THE INFLUENCE OF PARTIAL WEIGHT BEARING ON NORMAL GAIT: A NOVEL APPROACH FOR GAIT RETRAINING IN NEUROLOGICAL PATIENTS.

L. FINCH
School of physical and Occupational Therapy McGill University, Montreal.

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ABSTRACT

Neurological rehabilitation emphasizes gait retraining, however, poor patterns often persist. Interactive training (partial progressing to full weight bearing (FWB) combined with treadmill stimulation) allowed recovery of locomotion in spinalized cats. Normal responses to partial weight should be known, before applying this strategy to patients. Thus, 10 normal males walked on a treadmill with 0, 30, 50 and 70% of their body weight supported (BWS). At each BWS, the subjects walked slower than normal. To dissociate speed from weight changes each subject walked at the same speed FWB and with BWS. Simultaneous electromyographic (EMG), footswitch and video data were collected. FWB and BWS gait appeared similar, except at 70% BWS. Significant differences between other BWS and FWB trials were a decrease in; percent stance, double support time, hip angular displacement, and the EMG amplitude of erector spinae, and gastrocnemius. A training strategy of partial weight support progressing to FWB was developed and should be tested on patients.

ABSTRAIT

La rêhabilitation neurologique accentue la rééducation de la marche, mais souvent un patron anormal persiste. Les chats ayant une lésion spinale complête récupèrent leurs fonctions locomotrices à la suite d'entrainements interactifs [(stimulation du tapis roulant combiné au support de poids progressif (SPP)] . Les réactions normales au support de poids partiel doivent être connues avant d'être utilisées sur le patient. Lorsque 10 mâles normaux marchent sur le tapis roulant à 30, 50 et 70% du poids du corps supporté (PCS), ils ralentissent à chaque PCS. Pour dissocier la vitesse aux changements de poids, chaque individu marche à la même vitesse avec SPP et PCS. L'activité électromyographique (EMG) et le vidéo sont enregistrés simultanément. La démarche SPP et PCS semble identique, sauf à 70%. Les changements (considérables) remarqués entre essais SPP et PCS sont une diminution du % phase d'appui, double phase d'appui, amplitude à la hanche et résultats d'EMG. Nous suggérons qu'un PCS au SPP soit expérimenté avec les patients.

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1. Introduction

Cerebral vascular accidents (CVA) rank second as a cause of hospitalization in Canada, but account for only 9 percent of deaths (Statistics Canada 1984). Many of the survivors exhibit a wide range of disability requiring specialized care (Gresham et al. 1975). The large number of cases and their relatively long survival emphasises the role of rehabilitation. Although neurological diseases initially cause great locomotor difficulties, 54-80% of survivors of CVAs recover some independent walking ability (Garraway et al. 1980, Gresham et al. 1975). The majority of patients, however, still walk poorly and as walking ability is often equated with independance and quality of life a major goal of rehabilitation should be the restoration of normal gait. (Chin et al. 1982).

Previously therapy aimed at helping patients cope with their disability, but more recently the focus has been to develop training techniques to restore function (Gloag 1985, Kottke 1982).

The major problem in neurological gait is the inability to support body weight while moving forward (Goldfarb and Simon 1983, Knutsson and Richards 1979, Knutsson 1972). One retraining approach is to teach proper weight bearing throughout the gait cycle (Bogarth and Richards 1981, Bobath 1978). The patients are taught proper weight bearing while bearing their full weight. The increased ability to control weight transference and the weight acceptance phase of gait seem to improve other gait components. Even after treatment, however, appropriate weight bearing is not always achieved and despite the patients' adequate muscle activation and underlying abilities, poor gait patterns often persist (Chin et al. 1982).

Another approach to retraining might be to use a progressive partial to full weight bearing technique combined with treadmill stimulation.

Experimental studies on adult spinalized cats demonstrated that normal gait can be restored with this interactive strategy (Rossignol et al. 1984).

2. OBJECTIVES

The overall objective of this study is to develop a gait training strategy for neurological patients, based on the spinalized animal model of partial weight support and treadmill stimulation. Fundamental to this development is to determine if gait parameters change significantly from the normal full weight bearing state when 30, 50 or 70% of body weight is supported. Consequently, kinematic (cycle time, percent stance time, percent double support time, cadence, stride length, and the angular displacement pattern of the hip and knee) and electromyographic (EMG) (on/off timing and normalized mean burst amplitude of the right: erector spinae, gluteus medius, vastus lateralis, medial hamstring, tibialis anterior and gastrocnemius), measures used in the animal model, will be adapted to the human experimental situation.

On the basis of these results a methodology for gait retraining in neurological conditions will be proposed.

3. Literature Review

3.1 NORMAL HUMAN GAIT.

Normal locomotion will be reviewed. Kinematic (temporal, distance, and sagittal joint motion), kinetic (ground reaction forces, and joint moments) and electromyography (EMG) factors will be discussed.

3.1.1 Temporal-Distance

The basic fundamental unit in human gait is the gait cycle. Figure 1 illustrates the basic temporal-distance (TD) relationships in a cycle. A gait cycle, outlined in figure 1B occurs from heel-strike, denoted as 0%, to the subsequent heel-strike of the same limb, denoted as 100% of the cycle. The components of a cycle are: a) stance, from heel-strike to toe-off, which takes approximately 60% of the cycle, and b) swing, comprising the remaining 40% from toe-off to heel-strike (Bowker and Hall, 1975). A period of double support exists when both limbs are in contact with the ground. Stance can be subdivided further into critical events consisting of heel-strike, foot-flat and mid-stance. The body moves forward during mid-stance balanced on a single stance limb to the next critical event - heel-off. Toe-off then occurs which marks the end of stance and the beginning of swing. Two other parameters (figure 1A) exist, namely a) cadence, or step frequency, defined as the number of steps taken per unit time and b) stride length, the distance covered by a limb during one cycle (Inman et al. 1981, Bowker and Hall, 1975, Murray, 1967).

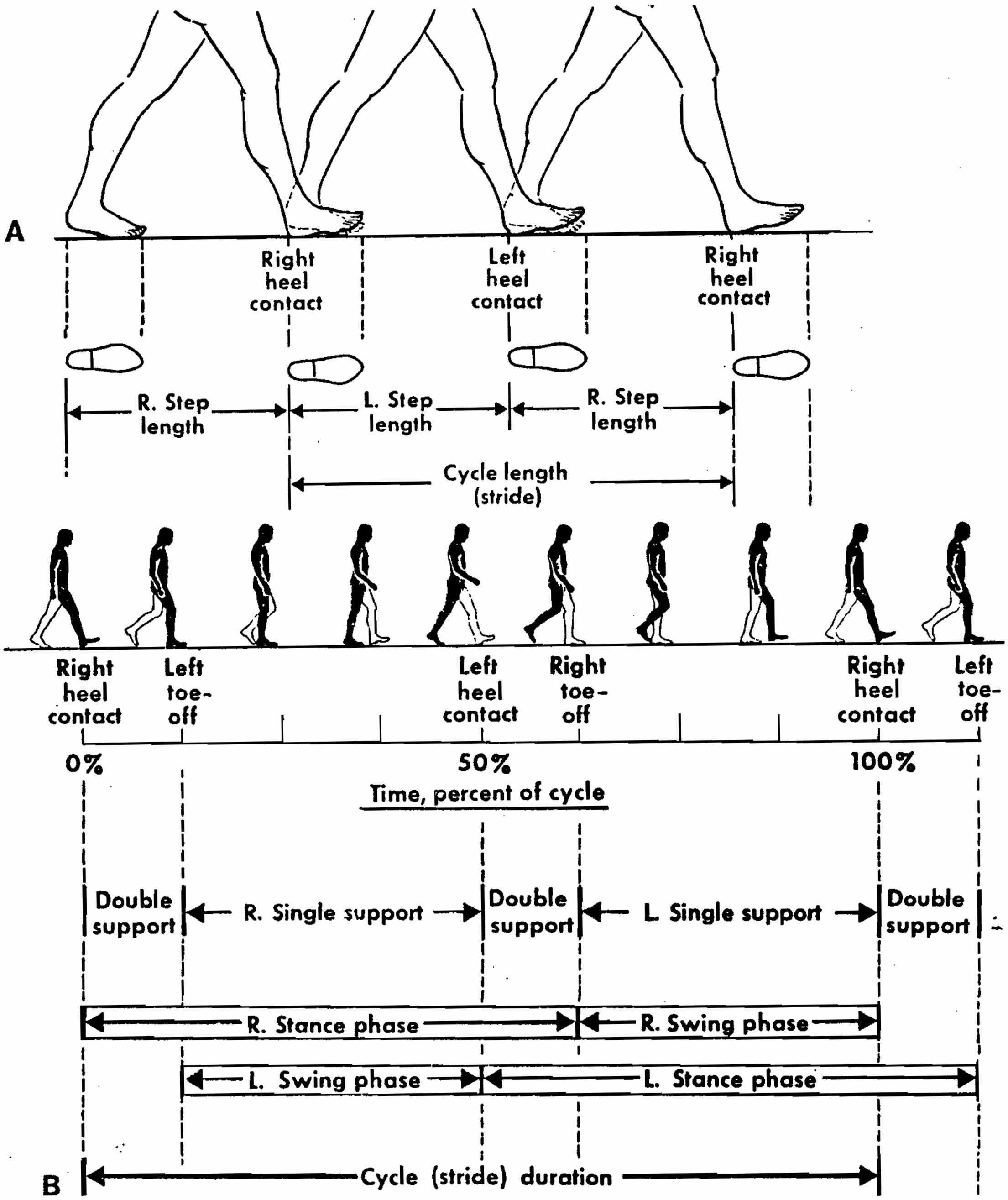
Identical cycles are rare. One cycle, obtained by averaging a number of cycles is usually taken as representative of a specific cycle pattern.

A stopwatch and instrumented walkway or surface capable of bearing the imprint of footsteps permits accurate documentation of temporal distance variables (Yack 1984, Robinson 1981). These measures may also

Figure 1A Distance parameters of the walking cycle

Figure 1B Time parameters of the walking cycle

Taken from Inman et al. (1981)


A
Right
heel
contact
Left
heel
contact
Right
heel
contact
R. Step
length
L. Step
length
R. Step
length
Cycle length
(stride)
Right
heel
contact
Left
toe-
off
Left
heel
contact
Right
toe-
off
Right
heel
contact
Left
toe-
off
0%
50%
100%
Time, percent of cycle
Double
support
R. Single support
Double
support
L. Single support
Double
support
R. Stance phase
R. Swing phase
L. Swing phase
L. Stance phase
B
Cycle (stride) duration

be obtained from imaging systems. Another more precise and reliable method is via footswitchs; tape switches attached to the shoe or a conductive floor are used. These devices linked in series to a set of resistors produce an electrical signal. The signal depicts the exact position of the foot depending on the number of sensors used. The footswitch data can be used as a time reference if recorded simultaneously with other data, for example EMGs (Dubo et al. 1976).

The choice and intrepretation of temporal distance parameters depends on the goal of analysis. Temporal distance measures are widely used outcome discriptors not only of normal, but also of pathological gait. Parameters selected should therefore be evaluated with regard to their reliability as well as meaningfulness in gait analysis. The test re-test relialibility of TD measures for normal individuals, both between and within subject using intermittant light photography was reported as "striking" (Murray et al. 1966) and in a number of studies with footswitches showed little variability (Larsson et al. 1981, Lyons et al. 1983, Winter 1984). Of the TD parameters available, Stanic et al.(1977) stated that cycle time, % stance and step length are statistically less variable than other TD variables in normal locomotion. They also stated that a gait evaluation can be performed with a minimum number of variables namely % stance, step duration and step length. Chao et al. (1983), while studying the kinetic and kinematics of normal knee motion, found the TD variables among the most significant. Dubo et al. (1976), on the other hand, felt that as many parameters as possible should be assessed to quantify normal gait, especially for rehabilitation purposes. Despite collecting a large number of parameters from footswitch, video and EMG recordings, Dubo et al. (1976) only reported on the phasic EMG activity and on TD parameters of cadence and stance swing ratios.

TD variables are highly related; step length and stride length, stance and cycle time are directly related to walking speed which is determined by cadence and stride length (Inman et al. 1981, Larsson et al. 1981, Murray et al. 1966). The absolute value of each above parameter can be influenced by gender and age, but the variability is not (Murray et al. 1967, Gabell and Nayak 1984). In addition, the different phases of normal gait, as studied by Larsson et al. (1981), demonstrate a linear relationship between duration of cycle and stance and between swing and double support time. Consequently, during a gait evaluation, velocity, which influences all parameters, must be stated and at least the minimum number of parameters as outlined by Stanic et al. (1977) should be measured. The phases of the gait cycle should also be quantified not as indicators of the quality of gait, but as indicators of postural control (Eke-Okoro and Larsson 1984, Gabell and Nayak 1984). A record of double support time (the weight transference phase of gait) is essential to record as an increase in its value points out the need for greater stability to compensate for poor balance and postural instability.

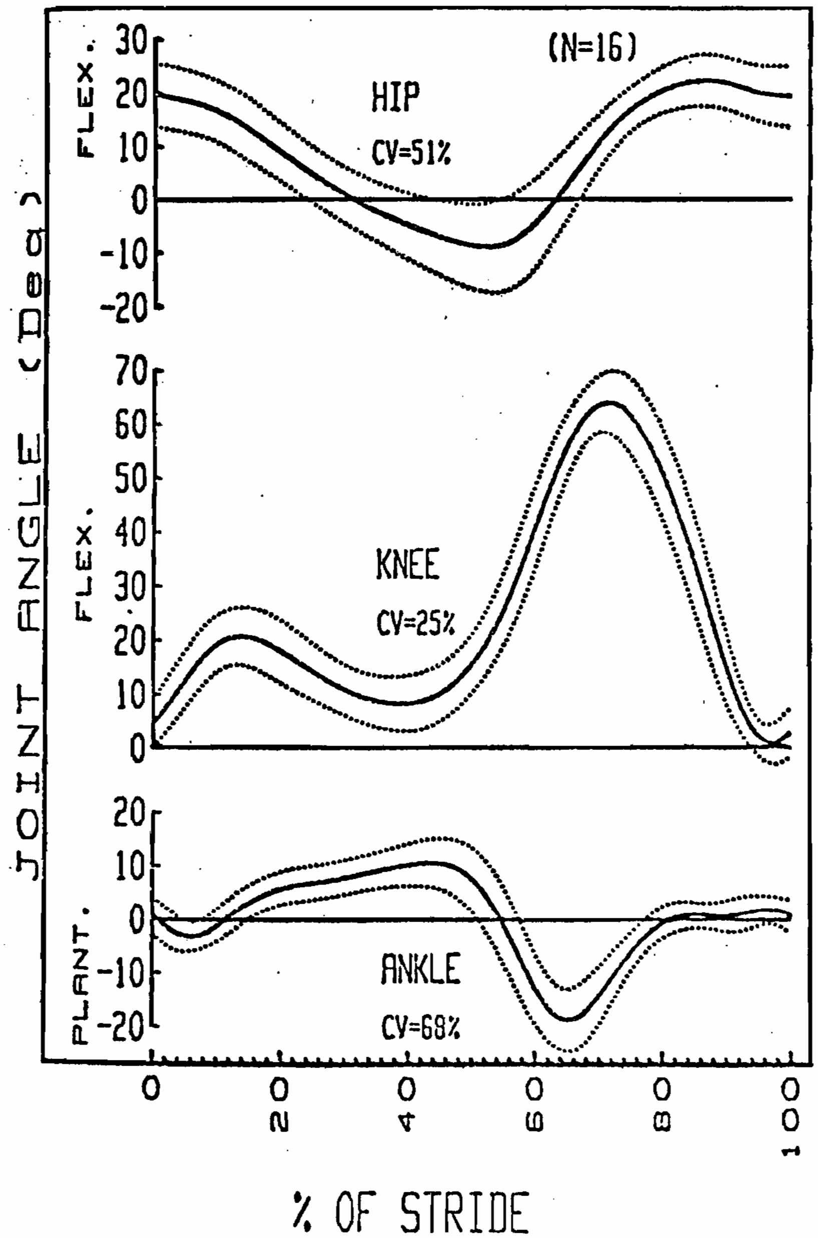
3.1.2 Angular displacements

Joint motion is the measurement of hip, knee and ankle angular displacement in the sagittal, or plane of progression. Movement occurs in two other planes; the coronal and transverse. These planes are not discussed in this thesis, but are well documented elsewhere (Murray et al. 1964, Inman et al. 1981). The angular joint displacements, as seen in figure 2, exhibit a series of curves with flexion and extension phases (Murray 1967).

At heel-strike, the ankle plantarflexes, the knee flexes and the hip extends. During the next phase, up to mid-stance, the ankle dorsiflexes and the hip and knee extend, while from mid-stance to

Figure 2 Ensemble average plots of normal hip, knee and ankle joint angles in saggital plane for 16 subjects at natural cadence. Solid line indicates the average with dotted line indicating one standard deviation.

Taken from Winter (1983)


(N=16)
HIP
CV=51%
KNEE
CV=25%
ANKLE
CV=68%
JOINT ANGLE (Deg)
FLEX.
PLANT.
30
20
10
0
-10
-20
70
60
50
40
30
20
10
0
20
10
0
-10
-20
0
20
40
60
80
100
% OF STRIDE

toe-off the ankle plantarflexes and the hip and knee flex. Throughout swing, the ankle dorsiflexes to a neutral anatomical position for heel-strike, the knee flexes and then extends in preparation for heel-strike, while the hip flexes (Inman et al. 1981, Winter 1983).

Obtaining and evaluating angular displacement data requires recording techniques, that are accurate, and that do not hinder movement (Stanic et al. 1977, Winter 1982). For example, one system used to analyze kinematic data involves goniometry, that requires lengthy preparation and calibration, provides relative displacement data, and can encumber movement during gait. On the other hand, video recordings require few body markers that are quickly and easily applied; provide a large volume of displacement data that are absolute in space for complex movement analysis; and provide a permanent record for later re-assessment (Winter 1982). Video recordings may also be played back immediately frame by frame to allow for freezing of movement to quantify specific patterns. For example, joint angles may be measured directly from the screen at specific points in time similar to the technique used in cinematography (Murray et al. 1964, Hewes et al. 1967). However, when limbs rotate during movement, measurement of angles in the sagittal plane from the screen leads to errors (Winter 1979). TD data may also be determined from video records.

Once analysed, angular data can be plotted as angle angle diagrams (Grieve 1969) or the data can be normalized in time and plotted as a function of time (Murray 1967, Winter 1984). These graphs are then used as normal or abnormal gait descriptors (Hershler and Milner 1980).

The usefulness of displacement data is being debated (Yack 1984). For example, in normal gait, displacement data were not used as major determinants (Saunders et al. 1958, Chao et al. 1983), but Stanic et al. (1977), (with goniometry), and Sutherland et al. (1981), (with video), stated that the maximum hip and knee swing angles were two of the major characteristics of gait. Chao et al. (1983) suggested that displacement variables may be redundant in normal gait. As angular displacement data are often very variable in pathological gait, the usefulness of these data appear therefore limited. Nevertheless extensive use of angular displacement data has been made in evaluating treatment techniques (Bogarth and Richards 1981, Sutherland et al. 1981).

The reliability of displacement measures and their variability over time and person in normal subjects has been documented (Murray et al. 1964, Murray 1967, Winter 1984, Nilsson et al. 1985). Intra subject variability is extremely low the root mean square standard deviation as measured by Winter (1984) was 1.50 at the ankle, 1.90 at the knee, and 1.80 at the hip. The meaning of angular displacement data beyond its discriptive abilities, however, needs to be explored.

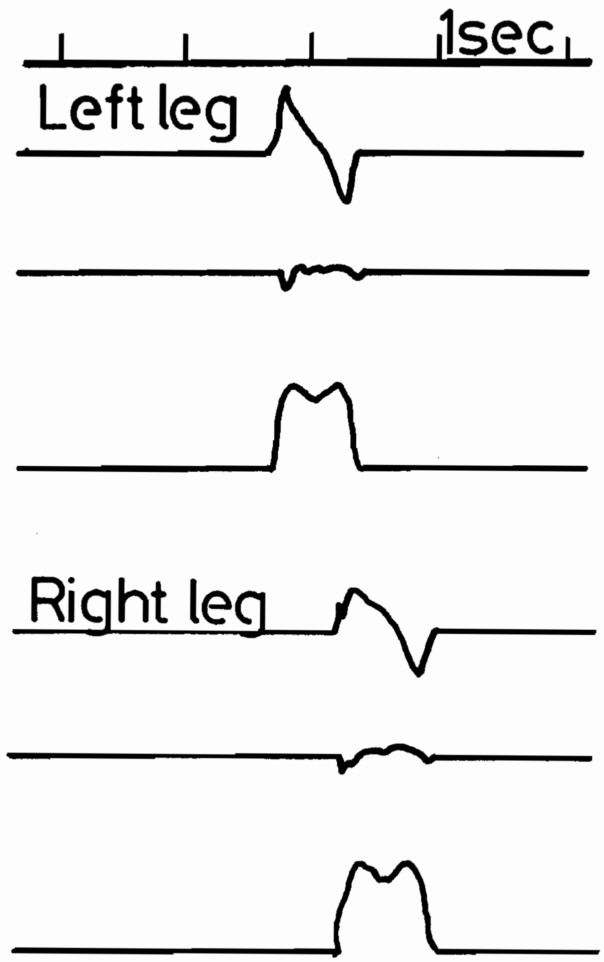
3.1.3 Kinetics

The ground reaction force in figure 3 is a three dimentional force with one vertical and two horizontal components. The normal vertical force exhibits a rapid rise at heel contact to a value 10-15% greater than body weight. Following heel-strike, as the knee flexes, the value drops to approximately 80% of body weight. At push-off the leg extends and produces a second force peak greater than body weight (Carlsoo et al. 1974, Winter 1979).

Moments of force about a joint represent the net muscle activity at that joint, but not the direction of the activity (lengthening or shortening). Calculations of moments about a joint show the

Figure 3 Components of the ground reaction forces. The upper force records the medio-lateral, antero-posterior and vertical forces of the left leg. The lower records the same components of the right leg. Distance between vertical lines 1 second.

Taken from Carlsoo et al. (1974)


1sec
Left leg
Right leg

contribution of gravitational, net muscular, and acceleration or deceleration forces (Winter 1979).

Moments about a joint reflect the overall muscle output for that joint and are usually extremely variable (Pedotti 1977, Winter 1981). Nevertheless, during a gait cycle the ankle has an initial small dorsiflexion moment followed by a planterflexion moment, the hip an extensor followed by a flexor moment, and the knee an inconsistent pattern as shown in figure 4 (Winter 1981). In spite of this variability the total support moment, a concept developed by Winter (1980), to describe the overall extensor support pattern of the leg, is consistently positive during stance. The implications of this are two-fold a) when one joint opposes or does not contribute to support, compensation at one or both other joints occurs to prevent collapse (Winter 1983) and b) this large variability may indicate that gait is less robotic than was once thought (Winter 1984).

3.1.4 EMGs in gait

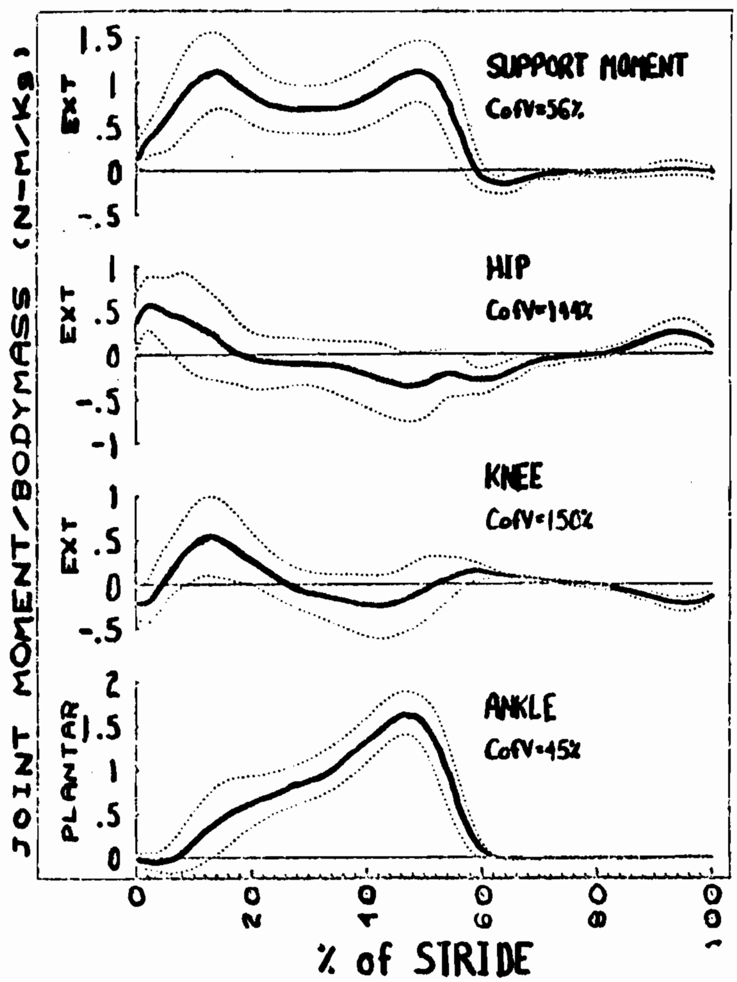
While moments represent the net muscle pattern at a joint, EMGs demonstrate individual muscle functioning. EMG signals have been related not only to the tension produced in muscles, (even under isotonic conditions (Bigland and Lippold 1954, Bouisset 1974), but also to the joint moment histories in gait (Pedotti 1977). Measures of EMG amplitude, duration and phasing can be used to obtain a profile of muscle activity, its appropriate phasing and the intensity of contraction necessary to achieve the desired movement.

EMGs are measured by electrodes. Surface electrodes are used to record overall muscle activity. The electrode type and anatomical placement affect the ability to accurately record any signal.

Normal locomotor muscle activity is centered around the beginning and end of stance and swing, the periods of limb acceleration and

Figure 4 Ensemble average of hip, knee and ankle joint moments of force/body mass for 16 subjects at natural cadence. Support moment is calculated by adding up the extensor moments at each of the joints. Solid line indicates the mean with dotted line indicating one standard deviation.

Taken from Winter (1983)



JOINT MOMENT/BODYMASS (N-M/Kg)
SUPPORT MOMENT
CofV=56%
HIP
CofV=144%
KNEE
CofV=150%
ANKLE
CofV=45%
EXT
PLANTAR
1.5
1
.5
0
-.5
-1
2
20
40
60
80
100
% of STRIDE

deceleration, as well as during weight acceptance periods (Inman et al. 1981, Yang and Winter 1985).

An EMG gait pattern is difficult to specify. A standard method of processing its phasic characteristics, shape, or amplitude has not been developed (Dubo et al. 1976, Battye and Joseph 1966, Grieve and Cavagna 1974). In addition, inter-subject measurement differences (e.g. placement of the electrodes) and biological differences (e.g. muscle type, amount of subcutaneous fat) exist (Winter 1984A, Yang and Winter 1985). Moreover, differences can be artifically reduced or amplified with different normalization or transformation procedures (Winter 1984A).

The phasic characteristics of the EMG signal have been interpreted via raw data (Pedotti 1977, Nathanson and Hershberg 1952) and by a temporal analysis of on/off timing (Battye and Joseph 1966, Mann and Hagy 1980). The amplitude and overall temporal shape of the EMG signal has been quantified by averaging the amplitude over a cycle(Milner et al. 1971), by the number of turning points (Grieve and Cavagna 1974), by integration of the signal (Brandell 1977), and by a linear envelope (Winter 1984A).

The choice of method depends on the objectives of the analysis. EMG recordings of on/off timing of muscle bursts have been studied extensively in a number of animals to investigate the neural control of locomotion (Grillner 1975). Medeiros (1978) used on/off timing of EMG patterns to investigate the neuronal mechanisms underlying human locomotion. The EMGs linked to footswitches, revealed a tendency to maintain a constant phase relationship between heel strike and onset of EMG activity in 3 out of the 4 muscles studied. The large within and between subject variability in Mederios' study was taken to represent normal human biological variation. Battye and Joseph (1966), also studying humans, tabulated the periods of on/off for different lower limb muscles. They concluded that the timing of muscle activity was very similar across subjects. The EMG timing linked to cine recordings of movement helped define specific muscle action during gait. Other researchers have linked EMG on/off timing to footswitch data (Soderberg and Dostal 1978, Lyons et al. 1983) to define muscle activity. A great deal of variability, both within and between subject was seen, nevertheless, they felt the variability did not obscure their findings.

Sutherland et al. (1981) using EMG on/off timing and mean amplitude measurements on children, on the other hand, concluded that EMGs were of limited value in the analysis of gait changes. The variability in the timing of muscles was large in their study and thus timing was an insensitive measure of change. To provide a more sensitive measure of change and adaptability in gait, Sutherland et al. (1981) suggested that EMG amplitude measures be used to analyse muscle contraction intensity.

On/off timing of EMG bursts provides information at two distinct points in time both of which are important in relation to function. Timing combined with a measure of EMG amplitude may be more beneficial. The amplitude of EMGs calculated as an average relative increase of EMG (Brandell 1977), or mean value of the average waveform over a cycle (Milner et al. 1971), measures the overall "turn on" of a muscle during gait. One alternative to these global methods (which mask amplitude changes within a burst), is to measure the mean EMG amplitude of specific muscle bursts as done in animal experimentation (Rossignol et al. 1985, Zomlefer et al. 1984).

A second alternative is to determine the shape of the amplitude change over time, (the linear envelope). This allows a better intrepretation of the EMG signal to link muscle functioning with mechanical events (Basmajian 1976, Winter 1984A). Yang and Winter (1985) used the shape of the EMG linear ensemble envelope, as well as the mean stance and swing amplitude to quantify gait changes. The EMG pattern was independent of cadence, but the mean EMG amplitudes for stance and swing differed significantly. The shape of the linear envelope thus may not be as sensitive a measure as the mean EMG amplitude in instances of large intersubject variability.

3.2 Speed and Normal Gait.

The parameters of gait are interrelated and velocity dependent (Winter 1984, Thorstensson et al. 1982, Grillner et al. 1979). The influence of speed should be dissociated from other influences to quantify gait changes. Normal walking speed is considered to range from 1.1 to 1.5 ms-1(Murray 1967).

Many investigators have analyzed temporal, kinematic, kinetic and EMG patterns in relation to speed (Murray et al. 1964, Herman et al.1976, Winter 1983, Larsson et al. 1980, Thorstensson et al. 1982, Yang and Winter 1985). These are discussed below.

3.2.1 Temporal-distance parameters and speed

Speed and range of speed chosen vary greatly as does the method of assigning speed. Comparisons are therefore often difficult. Nevertheless, with increasing speed cycle time decreases (Murray et al. 1964, Larsson et al. 1980, Nilsson et al. 1985), the relative contribution of swing increases and stance and double support time decrease (Larsson et al. 1980, Andraicchi et al. 1977). The relationship of stance and swing relative to cycle time demonstrates a large linear change for stance and a small linear change for swing (Nilsson et al. 1985, Larsson et al. 1980, Murray 1967). Herman et al. (1976) found a linear relationship between the square root of double support time and cycle time, while Murray (1967) and Larsson et al. (1980) found a linear relationship directly with double support time. Only Larsson et al. (1980) published their corelational values ($r2=.96$). Different walking speeds were used in these studies. Herman et al. instructed subjects to select brisk, natural, slow and extremely slow walking speeds, the speeds ranged from .55 m.s-1 to 1.86 m.s-1. Larsson et al. used a similar protocol, but their subjects speed ranged from .46 to 2.4 ms-1, while Murray's subjects walked in time to a metronome at a set pace. The negative linear relationship between cycle and double support time implies that as speed decreases, subjects spend more time with both feet on the ground probably for stability and control (Herman et al. 1976).

Increased walking speed is usually accomplished by increasing stride length and decreasing stride time. Cadence and stride length increase linearly with speed (Grieve and Gear 1966). The relationship between stride length and cadence is usually constant over the range of .7 to 2 ms-1 (Herman et al. 1976). The variability of these two parameters increases with decreasing speed (Larsson et al.1980)

3.2.2 Angular displacement and speed

Joint displacement patterns over a stride do not change with speed (Murray et al. 1966, Winter 1983). Increasing walking speed increases the joint angular displacement amplitude and the velocity of displacement (Mann and Hagy 1980), while decreasing walking speed decreases them both (Winter 1983). Among the ankle, knee and hip the latter two demonstrate the least variability with speed. The root mean square difference between slow, natural and fast cadence ranges from 2.10 to 4.10 (Winter 1983, Nilsson et al. 1985).

