HUMAN LOCOMOTION: TECHNIQUES FOR PROCESSING AND ANALYSIS OF EMG DATA

by

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ABSTRACT

Several techniques used by researchers in the area of human locomotion to process and analyse normal and pathological gait electromyographs (EMG) are discussed. Basic elements of neuro-muscular organization are described.

The thesis reports original work in several topics. The spectral analysis of dynamic EMG acquired during the locomotion of a normal subject was done to confirm the selected sampling frequency, and to determine a suitable low pass filter cutoff for smoothing EMG prior to data analysis.

Results of using two filters for smoothing EMG, a second order Butterworth low pass filter, and a mid-point moving window average filter are compared.

The cross correlation function is used in analysing EMG, since EMG signals are random. The results of cross correlation are compared with clinical observations in assessing the state of a patient following a stroke. Results for five normal and fourteen hemiplegic subjects are reported.

The conclusion is that cross correlation analysis quantifies the state of the patient and assesses post stroke recovery according to the neurological picture of central nervous system control.

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Glossary

EMG	electromyogram, electromyographic
TO	toe off
HO	heel off
FF	foot flat
HS	heel strike
FSL	left footswitch record
FSR	right footswitch record
Q	quadriceps muscle EMG record
H	hamstrings muscle EMG record
TA	tibialis anterior muscle EMG record
GA	gastrocnemius muscle EMG record
STRIDE	a walk cycle, heel strike to heel strike of the same leg
STEP	half a stride, heel strike to toe off or toe off to heel strike
CADENCE	steps/sec
SPEED OF WALK	metres/sec
BW	bandwidth
MF	median frequency

CHAPTER 1

INTRODUCTION

In the pursuit of life, we take for granted the complex phenomenon of human motion. The growing voice, and steadily emerging identity of those who refuse to accept as intractable a deficit or loss in the ability to move, has led to more active research in the area of human locomotion.

Since the development of concentric needle electrodes by Adrian and Bronk (1929) and the postwar accessibility of electronic instrumentation, many researchers have undertaken the study of human locomotion. In the 1920's neurophysiologists pioneered the technique of recording and studying the electrical potentials produced by nerves and muscles - electromyography. Today electromyography is a standard method of clinical examination and diagnosis of abnormal nerve and muscle function.

The research work for this thesis was conducted in the Locomotion Laboratory, Department of Biomedical Engineering, Chedoke Rehabilitation Centre, Hamilton, Ontario. The application of cross correlation to determine phasic interdependence and degree of co-activity of electrical signals has been encouraged by the successful use of this tool in Electrical and Communications Engineering. The assessment of locomotor function could be well served by a greater understanding of the phasic interdependence of muscle co-contraction and relaxation.

The goals of the original research presented in this thesis are:

1. the processing of electromyographic (EMG) signals collected during locomotion from major muscle groups in the lower limb.
2. the analysis of phasic interdependence of muscle activity by the application of cross correlation.
3. the interpretation of the results.
4. the demonstration of a viable link between the results and the clinical picture.

Chapter 2 gives the reader an overview of locomotion and electromyography, and describes some of the research which has been reported. The basis of neuromuscular function, and the resulting locomotion is described with reference to the stride cycle. Normal and pathological gait and EMG are characterized.

The experimental protocol followed in the acquisition of EMG and stride cycle data from five normal and fourteen hemiplegic subjects is explained in Chapter 3. A short discussion of the statistical properties of EMG concludes the chapter.

A comparison of processing methods for making the electromyographic signals amenable to mathematical analysis by cross-correlation is discussed in Chapter 4. The reliability of a second order Butterworth low pass filter and a mid-point moving window average filter is investigated. The effect of these filters on bioelectric signals is described and the filter characteristics are discussed.

The cross correlation function is described in Chapter 5. The results for the cross correlation of pairs of signals recorded from human muscle during locomotion are reported for both normal and hemiplegic subjects. A discussion of the results concludes this chapter.

All conclusions drawn from the research work are reiterated in Chapter 6. Some of the difficulties encountered during this work are presented, as are the merits of the work considered in the scheme of human curiosity.

CHAPTER 2

HUMAN LOCOMOTION

2.1 Introduction

The phenomenon of motion in humans is being investigated by utilizing many different techniques. Grieve (1968), Cavanaugh and Grieve (1973), and Murray (1967) have analysed movement while Basmajian (1974), and Battye and Joseph (1966) have analysed electromyographic muscle activity. The effect of walking speed on stride length, stride period, and period of swing has been described by Grieve (1968), and Andriacchi (1968) has identifed walking speed as a basis for normal and abnormal gait measurements.

Murray et al (1966), and Drillis (1958) have used interrupted light photography of subjects wearing anatomical markers. Serial anatomical point trajectory data were acquired and stick diagrams were then made to assist in gait analysis. Winter et al (1972) developed a television-computer analysis system. The television records the position of anatomical markers and the computer uses this temporal and spatial information to produce data which describes limb position and joint angles.

Peat et al (1976) have used this TV system to acquire gait data at a rate of about 50 Hz. A major improvement in electro-optical recording is the "Selspot" or Selective Spot Recognition System with a sampling rate of 322 Hz. Cameras monitor the changes in position of small LED's attached to the subject.

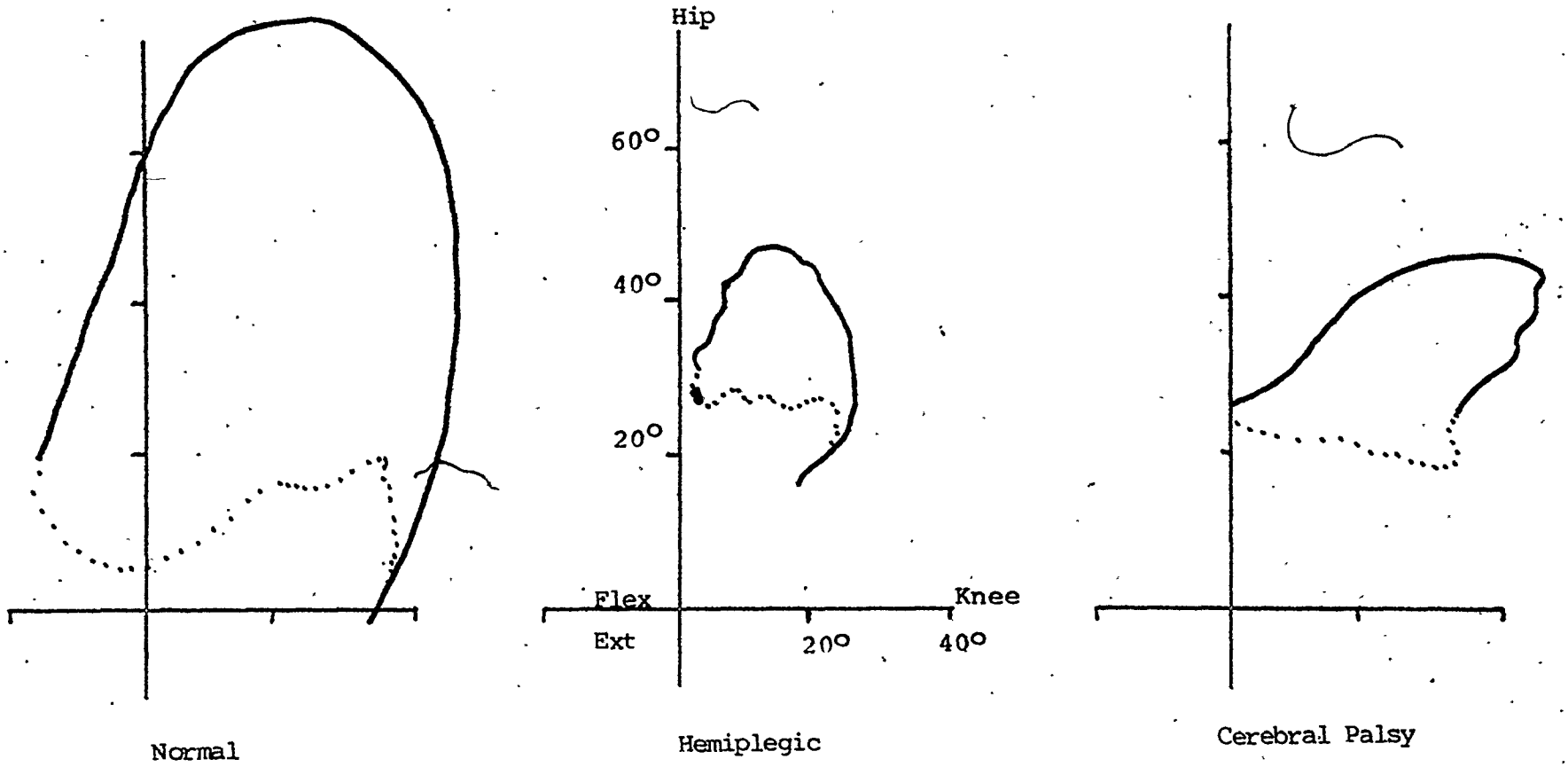


Figure 2.1 Hip Angle versus Knee Angle

Electromechanical devices such as the electrogoniometer are mounted at the joints. Voltage changes record changes in joint angles, and the display of angle versus angle for two joints produces distinctive shapes for normal, hemiplegic and cerebral palsy gait (Figure 2.1). Hershler and Milner (1977) have used this method with some success to describe normal, amputee and cerebral palsy locomotion.

A walkway with an electric conducting surface was devised by Drillis (1958) to obtain foot position timing data. Bloch and Szarka (1979), have proposed a similar method.

Marks and Hirschberg (1958) have analysed ground reaction forces, and energy expenditure in walking has been investigated by Beckett and Chang (1968).

These references are representative of the considerable research that has been reported concerning human locomotion. This research has concentrated heavily on the kinematics and kinetics of human locomotion. The activity patterns of the musculature producing this motion have not been as quantitatively analysed. The electromyographic (EMG) signal which is produced by the muscle during activation has only been analysed visually or by extremely simple processing. Consequently only some of the relationships between the electrical activity of muscles and the resulting forces and movement during locomotion have been identified. Most of the laws which relate bioelectric phenomena to motion have yet to be understood.

This chapter presents an abbreviated description of the electrical behavior of the anatomical elements which cause motion, and a description of the action of the four superficial lower limb muscle groups with which this study is concerned. Observable indicators of normal and pathological gait and EMG are discussed.

2.2 Basic Elements of Neuromuscular Organization

There exists across nerve and muscle membrane a potential gradient which results in a potential difference between the inner and outer cell of the order of 60 mV, called the resting potential of the cell. By convention, the outer surface is at zero potential with the inner cell negative. When the transmembrane potential of the nerve cell is reduced by about 15 mV, a swift change in the properties of the membrane takes place causing membrane currents (Figure 2.2) and local fibre currents (Figure 2.3) to flow. The current has a capacitive component C_m, ($1\ \mu F/cm^2$), and current flows due to ion transport. I_e and I_i are the longitudinal external and internal currents respectively, and are positive in the direction of increasing x as shown. E_K, E_{Na} and E_{Cl} are the electrochemical potentials due to the intra and extracellular differences in the concentrations of potassium, sodium and chloride ions respectively. The external and internal resistances per unit length of membrane are r_1 and r_2 respectively, and r_m is the shunt resistance. The response to depolarization of the membrane past threshold is called the "all-or-none" phenomenon since its

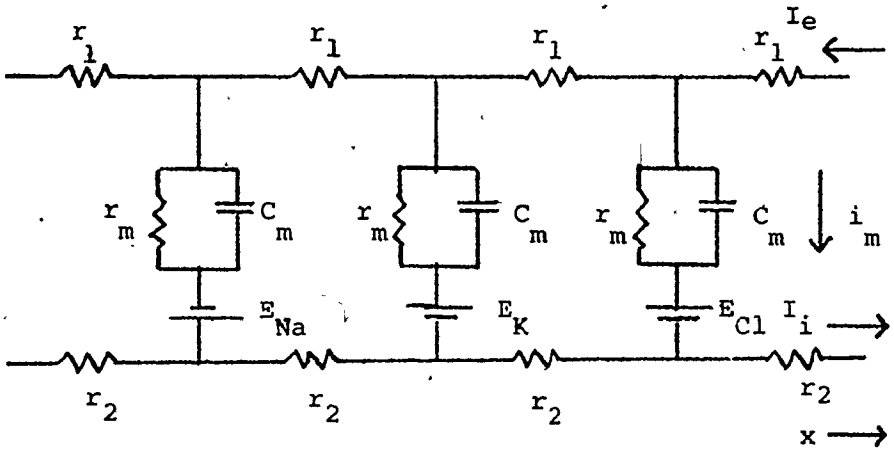


Figure 2.2 Membrane Currents

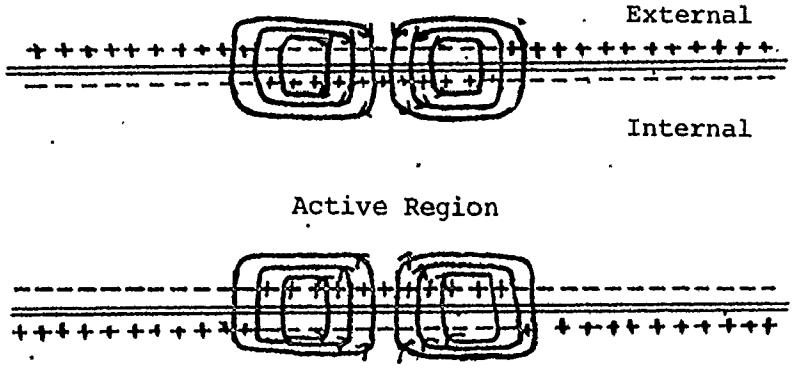


Figure 2.3 Local Fibre Currents

amplitude is independent of the size of the stimulus. The transmembrane potential changes swiftly from negative to positive and then recovers more slowly to resting potential. This series of events is called the action potential (Figure 2.4). The inflection represents the 15 mV depolarization threshold. The local depolarizing current reduces the membrane potential in adjacent regions and the action potential is propagated down the length of the fibre.

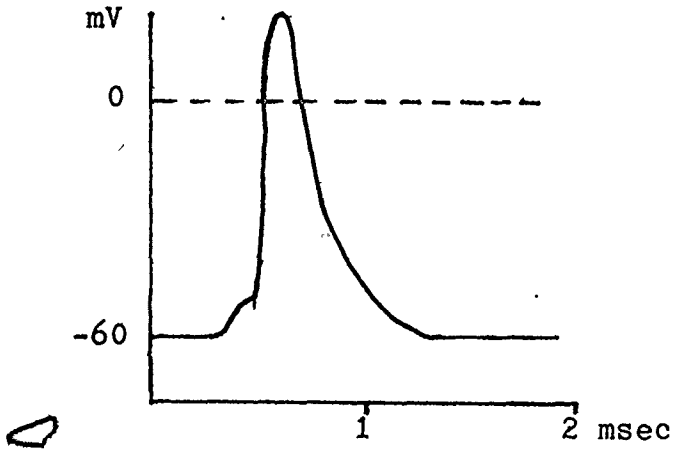
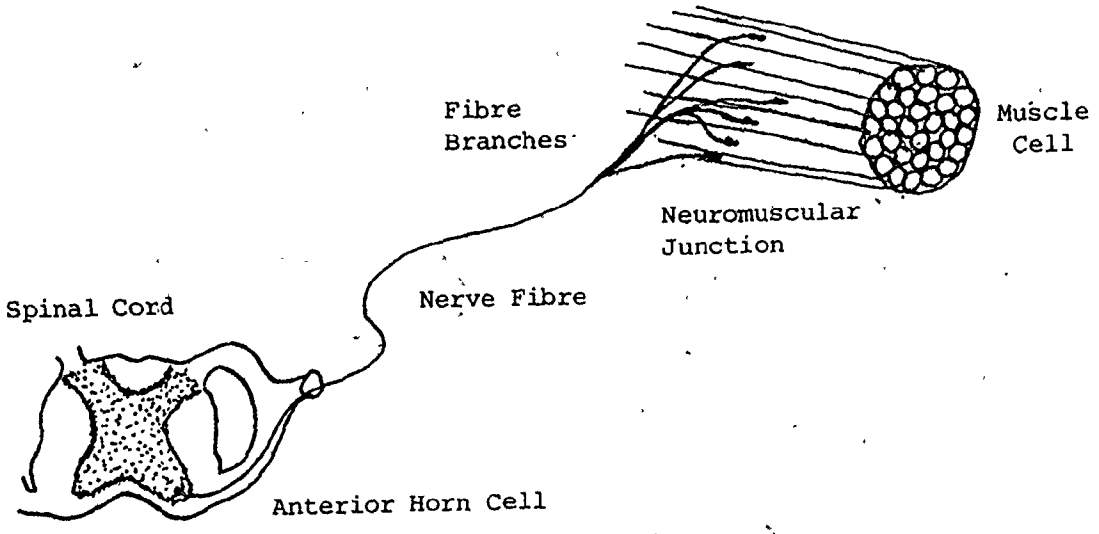


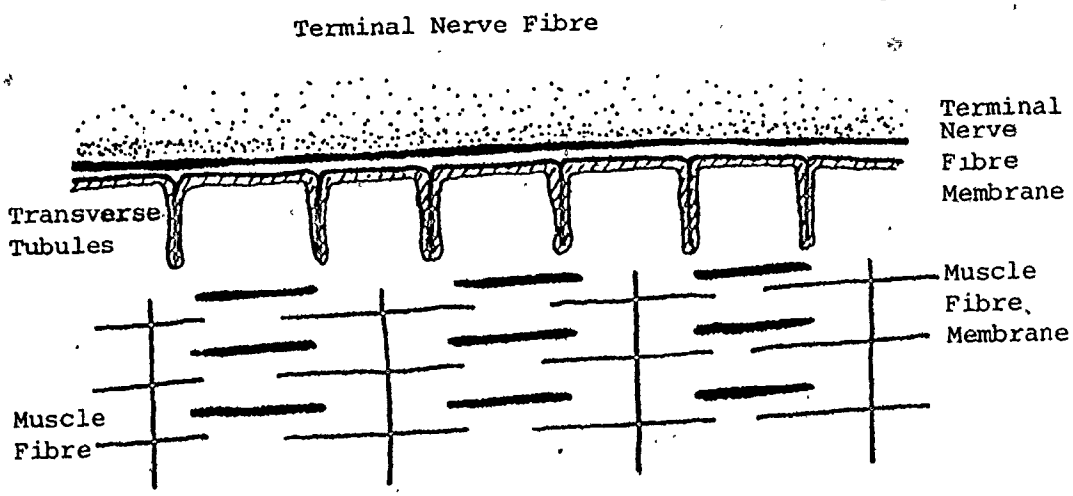
Figure 2.4 Action Potential

The electrical behavior of the muscle cell is similar to that of the nerve cell. Skeletal muscle fibre is composed of bundles of myofibrils enveloped by a double membrane, the sarcolemma. In large muscle fibres, there is a network of tubules called the sarcoplasmic reticulum and the transverse (T) system. These tubules make up a large fraction of a muscle fibre and serve to increase its surface area.

A muscle action potential is initiated at the neuromuscular junction (motor end plate) (Figure 2.5) by an end plate potential



(i) Motor Unit



(ii) Motor End Plate

Figure 2.5 Neuromuscular Junction

change of about 35 mV and is propagated by the superficial membrane. The action potential is transmitted to the interior of the muscle fibre by the tubular network and results in a brief mechanical contraction of the whole muscle fibre called a twitch.

Increased contraction in the muscle is produced by activating a larger number of fibres or by stimulating the cells at a higher frequency. The response of the muscle represents the sum of the contributions of all the stimulated fibres.

The motor unit (Figure 2.5) is the smallest unit of normal muscle response. It is comprised of an anterior horn cell, a neuron, (nerve fibre) and the family of muscle fibres which it subserves.

Any electrical impulse transmitted down the neuron is translated into a contraction of its "family" of muscle fibres. Throughout the muscle, motor units respond to irregular trains of impulses in an asynchronous fashion at rates of 1-50 Hz, and the smooth muscle action which we see is the summation of all these responses.

The electromyogram (EMG) is the electrical signal resulting from the temporal summation of those action potentials in close proximity to the recording electrode (Figure 2.6). The amplitude of the EMG record varies as the frequency of firing, and with the number of units activated, as well as the size of the activated fibres. Katz (1966) explains what is known about the transmission of messages in the living body. His discussion considers the underlying physics and physical chemistry of these phenomena.

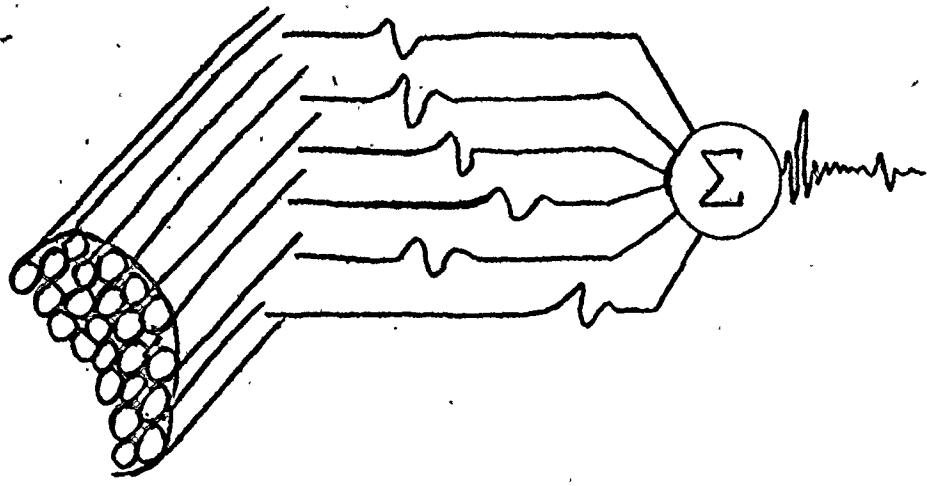


Figure 2.6 Temporal Summation of Muscle Fibre Responses

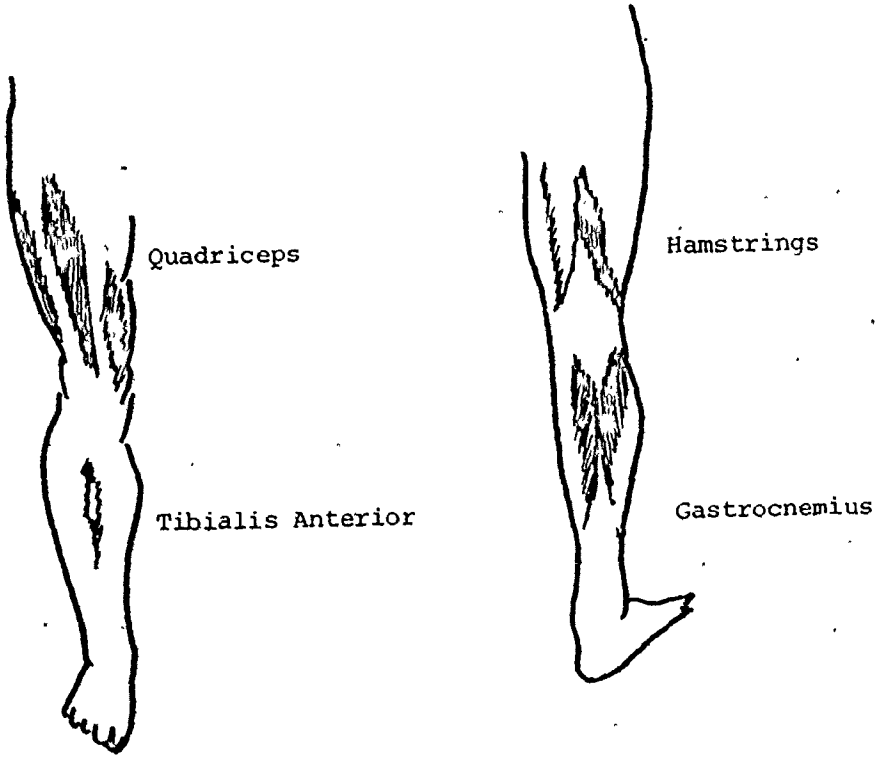


Figure 2.7 Main Muscle Groups of The Lower Limb

2.2.1 Muscles of the Lower Limb

In this study, the four main lower limb muscle groups which produce locomotion were selected for EMG study. These are the quadriceps, hamstrings, tibialis anterior and gastrocnemius muscles (Figure 2.7).

The quadriceps group includes the anterior (front) muscles of the upper leg which act to extend or straighten the knee, and flex or bend the hip. The tone or normal degree of tension in the quadriceps muscle plays an important role in strengthening the knee joint.

The posterior muscles of the thigh, the hamstrings, act to flex or bend the knee or to extend the hip.

The tibialis anterior, which runs along the shin, dorsiflexes the ankle and causes the foot to turn upward.

The gastrocnemius (calf) muscle plantarflexes (extends) the ankle and flexes the knee. Plantarflexion is the action of the ankle that precipitates a push against the ground. A propulsive force is generated by using the foot as a lever and raising the heel off the ground.

