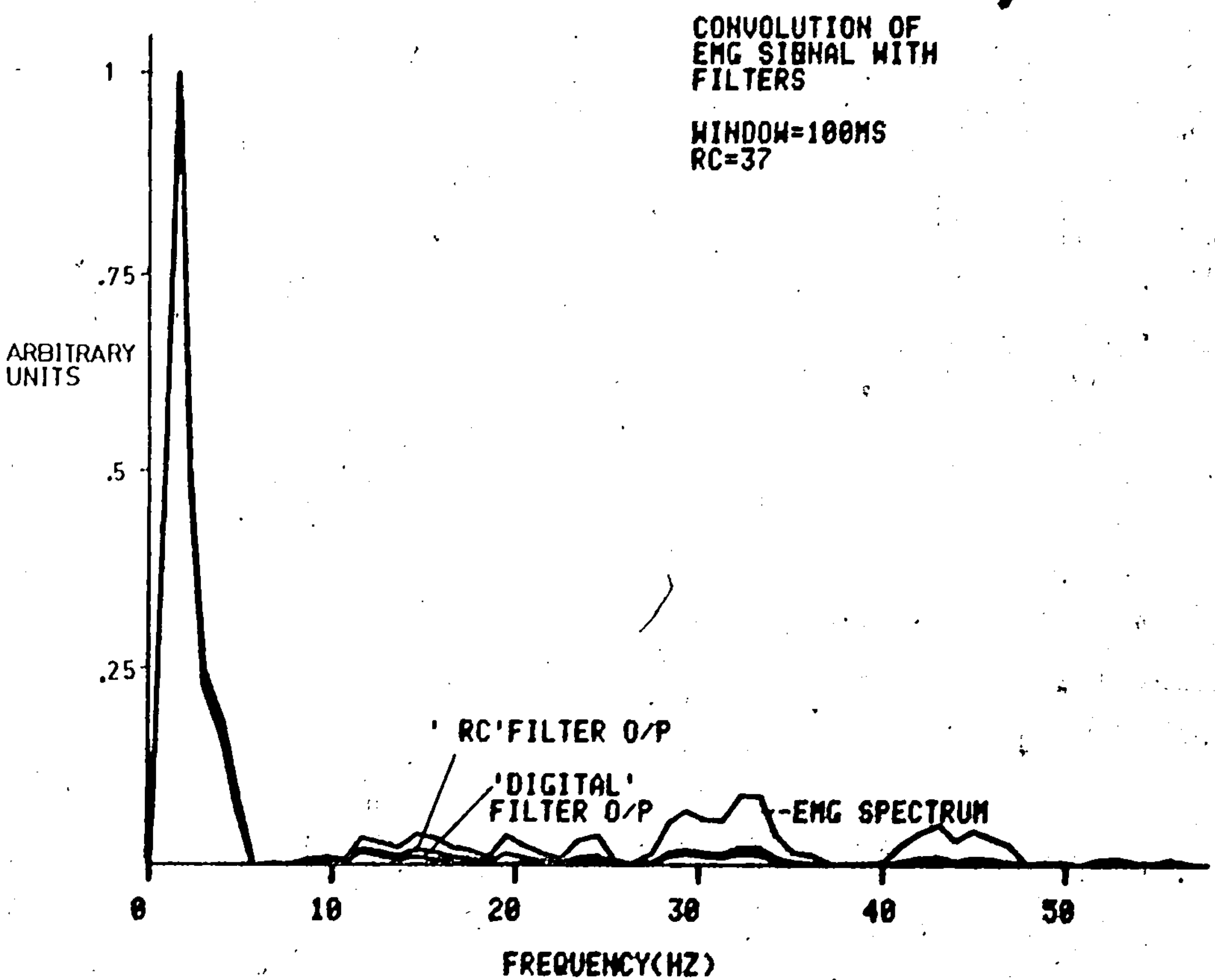


Figure 4.25 Frequency spectrum of rectified EMG signal and its convolution with the two filter transfer functions

CHAPTER 5

EMG CONTROL OF JOINT POSITION

5.1 Introduction

The feasibility of using EMG signals in the control of an FES orthosis depends in part upon the ability to ascertain for a given processor:

(i) The form of the EMG control signal that will produce the greatest repeatability between the evoked joint-angle and the control joint-angle time-histories.

(ii) Whether a high correlation exists between the processed forms of the volitional control EMG signals and the resulting evoked EMG signals. The establishment of such a relationship will suggest that the implementation of an adaptive EMG control system employing feedback from the stimulated muscle is plausible.

(iii) Whether (since it is joint position that is ultimately being controlled) the angular time-history of the joint is contained in the processed EMG signal recorded from the prime mover during a particular movement, whether it is artificially stimulated or not.

(iv) Whether, if (ii) and (iii) above are established, the inclusion of feedback quantitatively improves the repeatability between the evoked and volitional joint angles.

The first part of this chapter describes the experimental approach and procedures adopted. The following sections deal with open-loop EMG control of joint position and its ramifications. The last part of the chapter presents the results obtained when feedback from the stimulated muscle group is incorporated into the control system. Separate experiments are described for control of both plantarflexion and dorsiflexion of the ankle joint. The variance ratio of section 3.2 is used as the figure of merit in quantifying results.

The experiments to be described and their results apply to normal subjects only. Permission had been granted to use volunteer hemiplegic subjects since some of the practical problems involved in the design of such systems will only become apparent when the systems are tested and evaluated on the group of people they are ultimately meant to serve. However, the experiments performed here required the subject to be seated. Most hemiplegics have the ability to volitionally dorsiflect in this position but are unable to do so when dorsiflexion becomes part of an overall act such as during walking. Those patients who cannot dorsiflect while seated are incapacitated to a far greater degree than those who can. Because the adjustments to be made by these patients to their post-stroke situation are more traumatic, they were found to be extremely reluctant to participate in a project where no guarantee could be given that their condition would improve as a result of their participation. No significant difference exists between the ability to stimulate muscle paralyzed due to an upper motor neuron lesion (if muscle tone has been maintained), and unparalyzed muscle (Vodovnik et al, 1967). Stimulus parameters should therefore remain constant, possible differences being in the threshold level and the absolute values of stimulus intensity. For these reasons, the results obtained from the experiments to be described should apply to the paralyzed patient as well.

5.2 Open-Loop EMG Control: System Description

Figure 5.1 is a block diagram of the system arrangement where LEG1 contains previously recorded computer-stored outputs, $EMG_1(t)$, and $\theta_1(t)$, the volitional EMG and joint angle data respectively. LEG2 represents the ankle joint and the muscles acting about it that are to be stimulated. $\theta_2(t)$ and $EMG_2(t)$ are the evoked joint position and EMG time-histories respectively, resulting from stimulation. For open-loop EMG control of joint position (solid lines in figure 5.1), the joint angles $\theta_1(t)$ and $\theta_2(t)$ are used to calibrate the system. In this mode, $EMG_1(t)$ is used to modulate stimulus intensity. The controller (CONT.) is described in detail in Chapter 4. It supplies the threshold stimulus level to the muscle group being stimulated, and contains gain amplifiers and a modulator. The stimulator (STIM.) transforms the output voltages of the controller into a constant current source and ensures isolation of the subject from the 110^V mains supply ground.

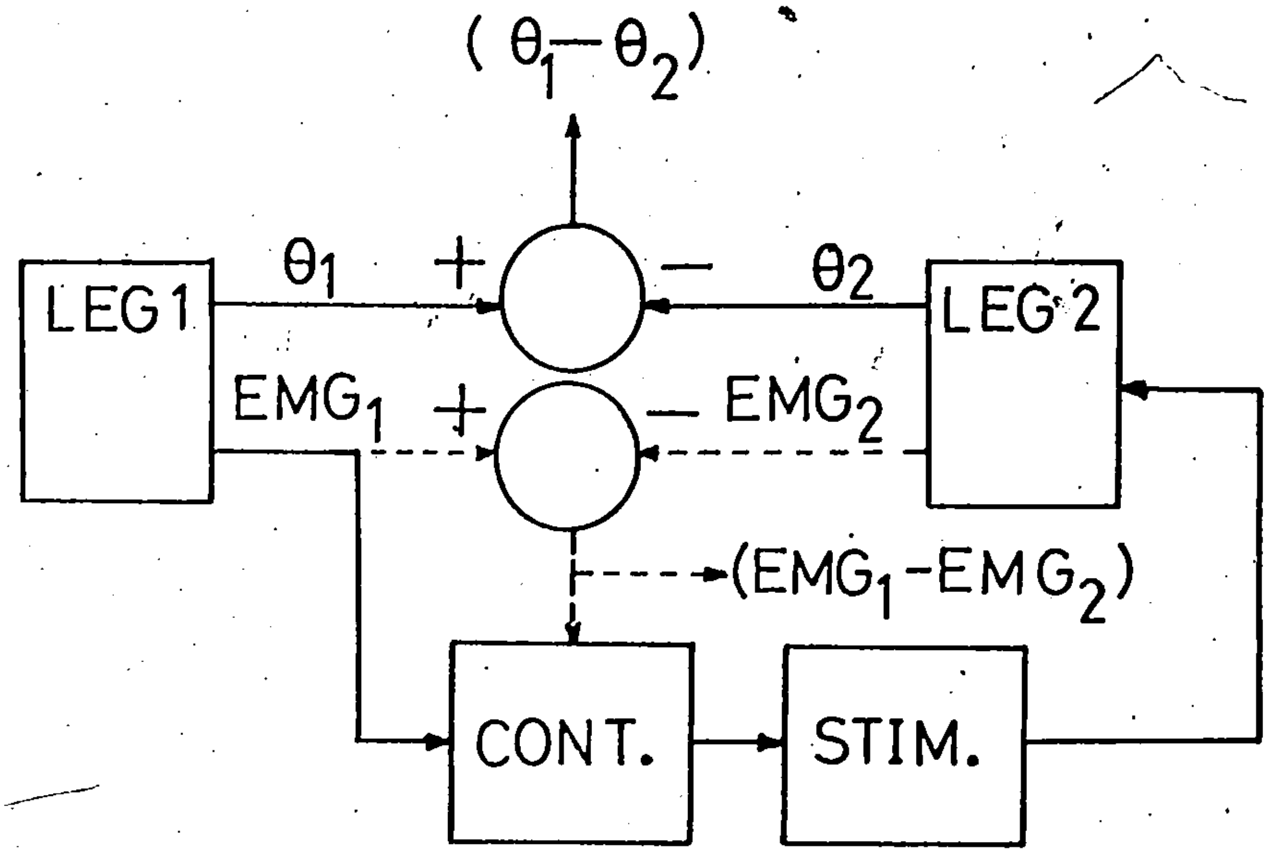


Figure 5.1 Block diagram of EMG control system

5.2.1 Recording of Control Signals

All experiments were performed on the ankle joint. Figure 5.2 shows the position of the subject during recording of the control signals and stimulation. The subject was seated and the limb of concern allowed to hang freely.

To obtain exactly repeatable records of the control signals ($\theta_1(t)$ was also used for this purpose. See Chapter 6), recording electrodes were placed on the muscles of concern (section 5.2.2) and an electrogoniometer (section 4.4) was attached about the ankle joint. The joint was then volitionally moved and the resulting analogue signals, $EMG_1(t)$ and $\theta_1(t)$, were sampled, digitized, and stored in a PDP11/10 computer. A sampling frequency of 1KHz was used as this is well above the nominal 250 Hz bandwidth of surface-recorded EMG signals (Hershler and Milner, 1977). To validate the assumption that a 1KHz sampling frequency is sufficient to prevent aliasing, EMG was recorded at a rate of 2 KHz from the tibialis anterior muscle during phasic volitional movements of the foot. Figure 5.3 is a display of the frequency spectrum of the recorded signal. No visible frequencies appear above 400 Hz. There was thus no need to band-limit the EMG signal by means of filters prior to data acquisition.

5.2.2 Placement of Electrodes

Electrode sites for recording volitional and evoked EMG data were kept the same throughout all experiments by marking their positions on the skin of the subject. For the dorsiflexors, one set of recording electrodes was placed on the tibialis anterior muscle close to where the

Figure 5.2 Position of subject during experiments

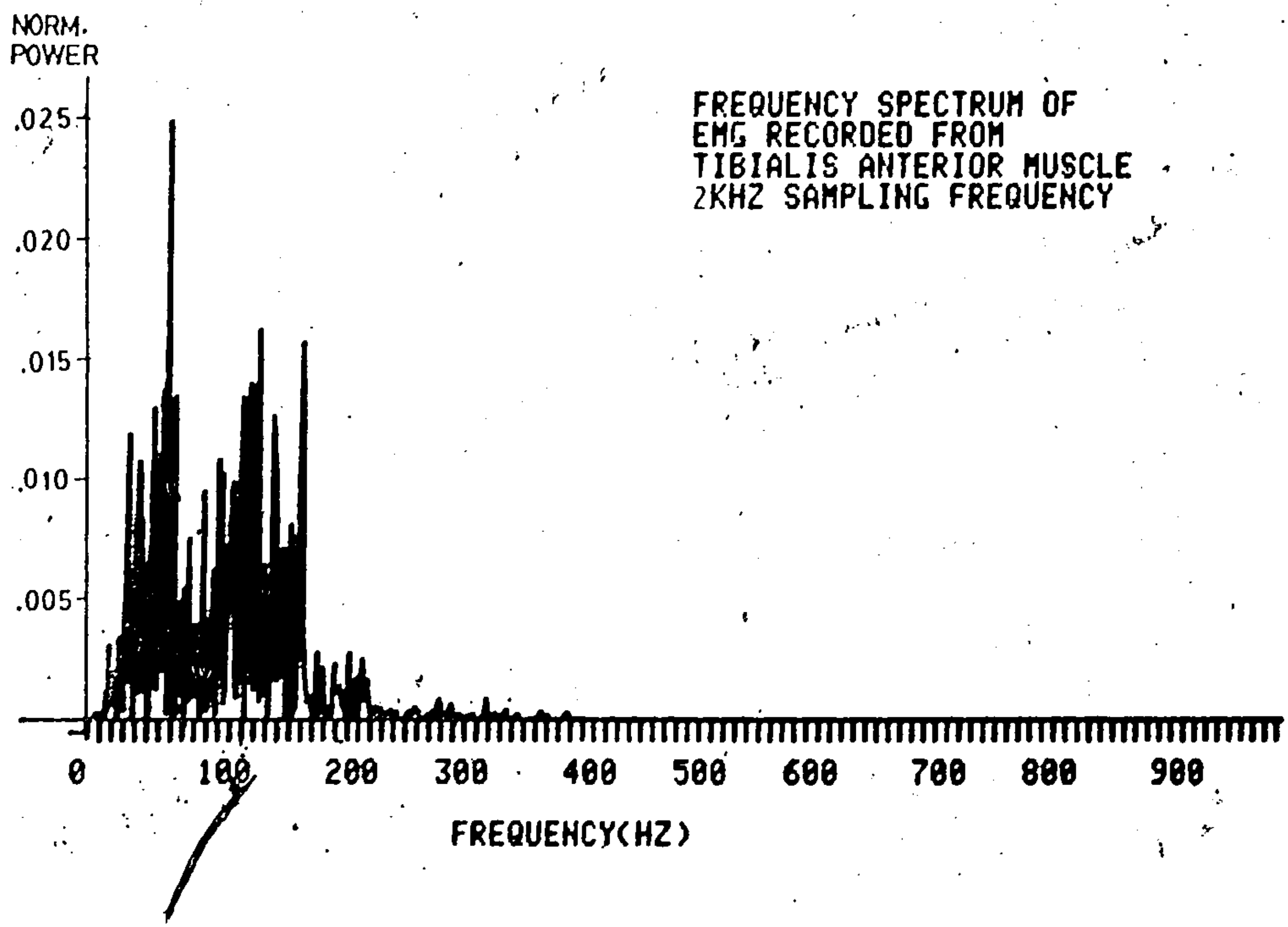


Figure 5.3 Frequency spectrum of EMG recorded from m. tibialis anterior

muscle runs into the distal tendon. For the plantarflexors, when only one pair of electrodes was placed between the heads of the gastrocnemius muscle, the processed EMG signal did not reflect the phases of the movements performed, and so proved to be a poor controller of ankle-joint position. Satisfactory control resulted from placing one set of recording electrodes on each head of the gastrocnemious muscle, the resultant EMG signals being fed into a summing amplifier shown in figure 4.7(c).

Stimulation electrodes (see sections 2.5 and 4.3) were positioned as follows:

(i) An inactive electrode of area 40cm^2 was placed on the distal anterior part of the lower leg so as not to impede movement of the ankle joint.

(ii) A search electrode was used to locate the positions where minimum stimulation produced dorsiflexion or plantarflexion without the occurrence of inversion or eversion.

(iii) For dorsiflexion, an electrode of area 14.5 cm^2 was placed over the motor point which was found to be a third of the length of the muscle measured from the proximal to the distal end.

(iv) For plantarflexion, an electrode of area 6.45 cm^2 was placed between the two heads of the gastrocnemius so that both contracted on stimulation.

5.2.3 Number of Repetitions of Each Experiment

Each set of experiments had to be completed in a single session so that electrode positions and electrogoniometer orientation remained the same. This allowed for comparison of results within each set of experiments. For example, an experiment to measure the efficacy of open-loop EMG control of dorsiflexion required 6 distinct experiments (in a set), each one employing a different time-constant of averaging for processing the EMG control signal. Recalibration of the system was required each time the window length was changed. The length of the experimental session therefore had to be minimized to prevent the subject from fatiguing and thus affecting the results of the experiments. This problem was compounded when feedback experiments were performed since the calibration procedure had to be repeated for each change in processor time-constants for both the volitional and evoked EMG signals.

It was therefore decided to limit the number of times an experiment was repeated to five.

5.2.4 Experimental Procedure

After the electrodes had been applied, the seated subject allowed his foot to hang freely, the ankle angle being about 105° with respect to the vertical. This position was taken as the 'zero' reference position. Figure 5.4 shows the stimulating and recording electrodes and the electrogoniometer attached to the leg.

The data stored in the computer was used as follows:

(i) $EMG_1(t)$ was first rectified and then smoothed with a boxcar window of specified length.

(ii) Both channels of data, $EMG_1(t)$ and $\theta_1(t)$, were read out of the computer via a digital-to-analogue (D/A) converter. Irrespective of window length, the EMG control signal was always adjusted to have the same peak amplitude. This simplified the calibration procedure for open-loop control since, irrespective of window length, the input calibration had to be set only initially.

(iii) The processed EMG signal was used to control stimulation, and the angle signals, $\theta_1(t)$ and $\theta_2(t)$, were used for calibration purposes. For open-loop control this meant adjusting the gain of the stimulus until maximum evoked joint angle, $\hat{\theta}_2$, was equal to the maximum of the recorded volitional joint angle, $\hat{\theta}_1$.

(iv) After calibration was completed, evoked EMG and angle data, $EMG_2(t)$ and $\theta_2(t)$, were read into the computer simultaneous with stimulation via analogue-to-digital (A/D) converters. The start of the process was initiated by means of a manual switch which triggered both the D/A and A/D to output and acquire data respectively. Each run was repeated a number of times for the same time-constant of averaging of the EMG processor. The acquired data were written into files, each file containing the volitional signal and the evoked signals for EMG and angle separately.

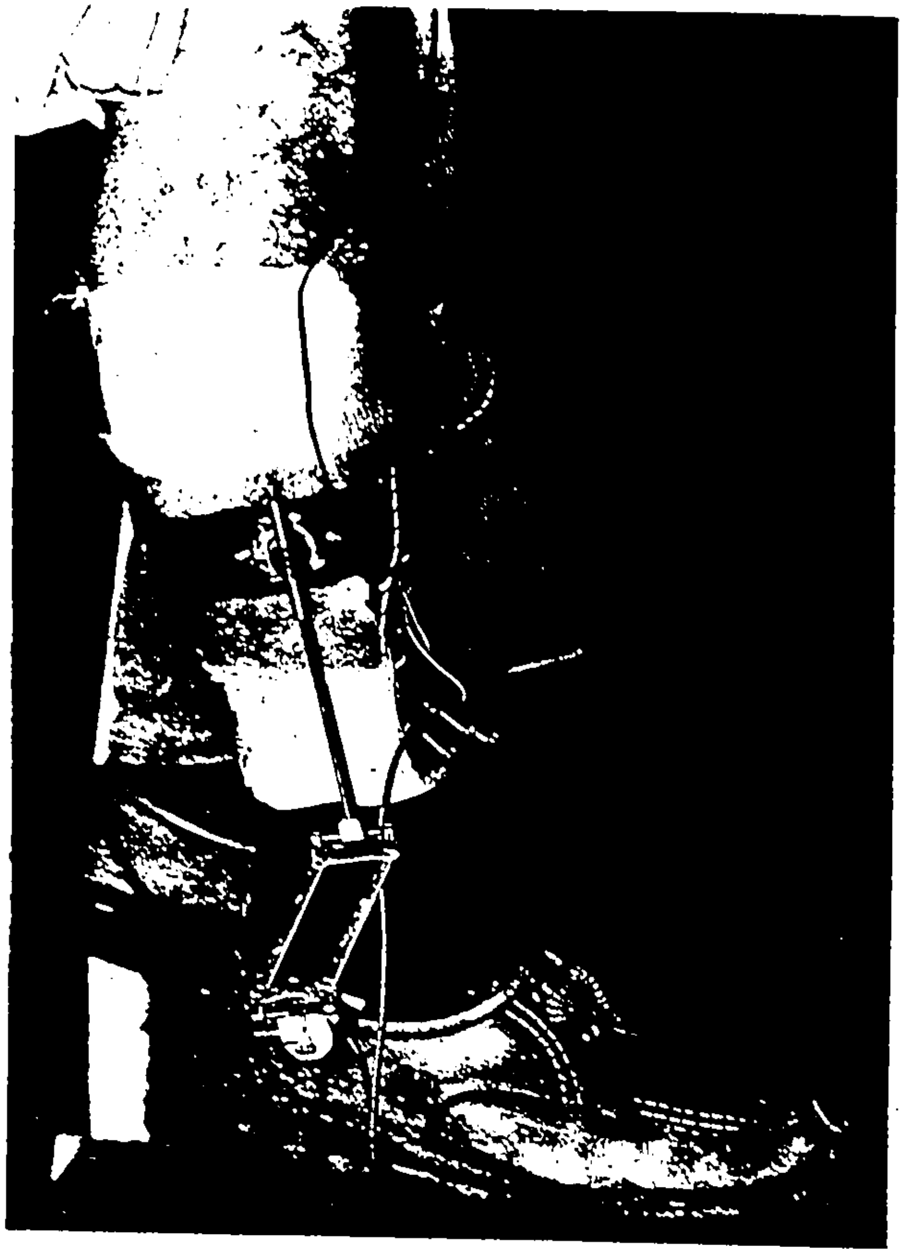


Figure 5.4 Recording and stimulating electrodes and EG attached to leg

Each experiment lasted for 2 seconds and was repeated 5 times for each window of averaging.

