INFLUENCE OF BODY WEIGHT SUPPORT ON SOLEUS H-REFLEX MODULATION IN NORMAL AND SPINAL CORD INJURED HUMAN SUBJECTS DURING STANDING AND WALKING

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ABSTRACT

The soleus H-reflex modulation pattern was measured in 8 normal and 8 spastic paretic subjects during standing and walking. under the conditions of 0% and 40% body weight support (BWS). In standing, both the normal and patient groups showed no significant difference ($p<0.01$) in the H-reflex amplitude between 0% and 40% BWS. Normal subjects had a phase dependent modulation of the H-reflex during gait, there being no significant difference ($p<0.001$) in this modulation with 40% BWS. The patients had an abnormally elevated H-reflex throughout the step cycle, although five showed some modulation. In the patient group, 40% BWS produced a decrease of the H-reflex amplitude mainly in the push-off phase. BWS produced a decrease in electromyographic (EMG) mean burst amplitude of the lower limb muscles investigated, with more appropriate EMG activity timing. BWS improved knee and ankle angular displacements in patients, which were associated with an improved locomotor pattern. However, these improvements in locomotor pattern were not reflected clearly by changes in the H-reflex modulation. Thus the relationship between H-reflex amplitude, EMG activity, and ankle position under 0% and 40% BWS, needs to be further investigated.

RESUME

Le patron de modulation du réflexe-H du muscle soléaire a été évalué chez 8 sujets normaux et 8 patients parésiques spastiques, en position debout et durant la marche, lorsque 0% et 40% du poids du corps était supporté (PCS). En position debout, les sujets normaux et les patients démontrent aucune différence significative ($p<0.01$) dans l'amplitude du réflexe-H entre 0% et 40% PCS. Durant la marche, les sujets normaux montrent une modulation phasique du réflexe-H à 0% PCS, qui n'est pas significativement différente ($p<0.001$) à 40% PCS. Les patients montrent un réflexe-H anormalement élevé durant tout le cycle de marche, alors que cinq d'entre-eux présentent une certaine modulation. Toutefois, à 40% PCS, ces derniers diffèrent par une amplitude du réflexe-H moindre qu'à poids total, surtout dans la phase d'appui terminale. La condition 40% PCS produit une diminution de l'amplitude électromyographique (EMG) moyenne des muscles du membre inférieur examiné, avec un profil d'activation plus approprié en relation avec le cycle de marche. Une amélioration du déplacement angulaire au genou et à la cheville, chez ce groupe de patients, est également observée. Néanmoins, ces changements du patron locomoteur ne sont pas intimement associés aux changements dans la modulation du réflexe-H. En conclusion, la relation entre l'amplitude du réflexe-H, l'activité EMG, et la position de la cheville, respectivement à 0% et 40% PCS doit faire l'objet d'une recherche plus approfondie.

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STATEMENT OF AUTHORSHIP

I certify that I am the primary author of the manuscript contained in this thesis. I do claim full responsibility for the content and style of the text included herein.

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CHAPTER 1: INTRODUCTION

Injury to the spinal cord, resulting in varying degrees of paralysis, is one of the most traumatic conditions a person can face. The prevalence of complete or partial paralysis caused by accidents in the United States is nearly 50 per 100,000 population, based on an annual incident rate of 3 per 100,000 population (Yashon, 1986). Men, havirg a mean age of 30 years, are five times more likely to be among those injured than are women (Jaeger et al, 1989). Half the injuries are due to motor vehicle accidents, one quarter are due to falls, and one in ten is a consequence of recreational activities, most commonly diving (Yashon, 1986). The estimated 50% survival rate for incomplete spinal cord injured (SCI) patients was calculated to occur in the twentieth year after the trauma (Kurtzke, 1975). However, as a result of much improved emergency care and comprehensive rehabilitation programs, SCI patients are living longer today.

One of the most important impairments, experienced by patients with a spinal cord injury, is spasticity, which is characterized by: hyperreflexia (increase in tendon jerk responses and velocity dependent stretch reflexes); hypertonia (increased resistance to passive rapid stretch); clonus (a series of repetitive muscle contractions initiated by a rapid and maintained stretch); and impairment of voluntary control (Ashby et al, 1987; Chapman & Wiesendanger, 1982; Young & Wiegner, 1987). Spasticity, along with the associated paresis/weakness, lack of dexterity, fatigability, and inappropriate and/or impaired ability to produce muscle contractions, cause most of the functional disabilities experienced by SCI patients (Katz & Rymer, 1989; Landau, 1980; Young & Wiegner, 1987). These deficits are most pronounced during movement and must be studied in greater detail during walking, as a common result of spinal cord injury is impairment of lower extremity function which disturbs the normal locomotor pattern. Therefore, the achievement of standing and walking by SCI subjects is one important goal of therapy, for it allows them to enjoy a higher level of functioning and provides a certain amount of psychological benefits. In addition, standing and walking confer several therapeutic benefits with respect to lower extremity contractures, osteoporosis, circulation, spasticity, and renal function

(Jaeger et al, 1989).

After the initial period of spinal shock due to trauma to the spinal cord, segmental reflexes associated with the muscle spindle (ie. the tendon jerk and H-reflex) slowly become hyperactive (Ashby et al, 1974; Little & Halar, 1985). The H-reflex, elicited most commonly in the soleus muscle, has been used as a means of indirectly assessing the excitability of the motoneuron pool associated with the predominantly monosynaptic stretch reflex, both at rest and during locomotion. In the normal subject it appears that the tonic and phasic modulation of the H-reflex may be under the control of different mechanisms; thus after injury to the spinal cord, the modulation may be affected differently (Capaday & Stein, 1986). It is known that the H-reflex in the normal individual is modulated to a great extent during the gait cycle (Capaday & Stein, 1986; Crenna & Frigo, 1987). This modulation appears to be functionally important for achieving smooth locomotion (Capaday & Stein, 1987b). In SCI patients the normal modulation of segmental reflexes is impaired, and the reason for this abnormal modulation has not been completely determined. Yang et al (1989) and Fung et al (1990) reported that, in the few SCI patients they studied, there was a high tonic level of reflex gain which seemed to mask the normal modulation pattern of the H-reflex elicited in the soleus muscle during gait.

Functionally, in SCI patients, the hyperactive spinal stretch reflexes cause altered responses to muscle stretch during the gait cycle (Barbeau et al, 1988; Knutsson, 1986). Spastic paretic gait is also characterized by impaired muscle activation patterns and difficulty in coping with weight bearing, speed and balance (Barbeau et al, 1988; Conrad et al, 1985; Knutsson, 1986). In addition to these deficits, there are other neurological changes that result from the interruption of inputs from supraspinal centers which lead to a large variation in the degree of gait disturbances (Knutsson, 1985).

Efforts to improve the locomotor pattern of SCI patients has led to the development of an interactive gait retraining strategy that involves the use of an overhead harness to support a percentage of the patient's body weight during locomotion, with a gradual progression to full weight bearing (Barbeau et al, 1987).

This technique reduces the loading response during stance, which may decrease the amount of stretch imposed upon the muscle, especially the triceps surae muscle group, which in many spastic patients exhibits a hyperactive stretch reflex during gait. The use of this body weight support (BWS) system has already been shown to be effective in improving the locomotor pattern of spastic paretic patients as measured by electromyographic and kinematic data (Visintin & Barbeau, 1989). However, the underlying neurophysiological mechanisms which lead to the expression of a more normal gait pattern, as a result of providing BWS during gait training, have not yet been elucidated. It has already been documented that : (i) SCI patients exhibit a hyperactive H-reflex, (ii) the modulation of the H-reflex in spastic patients has been found to be abnormal, in the few patients that have been studied, and (iii) the recruitment pattern of the triceps surae EMG activity is modified in SCI patients by providing BWS during locomotion. Therefore it is the purpose of this study, firstly to determine the soleus H-reflex amplitude modulation during standing and walking at full weight bearing in a group of SCI patients, and secondly to determine whether the effects that BWS has on the soleus EMG, both during standing and walking, is mediated by an alteration of the excitability of the soleus motoneuron pool, as measured by the H-reflex, in both a group of normal subjects and a group of SCI patients. This would provide insight into the functional role BWS plays at the spinal level in enabling SCI patients to exhibit a more normal gait pattern while walking on the treadmill. In addition, this information would strengthen the rationale for using progressive BWS during treadmill or overground walking as a gait retraining strategy for SCI patients.

1.1 OBJECTIVES

MAIN OBJECTIVE

The main objective of this research project is to determine the modulation pattern of the excitability of the soleus motoneuron pool, as measured by the H-reflex in the soleus muscle, in normal and spinal cord injured human subjects, during standing and walking, under the conditions of 0% and 40% body weight support.

SPECIFIC OBJECTIVES

1. To evaluate the change in soleus H-reflex amplitude while standing at full weight and when 40% of body weight is symmetrically removed.

2. To evaluate the change in soleus H-reflex amplitude during eight phases of the gait cycle while walking on a treadmill at full weight and with 40% of body weight removed.

3. To compare the modulation pattern of the H-reflex under each condition, in normal and spinal cord injured subjects.

CHAPTER 2: LITERATURE REVIEW

2.1 DESCRIPTION OF THE HOFFMANN REFLEX

Reflexes are nonvolitional responses elicited by particular types of sensory stimuli. One of the most well known reflexes is the stretch reflex, or tendon jerk reflex, which is induced by tendon percussion of extensor muscles primarily. The muscle spindles, when stretched, send impulses through the primary Ia afferent fibers, which synapse directly with the alpha motoneurons innervating the homonymous muscle, to produce a brisk contraction (Carew, 1985).

The Hoffmann reflex (H-reflex) was once considered as the electrical equivalent of the stretch reflex. It was suggested to be a monosynaptic reflex induced by electrical stimulation of Ia spindle afferent fibers, at intensities below the thresholds of motor fibers (Gottlieb & Agarwal, 1971; Hoffmann, 1918; Magladery & McDougal, 1950). Current review, however, suggested that in man the electrically induced afferent volley stimulates not only Ia afferent fibers but may also stimulate: group I muscle afferents from small muscles of the foot (Burke et al, 1983); cutaneous afferents from the skin of the heel and sole (Burke et al, 1983); and group Ib afferents from Golgi tendon organs (Pierrot-Deseilligny et al, 1981). These oligosynaptic inputs may have sufficient time to reach the motoneuron pool being excited by the Ia afferent input, during the rising phases of the excitatory postsynaptic potentials, thus influencing the amplitude of the reflex activity (Burke et al, 1984; Burke, 1985; Schieppati, 1987). This leads to the conclusion that the H-reflex is mono-oligosynaptic, rather than simply monosynaptic (Burke et al, 1984; Diamantopoulos & Gassel, 1965). Nevertheless, the H-reflex, in comparison to the stretch reflex, is less dependent on peripheral factors, as it bypasses the muscle spindle, and thus is believed to be a better means of assessing changes in synaptic efficacy between the muscle afferents and the alpha motoneurons, and of providing information concerning the excitability level of the alpha motoneurons (Capaday & Stein, 1986; Crenna & Frigo, 1987).

2.2 METHODOLOGICAL CONSIDERATIONS

2.2.1 Classical Methodology

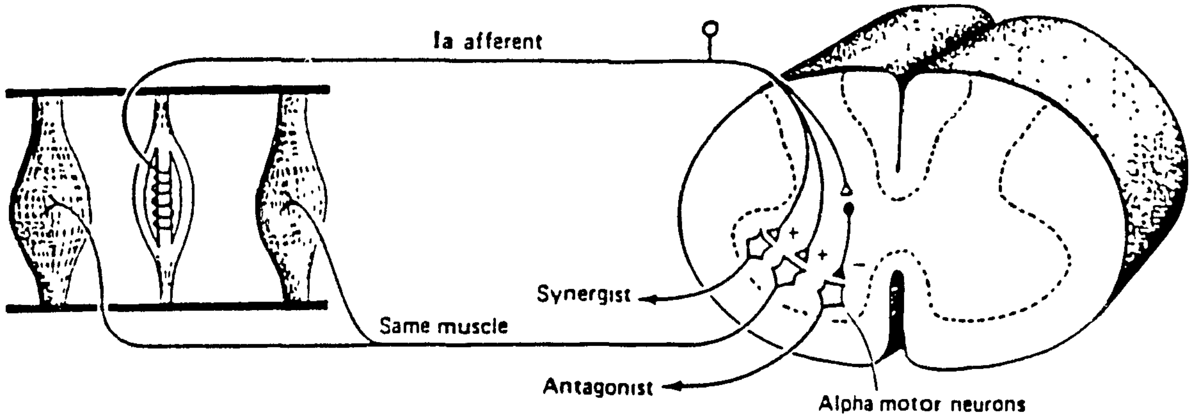
The H-reflex is most easily elicited in man when the posterior tibial nerve is electrically stimulated in the popliteal fossa, while recording the surface electromyogram (EMG) of the soleus muscle. This was shown first by Piper (1912), and then clearly described by Hoffmann (1918), in honour of whom Magladery and McDougal (1950) named the reflex which they explored systematically. The responses in the soleus muscle consist of two bursts of EMG activity when moderately intense electrical stimulation is applied to the posterior tibial nerve, with a duration of 1 ms and a minimum interstimulus delay of 3-5 sec (Hugon, 1973). The first response is identified as the M-wave and consists of a short latency (5-10 ms) activation of the motor units of the soleus muscle, due to direct stimulation of the alpha motoneuron axons. The second response is identified as the H-reflex, or H-wave, and consists of a longer latency (30-35 ms) reflex response, due to stimulation of Ia muscle spindle afferents (Magladery & McDougal, 1950; Schieppati 1987), (see figure 2.1).

The H-M recruitment curve relates the change in amplitude of the M- and H-response as a function of stimulation intensity. Occlusion between antidromic and orthodromic impulses in the same alpha motoneuron axons is believed to be the reason that the H-reflex disappears while the M-wave increases in size, as the stimulation intensity increases (Gottlieb et al, 1970).

In previous experiments the following protocol has been used to elicit and record the H-reflex in the lower limb during a dynamic activity, such as walking. The stimulating electrode was fastened to the skin more securely, and in such a way so as to avoid restricting normal movement of the knee joint. During walking, changes in stimulus effectiveness, due to movement of the stimulating electrode with respect to the tibial nerve, could not be avoided. Therefore, the stimulus intensity was chosen such that it elicited a near-maximal H-reflex and a minimal M-wave, with the variability of the M-response kept low, so that the effectiveness of the stimulus strength could be kept constant (Capaday & Stein, 1986; Crenna & Frigo, 1987).

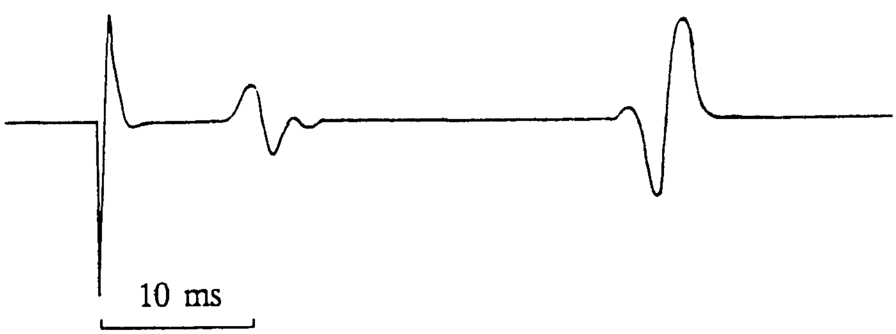
Crayton and King (1981) studied the variability of the H-reflex in normal

Figure 2.1. A diagram of the basic circuitry of the monosynaptic stretch reflex, above, and an example of the electrical response in the soleus muscle due to electrical stimulation of the tibial nerve, below.



The basic segmental circuitry for the stretch reflex.

- from Kandel & Schwartz, 1985.



Soleus M-wave and H-reflex elicited due to moderately intense electrical stimulation of the tibial nerve.

subjects and found that H-reflex parameters, (Hmax, Mmax, H-latency), were highly stable within subjects at different testing times, and that they were not correlated with certain demographic or personal habit variables such as age, sex, race, and exercise. However, there was inter-individual variability within the group of normal subjects, especially in the H/M ratio. Eke-Okoro (1982) reported that alcohol, caffeine, and fatigue, due to sustained isometric contractions, significantly potentiated the H-wave as measured in the soleus muscle.