Thus, the action of the quadriceps and gastrocnemius precipitates extension of the leg while the action of the hamstrings and tibialis anterior causes flexion (bending) of the leg (Figure 2.8). Just as quadriceps and hamstrings act to stabilize the knee, so too do the tibialis anterior and gastrocnemius give stability to the ankle. If muscles are not activated sufficiently during the proper period of the gait cycle, pathological gait results. For greater

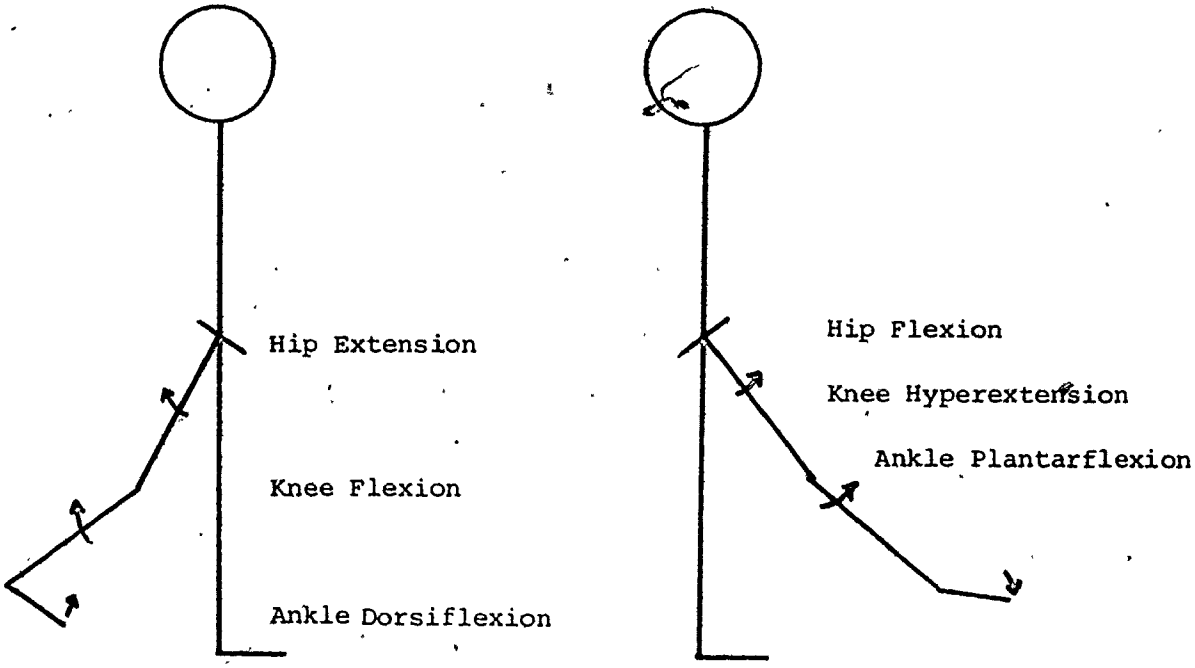


Figure 2.8 Flexion and Extension

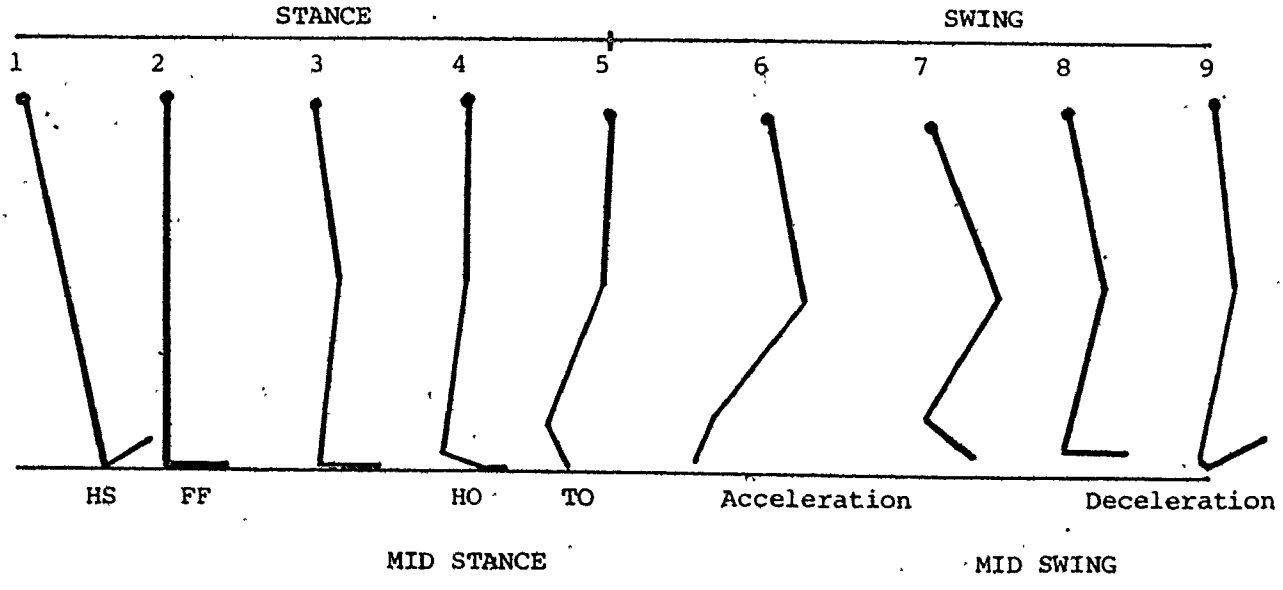


Figure 2.9 The Stride Cycle

detail, the reader is referred to Hollinshead's text (1976) on functional anatomy.

2.3 Human Gait

Dorlands (1974) defines gait as the manner or style of walking. Normal gait is as much characterized by individual personality as it is predicated on inherited bone and muscle structure. Gait is accomplished by the integration of all inputs describing body position and external forces and the mediation of this information by the higher centers of the central nervous system with modulated response as motion. Gait, then, is individual style of motion.

2.3.1 The Gait Cycle

By definition, walking is made up of a series of strides (Peizer et al, 1969). A stride cycle begins the instant one foot is in contact with the ground and ends with the next footstrike of the same limb (Figure 2.9). Each stride is characterized by two phases - stance phase, in which the foot is in contact with the floor, and swing phase, in which it is not. Stance occurs from "heel strike" to "toe-off", and swing takes place from toe-off to the next heel strike of the same leg.

At ordinary walking speeds, a leg is in stance approximately 60% of the stride, and in swing 40%. For about 25% of the stride cycle both feet are in contact with the floor (double support).

Speed of walking is inversely related to the double support period. Step frequency is defined as cadence.

2.3.2 Normal Gait

The co-ordinated activity of all the muscles acting at the hip, knee and ankle joints results in normal human gait. The smoothness and efficiency of forward locomotion is accomplished by appropriate strength and timing of muscle contractions.

In normal gait (Figure 2.10) the quadriceps are active at heel off to extend the leg. The hamstrings contract at the end of swing to decelerate the leg prior to heel strike and to stabilize the knee in stance. The tibialis anterior is active at heel strike and again just after heel off to dorsiflex the ankle.

The gastrocnemius is active at toe off where the plantar flexed foot provides a forward push off and again at the end of foot flat in order to impart stability to the ankle.

2.3.3 Hemiplegic Gait

One major problem in the management of hemiplegic patients is spasticity, or an increase in activity over the normal tone of a muscle. Heightened reflex activity results. Abnormal co-ordination characterizes the spastic patient.

Mild spasticity allows co-ordinated gross movements, but produces clumsy finer movements. In moderate spasticity

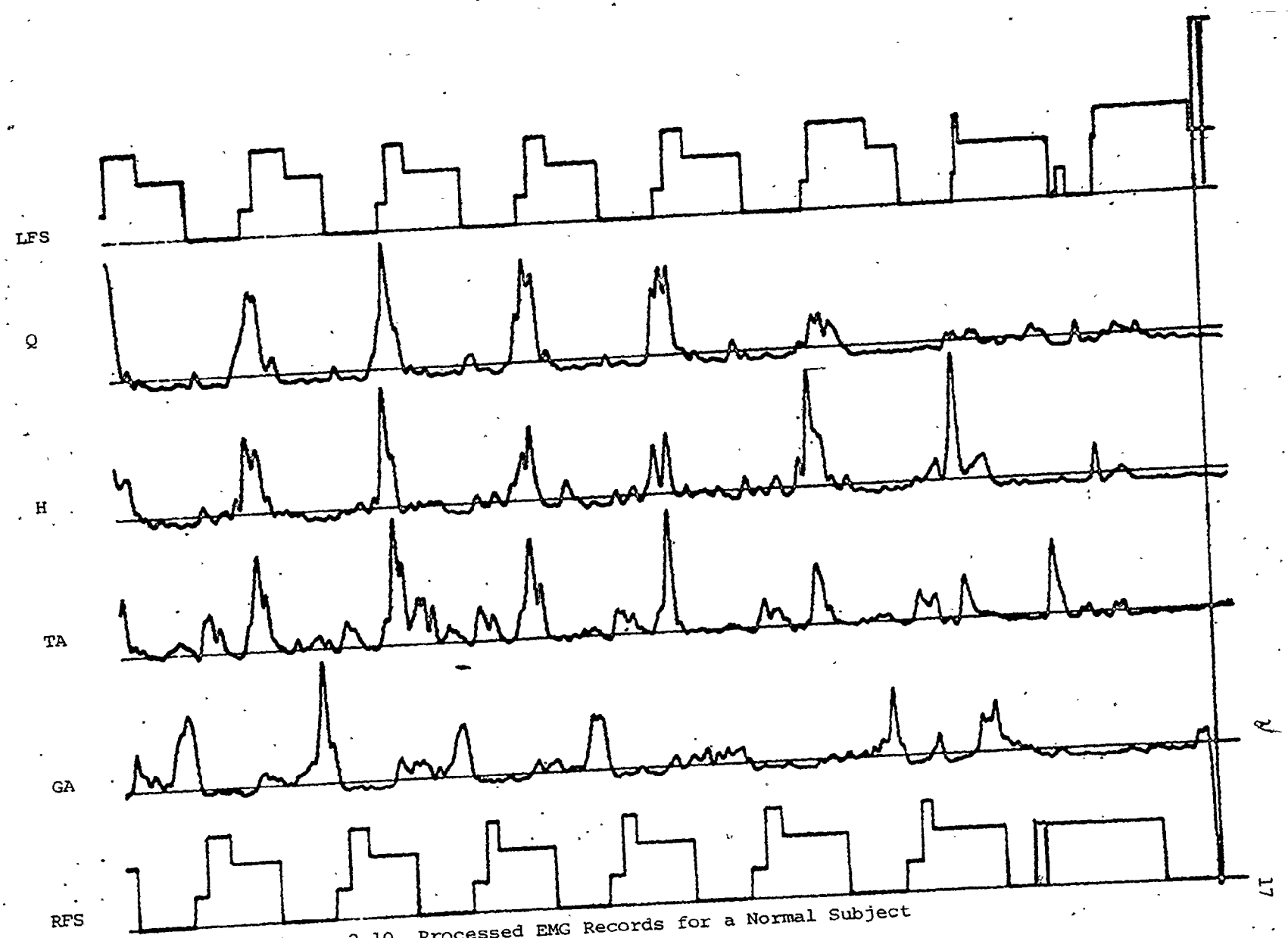


Figure 2.10 Processed EMG Records for a Normal Subject

slow movements are possible but the abnormal co-ordination requires more energy. Movement is impossible in severe spasticity.

Another problem is clonus, a series of abnormal reflex movements of the foot induced by sudden dorsiflexion causing a series of brief contractions and relaxations of the calf muscle. Automatic reflexes which are learned in childhood are underdamped as a result of cerebral insult to effect a closed loop positive feedback system.

A major problem in ambulatory hemiplegic patients is weakness or partial paralysis of some or all lower limb muscles. This results in an inability to perform basic movements or to maintain adequate control and stability of the affected limb, especially during weight bearing.

Hemiplegic patients exhibit these problems in varying degrees resulting in a broad spectrum of pathological gaits.

To accomplish smooth, co-ordinated motion, there must be normal muscle tone and graded muscle control of agonist/antagonist pairs of muscles integrated with synergist (other) muscles for appropriate timing and limb movement.

The agonist muscle initiates a movement while the antagonist exerts control to prevent associated undesired movements. Synergists act in concert with the agonist/antagonist pair to produce a movement, and relax while the agonist is working.

The problem of the hemiplegic patient is his inability to select combinations of learned patterns. Abnormal patterns must be supressed so that normal patterns may be relearned.

2.4 Electromyography

At the end of the 18th century, the scientist Galvani conducted experiments with nerve and muscle electricity. He discovered that muscles contract when stimulated electrically, and conversely, muscles produce a detectable voltage when they contract.

Early 20th century technological advances gave impetus to the study of muscle electricity - electromyography. In 1928, Adrian and Bronk published their first work on the discharge of impulses in motor nerve fibres. Since that time quantum advances in electronics have made possible detailed investigations of muscle function using electromyography. This tool is widely accepted as an adjunct to clinical diagnoses of nerve and muscle pathologies. Electromyography also reveals details of timing and co-ordination of muscle activity at the macroscopic level and allows non-invasive methods of studying dynamic muscle activity.

2.4.1 Normal EMG

There is no detectable electrical activity in normal relaxed muscle. Motor unit potentials appear only during voluntary or reflex muscle activity. During a weak contraction, single motor unit potentials can be discriminated in the electromyogram. However, during a strong contraction, motor unit potentials are recorded as an interference pattern in which individual motor unit activity cannot be discriminated (Figure 2.11).

The duration of a single motor unit potential is measured from the initial deflection to the final return to resting level.

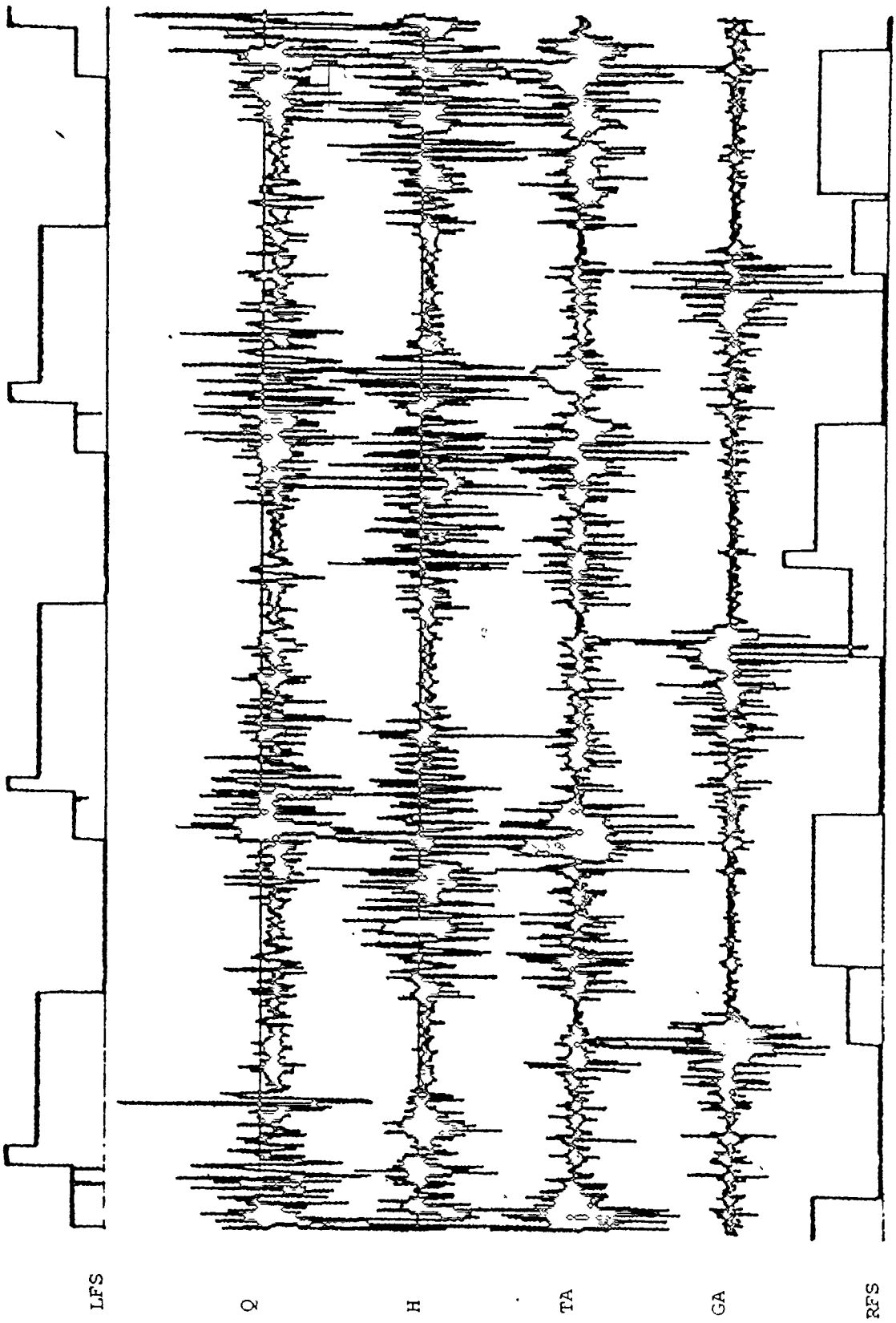


Figure 2.11 Raw EMG for a Normal Subject

Typically these durations are of the order 2 - 3 milliseconds. The duration does vary however, with the muscle studied, muscle temperature and age of the subject. The potentials of eye muscles are much shorter than those for say, skeletal muscles. Recorded action potential durations are also a function of electrode geometry and geography.

Maximal electrical stimulation of the nerve elicits a muscular response indicative of the number of muscle fibres which can respond to the stimulus. Normal gait study EMGs reveal a phasic motor unit potential interference pattern with regular contraction and relaxation of muscle groups (Figure 2.10). In normal muscle, almost 80% of the potentials are diphasic, that is there is an initial positive phase followed by a large negative one, or triphasic (having a third positive phase).

Petersen and Kugelberg (1949) found that action potentials recorded in the same muscle by unipolar and coaxial needles have a mean duration of 7.34 msecs., while those recorded with fine wire electrodes of the same dimension have a mean duration of 2.28 msecs. Since fine wire electrodes have a smaller recording area, only the potentials of a few fibres of an active motor unit are recorded. Dispersion effects of surface electrodes prolong action potential durations and render these electrodes less accurate than needle electrodes.

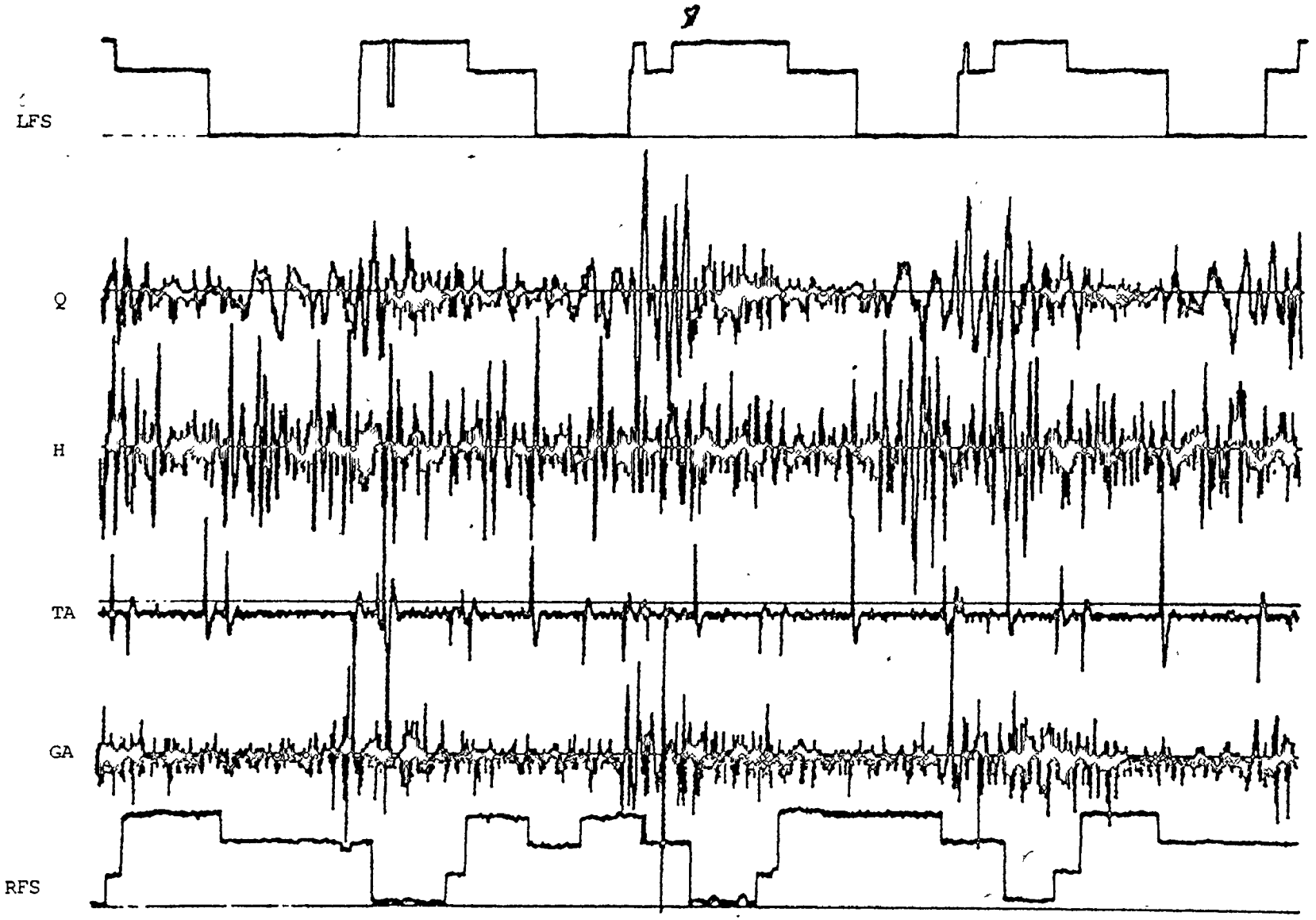


Figure 2.12 Raw EMG Records for a Severely Involved Hemiplegic Subject

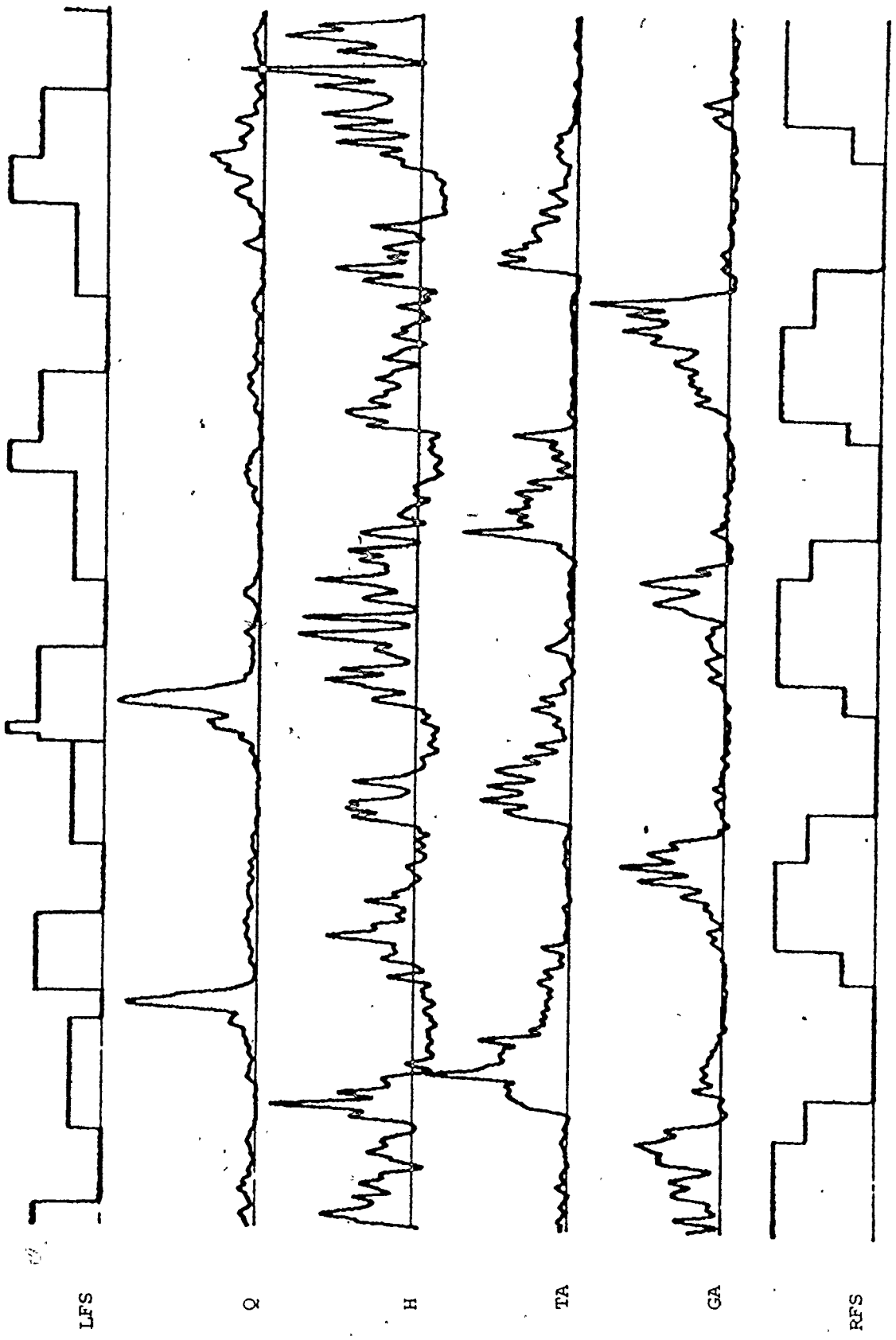


Figure 2.13 Processed EMG Records for a Severely Involved Hemiplegic Subject

Surface electrodes are adequate for gait studies because we are not concerned with individual motor unit responses. We are concerned with the presence of muscle activity, and the time relationship of the activity of different muscles during locomotion.