Figure 5.5 shows the calibration of the electrogoniometer output where 1 inch = 37.8 degrees. The potentiometer of the electrogoniometer is linear to within 1% of its nominal value.

Figure 5.6 shows the volitional angle and rectified EMG records, $\theta_1(t)$ and $EMG_1(t)$, used for controlling stimulation or for analytical purposes. From figure 5.6, maximum joint angle variation during dorsiflexion was 35.06°, and during plantarflexion, 15.83°.

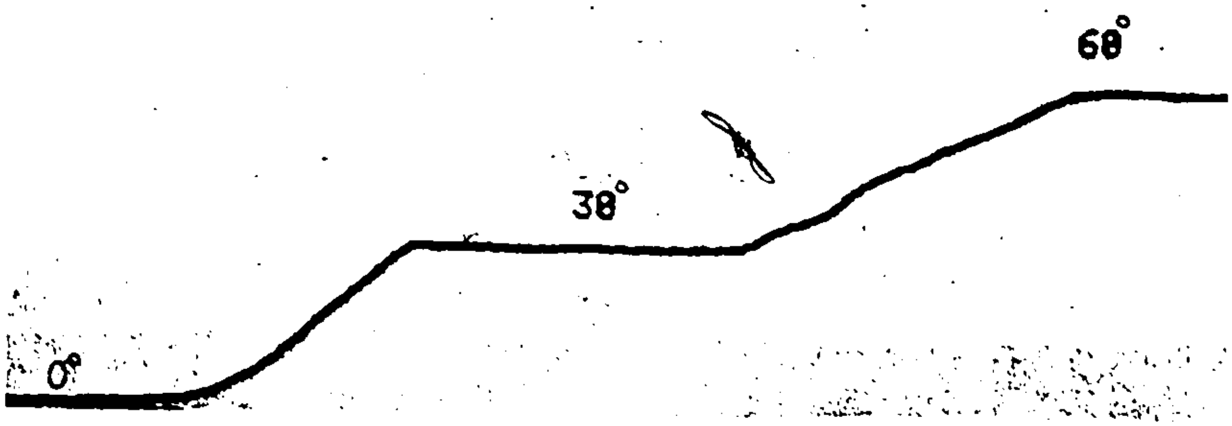


Figure 5.5 Calibration of Electrogoniometer

A deficiency in the electrogoniometer measuring system became apparant at smoothing windows of 50mS and below. Stimulation caused unpleasant prickling sensations and the foot trembled or oscillated. The electrogoniometer was not sensitive to these oscillations as demonstrated by the author when voluntarily causing the foot to tremble, and hence these oscillations are not reflected in the results. It is possible that the inertia of the goniometer plus movements of its attachments

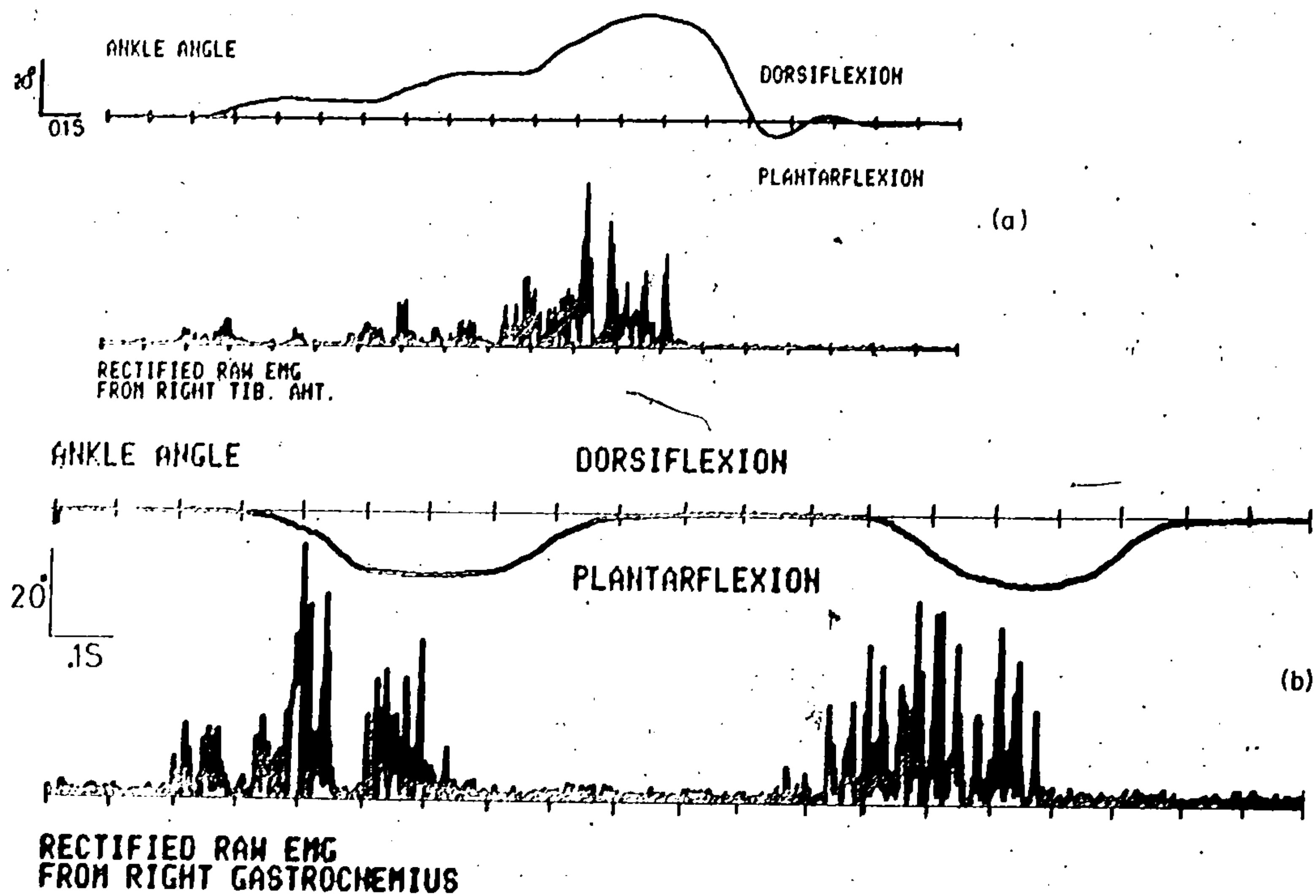


Figure 5.6 Volitional Angle and EMG records acquired simultaneously at 1 KHz sampling rate over 2 seconds
(a) Dorsiflexion (b) Plantarflexion

relative to the foot absorb high-frequency, low-amplitude variations.

5.2.5 Open-loop EMG Control: Results

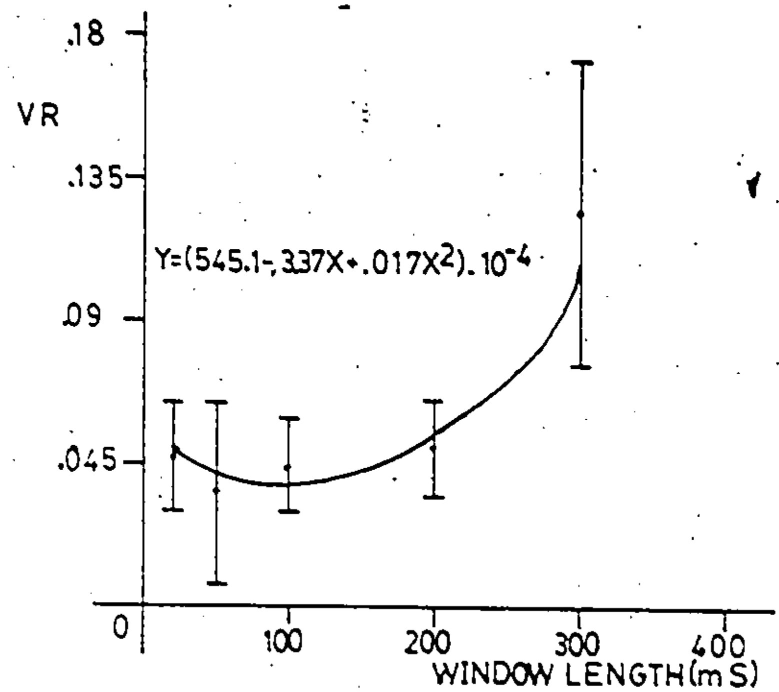
To compare the repeatability between volitional and evoked ankle-joint angles, $\theta_1(t)$ and $\theta_2(t)$, the mean value of each signal was subtracted from each data point within that signal. This avoided having to set the potentiometers of the electrogoniometers to identical positions. To align the two signals with respect to each other in time, a tolerance level was set which the signals had to exceed. All data points preceding the data point equal to the tolerance level were eliminated. Thus any delays which occurred between recording onset of volitional joint angle and the onset of the evoked response are eliminated. The numbers of sampled data points in the two records were equalized by adding a string of zeros to the end of each record. The variance ratio of section 3.2 was then invoked to compare $\theta_1(t)$ with $\theta_2(t)$.

Graphs 5.1 and 5.2 are plots of the variance ratio for all points in each stimulation of the dorsiflexors and plantarflexors respectively, as a function of the control EMG processor window length. The standard deviations are represented by bars above and below the means for each window length. First and second order polynomial regression curves were estimated from the experimental data. The data points were weighted by incorporating the sample variances when calculating the least-squares estimates of the polynomial constants. The data were best fitted by the second order polynomial as determined by the sum of squared errors normalized by the number of degrees of freedom. Confidence limits for the window length at which the variance ratio is a minimum were computed at a significance level of .05. The derivation of the equation is presented in Appendix B. The experimental data are to be found in Table 5.1. The windows yielding minimum variance ratios are 98.2 $\pm$ 5.01 mS for dorsiflexion, and 57.06 $\pm$ 5.16 mS for plantarflexion. As noted in section 5.2.2, the calculated window length of about 57 mS for plantarflexion may not be optimal due to the failure of the electrogoniometer to register the rapid oscillations which occurred at shorter window lengths.

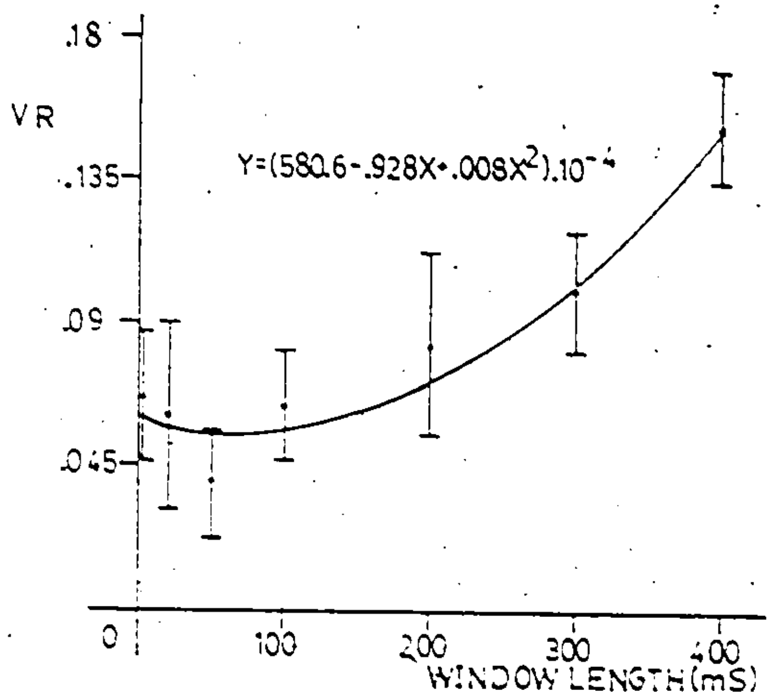
In dorsiflexion, the error between evoked and volitional angles is mainly due to the rapidity with which the foot dropped during the volitional movement. This is illustrated in figure 5.7 (a). A maximum error of 14.18° or 40.4% of the total angle variation during dorsiflexion occurred.

The primary source of error during stimulation to effect plantarflexion is the time taken for the foot to reach its equilibrium position at the end of stimulation. The evoked angle reaches this position about 100 mS after the volitional angle (Figure 5.8(a)). A maximum error of 5.43° or 34.3% of the total angle variation during plantarflexion occurred. The smaller error obtained here may be due to the composite EMG control signal used (see section 5.2.2) being more representative of the muscular activity as compared with dorsiflexion where only one set of recording electrodes was used.

One can also speculate that the larger time-constant of averaging needed to process the control EMG recorded from the dorsiflexors is due to the manner in which the EMG signals were obtained. Summing the



Graph 5.1 Variance ratio versus window length
Open-loop EMG control of dorsiflexion



Graph 5.2 Variance ratio versus window length
Open-loop EMG control of plantarflexion

activity recorded from two sets of electrodes, as was done during plantarflexion, and then rectifying the composite EMG signal, may result in a smoothing effect due to cancellation of individual spikes of activity of opposite polarity in the two records. For this reason, the ability to rectify each EMG signal prior to summation was later incorporated into the 4-channel stimulator of section 4.7.

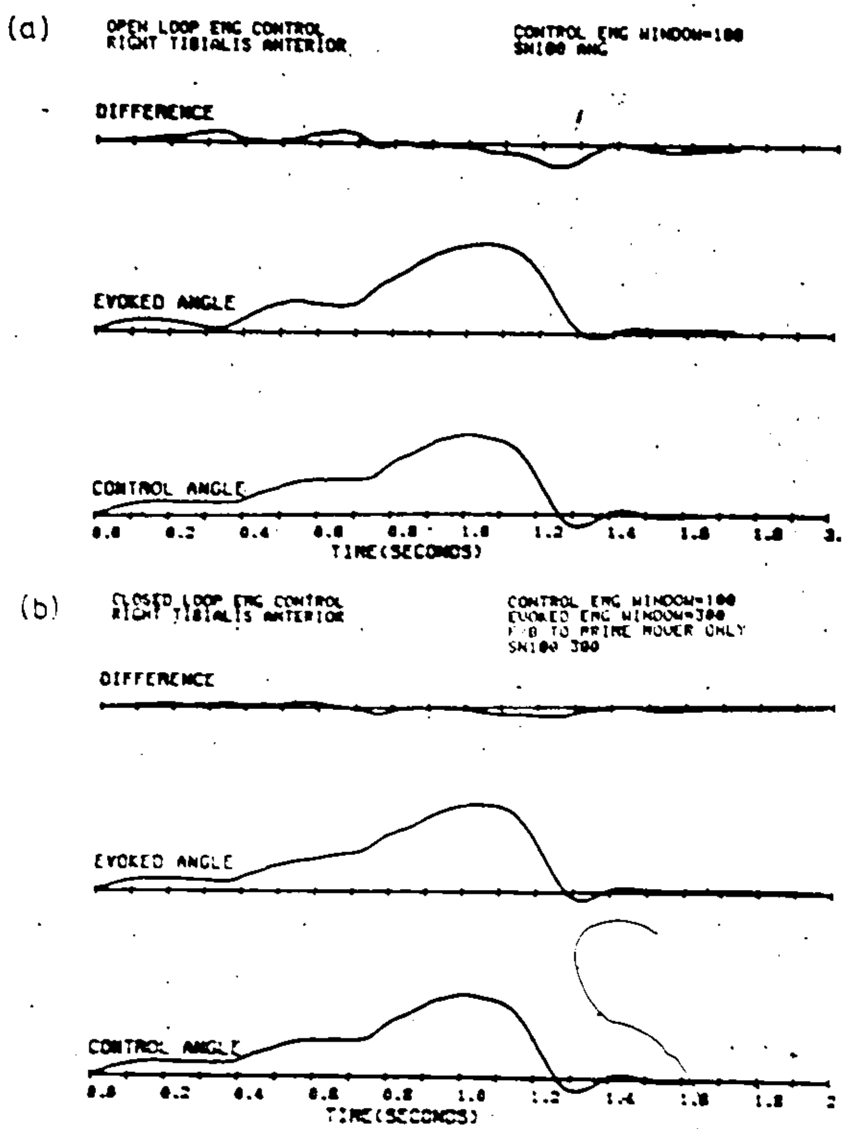


Figure 5.7 EMG control of dorsiflexion
(a) O/L (b) F/B

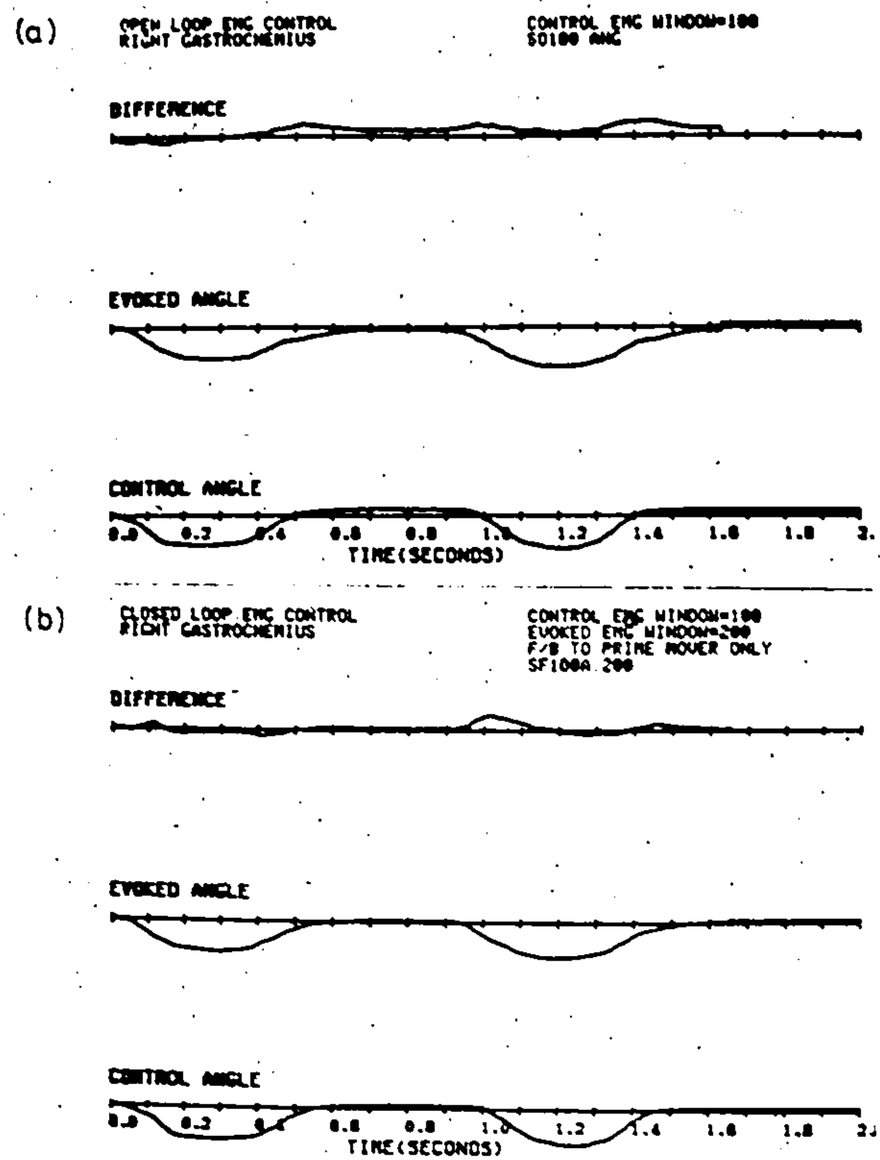


Figure 5.8 . EMG control of plantarflexion
(a) O/L (b) F/B

5.2.6 Comparison of Control and Evoked EMG Waveforms

During the experiments described in section 5.2.4, the evoked EMG singals, $EMG_2(t)$, were also digitized and stored in the computer during stimulation. These signals were later used to determine the degree of repeatability between them and the control EMG processed with the same window used during stimulation to collect each evoked EMG record. Each evoked EMG record was processed with a range of windows and the variance ratio of section 3.2 was used to compare each pair of processed control and evoked EMG signals. The existence of a high degree of repeatability between the two signals for particular time constants of averaging as reflected by the variance ratio will imply that the incorporation of feedback from the stimulated muscle in an EMG-controlled FES orthosis is plausible.