2.2.2 Muscle Under Study: Soleus

In the lower limb, for the most part, H-reflexes are elicited in muscles that are physiological extensors: calf muscles, quadriceps, and part of the hamstrings (Deschuytere et al, 1983). In the thigh, the H-reflex has been easily recorded in the quadriceps muscle group, specifically in the vastus medialis (Mongia, 1972), and the rectus femoris (Guiheneuc & Ginet, 1974). In the lower leg, the muscle typically investigated is the soleus. Generally, the H-reflex cannot be evoked in the pretibial muscles in normal human subjects at rest, although it has been recorded from the tibialis anterior (TA) muscle of a few subjects (Tanaka, 1974). In addition, an H-reflex has been elicited from the extensor digitorum longus muscle in humans (Deschuytere & Rosselle, 1971), when supraspinal inhibition is counteracted by facilitation (via a voluntary contraction) and a sufficient number of Ia fibers in the mixed peripheral nerve are stimulated (Deschuytere et al, 1983).

In this study the soleus muscle has been chosen because: (i) it has a simple mechanical arrangement (monoarticular), (ii) it is functionally homogeneous (slow-twitch motor units), (iii) it plays an important role in locomotion (Stein & Capaday, 1988), (iv) it is one of the muscles noticeably impaired in spastic gait (Dietz & Berger, 1983; Conrad et al, 1985), and (v) during locomotion, its EMG recruitment pattern and amplitude change under the body weight support condition (Visintin and Barbeau, 1989).

2.2.3 The Development of the Body Weight Support System

The use of body weight support (BWS) in the gait training of spinal cord injured

(SCI) subjects was initiated as a result of research using the spinal animal model. It has been determined that, following a complete transection of the low thoracic spinal cord, recovery of locomotion in kittens 1-2 weeks of age can take place without any type of training (Grillner, 1973; Forssberg et al, 1980; Smith et al, 1982). However, in cats spinalized at maturity it was initially found that, although there was return of rhythmical stepping movements in their hindlimbs, their gait on the treadmill was uncoordinated and they were unable to achieve hindquarter support, up to eight weeks post transection (Eidelberg et al, 1980). Recently, the importance of training in accelerating the recovery of quality locomotion, as well as re-establishing weight bearing in the adult spinal cat, has been studied (Barbeau & Rossignol, 1987; Rossignol et al, 1986; Smith et al, 1982). Barbeau and Rossignol (1987) reported that interactive locomotor training, which included progressive increase in weight support by the hindquarters, was found to be very important in accelerating the recovery of the locomotor pattern in the adult spinal cat following spinal cord transection (T_{13}). The degree of weight supported by the hindquarters was determined by the experimenter, who held the tail during the training. These findings encouraged researchers and clinicians to develop a better strategy of locomotor therapy for SCI patients. The gait training approach that developed, predominantly as a result of the spinal animal model, was a dynamic and task specific approach that involved progressive weight bearing during treadmill walking (Finch, 1986; Barbeau et al, 1987). This system involved the use of a body weight support harness which was similar to a mountaineering harness with a waist belt, two padded straps that criss-crossed underneath the pelvis, and shoulder straps connected to an overhead motor driven pulley system. The complete setup has been described in detail (Barbeau et al, 1987). This system is able to alleviate the difficulty neurologically impaired patients have in coping with weight bearing during ambulation, as the harness can support a percentage of the patient's body weight and allow for unassisted or assisted stepping movements. This permits training to be started more quickly post injury, and enables the three components of gait: weight bearing, balance, and stepping, to be retrained simultaneously (Visintin & Barbeau, 1989). The potential of this novel gait training strategy to improve the locomotor pattern of SCI patients is presently under investigation.

2.3 MODULATION OF THE SOLEUS H-REFLEX DURING STANDING AND WALKING

2.3.1 Modulation of the H-Reflex in Normal Subjects

Chan and Kearney (1982) studied the effect of static tilt, in the sagittal plane, upon soleus H-reflex modulation in standing man. They fou ' that the H-reflex amplitude was minimal when the subject was near vertical, but was increased as a result of either forward or backward tilt. It was postulated that the significant changes in soleus motoneuronal activity were due to the effect of tilting on the otolith receptors in the inner ear, which are responsible for the tonic labyrinthine reflexes. The reflex response would be functionally relevant in a situation, for example, when a forward sway would be counteracted by an increased soleus contraction to restore the body to equilibrium. However, Aiello et al (1983) found opposite results in normal suojects, leading to controversy concerning the influence of static tilt on soleus motoneuronal excitability, which may be attributable to differences in methodologies used (Chan & Kearney, 1984).

Capaday and Stein (1986) looked at H-reflex modulation in humans while standing and maintaining a tonic contraction of the soleus muscle at various levels. In this standing paradigm, these researchers found that the amplitude of the H-reflex response was larger when the voluntary EMG activity of the soleus was greater. In addition, the H-reflex response was larger when compared to the H-reflex amplitude while walking, at the same soleus EMG amplitude, the difference being greatest during low level activity. Both the reflex sensitivity and the reflex threshold were lower during standing than during walking, indicating a task specific reflex response which was modulated independently of the level of motor activity (Stein & Capaday, 1988).

In order to explain this reflex modulation, Capaday and Stein (1987a) developed a model of the motoneuron pool and simulated the reflex output of this pool in response to a Ia afferent input. These authors showed that the H-reflex is a linearly increasing function of the background EMG for a fixed stimulus strength and reflects the effectiveness of the synaptic transmission from the Ia afferents to the alpha motoneurons. This has been called 'automatic gain compensation' which means that the gain of the reflex increases with the excitation level of the motoneuron pool (Matthews, 1986). Capaday and Stein (1987a), using their model, showed that presynaptic inhibition is a possible mechanism for changing the central gain of the monosynaptic reflex.

Locomotion is an important functional motor task in which the soleus muscle plays a significant role. The role of the soleus is to control the degree of ankle flexion from foot flat to heel off, and to propel the body forward and upward at the end of stance (Stein & Capaday, 1988; Sutherland, 1966). During walking, the soleus shortens after heel contact until the foot is flat on the ground, and then lengthens during the phase of single limb support as the foot dorsiflexes, until the heel comes off the ground. During this lengthening phase, the soleus stretch reflex elicited may raise the excitability of the soleus motoneuron pool and assist in increasing the degree of contraction in this muscle. This is followed by a shortening phase between heel-off and toe-off, which pushes the body forward and lifts the limb in swing. During swing, the soleus is again lengthened as the foot is dorsiflexed, in order to clear the ground in preparation for the next heel contact. However, in this phase any stretch reflex elicited would be undesirable as it would counteract the dorsiflexion. Thus a modulation of the soleus motoneuron pool has an important functional role in the control of locomotion (Capaday & Stein, 1986).

The excitability of the triceps surae motoneuron pool during human locomotion has been tested by several researchers using the H-reflex technique. Garrett et al (1984) found that the gastrocnemius H-reflex amplitude was markedly reduced during the swing phase and reached a peak level of excitability at midstance. Capaday and Stein (1986), Crenna and Frigo (1987), and Fung et al (1990) found similar modulations of the H-reflex in the soleus muscles of their normal subjects during gait. The modulation of the H-reflex amplitude was found to vary approximately in proportion to the voluntary EMG amplitude of the soleus muscle.

There are several neural mechanisms that may play a role in the modulation of the H-reflex during the walking cycle. During early stance, the reduction in motoneuron pool excitation could be due to presynaptic gating of Ia afferent inputs, as hypothesized by Capaday and Stein (1987b). Morin et al (1982) supported this view and postulated that the inhibition was caused by the mixed composition of the electrically stimulated volley, which would activate fast Ib afferent impulses, thus gating the autogenetic excitation from the triceps surae, as it is stretched at heel contact. In addition, at heel strike, the increased TA activity initiated to lower the foot gently to the ground might inhibit the soleus muscle disynaptically by means of a Ia inhibitory interneuron pathway through reciprocal inhibition. That this occurs during concentric contractions has been verified experimentally by several researchers (Iles & Roberts, 1987; Kagamihara & Tanaka, 1985; Morin & Pierrot-Deseilligny 1977; Shindo et al, 1984). Shindo, et al (1984) demonstrated that dorsiflexion at the ankle facilitated the Ia inhibitory pathway from ankle flexors to ankle extensors, measured via the soleus H-reflex amplitude. Conversely, plantarflexion depressed this reciprocal inhibition in parallel with the amount of tonic voluntary contraction of the soleus.

During later stance, the soleus motoneurons fire at their maximum level and the H-reflex is found to be maximally facilitated, which could be explained both by the increased firing of the muscle spindles and by motoneuron depolarization caused by the central command (Crenna & Frigo, 1987). However, the oligosynaptic inputs may also contribute to the reinforcement of the ankle extensor activity, especially as the load on the limb is greater than body weight at the push-off phase, resulting in greater pressure on cutaneous receptors and increased stretch of intrinsic foot muscles. This hypothesis is supported by the fact that facilitation of the H-reflex during the stance phase in normal gait, in which body weight is supported by the limb, is accompanied by greater activation of the soleus motoneurons as compared to the stepping task, in Crenna and Frigo's (1987) study, in which no body weight support was provided by the limb.

Reciprocal inhibition may again play a role in the consistent inhibition of the H-reflex throughout swing as the TA contracts to dorsiflex the foot. The soleus motor units are being de-recruited at the end of stance and there may be a release-associated inhibition that would contribute to the gating of the soleus autogenetic excitation, as has been observed in static conditions (Crenna & Frigo, 1987; Schieppati, 1987). The H-reflex inhibition during swing may also be caused by descending facilitation of Renshaw cell activity exerted on soleus motoneurons, which is facilitated during the TA contraction. This serves to counteract the Ia effects of stretch of the soleus during the TA contraction and of the reciprocal Ia inhibitory pathway to the TA (Katz & Pierrot-Deseilligny, 1984; Pierrot-Deseilligny et al, 1977). There may also be a decrease in the excitability of the soleus motoneuron pool as a result of the unloading of the limb during swing. Crenna and Frigo (1987) reported that even the simple suspension of the flexed lower limb against gravity, in the absence of changes in EMG activity, produced a consistent H-reflex inhibition within the soleus muscle.

Functionally, the soleus H-reflex modulation during locomotion appears to facilitate extension at the ankle during stance and to inhibit unwanted plantarflexion during the swing phase (Garrett et al, 1984). At the end of stance, a large reflex is appropriate to assist in propelling the body forward (Stein and Capaday, 1988). Once the swing phase begins, the EMG activity of the soleus is minimal, and a substantial stretch reflex would inappropriately counteract the desired dorsiflexion of the foot, thus inhibition of the H-reflex is expected (Capaday & Stein, 1986).

2.3.2 Modulation of the H-Reflex in SCI Patients

Studies which have examined the H-reflex amplitude elicited in spastic lower limbs of patients with clinically complete and incomplete SCI, have revealed that in the prone position the H-reflex is elevated in amplitude compared to normal subjects (Angel & Hofmann, 1963; Ashby et al, 1974; Little & Halar, 1985). Taylor et al (1984) reported that the maximum H/M ratio, (the largest proportion of the motoneuron pool that is recruited via the reflex), was not significantly different from normal. However, when a moderately intense electrical stimulus (ie. below the threshold of motoneuron axons) was used, a greater proportion of the triceps surae motoneuron pool was excited in patients with spasticity (64%) than in normal subjects (43%, $p<0.01$), (Taylor et al, 1984).

In addition, it has been reported that recovery from reflex depression of spinal motoneuron excitability, due to conditioning afferent volleys of differing intensities, commenced much earlier in patients with upper motoneuron lesions, as compared to normal subjects (Magladery et al, 1952; Taylor et al, 1984). These authors postulated that this earlier recovery may result from removal of some depression of spinal motoneuron excitability, mediated through internuncial neuronal mechanisms, or from increased excitability of motoneurons.

Yang et al (1989) and Fung et al (1990) examined the soleus H-reflex modulation in a few patients with traumatic spinal or cerebral lesions while walking on a treadmill. They reported that the H-reflex amplitude was increased in both standing and walking, as compared to normal subjects, with no evident modulation during the gait cycle, when a stimulus intensity that elicited a maximal H-reflex was used. These authors postulated that the lack of reflex modulation was due to one of two mechanisms: (i) impairment of the normal modulating mechanisms, or (ii) presence of the normal modulating influence which, however, was masked by a high reflex gain that saturated the excitability of the motoneuron pool. The second hypothesis seems to be supported by the fact that in two patients there was a more normal H-reflex modulation during gait when a lower stimulus intensity was utilized. However, neither mechanism is mutually exclusive; thus further studies must be conducted to determine what mechanism is responsible for this abnormal reflex response in neurologically impaired individuals.

2.4 INFLUENCE OF BODY WEIGHT SUPPORT DURING LOCOMOTION

2.4.1 Influence of BWS in Normal Subjects

The effect of various levels (0%, 30%, & 50%) of body weight support (BWS) on the EMG, kinematic, and temporal distance data of seven normal humans walking on a treadmill, with their body weight supported by a modified parachute-type harness, has been investigated (Finch, 1986). It was found that as the level of body weight supported by the harness increased, the comfortable walking speed progressively decreased. With speed held constant, the differences measured when the subjects walked with BWS, as compared to full weight bearing (FWB), were a decrease in the percentage of the gait cycle spent in stance, a decrease in total double limb support time leading to an increase in single limb support time, and a decrease in maximum hip and knee angular displacement. In addition, knee flexion at the foot flat phase during stance was reduced because there was less body weight to support and because of harness constraints.

The changes in EMG activity in the lower limb, under the BWS condition, were a decrease in mean burst amplitude for the gastrocnemius (GA) muscle and an increase for the tibialis anterior (TA) muscle. For all muscles investigated in the lower leg, the on/off timing of the muscle bursts in relation to the gait cycle was similar with increasing levels of BWS (Finch, 1986). Thus, the walking pattern remained normal even as BWS was progressively increased. This is a very important finding, as it confirms that the use of progressive weight bearing can be used as a gait retraining strategy, to assist in overcoming the difficulty SCI patients encounter in coping with weight bearing during the stance phase of the gait cycle.

2.4.2 Spastic Gait and Influence of BWS in SCI Patients

Spastic gait is often characterized by knee and ankle joint flexion during foot-floor contact or stance. In some spastic patients during stance, the knee joint progressively extends and the foot is kept more in dorsiflexion, never reaching a neutral position of 90 degrees, thus keeping the triceps surae stretched (Benecke & Conrad, 1986). In other patients the foot is maintained in plantarflexion during swing, such that the ball of the foot, instead of the heel, makes initial contact with the ground (Dietz et al, 1981). As a result of spasticity these patients show an inability to increase locomotor speed (Dietz & Berger, 1983). The EMG activity of the lower limb muscles, such as GA and soleus, is prolonged and the amplitude profile is flattened, compared to normal subjects (Conrad et al, 1985). The authors attributed these changes in the EMG profiles to exaggerated stretch reflexes, as most of the activity in the muscles coincided with periods of muscle lengthening during the gait cycle. Benecke and Conrad (1986) further postulated that there may be a defective generation of centrally originated activity and a compensatory shift towards an increase in peripheral influences, in spastic patients.

Spastic paretic subjects have difficulty coping with loading of the lower limbs

(Barbeau et al, 1988; Carlsoo et al, 1974; Dickstein et al, 1984). Barbeau et al (1988) reported that, in severely spastic patients walking at full weight, there were enhanced stretch reflexes in the triceps surae and sustained clonus during the stance phase. However, when the subjects walked at 40% BWS, the decreased load on the lower extremities resulted in a more appropriate triceps surae activity and a decrease in clonus. This was attributed to the straighter trunk, hip, and knee angles which allowed the ankle to be in a neutral position in midstance, thereby decreasing the stretch of the triceps surae (Barbeau et al, 1988).

Visintin and Barbeau (1989), in their study of seven SCI patients, found that when 40% of the spastic paretic subjects' body weight was supported during gait, as compared to walking at full weight, there was a general decrease in EMG mean burst amplitude for all the major lower limb muscles with more appropriate timing of EMG activity. There were straighter trunk and knee alignment during the weight support phase and an increase in single limb support time, stride length, and speed.