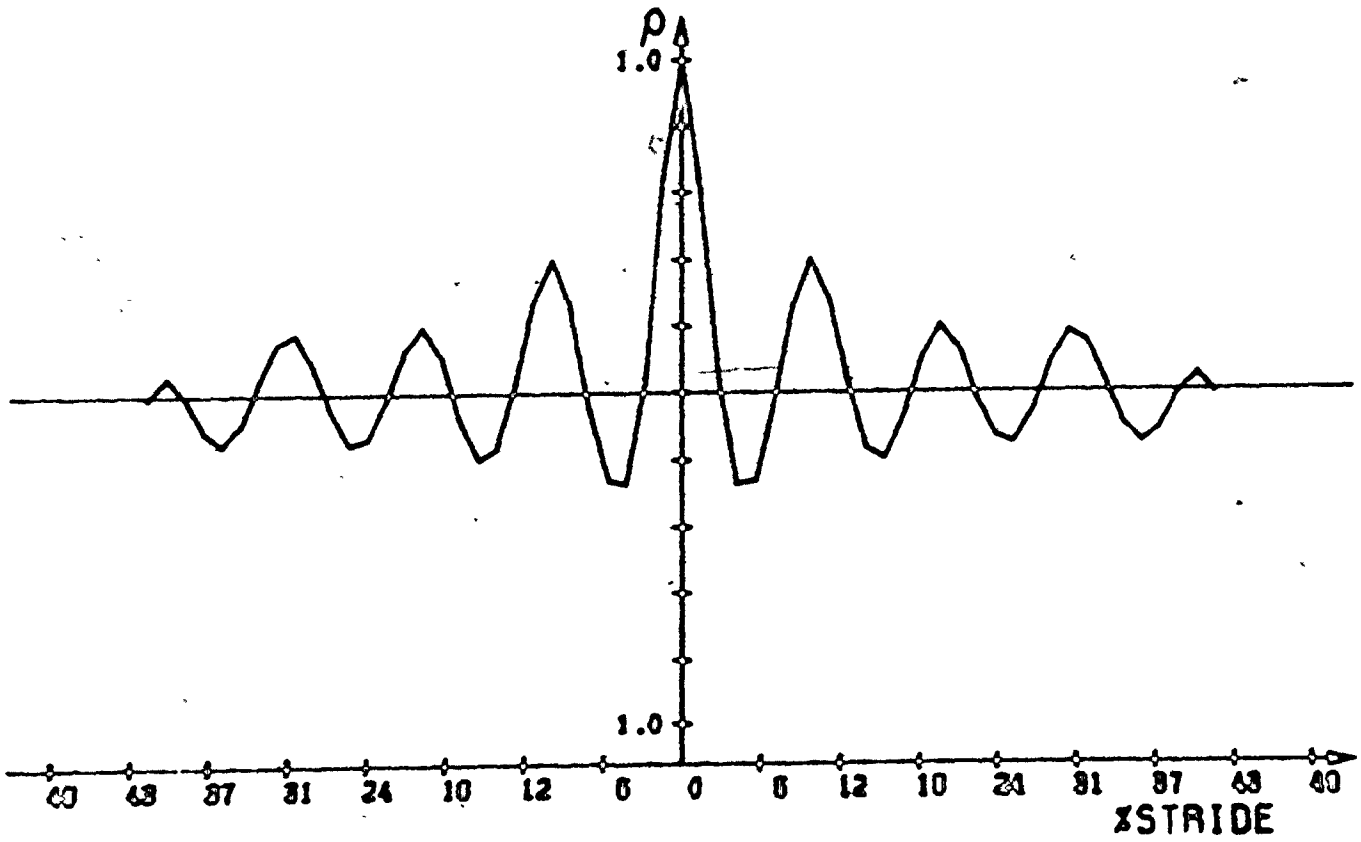
2.4.2 Pathological EMG

The individual action potentials recorded from a hemiplegic patient are similar to normal since the hemiplegia is a result of central nervous system damage. The individual motor units are not affected other than in their control. The interference patterns recorded from a hemiplegic patient during movement are only altered from normal because of loss of control. (Figure 2.12).

Hemiplegic gait study EMGs of subjects who have sufferd a stroke (Figure 2.13) are characterized by spastic activity (prolonged contraction), loss of phasic contraction/relaxation, and arbitrary or uncertain firing order.

The problem of clonus, described in section 2.3.3, is graphically illustrated when the EMG record of the calf muscle of a hemiplegic subject is autocorrelated (Figure 2.14). The characteristic repetitive contraction/relaxation of the calf muscle produces an autocorrelation curve with several peaks of decreasing amplitude. Figure 2.15 illustrates the single peaked autocorrelation curve for a normal subject.

Figure 2.14
Autocorrelation of EMG Record for Gastrocnemius Muscle of a Hemiplegic Subject Exhibiting Clonus



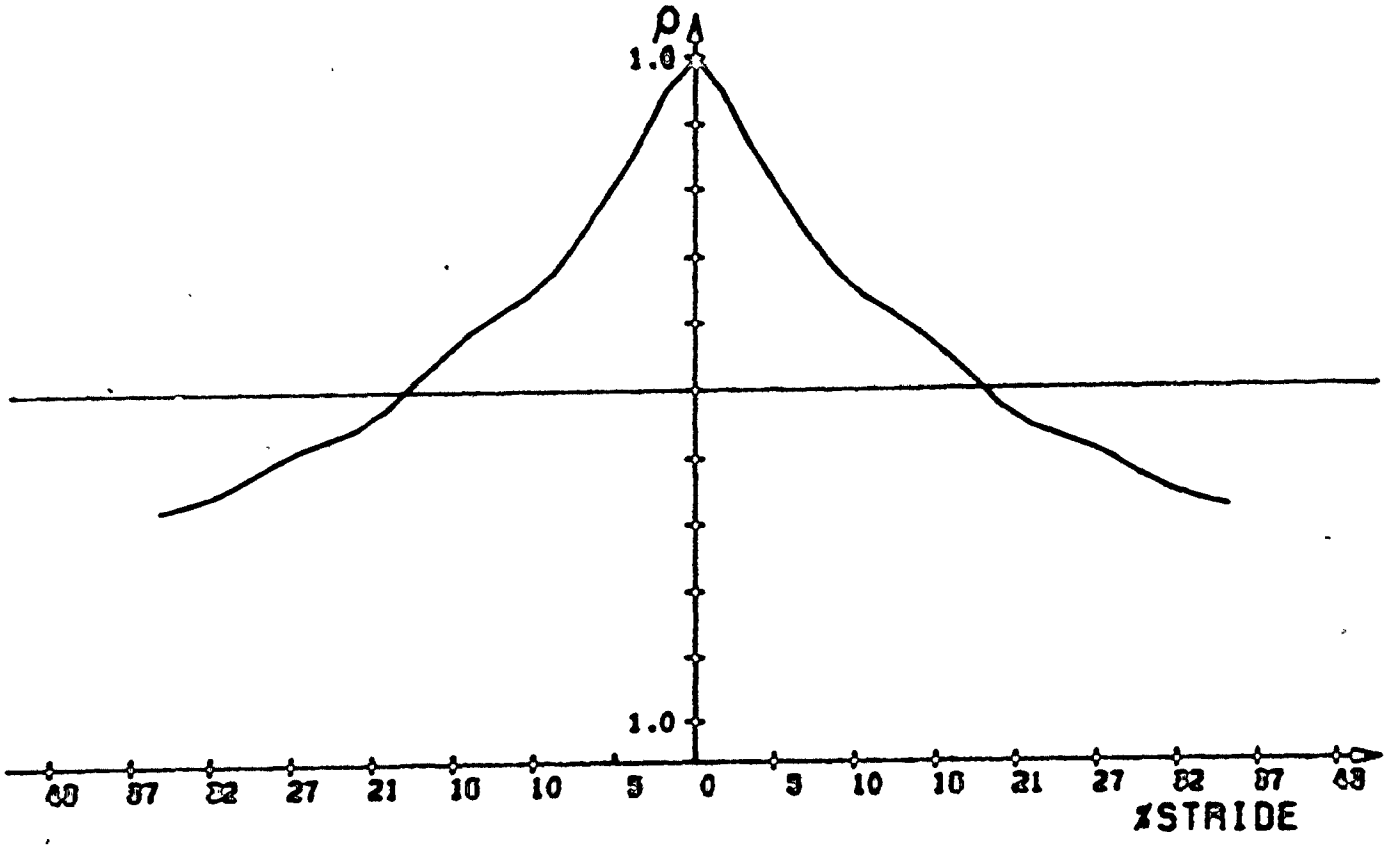


Figure 2.15 Autocorrelation of EMG Record for Gastrocnemius Muscle of a Normal Subject

2.5 Conclusion

A careful study of EMG records yields information about the relationship of bioelectric phenomena to locomotion, since EMG is the graphic representation of this bioelectric phenomena. The study of EMG activity recorded in pathological motion must then reveal some information about the nature of the pathology, the extent of neuro-muscular insult, and the effect of the pathology on natural motion.

The objective of EMG investigations is to deduce the electrical and so the physiological state of muscles from surface potential recordings. These surface recordings are evidence to be sifted to yield a manageable number of parameters which we may use to describe normal gait. Deviant patterns should describe pathological gaits.

CHAPTER 3

DATA ACQUISITION

3.1 Introduction

The purpose of this thesis is to describe a method of processing electromyographic signals collected from normal and hemiplegic subjects. In order to satisfy our requirements for reliable quantitative data, acquisition protocol observed by the Gait Research Group, Locomotion Laboratory, Chedoke Rehabilitation Centre, will be described in this chapter. The signal acquisition and processing system is described diagramatically in Figure 3.1.

3.2 Signal Acquisition and Processing System

EMG signals from four muscle groups of the lower limbs were recorded using surface electrodes made by Becton-Dickinson. These electrodes consisted of silver/silver chloride discs, one centimetre in diameter, filled with electrode jelly to minimize skin/electrode impedance (Figure 3.2).

The peculiar properties of electrodes in physiological potential measurements are discussed in depth by Plonsey (1969) and Geddes and Baker (1976). Silver/silver chloride electrodes are the usual choice in electrophysiological applications because the potential arising from the difference in chloride concentrations of the silver chloride electrode and the skin surface is nil.

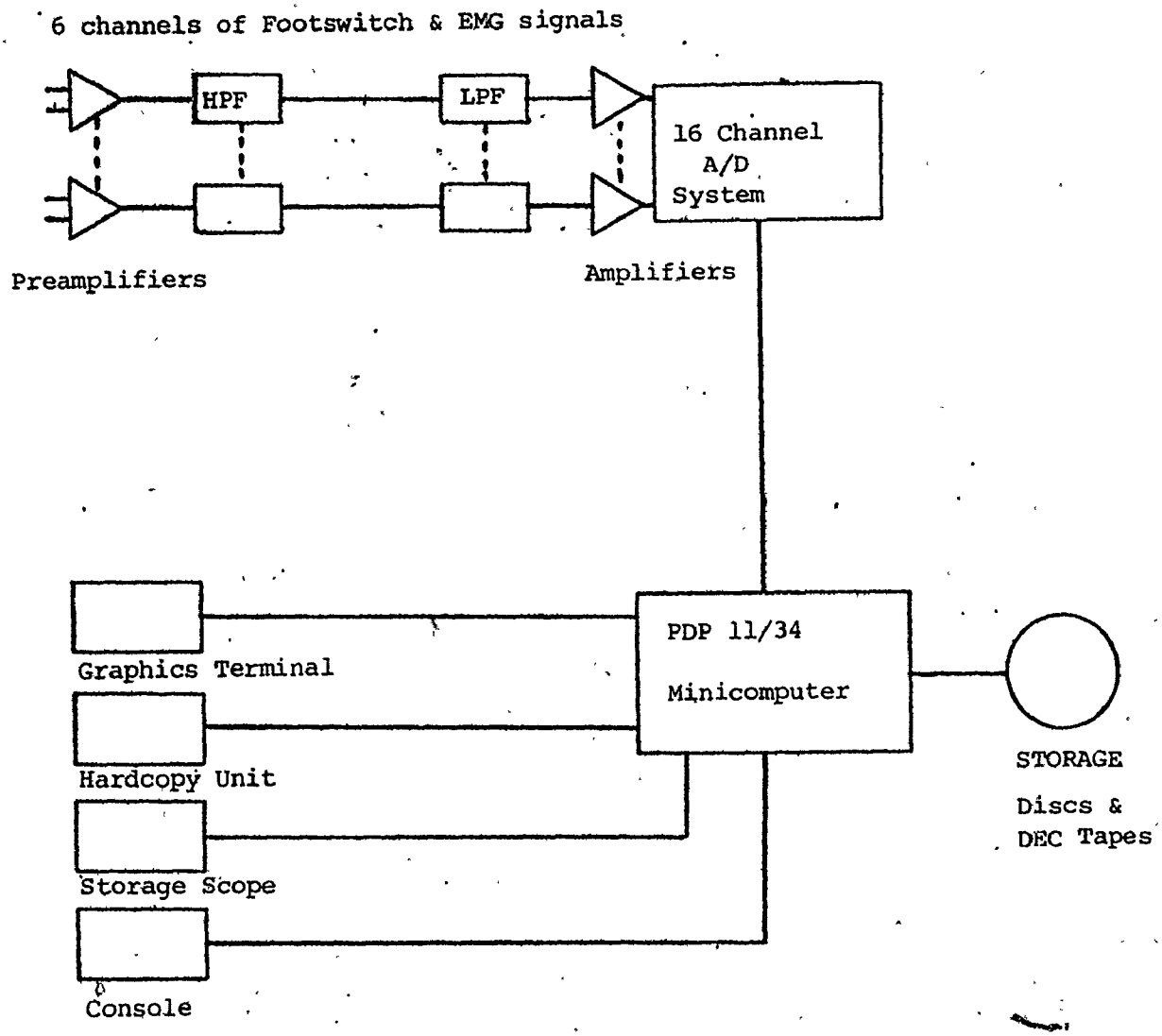


Figure 3.1 Signal Acquisition and Processing System

The skin was prepared by swabbing with alcohol to remove surface oil for optimum skin contact. Surface electrodes were the most practical choice in EMG recording in the gait laboratory despite the fact that only superficial activity is recorded. These electrodes can be applied by non physicians, are painless and record from a large area. They are therefore less affected by slight shifts in electrode position.

Indwelling fine wire electrodes detect the activity of a small pool of neurons. A slight shift may cause the next samples to be taken from the activity of a different pool of fibres. Repeatability of electrode placement is virtually impossible. In addition, these electrodes cause pain to the subject, and may cause the subject to move unnaturally. They also require sterilization and application by a physician.

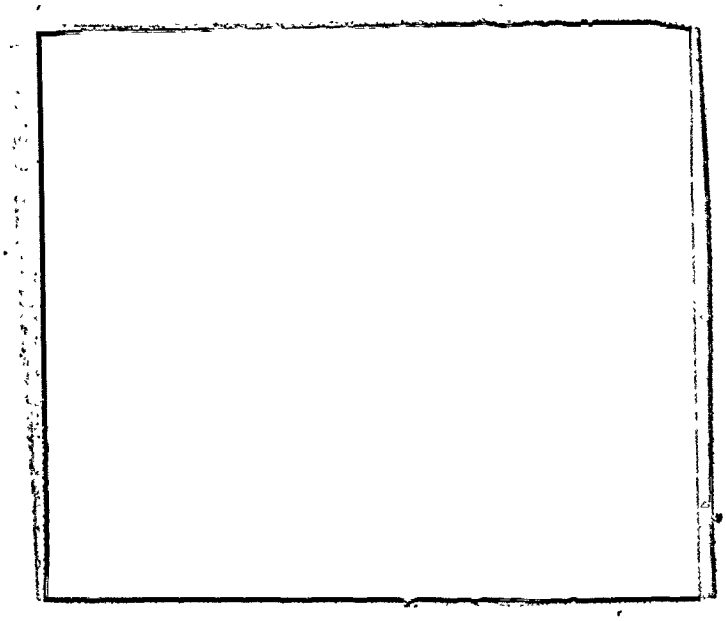


Figure 3.2 Surface Electrodes

The electromyographic signals from the electrodes were fed to miniature differential preamplifiers which were attached as close as possible to the electrodes. Signal preamplification (100 or 1000) at this point minimized interference from 60 Hz mains as well as from fluorescent lighting.

Amplifier outputs were connected to a connector block carried about the subject's waist. Overhead coaxial cables carried the six channels of footswitch and EMG data to front end hardware for preconditioning.

High pass filters with a 20 Hz cutoff, 24 dB roll off were used for all EMG channels to filter out motion artifacts due to electrode movement on the skin or relative to the underlying tissue. Low pass filters with a 500 Hz cutoff reduced instrumentation noise and provided sufficient band limiting to avoid potential aliasing of EMG signals. For very small signals, computer site amplifiers with gains of 1, 2, 5, 10 or 20 were also available. The EMG signals were then rectified and averaged to extract an envelope signal. A 12 bit analog to digital convertor (ADC) system was used to sample the incoming signals. The ADC has a $\pm$ 5 volts maximum input and the converted sample error is $\pm$ 1/2 the least significant bit.

Prior to a data acquisition run, signal quality was checked with a visicorder (ultra violet paper recorder) interfaced to the PDP system. Any faulty electrodes, connections or footswitches could be easily identified and the problem corrected.

Acquired data were displayed on the graphics terminal for inspection, and stored on disc if accepted. At the termination of

the data acquisition session, the data file was transferred to magnetic tape and stored until required for processing.

3.3 Footswitch Signals

EMG data recorded during a walk were referenced to each stride by means of footswitch records generated by pressure sensitive switches placed under heel and toe of both shoes. Closure of the heel switch generates a one-third volt level. Closure of a toe switch generates a two-thirds volt level, and closure of heel and toe switches generates a full volt level. A typical footswitch pattern is shown in Figure 3.3.

Interpretation of footswitch signals should be done cautiously. In many subjects, both normal and hemiplegic (especially in older participants), the stance phase begins with the simultaneous placement of heel and toe on the floor. This of course results in the absence of the 1/3 volt level. Some subjects scuff their toe during swing, causing an intermittent 2/3 volt level during the swing phase. One must of course be alert to the possibility of switch failure.

3.4 Experimental Procedure

For each data acquisition session, the electrodes were applied by the same physiotherapist. Although the skin was prepared by rubbing with alcohol to remove surface dirt and oil, (and shaving where excessive hair was a consideration), care was taken not to

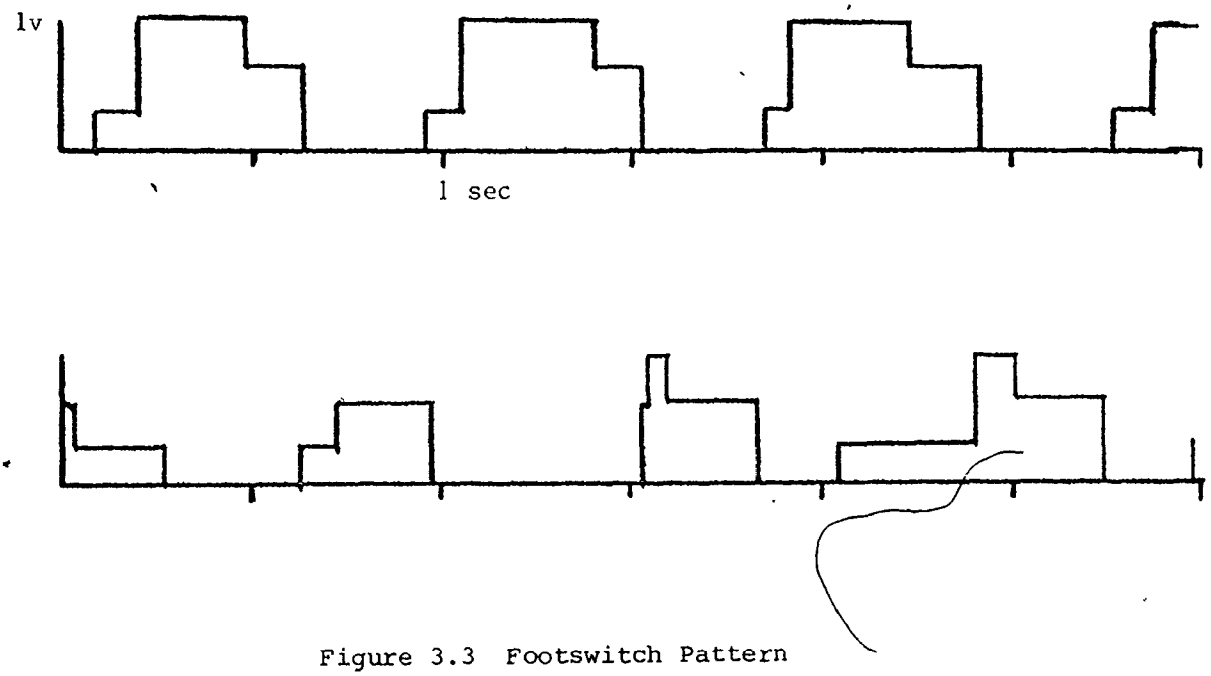


Figure 3.3 Footswitch Pattern

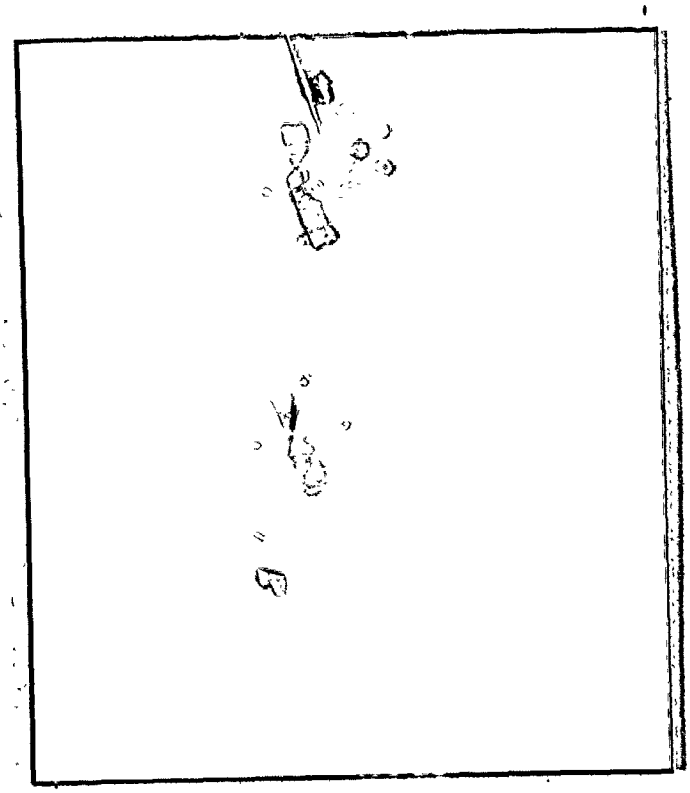


Figure 3.4 Instrumented Subject

abrade the skin. In older subjects, skin fragility must be considered, and in hemiplegic subjects one must be very mindful that any skin damage may result in ulcers. Obesity, muscle geometry and the degree of muscle atrophy influenced the placement of electrodes on each subject. Brown (1968) recommends consistent electrode placement since the waveform is dependent upon surface electrode position and size. Grieve and Cavanaugh (1974) found that differences in intensities and patterns of EMG are evident for different positions of electrode pairs on the muscle groups tested. The belly of each muscle was chosen as the most suitable site rather than a precise anatomical location since the anatomical geometry of the musculature changes from subject to subject.

Instrumented subjects (Figure 3.4) were asked to traverse a ten metre walkway. Data were recorded when the subject achieved a steady state gait. After collecting a resting record, normal subjects were asked to walk three times at comfortable, slow and fast walking speeds. Up to ten seconds of data could be collected for each walk. Hemiplegic patients were asked to walk at their most comfortable cadence. The speed of walk was calculated over a four metre section of the walk utilizing timers which were activated and deactivated by the interruption of an infrared beam set at this interval.

3.5 Statistical Properties of EMG

Surface recorded EMG contains both amplitude and frequency

information. The activation of a larger number of fibres or stimulating fibres at a higher frequency (Chapter 2.2), results in increased contraction. The order of recruitment of motor units is strictly random, and so an EMG signal is the result of a stochastic process. If the statistical methods described in this thesis are acceptable for processing EMG, the process must be shown to be stationary and ergodic. A stationary process is one in which the statistical properties are invariant with respect to time of observation. Stationarity can be safely assumed in a "walk" if the means and standard deviations are approximately equal for each stride period. Ergodicity of a stationary stochastic process is assured if the ensemble mean ("walk") is equal to single sample mean ("stride").

Thus, all the properties of an ergodic stochastic process can be determined by time averaging a single sample record ("stride"). Ergodicity is not a strict requirement for analysing gait data statistically because each stride is averaged and so ensemble averaging is already accomplished.

Most researchers assume stationarity and ergodicity of short term EMG signals. Subsequent descriptions of statistical treatment of EMG are based on this assumption.

CHAPTER 4

PROCESSING AND ANALYSIS OF EMG

4.1 Introduction

Even though we proceed with statistical analysis of EMG assuming it to be a record of a stationary process, we must remember that consecutive strides of normal human gait are not exactly repeatable. The challenge is to extract from the EMG record, consistent and reliable information which quantifies gait.

The preprocessing of acquired EMG signals results in a record of rectified, smoothed EMG which retains the phase and amplitude information of the original ("raw") record. This record also contains higher frequency components relating to instrumentation noise. These higher frequencies are eliminated by digital signal processing. The frequency content of the preprocessed record is, of course, different from that of the original "raw" record.

In this chapter the spectra of raw (Figure 2.11), preprocessed (Figure 4.1) and processed (Figure 4.2) (digitally low pass filtered) EMG are described. Two processing filters are discussed and the results of using each to process EMG are compared.