3.2.3 Kinetics and speed

Joint moment patterns change with speed. The magnitude of the moment peak increases with increasing cadence except at the ankle whose moment decreases with increasing speed (Winter 1984). A high degree of variability exists in the moments at the hip and the knee compared to those at the ankle, probably reflecting the greater number and flexibility of the two joint muscles at the hip and knee. The amount of flexibility and thus the variability at any one joint decreases with increasing speed (Winter 1984).

3.2.4 EMGs and speed

One group, via EMG, footswitches and videos, related findings on the back muscles and trunk movements at different speeds (Thorstensson et al. 1982, 1984). Trunk balance and maintenance of equilibrium is essential for efficient smooth locomotion. Thus, the findings by Thorstensson et al. (1982) that lumber back muscle contractions, (once per heel-strike, per cycle), with decreasing speed restrict excessive trunk movements are important. These results should be regarded with caution. They used ipsilateral knee flexor angles, which are not related to the stance or swing phases of the human gait cycle, to define phases of the step cycle, including heel strike. The knee flexes in stance before toe-off. A large variability can exist in determining heel-strike from knee flexion angles. The object of their study was to compare the human data with that of cats; however in the two situations the definition of step cycle differs. In addition, in reference to their EMG and movement figure, (figure 2, page 19), they state that the amplitudes of EMGs and movement curves cannot be compared quantitatively. However, other authors believe it is important to relate EMG amplitudes, which reflect underlying muscle tension (Bigland and Lippold 1954), to angular displacement at different speeds to understand adaptive neural control. Yang and Winter (1985) attempted such a comparison. Eleven subjects walked at the set cadence rates of 115, 95 and 75 steps per minute. The researchers postulated that, as muscles function to overcome gravity and control the speed of limb movement, only those muscles related to speed of limb movement, (the hip, and the knee), should be affected by speed changes. Their data substantiated this hypothesis. The EMG linear envelope of specific muscles were compared at the three speeds and later related to kinetic findings. The kinetic data were obtained from a different set of subjects, but the same population as the EMG subjects (although homogeneity of variances was not tested explicitly). The kinetic subjects walked at different, less controlled, cadences of 121, 105 and 84.7 steps per minute. Therefore, intrepretations should be guarded. Nevertheless, the joint moment data supported the EMG data. The muscles at the hip and knee showed amplitude changes in EMG and moment patterns. Their shapes remained the same, but the amplitude changed, especially at weight acceptance and push-off. The changes in muscle activity attributed to decreasing walking speed were a decrease in the linear envelope amplitude by 30% in soleus and tibialis anterior, by 50% in vastus lateralis and by 70%, in rectus femoris.

Although not extensively studied, the normalized on and off timing of EMGs appears consistent with speed changes (Grieve and Cavagna, 1974, Yang and Winter 1985).

Peak EMG amplitudes (Thorstensson et al.1982) and mean amplitudes taken under a linear envelope (Milner et al.1971) are also influenced by speed. Both progressively increase with velocity, each muscle showing a unique relationship with speed.

The muscles of the hip and knee are more sensitive to change of speed than the ankle (Brandell 1977, Winter 1983). The EMG linear envelope pattern at the hip and knee vary more than at the ankle for reasons similar to those discussed for kinetic parameters. The EMG variability, judged by linear envelope, increases with increasing speed. Hershler and Milner (1978), found EMG amplitudes were less variable if the subjects walked at a comfortable speed within a set range.

3.3 Different Loads and Gait

The response in gait to increased or decreased loads has been examined (Hewes et al. 1967, Pierrynowski et al. 1981). The effects are often inconclusive or incomplete as the definition of load and its placement differ widely depending on the study objectives. Load may be defined physically by the addition (Pierrynowski et al.1981) or subtraction of weight, sometimes called gravity (Hewes et al.1967), or load may be defined physiologically as increased muscle stretch, or exertion (Brandell 1977). The position of the loads also varies. Neumann and Cooke (1985) examined the effect of load and carrying position on the EMG linear envelope amplitude (normalized to maximum voluntary isometric contraction) of gluteus medius during walking. They indicated that the position of the load determined the amount of increased EMG activity. Inman et al. (1981), in their review of gait, stated that loading the body increases the metabolic cost of walking, but the effect will be greater the more distally the loads are placed.

Increased loads and gait; Pierrynowski et al.(1981), studying load carrying devices, found no alterations in kinematic or kinetic gait patterns, (EMG were not included), when subjects carried loads of 1.5 to 33.85 kg in a backpack.

Medeiros (1978) analysing the EMG on/off times, duration and mean amplitude found little difference between the loaded (15% of body weight) and unloaded state of subjects walking on a treadmill. EMG changes under various loads were also studied by Soderberg and Dostal (1978). They examined the role of gluteus medius during crawling, walking and stair climbing. Houtz et al. (1959) looked at ankle muscular activity under different physiological loads as defined by raw EMG recordings and extrapolated the findings to gait. Dietz and Berger (1981), with the linear envelope, and Brandell (1977), with integrated EMGs, studied the EMGs of gastrocnemius and vastii muscles as related to angular displacement while subjects walked on an inclined treadmill. Norman and Winter (1980) studied a number of muscles in their mechanical and metabolic analysis of men carrying loads. Relatively few muscles, (except for the study by Norman and Winter 1980), were examined in the preceeding studies. The data obtained were not always related to displacement nor was a detailed EMG analysis of their amplitude often performed. For example, Dietz and Berger (1981) and Brandell (1977) related EMGs to kinematics, but the amount of "load" is difficult to quantify. In another study (Soderberg and Dostal 1978) load was only presumed, and the muscles' function changed during the study. It should be noted that the purpose of many of the previously mentioned works was not to quantify gait changes under loads. Hence, drawing extensive conclusions on the effects of increasing loads is difficult.

All of the above authors concluded that increasing work loads correlated with increased EMG activity. Either the timing of EMG raw bursts changed (Houtz et al. 1959, Soderberg and Dostal 1978), or the magnitude of the signal did (Brandell, 1977, Dietz and Berger 1981).

Decreasing loads and gait; NASA commissioned studies into the effects of lunar gravity, (i.e. decreased loads), on walking and running. Simulation of lunar gravity was problematic, but Hewes et al. (1967) developed a unique, if unconventional, walkway set at

9.5° to the vertical. This walkway provided, in the plane of progression, a gravitational effect equal to that of the moon. Three subjects, supported by slings, walked and ran at various freely chosen speeds up to their maximum at both gravitational conditions. All the subjects walked and ran 60% slower than normally and a loping gait (3 m.s-1) was the most natural method of locomotion at simulated lunar gravity. The postulated reason was a decrease in weight and traction.

Hewes et al. (1967) also indicated that reduced gravity decreased the amplitude of hip, knee and ankle angular movements and increased the forward inclination of the body with increasing speed. The time history graphs of joint motion are confusing and, without normalization, comparison across speed and gravitational conditions is difficult to intrepret. The shape of the curves appears normal. It is unclear from the methodology whether the subject's data were compared at similar speed under each condition. Hewes et al. (1967) stated that the variability of the measures increased under lunar conditions. Data were not provided to support this statement. They demonstrated a definite trend although only a preliminary report. Clement et al. (1984) support the findings of Hewes et al.(1967) that decreasing gravity increases the forward lean of the body. Winter's discussion of gait (1983) reinforces Hewes variability results that as speed decreases variability increases. The separation of speed and weight factors was not their mandate; therefore, the influence of each factor was not evaluated.

A theoretical model of human locomotion in subgravity, by Margaria and Cavagna (1964),showed that a change in weight and inertial forces would be the main factors responsible for locomotor changes at reduced gravity. Assuming the inertial forces remained the same while body weight decreased (as reported by Hewes et al. 1967) and the vertical component of push-off equalled body weight then while walking under lunar gravity, the force at push-off will be less. A decrease in force at push-off would produce a proportional decrease in potential energy and kinetic energy with which to over come inertial forces. Forward speed would, therefore, be slower on the moon. Alternativaly, if the vertical component of push-off is to be greater than body weight (which is normal), there should be an increase in forward inclination of the trunk with an increase of speed, and a decrease in ground contact time. The increased speed and decreased contact time would imply a decrease in cadence and an increase in stride length. With reduced gravity the only change would be due to inertia and the speed would be slower. The NASA data indicated a reduced speed with a decrease in cadence and an increase in stride length which supported Margarita and Cavagna's model prediction. Additional evidence from EMG and kinetic data to test the theory would be valuable.

3.4 Treadmill Gait

The treadmill has been used in a variety of human physiological studies, including gait (Brandell 1977, Dietz and Berger 1983). The reports on gait have been descriptive (Brandell 1977), comparative (comparing the kinetics of treadmill walking to overground (Taves 1982), or analytical, defining the habituation process (Charteris and Taves 1978).

Habituation to the treadmill by a subject is required to compare treadmill data to normal walking. Charteris and Taves (1978) investigated habituation and demonstrated a marked initial stride to stride variation for 10-15 minutes. Later, Wall and Charteris (1980) observed different habituation periods for varying treadmill speeds. The degree of variability was velocity dependent normal walking speeds showed the least variability and lower speeds the most. Whether this variability is greater than the variability attributed to speed alone is not known. Arsenault (1982) compared the EMGs of treadmill walking to overground walking. The reported differences were small.

The advantages of the treadmill in gait can be summarized as follows:

1. variable speeds allow for stimulation of muscle activation,

2. effective stretch input provides stimulation of gait stepping mechanisms; and

3. instrumentation is not required to follow the subject.

3.5 Gait Retraining in pathological gait

Gait re-education programs include the three basic components of locomotion: postural stability; balance; and the ability to alternately flex and extend the lower limbs or step.

The conventional regimes concentrate on the preparation for walking and often devote less time to the actual retraining of gait (Bobath 1978, Brunstruum 1965, Johnstone 1983). A progressive set of balance and postural exercises are practised first while in lying then while in standing to develop the basics of gait. Some therapists advocate early resumption of the upright position, but often do not allow patients to walk until they acquire balance and the control of their limbs (Bobath 1978, Kottke 1982). Initially, some external stabilization may be necessary to support the patient in this upright position. The patient may concentrate on developing motor patterns without the added effort of maintaining balance. (Lehman 1982).

Once balance and postural stability are achieved, the components of walking are taught. The elements of stance and swing necessary for safe, efficient gait are learnt separately, then integrated into a gait pattern. Sensory inputs are frequently used to facilitate voluntary efforts or inhibit unwanted movement

Balance, postural stability and stepping are not sufficient to develop walking. The ability to bear weight through the affected limb and transfer weight from one limb to another are essential pre-requisites for ambulation. Optimal or even adequate weight bearing on the limbs during gait is not always achieved (Wannstadt and Herman 1978).

Treatment concepts have long advocated the need to improve the weight bearing capacity of the hemiplegic limb (Brunstruum 1965, Bobath 1978, Johnston 1983). Wannstedt and Herman (1978), using Krusen limb load monitor feedback, trained patients' weight bearing patterns in standing. Only those patients who were successful during the initial session could control their weight bearing ability without the feedback at the end of training. Hockerman et al. (1984) found an improvement in weight distribution during stance after platform training of hemiplegics. This improved weight distribution enhanced their postural stability in standing. If the amount of weight bearing in gait increased, paraplegics, treated with functional electrical stimulation, were found to have more normal TD values (Mizrahi et al. 1985). Bogardh and Richards (1981) objectively quantified the effects of another weight bearing treatment regime. The emphasis throughout treatment was placed on hip control to facilitate weight transference during stance. Although only an observational analysis of the EMG amplitudes and angular displacement was presented, post-treatment effects revealed improved stance control of knee flexion-extension, increased ability to support body weight normally, and a more adequate and smooth weight acceptance phase. These approachse often meet with limited success, especially in those patients with markedly increased tone who are unable to cope with the unmodulated or uncontrolled stretch of full weight bearing. Despite adequate muscle activation (Knutsson and Richards 1979) and underlying abilities (Dietz et al. 1981), gait deviations can persist even after periods of treatment.

All of the above studies trained or evaluated their patients statically with both feet on the ground. A successful training method is needed for dynamic single limb stance balance combined with gait training. A technique of supported partial-weight bearing that allows for a progression to full weight bearing may be of benefit. The amount of load carried, that is, the amount of stretch put on the muscle (especially the gastrocnemius), could be controlled to meet the patient's capabilities. The retraining of load compensating mechanisms could then progress smoothly. In addition, the patient, once supported, could deal with the demands of controlling balance at his own pace. Gait training, based on proper increased weight bearing, may benefit not only that group of patients with increased tone, but also that group with weak or poor muscle activation.

However, this procedure only helps to train posture and balance. The stepping mechanisms required in walking are not stimulated and other peripheral inputs may be necessary.

Peripheral afferent stimulation has been found to influence locomotor patterns in cats, as has the stimulating effect of the treadmill (Rossignol et al. 1981). Speculation has been made on the effect of cutaneous inputs on human gait. Pierrot-Deseilligny et al.(1983) stated that the cutaneous stimulation of the sole of the foot, in contact with the ground during locomotion, depressed the Ib pathway to the motor neurons supplying the knee muscles. They further speculated that foot contact could play a role in switching the Ib effects to either facilitate or inhibit knee muscles, depending on the phase of the gait.

A combination of inputs may be needed to effectively retrain the complicated movements required for locomotion in patients.

3.6 Spinal Animal Model as a Basis of Gait Training

A spinal animal model exists that demonstrates that recovery of locomotor function is possible. Adult spinalized cats trained by Rossignol et al. (1984) recovered the ability to walk on a treadmill. Other experimenters (Smith et al.1982, Eidelberg et al.1980) had difficulty in restoring adequate locomotor function to adult spinalized animals. The inadequate recovery may be the result of poor or insufficient training, or too long a delay after spinalization before training started, (upwards of one week in Smith et al.1982). Edgerton et al.(1983) stated that, without treadmill training, his 12-week old cats could not learn to walk. The specific training techniques are not described explicitly in most papers. They merely state that the animals were supported, amount of support unknown, and that the animal walked on the treadmill. Some did not even mention training, merely evaluation sessions which may or may not be consistent with training.

Previously, Shurrager and Dykman (1951) demonstrated that the recovery of locomotion in kittens was related to the type of training received. The apparent success of Rossignol et al.(1984) may be due in part to their interactive training technique. The effective technique utilized early graded weight support. The amount of support was decreased as the animals were capable of proper foot placement. A greater emphasis on proper weight support, adequate positioning to minimize postural defects, and the treadmill facilitating the stepping action, accelerated the animals' recovery of locomotor functioning. The force produced by these cats while walking is not known. The treadmill may be stimulating a reflexive walking pattern that in man would not allow forward progression.

These spinalized animal studies, nevertheless, indicate that treadmill training with partial weight progressing to full weight bearing allows for recovery of locomotion after transection of the spinal cord.

Before these findings can be extrapolated to man, normal gait studies on the effects of varied weight loads, in conjunction with treadmill stimulation, are needed. A training strategy can then be postulated and validated.

4. MEHODS AND MATERIALS

4.1 The Gait Laboratory

The experiments were conducted in the Human Gait Laboratory, School of Physical and Occupational Therapy, McGill University. Figure 5 is a picture of the laboratory, equipped with a treadmill, TV video equipment, EMG recording equipment, a weight support system and a PDP 11/34 computer. The apparatus used is described in the procedures section.

4.2 PRELIMINARY INVESTIGATIONS

4.2.1 Protocol Determination

Seven volunteers were used to determine the body weight support levels (BWS). After walking at different BWS levels, of the 7 subjects, only 3 could walk with proper heel contact at 80% BWS. Therefore, the upper BWS level was set at 70%. Thirty and 50% BWS were arbitrarily defined as the middle and lower limits.

At each BWS level tested, the subjects were unable to walk at their natural, full weight bearing (FWB). Subjects where then allowed to chose sucessively slower speeds at increased BWS levels. (The freely chosen mean treadmill speed for each BWS level is in appendix 1.). The time taken to determine a comfortable speed at each level was 10 minutes. The subjects walked at every BWS level, and 4 of the 7 subjects were retested 3 to 4 times. At increasing BWS levels the time taken for habituation was excessive and proved extremely uncomfortable at the 70% BWS level. In addition, subjects who were habituated to the treadmill walked consistently faster at each BWS level, compared to the other subjects. Hence, it was decided to habituate the subjects at 0% BWS (which is equivalent to FWB) and dictate a range of walking speeds for each BWS level. The set speed range limits for each level were as follows: for 0% BWS, 1.2-1.5 m.s-1; for 30% BWS, .90-1.00

Figure 5 Human Gait Laboratory
The subject is supported by a harness over a treadmill, with EMG and video recording equipment in the background and camera in the foreground.

m.s-1; for 50% BWS, .79-.89 m.s-1; for 70% BWS, .65-.75 m.s-1.

Two independent variables were introducted in this way. Therefore, to dissociate changes due to slower walking speed and changes due to amount of weight supported, the subjects walked first bearing their full weight, at three different set speeds, and then at the BWS levels of 0, 30, 50, or 70% and four set speeds. The set speeds, in m.s-1, were matched, that is each subject walked at .90-1.0 FWB and 30% BWS; at .79-.89 FWB and 50% BWS; and .65-.75 FWB and 70% BWS, as well as at 1.20-1.50 at 0% BWS.

4.3 CHOICE OF MUSCLES

The present investigation examined not only muscles representative of each joint involved in gait, but also muscles of interest in training pathological cases. The selection of muscles was based on their function, amount of variability and least possibility of cross talk between them. The following muscles on the right side were chosen; lumbar erector spinae, gluteus medius, vastus lateralis, medial hamstrings, tibialis anterior and medial gastrocnemius.

4.4 SUBJECTS

Ten male volunteers took part in the study. All subjects were novice, (less than 2 hours experience), treadmill walkers. All subjects wore shorts and a similar type of running shoe. Anthropometric data was collected and recorded on two separate sheets, (appendix 2.). The first included the subject's age, height, weight, marker placement and inter-marker distances (Winter 1979). The second document was a modified physical examination; to rule out potential cardiac problems, and to control for confounding effects of previous back, or lower limb injury. The mean age, height and weight of the subjects are presented in table 1. None of the subjects had a pelvic

	Mean	S.D.
Age (yrs)	31	3.68
Height (M)m	1.76	.04
Weight (Kgm)	72.73	7.08
Leg length (M)m	.88	.03
Foot length (M)m	.26	.01

TABLE 1 - Mean anthropometric data with standard deviations (SD) for the 10 subjects studied.

asymmetry, a leg length discrepency greater than 1 cm, a history of foot, ankle, knee, hip, or back problems, excessive hypertension or cardiac ailment. All procedures were explained and an informed consent signed, prior to each experiment (appendix 3.).

4.5 EXPERIMENTAL PROTOCOL

Data for each subject was collected in a single three hour testing session. Table 2 representes a typical session.

4.5.1 Training Session

Preliminary to data collection, the subjects were habituated to the treadmill, a W.E.Collins #101, (1.15 m long by .37 m wide), with a range of speeds from .26 to 2 m.s-1. Each subject walked for 20 minutes on the treadmill increasing their speed over a 5 minute period from .26 m.s-1 to a comfortable speed within the set range of 1.2 to 1.5 m.s-1. Each subject controlled the speed selection himself.

Although both legs of the subject were instrumented, information collected from the right leg only will be reported in the present document.

4.5.2 Muscle Activity Recording

Simultaneous recording of EMG activity was obtained via two electrodes, (Meditrace Pellet electrodes). These were applied 2 cm apart, center to center, longitudinally to the direction of the muscle fibers, on the skin centered over the muscle belly. The placement on the investigated muscles is as follows: (the electrode position follows the name of each muscle) erector spinae (ES), 2 cm lateral to and in line with the 4-5 lumbar disc space; gluteus medius (GM), 4 cm posterior to and in line with the anterior superior iliac spine and 2 cm below the iliac crest; vastus lateralis (VL), 10-12 cm superior to the upper edge of the patella and 5-6 cm lateral to the superior mid-line of the thigh; medial hamstrings (MH), posterior medial 1/3 of

TRIAL	BWS	Speed
ONE		
	0	1.20-1.50
	30	1.00- .90
	70	.75- .65
	50	.89- .79
TWO		
	FWB	1.20-1.50
	FWB	.89- .79
	FWB	1.00- .90
	FWB	.75- .65
THREE		
	70	.75- .65
	50	.89- .79
	30	1.00- .90
	0	1.20-1.50

TABLE 2 - Typical random experimental protocol for one subject with body weight support (BWS) and full weight bearing (FWB) speed ($m.s^{-1}$) trials.

the thigh, (no attempt was made to differentiate EMG activity from semimembranosus and semitendinosus); tibialis anterior (TA), 2 cm lateral and 4 cm below the tibial tubercle; medial gastrocnemius (GA), 2 cm superior to the lower edge of the muscle and 5 cm medial to the center line of the calf. One surface electrode, used as a ground, was placed medially on the right leg over the bony surface of the tibia. Leads, (incorporating a buffer, encased in epoxy, to decrease low frequency artifacts) were snapped onto the electrodes, taped to the skin and inserted sequentially into a control box. The EMG signals from the box were pre-amplified, 10 times, (custom designed pre-amplifier, with CMRR = 92dB; impedence > 100M ohms; band-width 1 to 10 KHz, -3dB), before relay to an EMG differential amplifier (also custom made with the same specifications as the pre-amplifier).

4.5.3 Temporal Parameter Recording

To relate EMG signals to the gait cycle and to collect temporal gait parameters three footswitches, (Tapeswitch systems of America), were taped to the sole of each shoe at the heel, fifth metatarsal - phalangeal joint and great toe. These footswitches recorded heel-strike and toe-off of each limb.

The subject then walked at his FWB chosen set speed for 10 minutes. The EMG, and footswitch signals were checked on the Nihon Kohden monitor, (VC 680g, 8 channel; CMRR > 28 dB at 1 KHz) while the appropriate gain for each muscle was set. A trial session was then recorded on the Honeywell FM, 16 channel, tape recorder at 3.75 ips (mid-band recording level 2500 Hz). After verification of the signals, the subject was placed in a modified Tyrollian climbing harness with extra padding. The subject's ability to walk comfortably without hinderance was assessed as was his ability to freely move his lower limbs through full pain-free range. The low pass filter was set at 10

Hz and the high pass at 1 KHz for all signals. Because of baseline artifacts the low pass filter was set at 30 Hz for ES in four subjects for each session and for GM in one subject for each session.

4.5.4 Angular Displacement Recording

To record saggital angular displacement data half a ping-pong ball, filled with polyfoam, marked with a 1 cm dot at its center, was attached to the outside of the right ankle, knee, hip, thorax or shoulder, and to the inside of the left ankle and knee of the subject. In addition, large 3 cm reflective dots, centered with 1 cm dots, were attached to the subject's shoes at each heel, 5 th, or first metatarsal-phalangeal joint and 2 cm above the sole in line with the front of the shoe. A camera, (Sony Rotary Shutter Camera #1010, exposure time 0.2 ms), was placed laterally on the right side, 4 meters from the center and perpendicular to the treadmill to record movement on 3/4 inch video tape at 60 fields per second (Panasonic Video Cassette Recorder NV-9240). A black screen provided the background for videotaping. Lighting, (Quartz Studio Lite #71052), was adjusted prior to recording and the video monitor screen checked for clarity of picture.

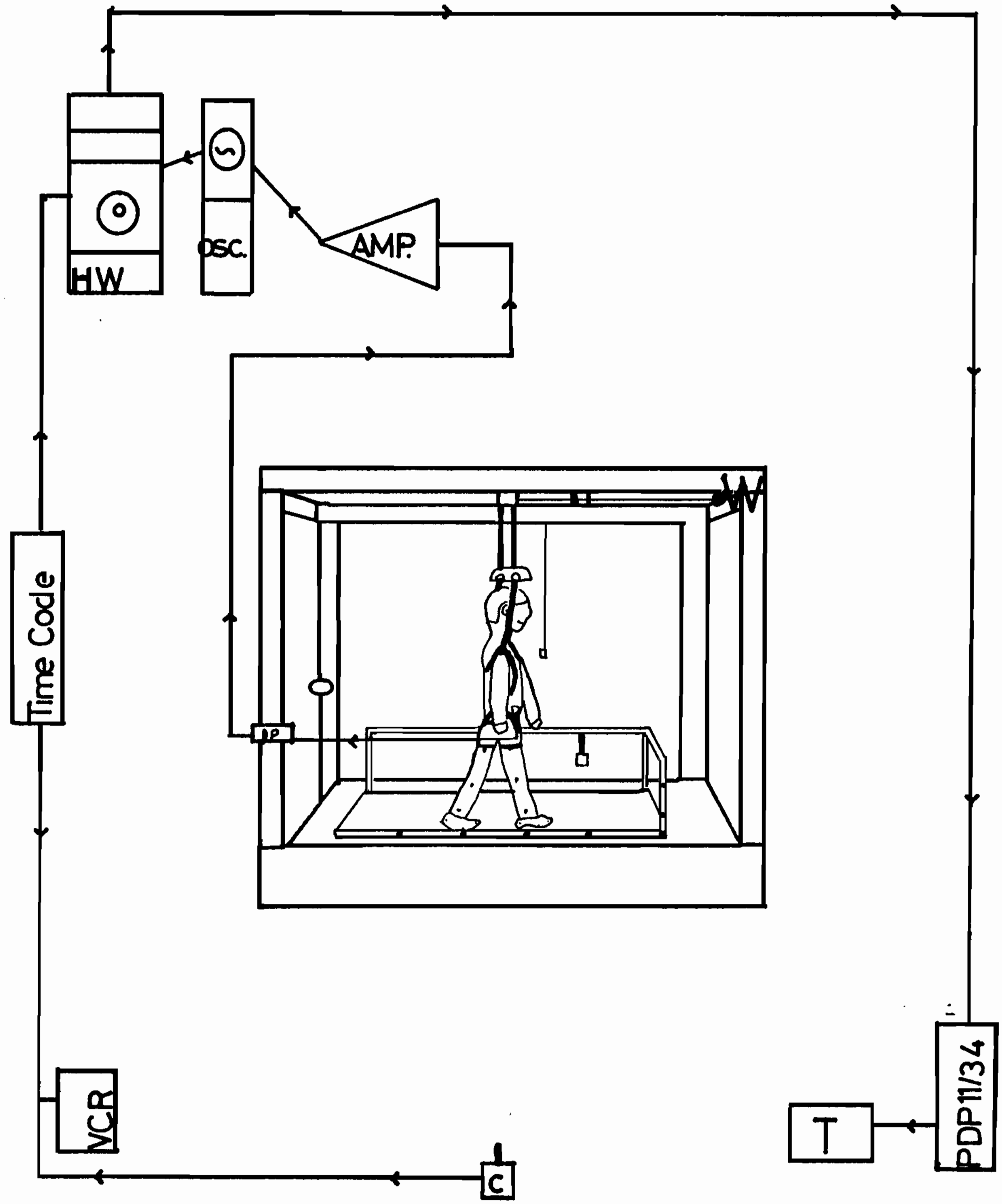
4.6 EXPERIMENTAL TRIALS

Each subject participated in 3 trials. The first trial was a preparatory BWS trial and was not analysed, the next trial was a FWB speed trial and the last another BWS trial. A latin square design was used to randomly assign BWS levels and FWB speeds. Subjects walked at the set speeds for each BWS level and at the same speeds bearing their full weight.

Throughout the weight support trials, the weight support system (a motor driven pulley system) was attached, via straps and quick release hooks to the harness on the subject. The weight support system

Figure 6 Schematic flow diagram of data acquisition and processing. Data flows in the direction of the arrows

HW	Honeywell
OSC	Oscilloscope
VCR	Video recorder
C	Camera
T	Transiac
AMP	EMG differential amplifier
P	Pre-amplifier
W	Weight support system



HW
OSC.
AMP.
Time Code
VCR
C
T
PDP11/34

transducer was calibrated to within 2%; and the transducer output voltage set at .5v per 45.4 kgm. The amount of weight supported was calibrated separately for each subject. The experimenter set the weight dial to 100 with the subject totally supported in the air; and set the dial to 0 with the subject fully on the ground. At each BWS level, the subject was first totally supported while the dial was rechecked and then lowered to the appropriate weight level. The subject then ran on the spot to remove any slack in the system and, if necessary, the BWS level was re-adjusted.

One to two minutes of footswitch, EMG and video recordings were obtained at each BWS level and speed after the subject was comfortable. To avoid fatigue, 10 minute rest periods were provided between each session. Blood pressure and pulse recordings were monitored throughout the experiment to ensure the subject was not under stress.

To ascertain if the height of the body fluctuated during the experiment, the height of 5 subjects was measured at each BWS level from the trochanter, to the floor. In addition, in two subjects the distance between the toes of the left foot and heel of the right foot was measured while both feet were on the ground. This distance is defined as the contact distance.

4.7 DATA PROCESSING

Figure 6 is a schematic representation of the laboratory with a flow diagram of data aquisition and processing. Raw EMG and foot switch signals, from the FM tape, were played back at recording speed onto the oscilloscope, and selected portions digitized by computer with a sampling rate of 1 KHz.

4.7.1 Temporal Distance Factors

Interactive computer programs were used to display all data channels, on a high resolution terminal (Transiac) in 10 second sections, and to detect the cycle components of stance, swing and double support time. A minimum of 10 cycles were chosen for averaging from each BWS and FWB speed trial, based on the clarity of footswitch signals and absence of movement artifact across all channels. Heel-strike to heel-strike was considered 100% cycle time for each stride selected. Percent stance time was calculated from absolute stance time divided by cycle time. The two double support times (DST) (right to left and left to right) were summed to yield a total double support time (TDST), which was then normalized to cycle time for statistical analysis. That is; right DST/cycle time + left DST/cycle time x 100% = TDST

The number of cycles in 3, 10 second sections of the screen were averaged to determine cadence.

Stride length was calculated from the video recordings. The mean number of frames per cycle (n=10) were multiplied by time per frame (.01 s) and multiplied again by the treadmill speed. That is; # frames x .01 (s) x treadmill speed (m.s-1) = stride length (m).

4.7.2 Angular displacement data

To obtain saggital displacement data the segment of video tape corresponding in time, determined by a Time Code generator (Skotel TCG-80), to the EMG segment was viewed on the video monitor frame by frame using a remote search controller (NVA 505), and one cycle, per subject, per trial was processed by hand. Sampling rate per trial was between 25 and 30 Hz (considered sufficient for kinematic data, Winter 1982). Figure 7 depicts the body angles and how they were measured. A protractor was centered on the marker over the joint to be measured. The angle between the two limb segments was determined as the angle between the two lines formed by joining the joint marker to

Figure 7 Illustration of various body angles. All angles are positive as shown.