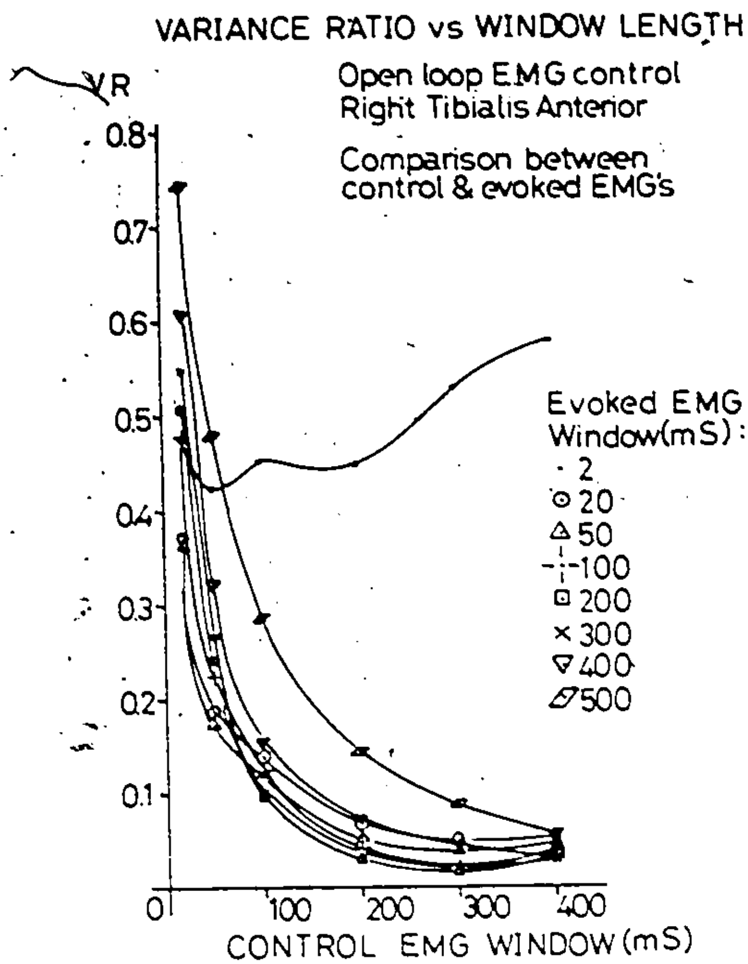
Graphs 5.3 and 5.4 show the plotted results for stimulation of the dorsiflexors and plantarflexors respectively. The curves have been drawn through the means of the data points, and indicate that a range of processing windows exist over which the two waveforms are highly correlated. For example, from graph 5.3 it can be seen that choosing a window length of 200 mS instead of 300 mS for the control EMG changes the variance ratio by about 0.01 so that window lengths of 200 mS for the four signals may provide satisfactory control. There are therefore three factors which can affect the final choice of the processor time constants. They are:

(i) The form of the control EMG when used in a feedback system that will produce the greatest repeatability between evoked and volitional joint angles, $\theta_2(t)$ and $\theta_1(t)$.

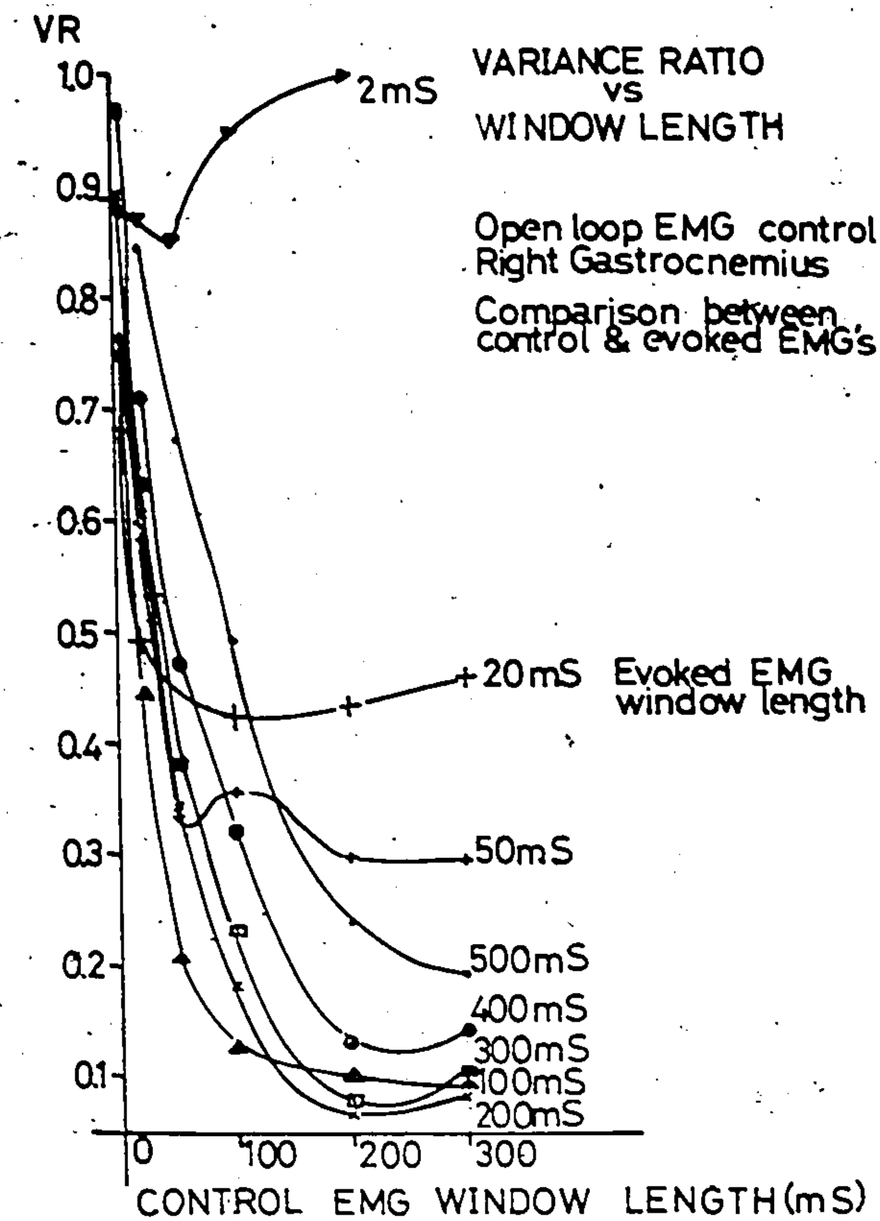
(ii) The forms of the control and evoked EMG signals, $EMG_1(t)$ and $EMG_2(t)$ that will result in the greatest similarity between these two waveforms.

(iii) The form of the evoked EMG_1 $EMG_2(t)$, when used in an EMG feedback control system that will produce the greatest repeatability between volitional and evoked joint angles, $\theta_1(t)$ and $\theta_2(t)$.

The experiments incorporating EMG feedback control are described in section 5.3.



Graph 5.3 Comparison between volitional EMG and evoked EMG (dorsiflexion)

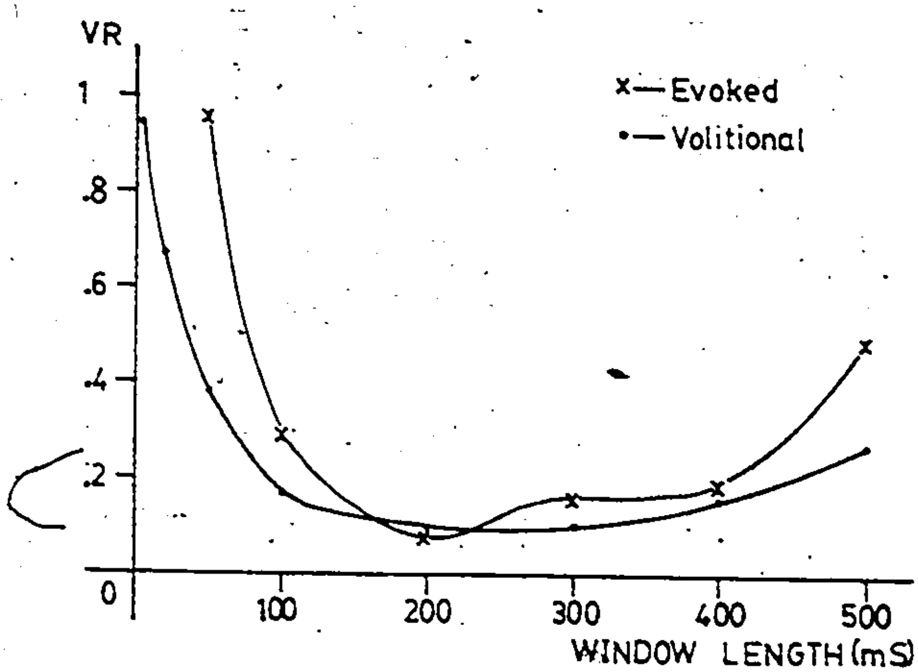


Graph 5.4 Comparison between volitional EMG and evoked EMG (plantarflexion)

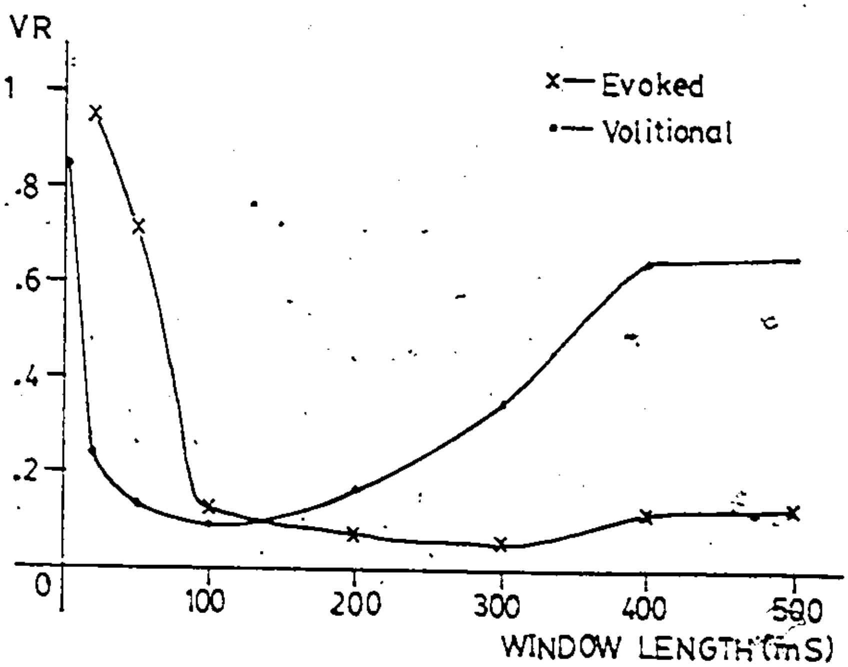
5.2.7 Relationship Between Joint Angle Variations and EMG Waveforms

Radonjic and Long (1970) (see section 3.1.8) ascribe the fact that voluntary control is the control of position and not of muscle tension to which EMG is related, to one of the reasons why difficulties are encountered when myoelectric control is used. As noted in section 4.6, a definite similarity exists between joint-angle time histories and their corresponding EMG envelopes. These two facts justified examining whether indeed position information is contained in concomitant EMG signals. Also of interest, then, will be the comparison between the forms of the processed EMG signals (in terms of the time-constant of averaging used) that produce the greatest repeatability between volitional and evoked joint angles, and the form of the processed EMG signals that contain the greatest amount of joint-angle information. Again, the variance ratio of section 3.2 was used as a means of quantifying the results.

The signals shown in figure 5.6 were used in the comparison between volitional angle and EMG trajectories. The EMG signals were processed with a range of window lengths and each resultant waveform was compared to its corresponding joint angle trajectory using the variance ratio. Similarly, for the comparison between evoked EMG and angle waveforms, the evoked signals $EMG_2(t)$ and $\theta_2(t)$ resulting from stimulation with a control EMG window length of 100 mS, were used. The results are shown in graphs 5.5 and 5.6 for dorsiflexion and plantarflexion respectively. The variance ratio is a minimum at window lengths of 250 mS and 100 mS for the volitional case, and for the evoked case a window lengths of 200 mS and 270 mS for dorsiflexion and plantarflexion respectively.



Graph 5.5 Comparison between EMG and joint angle waveforms (dorsiflexion)



Graph 5.6 Comparison between EMG and joint angle waveforms (plantarflexion)

Comparing the form of $EMG_1(t)$ that produces the greatest repeatability between $\theta_1(t)$ and $\theta_2(t)$ during stimulation with the form of $EMG_1(t)$ that contains the greatest amount of information on $\theta_1(t)$, the window lengths in the latter case are about twice those in the former case. This implies some form of filtering between the input, $EMG_1(t)$ when used to modulate stimulus intensity, and the resultant evoked joint angle, $\theta_2(t)$. As can be seen from figure 5.6, the electrical activity of the muscle preceeds the mechanical movement of the joint. Thus the filtering action most probably is due to the mechanical properties of the muscle and the inertia of the limb.

Examples of the processed EMG and angle waveforms for dorsiflexion are shown in figure 5.9. Note the similarity between these waveforms. Thus the plausibility of employing processed EMG in the control of an FES orthosis has been demonstrated. As noted in section 5.2.6, the final forms of the EMG waveforms must depend upon the performance of the system when feedback control is incorporated.

5.3 EMG Feedback Control of Ankle-Joint Position

The incorporation of feedback is represented by the dashed lines of figure 5.1. $EMG_1(t)$ and $\theta_1(t)$ were again read out of the computer via the D/A, $EMG_1(t)$ having the same peak amplitude irrespective of window length. $EMG_2(t)$, the evoked EMG due to stimulation, was processed by the analogue rectifier-averager of section 4.5. Calibration consisted of ensuring that both $EMG_1(t)$ and $EMG_2(t)$ were equal in amplitude, as well as $\theta_1(t)$ and

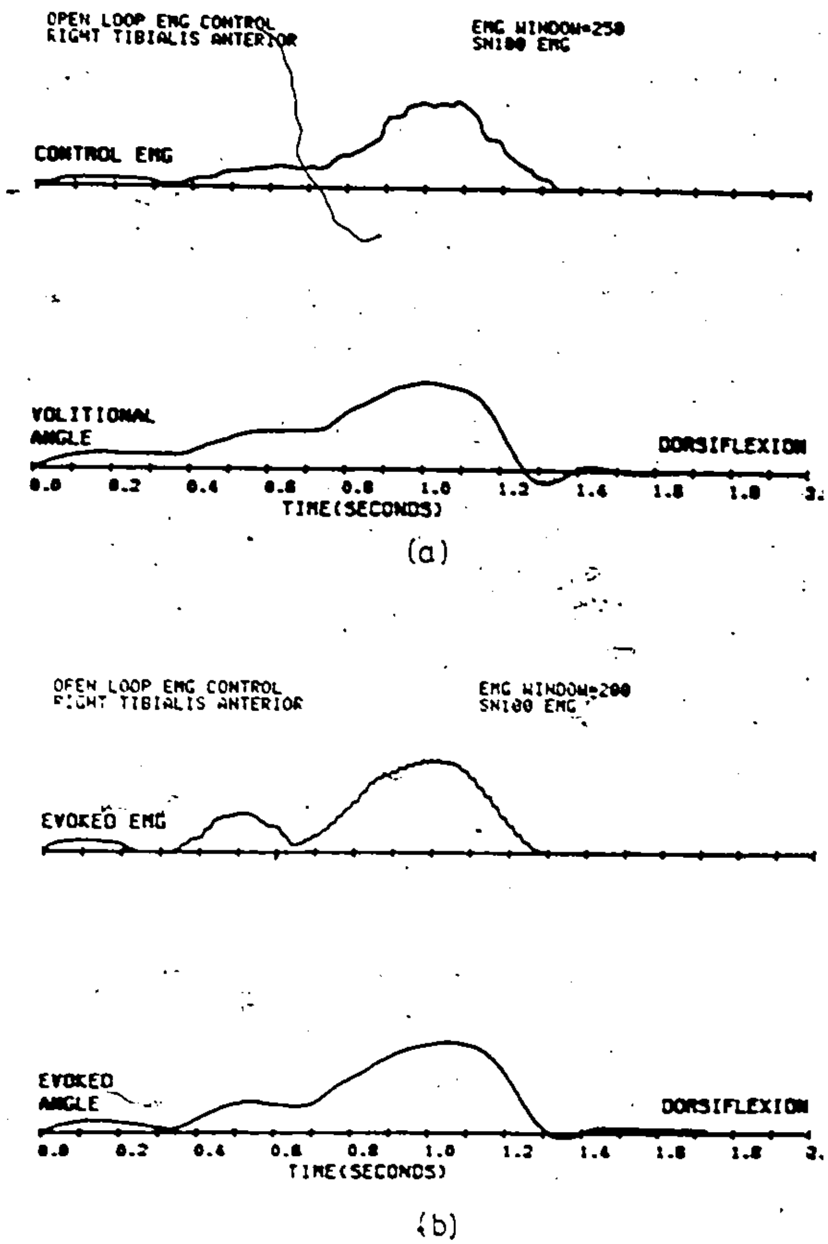


Figure 5.9 Comparison of processed EMG and angle waveforms for dorsiflexion

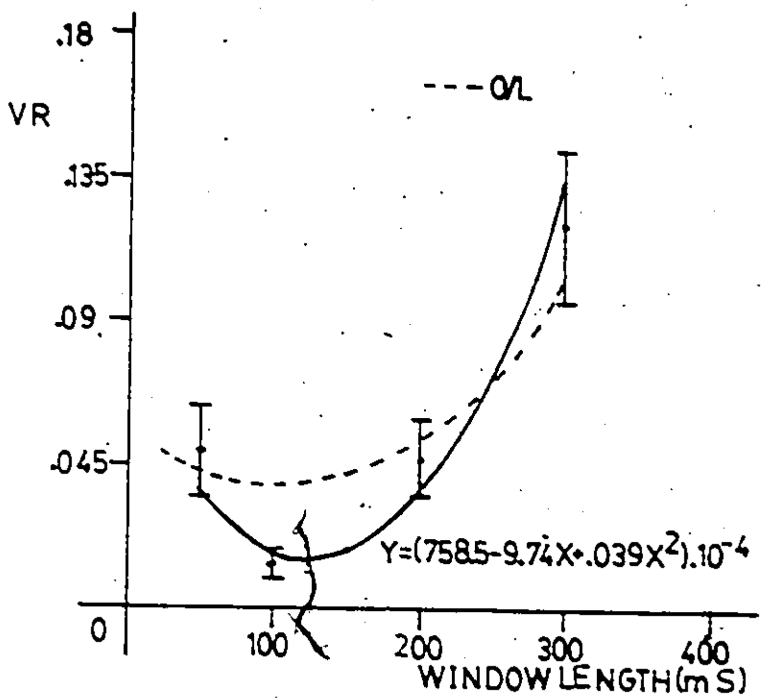
(a) volitional
(b) evoked

$\theta_2(t)$. Thus calibration was achieved by sequentially and repeatedly adjusting the feedback signal calibration control, the level shift and the gain control, and had to be repeated for each change in window length or RC time-constant.

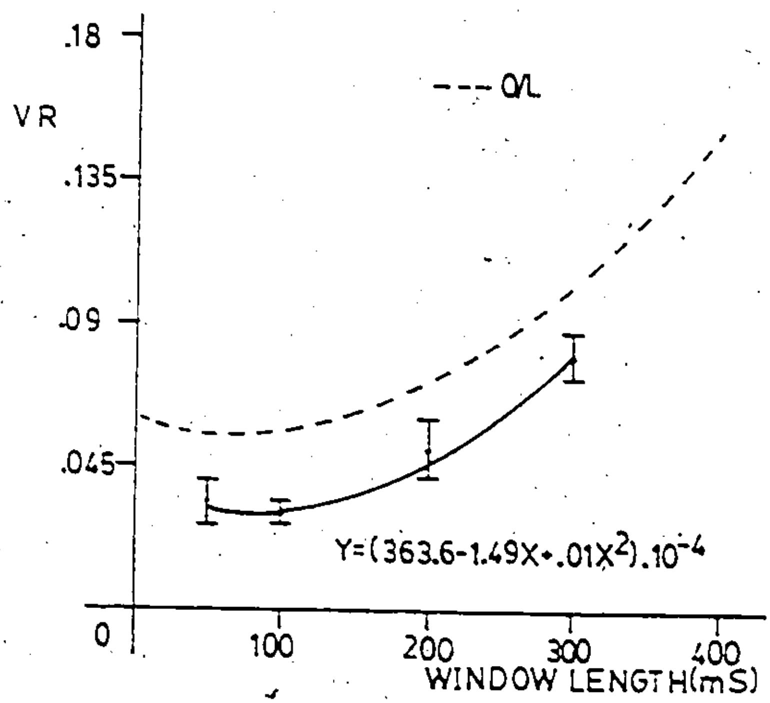
Boxcar window lengths of 50, 100, 200 and 300 mS were used on $EMG_1(t)$ for each RC time-constant. The RC time-constants used were 24.2, 48.4 and 72.6 mS. For both plantarflexion and dorsiflexion, the 72.6 mS time-constant yielded the most efficacious control. Graphs 5.7 and 5.8 show the estimated regression curves of the variance ratio as a function of the control EMG processor window length for both the open-loop case and feedback resulting from the 72.6 mS RC time-constant of processing for $EMG_2(t)$. (See Appendix B). The windows yielding minimum variance ratios, in the case of feedback, are 125.27 $\pm$ 9.37 mS for dorsiflexion, and 74.95 $\pm$ .56 mS for plantarflexion at a significance level of .05.

5.3.1 Comparison between Open-Loop and Feedback EMG Control of Joint Position

Table 5.1 contains the experimental data resulting from the open-loop and feedback experiments. An analysis of variance was performed (Winkler and Hayes, 1975) to ascertain whether the choice of the processor time-constant and open-loop or feedback control affects the correlation between the volitional and evoked joint angles (i.e. the variance ratio). For both plantarflexion and dorsiflexion, the choice of the processor time-constant significantly affects the variance ratio far beyond the .01 level. However, the choice of open-loop or feedback



Graph 5.7 Variance ratio versus window length
EMG feedback control of dorsiflexion



Graph 5.8 Variance ratio versus window length
EMG feedback control of plantarflexion

control is only significant ($<.01$) in the case of plantarflexion. A possible explanation for this can be obtained by examining graphs 5.7 and 5.8. Over the 50 mS to 300 mS range of window lengths, the open-loop and feedback curves for plantarflexion are distinctly separate. In the case of dorsiflexion, the two curves intersect. Since analysis of variance does not distinguish between different parts of the curves, the obtained result is expected. However, in the region of the window lengths where the variance ratio is a minimum, the variance ratio for feedback control is significantly different from the variance ratio for open-loop control.