Figure 4.1 Preprocessed EMG for a Normal Subject

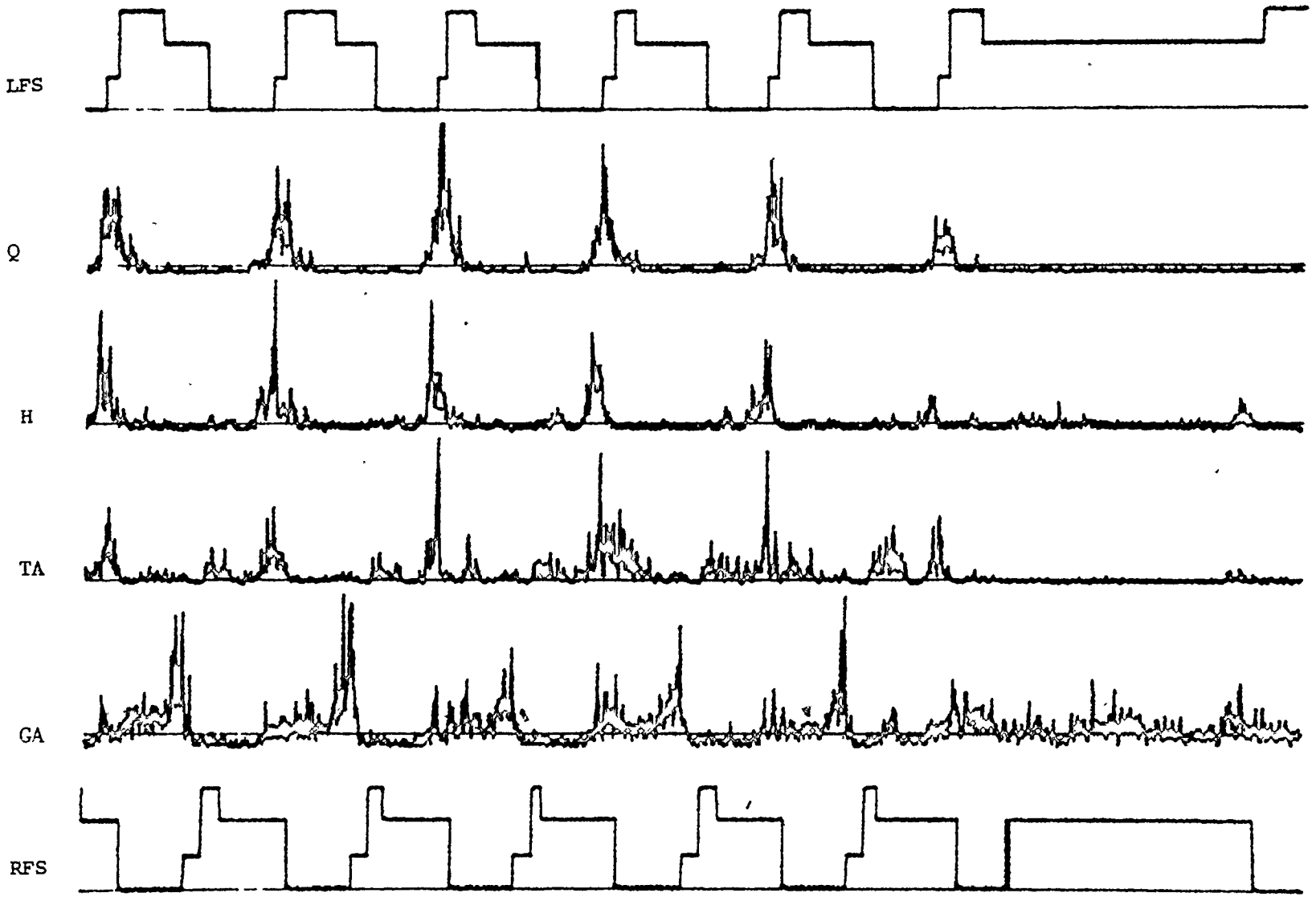
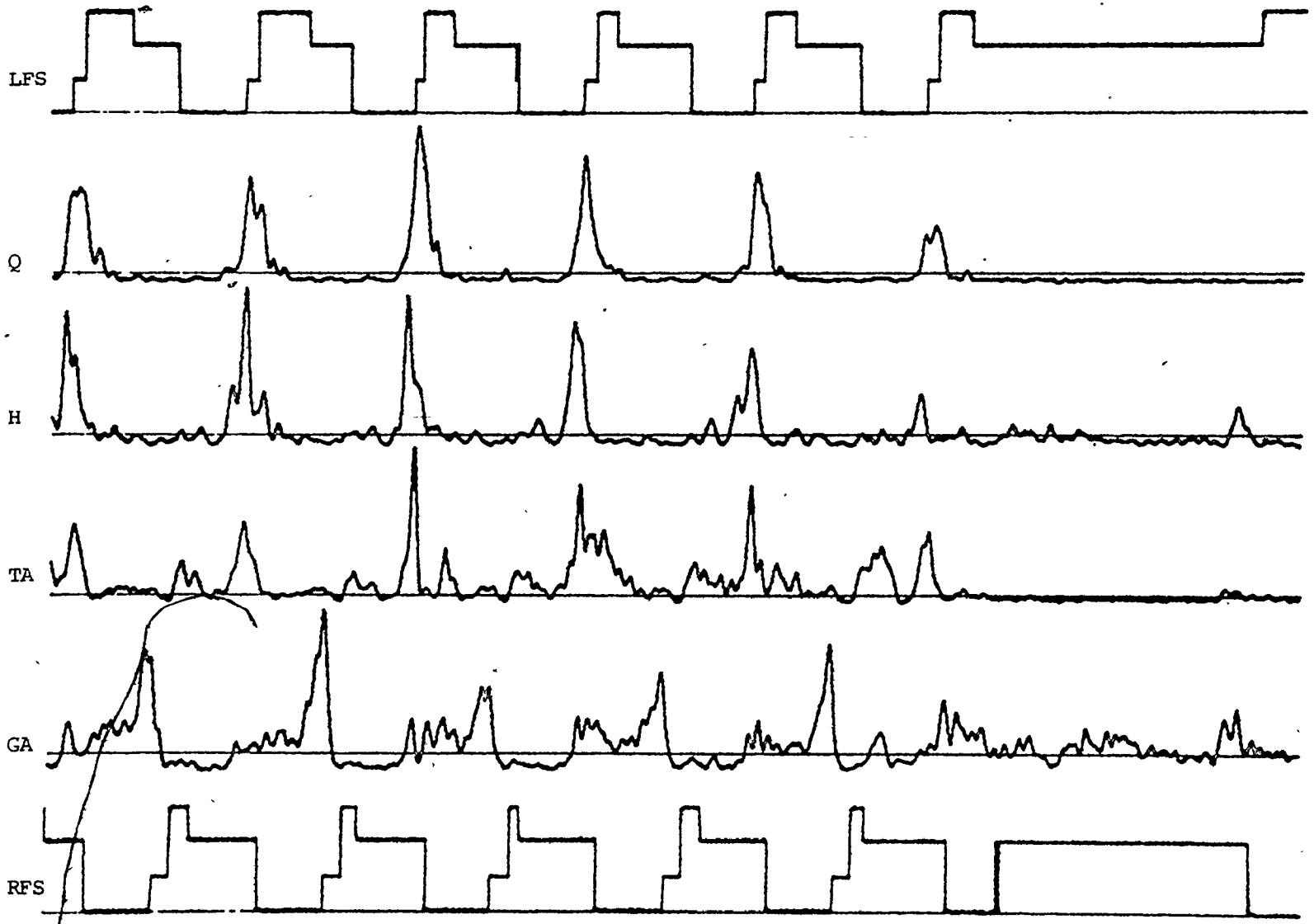


Figure 4.2 EMG of Figure 4.1 Processed with a Digital Low Pass Filter



4.2 Spectral Analysis of EMG

In a study by Gersten et al (1967) using needle electrodes, the results of harmonic analysis of EMG for normal subjects and subjects with known neuropathies (nerve disorders) and myopathies (muscle disorders) were reported for isometric contractions. The spectra of patients with myopathies could be easily distinguished from normals and the spectra of patients with neuropathies could be distinguished from spectra of patients with myopathies. Kwatny et al (1970) have used spectral analysis to study muscles at two levels of isometric contraction, before and during fatigue. Surface electrodes were used in this study.

The spectral analysis of normal EMG collected during gait was included in this study for three reasons:

1. to determine the power spectrum for surface EMG signals recorded during dynamic movement such as locomotion. Previously reported spectral results are for constant isometric contractions only.
2. to confirm the 200 Hz sampling rate for rectified, averaged EMG.
3. to confirm the selection of a 9 Hz low pass filter cutoff and a 60 millisecond window for a mid-point moving window average filter.

The normal EMG was acquired from a twenty four year old male at comfortable (1.08 m/sec), fast (1.38 m/sec) and slow (0.86 m/sec) walking speeds.

Power spectra were calculated using a Fast Fourier Transform (FFT2,International Mathematical and Statistical Library) with a base two algorithm. The Fourier Transform of f(x) is defined as

$$F(s) = \int f(x)e^{-i2xs}dx \qquad 4\text{-}1$$

To reduce side lobe levels from the normal left side records the input function f(x) by a cosine bell on a pedestal

$$x_t * 1/2(1-\cos 2\ t/T) \quad 0 \ t \ T \qquad 4\text{-}2$$

where x_t is the input data element, T is the total period of the signal and t is the number of the output element.

4.2.1 Spectral Analysis of Raw EMG

Representative power spectra for raw surface EMG recorded from the four muscle groups of interest are shown in Figures 4.3, 4.4, 4.5, 4.6. The features of interest for the power spectra of these muscles at three walking speeds are summarized in Table 4.1. The bandwidth was determined as that portion of the spectrum which is greater than 10% of the maximum amplitude of the spectrum. The raw data , acquired at the rate of 500 Hz, resulted in Fourier coefficients for 250 Hz.

Figure 4.3 Power Spectrum for Raw EMG - Quadriceps Muscle
Walking Speed 1.38 m/sec

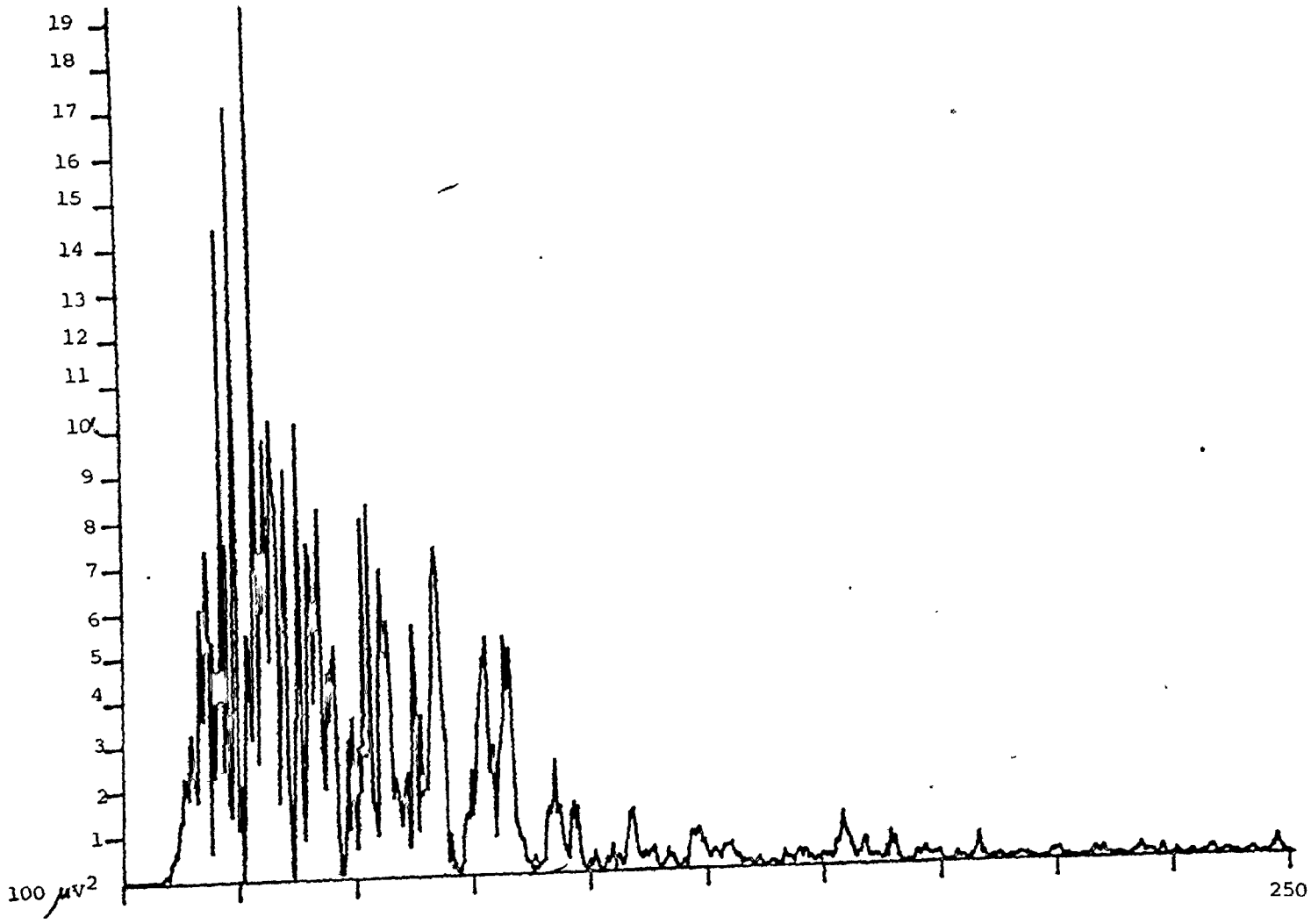
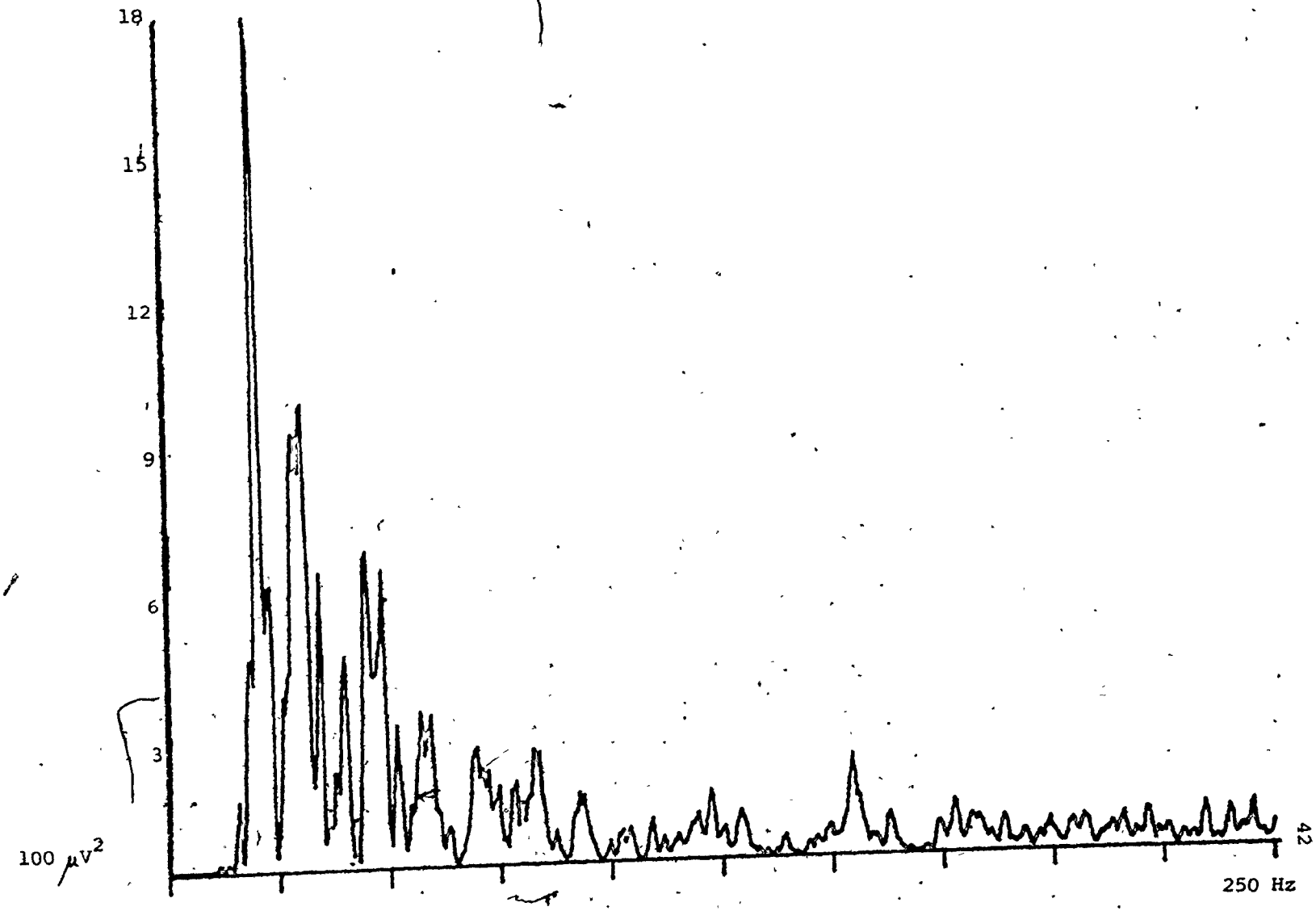


Figure 4.4 Power Spectrum for Raw EMG - Hamstrings
Walking Speed 1.38 m/sec



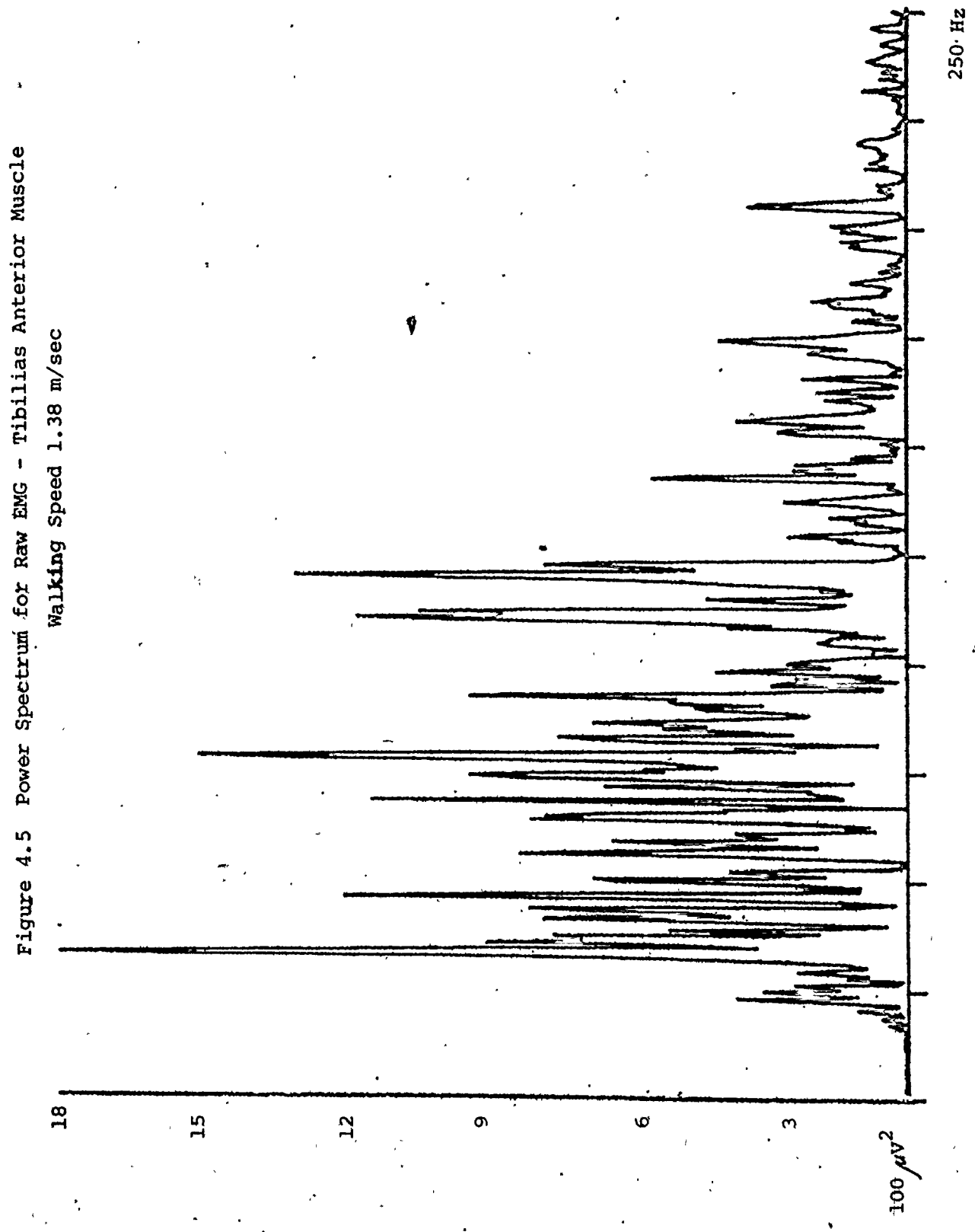
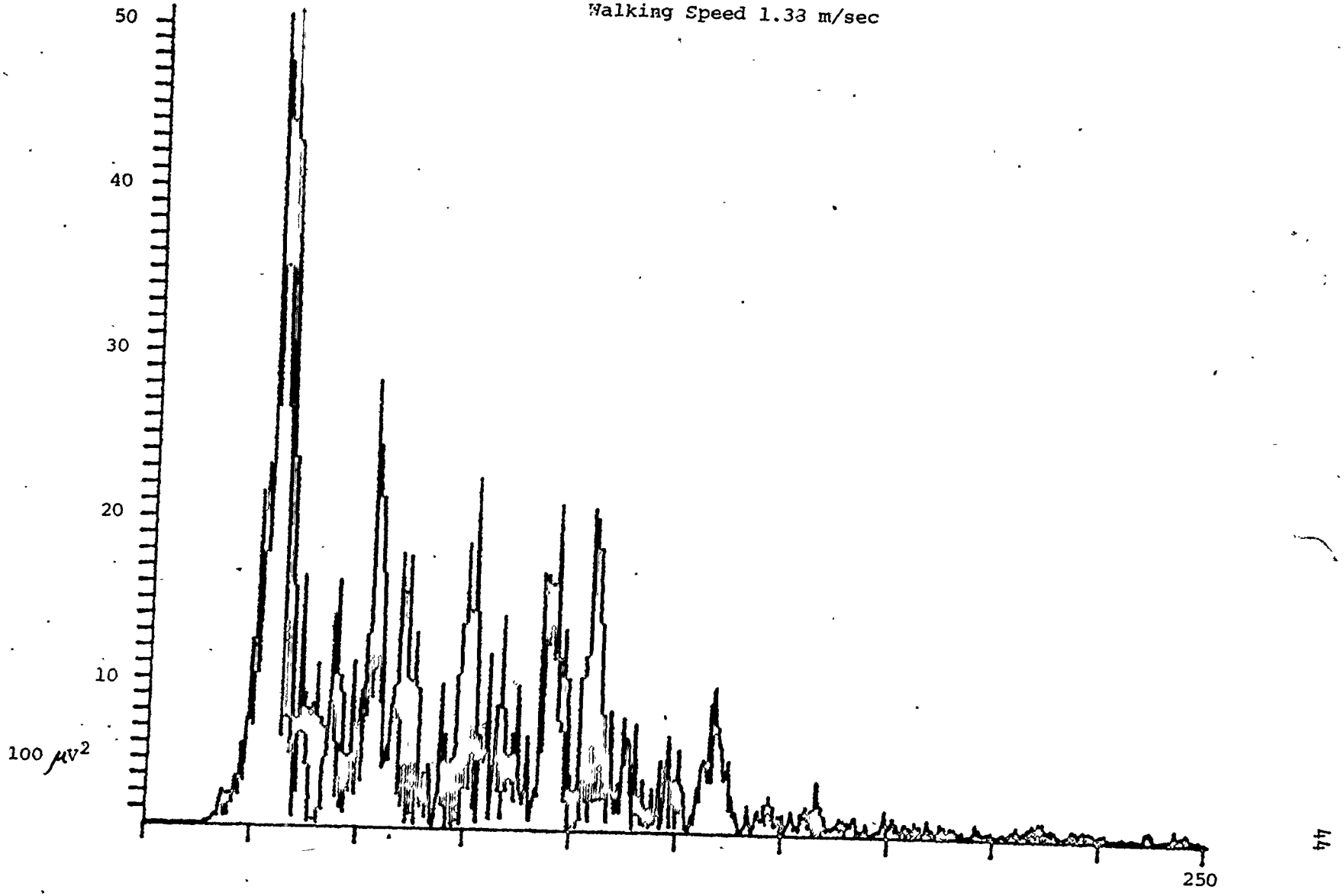


Figure 4.5 Power Spectrum for Raw EMG - Tibilias Anterior Muscle
Walking Speed 1.38 m/sec

Figure 4.6 Power Spectrum for Raw EMG - Gastrocnemius Muscle
Walking Speed 1.33 m/sec



The amplitudes were calculated with respect to the skin. All spectral components were considered in determining the median frequencies, that is, the calculation of the median frequencies was not band limited.

$$f_{med} = \frac{f_i * f_s / 1024 * i}{\text{Total Power}} \qquad 4\text{-}3$$

where f_i is the spectral power for coefficient i and f_s is the sampling frequency.

Table 4.1

Median Frequency, Bandwidth and Amplitude
of Maximum Fourier Coefficients for Spectra of
Quadriceps (Q) Hamstrings (H) Tibialis Anterior (TA)
and Gastrocnemius (GA) Muscles
500 Hz Sampling Rate, Walking Speed (WS) in metres/second
For Normal, Raw EMG

	MEDIAN FREQUENCY(Hz)			BAND WIDTH(Hz)			AMPLITUDE*		
WS	1.38	1.08	.86	1.38	1.08	.86	1.38	1.08	.86
Q	54	66	64	100	100	100	19	6	1
H	70	61	79	100	80	75	180	47	12
TA	91	92	82	210	135	135	180	100	63
GA	67	71	75	150	150	150	49	10	8

* 100 $volt^2$

There are no consistent shifts to higher or lower frequencies with change in walking speed but the median frequency for each muscle is different. Walking speed has no effect on the bandwidth of the spectra of the quadriceps and gastrocnemius but there is an increase in bandwidth with increased walking speed for the hamstrings and tibialis anterior. This reflects the increased activity in these muscles to control knee and ankle flexion at higher speeds.

The change in maximum amplitude with walking speed is quite pronounced with greater power in the spectra at the fastest walking speed. This analysis substantiates the use of a 500 Hz sampling rate for "raw" EMG.

It must be stressed that the results apply only to the subject chosen. Changes in bandwidth, median frequency and maximum amplitude may be due to the unique recruitment patterns of motor units for this subject.

4.2.2 Spectral Analysis of Preprocessed EMG

Raw EMG that is full wave rectified and low pass filtered is referred to as preprocessed EMG. Figures 4.7, 4.8, 4.9, and 4.10 are power spectra of preprocessed EMG for the four muscle groups for the subject's most comfortable speed of walk. The sampling rate was 200 Hz, so the Fourier coefficients for 100 Hz were calculated. Since there is no significant activity beyond 20 Hz only these coefficients are displayed. The features of interest for the spectra of these muscles for three walking speeds are summarized in Table 4.2 in the column labeled "PRE".