2.4.3 Influence of BWS on Soleus H-Reflex Modulation

The effect that BWS may have on H-reflex modulation, in the lower limb during gait, has not been directly addressed. One study has made a preliminary attempt to investigate this question. Crenna and Frigo (1987) elicited the H-reflex in the soleus muscle of subjects who stepped on one spot with the ipsilateral leg (50 strides/min) while the contralateral, nonperforming leg supported the body weight throughout the entire step cycle. In this movement, foot-ground contact was made with the ball of the foot; the heel never made contact with the ground at any time. They found that in early stance, the soleus muscle exhibited a peak EMG activity at foot contact. During midstance (15-30% of step cycle) the H-reflex was significantly reduced in amplitude, which coincided with soleus EMG activity of low amplitude. The late stance phase (30-55% of step cycle) involved a facilitation of the H-reflex, concommitant with a consistent EMG soleus muscle burst, which served to push the leg up off the treadmill. Throughout the swing phase there was a marked inhibition of the H-reflex which then recovered to reference level near the end of swing. This non-weight bearing stepping movement is different from walking, but it does represent a rhythmical activity with alternating flexion and extension. Thus, the modulation of the H-reflex during this stepping movement serves to confirm results found in normal walking and to reveal a pattern of H-reflex suppression in a non-weight bearing condition.

2.4.4 Functional Implication of Providing BWS During Locomotion

Providing BWS during treadmill walking, to SCI patients suffering from spastic paresis, appears to alleviate some of the problems of early stretch and abnormal muscle activation, produced because of the difficulty in coping with loading under the full weight bearing condition. One of the goals of this gait retraining strategy is to allow the patient to bear only the amount of weight that he is capable of supporting, such that the locomotor pattern can be facilitated with the least deformities. The conventional therapeutic approach to gait retraining does not provide this assistance, as the patient must walk under full weight using either parallel bars or walking aids to support the load on the lower extremities. However, with BWS, deviations in the patient's gait can be instantly corrected and peripheral stimulation given to assist muscle activation during stance and swing. The amount of BWS provided can be progressively decreased through different stages of gait retraining, ultimately enabling the patient to walk at full weight with the most optimal locomotor pattern he can achieve.

CHAPTER 3: METHODOLOGY

3.1 POPULATION

Eight normal subjects and eight SCI patients participated voluntarily in this research study (table 3.1 and 3.2). Selection of the patients was based on the following inclusion/exclusion criteria: (i) presence of an incomplete spinal cord lesion resulting in clinical symptoms of spasticity, including ankle clonus, hyperreflexia, and spasms, (ii) ability to ambulate overground, with or without the use of aids or braces, for at least 30 steps at a minimum speed of 0.2-0.25 m/s, (iii) absence of lower extremity pathology, (iv) absence of cardiovascular irregularities, and (v) age in the range of 18-60 years. The normal subjects were selected to participate if they were free from neurological disorders affecting the central nervous system, chronic musculo-skeletal disorders of the lower extremities, and cardiovascular irregularities.

3.2 RESEARCH DESIGN

Figure 3.1 provides a schematic overview of the research design. A repeated measures design was used, with a comparison made within subjects and between groups for the effects of altering the degree of body weight supported by the subject on the soleus H-reflex during the two conditions, standing and walking.

The independent variables in the standing paradigm were:
(1) the groups: one normal group and one patient group, with 8 subjects in each.
(2) the percentage of BWS removed: 0% BWS and 40% BWS.
The dependent variable was the soleus H-reflex amplitude. Soleus M-wave amplitude and EMG activity in the GA and TA, were recorded as control variables, to ensure that any modulation in the H-reflex amplitude was not due to significant changes in these variables.

The independent variables in the walking paradigm were:
(1) the groups: one normal group and one patient group, with 8 subjects in each.

Table 3.1. Demographic details of the normal subjects.

Demographic Data

Subject	Age	Sex	Comfortable treadmill speed FW (m/s)	Cycle Duration (s/cycle)
ES	24	M	1.09	1.15
JF	30	F	1.07	1.24
HB	36	M	0.74	1.32
MB	30	F	0.93	1.34
KN	25	F	1.00	1.24
JB	26	M	1.05	1.30
AP	29	M	1.05	1.29
NP	25	F	0.75	1.45

Table 3.2. Demographic details of the patients.

Demographic Data

Patient	Age	Sex	Diagnosis (years)	Chronicity	Ankle clonus	Comfortable treadmill speed FW (m/s)	Cycle Duration (s/cycle)	Overground walking aids used
CH	54	M	T_8 SCI	2.0	S	0.16	3.30	Walker *, SLB (R)
LR	24	F	T_{10} SCI	3.5	U	0.15	2.70	1 cane, SLB (R)
RM	35	M	C_6 SCI	19.5	U	0.26	2.84	Canadian crutches
AN	22	M	SP	10.0	U	0.30	2.05	Independent
RG	43	M	C_{3-4} SCI	0.2	U	0.25	3.04	Walker
MB	59	M	SP	10.0	U	0.32	2.02	1 cane
BP	34	M	SP	5.0	U	0.50	1.70	Independent, SLB (bilateral)
SH	29	M	C_{7-8} SCI	4.5	S	0.16	6.53	Walker *, SLB (bilateral)

SP : spastic paralysis of non-familial origin

S = sustained
U = unsustained

* could manage 5-10 steps overground even with effort
SLB : short leg brace

Figure 3.1. Schematic representation of the research design.

A. Standing paradigm.

B. Walking Paradigm.

RESEARCH DESIGN

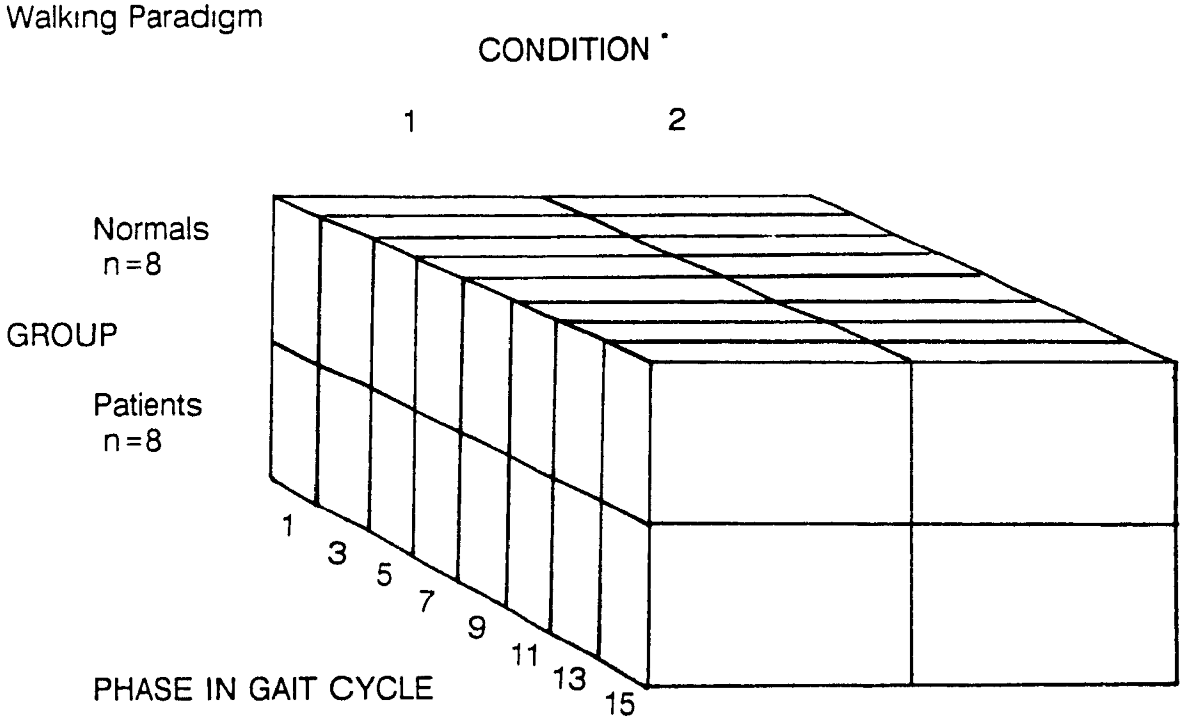
Standing Paradigm

		CONDITION	
		0% BWS	40% BWS
GROUP	Normals n=8	y	x
	Patients n=8	y	x

y = Soleus H-reflex amplitude

x = % change in soleus H-reflex amplitude compared to (y).

Walking Paradigm



* Condition (1) 0% BWS at comfortable cadence
(2) 40% BWS at same cadence at 0% BWS

(2) the experimental trials:

(a) 0% BWS at comfortable cadence.

(b) 40% BWS at the same cadence as at 0% BWS.

(3) the phase of the gait cycle: 8 phases normalized to the gait cycle.

The dependent variable used to measure the H-reflex modulation was the soleus H-reflex amplitude. Several control variables were also recorded in order to determine whether any modulation in the H-reflex amplitude was due to significant changes in these variables:

A) The soleus M-wave amplitude.

B) Soleus EMG amplitude, averaged in ten unperturbed cycles.

C) EMG amplitude of GA and TA muscles.

D) Kinematics: angular displacement at the trunk, hip, knee, and ankle of the tested side throughout two gait cycles.

E) Temporal distance parameters: cycle duration, percentage of stance and swing, and percentage of total double support time.

3.3 ORIENTATION SESSION

Each subject visited the lab for one orientation session during which the details of the study were explained, both verbally and in writing, and an informed consent form was signed. During this session an evoked ankle clonus test was conducted with the patient seated, both hip and knee angles in 90 degrees of flexion. The patient's ankle was briskly dorsiflexed and held until the clonus had ceased. Clonus was recorded as either sustained or unsustained. Also during this session, the subject's comfortable walking speed was determined, during habituation to treadmill walking, and the subject became familiar with the electrical stimulation as the presence of an acceptable H-reflex was determined.

3.4 EXPERIMENTAL SESSION

The experimental session was divided into two parts: (i) standing and (ii) walking, and in each trial the H-reflex was evoked by electrical stimulation of the posterior tibial nerve within the popliteal fossa of one leg.

3.4.1 Part I: Standing

After the electrodes were placed on the subject, he/she was strapped into the BWS harness. The modulation of the H-reflex was recorded during two quiet standing conditions. In the first condition, after appropriate calibration of the BWS system, the subject stood on the ground at 0% BWS, with weight equally distributed on each leg, and an H-M recruitment curve was recorded by increasing the intensity of stimulation, delivered in the popliteal fossa, by increments of 2-5 volts. This was followed by one set of ten stimulations at a stimulus intensity that evoked a near-maximal H-reflex and small M-wave (± 15% variation), which was determined from the recruitment curve. This enabled the acquisition of a stable baseline measure of soleus motoneuron excitability, so that changes in H-reflex amplitude in subsequent trials could be expressed as a percentage of the baseline measure. In the second condition, the subject stood with 40% of his body weight supported by the harness. Again an H-M recruitment curve was recorded, followed by one set of ten stimulations at an intensity that evoked a near-maximal H-reflex and minimal M-wave (± 15% variation), as determined from the recruitment curve. In four of the normal subjects ten stimuli were delivered at random inter-stimulus intervals of 10-15 seconds as the subject progressively contracted his soleus muscle against manual resistance, from 0% to maximal isometric contraction. A two channel oscilloscope was used to provide visual feedback to the subject. In one channel was a reference line which was set by the experimenter at the level of EMG activity desired. The other channel contained the soleus EMG activity which was low-pass filtered (3 Hz), linear enveloped, and displayed at 0.2 ms/div, such that it was a straight line. The subject had to superimpose the line representing the EMG activity in his soleus muscle over the reference line for a few seconds when the command was given. Care was taken to stabilize the ankle so that it was maintained at 90 degrees during the contraction.

3.4.2 Part II: Locomotion

Before each of the two experimental trials, the subject spent several minutes becoming habituated to walking on the treadmill. The subject then walked on the treadmill under two conditions: (i) 0% BWS at comfortable cadence and (ii) 40% BWS at the same cadence as at 0% BWS. In order to keep the cycle duration consistent across the two trials, a metronome was set at the appropriate frequency during the 0% BWS trial and the treadmill speed was adjusted during the 40% BWS trial, so that the subject's cadence matched that of the 0% BWS trial. During the experimental trials the soleus H-reflex was electrically evoked at the intensity that elicited a near-maximal H-reflex as determined during the 0% BWS standing condition, during eight phases of the walking cycle, once every two to four cycles, and ten times in each phase. At the end of the experimental session, one set of ten stimulations was administered to the subject while standing at full weight, at the same intensity as that used in the standing 0% BWS condition, in order to verify the stability of both the reflex response and the experimental conditions.

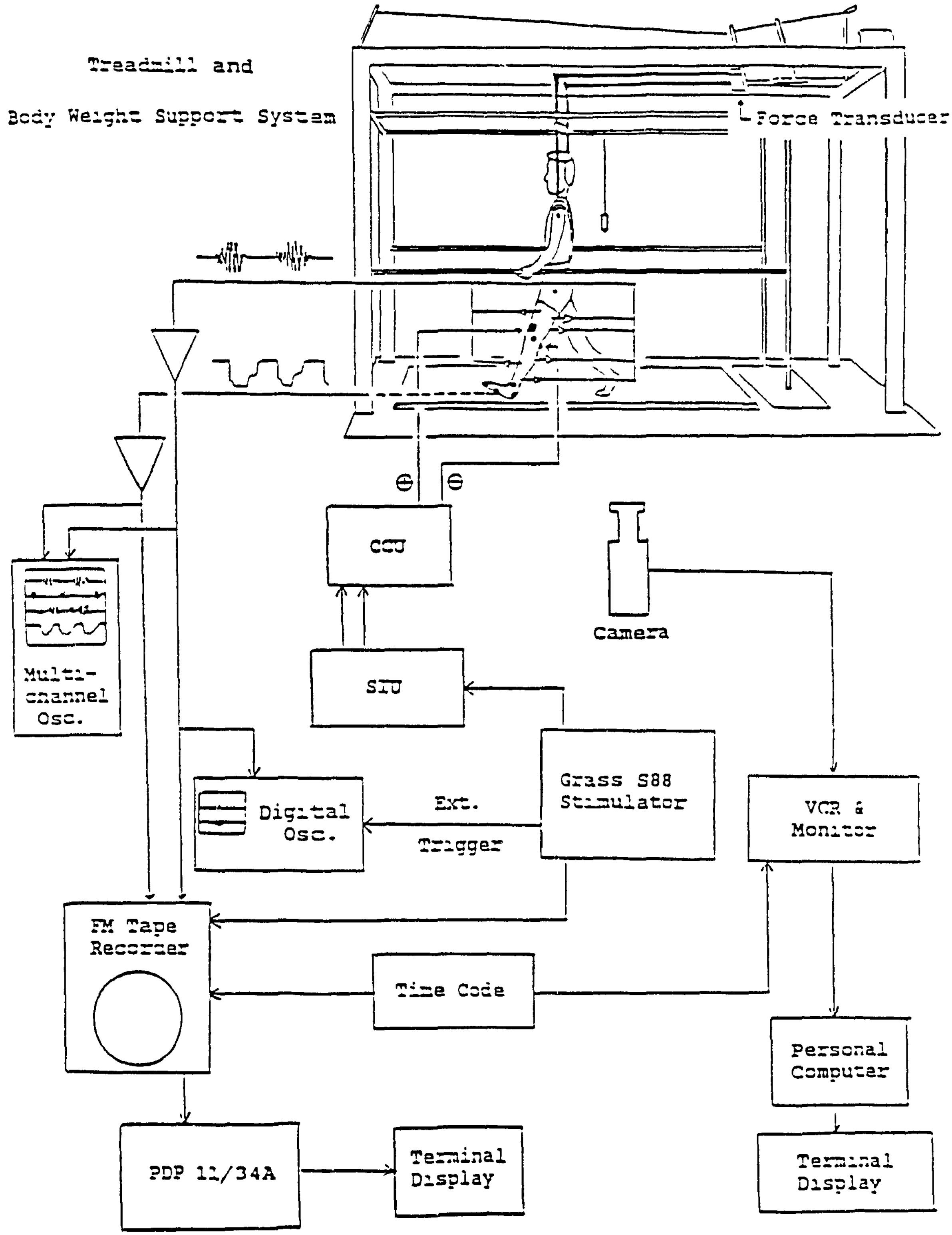
3.5 DATA RECORDING AND INSTRUMENTATION

The data necessary in this research project was collected in the Human Gait Laboratory at the School of Physical and Occupational Therapy, McGill University. The experimental configuration is found in figure 3.2.