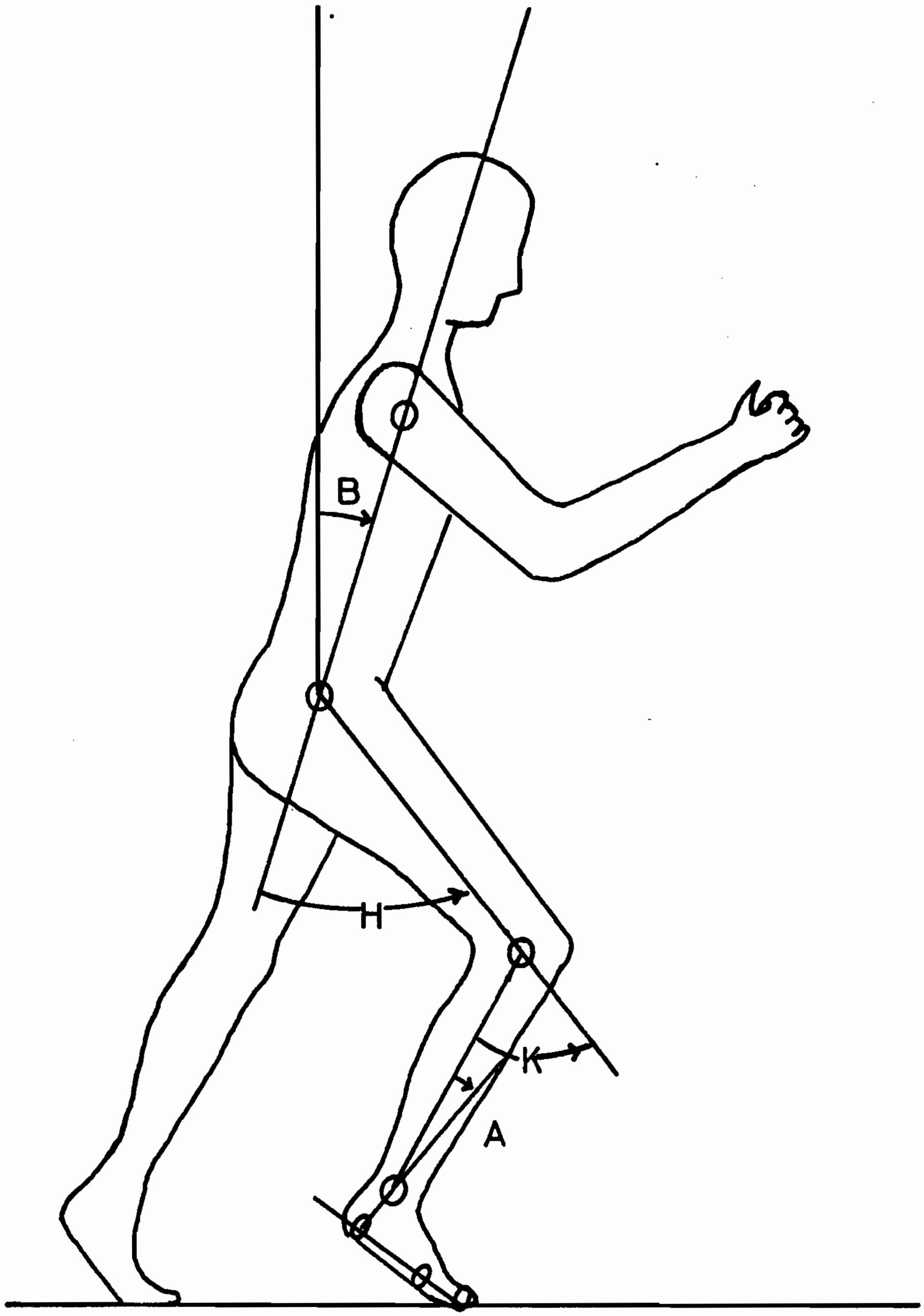
A Ankle

B Body

H Hip

K Knee

Adapted from Hewes et al. (1967)


B
H
K
A

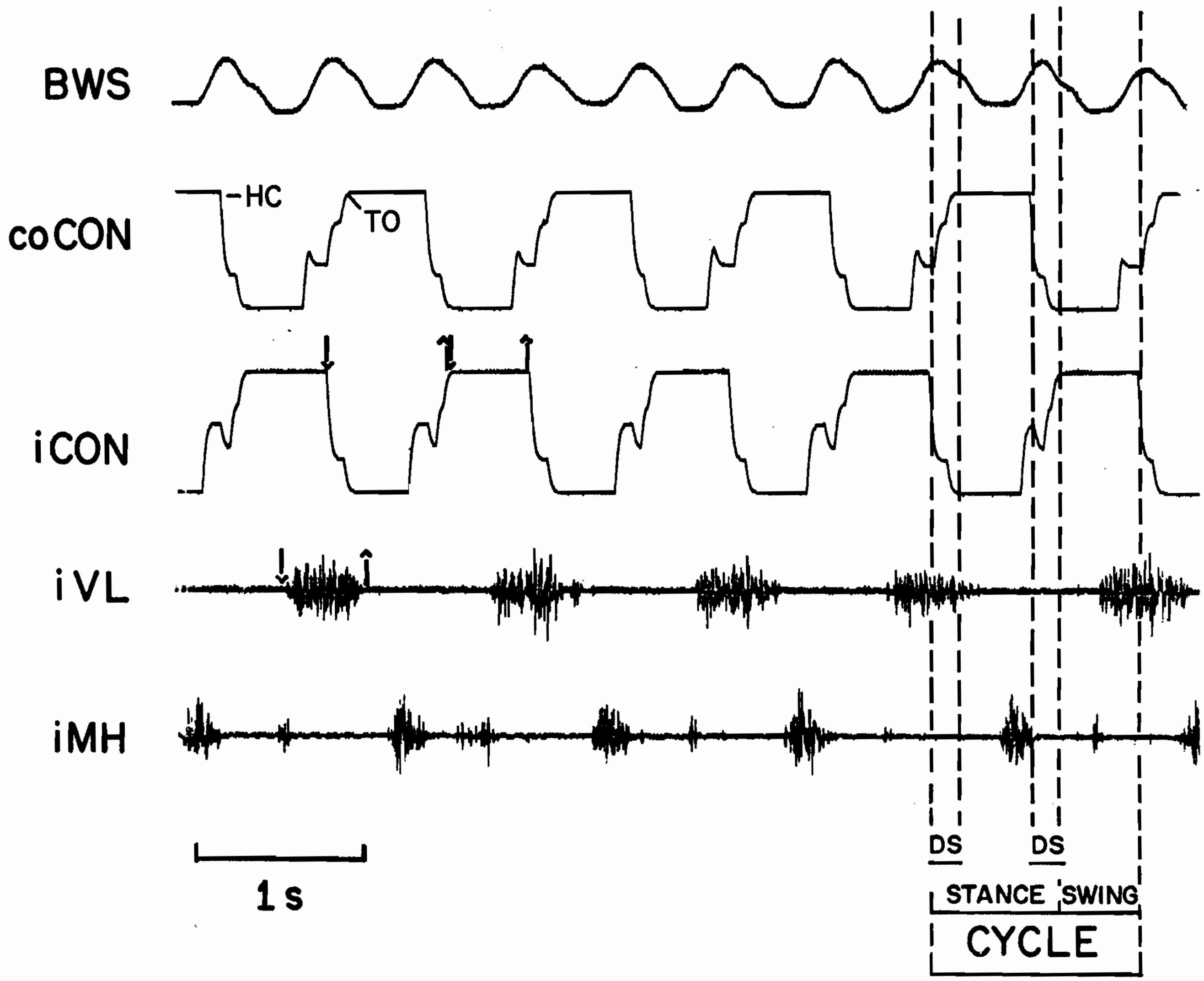
the marker on the joint above, and joining the joint marker to the marker on the joint below. All three joints were measured, but only the hip, and knee displacement will be reported in this document. The angular positions attained at the critical events of heel-strike, foot-flat, mid-stance, toe-off, and the maximum flexor swing angle, were plotted on a relative time scale. From this data, the total range of movement for each joint was calculated. The accuracy of the angular measurements using this technique was considered to be about +/-5 degrees, which was felt to be adequate for this investigation. The ankle was an exception to this fact, the amount of ankle movement was small and the error large in the calculation of joint movement history. In addition, the amount of arm swing, trunk rotation, and general quality of gait was subjectively recorded from the video tapes.

4.7.3 EMG Activity

Figure 8 is a representative example of raw EMG activity observed from muscles of one subject at 30% BWS. Interactive computer programs (Zomlefer et al. 1984) allowed placement of arrows, by hand, to define onset and offset of each EMG burst, 1 to 2 per channel. The EMG potentials between the arrows in figure 8 indicate when vastus lateralis was considered to be "on"; the first arrow indicates the on time, the second the off time. The on/off timing for each muscle was normalized as a percentage of the gait cycle. The normalized onset of a muscle is equal to the time from the immediately preceeding right heel strike to the start of the muscle burst, divided by the cycle time and multiplied by 100. The off time was determined in a similar manner. The burst duration was determined as the total amount of time, in milliseconds, between the two arrows defining a burst.

The amplitude of each EMG burst, from on to off, was determined by

Figure 8 Representative Footswitch and EMG recording. EMG's of 2 muscles of one subject walking with 30% BWS at .97 $m.s.^{-1}$ recorded on a electrostatic pen recorder from Honeywell 100 Magnetic tape recorder. The EMG gain on Vastus lateralis was 5 and on medial hamstrings 1. The high pass filter was set at 10 Hz and low pass filter at 1 KHz The top trace is the transducer readout at 30% BWS followed by the left (COCON) footswitch, right (ICON) footswitch, right vastus lateralis (IVL), and right medial hamstrings (IMH). The arrows in the IVL trace demonstrate the on and off times. The arrows in ICON represent one cycle with first 2 arrows representing stance the next 2 swing. The double support times and 1 second time reference are also included.


BWS
coCON
HC
TO
iCON
iVL
iMH
1 s
DS
DS
STANCE
SWING
CYCLE

the computer as the area under the rectified EMG signal (mVs). To obtain the mean burst amplitude of each muscle for each BWS and speed level, the area under the burst was divided by the burst duration to obtain the mean burst amplitude in millivolts. All mean burst amplitudes at each BWS and FWB speed were then normalized, within subject, to the 0% BWS mean burst amplitude of that muscle. The normalized mean burst amplitudes for each muscle were then averaged across all subjects to facilitate between trial analysis. The number of bursts in the two burst muscles (MH, TA, and ES) at any one speed or weight for any one subject can vary in number. Only discrete bursts were used and only the burst closest to HS sequentially or the first TA, MH and the first and last bursts of ES were analyzed.

4.8 ANALYSIS

Four different analyses were performed using kinematic data, EMG amplitude and on/off timing and footswitch data from the 10 subjects walking at 4 BWS levels and 4 speeds.

First a qualitative analysis of each subject's gait at each BWS compared to the similar speed FWB was performed; noted was the subject's ability to freely move his arms and legs, the amount of trunk rotation used, the change in hip height and the distance between his feet.

4.8.1 Temporal Distance Data

Second, five repeated measures ANOVAs tested whether the mean differences in cycle time, % stance, normalized total double support time, cadence and stride length, recorded from 10 subjects at different BWS (n=4) and speeds (n=4), were statistically significant ($p<.01$) for amoung group effects.

4.8.2 Angular Displacement Data

Third, four ANOVAs for repeated measures were used to determine if the mean range of movement for hip and knee, and if the maximum swing flexor angle of hip and knee were statistically different ($p<.01$) across BWS or speed levels (n=7).

4.8.3 EMG Data

Fourth, Repeated measures ANOVAS were used to test whether the results of the mean differences in normalized mean burst amplitude of EMG activity from each of the 6 muscles from the 10 subjects at 7 different conditions were statistically significant ($p<.01$). The on/off timing of EMG signals will be dealt with descriptively.

An F Max. test (F= largest variance/smallest variance) was used to confirm (appendix 8.) the homogeneity of variance ($p<.05$) for each variable (Snedecor and Cochran 1982). If variances were non-homogeneous a Friedman's ANOVA by ranks, with the Wilcoxon signed rank test as a post-hoc comparison, was used (Huck, et al. 1974). The Scheffe Multiple Comparison test was used as a post-hoc procedure with the repeated measures ANOVAS.

5. RESULTS

The results have been divided into four sections; one general section on the quality of gait at different slow speeds and BWS levels and 3 others one each on; TD, angular displacement, and EMG factors. Each section is further subdivided into speed and weight effects.

All tables report mean data with their standard deviations (SD), while all graphs depict mean data with standard error of the mean (SEM or SD/N). The individual subject data (appendix 4 to 7), ANOVA tables, and post-hoc tests (appendix 9 to 11) are in the appendicies.

5.1 QUALITATIVE GAIT FEATURES

With decreasing speed, from 1.3 to .70 m.s-1, each subject progressively decreased his arm swing. The other features of gait examined, (trunk rotation, height of trochanter, and distance between feet), did not appear to change with decreasing speed.

With increasing BWS levels, from 0% to 70%, the amount of arm swing progressively decreased in a manner similar to that seen with decreasing speed. At 70% BWS, however, 3 subjects' arms swung with the homolateral rather than the heterolateral limb. Other gait features changed at 70% BWS; four of the subjects leant forward 100, two of these four started leaning, (50), during 50% BWS; eight subjects increased their pelvic rotation at 70% BWS, five slightly and three excessively. The upper and lower trunk rotated excessively in the same direction. The average height measured from the trochanter to the floor, in 5 subjects, increased by 1.5 cm. at 70% BWS. The distance between the toe of the left foot and the heel of the right foot was measured in two subjects at each BWS level, and decreased sequentially by 15% at each level from 73.5 cm, at 0% BWS to 22.5 cm. at 70% BWS. Each subject's ability to move his limbs, in any direction, however, was not hampered. An additional feature of BWS trials was comfort, 5 subjects appeared uncomfortable at 70% BWS and complainted of sore shoulders, "pins and needles" in their hands and tight groin straps.

5.2 TEMPORAL DISTANCE RELATIONSHIPS

5.2.1 Cycle Time

The cycle time for the session at 1.36 m.s-1, FWB natural speed or 0% BWS, was 1.08 s. The means and SD are presented in table 3, the graph in figure 9. There was a significant decrease in cycle time with decreasing speed and increasing BWS levels as determined by Freidman's ANOVA ($x2=29.6$; $df=6$, $p<.001$).

Wilcoxon sign rank tests demonstrated that each cycle time, from the FWB natural speed to the slowest .70 m.s-1 speed, decreased significantly from the other (figure 9).

A similar significant difference exists between each BWS levels from 0% to 70% BWS. Significant differences were not found between slow FWB and BWS levels for a given speed. The p values ranged from .04 at the 70% BWS and .70 m.s-1 FWB speed to .33 at the 30% BWS and .97 m.s-1 FWB speed.

5.2.2 Percent Stance

Table 3 reports the 7 means and SD for % stance. A significant difference exists among the 7 means tested by a repeated measures ANOVA ($F=32.59$; $df= 6,54$; $p<.001$).

Although the % stance time increased with decreasing speed, the Scheffe multiple Comparison Test, demonstrated no significant differences between the means across decreasing speed ($n=4$), as illustrated in figure 10.

The % stance time decreased with increased BWS ($n=4$) and in figure 10 appears, paradoxically, opposite to that observed at FWB levels slow speeds. The decrease in % stance from 30, to 70% BWS was 5, 7, and 14% respectively.

Examining the means across increasing BWS levels, the 70% BWS mean

BWS%	0	30	50	70	FWB	FWB	FWB
Speed $m.s^{-1}$	1.36	.97	.85	.70	.97	.85	.70
Cycle Time							
Mean	1084.50	1238.30	1396.10	1680.40	1268.90	1361.60	1527.90
SD	62.10	97.60	163.80	249.50	40.20	82.70	126.40
CV	.57	.78	1.17	1.48	.31	.60	.82
% Stance							
Mean	59.90	56.90	55.60	51.70	62.30	63.10	63.50
SD	2.90	2.30	2.20	5.10	2.60	2.40	2.50
CV	.48	.40	.39	.98	.41	.38	.39
*TDST							
Mean	21.70	17.10	13.40	8.60	27.40	27.70	29.00
SD	4.40	4.40	4.20	5.70	3.80	3.00	3.40
CV	.26	.26	.31	.66	.14	.11	.12
+SLST							
Mean	38.20	39.80	42.20	43.10	34.90	35.40	34.50
++SD	7.30	6.70	6.40	10.80	6.40	5.40	5.90
CV	.19	.17	.15	.26	.18	.15	.17
Cadence							
Mean	111.10	98.30	87.90	73.90	95.60	89.50	78.90
SD	6.20	7.60	7.80	10.90	3.80	3.00	3.40
CV	.05	.07	.08	.15	.04	.03	.04
Stride Length							
Mean	1.30	1.07	1.03	1.05	1.09	1.04	.94
SD	.07	.10	.13	.14	.06	.06	.06
CV	.05	.09	.13	.13	.06	.06	.06

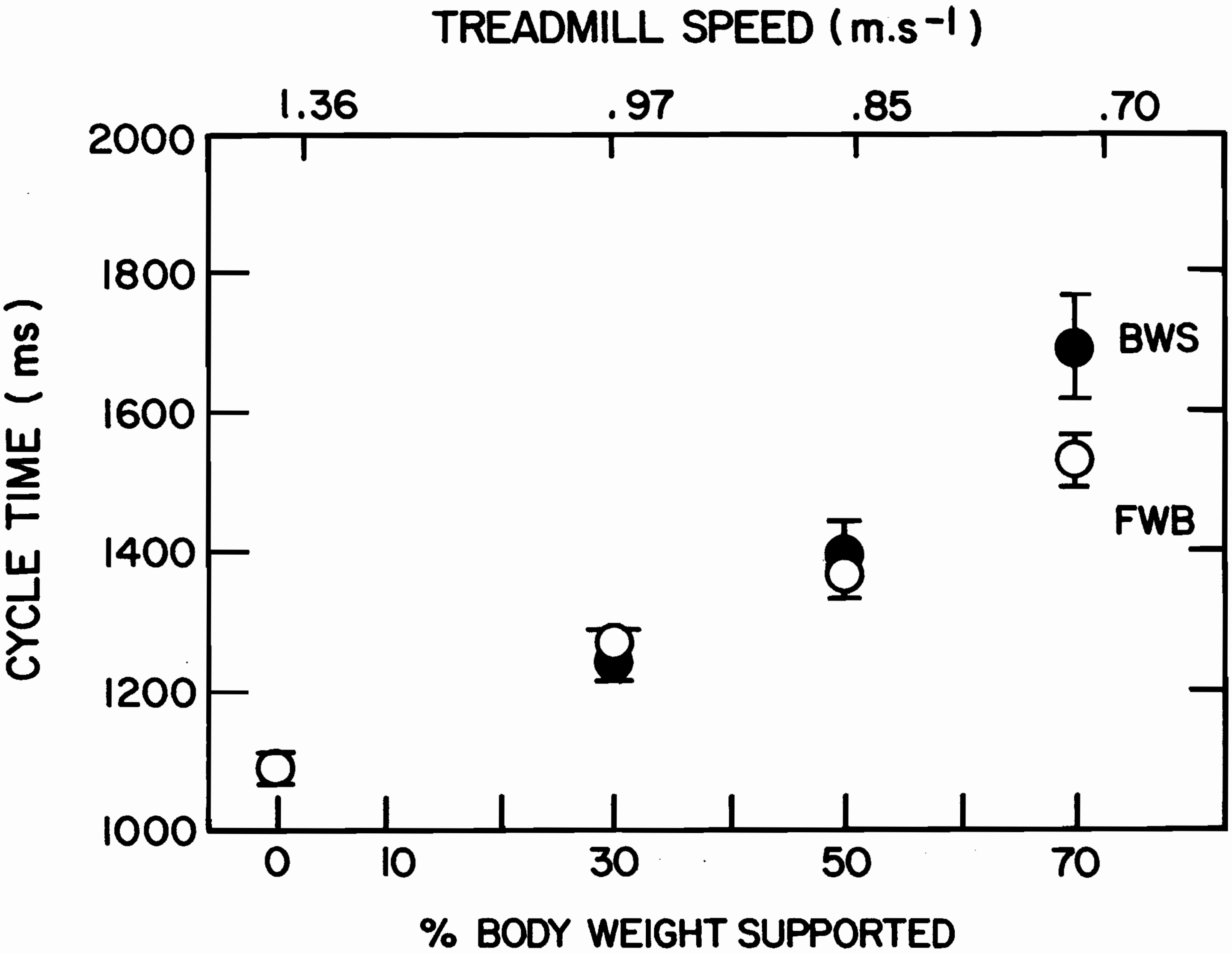
TABLE 3: Temporal distance results. Means, standard deviations (SD) and coefficients of variation (CV) of cycle time (ms), % stance (% of cycle time), total double support time (TDST as % of cycle time), single limb support time (SLST as % of cycle time), cadence (steps/minute) and stride length (m), at each body weight support (BWS) and full weight bearing (FWB) speed. During each BWS session 0, 30, 50 or 70% of the subject's body weight was supported. During FWB sessions, the subjects walked bearing their full weight, but at the same speed as during BWS sessions.

* N=9

\+ SLST means obtained by subtracting TDST from % stance

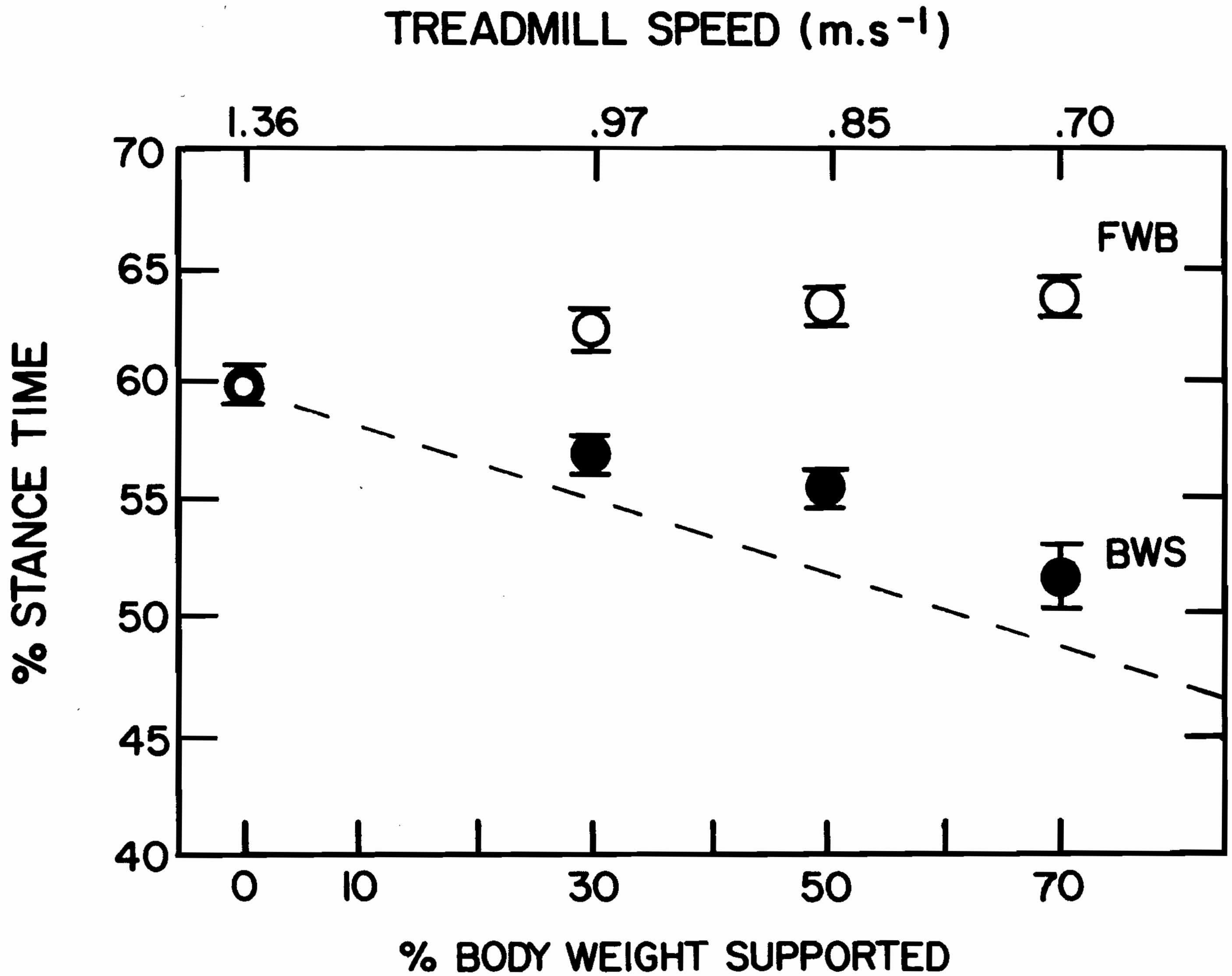
++ S.D. = $\sqrt{\text{variance}_{TDST} + \text{variance}_{\%\ stance}}$

Figure 9 Cycle Time means
As a function of full weight bearing
(FWB) treadmill speed and body weight
support (BWS).



CYCLE TIME AS A FUNCTION OF TREADMILL SPEED AND WEIGHT SUPPORT (± SEM)

Figure 10 Percent Stance means
As a function of full weight bearing (FWB) treadmill speed and body weight support (BWS). The dashed line represents the true weight support effect, that is minus the effect of speed.



PERCENT STANCE AS A FUNCTION OF TREADMILL SPEED AND WEIGHT SUPPORT (± SEM) differed from those at 0 and 30%. All BWS % stance values differed from their FWB means at equivalent speeds and were 10, 12, and 18% less respectively.

5.2.3 Normalized Total Double Support Time (TDST)

The data in table 3 for TDST are from 9 subjects only, one left footswitch recording was unusable. The TDST differed significantly across the 7 means ($F = 42.86$; $df = 6,48$; $p < .001$).

With decreasing speed an effect similar to that of % stance time data was noted.

TDST showed a pronounced decrease across BWS levels (n=4) as noted in figure 11. The percent decrease from 30 to 70% BWS was 21, 38, and 61% respectively and the percent decrease compared to their respective FWB slow speed TDST was progressively 37, 51 and 69%. Across the four BWS levels a significant difference appeared. The 0 and 30%, the 30 and 50%, and the 50 and 70% BWS levels TDST did not differ, while the 0, 50 and 70% and the 30 and 70% BWS TDST did.

It appears that the consistent cycle time produced at the same speed but different weight conditions (BWS vs. FWB) is associated with a decrease in relative double support time (e.g. 27.7% at FWB, .85 $m.s^{-1}$ to 13.4% at 50% BWS, .85 $m.s^{-1}$).

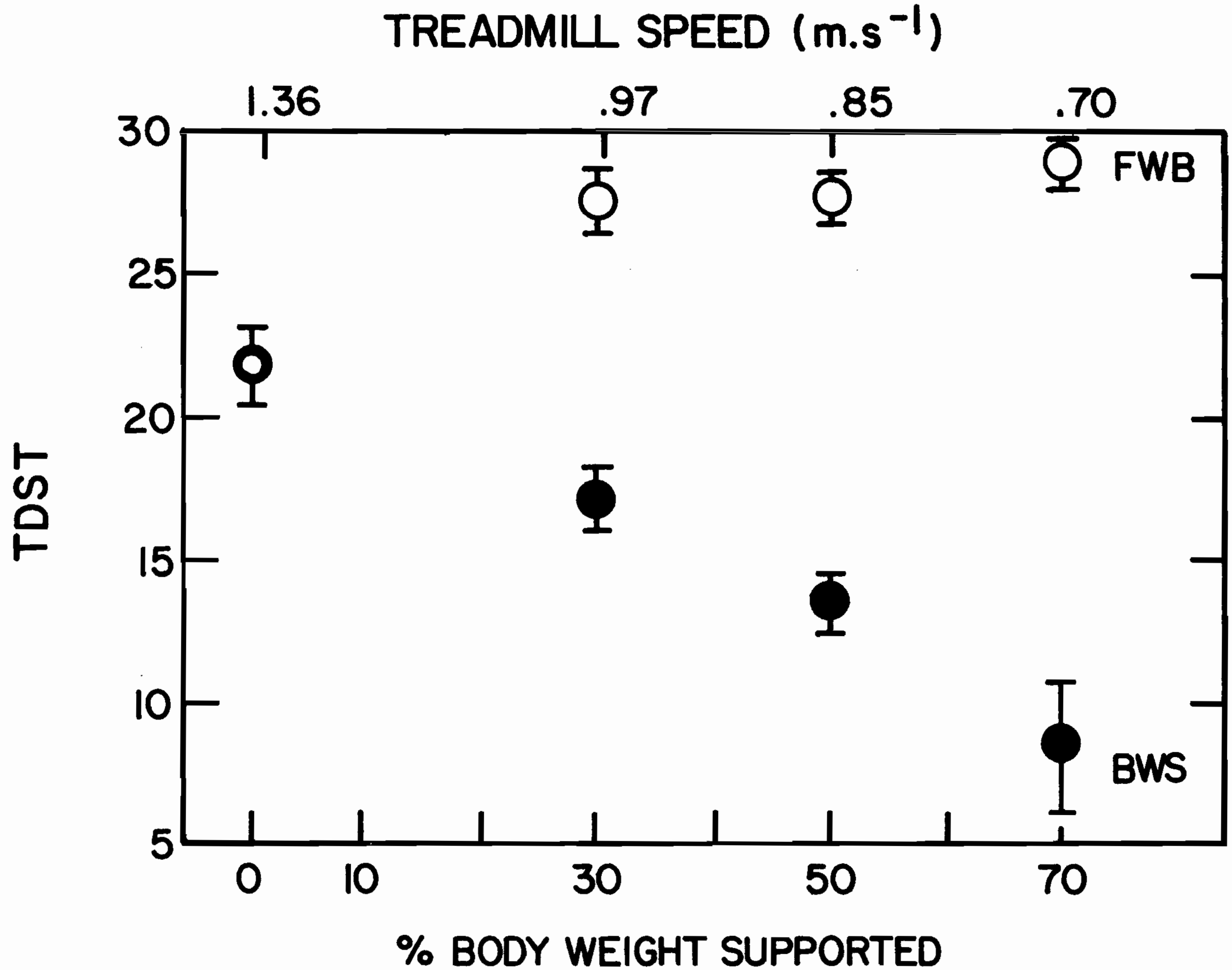
5.2.4 Normalized Single Limb Support Time (SLST)

SLST was calculated by subtracting the TDST values from the % stance values in table 3. The standard deviations were calculated by square rooting the sum of the TDST and % stance variances.

By inspection, the resultant SLST values in table 3 remain consistent across decreasing speed.

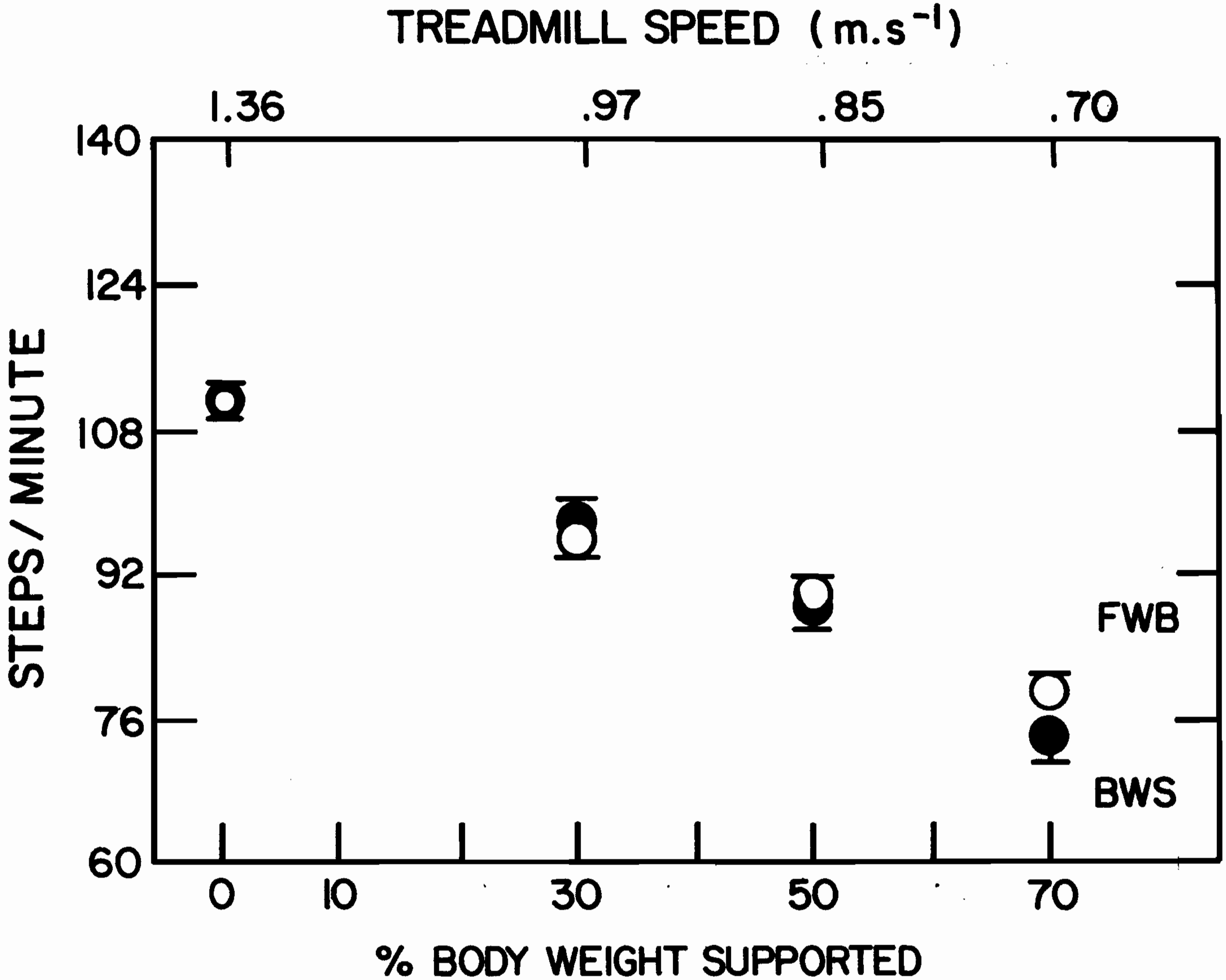
Increasing BWS, however, increased, slightly, the amount of time spent on a single limb from 39.8 at 30% BWS to 43.1% at 70% BWS, an insignificant gain of 4%.