Figure 4.7 Power Spectrum for Preprocessed EMG Quadriceps Muscle
Walking Speed 1.38 m/sec

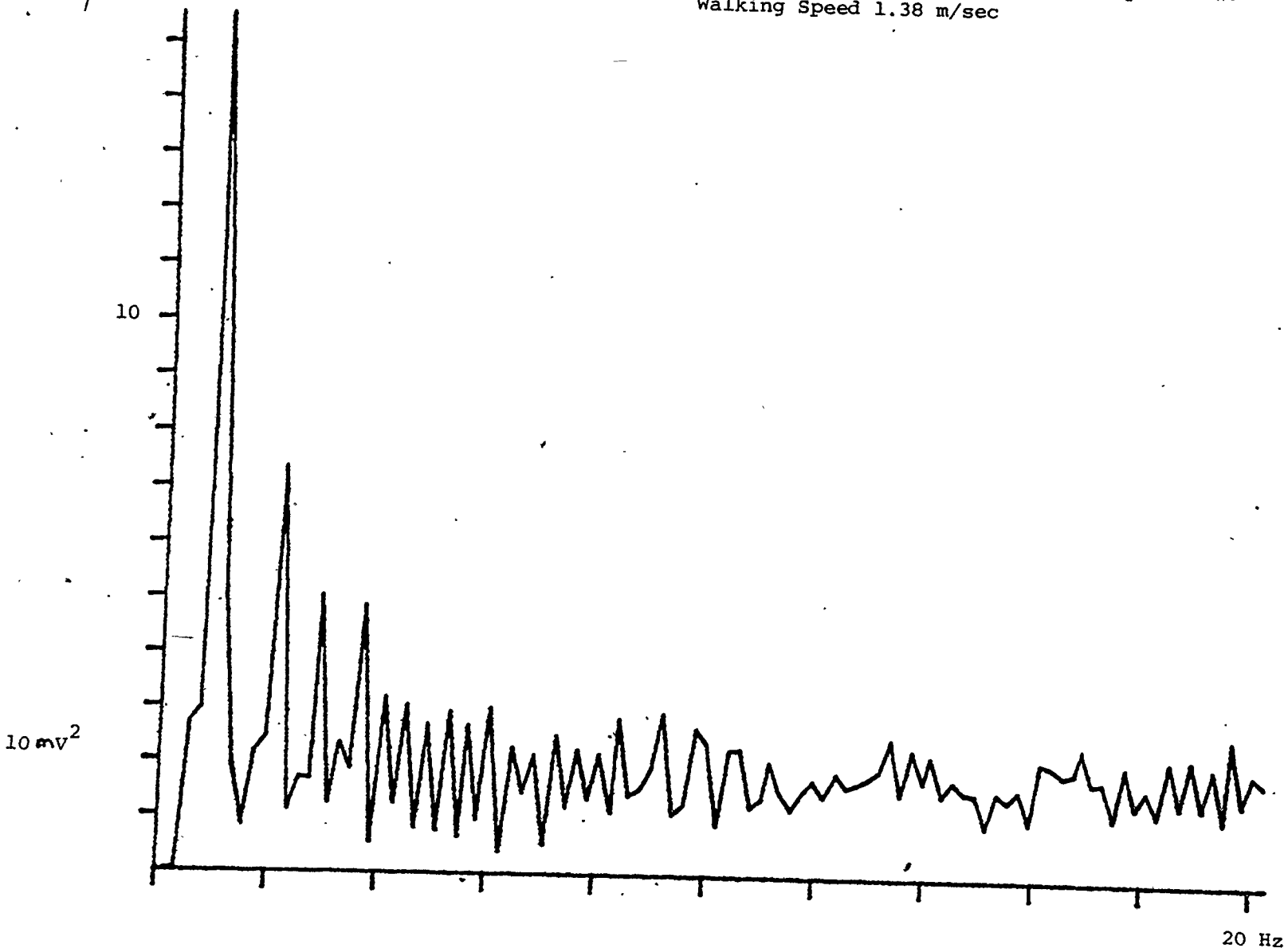


Figure 4.8 Power Spectrum for Preprocessed EMG Hamstrings Muscle
Walking Speed 1.38 m/sec

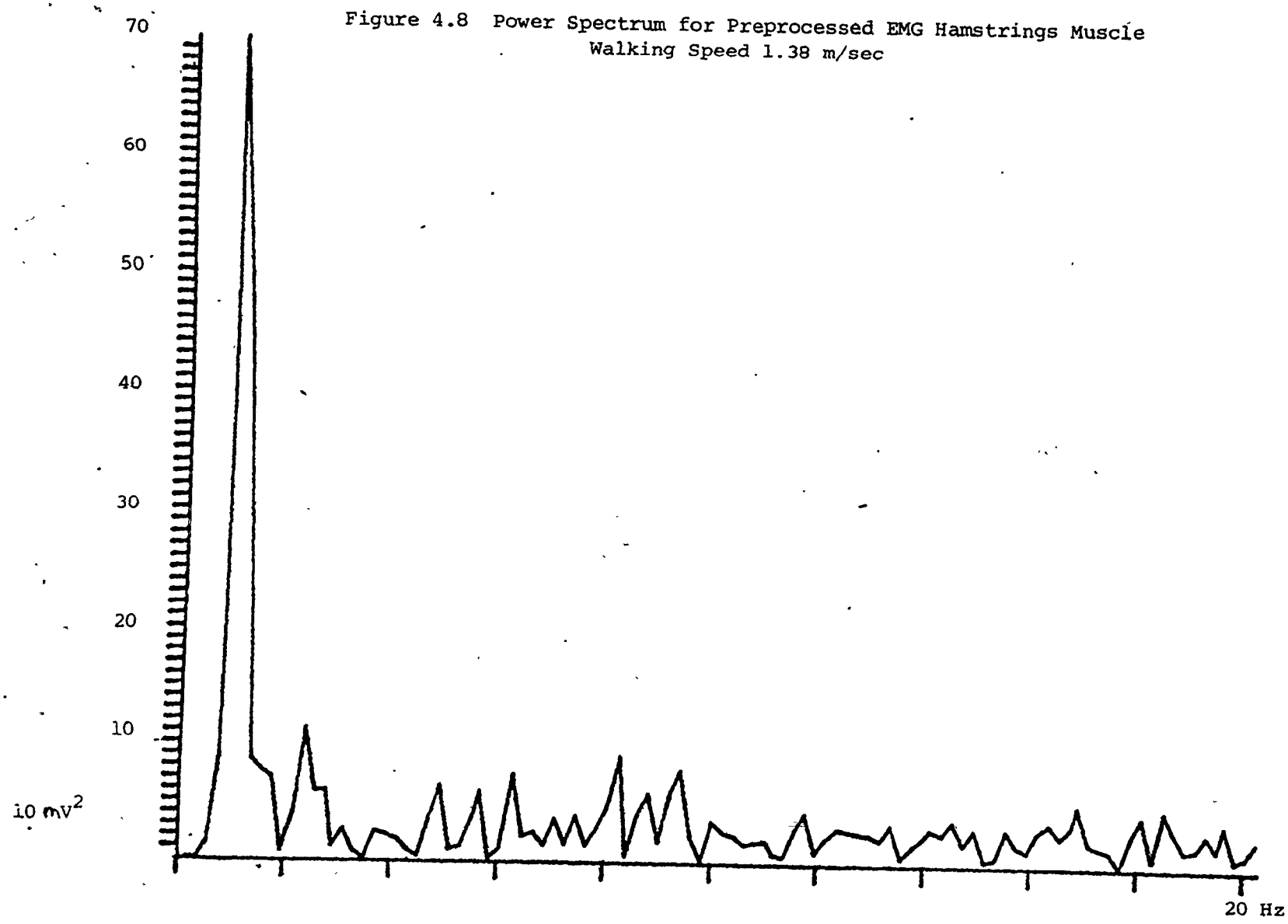


Figure 4.9 Power Spectrum for Preprocessed EMG Tibilias Anterior Muscle
Walking Speed 1.38 m/sec

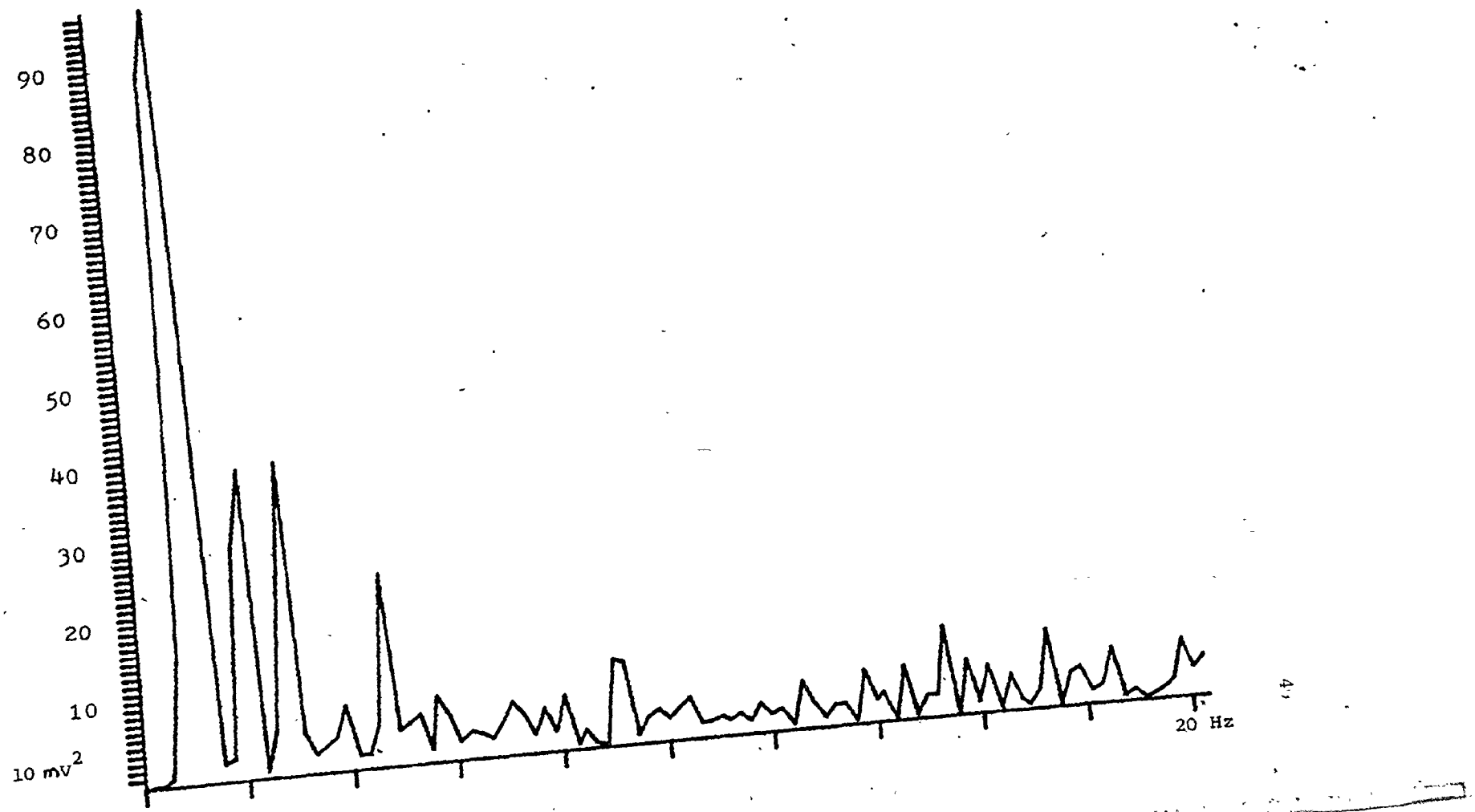
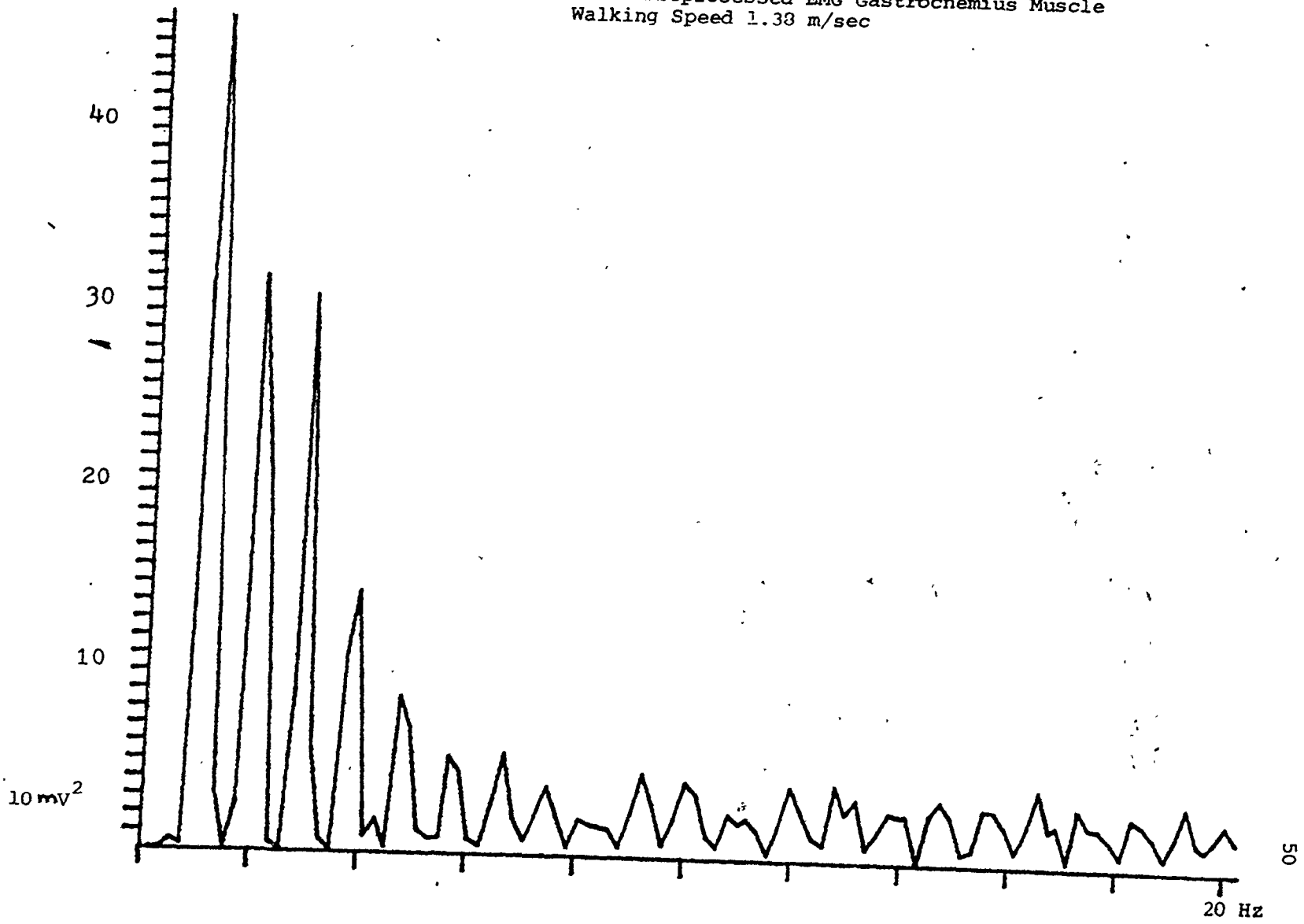


Figure 4.10 Power Spectrum for Preprocessed EMG Gastrocnemius Muscle
Walking Speed 1.38 m/sec



	1.38 m/sec									1.08 m/sec									.86 m/sec								
	PRE			LPF			AVG			PRE			LPF			AVG			PRE			LPF			AVG		
	MF	BW	A	MF	BW	A	MF	BW	A	MF	BW	A	MF	BW	A	MF	BW	A	MF	BW	A	MF	BW	A	MF	BW	A
Q	48	4	67.0	4	4	8.7	6	4	11.6	11	4	5.8	2	4	2.0	2	4	2.1	12	4	4.5	2	4	1.5	2	4	1.7
H	41	3	70.5	4	3	10.9	4	3	13.7	11	3	13.5	2	3	4.6	2	3	4.8	12	2	13.3	2	2	4.4	2	2	4.7
TA	34	5	75.9	3	5	14.6	3	5	17.2	11	6	27.6	2	6	9.8	2	6	10.1	11	5	27.9	2	5	10.0	2	5	10.4
GA	41	5	69.7	3	6	10.9	4	6	13.6	8	8	15.5	2	8	6.2	2	8	6.3	8	4	9.9	2	4	3.7	2	4	3.9

TABLE 4.2

Median Frequency, Bandwidth and Amplitude of Maximum Fourier Coefficients For the Spectra of Quadriceps (Q), Hamstrings (H), Tibialis Anterior (TA) and Gastrocnemius (GA) Muscles.

200 Hz Sampling Rate, Walking Speed in Metres per Second for Normal Rectified EMG

PRE - preprocessed EMG

LPF - low pass filtered EMG (9 Hz)

AVG - mid-point moving window average EMG (60 msec window)

MF - median frequency

BW - band width

A - amplitude (10 mvolt^2)

Immediately apparent is the higher median frequency for all muscles at the fastest speed of walk. In the preprocessed EMG, median frequencies (rounded to one significant figure) are greater than the bandwidth because of the criterion for selecting bandwidth (Section 4.2.1). This result shows that the power is distributed throughout the 100 Hz frequency band at this subject's fastest walking speed.

The 200 Hz sampling rate for preprocessed EMG is more than sufficient if 3 db bandwidths are considered. The increased power at higher frequencies may be a result of greater discharge rates of motor units at faster speeds of walking. Some of the power at higher frequencies in all the spectra obtained is a result of electrode and instrument noise. To remove the unwanted higher frequency components of the preprocessed signal, the low pass cut off frequency of the hardware filter should be reduced, allowing lower sampling rates.

The bandwidth occupies the 2-8 Hz range clearly demonstrating that most of the power lies in the low frequencies. This result justifies the selection of a 9 Hz cutoff for further processing of preprocessed EMG, if only phase information is to be obtained. The 6 Hz cutoff used by Arsenault and Winter (1980) may reduce the information content of the signal relating to variations in phasic muscle activity. Repeatable EMG in normals reported by these authors may be a result of their filter cutoff frequency .

The amplitude of the maximum Fourier coefficient increases with increased walking speed because more muscle fibres are active.

4.3 Mid-Point Moving Window Average Filter

Moving averages are used to eliminate randomness in a time series by averaging several data points together so that the record is smoothed. The averaging is done over a constant number of data points called a "window". The term "moving average" is used because for each observation the oldest observation in the "window" is dropped and a new average is computed.

Where the sample window is out of data range, that is, at the beginning and end of the record, the mirrored image of the first or last data points is used.

Papoulis (1977, pp. 343-4) showed that mere averaging can lead to a satisfactory estimate of a random signal if the noise is sufficiently small. The preprocessing described in Section 3.2 has effectively reduced signal noise. It is now necessary to smooth the data further to extract the phasic representation of the muscle activity.

A mid-point moving window average filter was implemented by a simple Fortran subroutine according to the equation:

$$y_i = \sum_{j=i}^{k} x_{i-k/2+j} \quad \text{for } i = 1,2 \ldots n \qquad 4.4$$

where there are n data points with k points in the window.

4.3.1 Window Width

Although smoothing EMG signal records with a mid-point moving window average filter is an accepted method of analysis, there is no agreement as to what constitutes good technique. Windows vary from 50 msecs (Bruce, 1977) to 100 msecs and larger (Hershler, 1977). Selection of window width, where documented, describes choices based on subjective criteria or simply by intuition.

The efficacy of window widths of 20 to 100 msecs was assessed. Figure 4.11 graphs the change in the magnitude of the coefficient for the cross correlation of EMG records of the left quadriceps and hamstrings of a normal subject, versus window width in milliseconds. The value of R(0) increases as window width, but the change in R(0) is least at a 60 msec window. We expect that, as the window width approaches the limit of the record, R(0) will tend to 1. The stride to stride standard deviation for the two signals is stable at 60 msecs. This is important for stationarity. The criterion selected for the choice of optimum averaging window was minimum change in the cross correlation coefficient at time zero, with change in window width. On the basis of this criterion, a window of 60 msecs was selected. At a sampling period of 5 msecs, the window constitutes an average of twelve samples.

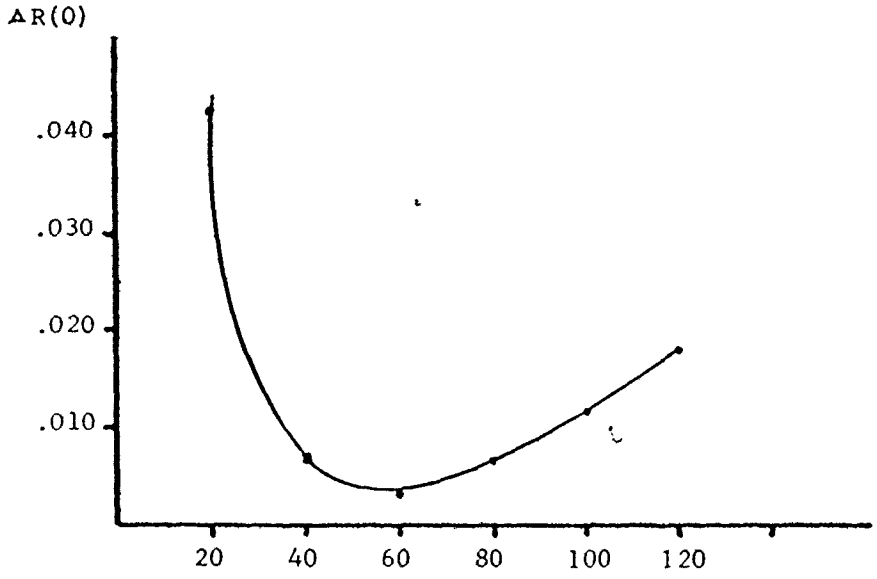


Figure 4.11 ΔR(0) Versus Window Width

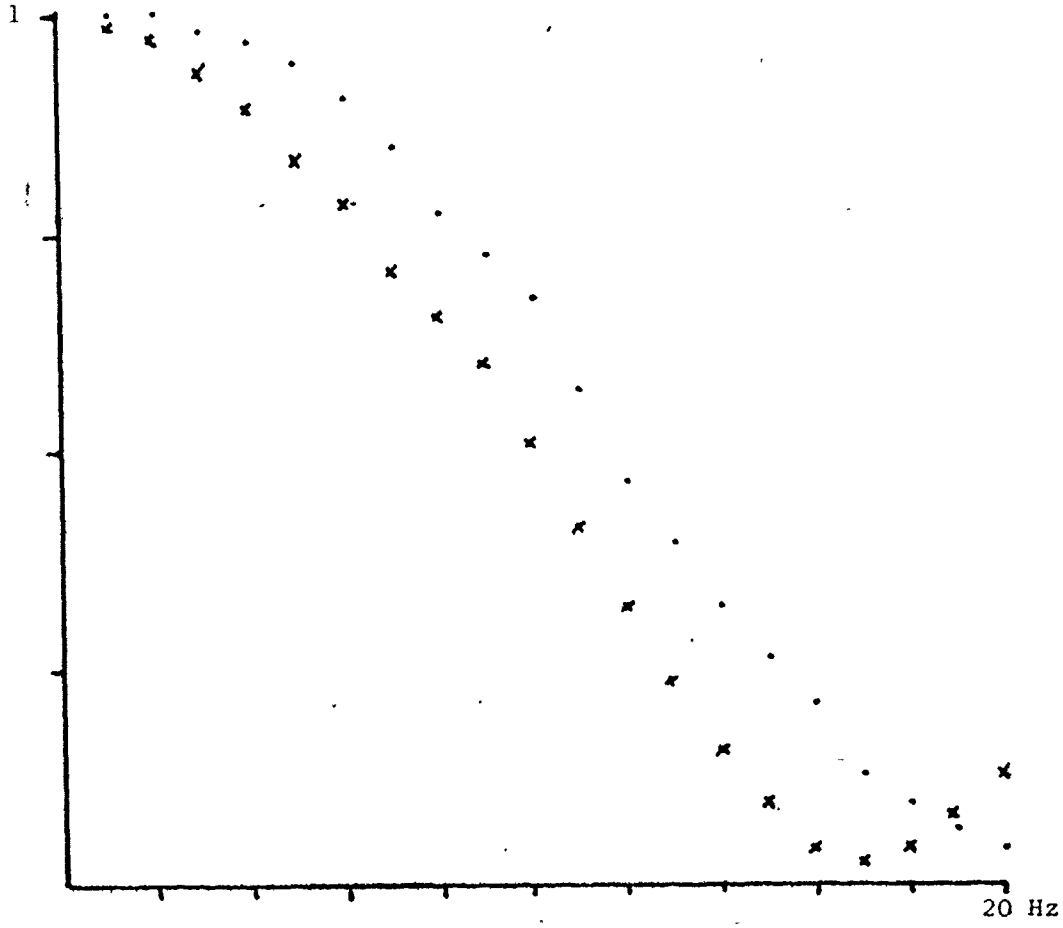


Figure 4.12 Amplitude Response for Two Smoothing Filters

xxx Moving Window Average

... LPF

4.3.2 Amplitude and Phase Response of the Mid-Point Moving Window Average Filter

A sine wave test pattern of a variable number of cycles was created in order to confirm the operation of this filter. The amplitude and phase responses for both this filter and the digital low pass filter of section 4.4 are shown in Figures 4.12, 4.13 respectively. The effective cutoff is about 7 Hz with a 24 dB per octave roll off. The appearance of the lobe above 17 Hz is expected because the Fourier transform of a rectangular function (window) is the $sinc^2$ function (Figure 4.14). In general, this filter has a near constant phase characteristic because mid-point averaging is used and the phase shift should be close to zero.

4.4 Digital Low Pass Filter

A simple second order Butterworth digital filter, designed by Abdel-Azim (1979) after Rader and Gold (1967) was available in a Fortran subroutine. Due to a phase lag of 90° introduced by the filter, data were filtered a second time in reverse order to cancel the lag. The recursive filter equation is

$$y(nT) = 0.0036x(nT) + 0.0072x(nT-T) + 0.0036x(nT-2T) + 1.823y(nT-T) - 0.837y(nT-2T) \qquad 4.5$$

where x and y are the input and output signal data points respectively and T is the sample period.