3.5.1 Assessment of Treadmill Locomotion

Each subject walked on a motorized treadmill (W.E. Collins #101) while wearing the modified overhead harness (Kanuk Co.) which was used to provide balance and body weight support during locomotion on the treadmill. This BWS system, as shown in figure 3.3, has been described in detail elsewhere (Barbeau et al, 1987). However, the harness itself was modified prior to this study, such that padded thigh rings replaced the straps that had previously criss-crossed beneath the pelvis, thus providing more comfort and stability. The weight support system was calibrated to the subject's weight at the beginning of each trial. The percentage of BWS provided was calculated using a force transducer. While the subject was sitting, the harness was lowered so that there was no tension on it and the weight support was set to 0%. The subject was then lifted up so that his feet left the treadmill surface and the weight support calibration was set to 100%. Then the subject was lowered until the

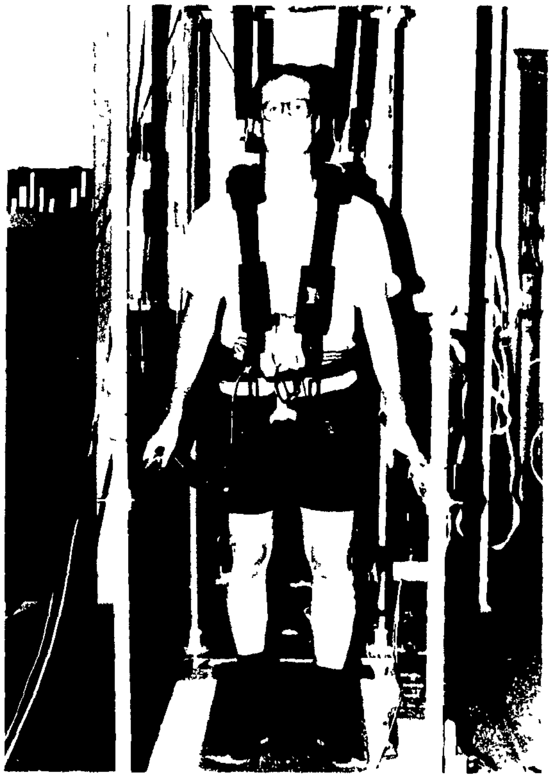
Figure 3.2. The experimental configuration for data acquisition and processing.



Treadmill and
Body Weight Support System
Force Transducer
CCU
Camera
SIU
Multi-
channel
Osc.
Digital
Osc.
Ext.
Trigger
Grass S88
Stimulator
VCR &
Monitor
FM Tape
Recorder
Time Code
Personal
Computer
PDP 11/34A
Terminal
Display
Terminal
Display

Figure 3.3. Anterior (A), posterior (B), and lateral (C) views of the body weight support (BWS) system and treadmill. The BWS harness consists of two padded thigh rings attached in three places to a waist belt which is supported by shoulder straps. The shoulder straps are in turn connected to the motor driven pulley system which provides the lift. Transducers are used to quantify the BWS provided; a gauge indicates the ongoing percentage of BWS. In these three pictures the location of the reference electrode (above the patella), some of the EMG recording electrodes, and the reflective joint markers are clearly visible.

A



B

C

3.5.3 Electrical Stimulation

The stimulus electrode (Medi-Trace Pellet Electrode CH307) was placed over the posterior tibial nerve in the popliteal fossa. The optimal location was determined manually by using a stimulating probe. The anode (reference electrode), of dimensions 4.5 cm x 10 cm, was placed on the thigh, proximal to the patella, after adequate skin preparation. A square pulse stimulator (Grass model S88) delivered the monopolar electrical stimulus which varied from 10 to 150 volts, with a stimulus duration of 1 ms and an interstimulus interval of 10 sec. This interstimulus interval was chosen because studies have shown that the H-reflex is progressively reduced in amplitude at rates of stimulation greater than 0.1 Hz (Burke et al, 1989; Ishikawa et al, 1966). The electrical stimulus passed through a stimulation isolation unit (Grass model SIU5B) and a constant current unit (Grass model CCU1A) before reaching the subject. The electrical stimulation was delivered manually in the standing condition. In the walking trials, feedback from the footswitches of the tested leg was used to divide the gait cycle into 16 phases or bins. A gait cycle is defined as the time interval from one heel contact to the next with the same foot, and thus includes a stance phase followed by a swing phase. The average cycle duration was calculated from the footswitch signal and this duration was then divided by 16 to arrive at the average duration of each bin. During the trial, the footswich signal was fed into a delay unit which could set the latency between the step marker (heel strike) and the stimulus marker, such that the stimulus could be delivered in any one of the 16 bins. Stimuli which were delivered in the same bin width of the gait cycle were averaged together (usually 10 stimuli per bin).

In this experiment the H-reflex was elicited randomly in only 8 of the 16 bins, more specifically in bin 1, 3, 5, 7, 8, 9, 11, and 13 for normals and in bins 1, 3, 5, 7, 9, 11, 13, 15 for patients. These bins adequately cover the range of the gait cycle and enabled collection of the required information concerning the modulation of the H-reflex during gait. The intensity of the stimuli administered during each bin was manually controlled such that the amplitude of the M-wave was kept constant (± 15% variation) during each phase, at a magnitude determined during the standing condition. This ensured that the stimulus intensity was kept constant, and therefore that the same percentage of the soleus motoneuron pool was being excited.

The preliminary method for investigating changes in the soleus H-reflex, during 16 bins of each gait cycle, and the computer program created to analyse the data, were devised by Dr. R.B. Stein and Mr. R. Rolf at the University of Alberta.

3.5.4 Footswitches

Pressure sensitive footswitches (Tapeswitch Systems of America) were taped under the heel, fifth metatarsal, and big toe of the subject's shoes to provide temporal distance factors of gait, such as step cycle duration and percentage of stance and swing, which could be temporally linked to muscle activation patterns. Each footswitch had a distinct voltage drop activated by pressure which was displayed visually on the oscilloscope and recorded on the FM tape. These footswitch signals were used to control the timing of the stimulation of the soleus H-reflex during gait and were synchronized with EMG data to correlate muscle activity at different times during gait. Footswitches indicated heel contact, foot-flat, and toe-off, and defined the different components of stance and swing during the gait cycle.

3.5.5 Video Recordings

Angular displacements of the tested side's trunk, hip, knee, and ankle in the sagittal plane were collected by filming the subject during locomotion with a rotary shutter video camera (Sony: RSC-1010) placed central and perpendicular to the treadmill, at a distance of four meters. One 850 watt halogen lamp (Hedler Co. D6251) illuminated the treadmill area. Reflective joint markers (2 cm in diameter circles) were placed at the shoulder, hip, knee, and ankle, as well as the heel, metatarsal, and toe region of the lateral border of the subject's shoe. (Refer to Appendix 3A for placement of the joint markers). Additional markers were placed on the horizontal bar of the treadmill frame to serve as absolute coordinates for video analysis. A video monitor (Panasonic: WV-5470) was used to view the trials which were recorded on 3/4 inch VHS videotapes using a videocassette recorder

(Panasonic VCR: NV-9240) at a speed of 60 fields/sec. This allowed visual verification of the gait pattern of each subject.

3.6 DATA ANALYSIS

3.6.1 Electromyographic and Kinematic Analysis

A PDP 11/34A computer was used in the analog to digital (12 bit A/D) conversion of the soleus raw EMG recording, which was then digitized at 3 KHz for off-line computer analysis, using an Ogivar Technologies System V computer. Signal processing software, developed in our laboratory, was used to analyze the M- and H-waves evoked by the electrical stimulation. Two windows of 6-10 ms were specified to determine the peak to peak and mean amplitude of both the M- and H-waves with predetermined latencies. In addition this program determined the standing H-M recruitment curve and the walking H-reflex modulation curve, in real time.

The PDP 11/34A computer was used to convert the SOL, GA, and TA EMG and footswitch recordings from analog to digital signals, off-line. The EMG signals were full wave rectified and passed through an anti-aliasing filter (low pass cut off frequency at 470 Hz) before being digitized at 1 KHz and displayed on a computer terminal (Transiac TRI 1024), where interactive computer programs were used to analyze the data of ten representative gait cycles for each trial. Placement of arrows on the footswitch signal served to determine the temporal parameters of gait, such as cycle duration and percentage of stance and swing, and thus enabled determination of the EMG activation profiles and peak activity within each stride. The digitized EMG data could be further processed so that the gait cycle was normalized to the average cycle duration, in order to allow for pooling of the data across strides and across subjects.

Angular displacements of the trunk, hip, knee, and ankle in the sagittal plane were analyzed digitally by a computer program developed by Peak Performance Technologies Inc., which captured each of the two fields of each frame separately for semi-automatic analysis. To do so it was necessary to copy the relevant videotape sequences from 3/4 to 1/2 inch tape, which was then played on a

Panasonic VCR (AG 7300), the image being displayed on a Samtron monitor (SC 431V) which, in conjunction with the software program installed in a Virage Tech computer, allowed the kinematic data to be analysed (see Appendix 4A for method of joint angle calculations).

3.6.2 Statistical Analysis

The statistical analysis of the data was carried out using the following analyses. A two-way analysis of variance (ANOVA), using level of weight support (0% and 40% BWS) as the repeated factor and groups as the between subjects factor, was done in order to determine whether there was a significant difference in the amplitude of the M-wave across the different trials.

Paired T-tests of the means of SOL and GA EMG activity: (i) in normals and patients under the two standing conditions, 0% BWS and 40% BWS, and (ii) in normals between the soleus EMG activity in standing at FWB and that in phase 1 of the gait cycle, were done to determine whether there was a significant difference in the EMG activity between these conditions within each group.

A two-way ANOVA, using level of BWS as the repeated factor and groups as the between subjects factor, was done to determine whether there was a significant difference in the H-reflex amplitude between the two standing conditions.

A three-way ANOVA, using level of BWS and phase in the gait cycle (early stance, push-off, and swing) as repeated factors and groups as the between subjects factor, was done to determine whether there was a significant difference in the amplitude of the H-reflex between the two walking conditions.

Finally, four two-way ANOVAs were done, using the same between subjects factor (groups) and the same repeated factors (level of BWS), for each of the following independent variables: cycle duration, percentage of the step cycle spent in stance, percentage of the cycle spent in swing.

CHAPTER 4: RESULTS

4.1 Reliability of the Measure

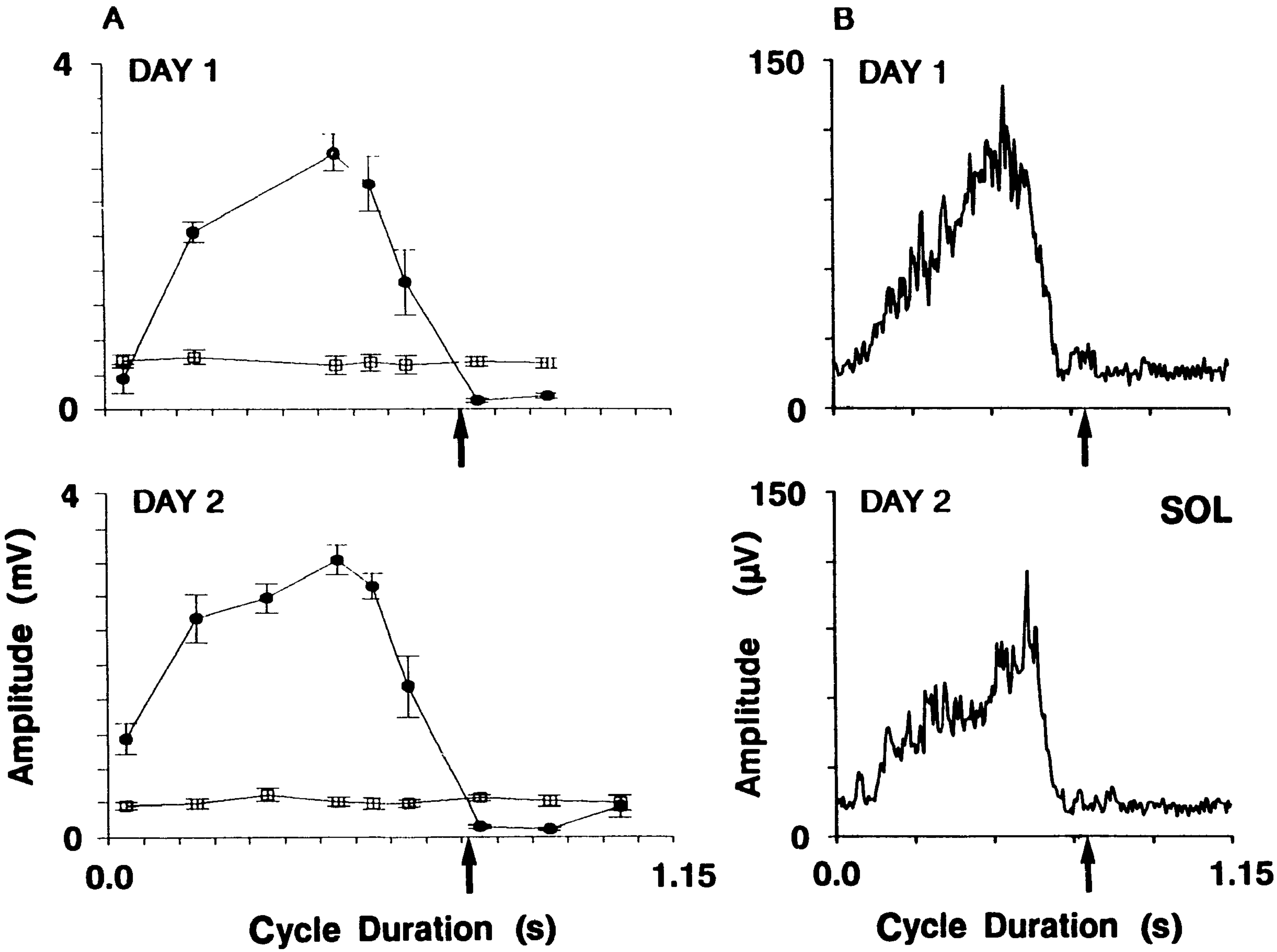
In order to verify the reliability of the H-reflex as a measure of the excitability of the soleus motoneuron pool across time for the same experimental protocol, the data from one representative normal subject, collected on two occasions separated by a time period of two months, was illustrated. The data recorded on the first day, as compared to that recorded on the second day, is depicted in figures 4.1 and 4.2. In figure 4.1A it can be noted that the same range of intensities of stimulation was used on each day to elicit the H- and M-wave recruitment curves, with a slight shift to the right in the curve on day two. This shift may simply be a result of stimulation of a greater number of sensory fibers due to a difference in the placement of the stimulating electrode, as compared to that of day one. In figure 4.1B the mean amplitude of the H-reflex elicited during quiet full weight standing on day one (3.84 ± 0.18 mV) is similar, but slightly larger, to the mean recorded on day two (3.56 ± 0.14 mV), as the mean M-wave amplitude was 0.509 ± 0.057 mV on day one and slightly smaller on day two, 0.344 ± 0.031 mV. In figure 4.2A it can be seen that the modulation pattern of the soleus H-reflex during each phase of one average step cycle is very similar between the two days, except for a slightly more elevated H-reflex amplitude in phase one on day two. The corresponding mean ensemble soleus EMG activity (figure 4.2B) was similar, with a slightly smaller peak on day two. The average occurrence of the stance-swing transition was the same, for the same cycle duration, across the two days.