Figure 11 Total double support time means (TDST) as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS)



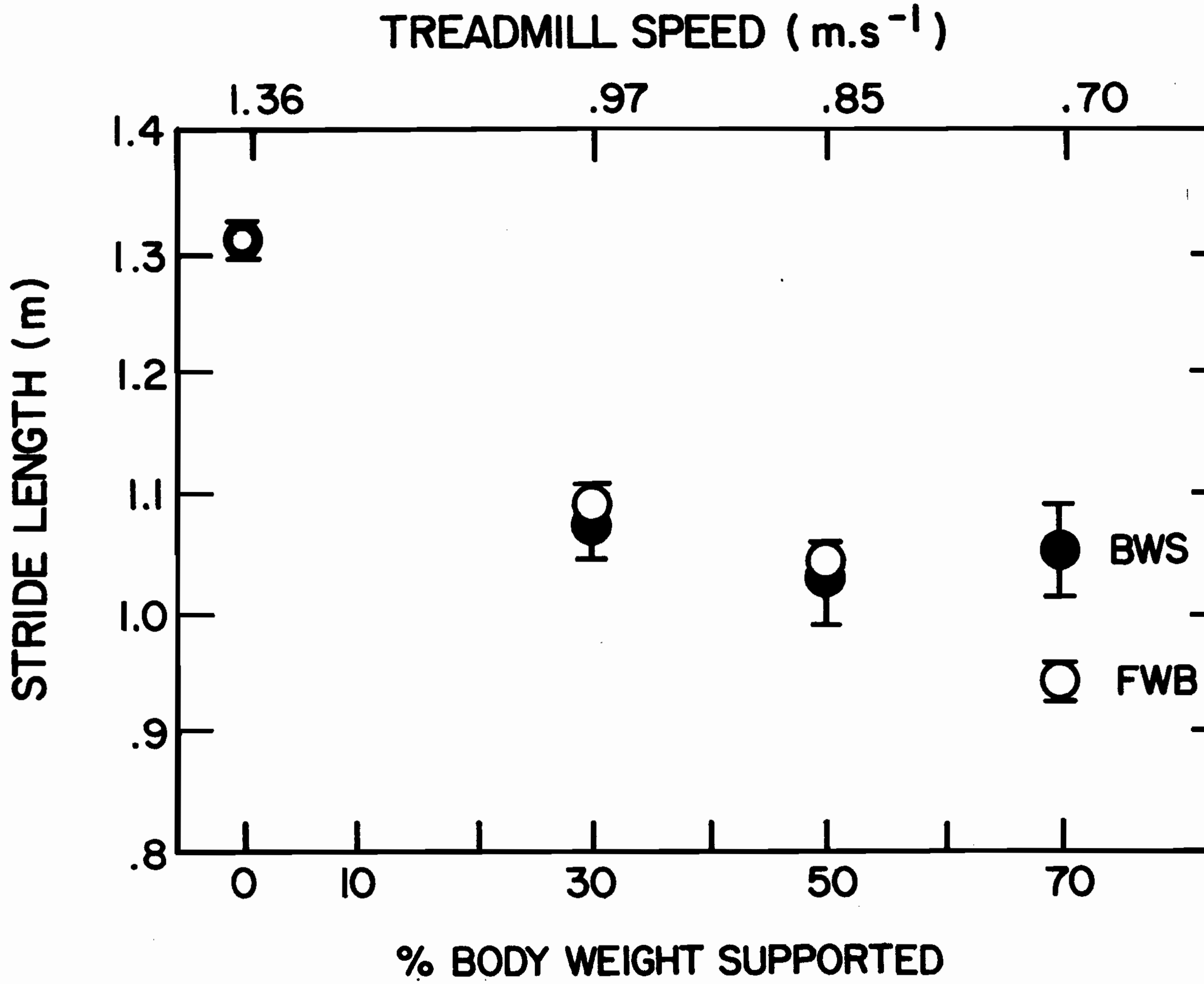
TOTAL DOUBLE SUPPORT TIME VS. TREADMILL SPEED AND WEIGHT SUPPORT (± SEM)

Figure 12 Cadence means as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS).



CADENCE AS A FUNCTION OF TREADMILL SPEED AND WEIGHT SUPPORT (± SEM)

Figure 13 Stride length means as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS).



STRIDE LENGTH AS A FUNCTION OF TREADMILL SPEED AND WEIGHT SUPPORT (± SEM)

5.2.5 Cadence and stride length

The trends for cadence and stride length with speed for the two experimental conditions are noted in figure 12 and 13. Cadence and stride length both differ significantly across the 7 means tested ($F=78.9$, 27.9; $df=6,54$; $p<.005$).

Cadence decreased with decreasing speed as seen in figure 12. Significant differences were noted between all decreasing cadences, except between cadences of 95.6 and 89.5 steps/minute at .97 and .85 m.s-1.

Cadence decreased with increasing BWS, but significant differences (Scheffe) were not found between the slow FWB and BWS levels at equivalent speeds. However, BWS levels were significantly different from each other. These cadence results are similar, both with increasing BWS and decreasing speed, to those of cycle time.

Although a decrease in stride length with decreasing speed exists, only the FWB natural or 0% BWS level stride length of 1.31 m differed statistically from any other ($N=4$), as seen in figure 13.

Stride length decreased with increasing weight support up to the 50% level, from 1.31 m to 1.03 m, then increased again to 1.05 m at 70% BWS. Despite this, as seen with decreasing speed, only the 0% BWS level (FWB natural speed) differed significantly (Scheffe $p<.01$) from the others. A weight effect above that attributed to speed was not noted. Although not significantly different, subjects took longer strides (1.05 m) and fewer steps per minute (73.9) at the slowest speed, highest BWS level, than at the equivalent FWB speed (.94 m, 78.9 steps/min).

5.3 ANGULAR DISPLACEMENT DATA

Figures 14 and 15 plot the hip and knee average angular displacement curves for the BWS and FWB speed trials. Curves were faired through the points to denote trends. The points represent the mean joint

Figure 14 Mean angular hip displacement as a function of gait events. The data is plotted normalized to critical events of gait with cycle normalized to 100%.

HS	Heel Strike
FF	Foot Flat
MS	Mid Stance
TO	Toe off
MSA	Maximum swing angle

MEAN ANGULAR HIP DISPLACEMENT AS A FUNCTION OF GAIT EVENTS

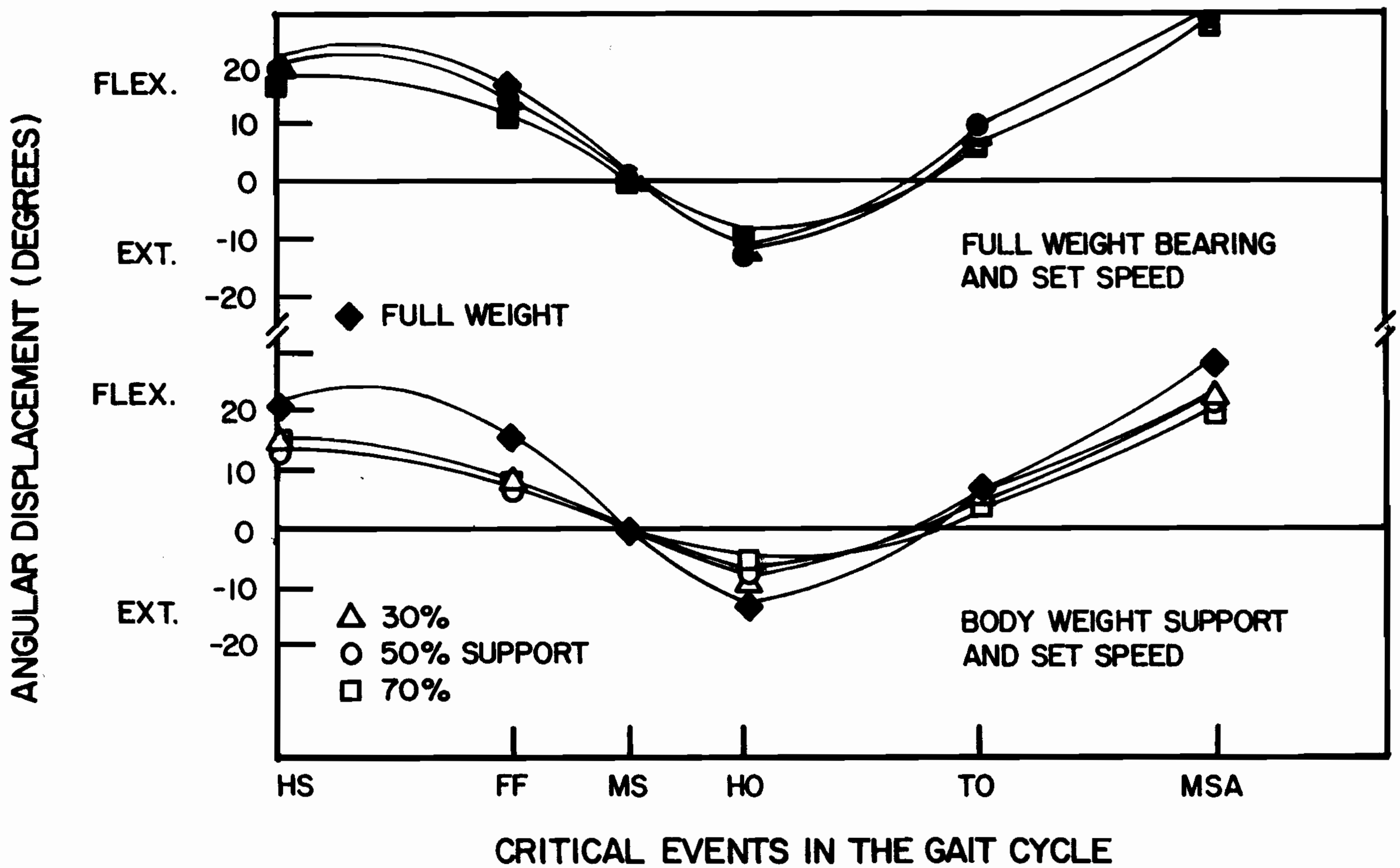
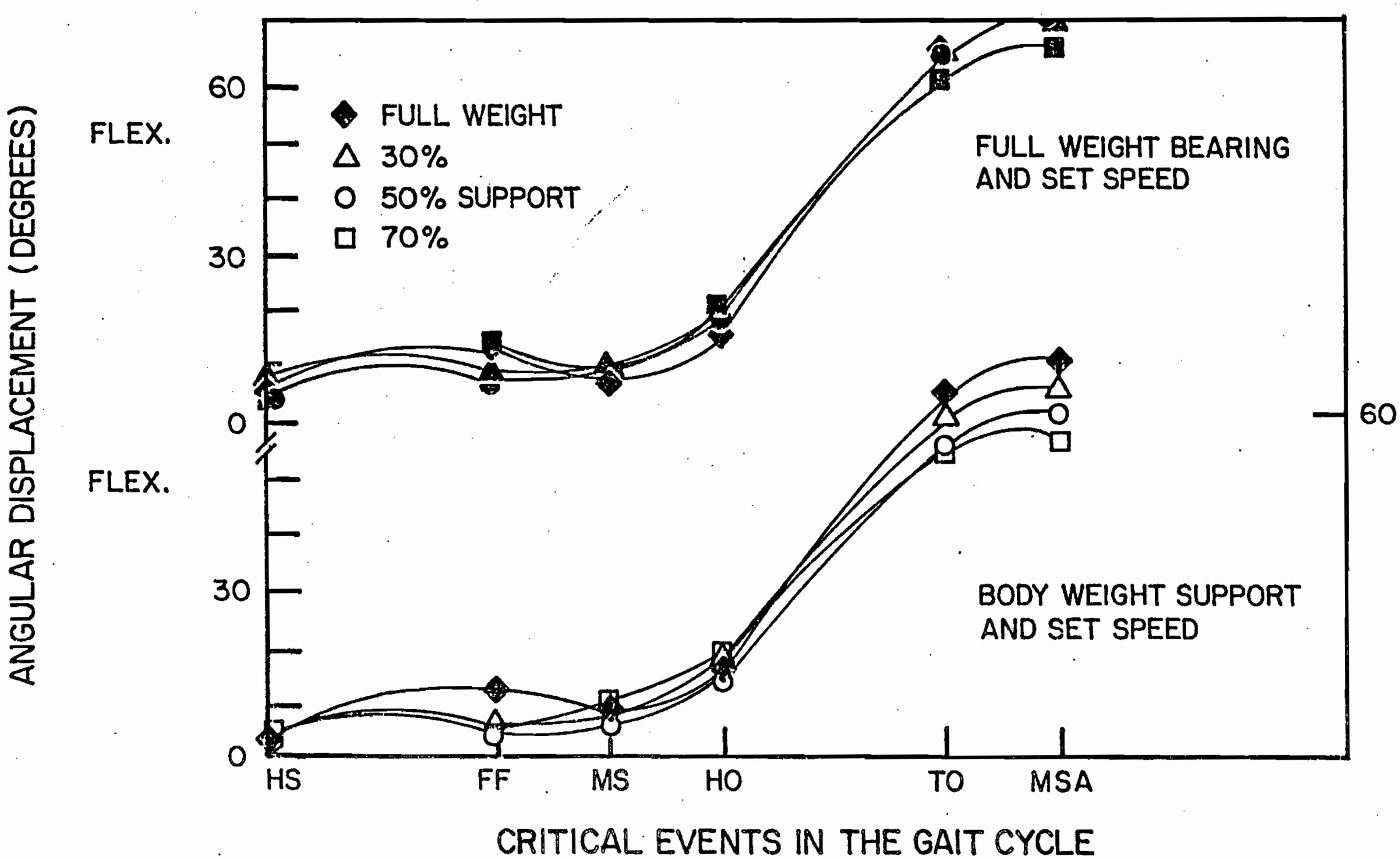


Figure 15 Mean angular knee displacement as a function of gait events. The data is plotted normalized to critical events of gait with the cycle normalized to 100%.

HS Heel Strike
FF Foot Flat
MS Mid Stance
TO Toe off
MSA Maximum Swing Angle



MEAN ANGULAR KNEE DISPLACEMENT AS A FUNCTION OF GAIT EVENTS angles of 10 subjects at each trial; for simplicity of illustration the 3-5 O standard deviation observed were not plotted. The 0% or full weight bearing condition at 1.36 m.s-1 speed, is included in both sets of curves as a reference line. All curves are plotted normalized to the critical gait events of the 0% BWS speed with the cycle normalized to 100%. Heel-strike (HS) occured at 0%, toe-off (TO) at 60%, maximum swing angle for the knee at 65% and for the hip at 85% of the gait cycle. The hip and knee joints demonstrate a similar pattern across all conditions with the exception of amplitude of movement. Statistical analysis was performed on the total mean hip and knee angular displacement and the maximum swing flexor angle of the hip and knee.

5.4 Total Mean Angular Displacement

The total mean amount of hip and knee angular displacement reported in table 4 and plotted in figures 16 and 17 decreased significantly across the 7 means tested ($F=28.54$, 14.58; $df=6,54$; $p<.01$) as determined by a repeated measures ANOVA.

5.4.1 Hip and Knee Speed Effects

Although with decreasing speed the total mean amount of angular displacement at the hip and knee decreased throughout the gait cycle, the amount of decrease was not significant across any of the 4 means tested via the Scheffe multiple comparison test. There was one exception, the mean total hip angular displacement at the slowest speed (.70 m.s-1) differed significantly from the 1.36 m.s-1 baseline.

5.4.2 BWS Hip Effects

Increasing BWS levels decreased the mean total angular displacement more at the hip and knee than that attributed to speed alone. The total mean angular displacement (table 4, and figure 16) did not differ with increasing BWS from 30 tO 70%. The Scheffe revealed

	BWS%	0	30	50	70	FWB	FWB	FWB
	Speed $m.s^{-1}$	1.36	.97	.85	.70	.97	.85	.70
<u>Hip</u>								
	Mean	46.20	32.70	31.60	27.60	40.20	42.00	38.70
	SD	3.30	7.70	4.50	4.90	4.30	3.60	4.30
	CV	.07	.23	.14	.17	.10	.90	.11
<u>Knee</u>								
	Mean	68.90	63.50	59.10	51.70	66.90	67.20	61.70
	SD	5.50	6.90	5.70	6.20	5.60	4.40	5.90
	CV	.08	.10	.09	.11	.08	.07	.10

TABLE 4: Hip and knee mean total angular displacement with standard deviations (SD) and coefficients of variation (CV) at each body weight support (BWS) and full weight bearing (FWB) speed. During each BWS session 0, 30, 50 or 70% of the subject's body weight was supported. During FWB sessions, the subject walked bearing full weight, but at the same speed as during BWS sessions.

Figure 16 Total mean range of hip angular displacement as a function of body weight support (BWS) and full weight bearing (FWB) treadmill speed.

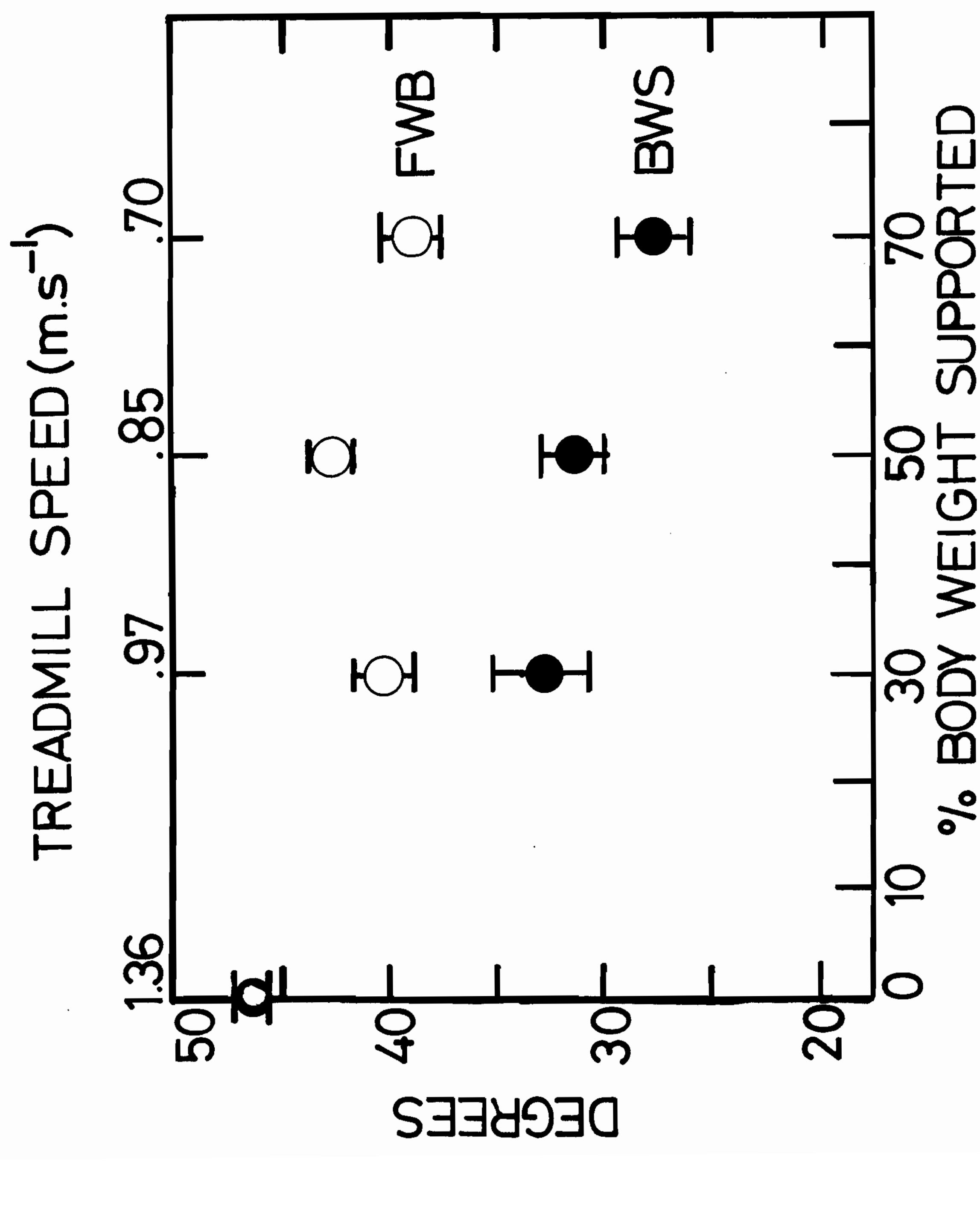
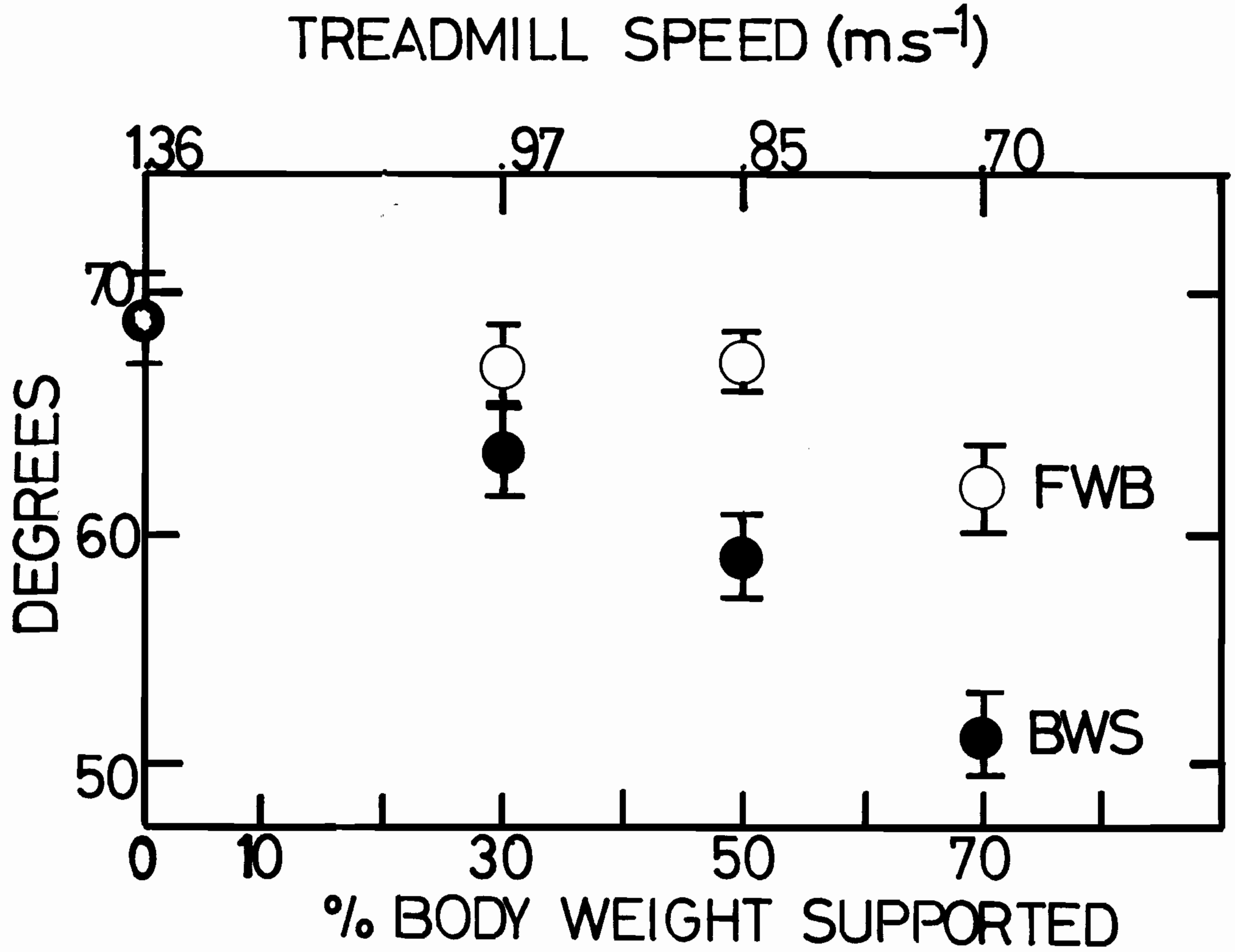

TREADMILL SPEED (m.s^{-1})
1.36
.97
.85
.70
DEGREES
50
40
30
20
0
10
30
50
70
FWB
BWS
% BODY WEIGHT SUPPORTED

Figure 17 Total mean range of knee angular displacement as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS).


TREADMILL SPEED (ms-1)
1.36
.97
.85
.70
DEGREES
70
60
50
FWB
BWS
0
10
30
50
70
% BODY WEIGHT SUPPORTED

significant decreases between BWS and FWB levels at equivalent speeds. The 30, 50, and 70% BWS total angular displacement being respectively 19, 24, and 28% less than the equivalent FWB speed angular displacement.

5.4.3 BWS Knee Effects

The total mean angular displacement (figure 17) differed across the BWS levels (n=4). The 0% BWS total angular displacement was significantly larger than the 50 and 70% BWS angular displacement. While the 70% BWS total angular displacement was less than at any FWB slow speed the total angular displacement did not differ (Scheffe) between the 30 and 50% BWS levels and their FWB slow speed equivalents. The total mean angular displacement for increasing BWS levels decreased by 5, 12, and 16% from their FWB slow speed equivalent.

5.5 Maximum Swing Angle

5.5.1 Hip and Knee Speed Effects

Subjectively, the angular displacement at each critical event for each decreasing speed appeared to differ little from the 0% BWS 1.36 m.s-1 baseline (figures 14 and 15). The Scheffe Comparison, however, revealed that the maximum swing angle (MSA) at the knee and hip were not affected significantly by decreasing speed (n=4, table 5).

5.5.2 BWS Hip Effects

The angular displacement for each BWS, at each critical event, appeared to differ from the 0% BWS baseline as seen in figure 14. The largest decreases being at HS, FF and the MSA. The 0% BWS, MSA was significantly larger than any BWS level MSA as seen in table 5. While the 50% BWS, MSA differed only from its FWB speed equivalent, the 70% BWS, MSA was significantly less than any FWS speed MSA. No other MSA differences were noted with the Scheffe multiple comparison test.

	BWS%	0	30	50	70	FWB	FWB	FWB
	Speed $m.s^{-1}$	1.36	.97	.85	.70	.97	.85	.70
<u>Hip</u>								
	Mean	29.90	23.80	22.50	20.20	27.90	29.30	27.20
	SD	3.00	7.50	6.50	5.10	3.30	3.40	3.40
	CV	.10	.32	.29	.25	.12	.12	.13
<u>Knee</u>								
	Mean	72.40	65.00	61.20	55.90	70.90	69.90	66.20
	SD	3.90	7.30	5.70	7.50	5.00	4.50	5.70
	CV	.05	.11	.09	.13	.07	.06	.09

TABLE 5: Hip and knee mean maximum flexor swing angles with standard deviation (SD) and coefficients of variation (CV) at each body weight support (BWS) and full weight bearing (FWB) speed. During each BWS session 0, 30, 50 or 70% of the subject's body weight was supported. During FWB sessions the subjects walked bearing their full weight, but at the same speed as during BWS session.

5.5.3 BWS Knee Effects

The knee angular displacement for each BWS level at each critical event appeared to differ little from the baseline 0% BWS except at FF, TO, and MSA (figure 15). Again similar to the hip, the 0% BWS, MSA was larger than any BWS, MSA. While the 50% BWS, MSA was significantly less than the .97 and .86 m.s-1 FWB speed MSAs, the 70% BWS, MSA was less than the 30% BWS and all FWB slow speed MSAs.

5.6 EMG ACTIVITY

The normalized on and off timing, and the normalized mean burst amplitude, of 6 right leg muscles (ES, GM, VL, MH, TA, GA) were examined (figure 18 and table 6a and 6b). The decreasing speed and increasing BWS levels affected the number of bursts per muscle and the number of subjects using each muscle. These facts made a statistical analysis of timing difficult, therefore the utilization and the timing of the muscles will be presented descriptively.

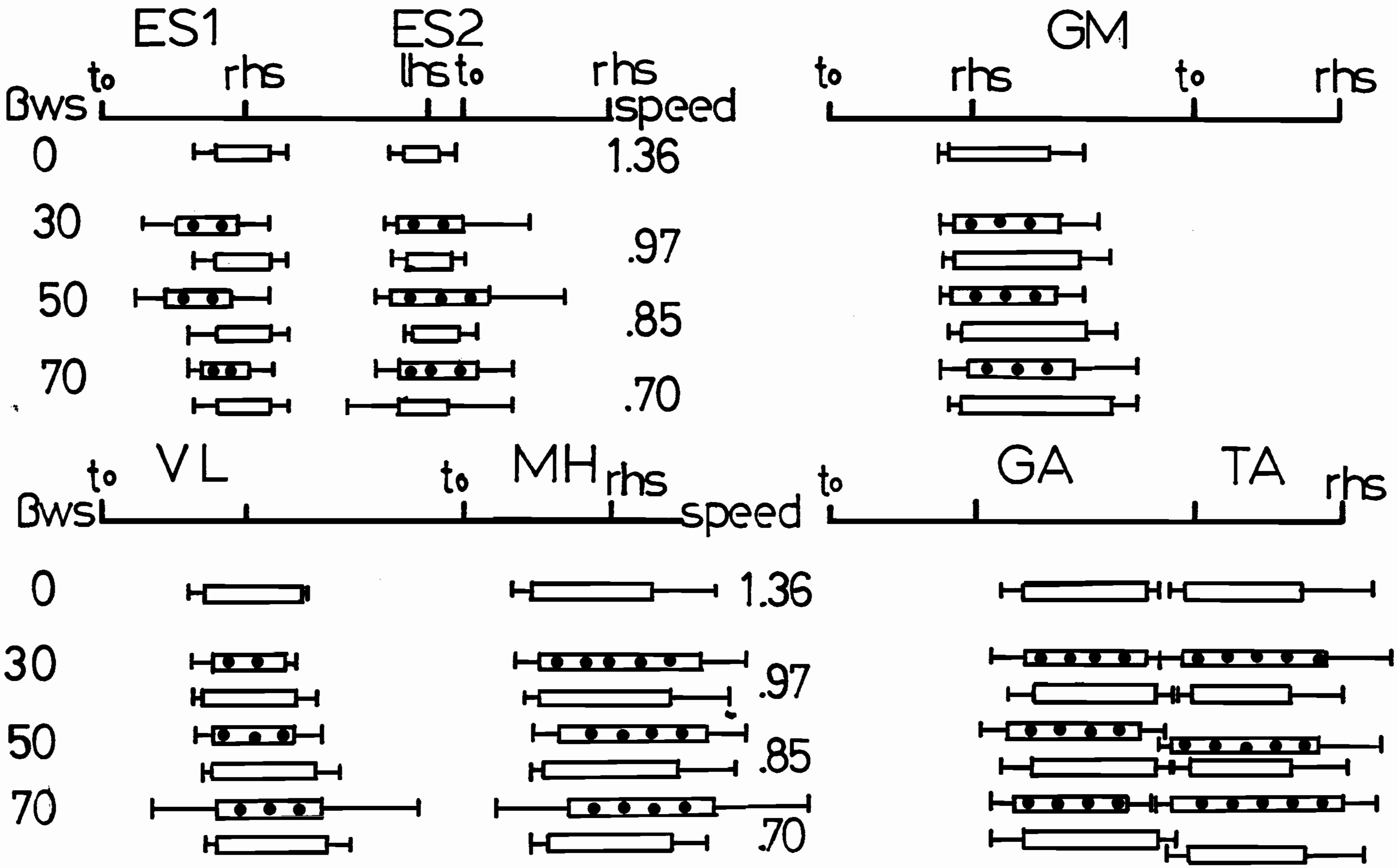
All ten subjects consistently used MH, GA, and TA at every BWS and FWB speed. Activity in ES, GM, and VL, however, varied according to speed or BWS level.

ES use did not vary with speed. With increasing BWS, however, the number of subjects with ES1 decreased from 7 at 30% BWS to 4 at 70% BWS. Two subjects did not use ES1 or ES2 at 70% BWS, while one other who did not use ES1 at 30 or 50% BWS developed a burst at 70% BWS. The decreased use of ES with BWS is in itself significant.

Only one person did not use GM either at 70% BWS or at .70 m.s-1 FWB speed.

The use of VL varied greatly, one subject never used this muscle, while the remaining 9 varied their use depending on the speed or BWS level. Only 8 subjects used VL at 1.36 m.s-1 and two additional subjects did not use VL at speeds less than 1.36 m.s-1. As with speed effects only 6 subjects (the same ones as above) used VL at 30

Figure 18 The normalized on and off timing of erector spinae (ES1 and ES2), gluteus medius (GM), vastus lateralis (VL), medial hamstrings (MH), gastrocnemius (GA), and tibialis anterior (TA). The timing was normalized to the gait cycle (to representing toe-off, rhs right heel strike, and lhs left heel strike). The full weight bearing treadmill speed (in $m.s.^{-1}$) conditions are represented by the open rectangles and body weight support (in percent) conditions the dotted rectangles. The 0% body weight support condition is represented by open rectangles. One standard deviation for each on and off timing is also included.