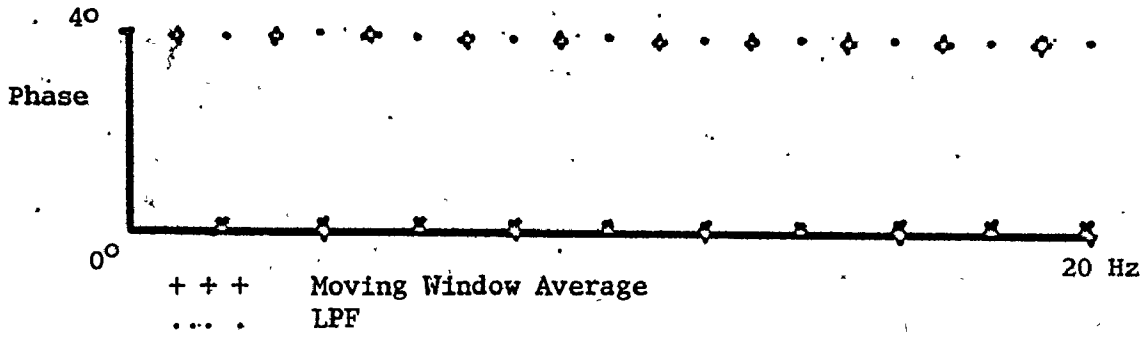


Figure 4.13 Phase Response for Two Smoothing Filters

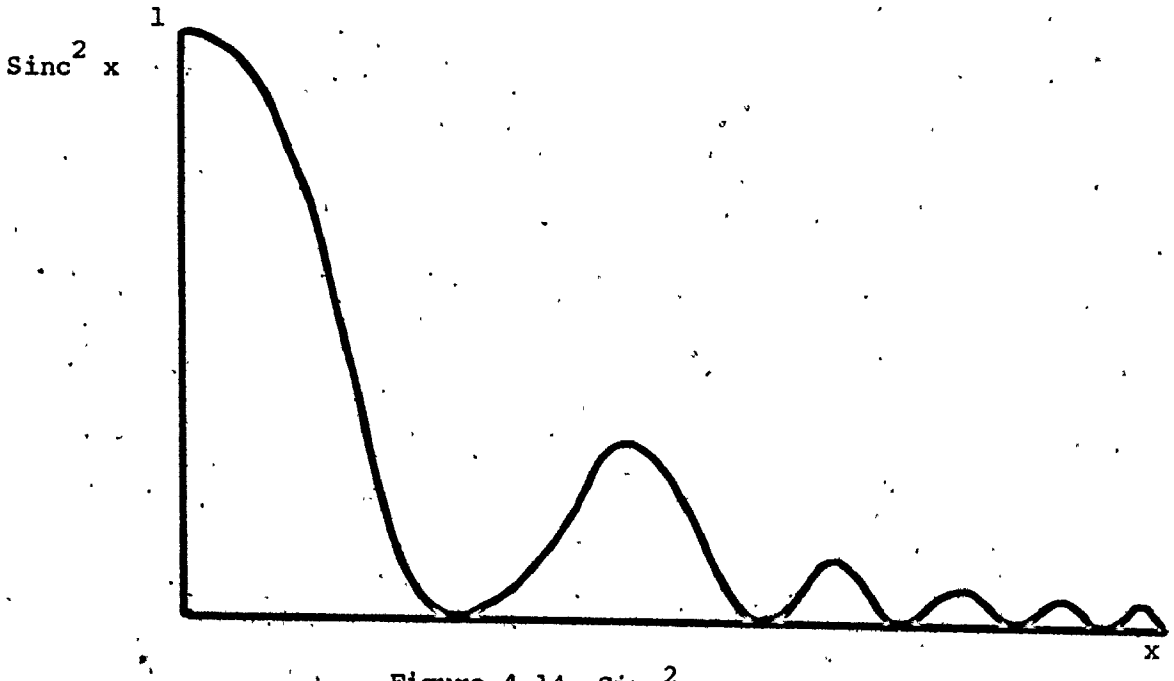


Figure 4.14 $Sinc^2$ Function

4.4.1 Amplitude and Phase Response of a Digital Low Pass Filter

The sine wave test pattern described in Section 4.3.1 was utilized to confirm the validity of this filter. Figures 4.12 and 4.13 show the resulting amplitude and phase response of the Azim filter, as well as the mid-point moving window average filter. The low pass filter operates effectively with a 9 Hz cutoff and a roll off of 24 db per octave. This is close agreement with Azim's plot.

The phase response shows a lead of approximately four degrees for all frequencies tested (0-20 cycles). This is close agreement with the predicted 0° lag after two filter passes.

4.5 Spectral Analysis of Filtered EMG

At medium and slow walking speed there is virtually no difference in the median frequency, bandwith and maximum amplitude of the spectra of signals processed by low pass or mid-point moving window average filtering. At fast speed of walk there are minor differences in median frequency and bandwidth for the quadriceps and hamstrings muscles. The low pass filter attenuates the maximum amplitudes to a greater extent than the averaging filter. The difference in the characteristics of the two filters is more apparent at a fast speed of walk.

4.6 Conclusion

The spectral analysis described in this chapter shows that the phasic nature of EMG results in a different power spectra for each muscle, at each walking speed. Kwatny et al (1970) reported the results of power spectrum analysis for constant isometric contraction. Variations in the power spectra for the raw EMG of muscles in the hands of several individuals were observed. Each subject had a characteristic region of high energy from 25-150 Hz which overlapped the high energy regions of other subjects. It was further shown that during fatigue, the region of high energy shifted to lower frequencies. The mean frequency for all subjects, rested and fatigued, occupied a narrow range of 48-58 Hz. Although the spectra of raw EMG collected from muscles in the lower limb during walking cannot be compared to these results, it should be noted that for the latter case the median frequency occupies the range 54-92 Hz. Spectral analyses of EMG for isometric and dynamic contractions are different. The results for isometric contraction cannot be applied to locomotion studies.

The sampling rate of 200 Hz for rectified averaged EMG was confirmed since the frequencies of interest do not exceed 100 Hz.

The digital low pass filter with a 9 Hz cutoff, and the mid-point moving window average filter with a 60 msec window were shown to operate with virtually the same effect. Since the signal of interest has a narrow band (0-8 Hz), the side lobes which appear above 20 Hz in the transfer characteristic of the averaging filter do not affect the results.

The digital low pass filter was selected for processing EMG on the basis of its superior transfer characteristics.

CHAPTER 5

QUANTIFICATION OF GAIT USING THE CROSS CORRELATION FUNCTION

5.1 Introduction

In engineering analysis the cross correlation function has proven a useful characteristic of random signals. The cross correlation function may be applied to the measurement of the time delay of the features of one signal relative to another. The cross correlation coefficient quantifies the extent to which two sets of data are related and has been found to be a sensitive measure of differences in phase and shape between signals.

Since the cross correlation function is self normalizing it is independent of the amplitudes of the signals. Figure 5.1 diagrams a flow chart for the digital computation of the cross correlation function. Gandy et al (1980) employed the correlation coefficient to determine repeatability of upper limb EMG patterns during a prescribed movement for both normal and hemiplegic subjects. This chapter investigates the use of the cross correlation function to provide an indicator of the phasic interdependence between pairs of muscles in the lower limb.

5.2 The Cross Correlation Function

Consider the EMG activity in say, the quadriceps (X(t)) and hamstrings (Y(t)) muscles in Figure 5.2.

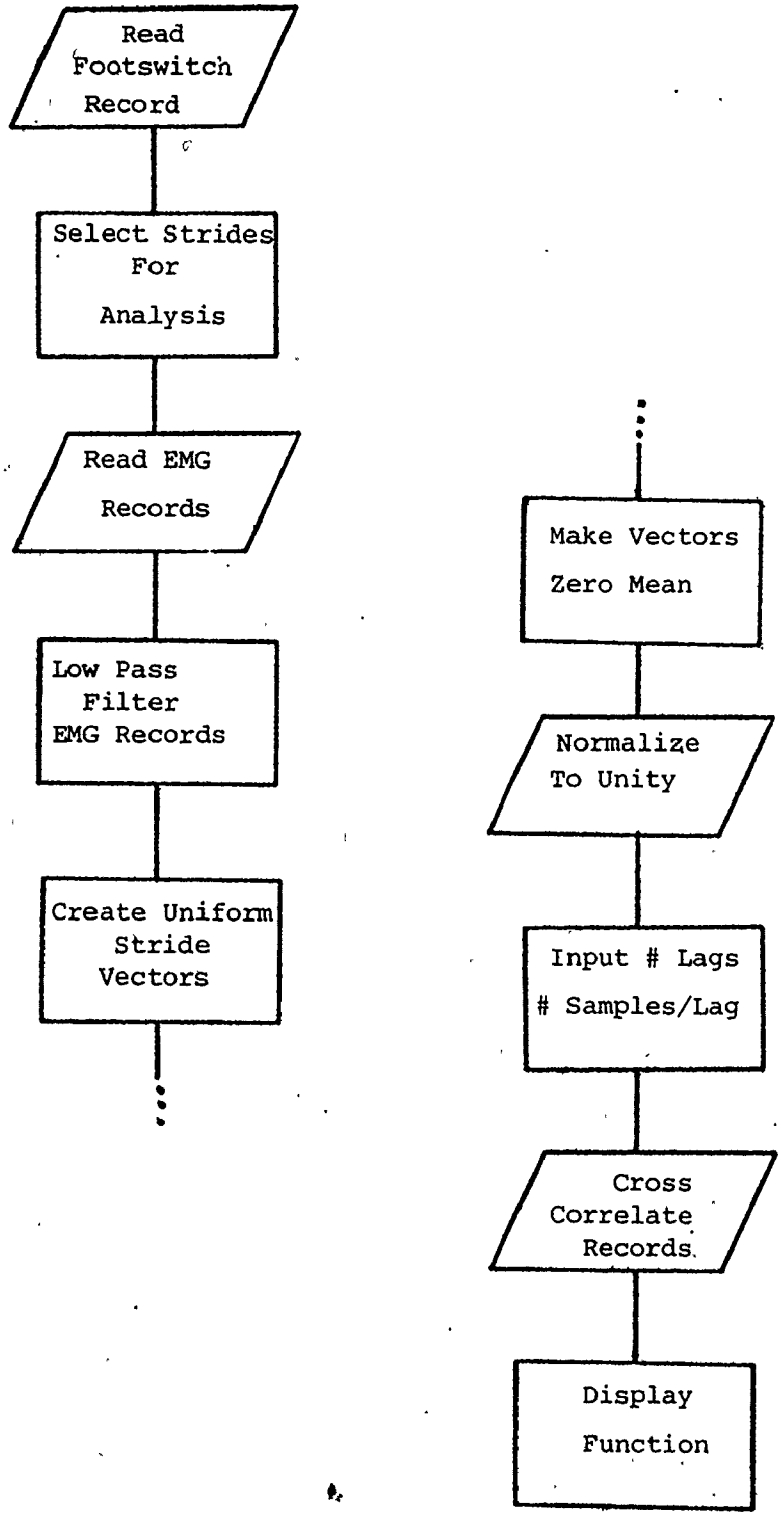
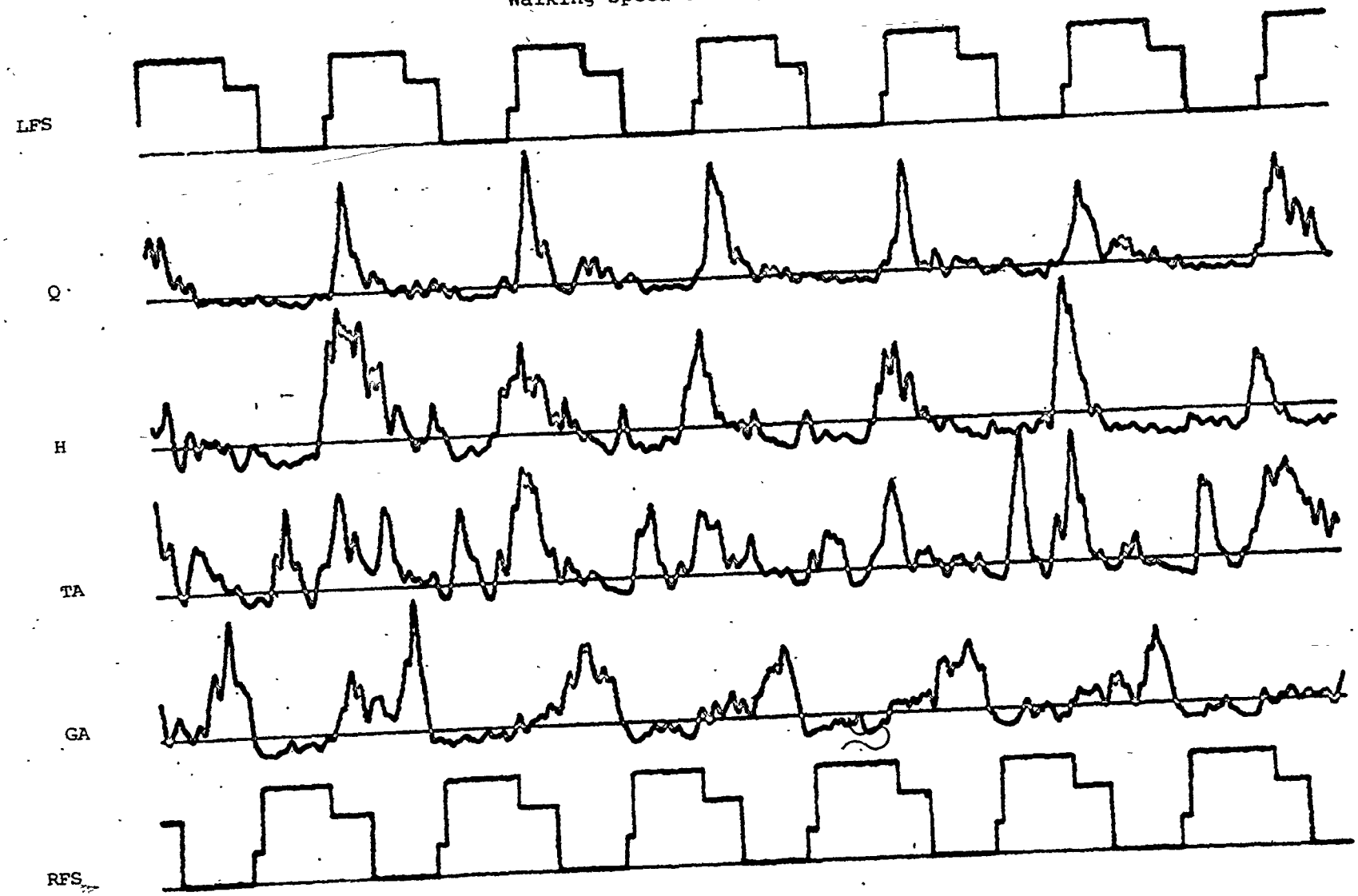


Figure 5.1 Digital Computation of the Cross Correlation Function

Figure 5.2. EMG and Footswitch Records for a Normal Subject
Walking Speed 0.66 m/sec



An estimate of the cross correlation of the signals X(t) at time t, and Y(t) at time $t+\tau$ is calculated by taking the average product of the two values over the observation time T. As T approaches infinity, the resulting average product approaches the cross correlation function.

$$R_{xy}(\tau) = \lim_{T\to\infty} T \int_{-\infty}^{\infty} x(t)y(t+\tau)\, dt \qquad 5\text{-}1$$

$\pm\tau$ is a positive or negative time delay in the Y record. The algorithm was implemented as a Fortran subroutine according to

$$R(X,Y) = \frac{E(XY) - E(X)\,E(Y)}{[(E(X^2)-E^2(X))(E(Y^2)-E^2(Y))]^{1/2}} \qquad 5\text{-}2$$

for $X = x_1, x_2, x_3 \ldots x_n$ (the X record)

$Y = y_1, y_2, y_3 \ldots x_n$ (the Y record)

E = expectation (mean) value.

The coefficient is normalized by means of the product of the standard deviations of X and Y.

The program was tested by using sine waves of known frequencies and then verifying the graphic display of the cross correlation function. The limits of the cross correlation function are +1 and -1 which implies perfect correlation with a phase difference of 0° and 180° respectively. Unrelated data, or similar data with phase differences of approximately 90° result in coefficients close to zero. Figure 5.3 illustrates the cross correlation function for the records in Figure 5.2. Sharp peaks indicate the existence of correlation beween X(t) and Y(t) at specific time delays.

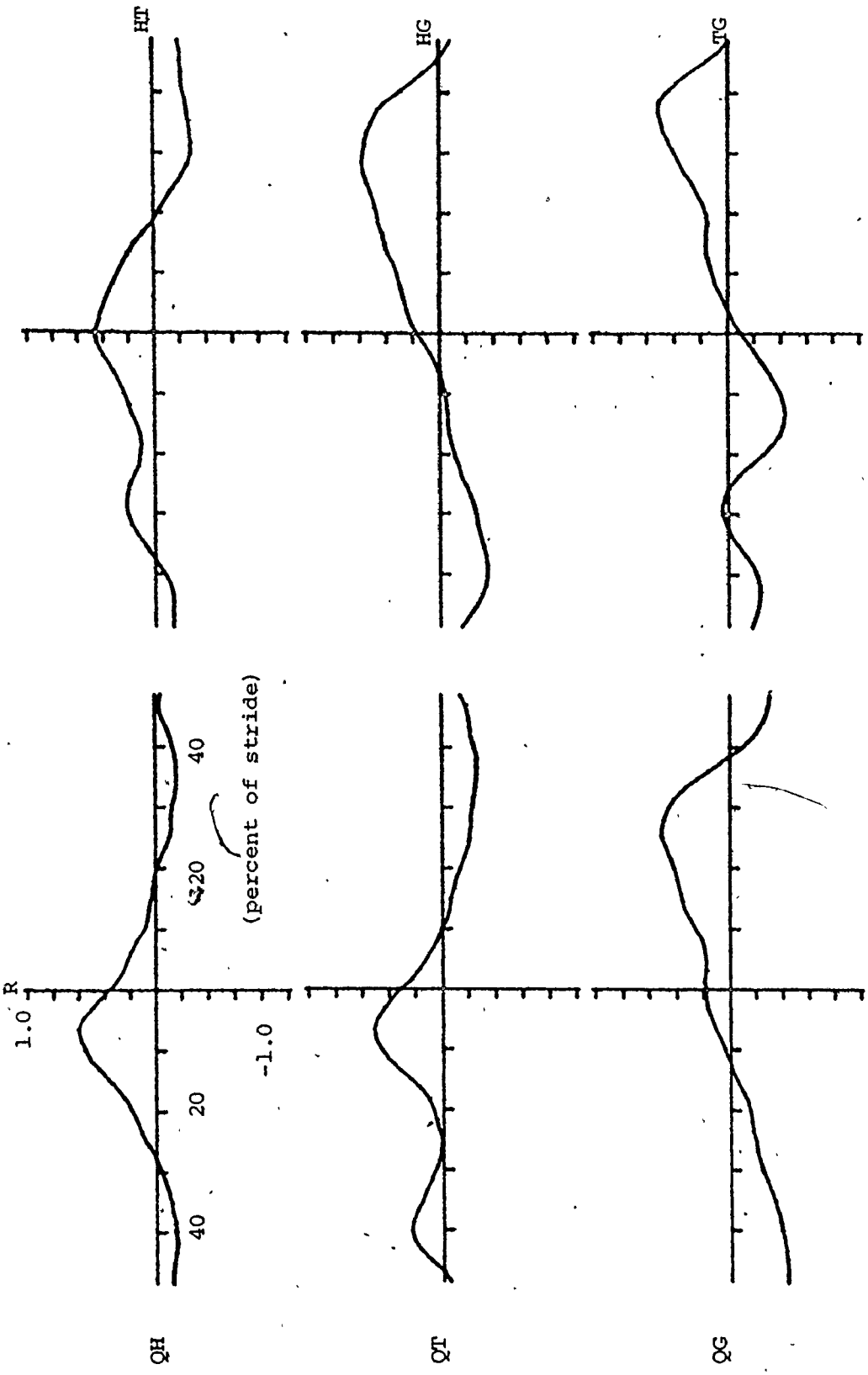


Figure 5.3 Cross Correlation Functions for the Records of Figure 5.2

5.3 Uniform Stride Period

For a realistic approximation to the cross correlation function, and to extract reliable estimates of the maximum cross correlation coefficient for positive and negative lags, we must consider the fact that the period varies from one stride to the next. The EMG signals recorded during locomoton are primarily event related, the event being a complete stride. The EMG signal should be redefined as a function of percent of stride period rather than as a function of time to accommodate varying stride periods.

Hershler (1977) averaged data from only those strides that were within 40 milliseconds duration of each other. He then added or deleted samples so that each stride had a uniform number of data points. A large number of strides were, by definition, deleted from the study.

Another approach to this problem was devised by Abdel-Azim (1979) for use in joint angle records. Using a variable length moving window and weighting coefficients, he redefined the data to a "normalized" walk cycle achieved by resampling the original cycle at a new sampling rate. All the normalized cycles were then averaged. This procedure would distort the original information unless the original sampling rate were much higher than twice the bandwidth of the signal. Since the signals were sampled at 500 Hz and the bandwidth of the information was 0-10 Hz, this procedure was satisfactory for Azim's work.

A different approach to creating strides of uniform period was implemented for this thesis. As a guide to the selection of the uniform number of samples to be created, the average number of samples in the strides selected for processing was calculated. If that number were 317, then the uniform number of samples to be created per stride would be 300.

A new sampling period was then calculated for each stride in the original record in order to resample the signal during that stride. The algorithm is listed in Appendix A.

The four EMG records from a walk were made into uniform stride records with the same periods and uniform numbers of samples.

There were run to run variations however. If, for example, in a very slow patient walk, the average number of real samples per stride were 394 or 415, then the number of uniform samples per stride would be 400.

The uniform stride records which were then independent of the stride duration and to some extent the speed of walk were cross correlated. The results of analysis for all subjects could then be compared regardless of walking speed.

5.3.1 Criteria for Selection of Strides

A condition for stationarity is a constant mean value. A means of insuring the stationarity of the EMG records to be cross correlated is to select only "steady state" strides for analysis.

"Steady state" strides are those which exhibit neither acceleration nor deceleration during the stride.

Only complete strides were selected, i.e., complete cycles from heel strike to heel strike. Strides in which the subject had stumbled or lost balance were not accepted.

5.4 Parameters Studied

From the cross correlation function for two EMG records, the following parameters were selected for discussion:

1) R(0), the cross correlation coefficient at zero lags. This value of R indicates the degree of co-contraction of the muscle pair over the whole stride sequence.

2) R+, the maximum value of the coefficient for positive time lags.

3) The time of occurence of the maximum R+ expressed as percent of stride

4) R-, the maximum value of the coefficient for negative time lags.

5) The time of occurence of the maximum R- expressed as percent of stride.

In addition, the cross correlation function curves were inspected for smoothness, number of peaks, and sharpness of peak. A sharp peak denotes a fast turn on/turn off of the muscles while a wide peak denotes a slow turn on/turn off of the muscles.

5.4.1 Results for a Normal Subject

The preprocessed, low pass filtered EMG records for the left side of a normal subject walking at a speed of 0.66 metres per second are shown in Figure 5.2.

The cross correlation curves for all pairs of records are shown in Figure 5.3. Each curve will be referred to by the initials of the cross correlated EMG records; for example QH refers to the cross correlation function for the quadriceps (X(t)) muscle record and hamstrings (Y(t)) muscle record.

In QH, the maximum activity of the hamstrings leads the maximum activity of the quadriceps by about 7% of the stride cycle. This occurs at heelstrike, where the quadriceps acts to stabilize the knee after the hamstrings have decelerated the forward swing of the lower limb.

The biphasic activity of the tibialis anterior is clearly revealed in the function for QT. The tibialis anterior has its first phase of activity at heelstrike to stabilize the ankle and again at the initiation of swing where the ankle must be dorsiflexed to prevent foot drag.

In QG, the maximum at positive lags indicates that the quadriceps' peak activity preceeds that of the gastrocnemius which is slightly active during foot flat to impart stability to the ankle, but has a burst of activity at toe off to help to push the leg off the ground for the swing phase.

This biphasic activity of the tibialis anterior is again reflected in the curve HT where the first phase of activity coincides

with that of the hamstrings muscle (peak at R(0)) at heelstrike.

The peak activity of the hamstrings leads that of the gastrocnemius (HG). The broad peak reflects the fact that one muscle is active for almost 50% of the stride. By inspecting QH which has a sharp peak at negative lags, we may deduce that it is the gastrocnemius which is on longer as it stabilizes the ankle during the entire time the foot is in contact with the floor.

Finally in TG, we see the influence of the biphasic tibialis anterior, and the sustained activity of the gastrocnemius muscle.

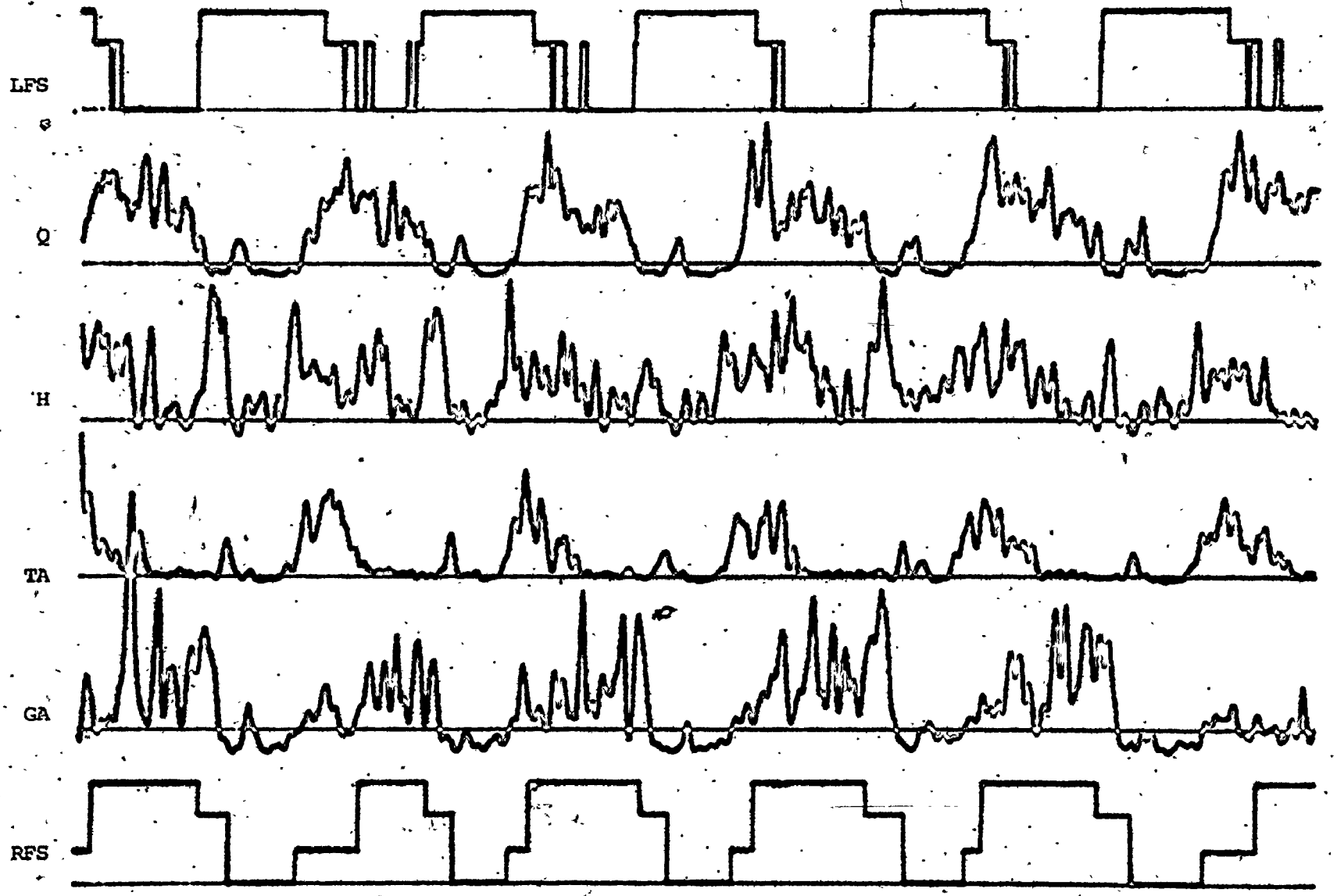
The EMG records in Figure 5.2 show the well defined phasic activity in all the muscles. The quadriceps, hamstrings and tibialis anterior achieve peak activity quickly after the initiation of a contraction. The biphasic activity of the tibialis anterior is easily distinguished and the gastrocnemius is active throughout the stance phase, peaking at toe-off.