Figures 5.7(b) and 5.8(b) are examples of the evoked angles obtained, the volitional angles, and their differences for dorsiflexion and plantarflexion respectively. The maximum differences between $\theta_1(t)$ and $\theta_2(t)$ are 3.7° during dorsiflexion, and 3.07° during plantarflexion. The effect of feedback control is most striking in the case of dorsiflexion where the maximum difference between $\theta_1(t)$ and $\theta_2(t)$ has been decreased by over 70% when compared with open-loop control. A decrease in error of about 40% was achieved in the case of plantarflexion. In both cases, the stimulated foot reached its equilibrium position more rapidly when feedback was included. Thus the effect of feedback has been to decrease the response time of the system.

The increase in window lengths needed to process $EMG_1(t)$ when feedback was included is expected since, from section 5.2.4, the volitional EMG window lengths needed to produce the greatest repeatability between $EMG_1(t)$ and $EMG_2(t)$ are much larger than those required to produce the greatest repeatability between $\theta_1(t)$

DORSIFLEXION							
	WINDOW						
MODE	-	20	50	100	200	300	
O/L		.0679	.0914	.0542	.0347	.1535	
		.0357	.0155	.0177	.0332	.0615	
		.0353	.0160	.0600	.0722	.0720	
		.0653	.0252	.0409	.0491	.1771	
		.0298	.0319	.0494	.0623	.1604	
F/B			.0317	.0109	.0317	.1329	
			.0704	.0114	.0622	.0788	
			.0369	.0216	.0351	.1109	
			.0550	.0191	.0550	.1469	
			.0525	.0092	.0525	.1332	

PLANTARFLEXION							
	WINDOW						
MODE	2	20	50	100	200	300	400
O/L	.0685	.0202	.0598	.0980	.1403	.1284	.1861
	.0657	.0330	.0299	.0525	.0698	.1060	.1448
	.0433	.0769	.0192	.0552	.0781	.1035	.1430
	.1011	.0949	.0298	.0547	.0638	.0712	.1406
	.0535	.0799	.0590	.0601	.0642	.0906	.1427
F/B			.0424	.0365	.0681	.0946	
			.0410	.0303	.0413	.0792	
			.0225	.0312	.0480	.0763	
			.0302	.0283	.0456	.0758	
			.0321	.0253	.0509	.0758	

Table 5.1 Variance ratios for EMG control of the ankle joint

and $\theta_2(t)$ during open-loop EMG control. Only part of this difference between the latter two cases can be accounted for by the filtering effect that occurs between translating the electrical input into a mechanical output as discussed in section 5.2.7. Thus the values of time constants of processing obtained for the case of feedback control are reasonable.

5.4 Conclusions

The author has demonstrated the feasibility of including EMG feedback, from the muscles being stimulated, into an EMG system for controlling joint position. In so doing, it has been demonstrated that the temporal angular variations of the joint position are largely contained within the EMG signals recorded from the prime movers during specific movements. (Naumann and Milner, 1978a). This result is significant in that, while a linear relationship has been established between processed EMG and force during isometric or isotonic contractions, (Ralston, 1961), this is the first time that such a relationship has been established between processed EMG and joint position for movements that are not necessarily isotonic. Practical values of RC time-constants for averaging the EMG signals are about 37 mS for the control EMG signals, and 73 mS for evoked EMG signals.

CHAPTER 6

POSTION CONTROL OF JOINT ANGLE

Open-loop and feedback experiments to control ankle joint position are described in this chapter. Similarly to Chapter 5 where open-loop and feedback EMG control of stimulation were compared, the variance ratio of section 3.2 has been invoked to assess the efficacy of different position control modes. The same angular time-histories used in Chapter 5 to determine EMG processor time-constants and the efficacy of EMG open-loop and feedback control, were used here so that comparisions between EMG and position control could be made.

6.1 Experimental Procedure

Figure 6.1 is a block diagram of the experimental control system where LEG1 contains the prerecorded computer stored volitional joint angle time-histories, $\theta_1(t)$, shown in figure 5.6. Prior to outputting from the computer, the first time-point of $\theta_1(t)$ was assigned zero value. The level of the remainder of the signal was adjusted by adding or subtracting (depending on whether the first time-point was negative or positive) the absolute value of the first time-point to or from all the other data points. θ_2 is the angle of the joint LEG2, to be stimulated and is measured by means of an electrogoniometer. Initially θ_2 was set by allowing the foot to

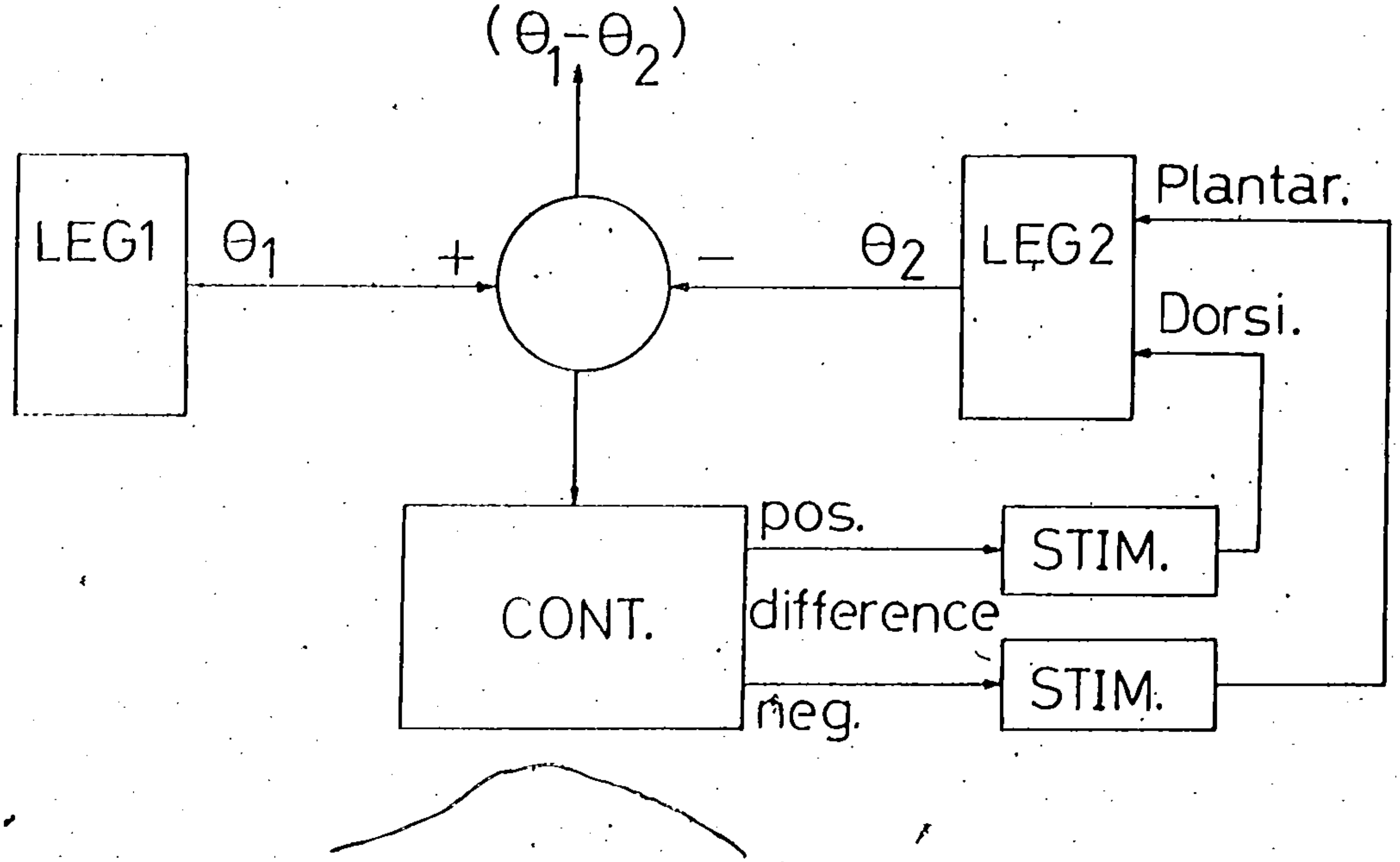


Figure 6.1 Block diagram of position controlled system

hang freely as shown in figure 5.2. With θ_1 set to zero, the level shift (section 4.2) was adjusted so that any DC voltage resulting from the potentiometer of the electrogoniometer not being set to zero resistance was compensated for. The controller then ensured that during stimulation a positive difference between $\theta_1(t)$ and $\theta_2(t)$ resulted in stimulation of the dorsiflexors of LEG2 whereas a negative difference resulted in stimulation of the plantarflexors. Simultaneous with stimulation, $\theta_2(t)$ was digitized and stored in the computer for later analysis.

6.2 Experimental Description

Four types of experiments were performed for both dorsiflexion and plantarflexion of the ankle joint. Open-loop and feedback positon control experiments, by stimulating only one muscle group, were performed so that the efficacy of EMG control (Chapter 5) could be compared to that of position control. Thus the experiments performed were (refer to figure 6.1) :

(i) Open-loop control of one muscle group only (i.e. $\theta_2(t)$ not fed to the summing point and either dorsiflexors or plantarflexors stimulated, not both).

(ii) Feedback to the protagonist ($\theta_2(t)$ now fed to the summing point but only one muscle group stimulated).

(iii) Feedback to, and stimulation of, both dorsiflexors and plantarflexors.

(iv) As in (iii) but with the PI-filter of section 4.5 included in the control circuit.

Calibration consisted of ensuring that $\hat{\theta}_1$, the maximum of the control angle was equal to $\hat{\theta}_2$, the maximum of the evoked joint angle. Prior to performing the two experiments where both plantarflexors and dorsiflexors were stimulated, calibration was performed for both directions of movement.

Each experiment was repeated 5 times. The stored evoked joint-angle time-history, $\theta_2(t)$ was then manipulated in the same manner as $\theta_1(t)$ by assigning the first data-point zero value and adjusting the levels of the subsequent data points accordingly.

6.3 Position Controlled Electrostimulation: Results

The means and standard deviations of the variance ratio for each type of experiment are given in Table 6.1 together with the maximum difference (in degrees) between the control and evoked ankle joint time-histories. Examples of the waveforms obtained are shown in figures 6.2 and 6.3 for stimulation to effect dorsiflexion and plantarflexion respectively.

The insensitivity of open-loop position control is evident from figure 6.2(a), the evoked angle waveshape failing to reflect the smaller variations present in the control signal. As in the case of open-loop EMG control (section 5.2.5), the primary source of error during stimulation to effect dorsiflexion (figure 6.3(a)) was the time taken for the foot to reach its equilibrium position once stimulation was terminated.

For dorsiflexion, feedback to the protagonist resulted in no improvement in the repeatability between $\theta_1(t)$ and $\theta_2(t)$ as reflected by the variance ratio. Also, the maximum difference between θ_1 and θ_2 increased from 8.9° (open-loop case) to 13°. The reasons for this are apparent from the waveshapes of figure 6.2(b). While the evoked waveshape now reflected the variations in the control signal more closely, the foot reached its equilibrium postion more rapidly than the control signal thus causing the large error difference between the two angles. That this effect was as a result of active stimulation can be seen from the absence of any plantarflexion of the stimulated foot. Applying feedback to the protagonist during stimulation to effect plantarflexion resulted in a marked improvement as reflected by both the variance ratio and visual inspection of the waveforms of figure 6.3(b). Here, the difference between the control and evoked signals resulted from both a more rapid rate of movement of the stimulated foot at the initiation, and a less rapid movement at the termination of stimulation when compared with the time-history of the control signal. The response time of the signal in this case had been decreased when compared to that of the open-loop system.

Stimulation of both muscle groups together with feedback resulted in the system becoming unstable and oscillations occurred. This effect was more pronounced in the case of plantarflexion (figure 6.3(c)), the excursions due to oscillation being larger than the maximum variation in the control signal.

CONTROL MODE	MUSCLE GROUP	EXPERIMENTAL MODE	MAXIMUM ANGULAR DIFFERENCE (degrees)	VARIANCE RATIO MEAN $\bar{X}$	VARIANCE RATIO STANDARD DEVIATION σ 2
EMG	DORSI.	O/L	14.18	.0444	.0148
	PLANT.		5.43	.0663	.0013
	DORSI.	F/B FROM PROTAG.	4.02	.0144	.0049
	PLANT.		3.07	.0292	.0037
ANGLE	DORSI.	O/L	8.90	.0501	.0187
	PLANT.		4.02	.0348	.0033
	DORSI.	F/B FROM PROTAG.	13.00	.0509	.0138
	PLANT.		3.78	.0240	.0063
	DORSI.	F/B FROM PROTAG. & ANTAG.	23.37	.0359	.0104
	PLANT.		19.80	.2194	.0396
	DORSI.	F/B WITH PI-FILTER	8.30	.0291	.0040
	PLANT.		2.61	.0110	.0030

Table 6.1 Comparison of Angle and EMG Control Modes

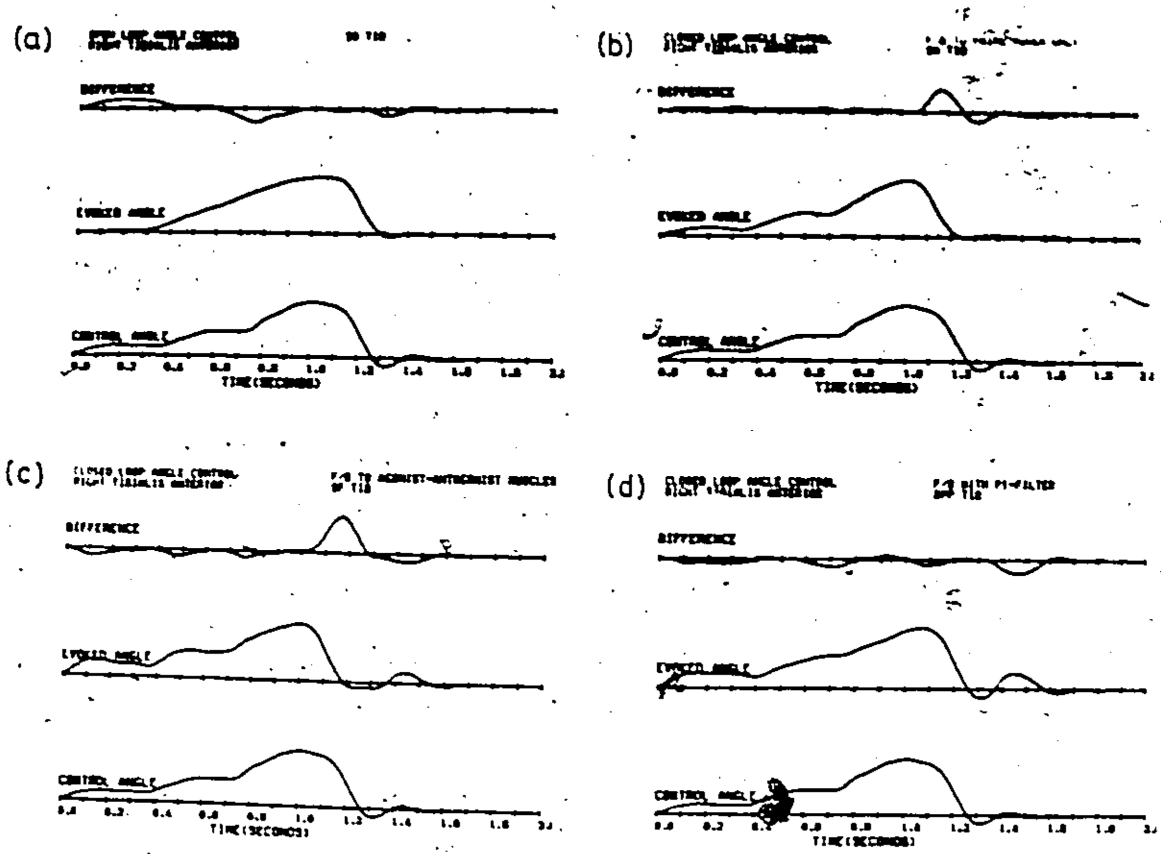


Figure 6.2 Position controlled stimulation to effect dorsiflexion

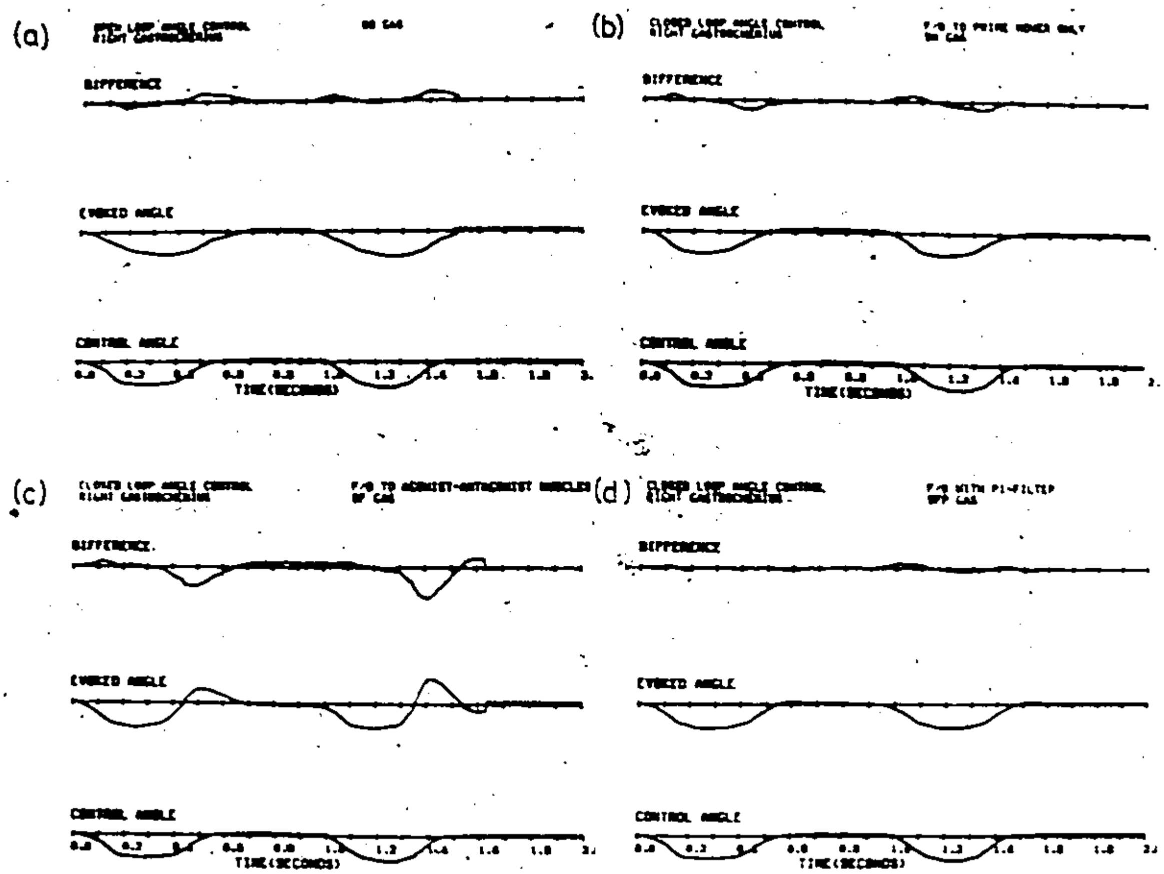


Figure 6.3 Position controlled stimulation to effect plantarflexion

Inclusion of the PI-filter when both muscle groups were stimulated resulted in a more stable system. From figure 6.2 it can be seen that plantarflexion of the stimulated foot occurred for this mode of control only (figure 6.2(d)). Figure 6.3(d) illustrates the effectiveness of the PI-filter in damping any oscillations.

6.4 Comparison of EMG and Position Control

Comparisons between EMG and position control efficacy can only be made in the cases of open-loop control and feedback to the stimulated muscle group. This is because EMG-controlled stimulation of the dorsiflexors and plantarflexors simultaneously was not performed. Although EMG and position control experiments were performed on different days, stimulating electrode positions were kept constant, so that a gross comparison between the two control modes is possible.

From table 6.1 it can be seen that EMG-controlled stimulation with feedback to the protagonist yielded values of the variance ratio that are at least comparable with those using position control. Thus EMG control is as efficacious as position control for the experimental configuration used.

The following are possible advantages of position control over EMG control:

(i) Less circuitry is required since only one control signal and one feedback signal are needed to control both dorsiflexion and plantarflexion of the ankle joint.

(ii) The input circuitry needed to process EMG signals is not needed (Of the three plug-in boards of the 4-channel stimulator, one of these boards is dedicated to this purpose).

(iii) Calibration is simpler since only manipulation of the gain control is needed. For EMG calibration, the gain control is adjusted together with the input calibration control and the level shift (see section 5.3).

Thus the greatest benefit to be reaped from position control is the reduction in the amount of circuitry required. This makes for a more compact device which is more portable and consumes less energy.

Possible advantages of EMG control over position control are:

(i) EMG control mimics in a gross fashion the desired programmatic sequences and levels of activation required on the paralyzed side.

(ii) Recording electrodes are less obtrusive than electrogoniometers. Patient acceptance of the orthosis should therefore be increased.

(iii) The absence of functional plantarflexion in stroke patients occurs infrequently (Kralj et al, 1971). For patients where the absence of active dorsiflexion is the only problem, the EMG-controlled system can be adapted to reduce the amount of stimulation to the dorsiflexors in proportion to the level of EMG activity in the plantarflexors. This should then mimic the process of reciprocal inhibition (section 2.6) whereby the muscular contractions occur without opposition.