ES1
ES2
GM
to
rhs
lhs
speed
Bws
0
30
50
70
1.36
.97
.85
.70
VL
MH
GA
TA

and 50% BWS, while all 9 subjects used VL at 70% BWS.

The two burst muscles ES, MH and TA were affected differently by speed and BWS.

A second MH burst appeared at 0% BWS which inconsistently appeared at slower FWB speeds and with BWS levels, in two of the 10 subjects studied.

Two distinct TA bursts, separated by a 10 ms silent period, were present in 5 subjects at 0% BWS, in 3 subjects at 30%, in 2 at 50% and in 3 subjects at 70% BWS. Decreasing speed increased the number of subjects with two bursts. Two TA bursts were present in 8 subjects at .97m.s-1, in 6 at .85m.s-1, and in 4 subjects at .70 m.s-1. A single burst was present at all other speeds and BWS levels in all other subjects.

An additional third burst, between the first and second ES bursts appeared in most subjects with slower FWB speeds, but not with increased BWS. Only the first and last ES bursts, first MH and TA bursts are considered in this thesis.

5.7 EMG Timing

A great deal of inter-individual variability existed in on/off times, but the averages across the 10 subjects revealed the same trend. The on and off timing showed a tight link relative to the events in the gait cycle. Table 6a and 6b report the means and standard deviations (SD) of the mean on/off timing across the 7 sessions for 10 subjects, while figure 18 graphically represents the same data.

5.7.1 Erector spinae

There were 2 bursts of activity in this muscle during each cycle at 0% BWS. The first burst (ES1) occured before HS (-9% of the gait cycle) and finished just after HS at 7%. The second (ES2) burst occured near mid-stance, at 43% of the cycle, close to HS of the

MUSCLE	BWS%	0	30	50	70	FWB	FWB	FWB
	Speed $m.s^{-1}$	1.36	.97	.85	.70	.97	.85	.70
ES1	Mean	-9.4	-19	-22.8	-12	-8	-7.5	-7
	SD	6	9	8	4	6	8	7
	N	10	7	5	4	10	10	10
ES2	Mean	42.8	41	41	43	45.4	45.9	43.2
	SD	3	4	4	7	4	2	14
	N	10	10	10	7	10	10	10
GM	Mean	-7	-5.7	-6.7	-2	-4.6	-4	-4
	SD	3	5	3	8	4	3	4
	N	10	10	10	9	10	10	9
VL	Mean	-11.8	-9.8	-8	-8	-11.5	-9.8	-8.5
	SD	4	6	7	18	3	3	4
	N	8	6	6	9	6	6	6
MH	Mean	78.6	81.3	86.3	89.2	81.4	82	83.8
	SD	5	7	7	20	5	5	6
	N	10	10	10	10	10	10	10
TA	Mean	57	56	53.9	53.5	59	58.8	58.3
	SD	4	5	4	5	6	4	6
	N	10	10	10	10	10	10	10
GA	Mean	13.7	13.7	8.4	11	16.7	15.4	13.4
	SD	6	10	7	7	8	8	8
	N	10	10	10	10	10	10	10

TABLE 6A: The means and standard deviations (SD) of EMG on timing of 10 subjects averaged over 10 gait cycles. The EMGs are normalized to % cycle and include: erector spinae (ES1 and ES2), gluteus medius (GM), vastus lateralis (VL), medial hamstring (MH), tibialis anterior (TA), and gastrocnemius (GA). The number of subjects using each muscle at each body weight support (BWS) and full weight bearing (FWB) speed is listed below the time for each muscle.

MUSCLE	BWS%	0	30	50	70	FWB	FWB	FWB
	Speed $m.s^{-1}$	1.36	.97	.85	.70	.97	.85	.70
ES1	Mean	6.9	-2	-4.4	.5	6.5	6.8	6.8
	SD	6	9	11	6	5	6	5
	N	10	7	5	4	10	10	10
ES2	Mean	55.6	60	67.5	63.7	57.6	58.7	55.5
	SD	5	20	21	10	4	5	18
	N	10	10	10	7	10	10	10
GM	Mean	20.5	23.8	22	27	28	30.6	37
	SD	10	11	10	16	9	10	6
	N	10	10	10	9	10	10	9
VL	Mean	13.4	11.2	13	21.8	14.5	19	22
	SD	2	4	8	26	6	6	7
	N	8	6	6	9	6	6	6
MH	Mean	111	124.4	126	128.4	116	118.7	116.5
	SD	16	13	12	26	16	16	10
	N	10	10	10	10	10	10	10
TA	Mean	89.9	95.6	93.3	99.4	85	85.4	89.8
	SD	19	17	16	10	15	16	17
	N	10	10	10	10	10	10	10
GA	Mean	46.9	46.9	45	42	49	49	49.7
	SD	4	4	6	6	5	4	6
	N	10	10	10	10	10	10	10

TABLE 6B: The means and standard deviations (SD) of the off timing of 10 subjects averaged over 10 gait cycles. The EMGs are normalized to % cycle and include erector spinae (ES1 and ES2), gluteus medius (GM), vastus lateralis (VL), medial hamstring (MH), tibialis anterior (TA), and gastrocnemius (GA). The number of subjects using each muscle at each body weight support (BWS) and full weight bearing (FWB) speed is listed below the time for each muscle.

heterolateral leg and terminated just before TO of the homolateral leg.

Decreasing speed, from 1.36 to .70m.s-1 (n=4), did not affect the on or the off timing of ES1 or ES2. Inter-subject variability was constant for the ES1 on/off timing, but increased with decreasing speed for the ES2 burst.

Increasing BWS affected the ES1 more than the ES2 burst. The first burst of ES at any BWS started and finished earlier than at any FWB speed. While the ES2 on/off timing did not differ with FWB slow speed or weight changes. There was, however, a slightly later ES2 off timing at BWS levels above 0% compared to those at FWB slow speeds.

5.7.2 Gluteus Medius

A single burst of activity in gluteus medius at 0% BWS started just before HS at -7% of the cycle, and ceased at 20% near mid-stance, during single limb support.

Generally, the on timimg did not differ with decreasing speed, but started slightly later than the 0% BWS or natural FWB speed time. The off timing was similar across all speeds (n=3), but occured later than the 0% BWS level. The variability of the off timing was greater than that of the on timing, but did not appear to increase with decreasing speed (figure 18).

Increasing BWS levels (n=4) did not affect the on timing or the off timing. The off timings were earlier than their FWB speed equivalents. The variability in timing across weight for GM was consistent.

5.7.3 Vastus Lateralis

Activity of this muscle at 0% BWS for 8 subjects commenced at 11% of the gait cycle before HS and ceased at 13% near FF when the knee was flexing. The results for VL vary greatly.

Beyond the actual number of subjects using VL, decreasing speed (n=4) did not affect the on times. The off times, however, progressively lengthened from 13% of the gait cycle at 1.36 m.s-1 to 22% at.70 m.s-1. Variability of timing across speed was greater for off timing than on timing.

The increased BWS levels did not affect on/off timing. The 70% BWS off times, however, were prolonged to 22% similar to the timing of its FWB speed equivalent. The variability of VL timing increased with increasing BWS, especially at the 70% BWS level.

5.7.4 Medial Hamstrings

This biarticular muscle functions both as a hip extensor and a knee flexor. The activity in medial hamstrings at 0% BWS or natural FWB speed, consisted of a single burst (except as noted in methods), starting at 79% of the gait cycle, continuing through to the start of the next HS and ceasing at 11% of the next gait cycle.

Decreasing speed (n=4) had no effect on the on/off timing of MH. The variability of the times were consistent across decreasing speed, but were much larger for the off than the on timing.

Increasing BWS had a marginal effect on MH timing. The MH on timing did not differ across BWS levels. Although the start of MH was progressively delayed from 78.6% at 0% BWS to 89.2% at 70% BWS, a difference either between BWS levels or between BWS and FWB speed equivalents was not seen for off timing. The variability of both on/off timing increased greatly at 70% BWS.

5.7.5 Gastrocnemius

Activity at 0% BWS, natural FWB speed, started at 14% near FF and ended at 47% near HO.

Despite the slight delay in on timing of GA with slower speed as seen in figure 18 decreasing speed had no effect on the on or off timing. The variability increased slighty with decreasing speed and was greater for the on than the off timing.

Despite the earlier on timing of GA with 50 and 70% BWS, increasing

BWS did not effect GA timing. The 70% BWS off timing, however, differed from its FWB speed equivalent, but was similar to the faster .85 m.s-1 off timing. Variability was similar to that seen with decreasing speed.

5.7.6 Tibialis Anterior

The TA activity at 0% BWS, FWB natural speed, started in late swing at 57% of the gait cycle and ended at 90% just before the next HS.

Decreasing speed did not affect the timing of the first TA burst. The variability, however, increased slightly with decreasing speed with the off timing varing more than the on timing.

Although the on timing of the first burst was slightly earlier and off timing later with increasing BWS levels compared to speed levels (figure 18), no differences appear between BWS levels or between BWS and FWB speed equivalents. Variability was similar to that of slower speed results.

5.8 NORMALIZED EMG BURST AMPLITUDES

The means and SD for the normilized mean burst amplitudes (n= 10 subjects) for each muscle burst are in table 7.

5.8.1 Erector Spinae

A Freedmans ANOVA by ranks demonstrated a significant difference across BWS and FWB speed levels (n=6) for both ES1 and ES2 ($X2=21.89, 23.37$; $df=6$; $p<.01$ respectively).

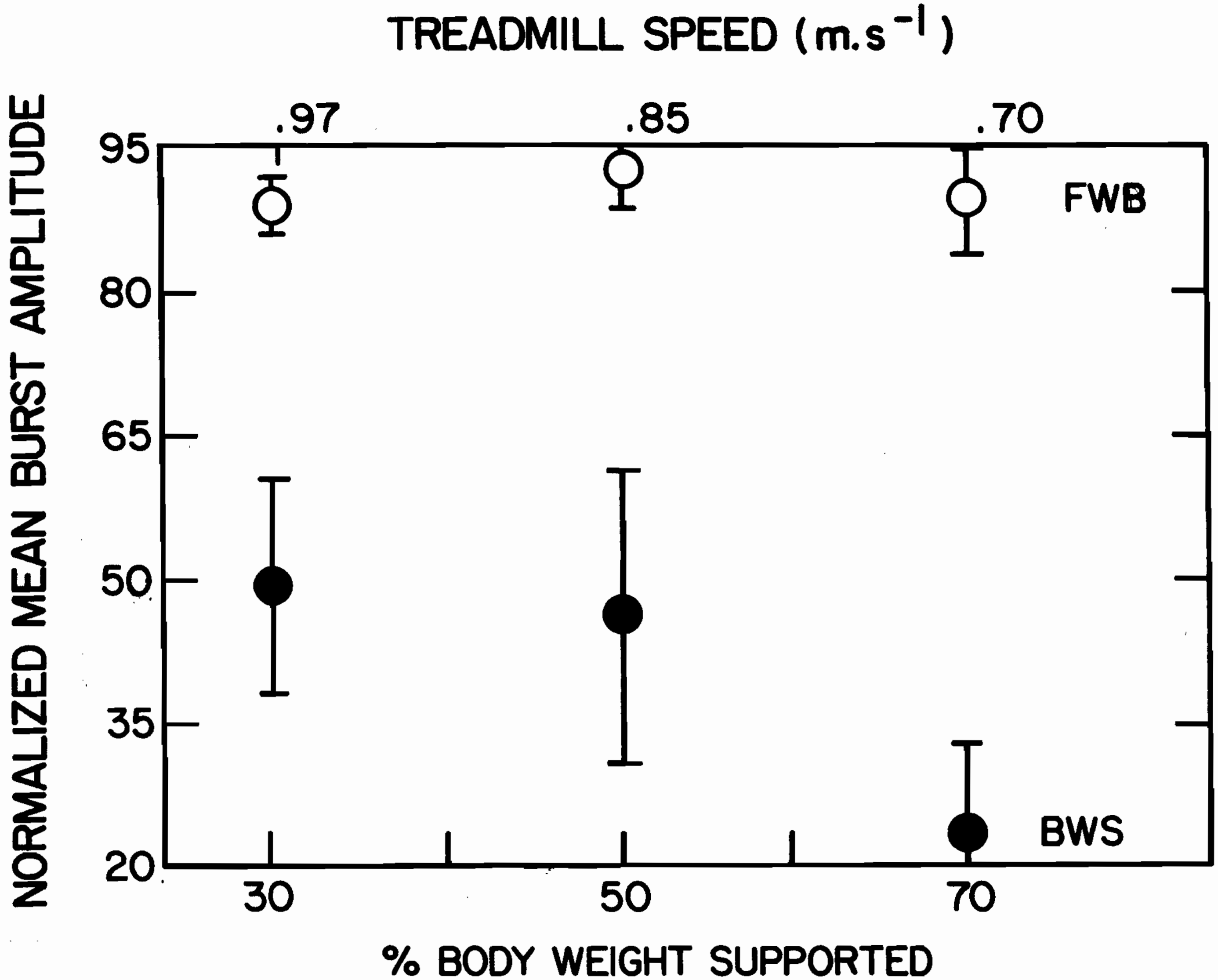
The Wilcoxon sign test revealed that decreasing speed did not affect the ES1 or ES2 burst amplitudes. One exception did exist, the ES2 .97 m.s-1 burst amplitude was greater than that of the ES2 .70 m.s-1. The mean amplitude of ES1 and ES2 appeared similar across decreasing speeds. The amplitude variability, for both bursts, increased with decreasing speed.

The amplitude of both ES1 and ES2 decreased compared to the FWB slower speed amplitudes as seen in figures 19 and 20. The percentage

MUSCLE	BWS% Speed $m.s^{-1}$	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
ES1	Mean	48.90	45.90	23.20	88.60	92.30	89.20
	SD	38.60	51.10	32.70	9.10	15.00	19.0
	N	.79	1.11	1.41	.10	.16	.21
ES2	Mean	69.50	63.20	32.30	84.70	80.60	70.60
	SD	23.60	35.50	26.30	12.10	16.80	13.20
	N	.34	.56	.81	.14	.21	.19
GM	Mean	65.00	57.70	31.80	89.60	94.80	91.50
	SD	17.60	20.00	20.00	19.30	18.00	26.30
	N	.27	.35	.65	.22	.19	.29
VL	Mean	35.50	28.00	33.60	53.50	49.20	39.40
	SD	35.50	27.60	33.10	34.70	31.50	26.10
	N	1.00	.99	.99	.65	.64	.66
MH	Mean	84.50	84.40	56.10	85.40	75.60	63.30
	SD	25.30	27.00	25.40	16.40	15.00	15.20
	N	.30	.32	.45	.19	.20	.24
TA	Mean	97.00	104.00	112.20	74.30	74.40	78.10
	SD	19.20	35.00	45.40	12.40	14.30	21.90
	N	.20	.34	.40	.17	.19	.28
GA	Mean	76.20	67.00	40.50	97.10	93.80	92.00
	SD	16.60	16.90	14.80	4.70	8.80	8.50
	N	.22	.25	.37	.50	.90	.90

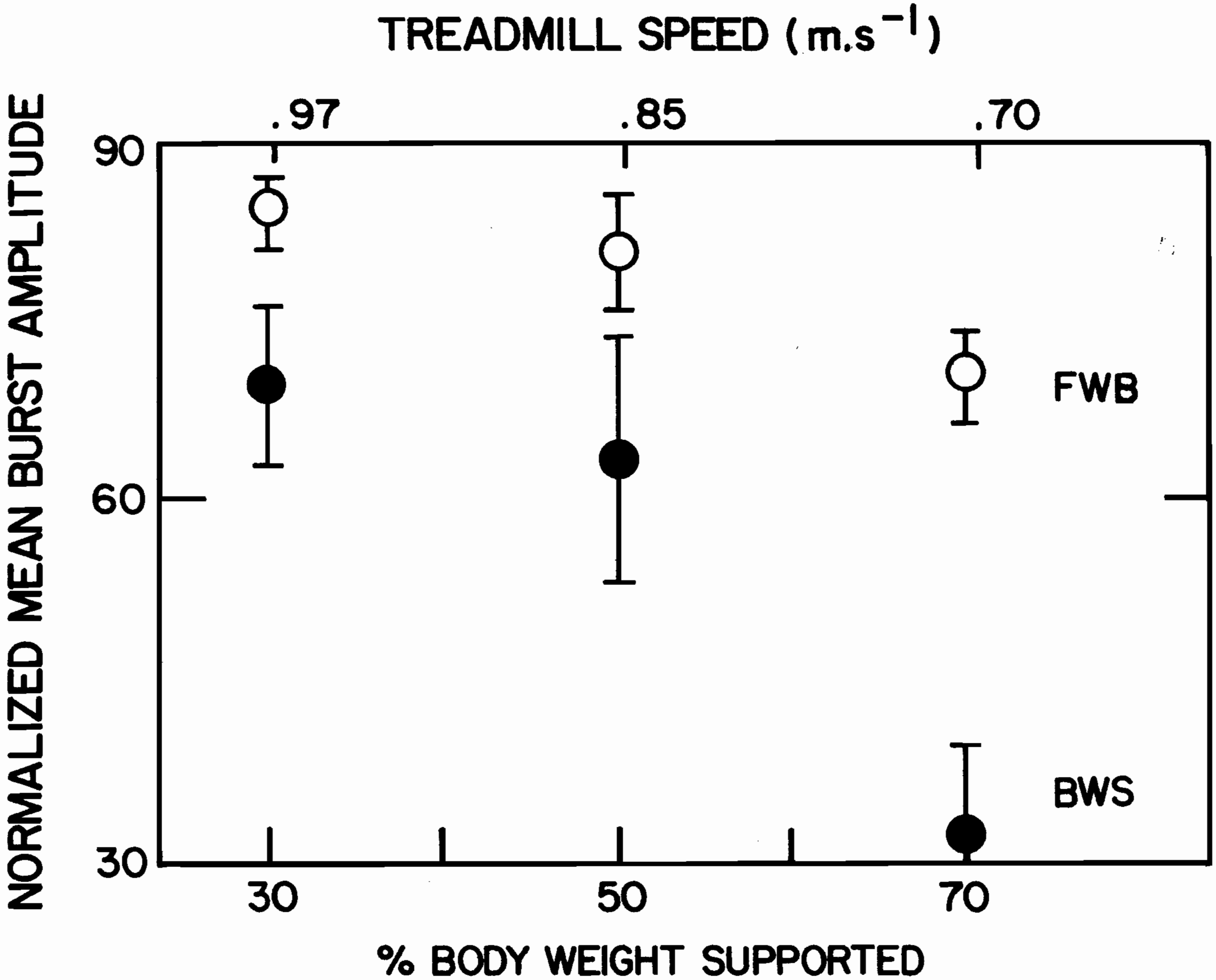
TABLE 7: The normalized mean burst amplitude with standard deviations (SD) and coefficients of variation (CV) of erector spinae (ES1, ES2), gluteus medius (GM), vastus lateralis (VL), medial hamstrings (MH), tibialis anterior (TA), and gastrocnemius (GA). The mean burst amplitude at each BWS and FWB speed were normalized to the 0% BWS 1.36 $m.s^{-1}$ speed mean burst amplitude before the data was pooled across the subjects.

Figure 19 First burst of Erector Spinae mean burst amplitude as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS)



ERECTOR SPINAE (I) MEAN BURST AMPLITUDE
TO TREADMILL SPEED AND WEIGHT SUPPORT (± SEM)

Figure 20 Second burst of Erector Spinae mean burst amplitude as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS)



ERECTOR SPINAE (2) MEAN BURST AMPLITUDE TO TREADMILL SPEED AND WEIGHT SUPPORT (± SEM) decrease at 30, 50 and 70% BWS compared to their FWB speed equivalents for ES1 was 44, 50 and 74% and for ES2 was 18, 21.5 and 54% respectively. The amplitude at 70% BWS for both bursts dropped sharply.

BWS levels affected ES1 and ES2 differently. Although the ES1 burst amplitudes did not differ with increasing BWS levels (n=3), both the 30 and 70%, but not the 50% BWS burst amplitudes were significantly less than any FWB slow speed burst amplitude (n=3; table 7 and figures 19 and 20).

While the ES2 burst at 70% BWS was significantly smaller than any other burst amplitude, except that at 50% BWS, the 30 and 50% BWS amplitudes were not (figure 20). Both the ES1 and the ES2 mean burst amplitudes varied greatly. The variability increased with increasing BWS such that the SD were often larger than the means themselves. The ES2 burst appeared greater than that of the ES1 for all BWS sessions.

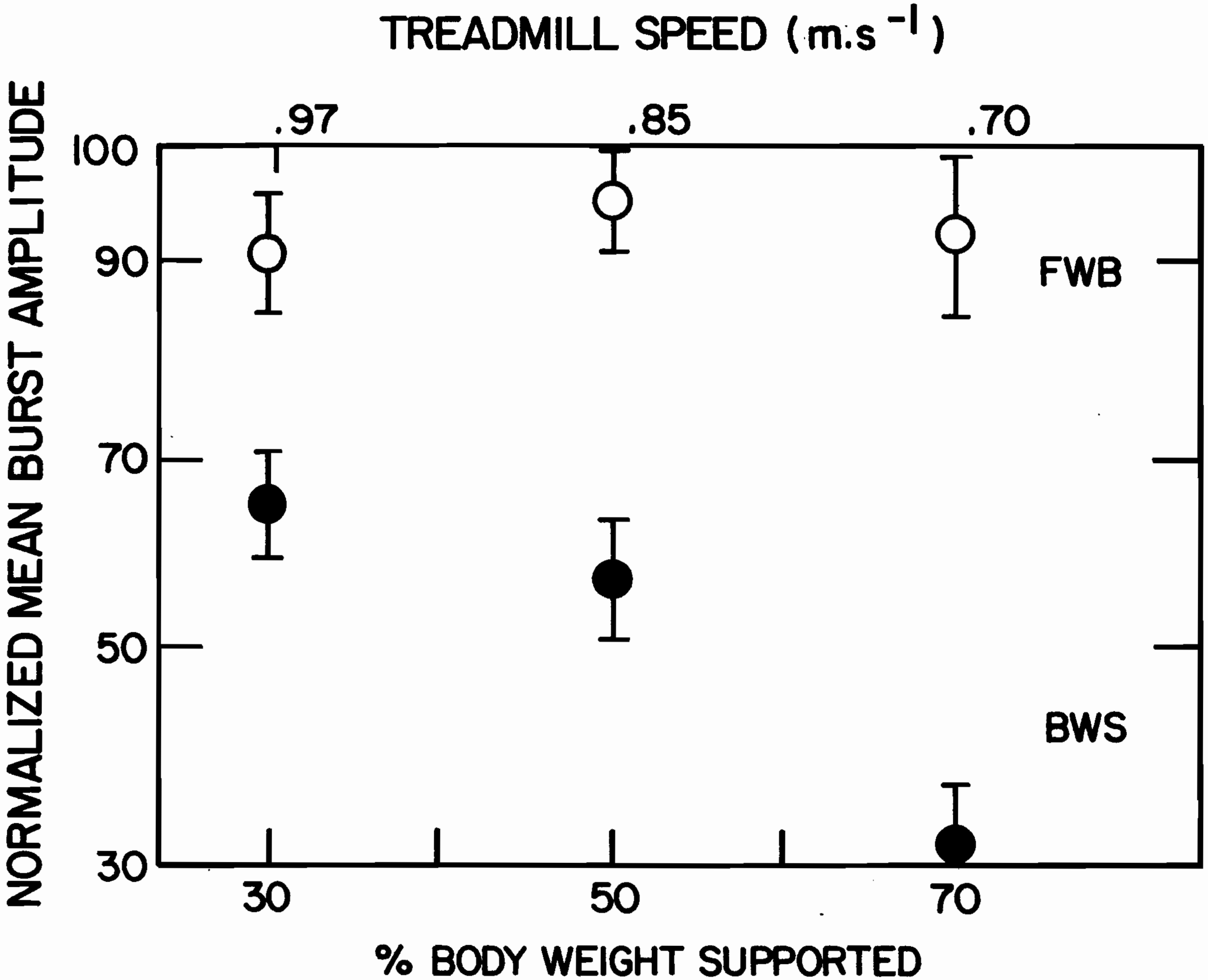
5.8.2 Gluteus Medius

The mean GM burst amplitude changes of figure 21 appear similar to those of the ES1 in figure 19. A repeated measures ANOVA (n=6) demonstrated a significant decrease in the mean burst amplitudes ($F=17.68$; $df=5,45$).

Decreasing speed had no effect on GM burst amplitudes. The variability across speeds (n=3) was similar with an increase at the slowest speed.

Despite a decrease in amplitude with increasing BWS, a Scheffe comparison revealed a significant decrease only in the 70% BWS mean burst amplitude compared to any BWS level or FWB slow speed. The 50% BWS mean amplitude, however, differed from its equivalent FWB speed amplitude. The percent decrease in BWS mean burst amplitude was 27, 39 and 65% for the 30, 50 and 70% BWS respectively compared to their FWB speed equivalent.

Figure 21 Gluteus Medius mean burst amplitude as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS)



GLUTEUS MEDIUS MEAN BURST AMPLITUDE TO TREADMILL SPEED AND WEIGHT SUPPORT (± SEM)

5.8.3 Vastus Lateralis

The mean burst amplitude results for VL in table 7 and figure 22 are the average of 8 subjects, because two subjects did not use VL while walking at 0% BWS. Despite a steady decrease in amplitude no significant differences (n=6) were found (F=1.18;df=5,35;p=.33)

Speed did not affect the amplitude, but the variability in the results were extreme and consistent across decreasing speed (n=3).

The mean amplitude decreased more in the BWS than the FWB condition. The percent decrease across BWS sessions compared to FWB speed equivalents was 35% for 30 and 50% BWS and 15% for 70% BWS level. The variability in amplitude was again extreme and greater than that at FWB slow speeds.

5.8.4 Medial Hamstrings

The mean amplitude results are tabulated on table 7 and shown in figure 23. The ANOVA results yielded a significant decrease in amplitude across the 6 means tested (F=5.06; df=5,45)

Despite a steady decrease in mean burst amplitude, no significant diffferences were found via the Scheffe test. The intersubject variability of the amplitude was consistent across speed.

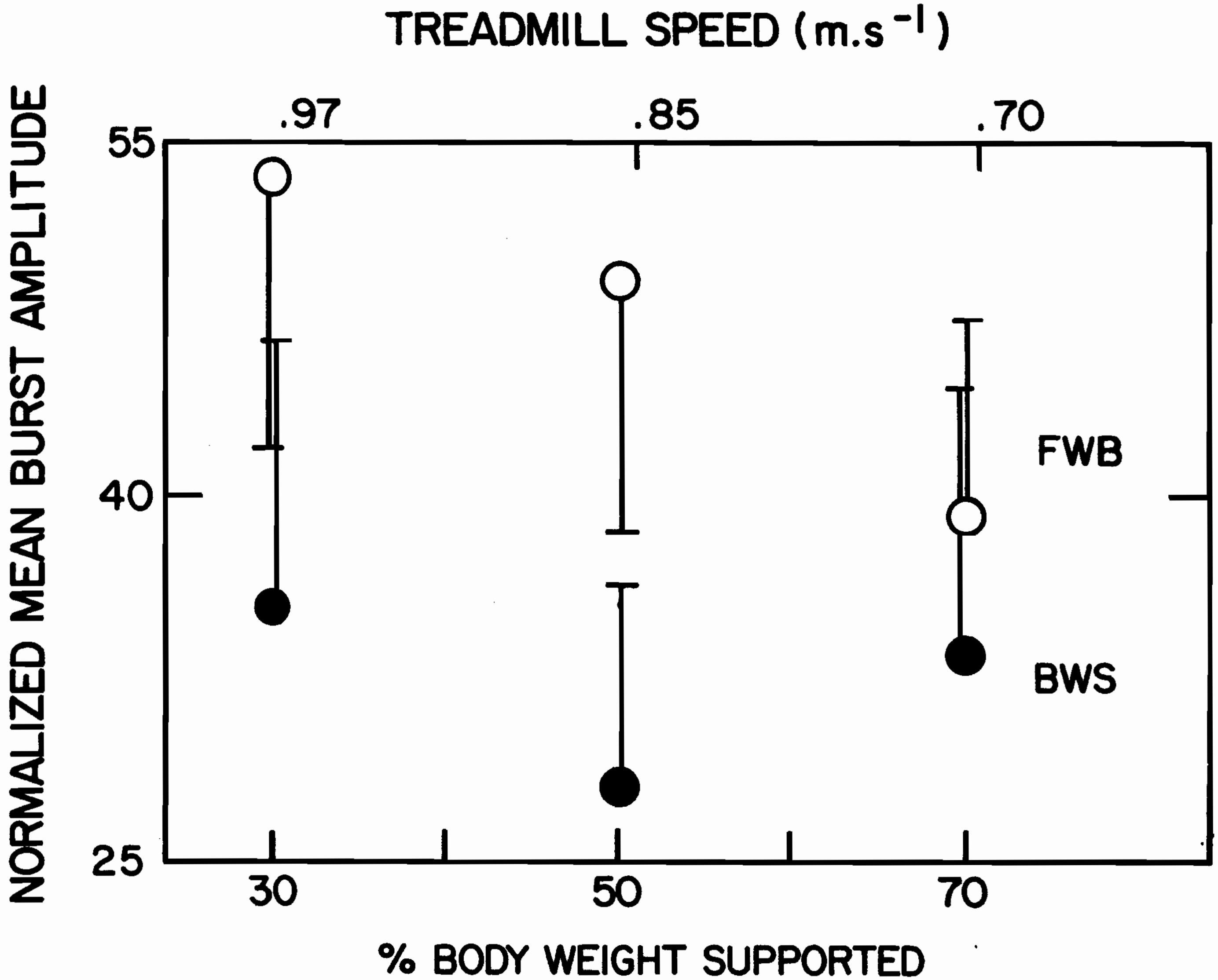
The mean burst amplitude did not decrease with increasing BWS, except at 70% BWS as seen in figure 23. The percent decrease at 70% BWS compared to its FWB speed equivalent was 13%, but a 33% decrease exised in the 70% compared to the 50% BWS level. The variability of the mean MH amplitudes were greater with increasing BWS than with decreasing speed.

5.8.5 Gastrocnemius

A Freedmans ANOVA uncovered a significant difference between the 6 means tested (X2=23.2;df=6)

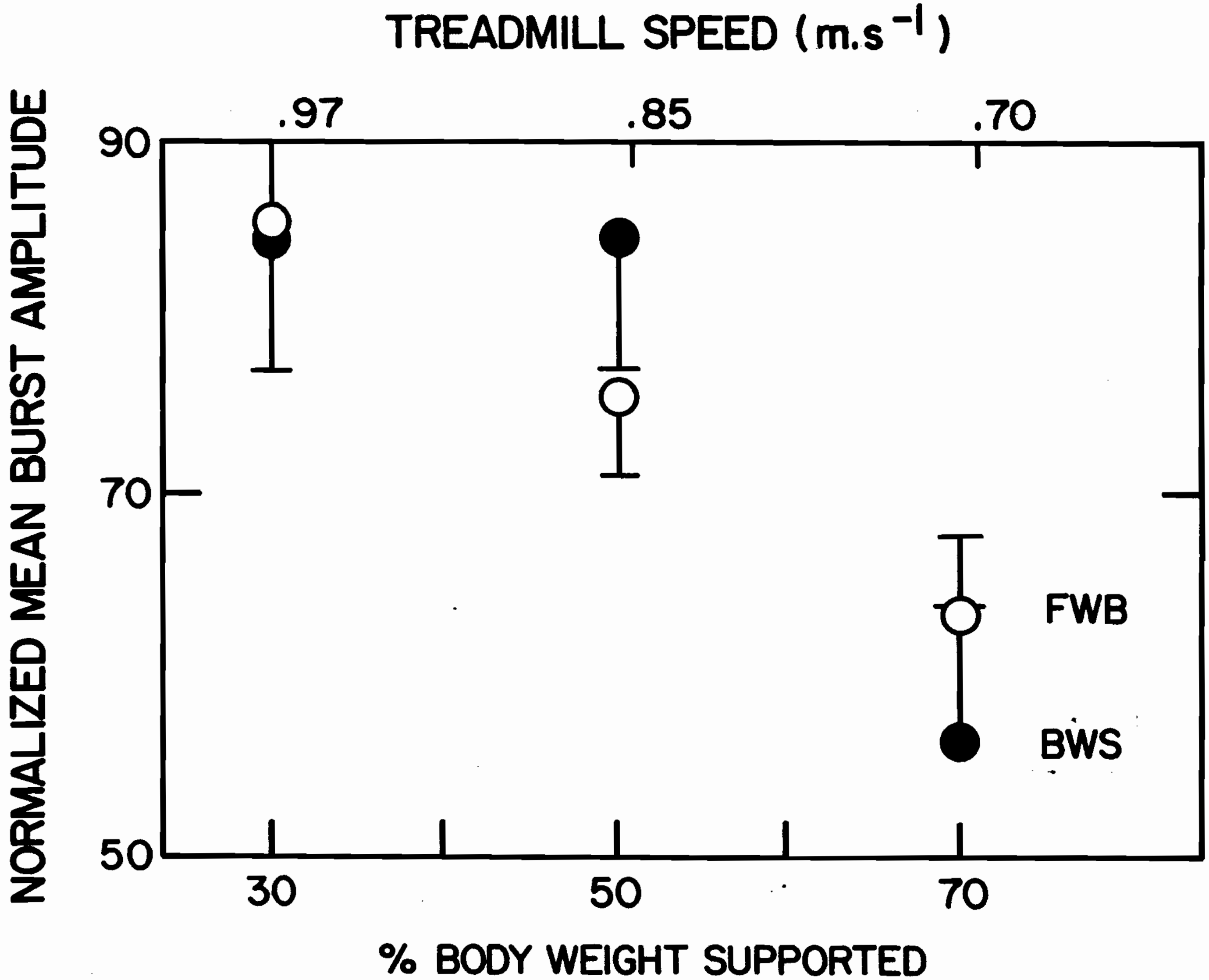
The mean GA amplitude remained stable across decreasing speed. The variance was low, but increased with decreasing speed.