5.4.2 Results for a Hemiplegic Subject

The EMG records for the unaffected and affected sides of a severely involved hemiplegic subject are shown in Figures 5.4 and 5.5 respectively. The cross correlation functions for these records are shown in Figures 5.6, and 5.7 respectively.

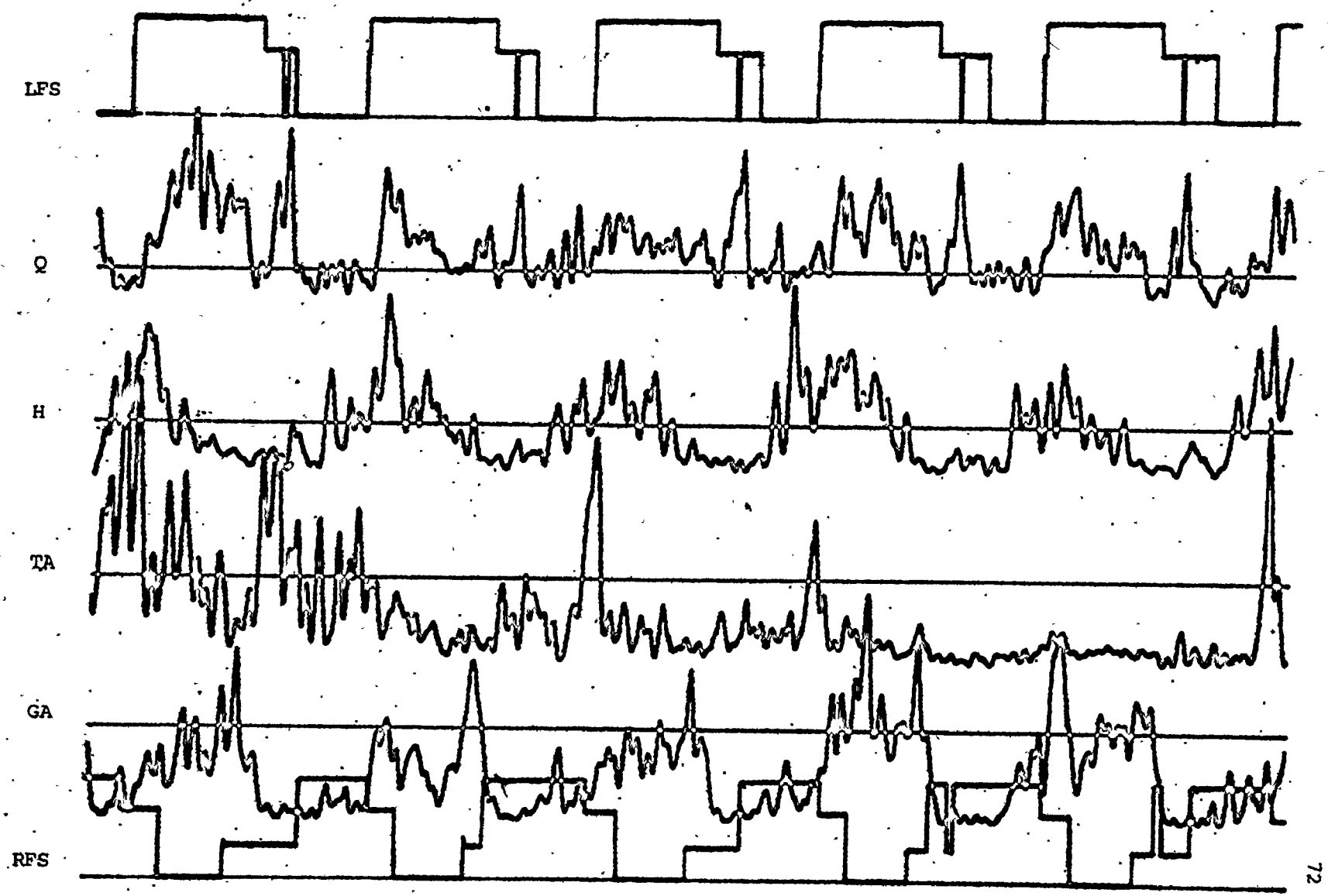
Inspection of the EMG records for the unaffected limb (Figure 5.6) shows some degree of control. The maximum values of R for QH and QT indicate a good phasic control.

Figure 5.4 EMG and Footswitch Records for a Severely Involved Hemiplegic Subject - Unaffected Limb



Walking Speed 0.34 m/sec

Figure 5.5 EMG and Footswitch Records for a Severely Involved Hemiplegic Subject - Affected Limb
Walking Speed 0.34 m/sec



A comparison of Figures 5.3 and 5.7 shows that the phasic relationship between most muscles for the unaffected limb is similar to normal control with the exception of hamstrings and gastrocnemius.

Although the functions for the unaffected side of the hemiplegic subject are not smooth, on the whole there are features in common with the functions for the normal subject. The values of R are comparable to the normal values, showing a good degree of co-contraction. The order of firing of the muscles is preserved with the exception of HG where a small degree (0.2) of correlation is observed at R(0).

An indication of the degree of loss of phasic control in the affected side is seen in the EMG of Figure 5.5. The muscles do not have well defined turn on and turn off phases. No second phase can be distinguished in the tibialis anterior.

In the cross correlation functions of Figures 5.6 and 5.7, the number of data samples in the records permitted time delays up to only 30% of a stride as compared to 50% of a stride processed for the cross correlations of Figure 5.3 (normal subject left side).

Inspection of Figure 5.7 reveals poorly defined cross correlation functions for the muscle records of the affected limb. The prolonged turn on of the muscles is reflected in the broad peaks of the curves. There is no suggestion of biphasic activity in the tibialis anterior in any of QT, HT or TG. Although the firing order of the quadriceps and hamstrings is preserved, the delay in the peak of activity is almost 18% whereas it is 7% in the normal and 9% in the unaffected limb.

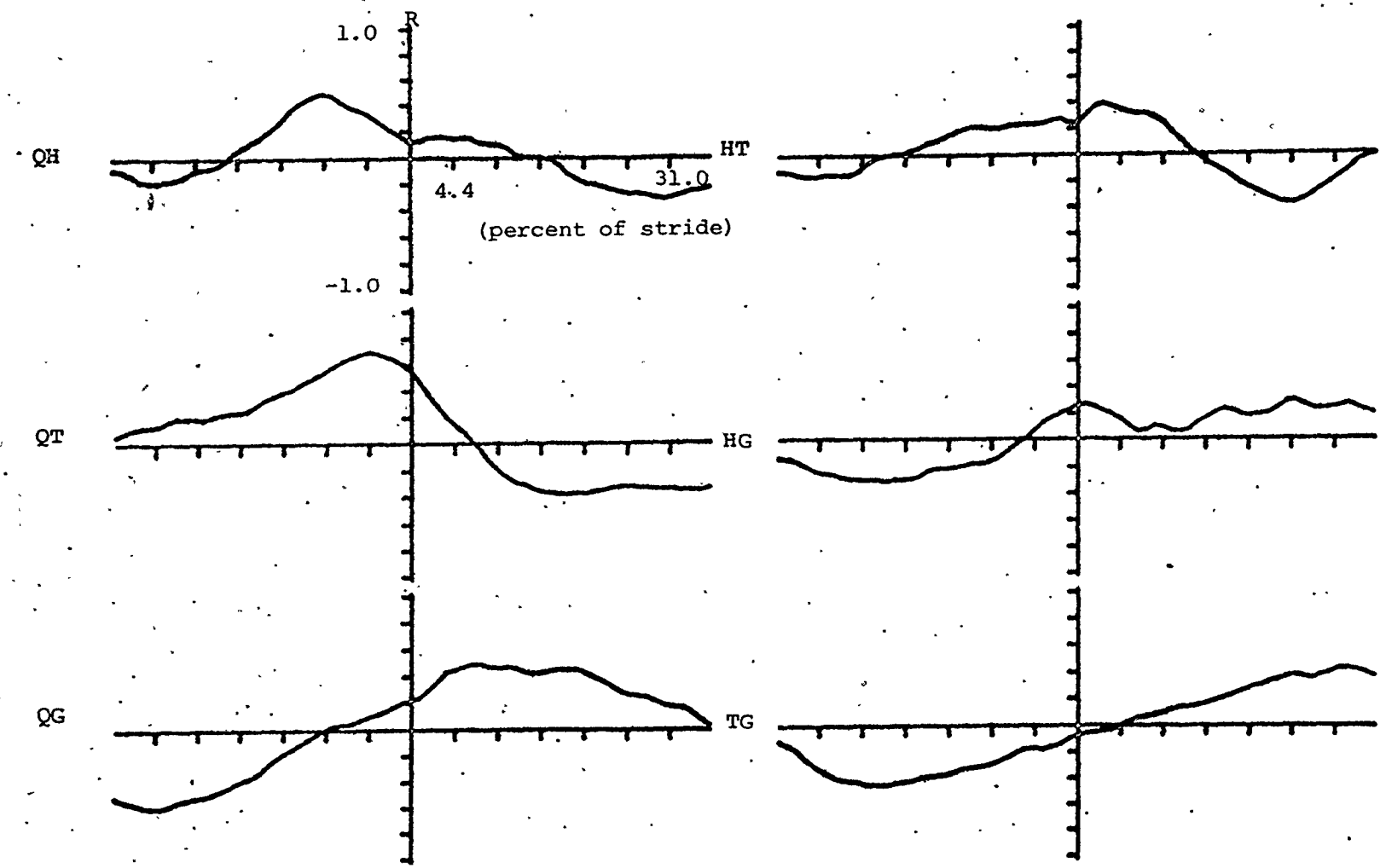


Figure 5.6 Cross Correlation Functions for the Records of Figure 5.4

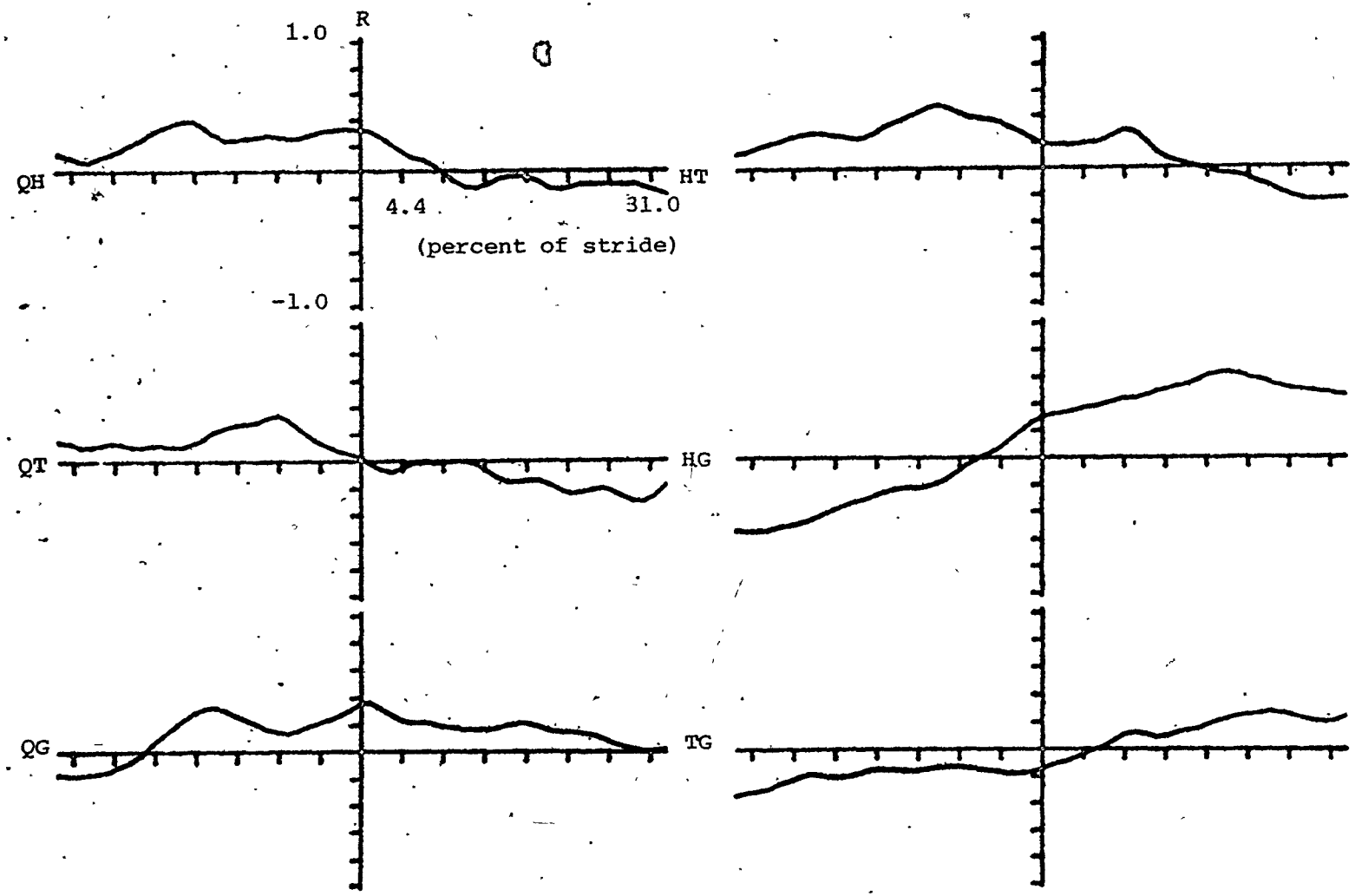


Figure 5.7 Cross Correlation Functions for the Records of Figure 5.5

The changes in balance, and stance and swing patterns due to altered neural control in the hemiplegic subject create abnormal demands on the unaffected side for compensation during locomotion. The extent of the compensation is reflected in the EMG records, and in the cross correlation functions.

5.5 A Comparison of Normal and Hemiplegic Results for the Population Studied

The use of the cross correlation function to analyse the inter-relationships of muscles of the lower limb during locomotion was applied to the EMG records of five normal subjects and fourteen hemiplegic subjects at different stages of recovery from stroke. The results for the parameters of interest (Section 5.4) are organized in Table 5.1.

In general, the values of R(0), R+ and R- for normals are larger than those of the hemiplegic subjects because normal records show more repeatable phasic activity. There are differences in the values for left and right side in normals due to the dominance of one side over the other. The differences are not significant. Dominance is readily observed in "handedness" for example.

TABLE 5.1

Means and Standard Deviations for the Results for
Normal (N) and Hemiplegic (H) Subjects
L - Left, R - Right, A - Affected, U - Unaffected Side

MUSCLES		R(0)	R+	+τ*	R-	-τ*
Q & H	N L	.38 ± .19	.21 ± .40	9 ± 14	.60 ± .16	8 ± 4
	R	.31 ± .19	.23 ± .40	22 ± 20	.47 ± .16	12 ± 4
	H A	.26 ± .19	.11 ± .43	16 ± 16	.39 ± .24	11 ± 11
	U	.19 ± .24	.16 ± .33	14 ± 14	.26 ± .33	12 ± 8
Q & T	N L	.39 ± .20	.29 ± .40	12 ± 12	.37 ± .25	11 ± 13
	R	.39 ± .15	.29 ± .30	8 ± 14	.45 ± .14	5 ± 8
	H A	-.06 ± .17	-.18 ± .32	22 ± 14	.03 ± .32	18 ± 15
	U	.32 ± .31	.34 ± .39	9 ± 11	.29 ± .40	9 ± 12
Q & G	N L	.07 ± .13	.57 ± .11	26 ± 8	-.17 ± .26	29 ± 9
	R	-.09 ± .08	.58 ± .16	34 ± 4	-.24 ± .36	35 ± 8
	H A	.25 ± .23	.29 ± .36	11 ± 12	.13 ± .37	15 ± 17
	U	.04 ± .16	.19 ± .37	21 ± 8	-.17 ± -.40	18 ± 11
H & T	N L	.30 ± .17	.39 ± .20	11 ± 10	.31 ± .14	6 ± 6
	R	.32 ± .11	.36 ± .24	9 ± 12	.22 ± .39	12 ± 17
	H A	-.06 ± .16	-.16 ± .26	20 ± 10	.10 ± .33	18 ± 12
	U	.06 ± .27	.10 ± .37	16 ± 14	.06 ± .31	12 ± 14
H & G	N L	-.12 ± .20	.42 ± .37	26 ± 16	-.10 ± .35	26 ± 16
	R	-.11 ± .21	.34 ± .30	29 ± 18	-.06 ± .46	25 ± 10
	H A	.16 ± .18	.38 ± .21	20 ± 10	-.07 ± .39	16 ± 13
	U	.21 ± .21	.23 ± .28	13 ± 13	.03 ± .39	14 ± 12
T & G	N L	-.00 ± .16	.31 ± .27	28 ± 11	-.02 ± .28	27 ± 18
	R	-.09 ± .24	.39 ± .28	28 ± 15	-.15 ± .42	19 ± 18
	H A	-.17 ± .20	-.06 ± .34	18 ± 16	-.15 ± .33	14 ± 13
	U	.00 ± .13	.22 ± .34	25 ± 9	-.12 ± .35	21 ± 14

* Expressed as % of stride

There are wide variations in the values of R for the affected and unaffected sides of the hemiplegic group. R(0) and R+ values are very similar in QH, QT and HT. It is apparent from inspecting Figure 5.3 that the maximumm R+ is simply the next data point to R(0). In QG, HG and TG, a true maximum for positive lags is determined.

The results tabulated for + and - , that is, the time of delay between the maximum activity of the two muscles expressed as percent of stride, are inconclusive because the data for cross correlation were processed with window periods from 28% to 50% of stride.

5.5.1 Regression Analysis

In order to determine the probability that a linear relationship exists for the values of R(0), R+ and R- with respect to walking speed, these data were further processed by linear regression analysis of each R with speed of walk. Only the R parameters for QH were considered. This analysis is a useful digression, not a study in itself. If there were no correlation between these quantities, then the analysis should yield a horizontal straight line (Bevington, 1969, pp. 119-122) with a slope of zero or a random scatter of points. The regression lines are plotted in Figures 5.8 to 5.13 inclusive. The values for R(0) and R- are tightly grouped, while the values for R+ are not. This is not surprising since the hamstrings activity leads the quadriceps activity. In all cases for normals, a positive correlation exists between walking speed and the