6.5 Conclusions

A method has been described whereby the efficacy of different position control modes can be assessed. The ability of the PI-filter to suppress oscillations arising from feedback control has been demonstrated. For the experimental configuration used, it has been concluded that EMG is as effective as joint position when used to control joint angle by means of electrical stimulation. Advantages and disadvantages of EMG and position control when compared with each other have been discussed. The final choice of control mode will eventually depend upon the results obtained from long-term assessments of the two systems.

CHAPTER 7

CONTRALATERAL CONTROL OF GAIT

7.1 Introduction

This chapter describes the development and implementation of an adaptive program to control stimulation of paralyzed muscles acting about the ankle joint during walking in a straight line by utilizing signals recorded from the contralateral side.

The first part of this chapter consists of a limited introduction to some of the descriptors of human gait. The concepts presented will be utilized in both the development of the control program and the analysis of results following from implementation of the program.

The final section of this chapter is a summary of the work presented in this thesis and includes recommendations for future possibilities and directions to be taken in the field of electrostimulation of paralyzed muscle.

7.2 Human Gait

Human locomotion can be defined as the displacement of the body from one point to another. This displacement is usually accomplished by expending the minimum amount of energy possible under the constraint that the gait is bipedal (Piezer et al, 1969). Because of this bipedal characteristic, movements of body segments occur in all planes. The electrogoniometers (section 4.4) used here to measure joint variations can only do so in the sagittal plane. However, the greatest displacements of the leg segments occur in the plane of progression i.e. the sagittal plane. Thus only movements in this plane will be considered.

7.2.1 The Gait Cycle

The gait cycle includes all events occurring from the moment one heel strikes the ground until the same heel contacts again. The gait cycle consists of two phases: the stance phase in which the leg receives support from the ground, and the swing phase in which the leg swings forward. The stance phase is further subdivided as follows:

(i) Heel-strike (HS) which indicates the start of the stance phase (ST).

(ii) Foot-flat (HT) where both the heel and toes are in contact with the ground.

(iii) Toe-only (TO) where the heel is now no longer in contact with the ground.

(iv) Toe-off (SW), the initiation of the swing phase.

The gait cycle can be represented diagrammatically by arbitrarily assigning different constants to each subdivision of the gait cycle. This is depicted in figure 7.1

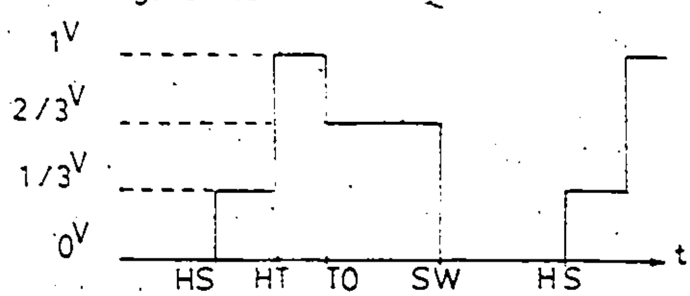


Figure 7.1 Phases of the gait cycle

The footswitch (implemented in a variety of forms in different gait laboratories) is a means whereby these phases of the gait cycle can be measured. Those used in this study are comprised of two switches which are attached to the heel and toe of the shoe sole. A DC voltage is applied to a resistive network such that different voltage levels (as indicated in figure7.1) result for the different gait-cycle phases due to the footswitches opening and closing.

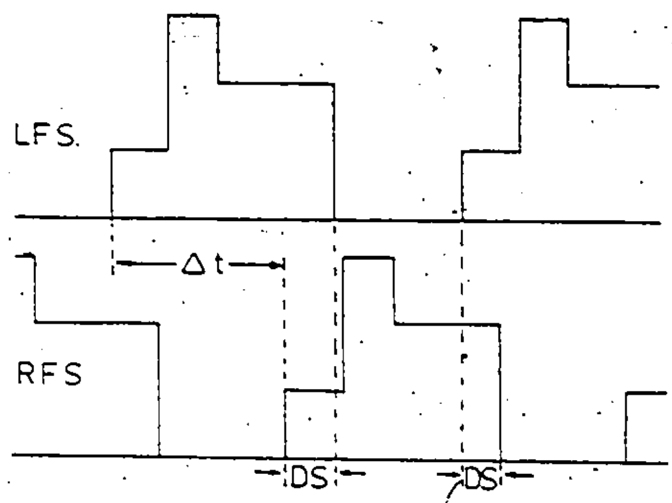


Figure 7.2 Schematic of footswitch patterns obtained from left and right sides during locomotion.

Figure 7.2 represents footswitch output levels obtained simultaneously from both the left and right sides during locomotion. For walking, the stance phase of one leg overlaps that of the other leg. This period, during which both feet are in contact with the ground, is called the double-support (DS) phase and is depicted in figure 7.2. For normal gait a symmetry exists between the two sides,the one pattern occuring time t after the other. An examination of the joint trajectories will yield the same observation. This symmetry exists during steady-state gait (see section 7.2.2). Minor differences between the two sides do occur and vary for different individuals.

7.2.2. Locomotion

The act of walking in a straight line can be subdivided into three phases. These are:

(i) Initiation of the walk. The person starts to walk from a stationary, standing position.

(ii) Steady-state gait. Initially, the body is accelereated from zero velocity. Within a few footsteps, a steady-state average speed for each step may be reached.

(iii) Termination of the walk. In order to again reach a stationary position, deceleration must occur.

Figure 7.3 is an example of typical footswitch patterns and ankle-joint angle variations recorded from a subject during the motions just described. As can be seen from the figure, symmetry between the two sides of the body does not exist at the initiation and termination of gait.

7.2.3 Angle-Angle Diagrams

A means is needed whereby the effects of stimulation during locomotion can be asssessed. One such means in the form of a graphical display is the angle-angle diagram. Since steady-state walking is a cyclic phenomenon, if the joint angle histories are plotted against each other (Grieve, 1968), the gait cycle appears as a closed loop as shown in figure 7.4.

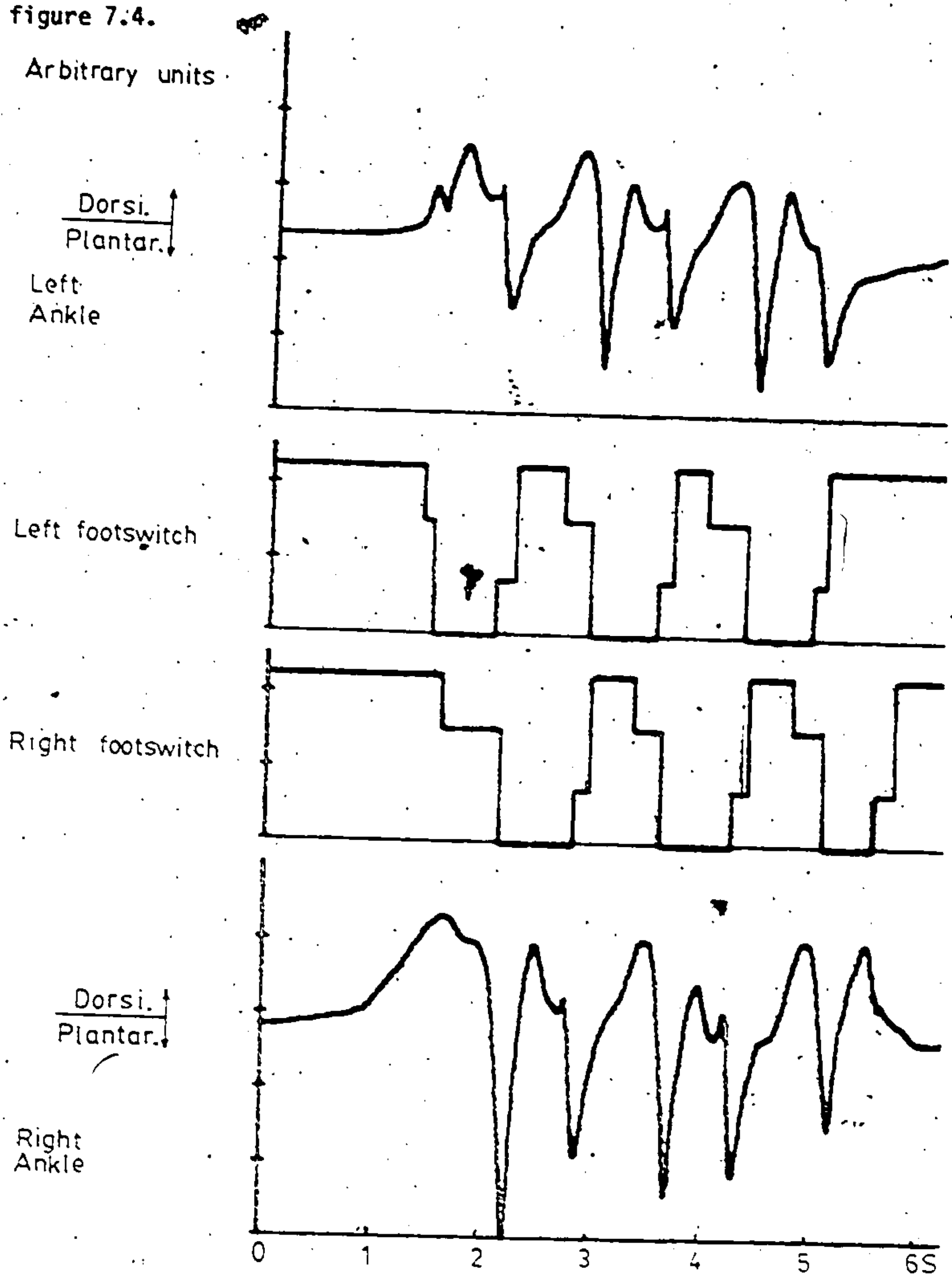


Figure 7.3 Footswitch and ankle angle records recorded from a normal subject.

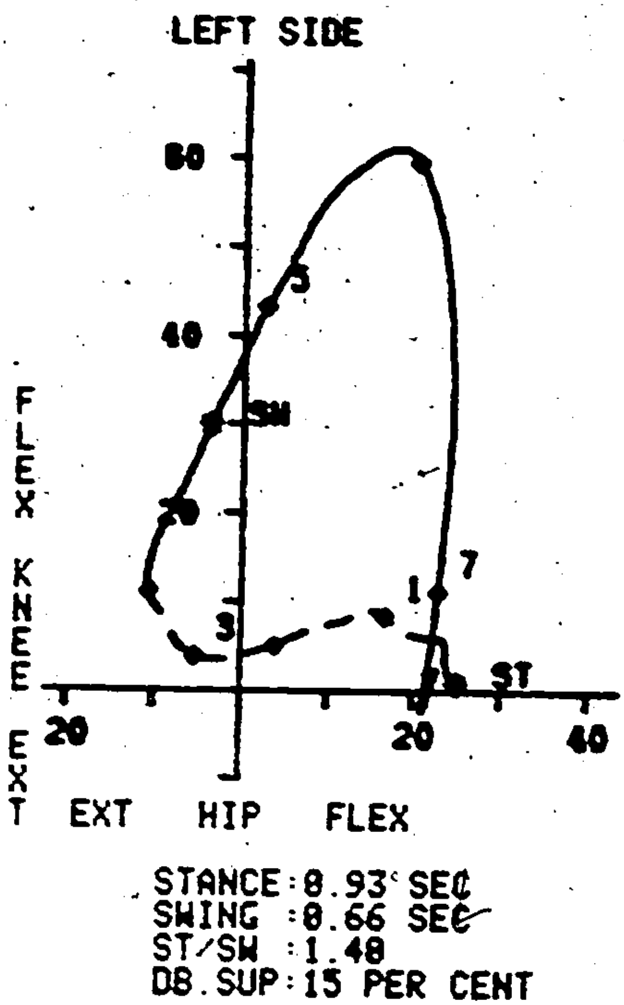


Figure 7.4 Knee angle versus hip angle from a normal subject.

The data are now presented in a form that contains both temporal and angular data allied simultaneously with two joints.

Consider figure 7.4, a plot of knee angle versus hip angle for normal gait. Temporal information is represented by:

(i) Stars indicating onset of the stance (ST) and swing (SW) phases.

(ii) Numerically sequenced time markers (0.2S apart in this case) whereby the times spent during different phases of the gait cycle can be visually observed.

(iii) Single support (dotted line) when only the left foot is in contact with the ground.

Angular information is presented in an easily digestible form in that:

(i) The overall ranges of motion are readily appreciated.

(ii) The shape of the loop is characteristic of various gaits. The effect of speed of walking or changes in the gait due to various pathologies result in characteristic changes in the shape of the loop.

Reading the display of figure 7.4 clockwise from heelstrike, the following information is obtained:

(i) Following heelstrike, a pronounced flexion of the knee occurs. The knee flexes so that the impact of the foot striking the ground is cushioned.

(ii) As the centre of gravity of the body moves from behind the leg to in front of it (weight-bearing portion), the knee angle remains almost constant as the hip extends.

(iii) After the angle-support phase has ended, the knee rapidly flexes as first the heel and then the toes lift off. The hip also flexes so that at toe-off, the whole leg is lifted off the ground.

(iv) During deceleration of the foot prior to heelstrike, the knee rapidly extends as the foot reaches forward, while the hip is maintained in a position of flexion.

Figure 7.5 is a plot of the ankle-joint angle versus knee angle. The various phases of the gait cycle as represented by this diagram are:

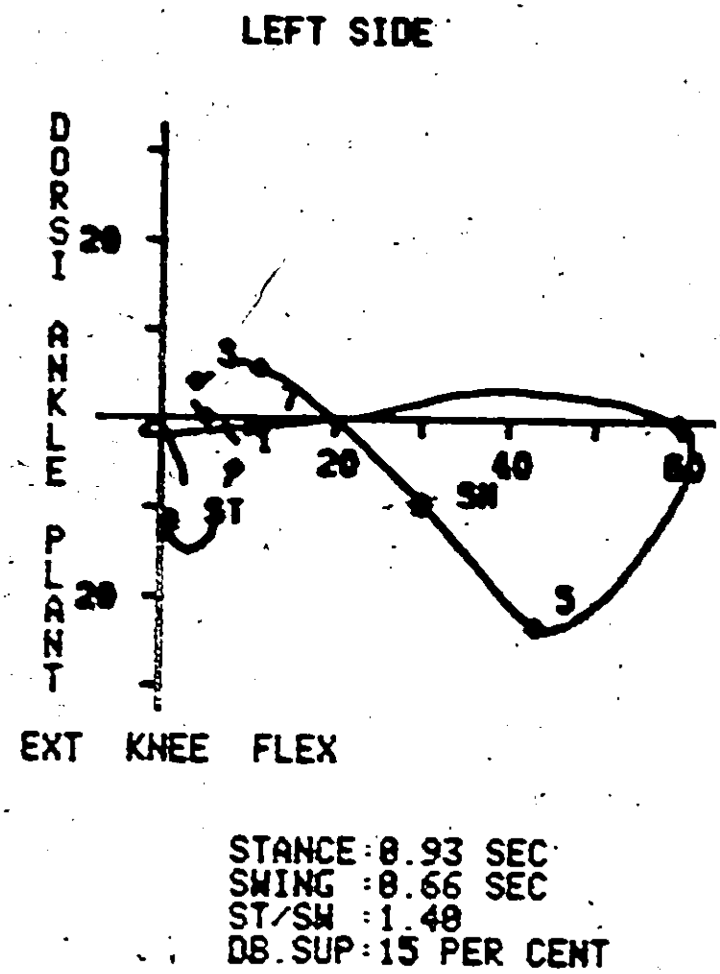


Figure 7.5 Ankle angle versus knee angle from a normal subject

(i) At heelstrike, the ankle is plantarflexing as the toes move downward to contact the ground.

(ii) During the single-support phase, dorsiflexion of the ankle occurs as the body swings over and forward, with the knee angle kept virtually constant.

(iii) Prior to, and just after swing, the ankle plantarflexes as the toes push off.

(iv) Dorsiflexion of the ankle is maintained as the foot is swung forward prior to heel strike. This ensures that the toes do not make contact with the ground during the swing phase.

7.2.4 Hemiplegic Gait

Variations in the gait of people having suffered a stroke can range from the extreme case where a person is non-ambulatory, to one which is not noticably different from normal gait.

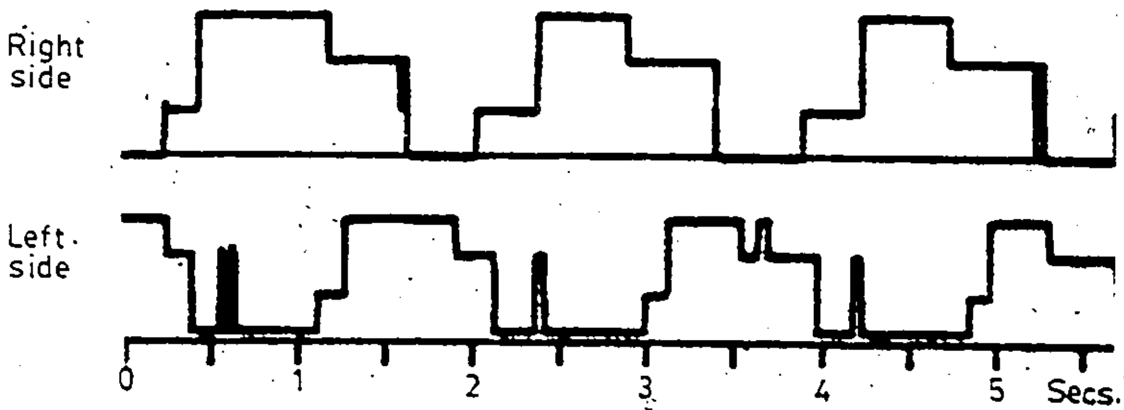


Figure 7.6 Footswitch data recorded from left hemiplegic subject

Figure 7.6 is a record of footswitch data obtained from a hemiplegic subject whose left side is affected. Abnormalities in the pattern are:

(i) A comparatively short swing phase on the right or uninvolved side due to the subject's inability to comfortably weight-bear on the affected side.

(ii) Instability at toe-off on the right side, again indicating the subject's reluctance to weight-bear on the affected side.

(iii) Toe-dragging on the left side due to footdrop.

(iv) Instability during weight-bearing on the affected side, the subject rocking from foot-flat to toe-only and back to foot-flat.

A further abnormality sometimes seen but not present in the records shown here, is the absence of heel-strike due to the inability of a subject to actively dorsiflect the foot.

7.3 Interactive Program to Implement Contralateral Control of Stimulation During Walking

An interactive program using available signals from the contralateral side to control stimulation of the involved side was written based on the information presented in section 7.2.

The following factors had to be accounted for:

(i) Asymmetry during initiation and termination of gait.

(ii) Recognition of different phases of the gait cycle.

(iii) Gait cycle phasic time differences between the two sides.

(iv) The existence of a DC voltage difference between the control and evoked signals.

(v) Adjustment of the stimulated side's movements to those of the control side over a number of footsteps.

(vi) Continuous processing of the control signal to reflect changes occurring in the gait.

In addition, footswitch and joint-angle data had to be recorded during stimulation for later evaluation.

A PDP 11/10 computer with 28K memory and an 8-channel A/D converter and 2-channel D/A converter were available. Footswitches placed on each shoe sole were used to supply gait timing information. One A/D channel was utilized to acquire the control signal, which was outputted via the D/A to the stimulator following processing. A second

A/D channel was used to acquire the evoked signal, the remaining 4 channels being used to acquire joint trajectory information.

7.3.1. Timing

Consider the two footswitch records depicted in figure 7.7, and Table 7.1.

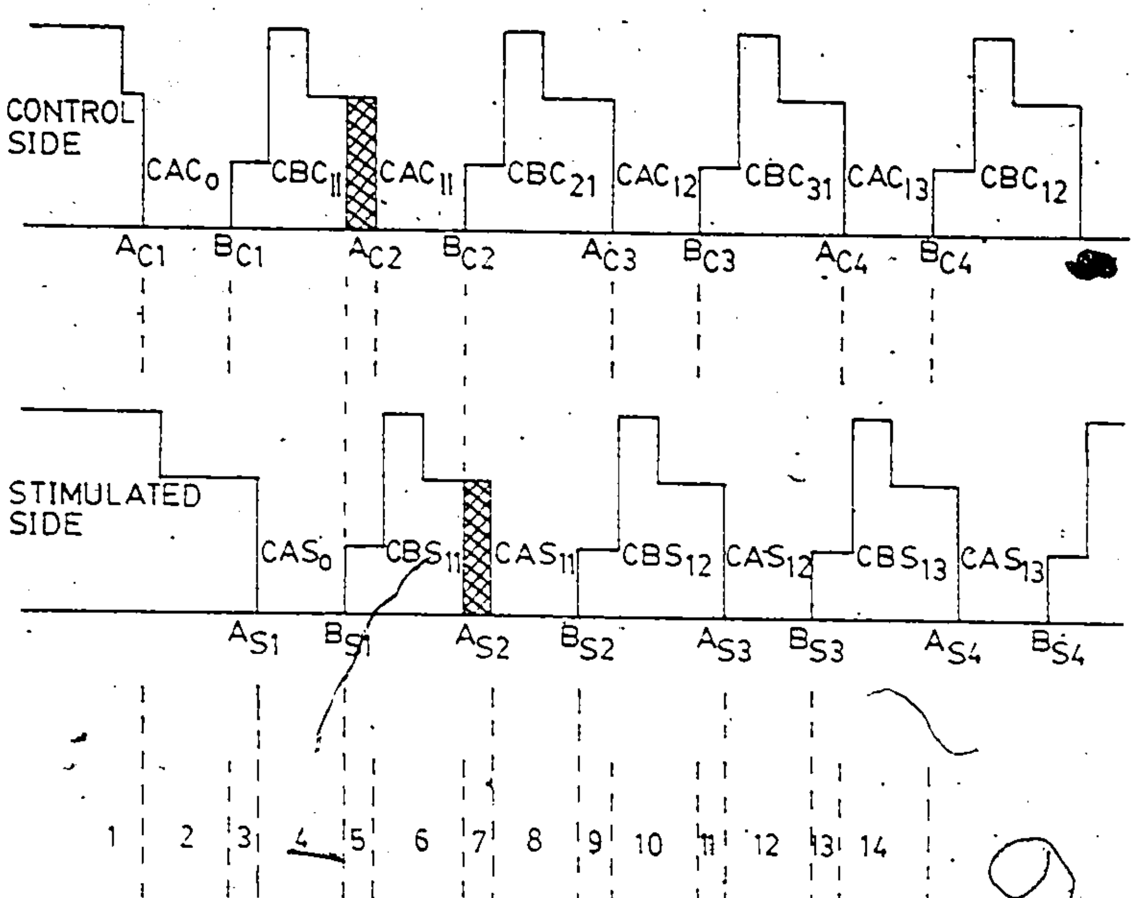


Figure 7.7. Schematic of normal footswitch patterns indicating swing and double-support phases

Each footstep can be divided into a swing phase, CA, and a stance phase, CB. The control signal recorded during time CAC_{11}, say, is used to control stimulation during time CAS_{11}. Two events must be accounted for:

(i) CAS_{11} is either longer or shorter in time than CAC_{11}. If CAS_{11}, say, is longer than CAC_{11} (the more likely case as can be seen from figure 7.6), then the last value in the record of the sampled control signal, CAC_{11}, is outputted until B_{S2} occurs. If the opposite occurs, then at time B_{S2}, data collected during the stance phase, CBC, is outputted to the stimulator.