Figure 22 Vastus lateralis mean burst amplitude as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS).



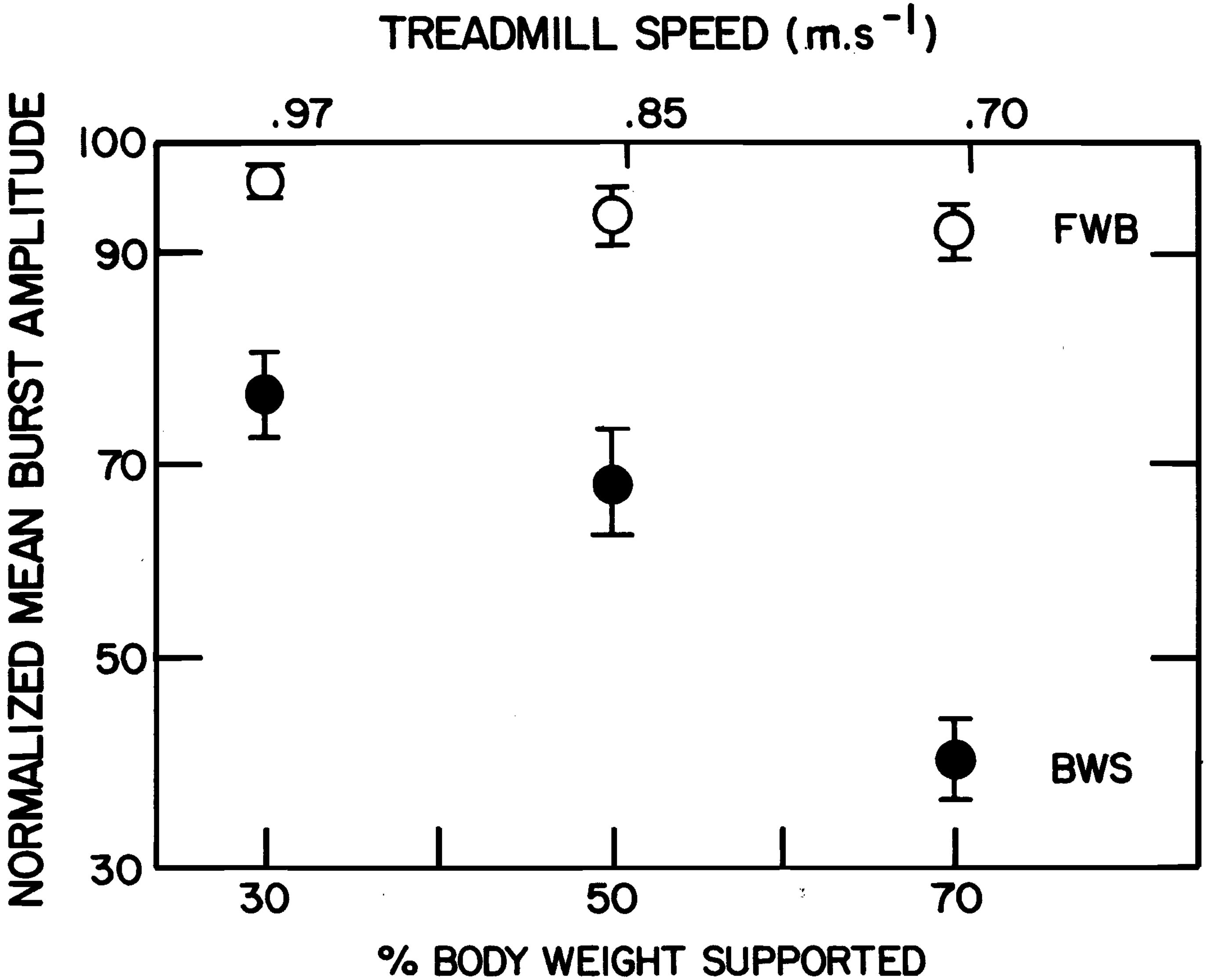
VASTUS LATERALIS MEAN BURST AMPLITUDE
TO TREADMILL SPEED AND WEIGHT SUPPORT (± SEM)

Figure 23 Medial hamstrings mean burst amplitude as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS).



MEDIAL HAMSTRINGS MEAN BURST AMPLITUDE
TO TREADMILL SPEED AND WEIGHT SUPPORT (± SEM)

Figure 24 Gastrocnemius mean burst amplitude as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS).



GASTROCNEMIUS MEAN BURST AMPLITUDE TO TREADMILL SPEED AND WEIGHT SUPPORT (± SEM)

Figure 24 depicts the mean GA amplitude decrease across BWS levels. The percent decrease in amplitude across BWS increases from 25% at 30 and 50% BWS to a 56% decrease at 70% BWS. The Wilcoxon test demonstrated that the 30 and 50% BWS mean amplitudes did not differ from each other, but both were significantly greater than the 70% BWS GA amplitude. In addition, the mean GA amplitudes at all BWS were significantly less than any FWB slow speed amplitude. The percent decrease in amplitude between BWS and FWB speed equivalents, from 30 to 70% BWS, was 21, 28, and 55% respectively. The amplitude variability increases with increasing BWS levels and was larger than FWB slow speed variability.

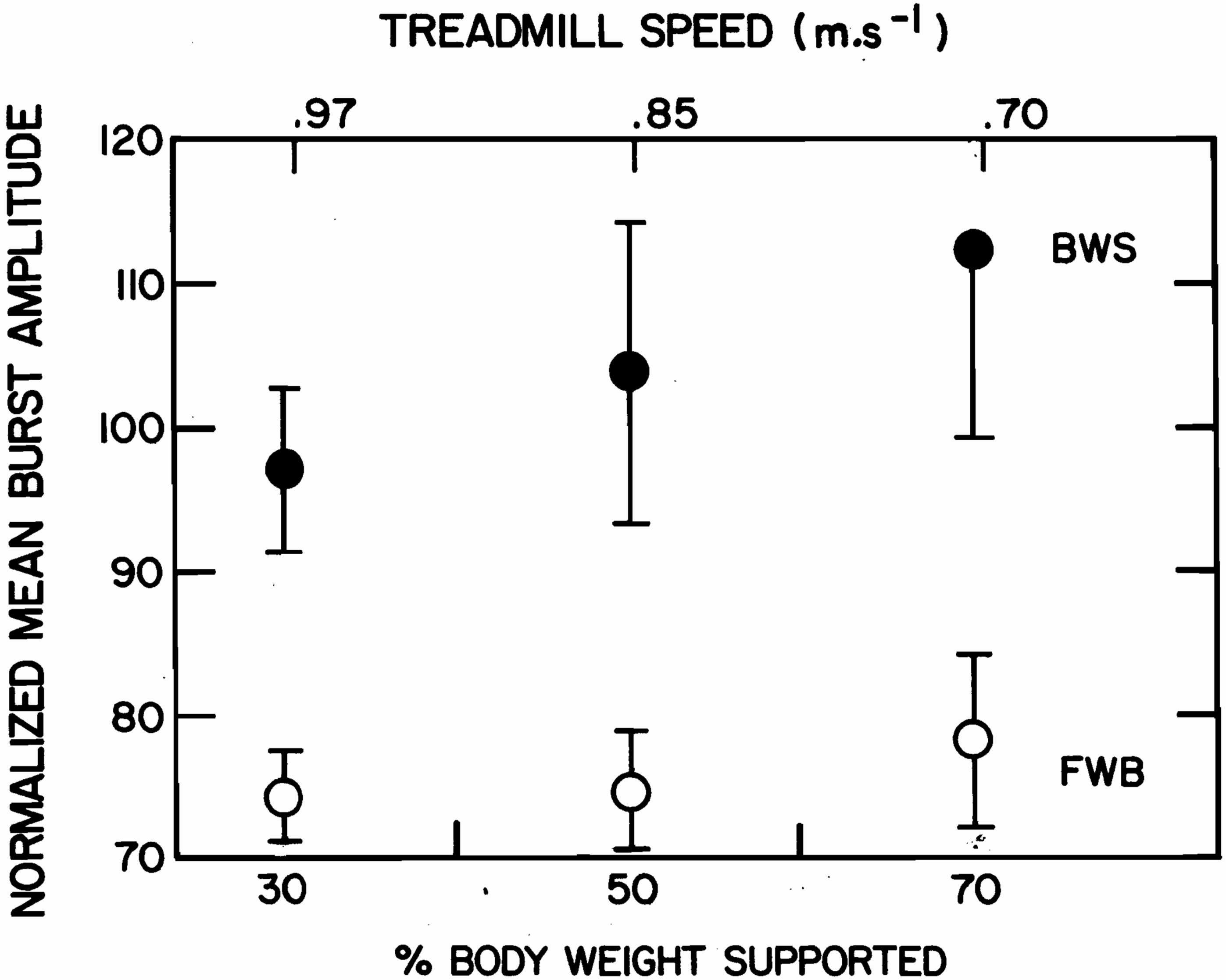
5.8.6 Tibialis Anterior

The mean TA amplitude changes in figure 25 appear to be the recripocal of those seen in figure 24 for GA. A Freedman ANOVA by ranks pointed out a significant difference scross the 6 means tested (X2=21.2; df=6).

While the variability of the mean amplitudes increased with decreasing speed, the amplitudes (n=3) themselves were not significantly different.

While the mean amplitudes increased with increasing BWS, the amplitudes were not significantly different. The percent increase in BWS amplitudes compared to their FWB slow speed equivalents was 30%. The 30% BWS mean amplitude did not differ from the slowest FWB speed (.70 m.s-1) amplitude. All BWS levels were significantly less than than those at the FWB slow speeds except for the one noted above. Variability increased with increasing BWS and was greater than the FWB speed variability.

Figure 25 Tibialis Anterior mean burst amplitude as a function of full weight bearing (FWB) treadmill speed and body weight support (BWS).



TIBIALIS ANTERIOR MEAN BURST AMPLITUDE TO TREADMILL SPEED AND WEIGHT SUPPORT (± SEM)

6 DISCUSSION

6.1 QUALITY OF GAIT

6.1.1 Speed Effects

Based on the subjective evaluation, it appears that speed has little effect on qualitative features of gait, except for arm swing. The decrease in arm swing supports previous work that the balancing effect of arm swing is required less with slower speeds. (Inman et al. 1981).

6.1.2 BWS Effects

Besides the diminishing need for arm swing increasing BWS affected other features of gait more than can be attributed to slow speed. BWS levels above 0% raised the height of the trochanter reaching a level 1.5 cm. above normal at 70% BWS. This decreased the contact distance. In addition, the raised trochanter and harness progressively restricted the natural downward displacement of the center of gravity (CG) with each step, especially at HS. This decreased range of vertical trunk movement in combination with a shortened contact distance may have repercussions on other gait parameters, for example: the decreased speed required with BWS; the decreased TDST; the pain felt at 70% BWS; and indirectly the decreased hip angular displacement.

Because the BWS system progressively increased the amount of weight supported, raising the height of the trochanter and shortening the contact distance, the natural stride length was effectively shortened. Therefore, to maintain the same speed at BWS and FWB the preliminary subjects either increased their cadence or could not follow the treadmill until the speed was reduced. Despite the dictated speed at each BWS level, 9 out of 10 experimental subjects increased their pelvic rotation to increase their stride length to walk at the FWB speed of .70 m.s-1 with 70% BWS, (the level with the shortest contact distance).

Although the amount of pelvic rotation could not be measured, it appeared normal at 30%, but increased progressivley at 50 and 70% BWS. Because of this fact, any conclusions related to rotation are speculative. The rotation may be related to speed - there may be an optimal speed per BWS level for each subject at which rotation would not increase - or rotation may be related to BWS.

The support system used by Hewes et al. (1967) did not raise their subjects. Nevertheless, the subjects, (freely choosing their speed), walked 60% slower than normally. The reason postulated for the slower speed (Margaria and Cavagna 1964) was a decrease in force at push off. The slower speeds at increasing BWS levels (especially at 70% BWS) may thus be the result of the decreased push-off force, as well as a shorter contact distance.

For each BWS condition there is a slow speed FWB control. The BWS results, therefore, include not only BWS effects, but also speed effects. This point will be discussed throughout the BWS sections.

6.2 Temporal Distance Factors

6.2.1 Speed Effects

The 10 subjects adapted to the decreasing speed by significantly increasing their cycle times, slightly, but not significantly increasing the % stance and TDST, and decreasing their stride length and cadence. Murray et al.(1966), and Larsson et al. (1980) reported similar adaptations to speed. Furthermore the values in table 3 closely resemble those of Yang and Winter (1985) and Murray et al.(1985) despite the difference in speed assignment and walking surface. The TDST, however, is slightly longer (27%) than normally (20%) reported for over ground walking, but is consistent with other treadmill data (Murray et al.1985). The longer TDST is thought to reflect a need for greater balance on a moving surface.

The increased variability of the data (table 3) increases with decreasing speed, the cycle time and TDST varying more than % stance and stride length as reported previously (Herman et al.1976).

6.2.2 BWS Effects

BWS had no effect on cycle time, cadence or stride length compared to equivalent FWB speeds (table 3).

Although the velocity of walking is determined by stride length and cadence a variety of cadences and stride lengths can produce the same walking speed (Grieve and Gear 1966). While the speed of walking was dictated during this investigation, no other constraints were imposed. The relationship between stride length, cadence and velocity, however, remained stable (figures 12 and 13). At 70%, however, subjects took fewer, longer steps similar to the subjects walking under lunar gravity. Nevertheless, even the individual parameters of cadence and stride length remained the same at FWB and BWS levels.

The decreased contact distance and raised trochanter made it difficult to maintain the same speed FWB and at 70% BWS without an increase in stride length. This increase in stride length might be brought about by the observed increasing pelvic rotation which could allow for a slight decrease in cadence. The increased pelvic rotation may not be required if the speed was reduced. The BWS effects on cadence and stride length are unknown as the speed effects confound the present BWS data. Further investigation into the relationship between the appropriate speed of progression for the amount of weight supported is necessary.

Although stride length appears unaffected by BWS levels, its method of determination may have led to misintrepretations. Stride length was inferred from time measures. Because of the probable progressive increase in pelvic rotation, the leg swung through an arc out of the plane of progression consequently swing time is increased. This increase in swing time leads to an apparent increase in stride length. An alternative method of measuring stride length - measuring the support length (sagittally measuring the horizontal amplitude of hip movement (Grillner et al. 1979)), however, would suffer from a similar problem. The pelvic rotation causes the hip to move out of the sagittal plane and a true measure of horizontal length can not be made. The values of the stride length in table 3 are not the true stride length, but the values (true and measured) are probably correct relative to each other.

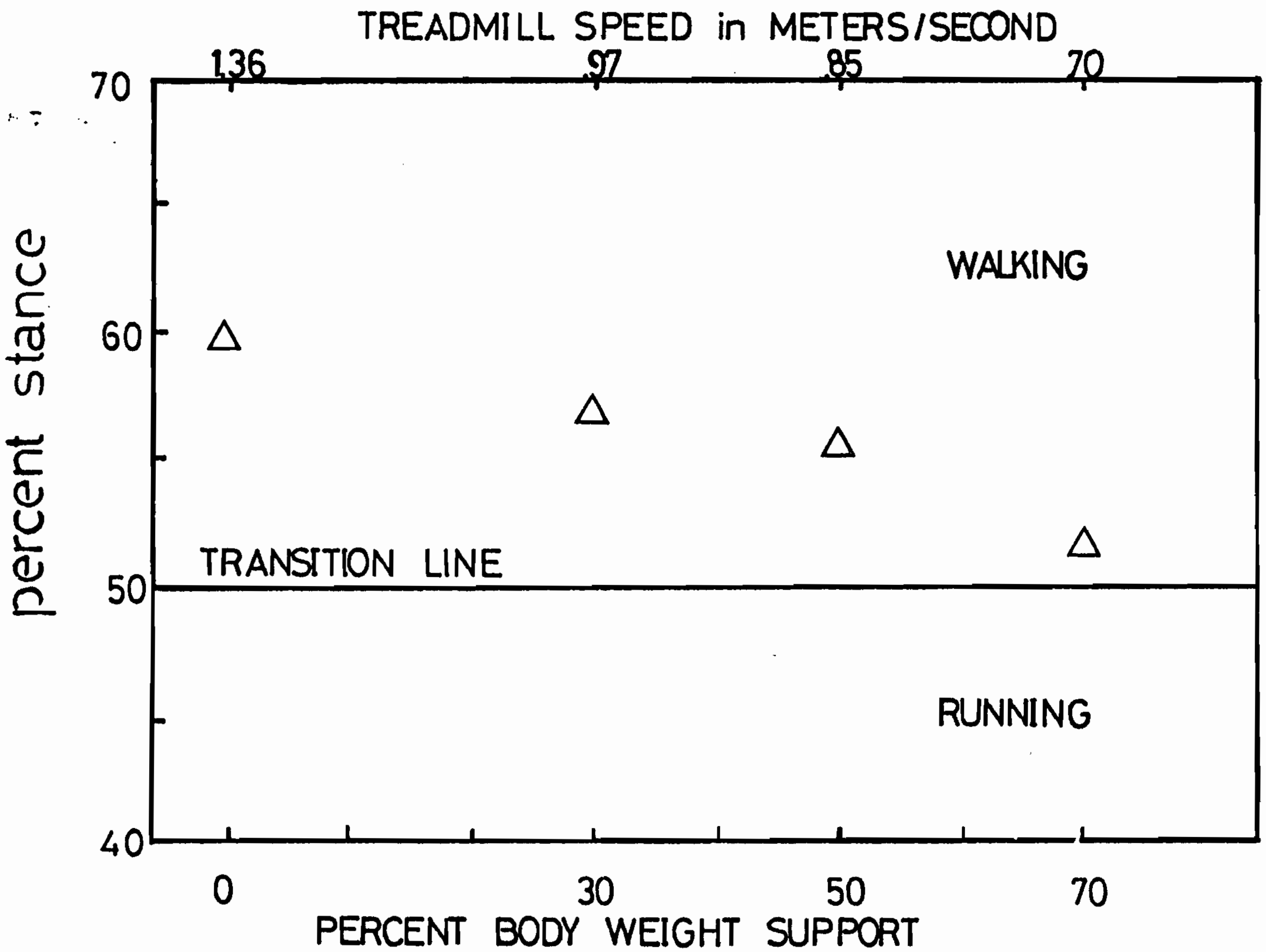
The effects of BWS, above those attributed to speed, were seen in % stance and TDST. Increasing BWS and decreasing speed have opposing effects on % stance and TDST. The effects increase proportionately with increasing BWS, but not with decreasing speed. The progressive decrease with BWS could reflect true BWS effects or effects resulting from the interaction between BWS levels and the set speed.

There are, however, two confounding effects within these results. First, the values in table 3, the result of increased BWS and decreased speed suggest that the weight effects are underestimated. For example, in figure 10, assuming that the effects of speed and weight are additive the lower dashed line would represent the true weight effect without the effect of decreasing speed. Second, within the % stance results there are two phases - single limb support time (SLST) and double limb support time (TDST). The weight effect on TDST is evident in figure 11, but when the TDST is subtracted from the BWS and speed % stance times, the resultant SLST increased (see table 3). Decreasing TDST had a greater influence in reducing overall stance time than in reducing the SLST. This makes balance more difficult, as subjects usually increase double support time for balance and equilibrium at slower speeds (Herman et al. 1976, Gabell and Nayak 1984). This is compounded on a treadmill. Murray et al. (1985) reported an increase in TDST required for balance while walking on a moving surface.

The subjects in this study walked at reduced speeds, while decreasing their double support time, and supporting their body weight, (albeit less body weight), on a single limb for longer periods of time, as is the case in running. The transition between walking and running usually occurs at 2 m.s-1 (Vaughan 1984). The speed at which the transition from walking to running occurs may be reduced, at reduced body weight in the normal. Under lunar gravity, or 70% BWS which can be considered roughly equivalent, the speed of transition between walking and running might be .80 m.s-1, if as stated by Hewes et al. (1967), gait under lunar conditions is 60% slower. The mean treadmill speed of the subjects walking at 70% BWS was .70 m.s-1. These subjects might then be considered as walking close to the transition speed. Figure 26 illustrates this. The line at 50% stance time portrays the theoretical transition between walking and running (defined as an absence of double support time when stance and swing times are equal). The transition point is approached progressively from 30 to 50% BWS and attaind at 70% BWS and is a result of the paradoxical decrease in double support time with decreasing speed. It consequently appears that subjects had to run to keep up with the treadmill at 70% BWS.

The parameters in table 3 are consistently and progressively more variable with increasing BWS than with decreasing speed. A decrease of afferent input with BWS may lead to more variable output, subjects may not be totally habituated to the system, and the speed of walking may

Figure 26 Theoretical transition line between walking and running. This graphic representation is not meant to be taken quantitively, but qualitatively.


TREADMILL SPEED in METERS/SECOND
1.36
.97
.85
.70
70
60
50
40
percent stance
WALKING
TRANSITION LINE
RUNNING
0
30
50
70
PERCENT BODY WEIGHT SUPPORT

not be appropriate to the amount of support. These factors can lead to a greater expression of biological variation.

6.3 ANGULAR DISPLACEMENT FACTORS

The angular displacement measure most sensitive to change is open to debate. Initially, the total mean range of joint movement was chosen to reflect changes due to BWS. In addition, this global measure was then to be compared to existing data by Hewes et al. (1967) for BWS, and Smidt (1971) for speed. The need for a more detailed analysis of the location of change within the gait cycle and its significance in relation to other parameters soon became obvious. The total mean movement diluted the specificity of BWS effects, therefore one critical event of significance, the hip and knee maximum swing angle (MSA), was analysed quantitatively, the others qualitatively.

6.4 Total Mean Angular Displacement

6.4.1 Hip and Knee Speed Effects

The results follow a trend similar to those reported by Murray et al. (1966), Smidt (1971), and Winter (1983). Yet only the hip angular displacement at the slowest speed differed significantly from the 1.36 m.s-1 speed. Smidt, studying overground walking, found a significant decrease in hip angular displacement between the speeds of .91 and 1.34 m.s-1. Hip angular displacement (mainly extension) (Murray et al.1985) decreased for subjects walking on a treadmill when compared with overground walking. Therefore, a decrease in hip displacement with treadmill walking may require larger intra-speed decreases to produce a significant decrease in displacement.

6.4.2 BWS Hip Effects

Hip displacement decreased with BWS and speed; however, there was an abrupt initial decrease in hip angular displacement between 0% and 30% BWS, followed by a steady decrease, while with speed, displacement decreased steadily (table 4 and figure 14). Pierrynowski et al. (1981) also reported an abrupt gait pattern change but with an increase in load, but without an additional change for increasing loads. Subjects may react in a parallel fashion with decreasing loads. Subjects walking under reduced gravity, however, did not demonstrate significant differences in total hip displacement. It may be that the initial decrease in movement at 30% BWS can be attributed both to decreasing weight and speed, while the further decrease with BWS is due to speed alone.

The harness did not restrict hip displacement, but it did affect the downward movement of the body. The unrestricted hip movement and decreased body movement along with the change in height of the trochanter may have affected the hip angular displacement. The exact relationship between decreasing speed, increasing BWS and decreasing downward movement, as well as, the effects on hip angular displacement is difficult to determine.

The increase in BWS makes gait variations more apparent. This could be a result of less constraint on walking with less weight and/or a result of increased forward trunk inclination and measurement error due to pelvic rotation.

6.4.3 BWS Knee Effects

The largest effect of BWS on knee angular displacement was a 16% decrease at 70% BWS (table 4). This decrease may be an indirect result of the decreased hip movement - less displacement is needed with removal of body weight. The total mean angular displacement measure only permits speculation into the origins of the decreased displacement. It reflects the whole range of knee displacement and as such may mask specific areas of change.

6.5 MSA

6.5.1 Hip and Knee Speed Effects

Figures 16 and 17 demonstrate a consistent pattern of angular displacement with decreasing speed similar to those of Winter (1983). Note the similarity of MSAs at the hip and knee with decreasing speed.

6.5.2 BWS Hip Effects

The major decrease in hip angular displacement was at HS, FF, HO and the MSA. The greater decrease at 70% BWS may be the result of increased pelvic rotation. The rotation decreased the amount of hip flexion required throughout swing to advance the limb.

Knee flexion progressively decreased during yield with increasing BWS. This lack of knee flexion during early stance, according to Inman et al. (1981), would cause the hip joint to extend from the beginning of stance, consequently reducing the hip flexion. Hewes et al. (1967) also noted a decrease in hip extension (masked in their total range measure). The hip displacement with BWS at HO decreased in a similar fashion. Murray et al. (1985) also reported a decrease in hip extension at HO for treadmill walking compared to overground walking; this decrease may be further decreased by BWS.

6.5.3 BWS Knee Effects

The major areas of decrease in knee displacement were TO, FF and MSA each being 6 - 10 degrees less compared to FWB levels (table 5). Hewes et al. (1967) found a decrease in knee flexor angles at FF attributed to the decreased amount of weight carried. The subjects in this experiment not only supported less weight on their limbs leading to a decrease in knee flexion at FF, but also were prevented from yielding downwards at FF by the harness. There might also be a decrease in the transfer of kinetic energy from push off proportionately decreasing momentum and subsequently decreasing knee flexion during swing.

6.6 EMG Timing Factors

6.6.1 Speed Effects

The timing of muscles relative to the gait cycle is similar to that reported for normal subjects walking at comfortable (Thorstensson et al. 1982, Basmajian 1976, Battey and Joseph 1966), or slower speeds (Nilsson et al. 1985, Herman et al. 1976, Mann and Hagy 1980). Even the considerable variation present both within and between subjects in the EMG on/off timing of every muscle at different speeds (appendix 6, table 6a and 6b and figure 18) is similar to the above authors.

The increased variability may be related to walking on a treadmill, especially at slower speeds. Both a difference in environmental cues and a decrease in vertical body movement (Murray et al. 1985) can lead to a decrease in the perception of movement and vestibular inputs influencing the neural control of gait. The greater intra-subject variability, may reflect the individual differences in response to the set speeds (Medeiros 1978).

Decreasing speed appears to have influenced the ankle muscles less than the more proximal hip muscles. The off timing of MH and ES demonstrate a varied response, probably as a result of the greater flexibility of MH, (a two joint muscle (Brandell 1977, Winter 1983)), and the different equilibrium demands of decreasing treadmill speed on ES (Thorstensson et al. 1982, 1984). The increased variability of TA off timing is likely a result of analysing only one TA burst.

The phase relationship between HS and on timing (Mederios 1978) was consistent. The strength of this phase relationship is lower (r=.72, Herman et al. 1976) than between other parameters. Nevertheless, the relationship between increasing cycle times and increased extensor duration (figure 18) has been reported (Pearson 1976, Herman et al. 1976, Mederios 1978) and may probably be related to the positive total support moment required in stance.

6.6.2 BWS Effects

Except for ES1, the on/off timing across increasing BWS levels is similar to that of decreasing speed (figure 18). The variability of EMG timing, both intra and inter-subject, however, increased more with BWS, the proximal muscles varing more than the distal. The increased variability may include an interaction between speed and weight that is arithmetic or unstable over the range of speeds used. Other factors involved could reflect individual responses to the different mechanics required in BWS walking (for example, decreased afferent input); the removal of body weight appears to affect individuals differently.

BWS affected the phase relationship between HS and timing slightly for all muscles and not at all for VL and GA. The adaptations required to walk at a set speed at BWS levels may have led to an earlier (ES1) and longer periods (ES2) of back muscle activity.

The relationship between stance time and extensor muscle timing appears to hold even with increasing BWS levels. MH, however, starts later and finishes much later with BWS which may reflect the greater need to control the limb as it swings forward. The relationship of MH to the swing phase may be worthwhile investigating. Though the TA, TDST relationship needs to be clarified and expanded, the decreased TDST in figure 11 may require an earlier TA burst for balance control.

6.7 EMG Mean Burst Amplitude Effects

If the alignment of the harness and trunk was not entirely satisfactory as sometimes when body weight was removed, weight was removed unequally. Muscle activity then reflects the adaptations produced by the uneven load distribution. To try to control this, great care was taken to ensure proper fitting and alignment of the harness. In addition, the harness straps never interfered with the electrodes. As there was no measure of weight distribution between limbs, equality of distribution was judged qualitatively. Despite this, the EMG mean burst amplitudes for all subjects demonstrated a similar trend.

While an EMG evaluation of muscle mean burst amplitude is specific as to both the timing and amount of activation, it does not measure the peak amplitude or time history of amplitude change. Although extremely variable in nature (Grieve and Cavagna 1974, Yang and Winter 1985), these latter two measures appear the best available to relate joint moment histories to specific muscle activation during locomotion.

6.7.1 Speed Effects

Because of the wide variability in muscle amplitudes and different normalizing procedures employed it is hard to make comparisons across studies. Nevertheless, the changes in amplitude (table 7) due to decreasing speed compare favourably with the existing literature (Longhurst 1980, Yang and Winter 1985). The effects of speed were more evident proximally in ES, VL and MH than at the ankle. While ES(1), GM and TA changed little, GA decreased slightly and ES(2), VL and MH showed moderate decreases with decreasing speed (table 7). Due to variability, significant statistical differences within a specific muscle between speed levels were not apparent in VL, MH, or ES(2).

Hershler and Milner (1979) found EMG amplitudes varied less if the subjects walked at a "set comfortable" speed, between .91 to 1.51 m.s-1. The set speed combined with the normalization techniques used here may account for the decreased variability of mean burst amplitudes in table 7 compared to other studies (Yang and Winter 1985, Mann and Hagy 1980). Although the variability of the amplitudes are relatively low, they increase with decreasing speed similar to those of Yang and Winter (1985) but still prevent statistical differentiation.

VL was not used by a number of subjects at slower speeds. Knee extension can be achieved without muscle activation in some cases (Battye and Joseph 1966, Brandell 1977, Yang and Winter 1985). Brandell (1977) and Pedotti (1977) found VL and VM (Vastus Medialis) to vary considerably under different speeds and loads. These two factors may account for the variability of VL mean burst amplitudes with speed (table 7). Also, there may be compensatory activity by other vastii that would add to the variability at the knee (Arsenault 1982, Pedotti 1977).

MH serves to decelerate the knee and as such the mean burst amplitude would tend to decrease with decreasing angular acceleration of the knee (Inman et al. 1981, Winter 1983).

ES2 brakes lateral trunk displacement (Thorstensson et al. 1982, 1984) and decreases in amplitude with decreasing speed. An additional feature of ES bursts is that the amplitudes of ES1 and ES2 (table 7) are of a similar magnitude, possibly indicating that there is more sagittal plane than frontal plane trunk movement (Thorstensson et al. 1982, 1984).

6.7.2 BWS Effects

Modification of muscle activity due to BWS should be dominated by gravitational forces. Each limb's weight remained the same, only body weight was reduced. The forces produced and countered while walking at specific BWS levels could not be measured, consequently, discussion of muscle activity is partly speculative.

Muscles affected by BWS will be those active at weight acceptance (HS) and push off phases of stance, that is; ES, GM, TA and GA.