In order to verify that any changes in the amplitude of the H-reflex were due to the experimental conditions and not due to a change in the stimulation intensity used to elicit the reflex, or to a change in the baseline level of activity in the soleus muscle (this for the standing condition only), two statistical comparisons were made. The data for the first comparison is depicted in table 4.1 which shows the M-wave amplitude within normal subjects and within patients between each of the experimental conditions. There was a significant difference ($p<0.05$) between the two groups, which is evident when the mean M-wave amplitude across all conditions

Figure 4.1. Data recorded from one representative normal subject (ES) during quiet standing, under 0% BWS, on two separate occasions separated by two months.

A: Soleus H-reflex (filled circles) and M-wave (open squares) recruitment curves, each point represents one stimulation.

B: Soleus H-reflex (filled circles) elicited ten times at a stimulus intensity that maintained a constant M-wave amplitude (open squares).



A
B
DAY 1
DAY 2
SOL
4
0
150
0
Amplitude (mV)
Amplitude (µV)
0.0
1.15
Cycle Duration (s)

Figure 4.2. Data recorded from one representative normal subject (ES) during treadmill walking with 0% BWS, on two occasions separated by two months. The arrow denotes mean occurrence of swing in the average gait cycle.

A: The modulation pattern of the soleus H-reflex (filled circles, mean ± sd, n≥10) and M-wave (open squares, mean ± sd, n≥10) during one average gait cycle. Note that on day 1 there is no data for phase 5 and 15.

B: The corresponding EMG mean ensemble average (across 10 strides) of the soleus (SOL) muscle.

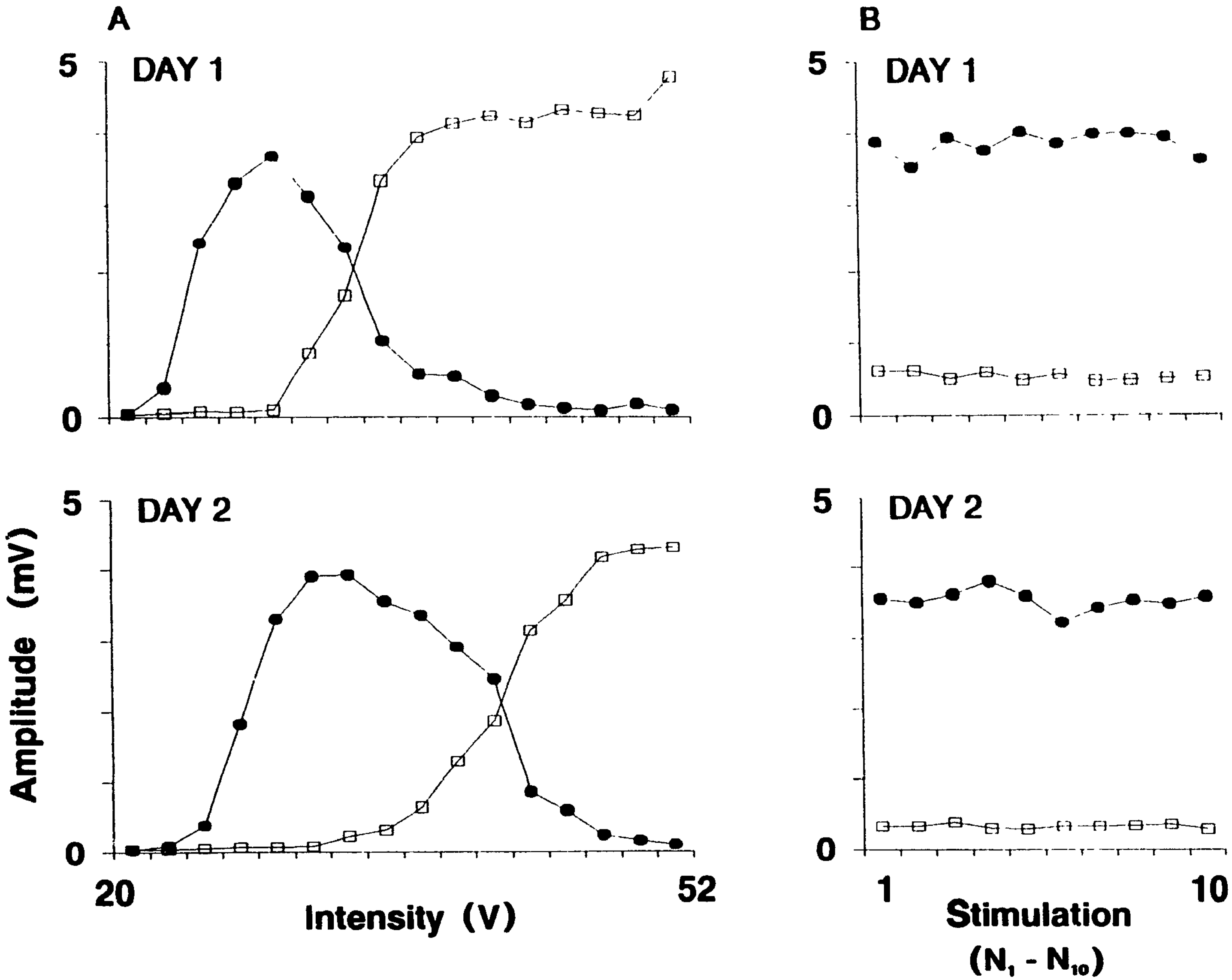

A
B
DAY 1
DAY 2
5
0
Amplitude (mV)
20
52
Intensity (V)
1
10
Stimulation
(N_1 - N_{10})

Table 4.1. Comparison of M-wave amplitude, **A:** within normal subjects (n=8) and **B:** within patients (n=8), between experimental conditions: (i) standing with 0% BWS, (ii) standing with 40% BWS, (iii) walking with 0% BWS, and (iv) walking with 40% BWS.

M-WAVE AMPLITUDE ACROSS EXPERIMENTAL CONDITIONS

Subject/	Standing FW $\bar{X}$	Standing FW SD	Standing BW $\bar{X}$	Standing BW SD	Walking FW $\bar{X}$	Walking FW SD	Walking BW $\bar{X}$	Walking BW SD	Across Cond. $\bar{X}$	Across Cond. SD
SE	.344	.031	.336	.047	.400	.038	.358	.033	.360	.028
JF	.543	.066	.567	.053	.615	.103	.536	.048	.565	.036
HB	.721	.067	.778	.063	.769	.066	.726	.083	.748	.029
MB	1.036	.102	1.070	.067	1.072	.037	1.083	.024	1.065	.020
KN	.252	.034	.235	.020	.235	.023	.222	.028	.236	.012
JB	.993	.095	1.028	.042	.902	.118	.847	.077	.942	.083
PA	.887	.101	.878	.077	.868	.073	.912	.095	.886	.019
NP	.332	.040	.394	.037	.329	.037	.291	.061	.336	.043
									$\bar{X}$= .642	.312
Patient/										
CH	.270	.072	.245	.032	.220	.030	.219	.022	.238	.242
LR	.299	.030	.301	.040	.338	.038	.308	.033	.312	.018
RM	.352	.040	.343	.043	.431	.058	.434	.082	.390	.049
AN	.465	.064	.480	.062	.540	.067	.529	.069	.504	.036
MB	.391	.060	.323	.028	.489	.041	.486	.064	.422	.080
BP	.460	.066	.354	.076	.310	.055	.285	.060	.352	.077
RG	.308	.025	.280	.047	.322	.045	.288	.053	.300	.019
SH	.153	.025	.220	.057	.216	.038	.158	.033	.187	.036
									$\bar{X}$= .338	.102

for the normal subject group (0.642 ± 0.312 mV) is compared with that of the patients (0.338 ± 0.102 mV). However there was no significant difference ($p < 0.01$) in the M-wave amplitude across each condition at full weight bearing (FWB) or BWS, for both normals and patients, which thus gives assurance that the intensity of stimulation remained constant between trials.

The data for the second statistical comparison is depicted in table 4.2 which shows the mean baseline EMG activity of the soleus and medial gastrocnemius, between the two standing conditions, within each group of subjects. In addition it shows the mean soleus EMG activity in the early stance phase (bin 1) of the gait cycle, for each normal subject under the 0% BWS condition. No statistical difference ($p < 0.05$) in mean EMG amplitude for either muscle during the two standing trials in either group was found, thus giving assurance that any change in soleus H-reflex was not due to a change in its muscular activity between the two conditions. There was, however, a significant difference ($p < 0.05$) between the mean soleus EMG activity during standing at FWB, as compared to its EMG activity during the early stance phase of the gait cycle, within the group of normal subjects.

In order to ascertain whether the difference in mean M-wave amplitude, which indicates stimulus efficiency, between the normal group pooled and the patient group pooled, may somehow be accountable for the difference in H-reflex amplitude modulation during walking, a subset of the normals (JF, ES, JB, NP) was selected who had a smaller mean M-wave amplitude (0.374 ± 0.126 mV) across all conditions comparable to the mean M-response of the patient group (0.338 ± 0.102 mV). The mean H-reflex amplitude, expressed as a percentage of the standing FWB control value, for this subset of normals for each of the three phases during walking was: early stance (15.99 ± 16.4%), push-off (77.63 ± 20.0%), and swing (3.75 ± 1.77%). These percentages of the H-reflex control value are within the normal margins, and thus indicate that the reduced M-response, elicited along with the H-reflex, is not responsible for the lack of modulation in the H-reflex recorded in these patients.

Table 4.2. Comparison of the mean baseline EMG activity of the soleus (SOL) and medial gastrocnemius (GA) between standing with 0% BWS and standing with 40% BWS, **A:** within normal subjects (n=8) and **B:** within patients (n=8). In addition, the mean soleus EMG activity across 10 strides, in phase one of the gait cycle, under the 0% BWS condition is shown for the normal subjects.

TRICEPS SURAE MEAN EMG ACTIVITY

Subject/	Standing 0% BWS		Standing 40% BWS		Walking 0% BWS
	$\bar{X}$ SOL	$\bar{X}$ GA	$\bar{X}$ SOL	$\bar{X}$ GA	$\bar{X}$ SOL PHASE 1
ES	20.64	12.44	18.71	9.33	24.77
JF	11.98	14.51	7.86	6.74	22.39
HB	15.21	8.85	10.45	8.43	9.44
MB	11.79	8.31	10.89	8.36	20.69
KN	9.68	9.14	10.56	9.95	15.58
JB	10.40	8.16	21.23	9.44	25.77
AP	10.82	10.21	9.50	9.16	19.22
NP	10.43	7.78	10.21	8.87	8.76
Patient/					
CH	7.88	9.21	8.06	6.14	
LR	7.41	7.17	6.94	11.94	
RM	14.24	9.87	13.83	6.68	
AN	11.92	9.16	12.92	6.49	
MB	23.62	12.43	15.76	5.87	
BP	9.23	7.87	6.77	6.28	
RG	14.85	5.97	8.12	5.58	
SH	13.51	12.54	19.44	8.58	

4.2 The Effect of Increasing Voluntary Soleus EMG Activity upon the H-reflex in Standing in Normal Subjects

The effect of increasing the background soleus EMG activity, without a change in ankle angle, upon the H-reflex amplitude, was examined in four of the normal subjects. Figure 4.3 shows, for each subject, the mean amplitude of the soleus H-reflex at increasing levels of background soleus EMG activity, from 0% to 100% maximum isometric contraction, with the M-wave amplitude maintained relatively constant. The H-reflex amplitude seemed to remain relatively stable from 0% to 40% maximal isometric contraction, after which the trend was that it increased in amplitude progressively with further increases in background EMG activity. Subject ES is the only subject whose data do not comply with this trend. That the soleus muscle was indeed attaining these percentages of contraction, in subject ES, was determined through off-line analysis. These data also gave assurance that the TA was not co-active with the triceps surae.

4.3 Effect of BWS on Normal Subjects and Patients Pooled as Groups

The effect upon the soleus H-reflex amplitude of providing 40% BWS to both normal subjects and patients, in the standing position and during walking, is depicted, in part, in figure 4.4. In this figure the mean amplitude of the H-reflex recorded in the standing 0% BWS condition for each individual was given a value of 100 percent, such that the amplitude of the H-reflex recorded in the subsequent standing or walking trials could be expressed as a percentage of this baseline value.

In figure 4.4 A and B it is apparent that there was no significant difference ($p<0.01$) in the H-reflex amplitude between groups and there was no significant difference ($p<0.01$) between conditions, 0% and 40% BWS. In addition there was no interaction between groups and level of BWS.

Figure 4.4 C and D depicts the soleus H-reflex modulation pattern in three phases of the step cycle, in both normals and patients under the 0% BWS and 40% BWS conditions. In this figure the normal subjects' data were pooled as were the patients' data, and again the change in H-reflex amplitude was expressed as a percentage of the 0% BWS standing condition. It can be seen in figure 4.4C that,

Figure 4.3. Soleus H-reflex amplitude (mean ± sd, n = 10) of four normal subjects (MB, ES, KN, AP) during increasing isometric contraction of the triceps surae muscle group, from 0% (standing quietly) to 100% maximal isometric contraction (MIC). The H-reflex amplitude is expressed as a percentage of the amplitude recorded at quiet standing, 0% MIC. The mean M-wave amplitude across the group of four subjects is depicted by the solid line, and the shaded area represents one standard deviation above and below the mean M-response.