(ii) To ensure that movements on the stimulated side converge towards those on the control side, the swing periods on each side, say, must be compared to each other (eg. CAC_{11} with CAS_0), and the control signal (CAC_{11}) then adjusted accordingly (and used to control stimulation during period CAS_{11}).

The swing periods on each side occur at distinctly different times. However, each stance phase on one side overlaps two separate stance phases on the other side. Thus CBC_{21} occurs simultaneously with parts of CBS_{11} and CBS_{12}. The approach adopted was to compare time CBC_{21} with time CBS_{12}, and then to use the modified signal to control stimulation during time CBS_{13}.

Adjustments to the control signals recorded during each swing or stance phase are accomplished as follows:

(i) The time period on the control side is compared to the corresponding time period on the affected side. This is achieved by subtracting the number of samples collected during phase CBS_{12}, say, from those collected during phase CBC_{21}, say.

(ii) If the difference in the number of samples collected during the two phases is zero, the control data is not modified. Thus data collected during phase CBC_{21} is used as is to control stimulation during time CBS_{13}, say.

(iii) If i more (or less) data points are contained in the control record (CBC_{21}) than in the record obtained from the affected side (CBS_{12}), and the number of footsteps selected over which the two sides are to converge is N, then every i/N^{th} point is dropped (repeated) from (in) the control record (CBC_{21}). The modified record (CBC_{21}) is then used to control stimulation during time CBS_{13}.

While the number of footsteps over which the timing values for the two sides are to converge is N, the above adjustment takes place each time the particular phase has occurred on both sides.

7.3.2. <u>Initiation and Termination of Gait</u>

The side on which the patient starts to walk does not affect execution of the program. If the affected side is first brought forward (as is usually the case since the hemiplegic tends in these circumstances to favour the unaffected side), the computer programming is such that data storage commences only after swing on the unaffected side has been completed (B_{C1} in figure 7.7.). Since feedback from the stimulated side is to be employed, the volitional signal from the side to be stimulated is fed to both input ports of the controller with zero delay between them. This ensures that no stimulation occurs until time A_{S2}.

The computer controls stimulation until a time, dependent upon the frequency of sampling chosen, has passed. At this point a bell in the CRT is rung, volitional information from the stimulated side is again fed to the control port of the controller with zero delay to prevent stimulation, and the stimulators are then switched off. Final termination of the program is effected by means of a hand-operated trigger.

To ensure that stimulation did not occur during the initial and final phases of the walk cycle as a result of a DC voltage difference between the control and feedback channels, the following procedure was followed:

(i) Just prior to the walk, the control and feedback channels were sampled with the subject in a standing position.

(ii) During execution of the control program, the recorded DC voltage difference was added to each sampled control data point.

7.3.3. Recognition of Gait Cycle Phases

The distinct voltage levels obtained from the footswitches promotes easy recognition of the different gait cycle phases. From section 7.2.4., however, abnormalities occur in the footswitch patterns of hemiplegic subjects. The control program must therefore be able to clearly recognize the start and end of each swing and stance phase in the presence of these abnormalities.

Consider the hypothetical footswitch patterns shown in figure

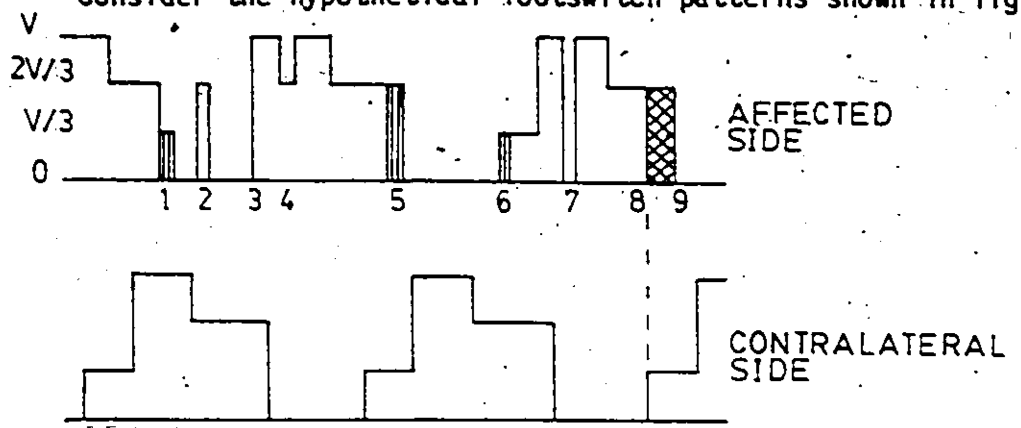


Figure 7.8. Hypothetical footswitch patterns illustrating potential abnormalities

7.8. Potential abnormalities are numbered 1 to 7. These are:

1) Noise due to contact-bounce of the switch. The program recognizes the first occurrence of a zero voltage output from the footswitch as the start of swing. Testing for the end of swing only occurs after a preselected period has elapsed. A period of 200mS was found to be satisfactory.
2) The occurrence of toe-only during the swing phase due to the subject's difficulty in maintaining dorsiflexion of the ankle. The program tested for the end of the swing phase by seeking voltage levels about the 1/3 level or above the 2/3 level (figure 7.8.) so that the occurrence of a voltage level between these two bands was ignored.
3) Absence of heelstrike. As in 2), either the presence of

heelstrike or foot-flat signified the end of swing and the start of stance.

4) Rocking from foot-flat to toe-only and back again to foot-flat. Toe-only was never tested for, thus this occurrence was ignored by the program.

5) Switch-bounce or instability of the subject at toe-off. As in 1), the first occurrence of a zero level was taken as the start of swing. Both 1) and 2) ensured that the noise had no effect on program execution.

6) Switch-bounce or instability of the subject at heelstrike. The first occurrence of a level indicating heelstrike was recognized as the start of stance. The program then switched to testing for the start of swing on the contralateral side thus ignoring any aberrations which might have occurred on the involved side.

7) As in 6), the program would have been testing for start or end of swing on the contralateral side.

The time periods 8 - 9, (see Table 7.1) indicated in figure 7.7. and 7.8. by hatched markings, occur during the period when the program is testing for the start of swing (double-support phase). If the footswitch output level drops to zero during this period, the program will assume that the swing phase has started. However, an examination of the footswitch patterns of 15 hemiplegic subjects revealed only the occurrence of instability, indicated in 5), during this time period.

7.3.4 Program Organization

Initial phases of the program consisted of:

(i) Defining the sampling frequency and hence the execution time of the program since the number of samples to be collected is fixed.

(ii) The total number of data channels to be sampled.

(iii) The number of electrogoniometer channels. At this point, a calibration procedure was undertaken whereby each electrogoniometer's output voltage levels at two preset joint angles were read into the computer. This permitted later transformation of the electrogoniometer voltage outputs into degrees of rotation (de Bruin, 1978).

(iv) Delays (section 7.3.3) were entered for each side before the end of which, testing for the end of swing on each side did not occur.

(v) The number of footsteps over which the two sides are to converge was entered.

(vi) The control and feedback channels were sampled (section 7.3.2) with the subject in a standing position in order that any DC voltage difference between the two channels could be compensated for.

The control program itself was written in ASSEMBLY language and was divided into fourteen separate service-routines. Table 7.1 is a summary of the data collected and outputted during each service-routine (refer to figure 7.7) where:

EGCA ≡ control data recorded or outputted during swing.

EGCB ≡ control data recorded or outputted during stance.

I3CHAN ≡ data recorded from transducer of side to be controlled.

A fifteenth subroutine ensures that at the termination of data collection, a bell in the CRT is rung and I3CHAN only is collected and outputted continuously to the controller until the stimulators have been switched off and the run terminated via the trigger.

The data is then written on a permanent memory medium (RK05 -disc). If another run is required, the program returns to step (iv) above.

7.4 Position Control of Stimulation During Locomotion: Trial on a Normal Subject

The program was first tested on a normal subject prior to applying it to a hemiplegic patient. Both EMG and position control of the ankle-joint position were performed.

Stimulating electrodes were applied to the dorsiflexor and plantarflexor muscle groups acting about the right ankle joint. Two ankle-goniometers placed about the joint were next used to calibrate the stimulator. The output of the electrogoniometer attached to the joint to be stimulated was then connected to I3CHAN, the channel to the computer designated for recording evoked data, and to the feedback input port of the stimulator. The output channel of the computer from the D/A was connected to the control input port of the stimulator. The stimulator was kept on a mobile cart so that during locomotion, the cart could be pushed alongside the subject. Care was taken to ensure that the control and feedback signals were passed through

SERVICE ROUTINE	IN (A/D)	OUT (D/A)	TIME PERIOD	COMMENTS
1	I3CHAN	I3CHAN	Upto A_{C1}	Input and output feedback channel, I3CHAN
2	I3CHAN	I3CHAN	A_{C1} - B_{C1}	Upto heelstrike on control side
3	EGCB1 (I3CHAN)	EGCB2 (I3CHAN)	B_{C1} - A_{S1}	I3CHAN only 1st time. EGCB2 after this
4	EGCB1 (I3CHAN)	EGCA1 (I3CHAN)	A_{S1} - B_{S1}	As in 3 except EGCA1 only after 1st time
5	EGCB1 (I3CHAN)	EGCB3 (I3CHAN)	B_{S1} - A_{C2}	1st control stance-phase. Data collected
6	EGCA1 (I3CHAN)	EGCB3 (I3CHAN)	A_{C2} - B_{C2}	1st control swing-phase. Data collected
7	EGCB2 (I3CHAN)	EGCB3 (I3CHAN)	B_{C2} - A_{S2}	
8	EGCB2	EGCA1	A_{S2} - B_{S2}	Stimulation begins at start of swing
9	EGCB2	EGCB1	B_{S2} - A_{C3}	Stimulation during stance phase
10	EGCA1	EGCB1	A_{C3} - B_{C3}	
11	EGCB3	EGCB1	B_{C3} - A_{S3}	
12	EGCB3	EGCA1	A_{S3} - B_{S3}	
13	EGCB3	EGCB2	B_{S3} - A_{C4}	
14	EGCA1	EGCB2	A_{C4} - B_{C4}	RETURN TO SERVICE ROUTINE No. 3

Table 7.1 Program data organization.

the computer with unity gain so that the off-line calibration of stimulus parameters remained valid. After calibration of the ankle, knee, and hip electrogoniometers on each side, footswitches were placed on the shoe soles. All data were fed to the computer via an umbilical cable. One end of the cable was attached to a control box fitted about the subject's waist, and into which the goniometers and footswitches were plugged. The other end of the multi-wire cable was plugged into the A/D inputs of the computer. A remote switch was used to trigger the start and end of data conversion.

A number of runs were performed, the first and last without stimulation for comparison purposes. During the runs with stimulation, the subject either walked normally or tried to simulate hemiplegic gait by not voluntarily dorsiflexing the ankle joint, and by circumducting the leg.

7.4.1 Position-Control Results

Figure 7.9 is a display of the footswitch patterns and joint angle variations during locomotion without stimulation. From this diagram it can be seen that the patterns occurring on one side of the body are repeated on the other side with a delay. This delay is proportional to the period of stepping.

Figure 7.10 is a display of the footswitch patterns and joint angle variations obtained from the subject walking normally but with stimulation applied to the right leg. In figure 7.9, the patterns of the ankle-joint angle variations on each side differ during the swing-phase, the right foot not dorsiflexing to the same degree as the left foot. This difference is due to the slight variability that exists between the left and right sides even in normals (see section 7.2). With stimulation (figure 7.10), this difference has been obviated.

Figure 7.11 was obtained during a simulation of hemiplegic gait accompanied by stimulation. The trajectories are more similar to those obtained from the unaffected side of a hemiplegic rather than from the affected side. Evidence of stimulation is present in the swing portion of the right ankle trajectory, with the foot seeming to oscillate slightly.

While no conclusions could be drawn from these experiments as to the final performance of the system when used to control hemiplegic gait, the system has been demonstrated to be feasible and the subject experienced no unpleasant sensations.

7.5 EMG Control of Stimulation During Locomotion: Trial on a Normal Subject

Stimulating electrodes were applied to the dorsiflexor muscle group on the right side, and recording electrodes were placed over each tibialis anterior muscle on both legs. Calibration was achieved by ensuring equal volitional and evoked joint angle and EMG maximum amplitudes. Since the evoked EMG amplitude is larger than the volitional one, this was compensated for by means of the input calibration control of the controller. As in section 7.4, all signal lines were fed to the computer, the evoked EMG channel, I3CHAN, also being fed to the feedback input port of the stimulator. Both volitional and evoked EMG channels were processed by means of

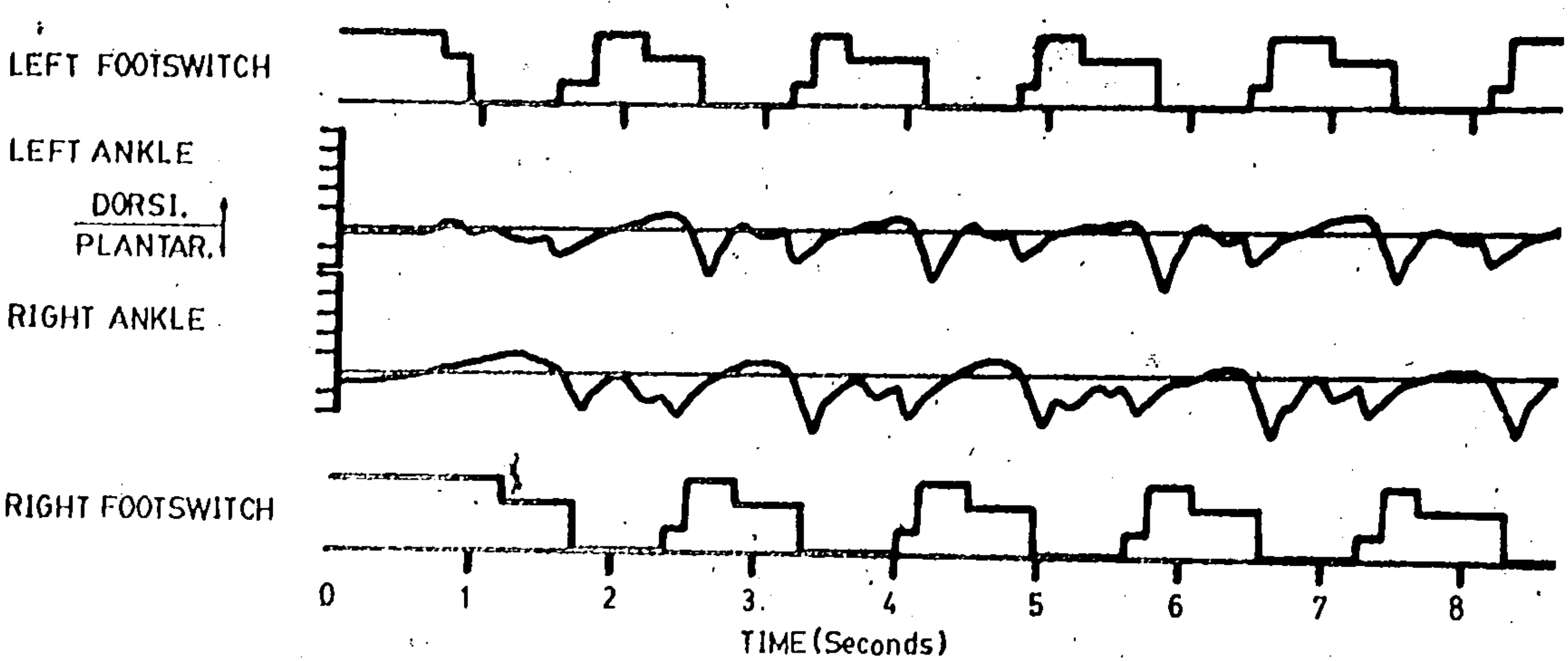


Figure 7.9 Footswitch patterns and joint angle variations. Normal subject

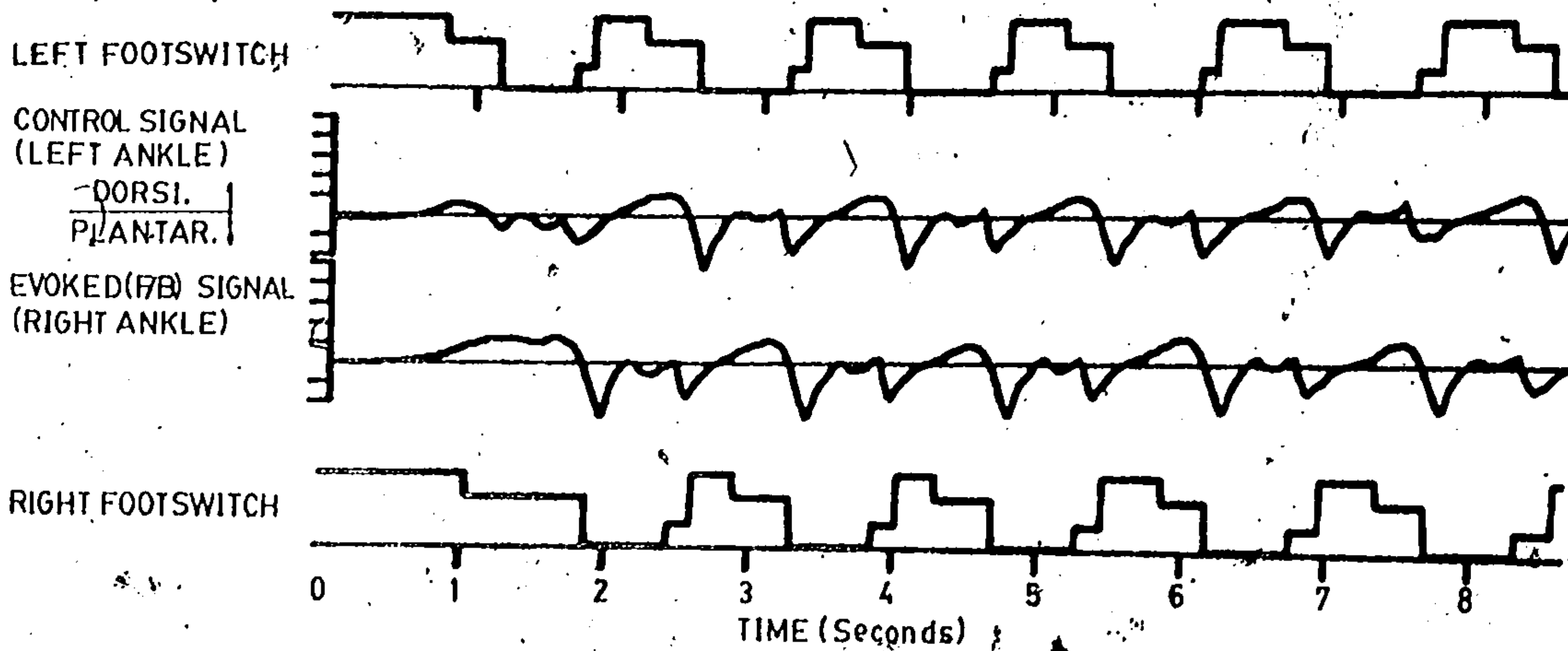


Figure 7.10 Footswitch and joint angle variations from normal subject with stimulation of right side

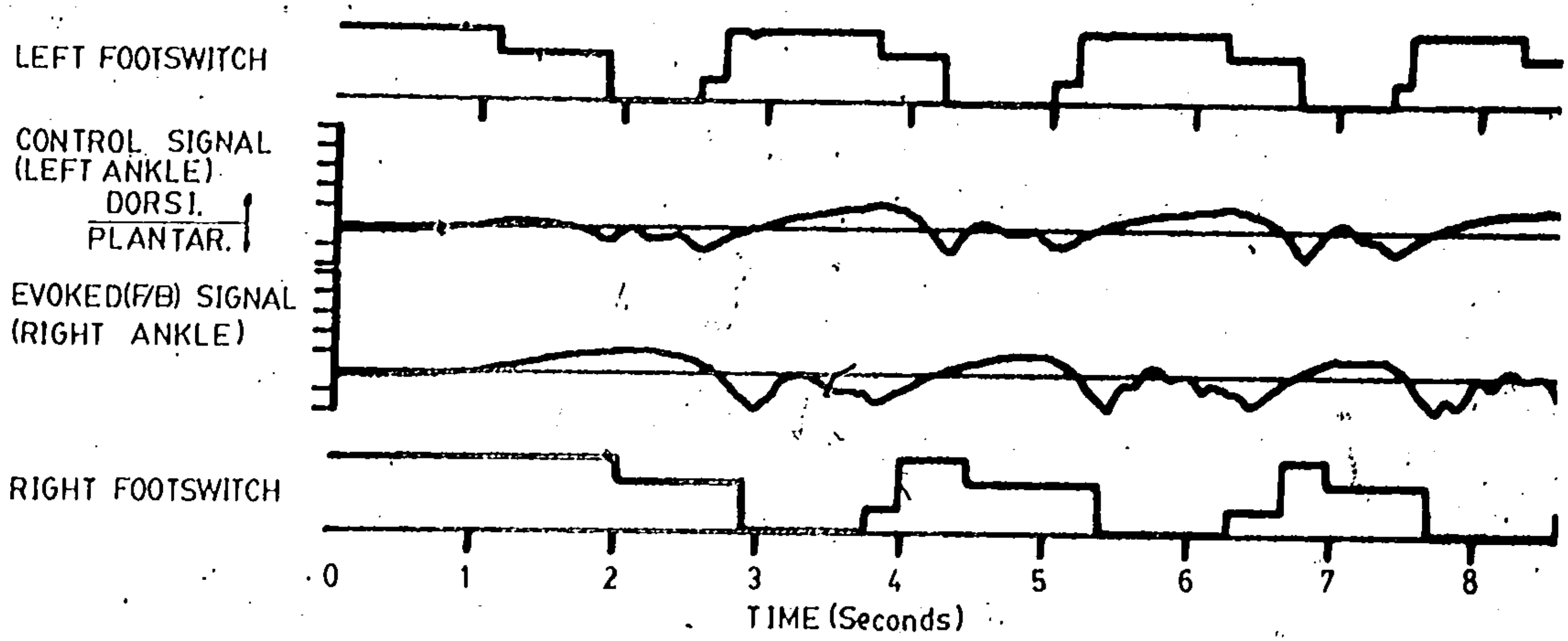


Figure 7.11 Gait patterns from normal subject simulating hemiplegic gait. With stimulation.

rectification and smoothing, the analogue averagers having RC time-constants of 72.6mS. Since only 8 channels of A/D converters were available, and four channels were required to record footswitch and EMG data, electrogoniometers were only attached about the ankle and knee joints.