The decrease in ES mean burst amplitude is likely due to weight alone (table 7). Both ES1 and ES2 decreased with increasing BWS. ES1 brakes forward trunk movement at HS (Thorstensson et al. 1982, 1984, Morris and Waters 1973), and with less weight a smaller braking reaction would be required significantly decreasing ES1 in amplitude. The large variability (table 7), and decrease in the number of subjects using ES1 (4 at 70% BWS), however, precludes a definitive statistical inference.

ES2 brakes body movement to the contralateral side. Only the sagittal plane was examined, but from the larger and consistently used ES2 mean burst amplitude it may be inferred that BWS has a smaller effect on lateral trunk movement than on movement in the sagittal plane. Although BWS decreases the ES2 mean burst amplitudes, (as more weight is removed less weight is transfered), ES2 is still required to control lateral weight transference (figure 20).

GM, a postural muscle used for stance stability during SLST, demonstrates a modification similar to ES1 (figure 21). There was a progressive decrease in mean burst amplitude with little activation at 70%. The need for body weight stability decreases as body weight declines and is supported by a harness. There remains a need for body on lower limb stability as demonstrated by the statistacially non-significant decrease in GM amplitude till the 70% BWS level.

TA controls ankle position during swing and HS. Figure 25 demonstrates that BWS, but not speed, affected TA activity. The mean burst amplitude increases (30%) with increased BWS. The effects of BWS compared to those of speed are difficult to intrepret. All subjects, bar 3-4, had a single TA burst of activity at BWS levels, while all subjects, again bar 3-4, had 2 bursts at slow speeds. Because only the first burst was analysed, the TA mean burst amplitudes with BWS may be artifically high. The trend remains, nonetheless, towards an increase of TA activity.

Hewes et al. (1967) noted that the ankle joint of subjects walking under lunar conditions oscillated between plantar and dorsiflexion. It may be that the increased TA activity seen in figure 25 reflects a similar lack of control at the ankle joint with BWS. Clement et al. (1984), studying postural adaptations to weightlessness, found that the control and erectness of posture was mainly due to contraction of ankle flexors. The increase in TA utilization (table 7) may, therefore, compensate for the decreases in ES1 pre-HS. Whatever the cause of the increase in TA mean burst amplitude, it appears to be related to the removal of BWS, as evidenced by Herman et al. (1976), who observed no such increase in activation for subjects bearing full weight.

Before push-off gastrocnemius discharges to propel the body forward. This discharge can be influenced by the duration, rate and magnitude of the applied forces (body weight) (Herman et al. 1976, Monster 1976). As a consequence the GA is expected to decrease and as figure 24 suggests this decrease is as much as 50%. As a result of removal of body weight other inputs decrease notably proprioceptive (stretch to the muscle spindle), cutaneous (Pierrot-Deseilligny et al. 1983, Conrad et al. 1983) and vestibular (decreased arc of CG movement), all of which may affect the GA mean burst amplitude. Further research into the correlations between decreasing EMG amplitude, joint moments and BWS levels would allow for better modelling of BWS effects.

Although not significant, the trends in VL seem to be towards a decrease in amplitude. VL contracts, by an amount dependent on weight acceptance, in the support phase prior to the knee extensor moment (Pedotti 1977). Because weight acceptance and knee flexion at FF are less the need for active extension should be less.

MH appears to be affected by speed and weight. The mean burst amplitudes in table 7 are very similar across BWS levels and do not decrease till the 70% BWS level. Speed of angular displacement, although not measured, may account for the decrease at 70%.

6.8 GENERAL CONSIDERATIONS

Two factors controlled in this experiment were the amount of decrease in speed and the amount of increase in body weight supported.

6.8.1 Speed

The adaptation to decreasing speed appears due to a decrease in both cadence and amplitude of leg movement. This agrees with the observations of others (Herman et al. 1976, Mann and Hagy 1980, Nilsson et al.1985, Winter 1984), who found these parameters to increase with increasing speed. Despite differences in determination of the gait cycle, the relative relationship between cycle time and stance time with decreasing speed (table 3) also agrees with those of Thorstensson et al. (1984), and Grillner et al. (1979). As cycle time increases with decreasing speed, the relative change in stance was larger than that of swing.

It is likely that the increased TDST is responsible for the % stance increase and may also modify the angular displacement and EMG parameters. The dictated slow speed would probably decrease limb acceleration and produce a shorter stride length and longer cycle time. Both Nilsson et al. (1985) and Smidt (1971) suggested that changes in hip sagittal motion contribute the most to changes in stride length. Thus the shorter stride length in table 3 is probably related to the decreased hip angular displacement (table 4). A decrease in pelvic rotation (not assessed quantatively) might also play a role in decreasing the stride length (Smidt 1971, Murray 1967).

The greatest areas of decreased hip and knee angular displacement (figures 14 and 15) occurred during periods of TDST, HS to FF and HO to TO. Although not statistically significant, the decrease in angular displacement at these points may reflect the influence of an increased TDST.

Winter (1983), and Yang and Winter (1985) studying slow walking, found proportional changes in joint acceleration (Winter 1983) and deceleration with decreasing speed which modified mainly the hip and knee muscle EMG linear envelope, the ankle being less affected (Yang and Winter 1985). An average amplitude EMG measure, the mean burst amplitude, was used and although a statistical difference was not evident, the means in table 7 follow the same trend. That is, VL and MH mean burst amplitudes decrease with decreasing speed, while GA and TA remain relatively stable.

An EMG analysis of where in specific bursts the peaks of activity occur might help delineate further changes. The variability seen with most EMG analysis, however, may preclude a definitive statement. As suggested by Winter (1984) and others (Herman et al.1976, Shik and Orlovsky 1976), it might be more beneficial to examine patterns of muscle activity in terms of their co-ordinated functions rather than their individual patterns.

Despite decreasing speeds, the TD results and angular displacement patterns were stereotypic. Even the EMG timings relative to the gait cycle remained stable with decreasing speed; the mean burst amplitudes, however, varied. The increased variability noted may reflect adaptations by the sensorimotor system and/or of the muscle itself to unaccustomed slower speeds (Herman et al. 1976, Medeiros 1978). The increased variability also may be in response to the increased postural control required, while walking slowly on a treadmill (Murray et al.1985). This is demonstrated by the small change in GM and ES, (postural muscles), mean burst amplitudes (figures 20 and 21); but an increase in their duration as TDST, an indicator of postural control, increased.

6.8.2 BWS

Adaptations to increasing BWS appear related to two factors, 1) changes in the height of the trochanter and the downward displacement of the CG combined with the dictated speed and 2) effects of removing body weight. For example, when trochanter height changed, subjects walked slower or progressively increased their pelvic rotation to compensate for the decreased contact distance, thus producing decreasing TDST and possibly the hip MSA. Associated with the removal of body weight is the initial decrease in hip angular displacement, the decreased knee flexion at FF and the decreased mean burst amplitudes of GM, ES and GA. The body appears to pivot about a central point with 70% of its weight suspended in the harness. Little muscular activity is required to move, as witnessed by a lack of activation in ES1, VL and GM in a number of subjects. Using a measure of force during walking, it could be ascertained if the subject is walking on the treadmill or the treadmill is walking the subject.

Regulation of cycle time and thus muscle activity depends on the speed of walking (Mann and Hagy 1980) and amount of weight supported (Neumann and Cooke 1985). Because BWS and FWB cycle times were similar and BWS decreased the TDST, the % swing increases. The decreased % stance and contact distance would produce an increase in the angular velocity of the leg. The expected velocity, however, would be less during swing, (but with a slight increase in stride length with BWS) since the limb must travel further. The stride length increased by 10%, while the % swing increased by 20%, at 70% BWS over that attributed to speed. The actual velocity was not determined, but it is certain that the decrease is not as great as the increase in time would suggest.

The presumed velocity change during stance would require a change in muscle force to decelerate the body, stabilize it, and then accelerate it again. But with BWS, as the body mass has decreased, the necessary force is lessened. This is observed in the progressively decreasing mean burst amplitudes of ES, GM and GA with BWS (table 7 and figures 19, 20, 21 and 24).

In contrast, during swing, the weight of the limb remains the same, thus more force may be required at the hip to swing the limb. Despite the slower gait and because of the mechanical constraints of the harness, a larger hip flexor torgue, combined with an increased pelvic rotation, may be necessary to swing the leg forward the required distance at a specific speed. The later off-timing (figure 18) and larger BWS mean burst amplitude of MH required to control hip flexion and pelvic rotation supports this point. In addition, the increased MH, working more at the hip just before HS, may compensate for the decreased ES activity. That the hip is responsible can only be inferred as there were no direct measures of hip flexor or pelvic activity. Future work to test this hypothesis should be undertaken.

The decrease in the number of subjects and ES activity reflects the adaptation needed to control and produce the trunk movement with BWS, while the hip and knee angular displacement patterns remained similar compared to those of slow speed. Winter (1983) suggested that the flexibility of the two joint muscles maybe responsible for the consistent angular displacement.

If the output patterns remain similar, the timing of the input producing the pattern would be expected to remain the same as shown in figure 18 (except for ES1 needed to control trunk movement). The EMG timings may, therefore, be a redundant parameter in normals. EMG timings may also be redundant for patients when displacement patterns are also available. In any event the variability is such that normal and pathological timings often overlap.

6.9 Implications for gait retraining

Gait studies demonstrate that TD (Murray 1967), angular displacement (Perry 1978), and EMG (Knutsson and Richards 1979), abnormalities are associated with a patients' limited walking speed and poor weight bearing capacity. Indeed, treatment concepts (Bogarth and Richards 1981) advocate the need to improve weight bearing and control of weight transference before training gait. Improvement post-treatment is then judged by a patient's ability to walk faster (Mizrahi et al. 1982) and bear more weight (Mizrahi 1985). Both the increased speed and increased weight bearing have been linked to further functional recovery (Brandstater et al. 1983, Holden et al. 1984).

Postural instability and lack of balance control are a large part of the reason neurological patients walk slowly and with difficulty. The training of stance balance is a major concern and a large part of any retraining program. Before a patient can walk, momentary single limb standing balance is taught, usually separately from walking. The training of single limb balance by decreasing the TDST combined with walking may be more beneficial. The SLST (table 3) increased slightly with BWS, while the TDST decreased drastically. Therefore, starting at an appropriate BWS level a patient learns to balance and to walk simultaneously. It may appear more difficult to train both balance and walking, but certain factors are favourable. For example, the patient only needs to balance and propel a portion of his body weight while having the security of the harness to relie on. The literature shows that, removing weights from normal subjects, previously weighted on one side, can lead to a resumption of normal TD parameters; stance, TDST time and stride length. (Eke-Okoro and Larsson 1984). Hemiplegic patients may have the same positive response.

Progressive decrease of BWS levels combined with treadmill stimulation should allow for gradual retraining of postural muscles necessary in gait. The mean burst amplitudes of ES and GM demomstrated a gradual decrease with increasing BWS (table 7). Patients lacking the muscle control for postural support can develop not only the necessary strength, but also the co-ordination required to maintain stability while their body weight is supported by the harness.

Patients need a proportional relationship between flexion and extension to prevent ill-timed and abnormal movement from developing during gait (Dimitrijevic et al. 1981). Patients lack the amplitude and smoothness of movement, but especially the ability to switch from extension to flexion. The pattern and smoothness of joint angular displacement was not significantly affected by BWS (figures 14 and 15). For man, the critical event of HO may be the turning point between flexion and extension, and if so it would be the initiation point for swing. BWS levels decreased hip extension insignificantly at HO. The hip angular displacement would thus appear adequate to facilitate the switch to flexion in neurological cases and possibly, as suggested by Andersson and Grillner (1983), facilitate flexor muscles of the entire limb. The other feedback signal switching extension to flexion, the amount of weight the limb supports at HO, would be less with BWS. The extensor muscles will not allow a release to flexion if a load in series with the achilles tendon is high (Duysens and Pearson 1980). The BWS system could facilitate flexion in patients unable to control the unloading of their limbs.

During swing, the limb weighs the same. However, the effective weight will increase if the spastic restraint of the extensors or weakness of the flexors is excessive. The decreased demand for postural stability combined with a decrease in vestibular stimulation (decreased displacement of CG) may help decrease the abnormal extensor activity responsible for patient's abnormal pelvic retraction and poor swing phase initiation (Dimitrijevic and Larsson 1981). Whether sufficient flexion will develop then depends on the amount of hip flexor facilitation.

If a patient is able to swing his leg adequate flexor mechanisms may be available, but the patient may lack adequate extension for support. Therefore, not only must flexion be facilitated, but also extension. Training using progressive BWS levels not only allows for facilitation of flexion, but also allows the gradual strengthening of extensor muscles as BWS decreases (figures 19, 20, and 21).

Muscles active during the walking cycle can be trained at their functional lengths; first starting with small loads and then progressing to larger loads. Thus the BWS treadmill stimulation approach appears ideal.

A variety of muscle patterns develop in neurological cases. While walking FWB patients develop abnormal early GA activity in response to body weight. The mean burst amplitude of GA decreased, however, in normal subjects with BWS and slow treadmill speed (table 7 and figure 24). This may be due to the decrease in the load size (decreased body weight), the rate of load application (slow speed), and the duration of the load (decreased TDST). The proper phasing of GA could be facilitated in patients with the BWS technique. One effective gait retraining program decreased the stretch on GA in early stance in combination with weight acceptance exercises (Richards and Knutsson 1974). BWS training can be as effective.

Any gait pattern produced depends on the pathology and is influenced by peripheral factors, such as the cycle phase and the muscle responses to stretch and cutaneous inputs. Therefore, re-education programs must consider deficits on an individual basis and choose an appropriate BWS level according to the clinical symptoms encountered. Initially, training at an appropriate BWS level and slow speed should decrease the effort and force of contraction required to walk.

The 70% BWS level, however, may reduce the inputs to such an extent that poor quality or little walking will be stimulated. The 70% level decreased the TDST drastically making trunk control difficult. Nonetheless, the decreased TDST, the observed increase in SLST and trunk pertubations were probably the result of the set speed and raised trochanter rather than the effect of reduced weight. Any BWS level, short of 100% support, may prove, therefore, beneficial for training, if the treadmill speed is adjusted to the patients abilities and BWS conditions.

Once started, the amount of BWS can be decreased and speed increased as the patient improves. Improvements should be judged, as suggested by Brandstater et al. (1983) and Holden et al. (1984) and others (Sutherland 1981, Mizrahi et al. 1982), by the patients ability to walk faster. Improved walking speed should also be reflected in other parameters; cycle time, % stance, and TDST, cadence and stride length as well as balance and function.

7. LIMITATIONS AND SUGGESTIONS FOR FUTURE RESEARCH

It is recommended that further study be directed towards the following topics;

Evaluate the weight distribution between the limbs. Without a way to evaluate weight distribution essential in neurological gait, the effects of training may be negated.

Investigate further the relationships between BWS levels, the height of the trochanter, contact distance, pelvic rotation and speed. The interaction between BWS levels and set speed influenced a number of parameters.

Investigate the role of the hip. The force required by the hip flexors for swing may be increased with BWS in normals. Thus the decreased extensor hip amplitude and unweighting of the limb may not be sufficient to facilitate hip flexion in patients.

There is a need for a detailed accurate link between the three sets of parameters to thoroughly assess the relationship between muscle activity and movement patterns with decreased weight loads.

Evaluate further the possibility of an increasingly sensitive EMG analysis via linear enveloping the mean burst amplitude.

Investigate the EMG timing swing phase relationship of flexor muscles.

Develop a method of weight support for overground walking to investigate the influence of treadmill stimulation on supported walking.

B. CONCLUSIONS

Normal gait under various BWS levels was compared to normal FWB gait to determine if a strategy of partial weight in combination with treadmill stimulation could be developed for gait retraining of neurological patients. Adaptations to BWS are related to two factors. Firstly modifications attributed directly to the removal of body weight are few and include;

An initial decrease in total mean angular hip displacement.

A decrease in knee angular displacement at FF.

The decreased load on the limb reduced the mean burst amplitude of ES1, GM, and GA. As well, an increase in mean burst amplitude of MH (for increased limb movement control) and possibly TA (for increased postural stability) developed.

The secondary modifications developed related to the mechanical constraints of the BWS system and the dictated speeds and include;

A change in the height of the throchanter. The support system raised the position of the trochanter, secondarily decreasing the contact distance between the feet and limiting the downward path of the CG.

The decreased contact distance decreased the TDST and increased the SLST. The amount of time both feet of a subject are in contact with the ground decreases, while the time for any one increases slightly. An increase in pelvic rotation may have occured to increase the stride length to maintain the set speed at any BWS level.

A change in ES1 timing occured probably to control the increased frontal trunk movement at HS.

These secondary modifications can be controlled (increased, decreased or eliminated) with a better understanding of BWS/speed relationships.

The results led to the development of a BWS treadmill training scheme. The postulated scheme offers several advantages over current methods.

The training technique does not produce abnormal walking.

As the weight supported by the limb is reduced, this technique should provide an easier progression from the stance to swing phase of gait.

The technique should provide for dynamic simultaneous training of balance, postural stability and stepping.

It should strengthen muscles at their functional lengths.

It should require less effort for the patient to master and therapist to learn.

Training techniques used need to be re-evaluated in the face of new developments in research. Patients with ability and those in the early stages of recovery need effective strategies to enhance their abilities. A human retraining strategy has been proposed similar to that used in animal experiments. It should now be applied to influence the training of patient's motor patterns to produce safe efficient gait.

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APPENDICES

APPENDIX 1.

INDIVIDUAL PRELIMINARY FREELY CHOSEN TREADMILL SPEEDS.

TABLE A2 - Freely chosen mean treadmill (TM) speed ($m.s^{-1}$) and range of speeds selected by the unhabituated and habituated subjects for each body weight support (BWS) level (in percent body weight support). N represents the number of sugjects tested per level.

BWS	N	TM speed	Range	Habituated TM speed
0	7	1.16	.81-1.16	1.34
20	5	.93	.72-1.23	1.26
30	5	1.05	.82-1.36	1.36
40	4	.90	.78-1.06	1.06
50	6	.75	.59-1.05	.95
60	6	.88	.54-1.05	1.04
70	5	.62	.40- .80	.75
80	7*	.44	.25- .66	unable

* only 3 subjects could walk at this BWS level.

APPENDIX 2.

SUBJECT ANTHROPOMETRIC DATA FORM.

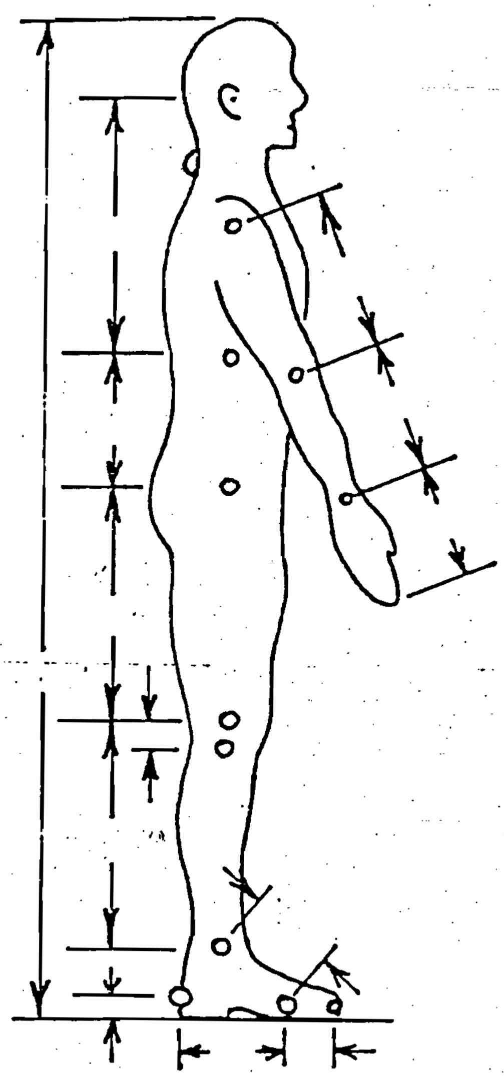
INDIVIDUAL SUBJECT ANTHROPOMETRIC DATA.

MODIFIED PHYSICAL EXAMINATION SHEET.

CODE NO.:_____ SUBJECT/PATIENT ANTHROPOMETRIC DATA SHEET DATE:_________

NAME_______________________________ HEIGHT:_____ WEIGHT:_____ AGE:____ SEX:___

*



Anatomical Location of Markers

Rib - midline of rib cage half way between illiac crest and shoulder

Hip - greater trochanter

Knee - lateral femoral epicondyle (about 2 cm. above knee line)

Ankle - lateral malleolus of fibula

Heel - about 2 cm above ground in line with rear of shoe

Meta - 5th metatarsal phalangeal joint

Toe - about 2 cm. above sole in line with front of shoe

Units & Timing

* Taken from Winter (1979)

TABLE A2a - Age, height, weight, leg length and foot length for subjects.

Subject Number	Age years	Height meters	Weight kgms	Leg Lenght meters	Foot Length meters
1	33	1.715	66.74	0.838	0.254
2	33	1.708	69.01	0.876	0.254
3	25	1.803	67.65	0.914	0.254
4	35	1.803	72.19	0.914	0.254
5	33	1.746	74.46	0.895	0.273
6	31	1.740	85.35	0.889	0.260
7	26	1.778	68.10	0.902	0.279
8	35	1.708	66.28	0.813	0.267
9	27	1.772	84.90	0.902	0.254
10	32	1.791	71.73	0.889	0.241

Subject Data Sheet.

Code #		Date

Age		T.M.experience

1. Are you in good physical condition?

2. When did you last see a doctor?

3. Have you ever injured your ankle?

4.Have you ever injured your knee or leg?

5.Have you ever injured your back?

6. Do you ever have recurring bouts of low back pain?

or sciatica?

or pain down your leg?

Measures to be taken

1. ASIS are they level?

2. Leg length from the ASIS to medial malleolus. rt cm.
lt cm.

3. SLR (degrees) rt lt

4. forward flexion cm from the floor?

Are there any pertinent scars?
Do you have any medical condition such as Diabetes, heart condition, kidney problem ?
If yes what?

Are you presently taking any medication?
Which one(s)?

Blood pressure and pulse readings

	BP	Pulse
Initial reading		
After 1st habituation		
After 2nd habituation		
After trial 1		
2		
3		

APPENDIX 3.

SUBJECT TREADMILL INFORMATION SHEET.

SUBJECT CONSENT FORM.

LABORATORY of LOCOMOTOR FUNCTION
SCHOOL of PHYSICAL AND
OCCUPATIONAL THERAPY
McGILL UNIVERSITY

Informed consent form

The nature and purpose of the present study have been clearly explained to me, that is to examine the influence of partial weight bearing on walking patterns.

I have been informed of the various techniques used in the analysis of treadmill walking (surface electromyography and video recording). The harness, the means of supporting me over the treadmill, and the operation of the treadmill have been explained to me.

I understand that the examinations and the training I shall undergo are intended to measure locomotor function on a treadmill. The experiment consisting of three trials should last approximately three hours.

I have been informed that there are no foreseeable dangers from this proposed study. I am aware that I can withdraw my consent and discontinue my participation in this study at any time without any prejudice to my well being.

Subject ------------------------ Witness ---------------------

The above mentioned person is aware of the nature of this study and can with draw at any time. I have assured him the information obtained wil be held in confidence.

signature --------------------- date ----------

THE COLLINS #101 TREADMILL

The treadmill speed varies from .25 m.s-1 to 3 m.s-1. The on/off and speed of the treadmill are controlled by a very sensitive remote console. A stop button is placed on the harness near the subject; once pressed the treadmill will stop smoothly in less than 5 seconds. A fail safe mechanism is built in to prevent treadmill restart except at the slowest speed. Parallel bars are also provided for stability during speed changes.

To accustom you to the treadmill a training period is required. The training period consists of walking on the treadmill for 15-20 minutes to determine your natural walking speed. A rest period of one hour follows where the electromyographic (EMG) electrodes and video markers will be applied. A second training period of 10-15 minutes is used to check that all video and EMG recordings are accurate. The experiment itself will start after a 10 minute rest period.

The experiment consists of three trials, each lasting 15 minutes separated by 10 minute rest periods. During each trial you will walk at a set speed on the treadmill. Fifteen step cycles will be recorded at each of four randomly ordered weight support levels from 0% to 70% of body weight.

APPENDIX 4.

MEAN I-D DATA FOR EACH SUBJECT.

TABLE A3 - Gait Cycle Times in Milliseconds

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	0 1.36	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		1067.00	1202.00	1433.00	1494.00	1220.00	1418.00	1525.00
2		1139.00	1278.00	1310.00	1526.00	1256.00	1291.00	1395.00
3		1150.00	1297.00	1522.00	1934.00	1346.00	1370.00	1506.00
4		1033.00	1188.00	1416.00	1627.00	1287.00	1319.00	1534.00
5		1163.00	1412.00	1762.00	2092.00	1326.00	1508.00	1778.00
6		1082.00	1166.00	1420.00	1866.00	1249.00	1404.00	1635.00
7		1009.00	1159.00	1231.00	1651.00	1229.00	1245.00	1356.00
8		982.00	1159.00	1213.00	1351.00	1261.00	1257.00	1401.00
9		1130.00	1146.00	1250.00	1390.00	1266.00	1386.00	1546.00
10		1090.00	1381.00	1404.00	1873.00	1249.00	1418.00	1603.00

TABLE A4 - Stance as a Percent of Gait Cycle

SUBJECT NUMBER	BWS% / Speed $m.s^{-1}$	0 / 1.36	30 / .97	50 / .85	70 / .70	FWB / .97	FWB / .85	FWB / .70
1		57.20	55.00	51.00	47.00	60.00	64.00	62.90
2		57.20	59.20	53.70	56.90	65.90	62.20	63.90
3		60.80	55.90	55.60	45.90	62.00	63.00	66.00
4		55.60	52.30	54.30	54.90	57.50	58.20	58.90
5		63.70	60.00	57.70	50.60	65.20	64.40	64.60
6		61.70	56.10	56.80	44.60	62.90	63.30	64.00
7		58.20	56.30	56.70	48.00	60.40	60.60	60.80
8		58.80	57.70	56.70	56.90	62.10	62.80	61.40
9		62.00	59.00	58.50	57.50	64.80	65.30	65.80
10		64.00	57.00	55.00	55.00	62.00	67.00	67.00

TABLE A5 - Total Double Support Time

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	0 1.36	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		19.00	21.00	9.00	3.00	22.00	25.00	27.00
2*		15.00	12.00	9.00	17.00	32.00	25.00	26.00
4		15.00	11.00	12.00	11.00	21.00	22.00	23.00
5		24.00	17.00	14.00	3.00	29.00	28.00	29.00
6		28.00	22.00	16.00	4.00	31.00	30.00	34.00
7		23.00	22.00	20.00	4.00	29.00	28.00	29.00
8		24.00	19.00	16.00	15.00	28.00	30.00	29.00
9		24.00	18.00	17.00	14.00	29.00	30.00	32.00
10		23.00	12.00	8.00	6.00	26.00	31.00	32.00

* no left footswitch for subject 3

TABLE A6 - Cadence in Steps per Minute

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	0 1.36	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		109.00	94.00	84.00	75.00	96.00	84.00	75.00
2		104.00	103.00	92.00	81.00	98.00	96.00	84.00
3		108.00	90.00	80.00	60.00	90.00	87.00	78.00
4		120.00	104.00	88.00	72.00	96.00	88.00	80.00
5		104.00	89.60	73.60	64.00	88.00	80.00	72.00
6		112.00	104.00	88.00	64.00	96.00	88.00	72.00
7		118.00	108.00	96.00	86.00	96.00	96.00	88.00
8		120.00	102.00	98.00	90.00	104.00	96.00	84.00
9		108.00	102.00	96.00	84.00	96.00	90.00	84.00
10		108.00	86.00	84.00	63.00	96.00	90.00	72.00

TABLE A7 - Stride Length of the Right Leg in Meters

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	0 1.36	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		1.27	1.07	1.05	.98	1.09	1.04	.99
2		1.34	1.13	1.16	.96	1.12	1.02	.95
3		1.36	1.14	1.19	1.32	1.17	1.11	.95
4		1.29	.95	.93	.94	1.05	.98	.88
5		1.33	1.17	1.22	1.00	1.13	1.13	.90
6		1.19	.95	.84	1.04	.98	.95	.93
7		1.20	.98	.90	1.10	1.00	.94	.83
8		1.40	1.13	1.00	1.05	1.17	1.06	1.03
9		1.34	1.02	.92	.89	1.11	1.06	.95
10		1.40	1.18	1.11	1.25	1.07	1.07	.99

APPENDIX 5.

MEAN ANGULAR DISPLACEMENT DATA FOR EACH SUBJECT.

TABLE A8 - Total Hip Angular Displacement

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	0 1.36	30 :.97	: 50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		50	34	32	26	40	42	40
2		50	37	32	27	40	42	42
3		47	37	36	26	45	48	45
4		45	25	35	35	35	40	30
5		47	39	38	32	40	40	40
6		43	35	30	20	45	45	40
7		40	20	25	25	32	35	35
8		45	25	25	25	40	40	35
9		45	45	30	25	45	45	40
10		50	30	30	35	40	43	40

TABLE A9 - Total Knee Angular Displacement

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	0 1.36	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		80	60	55	52	70	70	60
2		70	62	60	50	72	70	67
3		72	67	65	53	75	72	70
4		65	60	70	60	60	60	50
5		72	71	61	52	72	70	65
6		65	70	60	60	65	70	60
7		70	55	55	40	65	65	65
8		60	55	50	50	60	60	55
9		65	75	60	55	60	70	65
10		70	60	55	45	70	65	60

TABLE A10 - Means and standard deviations (SD) of hip and knee angular displacement for critical events of gait; heel stride (HS), foot flat (FF), midstance (MS), heel off (HO) and toe off (TO), for each BWS and FWB speed. Positive numbers are flexion, negative numbers extension.

	BWS%	0	30	50	70	FWB	FWB	FWB
	Speed $m.s^{-1}$	1.36	.97	.85	.70	.97	.85	.70
Hip	HS	23.7	17.2	15.8	17.2	23.9	21.2	19.5
	SD	4.9	4.2	3.6	6.5	4.2	4.6	4.9
	FF	18.0	11.5	8.9	10.0	16.0	15.0	12.7
	SD	7.8	5.8	4.7	5.0	5.7	7.5	6.5
	MS	.5	.5	.4	1.2	2.3	1.3	1.0
	SD	3.7	2.8	3.5	4.8	3.4	3.5	3.2
	HO	-12.0	-8.0	-7.2	-5.0	-10.3	-12.2	-9.5
	SD	-3.3	-4.8	-2.6	-5.8	-3.4	-2.7	-3.7
	TO	7.5	7.0	7.5	5.7	8.7	10.2	7.5
	SD	5.9	6.7	5.8	6.7	6.2	4.4	5.4
Knee	HS	5.7	4.2	4.1	5.9	6.0	6.1	5.5
	SD	4.8	5.3	4.5	4.2	5.2	5.3	4.4
	FF	13.7	6.9	5.1	7.6	9.8	7.1	8.2
	SD	9.2	3.6	4.5	4.4	6.1	5.8	7.4
	MS	8.7	7.2	6.9	10.7	10.0	7.6	8.5
	SD	4.4	4.5	3.7	8.5	4.7	4.4	3.3
	HO	16.2	17.3	13.2	17.4	18.9	17.7	19.7
	SD	10.4	7.1	9.2	8.4	8.2	5.8	6.7
	TO	64.9	60.5	56.2	53.5	65.7	64.9	60.5
	SD	3.8	9.2	6.3	9.1	6.6	6.7	4.9

APPENDIX 6.

MEAN EMG ON AND OFF TIMING FOR EACH SUBJECT FOR EACH MUSCLE.

TABLE A11 - On timing of erector spinae first and second burst (ES1, ES2), gluteus medius (GM), vastus lateralis (VL), medial hamstring (MH), gastrocnemius (GA) and tibialis anterior (TA1, TA2) for each body weight support (BWS) and full weight bearing (FWB) speed.