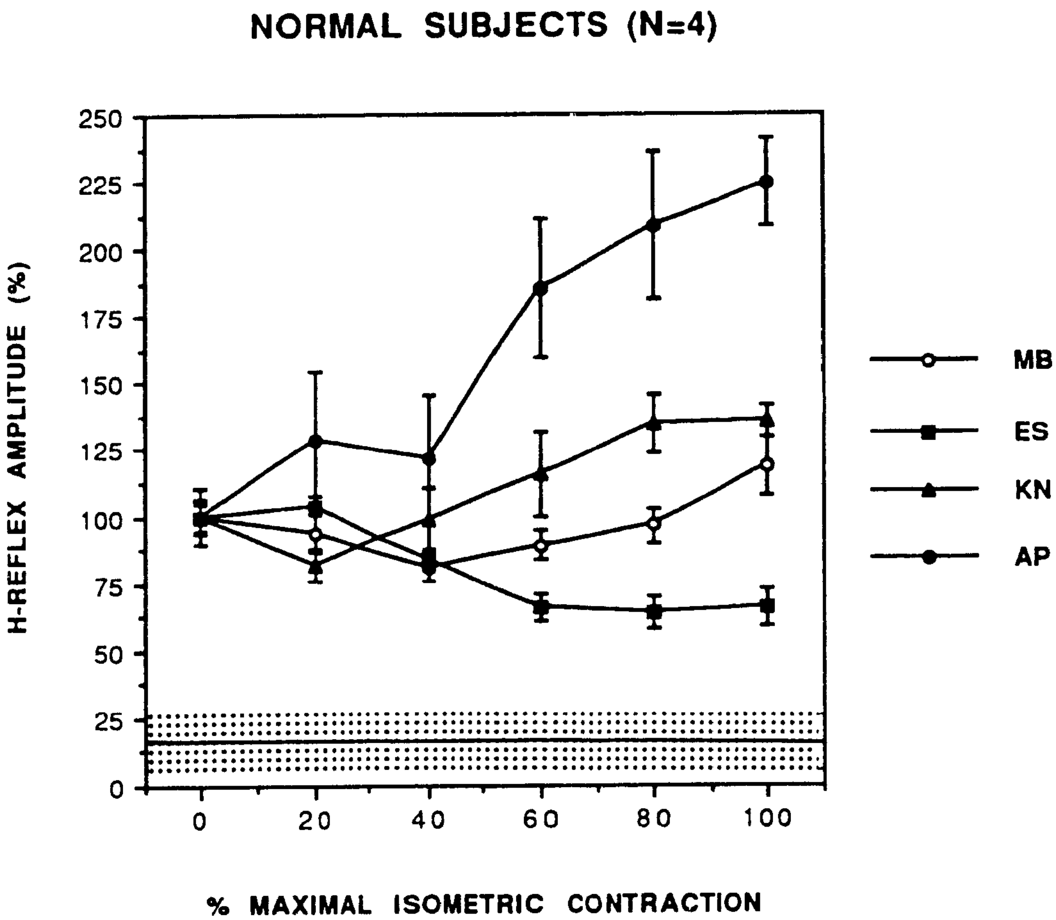
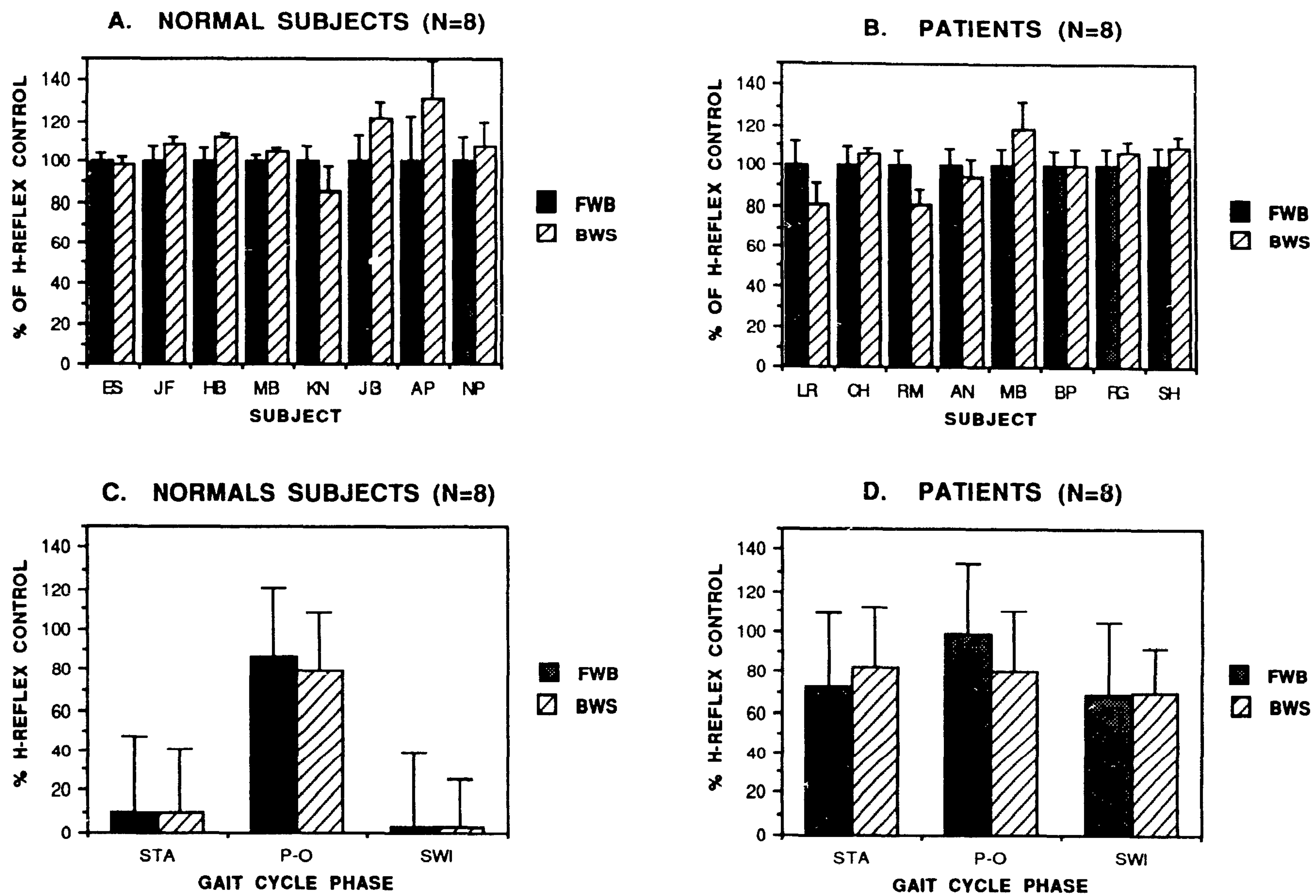

NORMAL SUBJECTS (N=4)
H-REFLEX AMPLITUDE (%)
250
225
200
175
150
125
100
75
50
25
0
0
20
40
60
80
100
% MAXIMAL ISOMETRIC CONTRACTION
MB
ES
KN
AP

Figure 4.4. Comparison of (i) soleus H-reflex amplitude (mean ± sd, n=10), between the two standing conditions 0% and 40% BWS, **A:** within normal subjects (n=8) and **B:** within patients (n=8), and (ii) soleus H-reflex amplitude (mean ± sd, n≥10) between the two walking conditions 0% and 40% BWS, as a function of the phase in the gait cycle: stance (STA), push-off (P-O), and swing (SWI), within **C:** normal subjects pooled as a group (n=8) and **D:** patients pooled as a group (n=8). The H-reflex amplitude (mean ± sd, n=10) recorded at 0% BWS was given a value of 100, in percent amplitude, and its amplitude recorded during the subsequent standing and walking trials was expressed as a percentage of this 0% BWS value.



A. NORMAL SUBJECTS (N=8)
% OF H-REFLEX CONTROL
140
120
100
80
60
40
20
0
ES JF HB MB KN JB AP NP
SUBJECT
FWB
BWS
B. PATIENTS (N=8)
% OF H-REFLEX CONTROL
140
120
100
80
60
40
20
0
LR CH RM AN MB BP RG SH
SUBJECT
FWB
BWS
C. NORMALS SUBJECTS (N=8)
% H-REFLEX CONTROL
140
120
100
80
60
40
20
0
STA P-O SWI
GAIT CYCLE PHASE
FWB
BWS
D. PATIENTS (N=8)
% H-REFLEX CONTROL
140
120
100
80
60
40
20
0
STA P-O SWI
GAIT CYCLE PHASE
FWB
BWS

in normals at FWB, there was a modulation of the H-reflex amplitude such that it was reduced in early stance, was elevated at the push-off phase, and then was virtually suppressed during swing. There was a significant difference ($p<0.001$) in the H-reflex amplitude as a result of the different phases in the gait cycle in which it was measured. Under the 40% BWS condition there was no significant difference ($p<0.001$) in the H-reflex modulation in these three gait phases, as compared to that at FWB, for the pooled group of normal subjects.

In the patients pooled as a group (figure 4.4D) there was an obvious decrease ($p<0.05$) in the degree of H-reflex modulation during the step cycle, in comparison with the same data recorded in normals. The H-reflex amplitude, under the FWB condition, in early stance and swing was abnormally high and the value at push-off was slightly greater than the other two phases, but within the normal range. Statistically, it was found that there was a significant difference ($p<0.01$) in the H-reflex amplitude modulation across different phases, however there was also a significant interaction effect ($p<0.01$) between phase and level of BWS provided. In figure 4.4D it is evident that this interaction is due most probably to the markedly decreased mean H-reflex amplitude in the push-off phase under the 40% BWS condition, as compared to FWB. In two patients (CH, LR), who had the most obvious change in H-reflex amplitude in the push-off bin, it was discovered that there was a significant difference ($p<0.001$) in mean H-reflex amplitude, as a function of the level of BWS provided, which interacted significantly ($p<0.025$) with the phase in the gait cycle.

Within the group of pooled patients, who all exhibited an abnormal H-reflex modulation during gait to some degree, it is interesting to report the extent to which the patients, individually, lacked this phasic modulation. The % inhibition or facilitation mentioned for a patient in this section is expressed as a percentage of this patient's mean H-reflex amplitude control value recorded in the standing 0% BWS condition. Of the group of eight patients under the FWB condition, RM was the only one who had a 90% inhibition of the mean H-reflex amplitude in the stance phase.Three of the eight patients (LR, AN, RG) had moderate inhibition of the H-reflex (35-55% inhibition) in the stance phase. Four of the patients (CH, MB, BP, SH) had minimum inhibition of the H-reflex (20% inhibition to 25% facilitation) in the stance phase. Again under the FWB condition RM was the only patient out of the eight who had a 95% inhibition of the H-reflex in the swing phase. Three patients (LR, MB, RG) had a moderate inhibition of the H-reflex (35-65% inhibition) and four patients (CH, AN, BP, SH) had minimum inhibition of the H-reflex (20% inhibition to 15% facilitation) in the swing phase.

Under the 40% BWS condition there was a slight change in the distribution of patients according to the level of percentage of H-reflex inhibition. In the stance phase, RM was the only patient to show a 75% inhibition of the H-reflex control value. Only one patient (LR) showed a moderate 50% inhibition of the control value and the other six patients (CH, AN, MB, BP, RG, SH) had a minimum inhibition of the H-reflex (25% inhibition to 20% facilitation) during the stance phase. In the swing phase both RM and LR had a 55-70% inhibition of the H-reflex control value. Three patients (MB, BP, RG) had a moderate inhibition (25-35% inhibition) and three patients had a minimum inhibition (5-10% inhibition) in the swing phase, with BWS.

BWS had a definite effect on the temporal distance parameters of gait, when compared to FWB walking (see table 1A and 1B in the appendix). Under the BWS condition, with cadence held constant, the normal subjects showed a decrease in the percentage of the stride spent in stance (% stance) and a subsequent increase in percentage swing (% swing), and a decrease in percentage of time in which both feet were in contact with the ground (% TDST). All of these differences in normals, pooled as a group, were significant ($p<0.01$). One exception to this trend was subject JF, who was one of the earliest subjects to take part in the study, at which time the cadence was not controlled. Thus, in her case there was an increase in % stance and % TDST under the BWS condition. In patients, the trend is exactly the same as that found in the normals, but to a greater degree. With BWS, the % stance was decreased, as was the % TDST. Subsequently, the percentage of the step cycle spent in swing was increased. Again these three differences in temporal distance factors, between FWB and BWS, in patients pooled as a group, were significant ($p<0.01$).

4.4 The Effect of BWS on One Representative Normal Subject

It is the purpose of the next three figures (figures 4.5, 4.6 and 4.7) to describe in detail the effect that removing 40% BWS, during standing and walking, has on the soleus H-reflex amplitude, the EMG activity of three lower leg muscles, and the angular excursions at four major joints, in one representative normal subject. This data will serve as a reference to which the spastic paretic patient data will be compared.

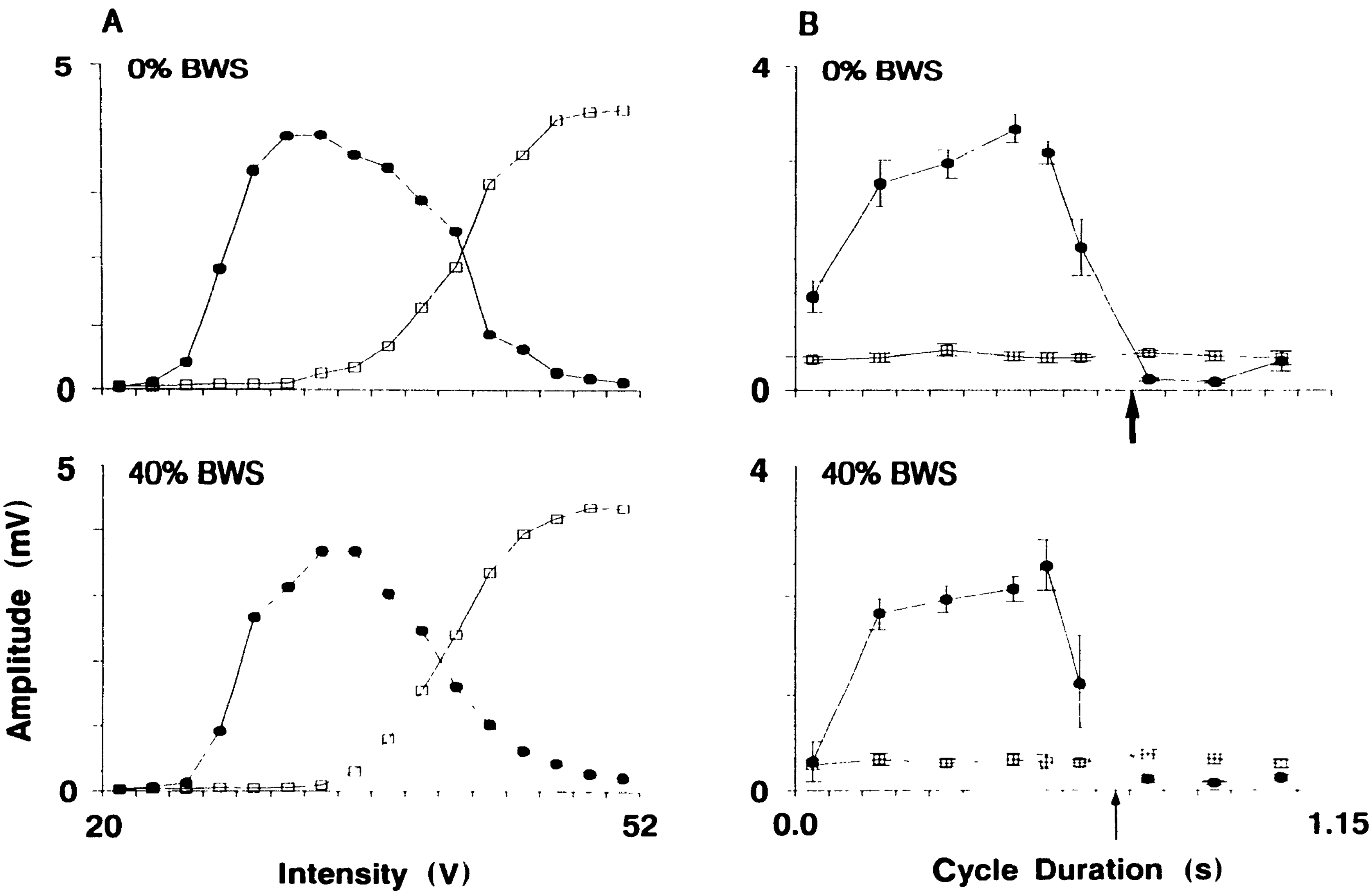
In figure 4.5A it can be seen that the H- and M-wave recruitment curves under the two conditions, 0% and 40% BWS, were very similar in shape and amplitude, with the H-reflex amplitude reaching a maximum of 3.9 and 3.7 mV, respectively. The normal phase dependent soleus H-reflex modulation, during one gait cycle under the two experimental conditions, is shown in figure 4.5B. It can be seen in this figure that, with a relatively constant M-wave amplitude, the soleus H-reflex amplitude was reduced in early stance, progressively increased from midstance to heel-off, and reached a maximum at push-off, and then was inhibited throughout swing. These changes in the excitability of the soleus motoneuron pool, as measured by the H-reflex, generally reflect the EMG activity in the soleus muscle, as shown in figure 4.6. This figure shows the normal mean EMG ensemble average patterns, during ten consecutive step cycles, for ankle flexors and extensors. The heavy line depicts the EMG activity of each muscle under the 0% BWS condition. As a plantarflexor, the soleus muscle (SOL) is active during stance, initially to control the forward rotation of the leg at the ankle, and then between 40-60% of the stride to generate a push-off force. The medial gastrocnemius (GA) functions in much the same manner as the soleus, with its peak of activity occurring at 50% of the stride. The tibialis anterior (TA) is reciprocally active to the triceps surae. It peaks immediately after heel contact to lower the foot gently to the ground after which it is generally silent until toe-off, at which point it produces its major activity, in order to dorsiflex the foot, to clear the foot during midswing.

Under the 40% BWS condition, the soleus H-reflex modulation during walking, as shown in figure 4.5B, was changed very little with respect to that at 0% BWS. However, the 40% BWS peak H-reflex amplitude (2.8 mV) occurred in phase 8 of

Figure 4.5. Comparison of H-reflex data recorded from one representative normal subject (ES) under the two experimental conditions, 0% and 40% BWS.