7.5.1 EMG - Control Results

Figure 7.12 is a display of EMG and joint variations and footswitch patterns during locomotion without stimulation.

Figure 7.13 was obtained during stimulation. From the figure it can be seen that:

(i) Cross-talk from the evoked EMG record is present in the control EMG record due to the spread of the stimulus through the body. This suggests that the switches used to eliminate stimulus artifact in the evoked EMG signal (section 4.6) be employed in the volitional EMG recording circuitry. The occurrence of cross-talk here does not invalidate the results obtained in Chapter 5 since the volitional signal there was recorded prior to the performance of the stimulation experiments.

(ii) Because of cross-talk from the stimulated side appearing in the control signal, the system has become unstable as evidenced by the exponential growth of the evoked EMG. Dorsiflexion of the right ankle prior to heelstrike can also be seen to increase with time. However, it did not

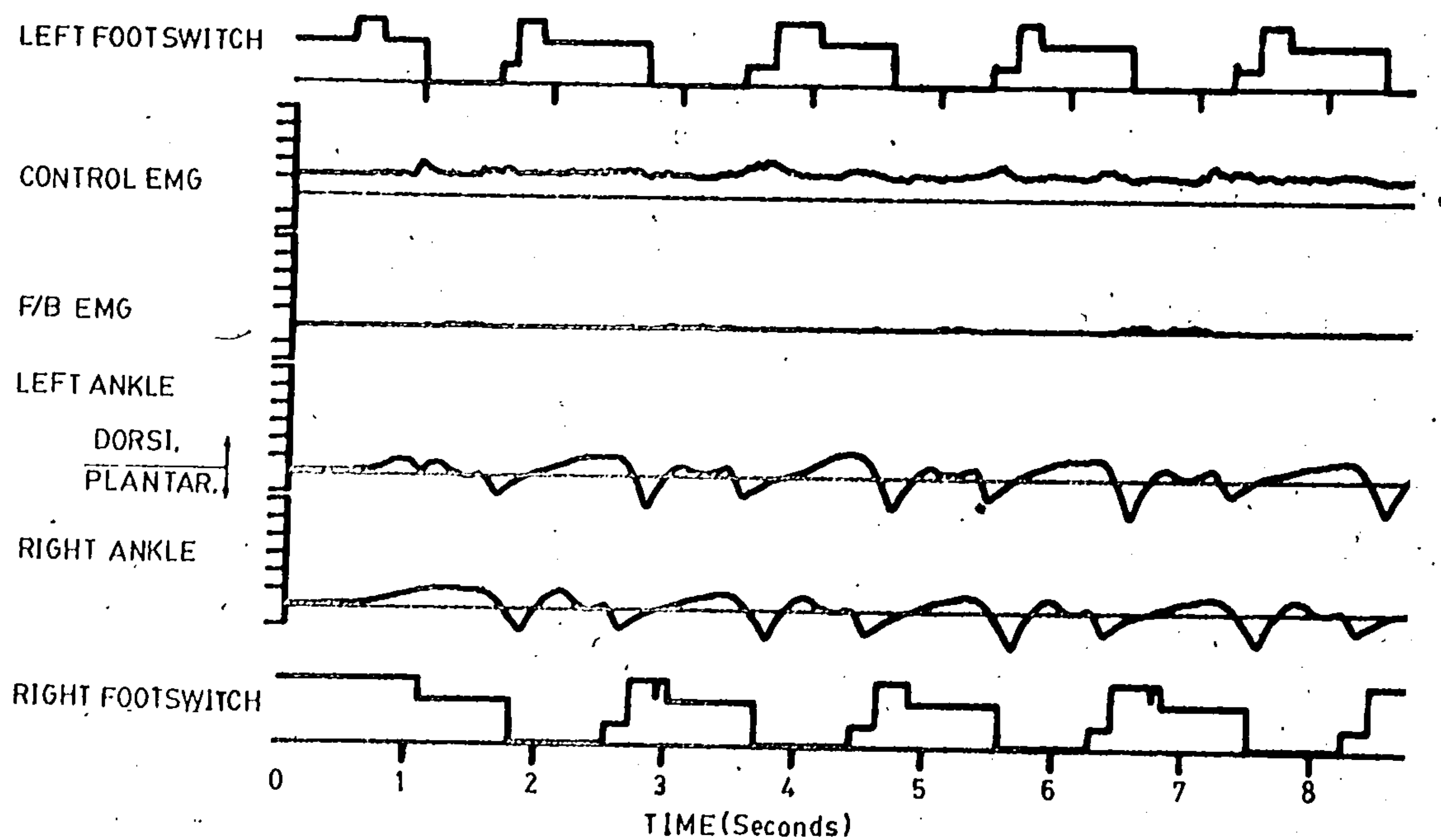


Figure 7.12 Footswitch patterns and EMG and ankle joint variations from normal subject

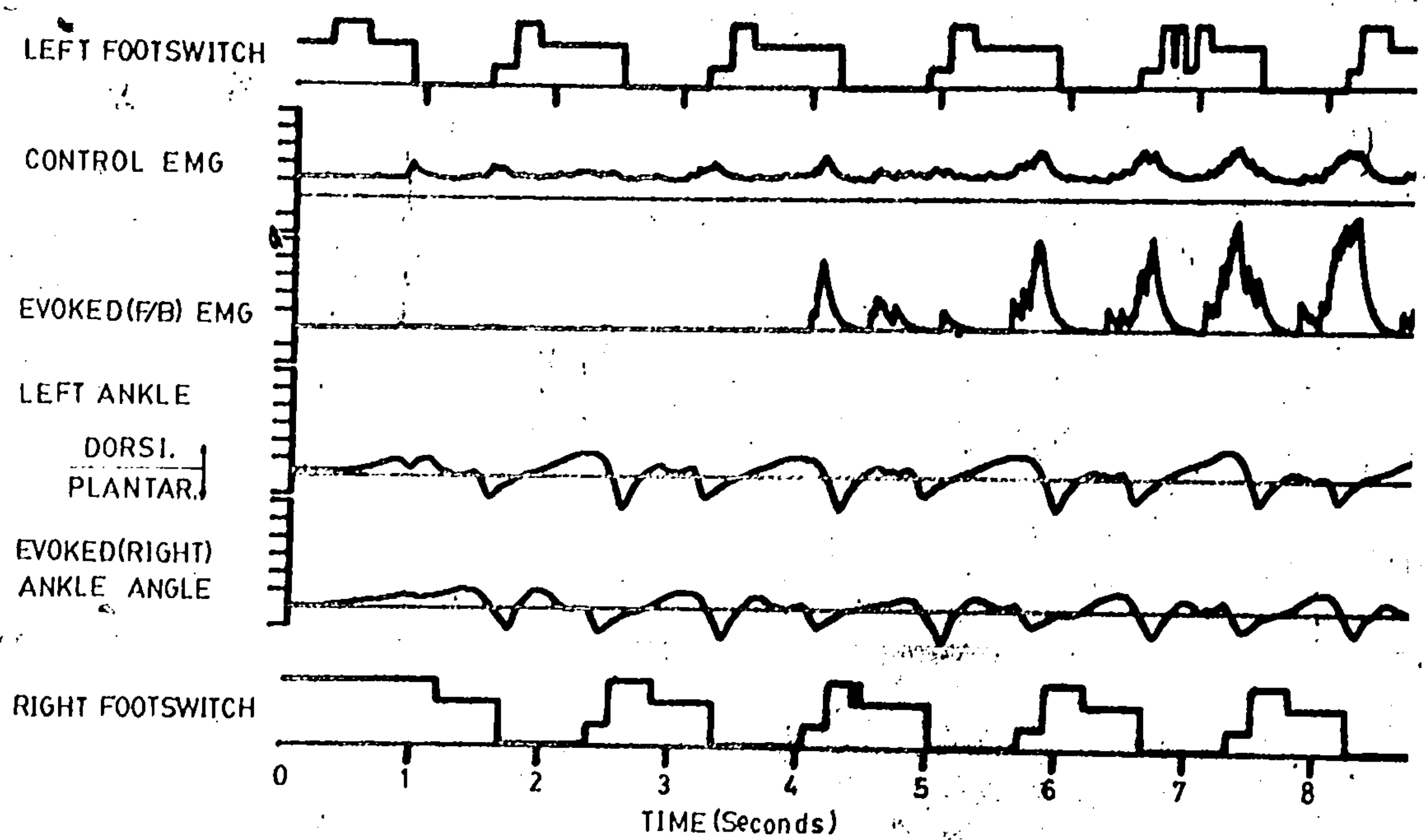


Figure 7.13 Footswitch patterns, EMG and ankle joint variations from normal subject. EMG controlled stimulation of right side.

become excessive during the period of the walk.

Thus further modification to the system is needed before contralateral EMG control can be effectively used to regulate joint position by means of electrical stimulation.

7.6 Contralateral Position Control of Hemiplegic Gait

The objectives of the study, attendant risks and discomfort, and expected benefits were detailed prior to obtaining the consent of the patient. The equipment to be used was shown and demonstrated. The patient was encouraged to ask questions at this stage and throughout the course of the study. The option of withdrawing from the study at any time was stressed both before and during the experiments. A copy of the consent form appears in Appendix C.

The author encountered difficulty in securing volunteer hemiplegic patient participation in the study. Of the several people approached, two agreed to participate but withdrew when confronted with the laboratory situation. Great care had to be taken over the manner in which the patient was approached, diplomacy being called for. Frequent encouragement by the medical staff was needed. One subject withdrew from the study because of personal family problems thus stressing the need to carefully assess the social situation of each patient beforehand.

The results presented in this section were obtained from a patient who agreed to participate in only two sessions. Another patient who had suffered damage to the sciatic nerve as a result of an operation to replace the hip joint, was found to be an unsuitable candidate in that only plantarflexion could be evoked upon stimulation, dorsiflexion function being totally absent.

7.6.1 Patient Case History

The subject was a 65 year old woman. In May, 1978, she developed a right hemiplegia due to an infarction in the left hemisphere. The patient had a severe expressive loss (verbal apraxia) but could understand conversation, follow commands, and express agreement or disagreement readily. The subject was alert and cooperative and in the opinion of the author, she has adjusted remarkably well to her situation.

At the time of the study, the patient was able to elicit volitional dorsiflexion and plantarflexion of her right ankle while seated, but had difficulty in actively dorsiflexing during the swing-phase of gait.

The physiotherapy discharge assessment indicated the following:

(i) Knee flexion during stance

(ii) Exaggerated knee flexion during swing

(iii) Poor balance and equilibrium

(iv) Cortical level absence of protection reflexes for both leg and arm

(v) Normal extension but poor flexor synergies

(vi) Normal proprioception

The Brunnstrom scale is invoked at the Chedoke Rehabilitation Centre to assess the stage of recovery of a patient following stroke (Brunnstrom, 1970). Six stages are defined. For the lower limb these are:

Stage 1 - Flaccidity

2 - Minimal voluntary movements of the lower limb

3 - Hip-knee-ankle flexion in sitting and standing

4 - Voluntary dorsiflexion of the ankle without lifting the foot off the floor.

5 - Standing, isolated nonweight-bearing knee flexion, hip extension; standing, isolated dorsiflexion of the ankle, knee extension, heel forward.

6 - Reciprocal action.

The patient was classified as a stage 4 on discharge meaning that spasticity had begun to decrease and she could perform some movement combinations that deviated from the basic limb synergies. The patient was discharged with a shoehorn brace to prevent foot-drop.

7.6.2 Experimental Procedure

During the first session, electrodes were applied to the right leg to effect dorsiflexion. An attempt to place a stimulating electrode on the plantarflexors failed due to pain resulting from stimulation. The second session took place 5 days later. The procedure outlined in section 7.4 was followed except that the PI-filter was not employed since only the dorsiflexors were stimulated.

The first and last of 4 runs were performed without the application of stimulation. During all 4 runs, the patient used a cane to assist her in walking. The session lasted for 2.5 hours, the most time-consuming and fatiguing part being the calibration of the electrogoniometers. The patient was often rested during the session but was visibly fatigued at its conclusion.

7.6.3 Description of Gait without Stimulation

Figure 7.14 and 7.15 are angle-angle diagrams obtained during locomotion of the hemiplegic subject prior to stimulation. When compared with the angle-angle diagrams of figures 7.4 and 7.5 obtained from a normal subject, the following abnormalities are noted:

(i) The knees on both sides were constantly in flexed positions.

(ii) At heelstrike on the affected (right) side, the knee buckled.

(iii) During single-support on the left side (dashed line), the knee flexed instead of extending thus illustrating the subject's poor balance and equilibrium (see section 7.6.1). This phenomenon occurred to a greater extent on the affect side.

(iv) During swing on the right (affected) side, the knee extended as the hip flexed, the lower leg swinging in a "pendulum" fashion. This movement is a result of the subject being unable to actively flex the hip, tilting of the pelvis being used to achieve t his

(v) The range of the knee on the affected side was half that of the left side.

(vi) At heelstrike, the ankle passively dorsiflexed due to an abnormally short heel-only phase. During single-support, the left ankle then actively plantarflexed as the knee extended to support body weight. This did not occur on the right side since, most probably body weight was being supported by the cane.

(vii) A grotesque pattern evolved on the affected side during the swing phase (figure 7.14). Instead of the ankle plantarflexing at toe-off and then dorsiflexing as the knee was first flexed and then extended in order to swing the leg forward (figure 7.5), the knee extended while the ankle joint remained fixed (ie. no push-off). The ankle then passively plantarflexed as the knee angle remained constant, the forward swing of the leg thus resulting from action at the hip only.

The extremely assymmetrical gait patterns observed between the left and right sides of this patient suggest that a stage 4 hemiplegic is not a suitable candidate for contralaterally controlled FES. However, no other hemiplegic subjects were available to the author.

7.6.4 Gait with Stimulation

The number of footsteps over which convergence with stimulation between the left and right sides was to occur were chosen as 3 and 5 for the two runs performed. No discernible differences were noted between the two runs.

Figure 7.16 and 7.17 are typical angle-angle diagrams obtained from the FES-assisted locomotion. The following effects of FES on the gait are noted:

(i) The right knee was more stable at heelstrike and actively

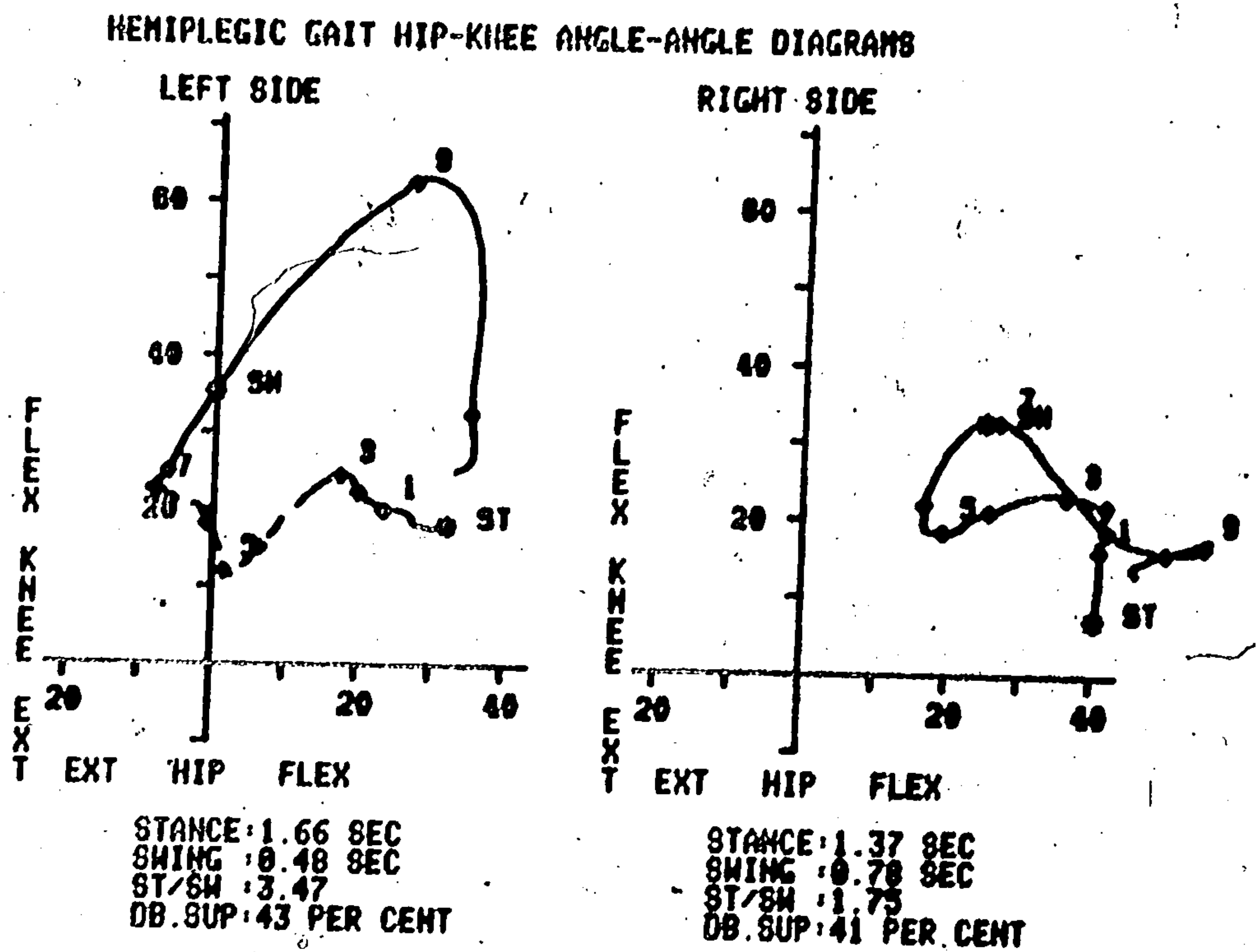


Figure 7.14 Knee versus hip angles for patient D.H. (right hemiplegic)

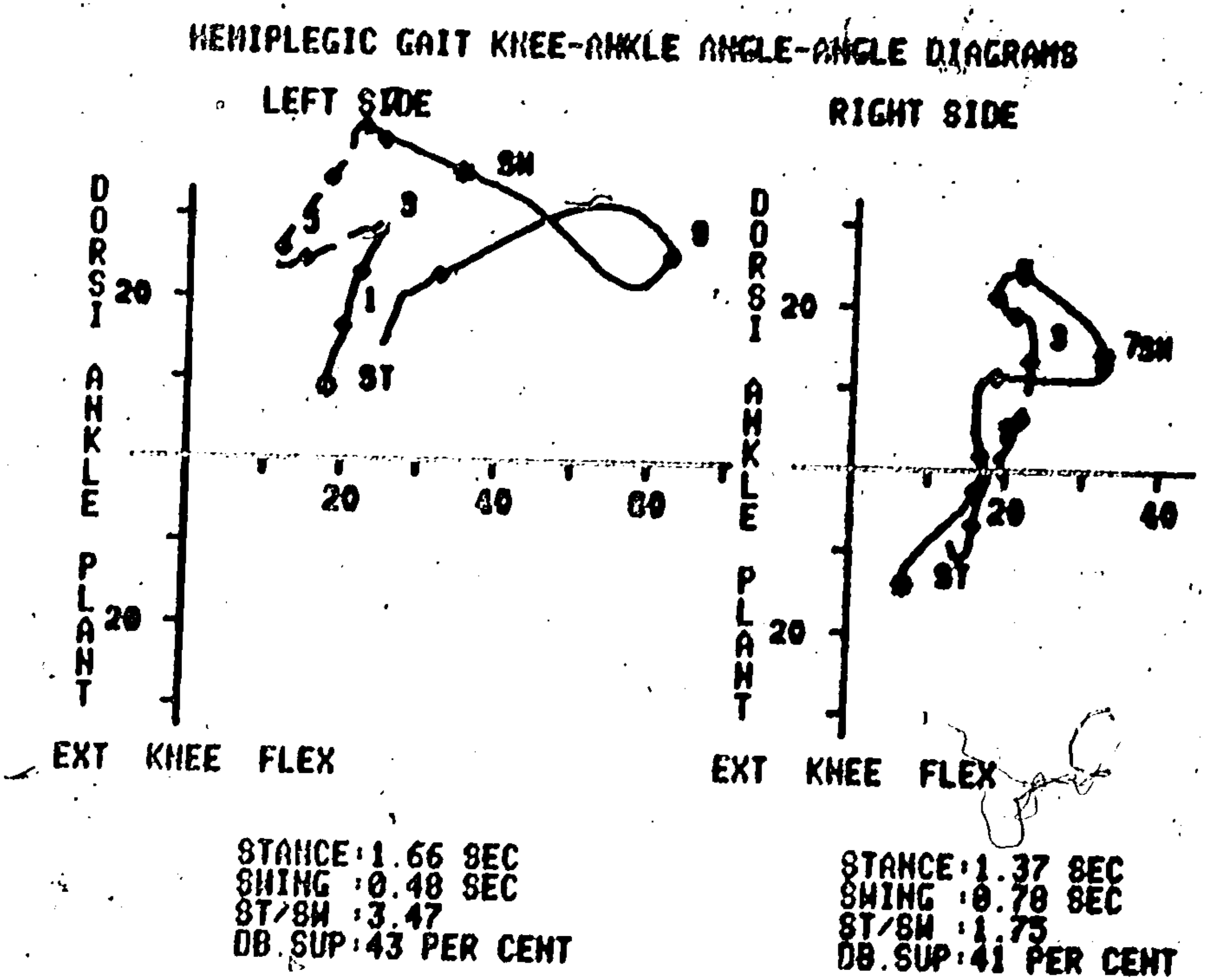


Figure 7.15 Ankle versus knee angles for patient D.H. (right hemiplegic)

maintained its position during hip extension.