Figure 5.8 Regression Analysis Normal Population: R(0) versus Speed of Walk

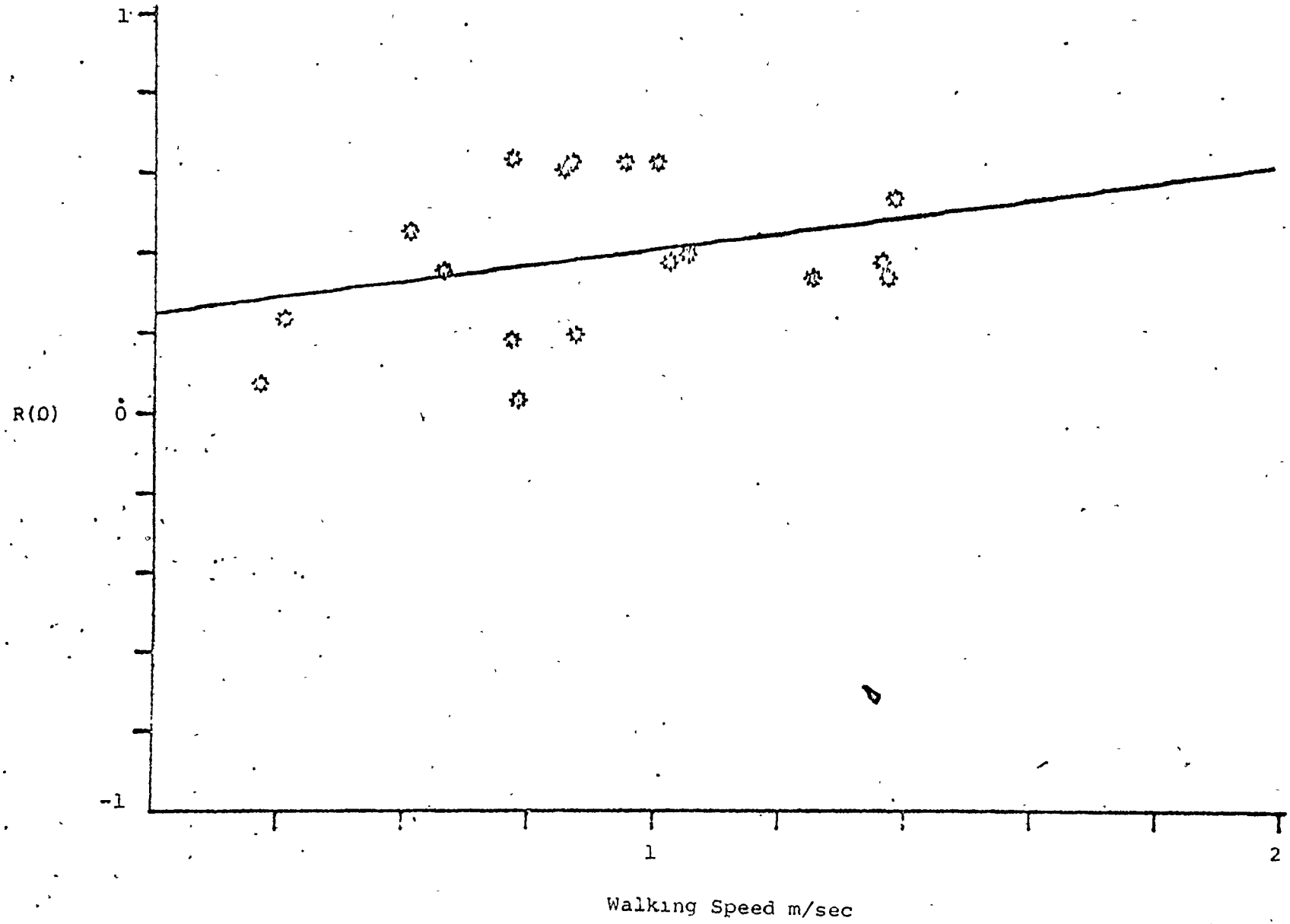


Figure 5.9 Regression Analysis Hemiplegic Population: R(0) Versus Speed of Walk

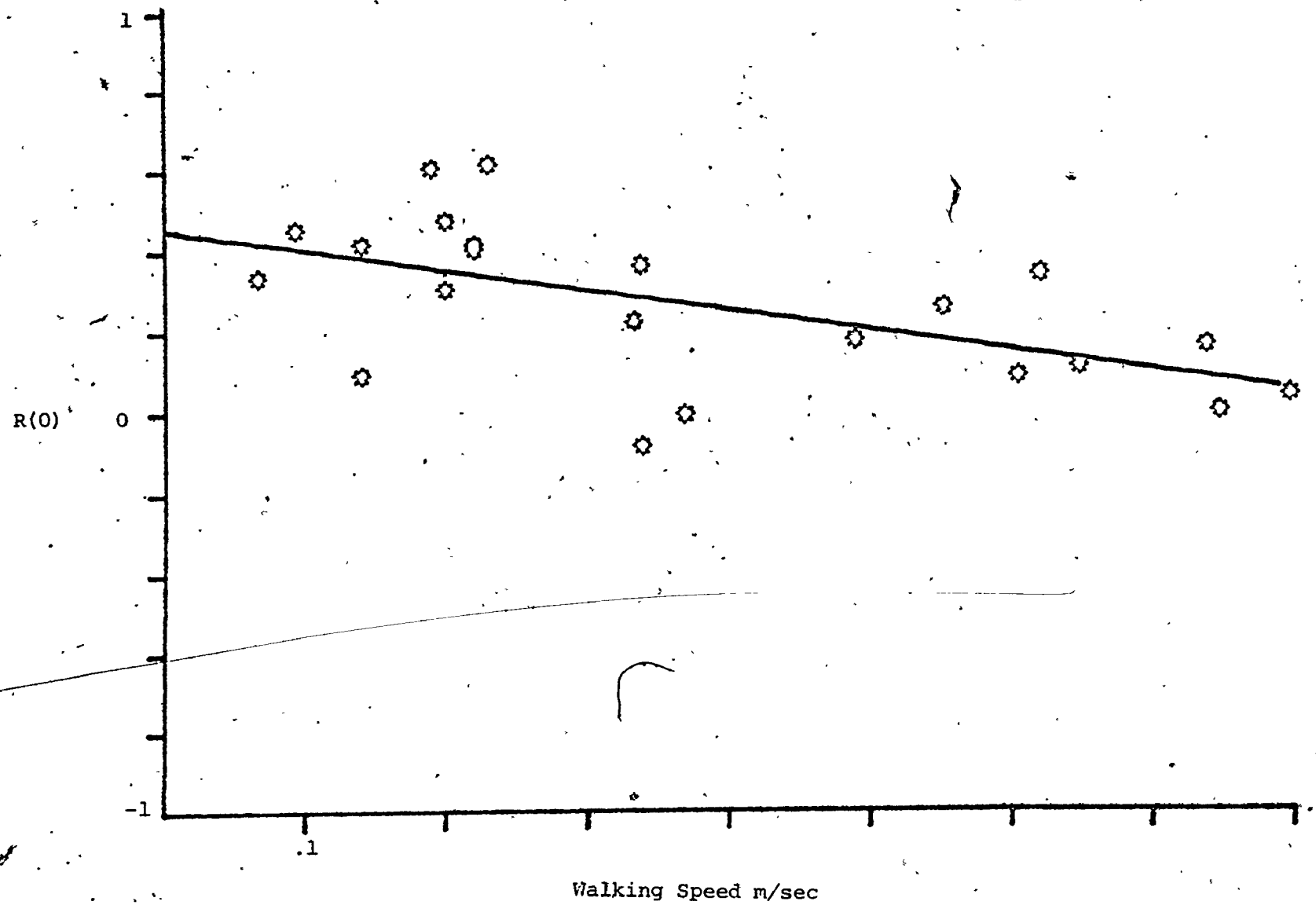


Figure 5.10 Regression Analysis Normal Population: R+ Versus Speed of Walk

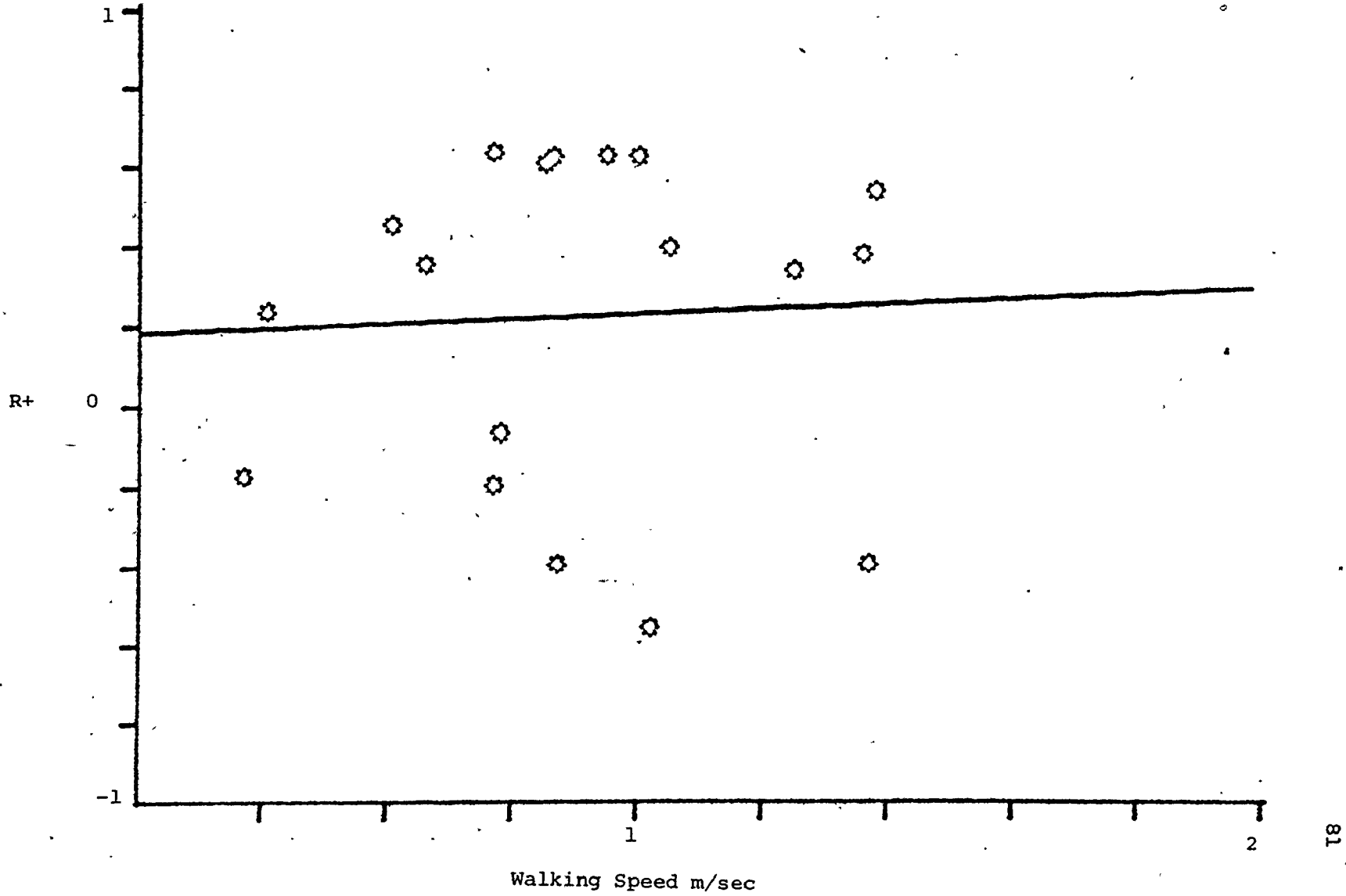


Figure 5.11 Regression Analysis Hemiplegic Population: R+ Versus Speed of Walk

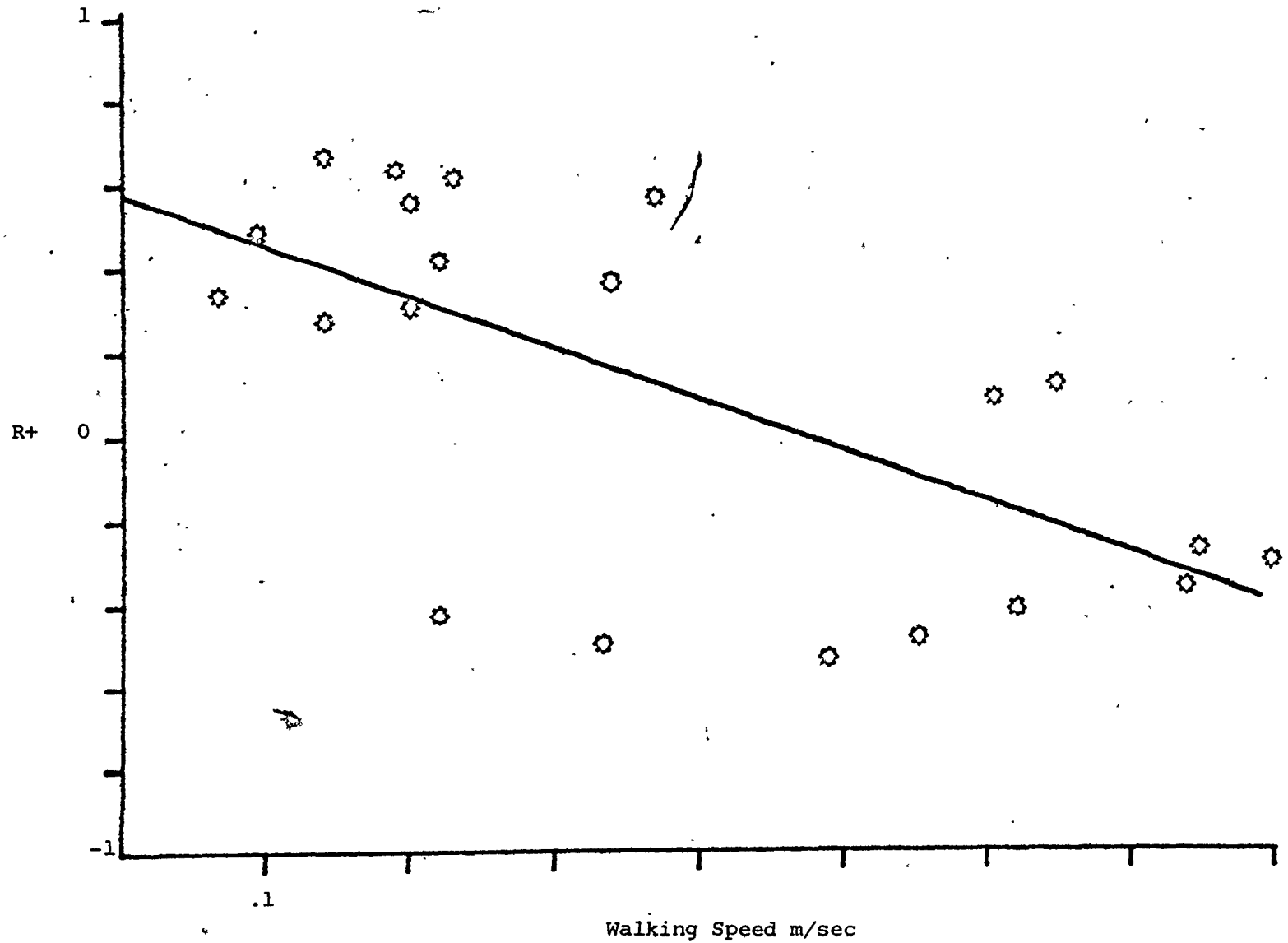


Figure 5.12 Regression Analysis Normal Population: R- Versus Speed of Walk

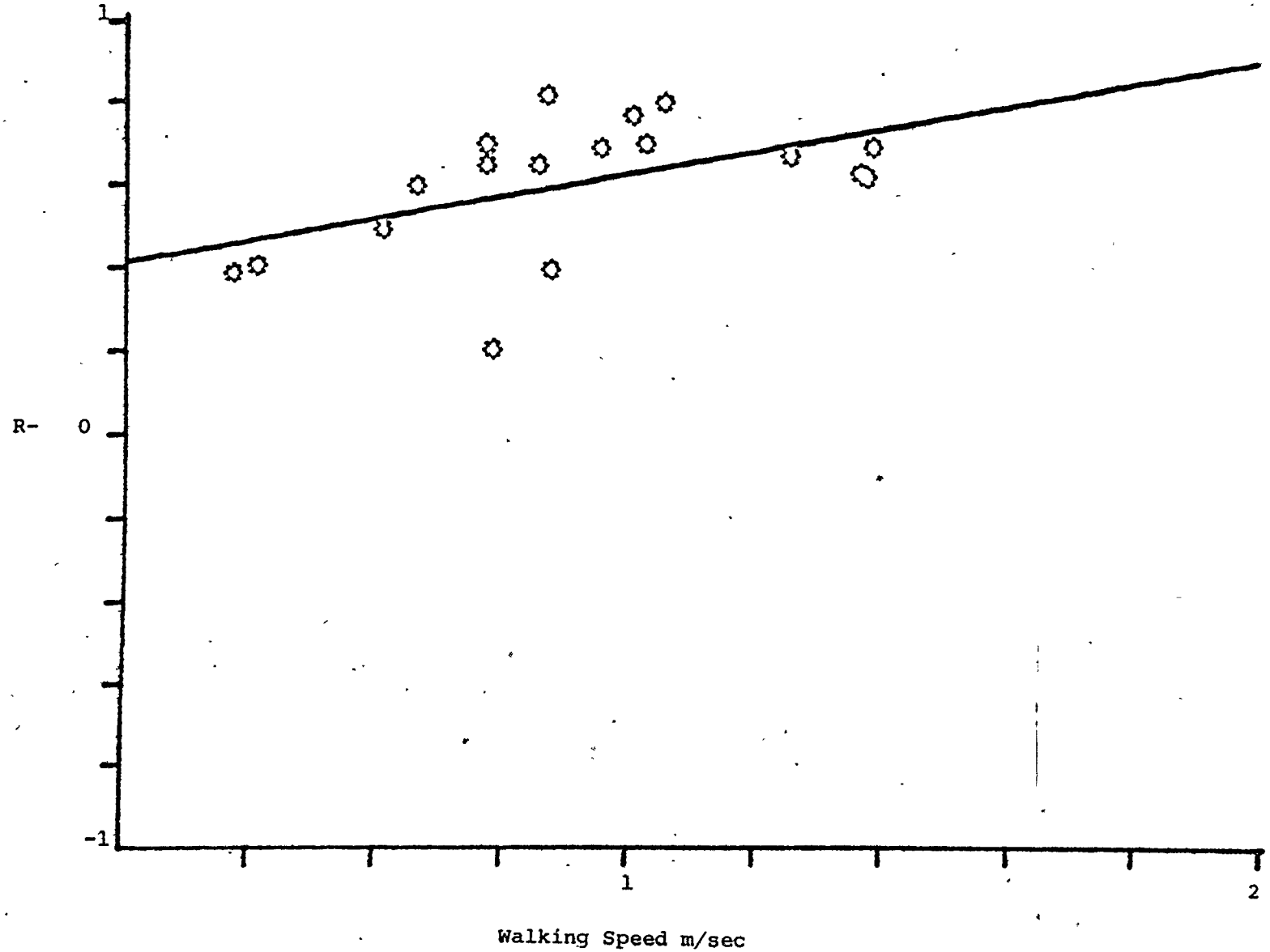
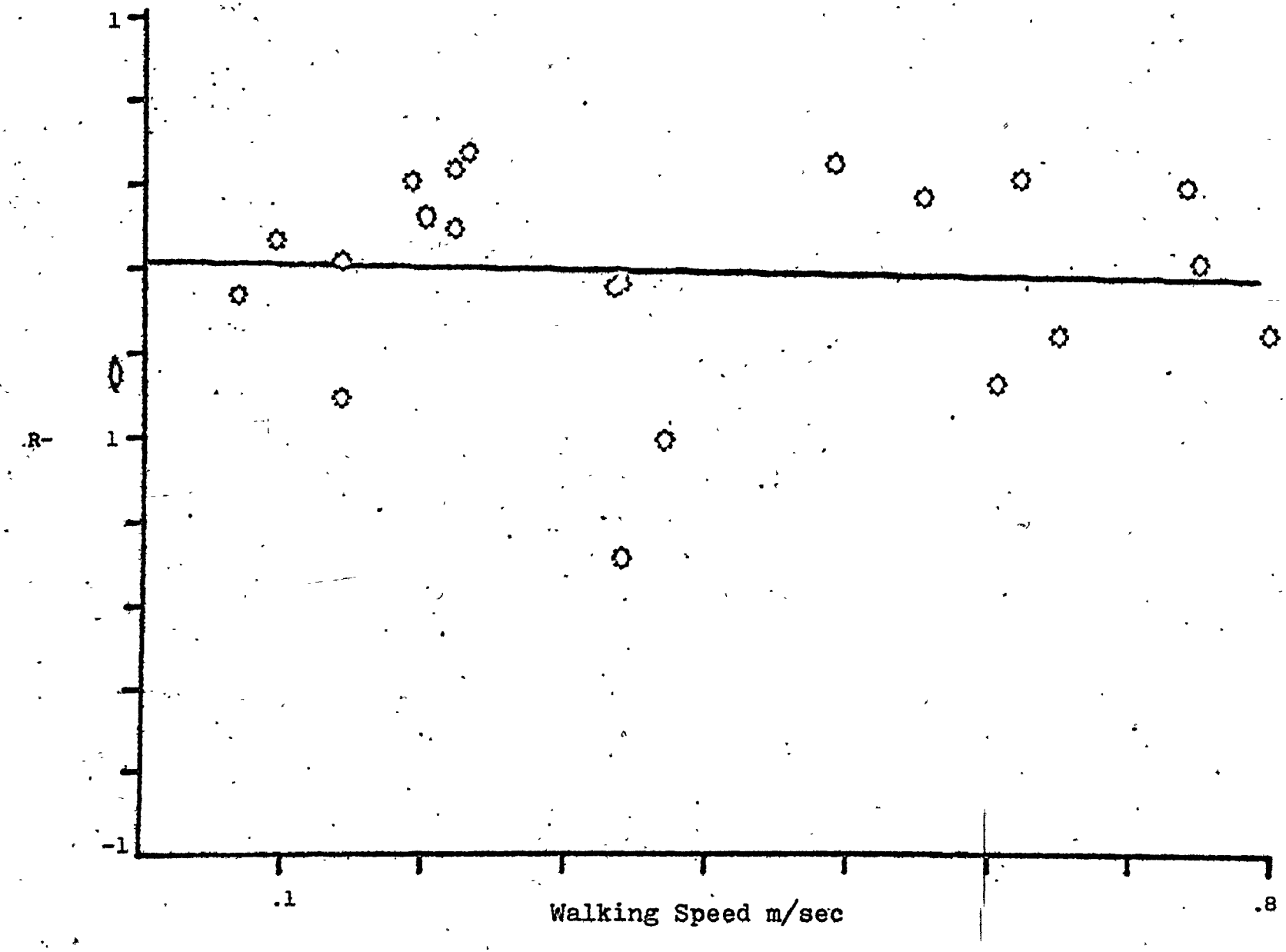
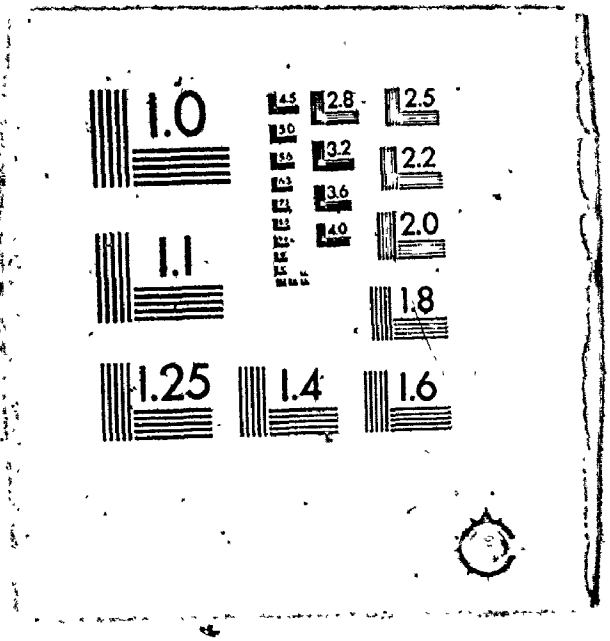


Figure 5.13 Regression Analysis Hemiplegic Population: R- Versus Speed of Walk





the R's . That is, the R's from the normal left side records increase with increasing speed of walk, because the phasic activity of the quadriceps and hamstrings tends to become more repeatable.

In R(0) and R+ the opposite trend is revealed for the hemiplegic subjects (affected side) - these values of R decrease with increased speed of walk.

The straight line in Figure 5.13 shows that there is no relationship between speed of walk and R- for the hemiplegic population

5.6 Discussion

The cross correlation function is used in the study of observations which exhibit randomness so it is quite appropriate for the study of EMG which is a random signal. The character of the function is more important than the magnitude, and should be studied in conjunction with the EMG records in order to relate various features such as delays, broadness of peak, and more than one peak, to the stride cycle.

There are few objective tools to describe the state of the patient following a stroke, or to assess the progress made as a result of physiotherapy and rehabilitation. At present, the patients are evaluated according to Brunnstom's method (1970) which identifies six stages of recovery after stroke. These stages are not linear increments, but groups of subjective assessments which are not supported by kinesiological EMG.

The cross correlation function reveals the phasic pattern of muscle activity. Therapists may utilize this information to prescribe a specific regimen to induce correct muscle activity patterns. Although instability at knee and ankle may readily be observed, the cross correlation coefficient provides a unitless measure of the instability. Abnormal muscle activity patterns (such as extension synergy or excessive co-contraction of antagonist muscles in order to stabilize a joint) which had been observed clinically in a patient were also revealed and quantified by the cross correlation function. Even in the case of very low levels of muscle activity, the cross correlation function reveals the micro level muscle function because the coefficients are independent of the amplitude of the signal.

This method of analysis allows the assessment of patient progress according to the neurological picture of improvement in control.

A wealth of information concerning the timing of muscle activity is available in the $\pm\tau$ parameters, that is, the time of occurence of peak activity at positive and negative lags. However, there must be sufficient data so that the period for cross correlation is consistently one full stride. The algorithm which identifies maxima should be amended to identify local maxima for positive and negative lags. This would overcome the problem of a coefficient adjacent to R(0) being designated as a local maximum.

It is possible that windows should be tailored to speed of walk. That is, a larger window may be appropriate for the slower walking speeds generally observed in hemiplegic gait. Section 4.3.1 reports the selection of a window on the basis of the study of one normal subject at one walking speed.

A complete spectral analysis of the EMG of normal and hemiplegic subjects may reveal different frequency components in these groups. This result would certainly require reconsideration of filtering techniques.

Many questions have been raised about the application of the methods of numerical analysis to data acquired in a gait study. This thesis examines the feasiblity of one such method of analysis. The cross correlation of EMG records of pairs of muscles on the lower limb suggests that this method is a useful adjunct in assessing the level of function, and the degree of improvement of patients with neuromuscular dysfunction following stroke. The method is practical and easy to implement. Results may be furnished within minutes of a gait study by the use of an automated program such as CRSSIX, which was written for this study and used to process the data and provide some of the illustrations for this thesis. This straightforward mathematical technique sifts a few numbers from a mountain of information supplied by the millions of nerve and muscle cells in the human lower limb. Clinical observations concur with the interpretation of the cross correlation functions.

CHAPTER 6

CONCLUSION

The present method of analysing patterns of activity for a single muscle consists of converting the phasic pattern of EMG into a rectangular waveform with at most four discrete amplitude levels (Grieve and Cavanaugh, 1974). Only the times of turn on and turn off of muscle activity, when the level of phasic EMG crosses some arbitrary threshold, are noted. Comparing the activity patterns in muscles is reduced to comparing these thresholds rather than amplitude history during the on period. Most of the phase information is lost.

Until now, there has not been a way of determining the phasic EMG pattern of one muscle with respect to the phasic EMG pattern of another, nor has there been a way of calculating the relative times of occurence of the strongest contraction during a stride cycle. There has been no method independent of the selection of an arbitrary threshold.

The task of determining turn on of a muscle has been difficult, considering the irregular pattern of electromyographic recordings.

The following questions arise in the consideration of methods of analysis of EMG

1. Given the irregular patterns of activity in EMG, is there a method for determining the phase inter-relationship between muscles? Can this method be automated?

2. Can unique pathologies such as agonist/antagonist co-contraction and flexion/extension synergies be identified? Is there a quantitative index for describing these pathologies?

3. Can times of peak activity of one muscle relative to another be identified? Can these patterns of activity be quantified?

One method which addresses these requirements affirmatively is the use of cross correlation, a statistical method which is easily automated and which permits the comparison of random wave forms. This straightforward mathematical technique describes the degree of co-activation in a pair of muscles and the time delay in peak activity of the muscles relative to each other. A pathology such as a flexion/extension synergy may be identified by high values for R(0) from the cross correlation of the flexor muscles (the hamstrings and tibialis anterior) or the extensor pair (quadriceps and gastrocnemius). The severity of the pathology may be quantified by the magnitude of this coefficient. The patterns of activity in muscles may be quantified by identification and extraction of features from the cross correlation function.

Every method of analysis has its limitations, and so does the technique of cross correlating pairs of EMG records recorded from muscles of the lower limbs during locomotion. This method works best for brief strong contractions. If contractions are prolonged, the shape of $R(\tau)$ makes it difficult to pinpoint the real maximum for

that peak, since several adjacent coefficients may have the same value. If there is insufficient data, too small a portion of a stride can be covered by $R(\tau)$ for a significant result. Comparison of results for whole strides is much more useful.

As a result of the topics considered in this thesis several changes in the protocol for data acquisition and analysis have been implemented by the Gait Research Group at the Chedoke Rehabilitation Centre.

A lower sampling frequency is now being used as a consequence of the conclusions drawn from the spectral analysis of EMG discussed in Section 1.2. This lower sampling frequency allows the collection of data from longer walks, and as a result whole strides may be processed by cross correlation.

Spectral analysis of EMG from normal and hemiplegic subjects is being done to determine whether there are shifts in the frequency contents of the signals. A faster sampling rate may be required for hemiplegic subjects if the frequency range shifts to higher frequencies in pathologies. In addition, window widths for various speeds of walking will be investigated.

The spectral analysis of EMG has also confirmed the use of surface electrodes for data acquisition. Perry et al (1980) have addressed this issue and state that needle electrodes should be used because surface electrodes pick up too much "cross talk" from nearby muscles not included in a study. The closest muscles geometrically are the tibialis anterior and gastrocnemius. If there were cross

talk, we would see very large, positive values for the cross correlation function for these muscles at time zero. This did not occur in the analysis of any normal or hemiplegic subject in this study. Raw records may be cross correlated to confirm the absence of cross talk. In bipolar records, motor unit potentials do not line up in time. A value of R(0) approximately equal to zero for the cross correlation of the tibialis anterior and gastrocnemius would dispel any concern about cross talk.

A second order Butterworth low pass filter with a 9 Hz cutoff has been implemented to smooth preprocessed EMG (Section 4.4) because of its superior transfer function.

Clinical members of the gait research team are using interactive computer programs to study muscle activity. The programs written for this thesis process EMG records acquired from subjects in the study population and produce graphic displays of the cross correlation function which may be photocopied. Pertinent data is streamed to the printer. The analysis proceeds quickly and automatically.

In processing EMG data acquired during the dynamic process of human locomotion, cross correlating EMG records from pairs of muscles provides a viable tool for gait analysis in the clinical setting. The method yields a few unitless parameters from a confusing mosaic of information. It allows normal/hemiplegic subject and normal/hemiplegic group comparisons. This method allows the progress of stroke patients to be monitored according to return of neurological control rather than subjective assessment of stage of recovery.

The cross correlation functions produced for each hemiplegic subject correspond to the subject history at the time of data acquisition. Hence, inspection of the function is clinically useful.

Although many techniques have been developed for quantification of human locomotion data, the determination of phasic interdependence of pairs of muscles has not been considered. The use of the cross correlaton function allows us to quantify these patterns and provide a rehabilitation index.

Further refinement of the algorithm should allow clinicians to assess patient function according to the degree of neurological control.

A method of classifying poststroke patients could be developed from the use of cross correlation analysis of EMG to monitor the rehabiitation process.

APPENDIX A

UNIFORM STRIDE ALGORITHM

NSTRS = the number of strides selected for processing

INSTART = first sample of a stride

IEND = last sample of a stride

IBEGIN = first sample of first stride -1

UIPER = unifrom stride period

J = real sample identifier

I = uniform sample identifier

TIME = fraction of real time period

SIG = real EMG sample vector

UNI = uniform sample vector

NUSAMP = # uniform samples/stride

IAST(N) = first sample of Nth stride

IAEND(N) = last sample of Nth stride

```
   IS          =1
   IL          = NUSAMP
   DO 10       N = 1, NSTRS
   ISTART      = IAST(N)
   IEND        = IAEND(N)
   UIPER       = (IEND-ISTART+1)/NUSAMP * RPER

               DO 20 I    = IS, IL
               J          = UIPER*(I-1)/RPER + IBEGIN
               TIME       = ((I-1)UIPER-(J-IBEGIN)*RPER)/RPER
               UNI(I)     = SIG(J)+(SIG(J+1)-SIG(J))*TIME
            20 CONTINUE

10 CONTINUE
```

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