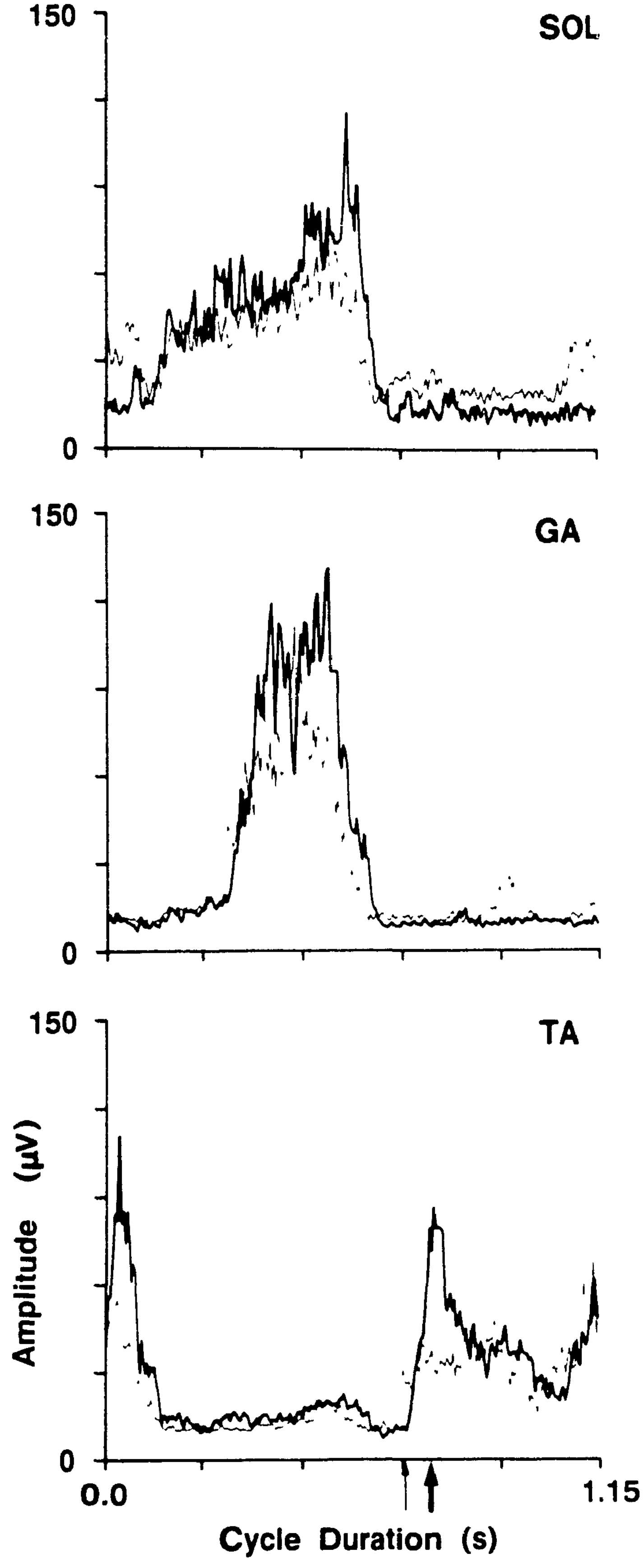
A: Soleus H-reflex (filled circles) and M-wave (open squares) recruitment curves, each point represents one stimulation.

B: The modulation pattern of the soleus H-reflex (filled circles, mean ± sd, n≥10) during an average gait cycle, with the M-wave (open squares, mean ± sd, n≥10) kept constant. The arrow denotes mean occurrence of swing in the average gait cycle: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS.



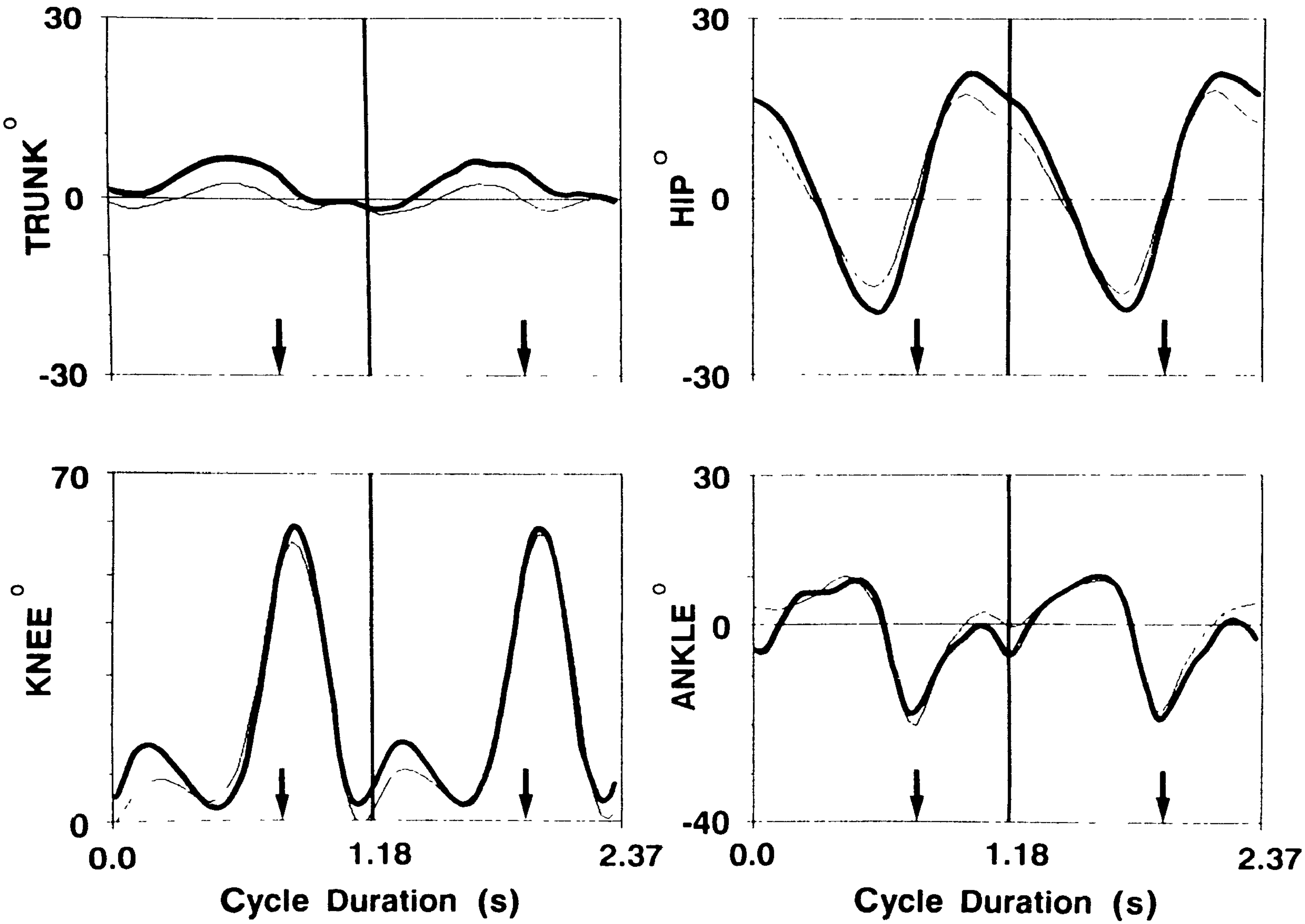
A
B
0% BWS
40% BWS
0% BWS
40% BWS
5
0
5
0
4
0
4
0
20
52
0.0
1.15
Amplitude (mV)
Intensity (V)
Cycle Duration (s)

Figure 4.6. Comparison of EMG mean ensemble averages (across 10 strides) of the soleus (SOL), medial gastrocnemius (GA), and tibialis anterior (TA) muscles of the tested leg of one representative normal subject (ES), normalized to the average gait cycle, under the two walking conditions, 0% and 40% BWS. The EMG activity recorded at 0% BWS is depicted by the heavy line and that recorded at 40% BWS is depicted by the light line. The arrows denote stance-swing transition: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS.



150
SOL
0
150
GA
0
150
TA
Amplitude (µV)
0
0.0
1.15
Cycle Duration (s)

Figure 4.7. Comparison of the sagittal angular excursions of two consecutive gait cycles (expressed in real time) for the trunk, hip, knee, and ankle of one representative normal subject (ES) under the two walking conditions, 0% and 40% BWS. The angular excursions recorded at 0% BWS are depicted by the heavy line and those recorded at 40% BWS are depicted by the light line. The solid vertical line denotes termination of the first gait cycle and the arrows denote the stance-swing transition, which in this subject is the same under the two conditions. Positive angular displacements indicate flexion at all joints and negative angular displacements indicate hyperextension at the trunk, extension at the hip, and plantarflexion at the ankle.



TRUNK °
30
0
-30
HIP °
30
0
-30
KNEE °
70
0
ANKLE °
30
0
-40
0.0
1.18
2.37
Cycle Duration (s)
0.0
1.18
2.37
Cycle Duration (s)

the gait cycle, as compared to a slightly larger peak of 3.2 mV which occurred in phase 7 of the FWB condition. The slight decrease in H-reflex amplitude under the BWS condition may be a result of the slight decrease in soleus EMG activity, as depicted in figure 4.6 (thin line). BWS had the effect of producing a decrease in the mean burst of activity in the stance muscles predominantly and a slight decrease in the mean burst of the TA, without altering the timing of the muscular activity.

The pattern of trunk, hip, knee, and ankle sagittal angular excursions for the representative normal subject, during two consecutive step cycles, is shown in figure 4.7. The kinematic data recorded under the 0% BWS condition are depicted by the thick line. The trunk was in 0 degrees of flexion just after heel strike and then slowly flexed during midstance, before finally extending at end of swing. The hip joint had its maximum flexion just prior to heel strike, after which it extended, as the leg passed underneath the trunk, until push-off, when it flexed to bring the leg forward for the next heel strike. The knee joint had two excursions of flexion and extension during one stride. Just after heel strike, the knee flexed mildly during load acceptance, after which it extended as the trunk and contralateral leg moved forward. Then it flexed prior to, and after toe-off, in order to provide foot-floor clearance early in swing, before finally extending the leg forward for heel contact. The ankle joint, as well, had two flexion/extension displacements in one stride. At heel strike, the ankle joint was slightly plantarflexed, after which it was dorsiflexed as the leg rotated over the joint. Then at push-off there was a gradual plantarflexion, followed by a rapid dorsiflexion at midswing, to provide foot-floor clearance.

Providing 40% BWS to the subject caused a decrease in flexion of the trunk, hip, and initial yield of the knee, and a decrease in maximum extension at the hip, but a slight increase in extension of the trunk. The angular displacement of the ankle was little changed; however, there was an increase in dorsiflexion at end of swing and at heel strike.

4.5 Effect of BWS on Spastic Paretic Patients

Due to the fact that there was no representative patient of the eight studied, the results from four patients will be presented in detail. This will provide a more complete understanding of the link between the soleus H-reflex modulation and the EMG activity and kinematics during standing and walking, and of the effects providing 40% BWS has upon these parameters.

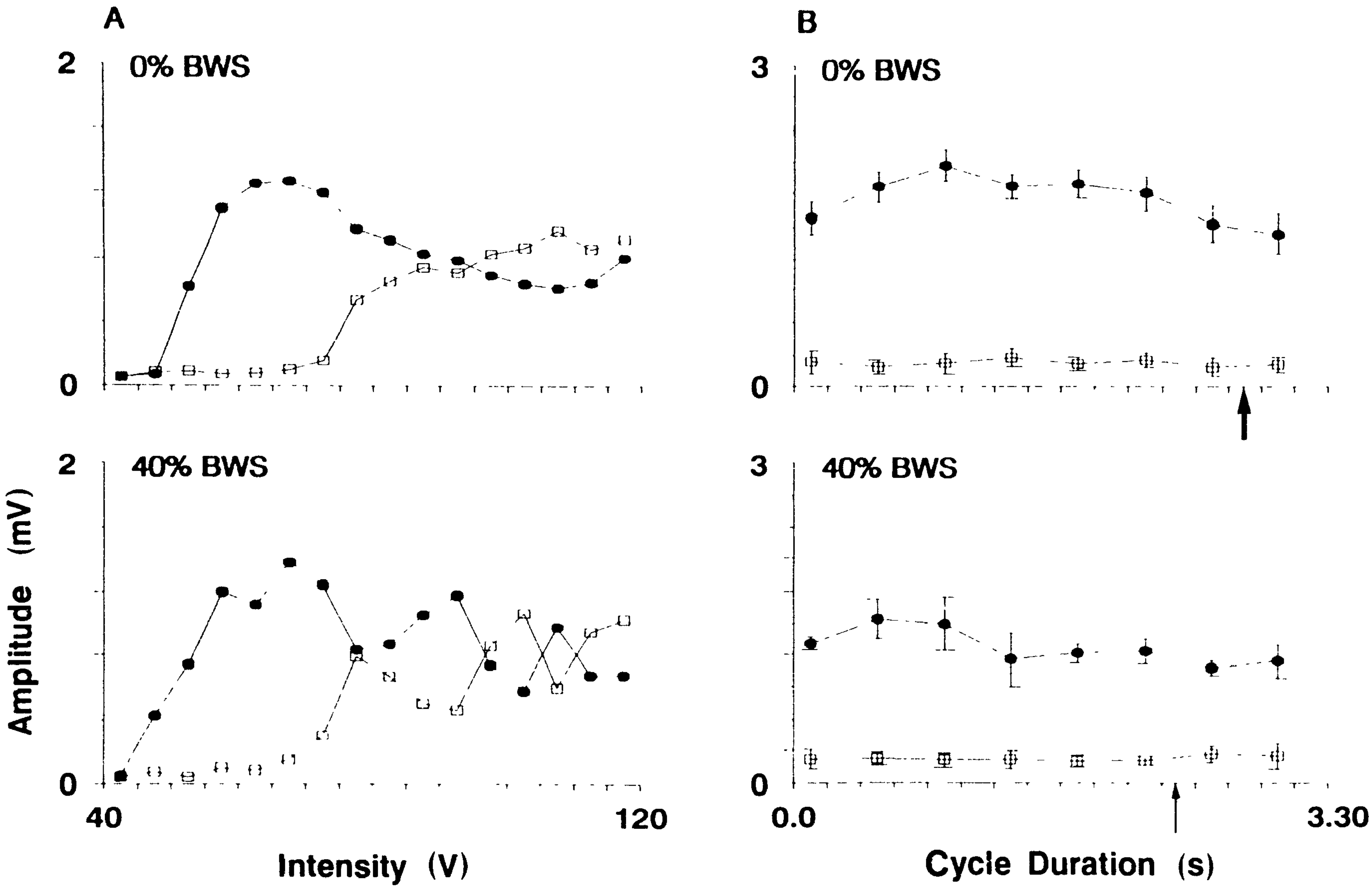
Figures 4.8, 4.9, and 4.10 show the results from one of the more severely affected patients (CH). In figure 4.8A, note that, contrary to normal, the H-reflex recruitment curve, under the FWB condition, after attaining its peak, reached a plateau and remained at a relatively high amplitude despite increasing stimulus intensity. The M-wave increased as well and then reached a plateau, but never attained an amplitude greater than the peak H-reflex amplitude. With 40% BWS the H- and M-wave recruitment curves were not as smooth as at FWB; however, the same amplitude as at FWB was achieved by each curve, with a subsequent variable plateau.

During walking, as shown in figure 4.8B, the normal phase-dependent modulation of the soleus H-reflex under the FWB condition was virtually absent, despite a stable and low M-response. The stance phase occupied a greater percentage of the gait cycle than was the case in normals, with swing occupying only a small percentage. With 40% BWS, there was a reduction in the amplitude of the H-reflex in all phases of gait, as compared with that under FWB, however the modulation of the H-reflex was still absent. The stance-swing transition occurred much earlier, which was a change towards a more normal gait pattern. The corresponding ensemble average EMG activity across 10 strides is depicted in figure 4.9 (FWB: thick line, BWS: thin line), and it is evident that the EMG activity in the soleus and GA, with 0% BWS, was low, with activity present throughout the gait cycle and clonus evident in the soleus burst. The TA EMG activity profile was close to normal, but it did not produce dorsiflexion at the ankle, as seen in figure 4.10. Under the BWS condition, the sustained clonic activity in the soleus muscle was markedly decreased and a more normal, but small, muscle burst appeared. The timing of the bursts of EMG activity in both the soleus and GA was more normal with 40% BWS. The GA and TA ensemble average EMG muscle bursts were also decreased in amplitude with BWS.

Figure 4.8. Comparison of H-reflex data recorded from one spastic paretic patient (CH) under the two experimental conditions, 0% and 40% BWS.