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	0 1.36	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
	MUSCLE ES1							
1		-13	-22	--	--	-11	-14	-14
2		-9	-9	-19	-17	-4	-4	-4
3		-6	-9	-11	--	-5	-5	-4
4		-13	-34	-29	-10	-13	-12	-12
5		-10	--	--	--	-5	-5	-2
6		-6	-13	--	-13	-6	-3	-3
7		-21	-24	-31	--	-21	-24	-20
8		3	--	--	--	2	5	5
9		-12	-22	-24	--	-12	-8	-13
10		-7	--	--	-8	-8	-5	-4
	ES2							
1		44	38	39	33	47	47	49
2		39	45	35	42	47	47	49
3		47	45	45	44	48	49	50
4		36	35	34	--	37	41	4
5		44	41	46	--	49	47	47
6		44	39	42	35	48	47	49
7		40	38	41	--	41	45	42
8		45	44	38	47	45	45	45
9		45	44	46	46	48	48	50
10		44	42	39	54	44	43	47
	GM							
1		-6	-6	-5	-6	-5	-5	-2
2		-8	-1	-1	-1	-2	-4	-3
3		-13	-15	-12	-6	-13	-11	-12
4		-11	-1	-7	47	-7	-6	-8
5		-6	-2	-3	--	-2	-1	--
6		-5	-10	-8	-11	-2	0	0
7		-4	-4	-8	13	-2	-3	-3
8		-4	-3	-7	-1	-2	-2	-3
9		-7	-9	-8	-11	-7	-6	-1
10		-7	-6	-8	-2	-4	-3	-4
	VL							
1		--	--	--	--	--	--	--
2		-19	-16	-15	-14	-16	-11	-10
3		--	--	--	-21	--	--	--
4		-18	-17	-12	39	-15	-14	-14
5		-9	-1	3	-2	--	--	--
6		-8	--	--	-14	--	--	--
7		-9	-8	-9	-13	-7	-8	-4
8		-12	--	-4	-14	-10	-10	-9
9		-10	-10	--	-15	-11	-9	8
10		-10	-7	-12	-18	-10	-7	-6

TABLE A11 (continued)

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	0 1.36	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
	MUSCLE MH							
1		75	80	91	94	85	89	91
2		75	81	79	91	81	78	79
3		80	79	95	103	83	82	82
4		67	67	71	37	70	74	71
5		83	79	91	92	87	87	90
6		78	83	84	92	82	82	84
7		81	82	86	100	80	79	83
8		83	87	86	86	82	81	86
9		81	81	85	87	82	82	83
10		83	94	95	110	82	86	89
	TA1							
1		59	55	52	60	59	59	49
2		49	58	55	57	59	59	59
3		56	53	54	52	60	60	63
4		55	44	47	49	49	54	48
5		61	58	54	52	68	62	60
6		55	56	51	45	58	56	58
7		60	62	62	60	63	63	67
8		56	55	52	55	54	52	56
9		63	64	59	50	65	65	64
10		57	55	53	55	59	58	59
	TA2							
1		87	86	--	76	93	91	91
2		--	--	--	--	87	87	88
3		--	--	--	--	--	--	--
4		98	72	73	--	83	78	79
5		87	82	--	--	89	85	85
6		87	--	--	--	91	--	--
7		--	--	--	--	87	87	89
8		--	--	--	--	84	84	--
9		86	--	--	--	90	89	89
10		--	--	--	--	--	--	--
	GA							
1		26	29	2	16	7	7	7
2		11	17	3	14	22	22	20
3		9	9	8	11	14	12	16
4		6	1	0	4	7	9	5
5		11	3	2	0	13	9	7
6		19	3	5	10	28	22	5
7		10	12	11	7	13	10	17
8		9	18	15	7	15	12	7
9		19	24	19	26	29	32	26
10		17	21	19	15	19	19	24

TABLE A12 - Off timing of erector spinae first and second burst (ES1, ES2), gluteus medius (GM), vastus lateralis (VL), medial hamstring (MH), gastrocnemius (GA) and tibialis anterior (TA1, TA2) for each body weight support (BWS) and full weight bearing (FWB) speed.

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	0 1.36	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
	MUSCLE <u>ES1</u>							
1		3	4	--	--	4	5	4
2		1	1	-4	-8	5	6	5
3		9	0	10		9	7	6
4		3	-20	-18	1	1	0	0
5		10	--	--	--	10	8	10
6		8	9	--	3	8	7	12
7		-2	-5	-13	--	-3	-1	-1
8		9	--	--	--	9	8	8
9		11	5	3	--	9	9	7
10		17	--	--	6	13	19	17
	<u>ES2</u>							
1		54	52	94	81	57	59	64
2		48	53	59	57	56	59	57
3		57	56	59	65	62	57	61
4		49	44	48	--	50	51	5
5		57	54	53	--	60	57	57
6		54	57	55	51	56	55	58
7		56	59	59	--	56	58	61
8		58	59	87	65	55	60	60
9		60	55	54	57	60	61	62
10		63	115	107	70	64	70	70
	<u>GM</u>							
1		32	32	29	22	38	44	46
2		5	31	23	21	18	31	29
3		31	32	36	41	38	38	45
4		26	1	16	65	25	25	28
5		19	10	7	--	24	12	--
6		19	27	17	3	29	18	37
7		9	28	26	34	11	32	35
8		15	17	19	18	25	33	35
9		34	30	27	24	40	41	40
10		15	30	27	15	35	32	39
	<u>VL</u>							
1		--	--	--	--	--	--	--
2		13	16	12	11	22	26	27
3		--	--	--	22	--	--	--
4		10	10	27	87	14	14	17
5		15	11	12	10	--	--	--
6		13	--	--	18	--	--	--
7		14	15	5	5	13	11	12
8		14	--	16	29	15	17	30
9		16	9	--	3	19	27	28
10		12	-6	-4	18	-6	11	20

TABLE A12 (continued)

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	0 1.36	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
	MUSCLE MH							
1		100	120	119	119	110	110	112
2		102	121	116	146	114	115	120
3		117	128	126	134	124	130	135
4		100	108	114	73	93	98	97
5		124	122	124	117	125	122	122
6		104	118	127	134	107	107	110
7		100	143	148	156	116	120	114
8		107	122	120	116	108	118	126
9		105	111	121	124	110	110	117
10		152	151	146	165	154	157	112
	TA1							
1		60	79	73	81	80	82	81
2		96	105	102	105	81	82	82
3		114	108	105	108	112	112	113
4		73	63	65	88	68	69	66
5		79	74	73	91	81	79	77
6		78	101	98	104	81	77	100
7		106	102	103	100	80	80	81
8		104	102	100	103	75	78	107
9		76	109	108	104	80	79	79
10		113	113	106	110	112	116	112
	TA2							
1		112	105	--	100	110	114	115
2		--	--	--	--	110	111	105
3		--	--	--	--	--	--	--
4		112	89	88	--	95	106	91
5		104	94	--	--	107	106	105
6		106	--	--	--	113	--	--
7		--	--	--	--	104	103	105
8		--	--	--	--	101	102	--
9		105	--	--	--	117	117	113
10		--	--	--	--	--	--	--
	GA							
1		49	46	44	41	49	50	53
2		41	50	47	50	49	47	47
3		51	50	51	36	55	56	60
4		39	37	35	41	41	42	41
5		51	47	45	33	56	51	50
6		47	49	34	44	49	49	45
7		45	47	49	47	42	45	46
8		48	48	46	48	48	49	48
9		51	52	53	47	54	56	58
10		47	43	47	33	47	49	49

APPENDIX 7.

MEAN EMG MEAN BURST AMPLITUDES FOR EACH SUBJECT FOR EACH MUSCLE.

TABLE A13 Normalized Mean First Burst Erector Spinae Burst Amplitude

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		68.00	.00	.00	102.00	85.00	81.00
2		82.00	147.00	82.00	79.00	100.00	115.00
3		29.00	16.00	.00	74.00	79.00	94.00
4		54.00	53.00	72.00	92.00	92.00	84.00
5		.00	.00	.00	89.00	115.00	87.00
6		81.00	87.00	43.00	89.00	87.00	81.00
7		75.00	70.00	.00	90.00	77.00	75.00
8		.00	.00	.00	79.00	77.00	79.00
9		100.00	86.00	.00	100.00	119.00	129.00
10		.00	.00	35.00	92.00	92.00	67.00

TABLE A14 Normalized Mean Second Burst Erector Spinae Burst Amplitude

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		86.00	53.00	53.00	86.00	74.00	70.00
2		51.00	84.00	49.00	71.00	55.00	47.00
3		52.00	52.00	21.00	62.00	62.00	59.00
4		29.00	66.00	.00	86.00	89.00	71.00
5		53.00	29.00	.00	96.00	106.00	76.00
6		100.00	.00	74.00	91.00	83.00	94.00
7		83.00	67.00	.00	99.00	103.00	75.00
8		60.00	60.00	36.00	75.00	68.00	61.00
9		84.00	88.00	36.00	84.00	76.00	84.00
10		97.00	133.00	54.00	97.00	90.00	69.00

TABLE A15 Normalized Mean Gluteus Medius Burst Amplitude

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		78.00	36.00	21.00	115.00	104.00	108.00
2		29.00	45.00	49.00	69.00	67.00	71.00
3		66.00	57.00	43.00	84.00	84.00	99.00
4		47.00	40.00	15.00	114.00	109.00	122.00
5		86.00	56.00	48.00	70.00	94.00	49.00
6		73.00	60.00	32.00	100.00	123.00	88.00
7		83.00	104.00	18.00	95.00	104.00	114.00
8		59.00	52.00	3.00	103.00	105.00	120.00
9		74.00	77.00	68.00	87.00	90.00	90.00
10		55.00	50.00	21.00	59.00	68.00	54.00

TABLE A16 Normalized Mean Vastus Lateralis Burst Amplitude

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
*							
2		51.00	35.00	35.00	78.00	59.00	60.00
4		66.00	45.00	43.00	88.00	80.00	61.00
5		.00	9.00	33.00	14.00	18.00	.00
6		.00	.00	29.00	.00	.00	13.00
7		61.00	52.00	13.00	66.00	66.00	17.00
8		.00	12.00	.00	78.00	66.00	38.00
9		17.00	.00	9.00	79.00	83.00	62.00
10		89.00	71.00	107.00	25.00	21.00	64.00

* 1 & 3 no VL burst

TABLE A17 Normalized Mean Medial Hamstring Burst Amplitude

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		62.00	65.00	81.00	103.00	73.00	57.00
2		101.00	123.00	99.00	89.00	70.00	57.00
3		94.00	96.00	72.00	91.00	85.00	85.00
4		59.00	69.00	56.00	90.00	86.00	78.00
5		122.00	100.00	44.00	103.00	103.00	88.00
6		93.00	79.00	43.00	92.00	88.00	67.00
7		76.00	78.00	72.00	50.00	56.00	49.00
8		56.00	68.00	11.00	68.00	56.00	52.00
9		61.00	40.00	46.00	76.00	65.00	52.00
10		121.00	126.00	37.00	92.00	74.00	48.00

TABLE A18 Normalized Mean Tibialis Anterior Burst Amplitude

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		123.00	173.00	175.00	88.00	100.00	118.00
2		87.00	73.00	65.00	65.00	79.00	57.00
3		74.00	95.00	104.00	74.00	69.00	86.00
4		74.00	72.00	78.00	75.00	65.00	65.00
5		92.00	139.00	84.00	87.00	92.00	98.00
6		133.00	142.00	206.00	81.00	75.00	89.00
7		93.00	74.00	92.00	48.00	48.00	43.00
8		107.00	93.00	131.00	70.00	76.00	88.00
9		92.00	83.00	87.00	87.00	69.00	66.00
10		95.00	96.00	100.00	68.00	71.00	71.00

TABLE A19 Normalized Mean Gastrocnemius Burst Amplitude

SUBJECT NUMBER	BWS% Speed $m.s^{-1}$	30 .97	50 .85	70 .70	FWB .97	FWB .85	FWB .70
1		92.00	76.00	51.00	93.00	95.00	100.00
2		94.00	79.00	48.00	107.00	112.00	100.00
3		59.00	40.00	17.00	94.00	99.00	84.00
4		51.00	43.00	27.00	94.00	85.00	84.00
5		81.00	71.00	46.00	96.00	103.00	89.00
6		76.00	74.00	49.00	92.00	91.00	98.00
7		60.00	60.00	27.00	98.00	89.00	107.00
8		95.00	97.00	67.00	100.00	93.00	89.00
9		90.00	66.00	32.00	95.00	88.00	84.00
10		64.00	64.00	41.00	102.00	83.00	85.00

APPENDIX 8.

F-MAX TEST RESULTS FOR HOMOGENEITY OF VARIANCE

TABLE A20 - F max test results for homogenity of variance.

Parameter	DF	F	p	Critical F
% Stance	9	4.70	-	8.41
TDST	8	3.49	-	9.78
Cadence	9	6.44	-	8.41
Stride	9	5.66	-	8.41
Cycle time	9	38.5	<.05	8.41
Total mean hip	9	5.35	-	8.41
Total mean knee	9	2.43	-	8.41
Max swing angle hip	9	6.25	-	8.41
Max swing angle knee	9	3.69	-	8.41
Normalized Mean Burst Amplitudes				
ES_1	9	32.07	<.05	7.8
ES_2	9	8.5	<.05	7.8
GM	9	2.19	-	7.8
VL	7	3.65	-	10.8
MH	9	2.85	-	7.8
TA	9	7.9	<.05	7.8
GA	9	15.4	<.05	7.8

APPENDIX 9.

I-D ANOVA AND POST-HOC TESTS.

TABLE A21 - Summary of Freidman's analysis of variance by ranks for cycle time means from 10 subjects at each body weight support (BWS) and full weight bearing (FWB) speed.

BWS%	0	30	50	70	FWB	FWB	FWB
Speed $m.s^{-1}$	1.36	.97	.85	.70	.97	.85	.70
rank sum	10	23	45	66	30	45	61

$x^2 = 26.6$ df = 6
p <.001   n = 10

TABLE A21 (continued) - Results of selected individual comparisons using the Wilcoxon signed rank test for mean cycle time.

Group	subgroup	Z	p
*BWS 0%	BWS 30	2.8	.005
	" 50	2.8	
	" 70	2.8	
	++FWB .97	2.8	
	" .85	2.8	
	" .70	2.8	
FWB .97	BWS 30	-.97	.33
	" 50	2.2	.02
	" 70	2.8	
	FWB .85	2.7	.007
	FWB .70	2.8	
BWS 30%	BWS 50	2.8	
	" 70	2.8	
	" .85	2.8	
	" .70	2.8	
FWB .85	BWS 50%	1.07	.28
	BWS 70%	2.65	.008
	FWB .70	2.8	.005
FWB .70	BWS 70%	1.9	.04

*BWS(body weight support)
++FWS(full weight bearing speed $m.s^{-1}$)

TABLE A22 - Summary of repeated measures ANOVA among mean differences in % stance from 10 subjects at each body weight support (BWS) and full weight bearing (FWB) speed.

Source	df	sum of squares	mean squares	F	p
speed by weight	6	1181.35	196.89	32.56+	.001
subjects	9	242.38	26.93	4.45	
error	54	326.53	6.05		
total	69	1750.27			

$+F_{.01}(df6,54) = 3.15$

TABLE A22 (continued) - Results of the Scheffe multiple comparison test for % stance

Group	subgroup	F	p	$F_{.01}$ Critical
+BWS 0%	BWS 30	7.78	-	
	" 50	15.4	-	19.14
	" 70	55.4	.01	
	++FWB .97		-	
	" .85		-	
	" .70	10.71	-	
BWS 30%	BWS 50	1.2	-	
	" 70	21.6	.01	
	FWB .97	24.0	.01	
	" .85	33	.01	
	" .70	37	.01	
BWS 50%	BWS 70	12.5	-	
	FWB .97	36.3	.01	
	" .85	46	.01	
	" .70	51.9	.01	
BWS 70%	FWB .97		.01	
	" .85		.01	
	" .70		.01	
FWB .97	FWB .85	.58	-	
	" .70		-	
FWB .85	FWB .70		-	

+BWS(body weight support)

++FWB(full weight bearing speed $m.s^{01}$)

\- not significant

TABLE A23 - Summary of repeated measures ANOVA for total double support time from 9 subjects at each body weight support (BWS) and full weight bearing (FWB) speed.

Source	df	sum of squares	mean squares	F	p
speed by weight	6	3391.71	565.29	42.8+	.001
subjects	8	358.41	44.8		
error	48	633.14	13.19		
total	62	4383.27			

$+F_{.01}(df6,48) = 3.04$

TABLE A23 (continued) - Results of the Scheffe multiple comparisons test for total double support time.

Group	subgroup	F	p	$F_{.01}$ Critical
+BWS 0%	BWS 30	7.99	-	18.24
	" 50	26	.001	
	" 70	66	-	
	++FWB .97	12.8	-	
	" .85	13.8	-	
	" .70	20.6	-	
BWS 30%	BWS 50	.51	-	
	" 70	28	.001	
	FWB .97	41		
	" .85	42.8		
	" .70	42.8		
BWS 50%	BWS 70	9.15	-	
	FWB .97	75.3	.001	
	" .85	75.3		
	" .70	75.3		
BWS 70%	FWB .97	140		
	" .85	140		
	" .70	140		
FWB .97	FWB .85	-	-	
	" .70	-	-	
FWB .85	FWB .70	-	-	

+BWS(body weight support)
++FWB(full weight bearing speed $m.s^{-1}$)
- not significant

TABLE A24 - Summary of repeated measures ANOVA among mean differences in cadence from 10 subjects at each body weight support (BWS) and full weight bearing (FWB) speed.

Source	df	sum of squares	mean squares	F	p
speed by weight	6	9277.37	1546.23	78.91	.001+
subjects	9	2169.71	241.08	12.3	
error	54	1058.12	19.59		
total	69	12505.19			

$+F_{.01}(df6,54) = 3.15$

TABLE A24 (continued) - Results of the Scheffe multiple comparisons test for cadence.

Group	subgroup	F	p	$F_{.01}$ Critical
+BWS 0%	BWS 30	43.3	.01	18.9
	" 50	43.3		
	" 70	43.3		
	++FWB .97	63.2	.01	
	" .85	63.2		
	" .70	63.2		
BWS 30%	BWS 50	27.9	.01	
	" 70	156.1	.01	
	FWB .97	1.8	-	
	" .85	20.1	.01	
	" .70	20.1		
BWS 50%	BWS 70	52	.01	
	FWB .97	15.3	-	
	" .85	.8	-	
	" .70	21.6	.01	
BWS 70%	FWB .97	21.6		
	" .85	21.6		
	" .70	6.5	-	
FWB .97	FWB .85	9.7	-	+BWS(body weight support)
	" .70	73.3	.01	++FWB(full weight bearing speed $m.s^{-1}$)
				- not significant
FWB .85	FWB .70	29	.01	

TABLE A25 - Summary of repeated measures ANOVA among mean differences in right stride length from 10 subjects at each body weight support (BWS) and full weight bearing (FWB) speed.

Source	df	sum of squares	mean squares	F	p
speed by weight	6	.78	.13	27.9	.001+
subjects	9	.30	.03	7.18	
error	54	.25	.01		
total	69	1.34			

+$F_{.01}$(df6,54) = 3.15

TABLE A25 (continued) - Results of the Scheffe multiple comparison test for right stride length.

Group	subgroup	F	p	$F_{.01}$ Critical
+BWS 0%	BWS 30		.01	18.9
	" 50		.01	
	" 70		.01	
	++FWB .97	24.2	.01	
	" .85		.01	
	" .70		.01	
BWS 30%	BWS 50		-	
	" 70		-	
	FWB .97		-	
	" .85		-	
	" .70		-	
BWS 50%	BWS 70		-	
	FWB .97		-	
	" .85		-	
	" .70	4.05	-	
BWS 70%	FWB .97		-	
	" .85		-	
	" .70	6.05	-	
FWB .97	FWB .85		-	
	" .70	11.15	-	
FWB .85	FWB .70		-	

+BWS(body weight support)

++FWB(full weight bearing speed $m.s^{-1}$)

- not significant

APPENDIX 10.

ANGULAR DISPLACEMENT ANOVA AND POST-HOC TESTS.

TABLE A26 - Summary of repeated measures ANOVA among mean differences in total hip angular displacement.

Source	df	sum of squares	mean squares	F	p
speed by weight	6	2587.80	431.3	28.5	.001+
subjects	9	636.29	70.73	4.68	
error	54	815.91	15.1		
total	69	4040.00			

$^{+}F_{.01}(df6,54) = 3.15$

TABLE A26 (continued) - Results of the Scheffe multiple comparison test for hip angular displacement.

Group	subgroup	F	p	$F_{.01}$ Critical
+BWS 0%	BWS 30	58.96	.001	18.9
	" 50	69	.001	
	" 70	69	.001	
	++FWB .97	12.8	-	
	" .85	12.8	-	
	" .70	18.7	-	
BWS 30%	BWS 50	.96	-	
	" 70	8.6	-	
	FWB .97	33.3	.001	
	" .85	28.3	-	
	" .70	12.0	-	
BWS 50%	BWS 70	5.3	-	
	FWB .97	36	.001	
	" .85	36	.001	
	" .70	16.8	-	
BWS 70%	FWB .97	52	.001	
	" .85	69	.001	
	" .70	41	.001	
FWB .97	FWB .85	1.08	-	
	" .70	.75	-	
FWB .85	FWB .70	3.63	-	

+BWS(body weight support)
++FWB(full weight bearing speed $m.s^{-1}$)
- not significant

TABLE A27 - Summary of repeated measures ANOVA among mean differences in total knee angular displacement.

Source	df	sum of squares	mean squares	F	p
speed by weight	6	2119.29	353.21	14.58	.001+
subjects	9	802.86	89.21	3.68	
error	54	1308.14	24.22		
total	69	4230.29			

$+F_{.01}(df6,54) = 3.15$

TABLE A27 (continued) - Results of the Scheffe multiple comparison test for total knee angular displacement.

Group	subgroup	F	p	$F_{.01}$ Critical
+BWS 0%	BWS 30	6.07	-	18.9
	" 50	20.4	.01	
	" 70	61.6	.001	
	++FWB .97	10.8	-	
	" .85	10.8	-	
	" .70	10.8	-	
BWS 30%	BWS 50	4.26	-	
	" 70	.6	-	
	FWB .97	2.4	-	
	" .85	2.8	-	
	" .70	29	.01	
BWS 50%	BWS 70	11.1	-	
	FWB .97	13	-	
	" .85	13.9	-	
	" .70	1.5	-	
BWS 70%	FWB .97	48.1	.001	
	" .85	49.9	.001	
	" .70	20.8	.01	
FWB .97	FWB .85	5.6	-	
	" .70	5.6	-	
FWB .85	FWB .70	5.6	-	

+BWS(body weight support)

++FWB(full weight bearing speed $m.s^{-1}$)

- not significant

TABLE A28 - Summary of repeated measures ANOVA among mean differences in maximum flexor swing angle of the hip.

Source	df	sum of squares	mean squares	F	p
speed by weight	6	963.37	160.56	16.95	.001+
subjects	9	992.91	110.32	11.65	
error	54	511.49	9.47		
total	69	2467.77			

$+F_{.01}(df6,54) = 3.15$

TABLE A28 (continued) - Results of the Scheffe multiple comparison test for maximum flexor swing angle of the hip.

Group	subgroup	F	p	$F_{.01}$ Critical
+BWS 0%	BWS 30	19.68	.01	18.9
	" 50	28.97	.01	
	" 70	49.78	.001	
	++FWB .97	2.1	-	
	" .85	2.1	-	
	" .70	2.1	-	
BWS 30%	BWS 50	1.03	-	
	" 70	6.85	-	
	FWB .97	8.46	-	
	" .85	16.0	-	
	" .70	5.83	-	
BWS 50%	BWS 70	2.9	-	
	FWB .97	8.89	-	
	" .85	24.4	.01	
	" .70	6.11	-	
BWS 70%	FWB .97	31.3	.001	
	" .85	43.8	.001	
	" .70	25.9	.001	
FWB .97	FWB .85	1.8	-	
	" .70	1.08	-	
FWB .85	FWB .70	1.08	-	

+BWS(body weight support)

++FWB(full weight bearing speed $m.s^{-1}$)

- not significant

TABLE A29 - Summary of repeated measures ANOVA among mean differences in maximum flexor swing angle of the knee.

Source	df	sum of squares	mean squares	F	p
speed by weight	6	2339.57	389.93	24.66	.001+
subjects	9	1369.94	152.22	9.63	
error	54	853.86	15.81		
total	69	4563.37			

+$F_{.01}$(df6,54) = 3.51

TABLE A29 (continued) - Results of the Scheffe multiple comparison test for maximum flexor swing angle of the knee.

Group	subgroup	F	p	$F_{.01}$ Critical
+BWS 0%	BWS 30%	17.32	-	18.9
	" 50	40.6	.001	
	" 70	86.1	.001	
	++FWB .97	.71	-	
	" .85	1.97	-	
	" .70	12.16	-	
BWS 30%	BWS 50	4.6	-	
	" 70	26.2	.01	
	FWB .97	11.0	-	
	" .85	7.5	-	
	" .70	.45	-	
BWS 50%	BWS 70	8.8	-	
	FWB .97	29.8	.01	
	" .85	24.06	.01	
	" .70	7.79	-	
BWS 70%	FWB .97	71.2	.001	
	" .85	62	.001	
	" .70	34.2	.001	
FWB .97	FWB .85	.3	-	
	" .70	6.9	-	
FWB .85	FWB .70	6.9	-	

+BWS(body weight support)

++FWB(full weight bearing speed $m.s^{-1}$)

- not significant

APPENDIX 11.

EMG MEAN BURST AMPLITUDE ANOVA AND POST-HOC TESTS.

TABLE A30 - Summary of Freidman's analysis of variance by ranks for erector spinae first burst normalized mean burst amplitude.

Sample	N	Rank Sum	x^2	df	p
+BWS 30%	10	25.5	21.89	5	.01
" 50%	10	24.5			
" 70%	10	18			
++FWB .97 $m.s^{-1}$	10	47			
" .85	10	48.5			
" .70	10	44.5			

++FWB (full weight bearing)
+BWS (body weight support)

TABLE A30 (continued) - Results of selected individual comparisons using the Wilcoxon signed rank test for erector spinae first burst normalized mean burst amplitude.

Group	subgroup	Z	p
+BWS 30%	BWS 50	-.85	.4
	" 70	-1.69	.08
	++FWB .97	-2.55	.01
	" .85	-2.80	.005
	" .70	-2.52	.01
BWS 50%	BWS 70	-1.52	.124
	FWB .97	-2.19	.02
	" .85	-2.19	.02
	" .70	-2.19	.02
BWS 70%	FWB .97	-2.70	.007
	" .85	-2.80	.005
	" .70	-2.80	.005
FWB .97	FWB .85	.84	.4
	" .70	-.06	.49
FWB .85	FWB .70	-.51	.6

+BWS(body weight support)
++FWB(full weight bearing speed $m.s^{-1}$)

TABLE A31 - Summary of Freidman's analysis of variance by ranks for erector spinae second burst normalized mean burst amplitude.

Sample	N	Rank Sum	x^2	df	p
+BWS 30%	10	37	23.37	5	.001
" 50%	10	32.5			
" 70%	10	12.5			
++FWB .97 $m.s^{-1}$	10	49.5			
" .85	10	44.5			
" .70	10	34			

++FWB (full weight bearing)
+BWS (body weight support)

TABLE A31 (continued) - Results of selected individual comparisons using the Wilcoxon signed rank test for erector spinae second burst normalized mean burst amplitude.

Group	subgroup	Z	p
+BWS 30%	BWS 50	.07	.05
	" 70	2.80	.005
	++FWB .97	2.20	.02
	" .85	.92	.3
	" .70	.18	.6
BWS 50%	BWS 70	1.72	.08
	FWB .97	1.58	.1
	" .85	1.07	.2
	" .70	.97	.33
BWS 70%	FWB .97	2.80	.005
	" .85	2.80	.005
	" .70	2.70	.007
FWB .97	FWB .85	1.48	.1
	" .70	2.49	.01
FWB .85	FWB .70	1.73	.07

+BWS(body weight support)
++FWB(full weight bearing speed $m.s^{-1}$)

TABLE A32 - Summary of repeated measures ANOVA among mean differences in gluteus medius normalized mean burst amplitude.

Source	df	sum of squares	mean squares	F	p
speed by weight	5	30789.5	6157.91	17.68	.001+
subjects	9	6719.73	746.64	2.14	
error	45	15674.47	348.32		
total	59	53183.73			

$+F_{.01}(df5,45) = 3.44$

TABLE A32 (continued) - Results of the Scheffe multiple comparison test for gluteus medius normalized mean burst amplitude.

Group	subgroup	F	p	$F_{.01}$ Critical
BWS 30%	BWS 50	.76	-	17.2
	" 70	15.8	-	
	FWB .97	8.69	-	
	" .85	12.7	-	
	" .70	10.08	-	
BWS 50%	BWS 70	9.6	-	
	FWB .97	14.6	-	
	" .85	19.7	.01	
	" .70	16.4	-	
BWS 70%	FWB .97	47.79	.001	
	" .85	56.7	.001	
	" .70	51.2	.001	
FWB .97	FWB .85	17.2		
	" .70	17.2		
FWB .85	FWB .70	17.2	-	

+BWS(body weight support)
++FWB(full weight bearing speed $m.s^{-1}$)
- not significant

TABLE A33 - Summary of repeated measures ANOVA among mean differences in vastus lateralis normalized mean burst amplitude.

Source	df	sum of squares	mean squares	F	p
speed by weight	5	3765.35	753.07	1.18	.33
subjects	7	19249.48	2749.93	4.32	
error	35	22295.15	637		
total	47	45309.98			

TABLE A34 - Summary of repeated measures ANOVA among mean differences in medial hamstring normalized mean burst amplitude.

Source	df	sum of squares	mean squares	F	p
speed by weight	5	7811.48	1562.30	5.06	+.001
subjects	9	10700.02	1188.89	3.85	
error	45	13896.68	308.82		
total	59	32408.15			

$+F_{.01}(df5,45) = 3.44$

TABLE A34 (continued) - Results of the Scheffe multiple comparison test for medial hamstring normalized mean burst amplitude.

Group	subgroup	F	p	$F_{.01}$ Critical
+BWS 30%	BWS 50	13.09	-	17.2
	" 70	13.09	-	
	FWB .97	1.28	-	
	" .85	1.28	-	
	" .70	7.29	-	
BWS 50%	BWS 70	13.1	-	
	FWB .97	13.1	-	
	" .85	1.25	-	
	" .70	1.25	-	
BWS 70%	FWB .97	14.08	-	
	" .85	13.09	-	
	" .70	.84	-	
FWB .97	FWB .85	7.92	-	
	" .70	7.92	-	
FWB .85	FWB .70	7.22	-	

+BWS(body weight support)
++FWB(full weight bearing speed $m.s^{-1}$)
- not significant

TABLE A35 - Summary of Freidman's analysis of variance by ranks for tibialis anterior normalized mean burst amplitude.

Sample	N	Rank Sum	x^2	df	p
+BWS 30%	10	45	21.27	5	.01
" 50%	10	44			
" 70%	10	49			
++FWB .97 $m.s^{-1}$	10	24			
" .85	10	23			
" .70	10	25			

TABLE A35 (continued) - Results of selected individual comparisons using the Wilcoxon signed rank test for tibialis anterior normalized mean burst amplitude.

Group	subgroup	Z	p
+BWS 30%	BWS 50	.41	.66
	" 70	1.22	.219
	++FWB .97	2.55	.01
	" .85	2.67	.007
	" .70	2.19	.026
BWS 50%	BWS 70	1.38	.16
	FWB .97	2.5	.01
	" .85	2.7	.01
	" .70	2.8	.005
BWS 70%	FWB .97	2.4	.016
	" .85	2.4	.016
	" .70	2.5	.012
FWB .97	FWB .85	.06	.4
	" .70	.82	.42
FWB .85	FWB .70	.98	.3

+BWS(body weight support)

++FWB(full weight bearing speed $m.s^{-1}$)

TABLE A36 - Summary of Freidman's analysis of variance by ranks for gastrocnemius normalized mean burst amplitude.

Sample	N	Rank Sum	x^2	df	p
+BWS 30%	10	32	23.2	5	.001
" 50%	10	24			
" 70%	10	10			
++FWB .97 $m.s^{-1}$	10	53			
" .85	10	47			
" .70	10	38			

TABLE A36 (continued) - Results of selected individual comparisons using the Wilcoxon signed rank test for gastrocnemius normalized mean burst amplitude.

Group	subgroup	Z	p
+BWS 30%	BWS 50	2.31	.019
	" 70	2.80	.005
	++FWB .97	2.80	.005
	" .85	2.50	.012
	" .70	2.40	.016
BWS 50%	BWS 70	2.80	.005
	FWB .97	2.80	.005
	" .85	2.70	.007
	" .70	2.70	.007
BWS 70%	FWB .97	2.80	.005
	" .85	2.80	.005
	" .70	2.80	.005
FWB .97	FWB .85	1.27	.2
	" .70	1.89	.056
FWB .85	FWB .70	.46	.6

+BWS(body weight support)
++FWB(full weight bearing speed $m.s^{-1}$)