(ii) Knee flexion on the affected side extended beyond the start of the swing phase.

(iii) The ankle plantarflexed at a slower rate prior to heelstrike.

(iv) The patient indicated that stimulus intensity increased to uncomfortable levels as each walk progressed thus suggesting that divergence between the joint angle movements of the left and right sides was occurring.

7.6.5 Discussion

The results presented indicate that FES had little effect on improving the gait of the subject. An experimental restriction was the limited storage capability of the computer, each run being limited to 10 seconds in length at a sampling rate of 200 Hz. Therefore, only the initial effects of contralaterally controlled stimulation could be observed. The patient also exhibited knee instability in addition to the absence of active dorsiflexion. Thus multichannel stimulation to control both knee and ankle positions is indicated for this particular patient.

While divergence between the control and stimulated sides did not occur when joint-position control was applied to a normal subject, the possibility of divergence occurring for a particular group of hemiplegic patients exists. Further experimentation on a large group of hemiplegic subjects is therefore required to delineate that group of patients which can best benefit from contralaterally controlled gait

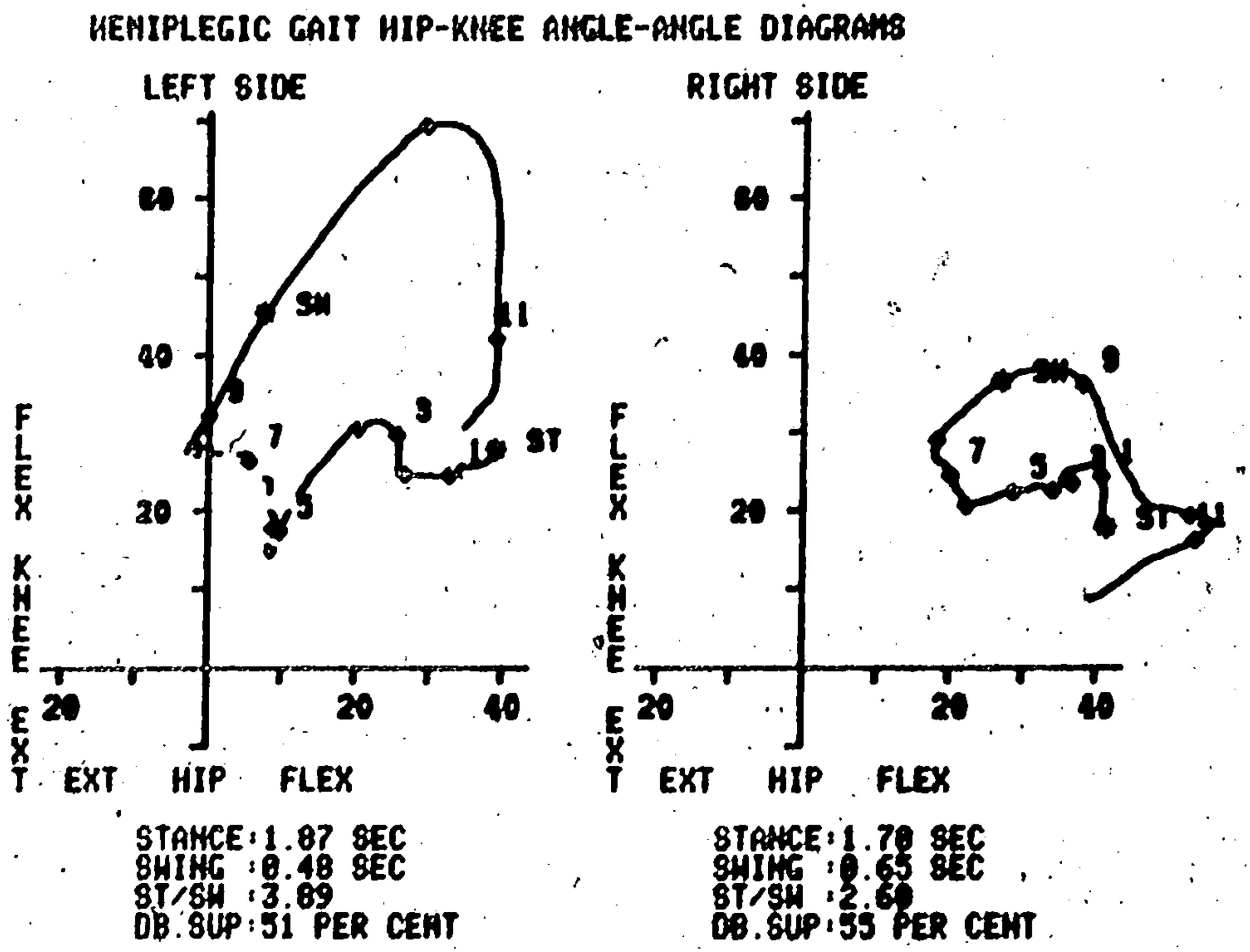


Figure 7.16 Knee versus hip angles for patient D.H. with stimulation of right side.

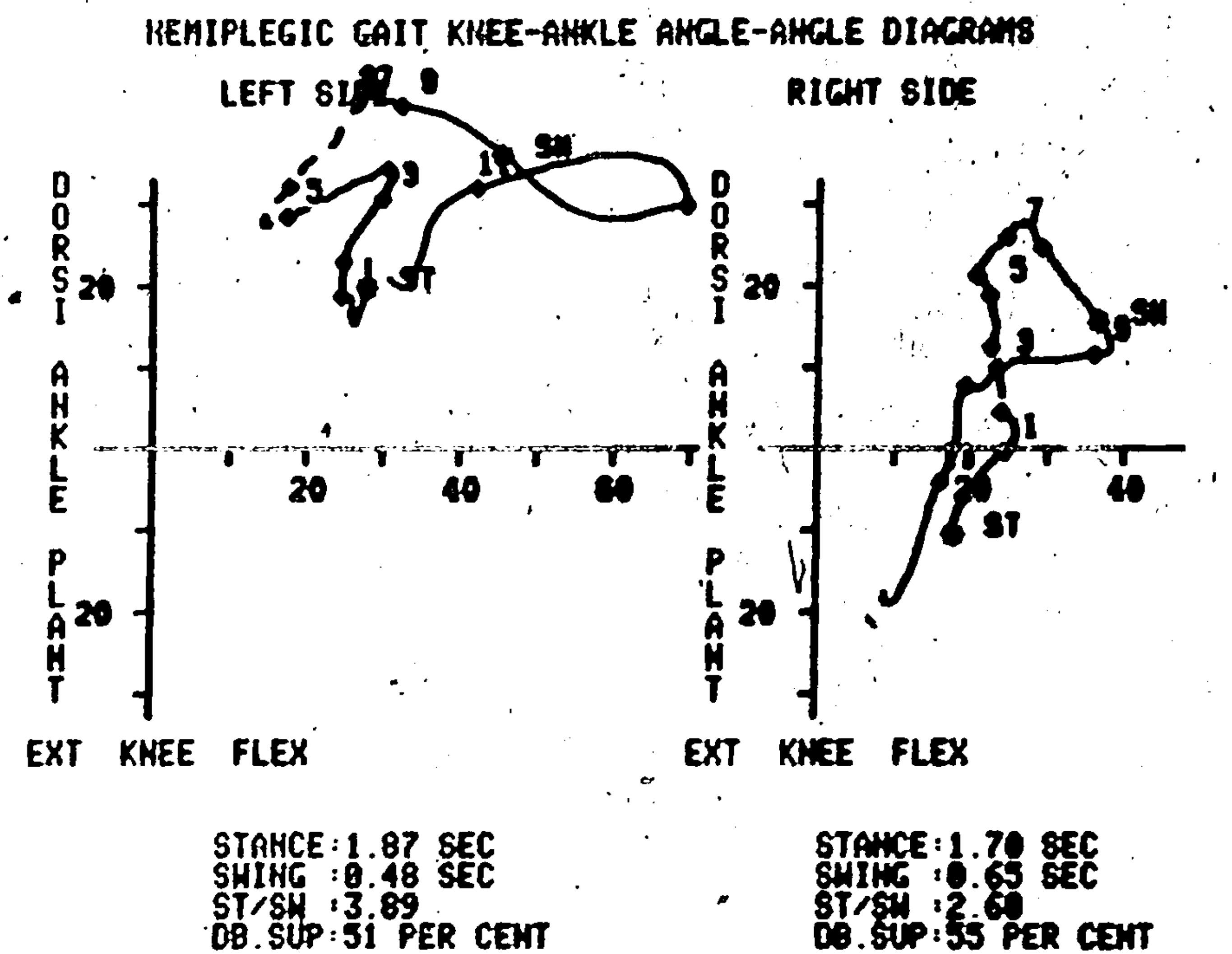


Figure 7.17 Ankle versus knee angles for patient D.H. with stimulation of right side.

by means of electrostimulation. These experiments must ascertain whether divergence between the control and stimulated sides was as a result of knee instability, and thus, whether the use of multichannel stimulation will increase the number of patients who can benefit from contralateral control of electrostimulation.

7.7 Conclusions

The concept of contralateral control for evoking dorsiflexion of the ankle-joint of hemiplegics during locomotion has been presented. During the action of walking in a straight line, signals generated by the walking action itself are potentially available from the unaffected side to control stimulation intensity, and hence movements, of the affected side. A requirement is that a delay proportional to the period of stepping be imposed between recording of the control signals and activation of the stimulators.

Two potential control signal forms have been investigated: the electromyographic activity of pertinent muscles; and joint-angle variations. A mini-computer was utilized in enabling repetition of experiments, and to analyze results. A method has been proposed and demonstrated whereby the effects of different control strategies can be assessed. The investigations into the efficacy of EMG and joint position as controllers of joint position have led to:

(i) The development of a method whereby evoked EMG can be recorded by eliminating stimulus artifact.

(ii) A quantitative means for comparing the effects of a digital and analogue filter on a particular signal.

(iii) The design of a proportional-integral filter to eliminate static errors between the input and output signals of the control system, and to increase the stability of the system when feedback is incorporated.

(iv) The determination of practical time-constants of averaging for the processing of control and evoked EMG signals to be used in an FES-based feedback control system.

(v) The demonstration that joint position information is contained in the EMG records obtained from the prime movers during particular movements.

A stimulator was constructed and computer programs were developed to implement contralateral control of foot-drop. Experiments on a normal subject demonstrated the feasibility of contralateral FES control. Conclusive results as to the effectiveness of the control system when applied to one hemiplegic subject are lacking and indicate the need for further experimentation on a larger hemiplegic population.

7.8 Future Possibilities

The 4-channel stimulator constructed by the author has the capability of controlling the position of two joints. A natural extension of the work presented in this thesis is the employment of multichannel contralateral position control of more than one joint. The system developed by the author also needs to be portable so that a microprocessor dedicated to controlling the system must be developed. With the current progress in the field of microelectronics, the author envisages the development of completely implantable, self-contained control systems. These systems will derive their power from electrochemical sources within the host.

Possibilities exist whereby prerecorded sequences of activation derived from normal subjects can be used in conjunction with the microprocessor-based system to control paraplegic gait. The problem of bracing these patients may be solved by implementing a hybrid system incorporating both mechanical and electrostimulation forms of bracing.

The method of "one-to-one" contralateral control has made available the potential to design systems with the ability to execute a multitude of different movements, and to limit the conscious effort required to operate such systems to the selection of a particular movement or sequence of movements, and to its initiation and termination.

APPENDIX A

THRESHOLD CHARGE

Consider the approximate equivalent circuit of a section of a fibre(Ruch and Fulton, 1961, p. 49) as shown in figure A1 where I is the stimulating current and

E_s = steady potential

C_m = membrane capacitance

r_m = membrane resistance

r_i = axoplasm resistance

The requirement for threshold stimulation is that E_m, the membrane potential, be depolarised a fixed amount $E_m = E_m - E_t$. This can be described by the following equation:

$$E_m = I_s\, r_m(1 - e^{-t/T})$$

where I_s = portion of I flowing through r_m at the site of the stimulating electrode.

At $E_m = E_{TH}$, the fibre fires and

$$I_s = \frac{\Delta E_{TH}}{r_m(1 - e^{-t/T})}$$

I_s becomes a minimum as $t \to \infty$

Therefore

$$I_m = \frac{\Delta E_{TH}}{r_m}$$

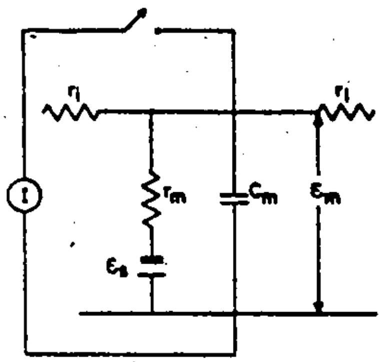


Figure A1 Approximate equivalent circuit of uniform fibre section (taken from Ruch and Fulton (1961))

For $t << T$

$$I_s = I_m T_m / t$$

or $\quad I_s t = I_m T_m$ which is the charge required to fire the fibre

Time constant

$$T_m = C_m r_m$$

$$I_s t = I_m\, T_m = \frac{\Delta E_{TH}\, T_m}{r_m}$$

$$= \Delta E_{TH}\, C_m$$

Therefore the charge required to excite the fibre is proportional to the membrane capacitance.

APPENDIX B

CURVILINEAR REGRESSION AND WINDOW LENGTH CONFIDENCE LIMITS

Consider a second-order polynomial

$$y = b_o + b_1 x + b_2 x^2 + \hat{e}$$

where $\hat{e}$ devotes a random-error term.

$$\hat{e} = y - (b_o + b_1 x + b_2 x^2)$$

Given n pairs of values (x_i, y_i)

$$\hat{e}_i = y_i - (b_o + b_1 x_i + b_2 x_i^2)$$

According to the least-squares criterion, b_o, b_1 and b_2 should be selected so that the sum of squared errors is a minimum. ie.

$$\text{minimize} \sum_{i=1}^{n} e_i^2 = \sum_{i=1}^{n} [y_i - (b_o + b_1 x_i + b_2 x_i^2)]^2 = S_e^2$$

For repeated experiments at each x_i, the sample variance S_i^2 can be used to weight the x_i's so that the influence of those x_i's whose variances are smaller, is larger than the influence of those x_i's whose variances are larger, when determining the regression curve.

The expression for the sum of squared errors becomes:

$$S_e^2 = \sum_{i=1}^{n} \frac{[y_i - (b_o + b_1 x_i + b_2 x_i^2)]^2}{S_i^2}$$

Differentiating the expression for S_e^2 with respect to b_o, b_1 and b_2 and setting the results equal to zero will yield three equations that can be solved to obtain the regression coefficients.

Using matrix notation, the equations become

$$y = Xb + \hat{e}$$

Since $\hat{e}^t \hat{e} = [\hat{e}_1 \ \hat{e}_2 \ldots\ldots \hat{e}_n] \begin{bmatrix} \hat{e}_1 \\ \hat{e}_2 \\ \vdots \\ \hat{e}_n \end{bmatrix} = \sum_{i=1}^{n} \hat{e}_i^2$,

the expression to be minimized becomes:

$$(y - Xb)^t (y - Xb)$$

and following differentiation, the least-squares regression coefficients are:

$$b = (X^tX)^{-1} X^ty$$

Now

$$X^tX = \begin{bmatrix} \Sigma 1/S_i^2 & \Sigma x_i/S_i^2 & \Sigma x_i^2/S_i^2 \\ \Sigma x_i/S_i^2 & \Sigma x_i^2/S_i^2 & \Sigma x_i^3/S_i^2 \\ \Sigma x_i^2/S_i^2 & \Sigma x_i^3/S_i^2 & \Sigma x_i^4/S_i^2 \end{bmatrix}$$

so that its inverse, $(X^tX)^{-1}$, can be found.

$$X^t y = \begin{bmatrix} \Sigma y_i/S_i^2 \\ \Sigma x_i y_i/S_i^2 \\ \Sigma x_i^2 y_i/S_i^2 \end{bmatrix} \qquad y = \begin{bmatrix} y_1 \\ y_2 \\ y_3 \end{bmatrix}$$

The error variance is given by:

$$S_e^2 = \frac{y^t y - b^t x^t y}{n}$$

If the error terms are assumed to be normally distributed with zero mean and variance σ_e^2 then

$$t = \frac{b_i - \beta_i}{S_e \sqrt{n/(n-k)} \sqrt{a_{ii}}}$$

is a value drawn from a t-distribution with n-k degrees of freedom where:

n = number of equations

k = number of independent variables

b_i = the estimate of β_i

a_{ii} = element of the i^{th} row and column of matrix $(X^tX)^{-1}$.

$$a_{11} = \frac{\Sigma x_i^2/S_i^2 \Sigma x_i^4/S_i^2 - (\Sigma x_i^3/S_i^2)^2}{|X^tX|}$$

$$a_{22} = \frac{\Sigma 1/S_i^2 \Sigma x_i^4/S_i^2 - (\Sigma x_i^2/S_i^2)^2}{|X^tX|}$$

$$a_{33} = \frac{\Sigma 1/S_i^2 \Sigma x_i^2/S_i^2 - (\Sigma x_i/S_i^2)^2}{|X^tX|}$$

The $100(1-\alpha)\%$ confidence interval for β_i is

$$b_i \pm aS_e\sqrt{a_{ii}}\sqrt{\frac{n}{n-k}} \quad \text{where}$$

$a = (1 - \alpha/2)$ fractile of the t distribution with n-k degrees of freedom.

Having calculated percentage confidence intervals for the b_i's, the confidence limits for the window length of processing producing the minimum variance ratio is calculated as follows:

Let the estimated curve be:

$$y = b_o + b_1x + b_2x^2$$

The value of x for which y is a minimum is:

$$\frac{dy}{dx} = b_1 + 2b_2x = 0$$

or $$x_{min} = \frac{-b_1}{2b_2}$$

$$\Delta x = \pm\sqrt{\left(\frac{\partial x_{min}}{\partial b_1}\cdot\Delta b_1\right)^2 + \left(\frac{\partial x_{min}}{\partial b_2}\cdot\Delta b_2\right)^2 + \frac{\partial x_{min}}{\partial b_1}\frac{\partial x_{min}}{\partial b_2}\Delta(b_1b_2)}$$

$$= \pm\frac{1}{2b_2}\sqrt{(\Delta b_1)^2 + \left(\frac{b_1\Delta b_2}{b_2}\right)^2 + \left(\frac{-b_1}{b_2}S_e^2\, a_{23}\frac{n}{n-k}\right)}$$

so that the confidence limits of the window length are given by:

$$x_{min} \pm \frac{1}{2b_2}\sqrt{(\Delta b_1)^2 + \left(\frac{b_1 \Delta b_2}{b_2}\right)^2 + \left(\frac{-b_1}{b_2}\, S_e^2\, a_{23}\, \frac{n}{n-k}\right)}$$

APPENDIX C

CONSENT FORM:

The objective of this study is to evaluate a device being developed for the purpose of replacing lost function to paralyzed muscle.

If you agree to participate in the study; you will have electrodes taped to your leg which will be used to stimulate the nerves which cause the muscles in your leg to pull the foot up or down. We also attach small instruments to your limb to enable the measurement of ankle angle during stimulation. On occasion we will measure the electrical currents generated by your leg muscles during stimulation. This will involve sticking surface electrodes on your skin.

ATTENDANT RISKS AND DISCOMFORTS:

Practically speaking there are no real risks associated with your participation in this study. The stimulators are isolated from all power lines.

A source of minor discomfort exists to which you may be subjected. The stimulating electrodes, if incorrectly placed, will produce an electric sensation on the skin. You will find that when the electrodes are properly placed, the sensation is not at all irritating.

EXPECTED BENEFITS:

The device is still in an early stage of development and it is unlikely that you will personally benefit from it at this time.

OTHERS MATTERS:

If you have any questions about any part of this study you are encouraged to ask them before giving your consent. However, you are free to ask questions about the study at any time during it course.

If you wish to withdraw from the study you are free to do so at any time without any fear of prejudice regarding other and future treatment.

CONSENT FORM

I have had explained to me the nature of the study, had all questions thus far satisfactorily answered and hereby consent to participate in it.

Patient's Signature

Name in Block Letters

Witnessed by

Date

I have explained the details of the study.

Investigator

Witnessed by

Date

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