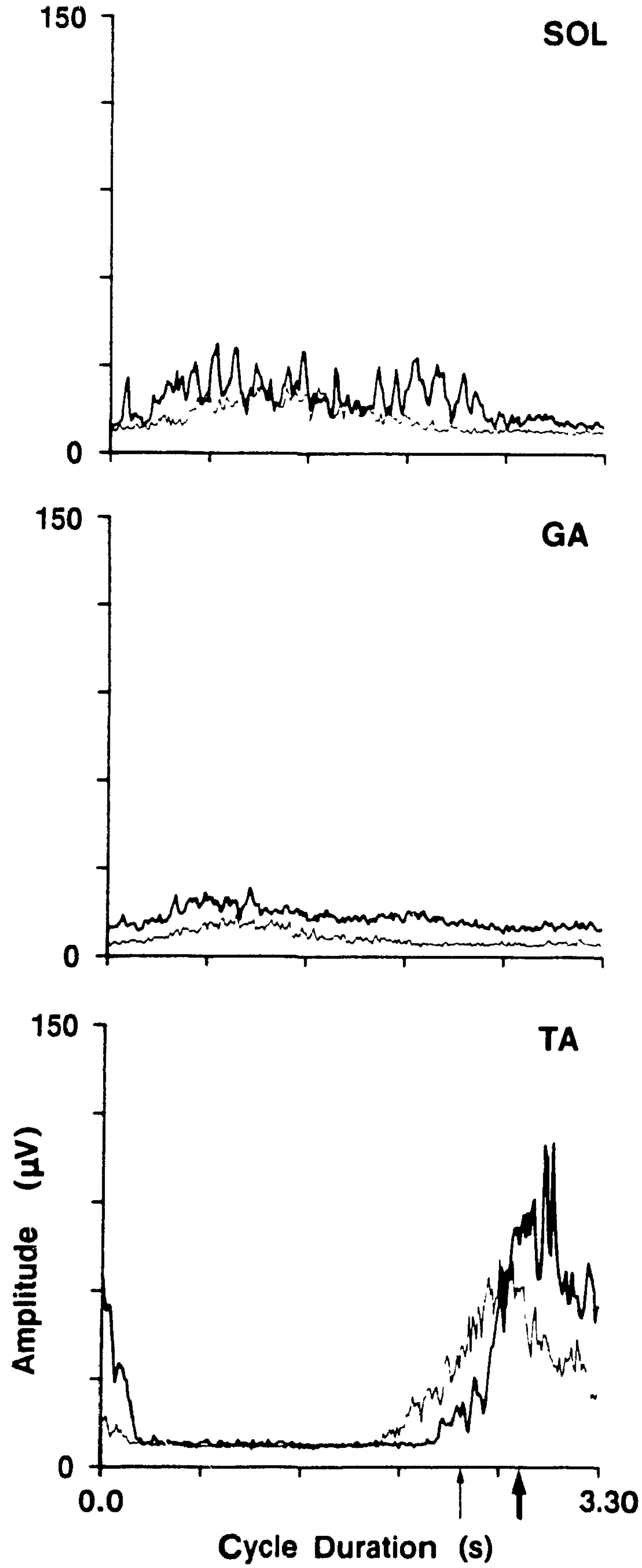
A: Soleus H-reflex (filled circles) and M-wave (open squares) recruitment curves, each point represents one stimulation.

B: The modulation pattern of the soleus H-reflex (filled circles, mean ± sd, $n \geq 10$) during an average gait cycle, with the M-wave (open squares, mean ± sd, $n \geq 10$) kept constant. The arrow denotes mean occurrence of swing in the average gait cycle: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS.



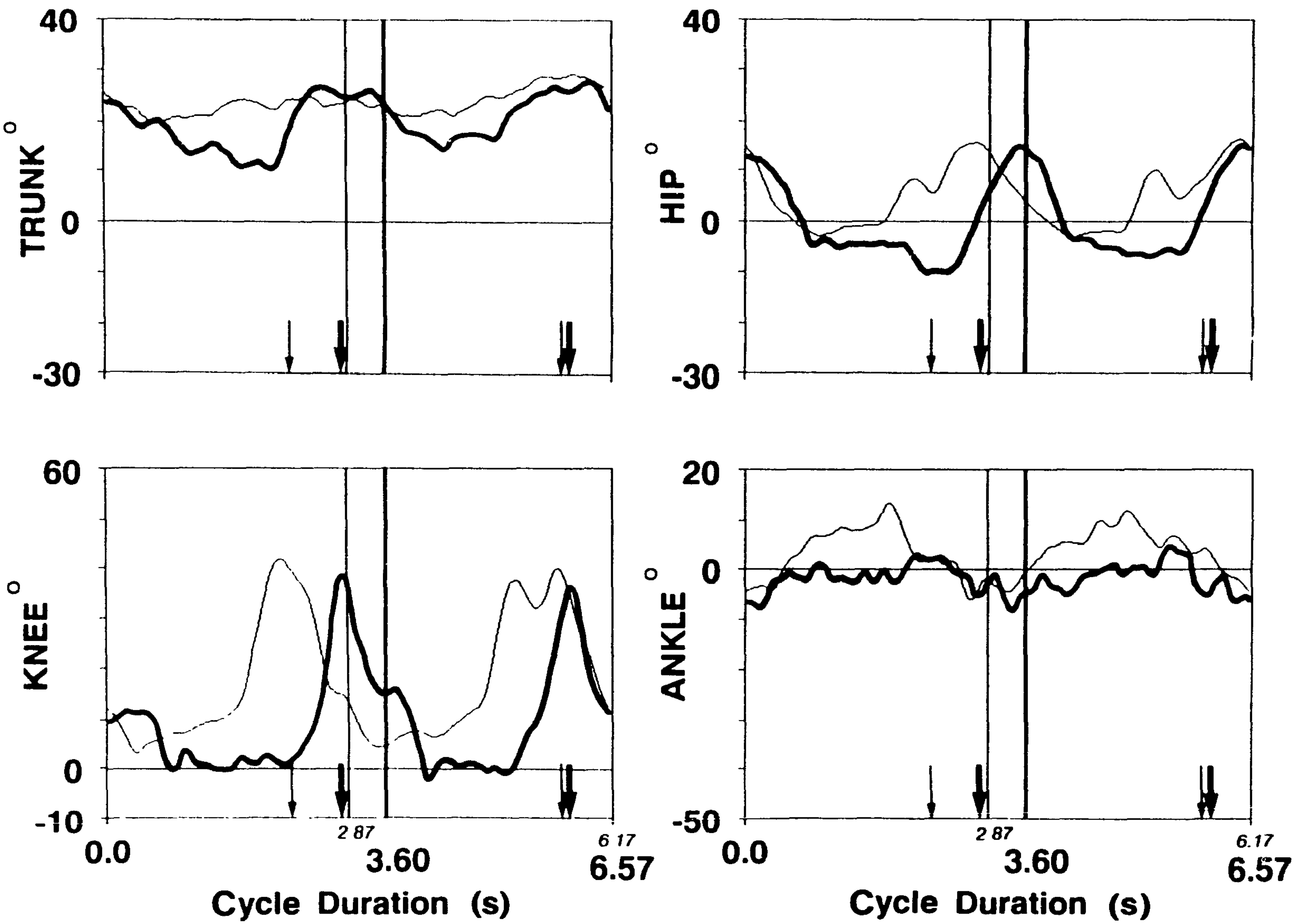
A
B
0% BWS
40% BWS
0% BWS
40% BWS
2
0
2
0
3
0
3
0
Amplitude (mV)
40
120
Intensity (V)
0.0
3.30
Cycle Duration (s)

Figure 4.9. Comparison of EMG mean ensemble averages (across 10 strides) of the soleus (SOL), medial gastrocnemius (GA), and tibialis anterior (TA) muscles of the tested leg of one spastic paretic patient (CH), normalized to the average gait cycle, under the two walking conditions, 0% and 40% BWS. The EMG activity recorded at 0% BWS is depicted by the heavy line and that recorded at 40% BWS is depicted by the light line. The arrows denote stance-swing transition: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS.



150
SOL
0
150
GA
0
150
TA
Amplitude (µV)
0
0.0
3.30
Cycle Duration (s)

Figure 4.10. Comparison of the sagittal angular excursions of two consecutive gait cycles (expressed in real time) for the trunk, hip, knee, and ankle of one spastic paretic patient (CH) under the two walking conditions, 0% and 40% BWS. The angular excursions recorded at 0% BWS are depicted by the heavy line and those recorded at 40% BWS are depicted by the light line. The heavy vertical line and the light vertical line depict termination of the first gait cycle under the 0% BWS condition and the 40 % BWS condition, respectively. The arrows denote the stance-swing transition: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS. The two representative gait cycles under the 0% BWS condition occupied a longer period of time than did the two gait cycles under the 40% BWS condition. Note that the trunk remained flexed throughout the two gait cycles. The positive and negative angular displacements at the hip indicate flexion and extension, respectively. The positive and negative angular displacements at the ankle indicate dorsiflexion and plantarflexion, respectively. The positive angular displacement at the knee indicates flexion and the negative displacement, hyperextension.



40
TRUNK °
0
-30
40
HIP °
0
-30
60
KNEE °
0
-10
0.0
2 87
3.60
6 17
6.57
Cycle Duration (s)
20
ANKLE °
0
-50
0.0
2 87
3.60
6.17
6.57
Cycle Duration (s)

The kinematic data of two step cycles in real time are displayed in figure 4.10. The trunk is maintained in an abnormally flexed position throughout the step cycle and BWS simply reduced the degree of flexion produced during FWB walking. The degree of hip joint extension is limited at FWB and is slightly more limited with BWS. The knee and ankle joints show the greatest improvement with BWS. At FWB, foot-floor contact was made with the knee flexed, with subsequent knee hyperextension being maintained throughout the loading phase. With 40% BWS, the degree of knee flexion at heel strike was reduced, and there was an improved yield phase followed by a greater degree of flexion at toe-off. These changes in the knee were associated with the ankle joint, such that with 40% BWS the ankle was more dorsiflexed at heel strike and achieved a substantially greater degree of dorsiflexion at midstance, rather than being locked into 90 degrees, with respect to the lower leg, throughout stance, as occurred at 0% BWS. In addition there was an improvement in the degree of plantarflexion at the ankle. This patient relied heavily on the handrails for balance and to assist in swinging his more affected right leg forward.

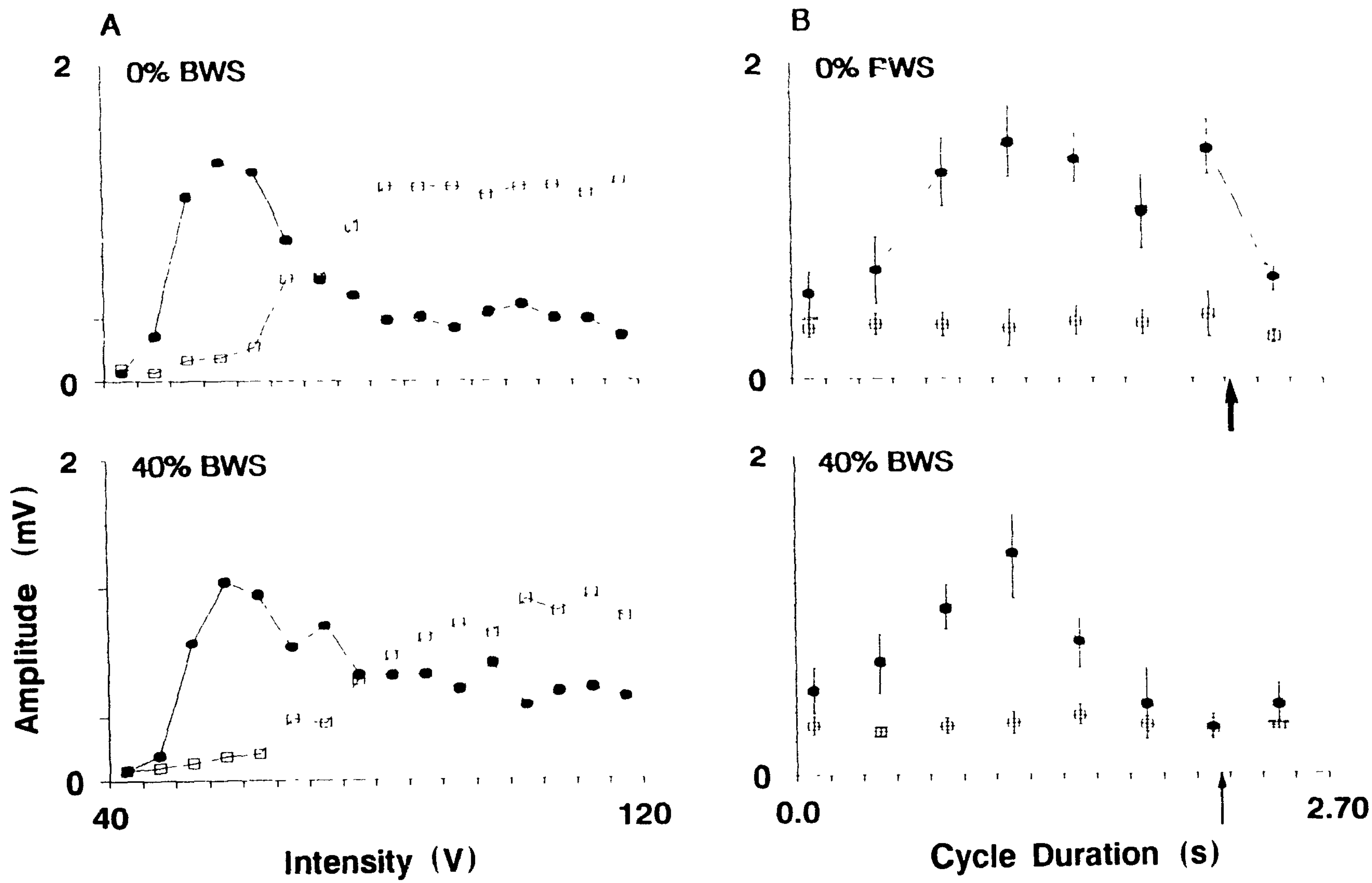
Figures 4.11, 4.12, and 4.13 depict the data of a moderately spastic subject (LR) whose gait was asymmetrical. The tested leg was the more functional of the two lower limbs. Her H- and M-wave recruitment curves under the 0% BWS condition, as shown in figure 4.11A, were close to normal, however, the H-reflex was never completely inhibited. With 40% BWS the recruitment curves were little changed, as compared to those at FWB, however the H-reflex amplitude remained slightly more elevated at the higher levels of stimulation intensity.

In figure 4.11B it is evident that there was some modulation of the soleus H-reflex during the step cycle with 0% BWS, as the H-reflex was suppressed in early stance, was increased during midstance, and was inhibited again in the one swing phase in which a stimulus was delivered. The shape of the H-reflex modulation followed somewhat the soleus muscle activation pattern, which is displayed in figure 4.12. Rather than the normal soleus ensemble average EMG profile, RL's soleus muscle activity peaked in early stance and then progressively diminished, with a small plateau in the activity in late stance, after which it became silent. With 40%

Figure 4.11. Comparison of H-reflex data recorded from one spastic paretic patient (LR) under the two experimental conditions, 0% and 40% BWS.

A: Soleus H-reflex (filled circles) and M-wave (open squares) recruitment curves, each point represents one stimulation.

B: The modulation pattern of the soleus H-reflex (filled circles, mean ± sd, $n \geq 10$) during an average gait cycle, with the M-wave (open squares, mean ± sd, $n \geq 10$) kept constant. The arrow denotes mean occurrence of swing in the average gait cycle: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS.



A
0% BWS
40% BWS
B
0% FWS
40% BWS
Amplitude (mV)
2
0
2
0
40
120
Intensity (V)
2
0
2
0
0.0
2.70
Cycle Duration (s)

Figure 4.12. Comparison of EMG mean ensemble averages (across 10 strides) of the soleus (SOL), medial gastrocnemius (GA), and tibialis anterior (TA) muscles of the tested leg of one spastic paretic patient (LR), normalized to the average gait cycle, under the two walking conditions, 0% and 40% BWS. The EMG activity recorded at 0% BWS is depicted by the heavy line and that recorded at 40% BWS is depicted by the light line. The arrows denote stance-swing transition: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS.

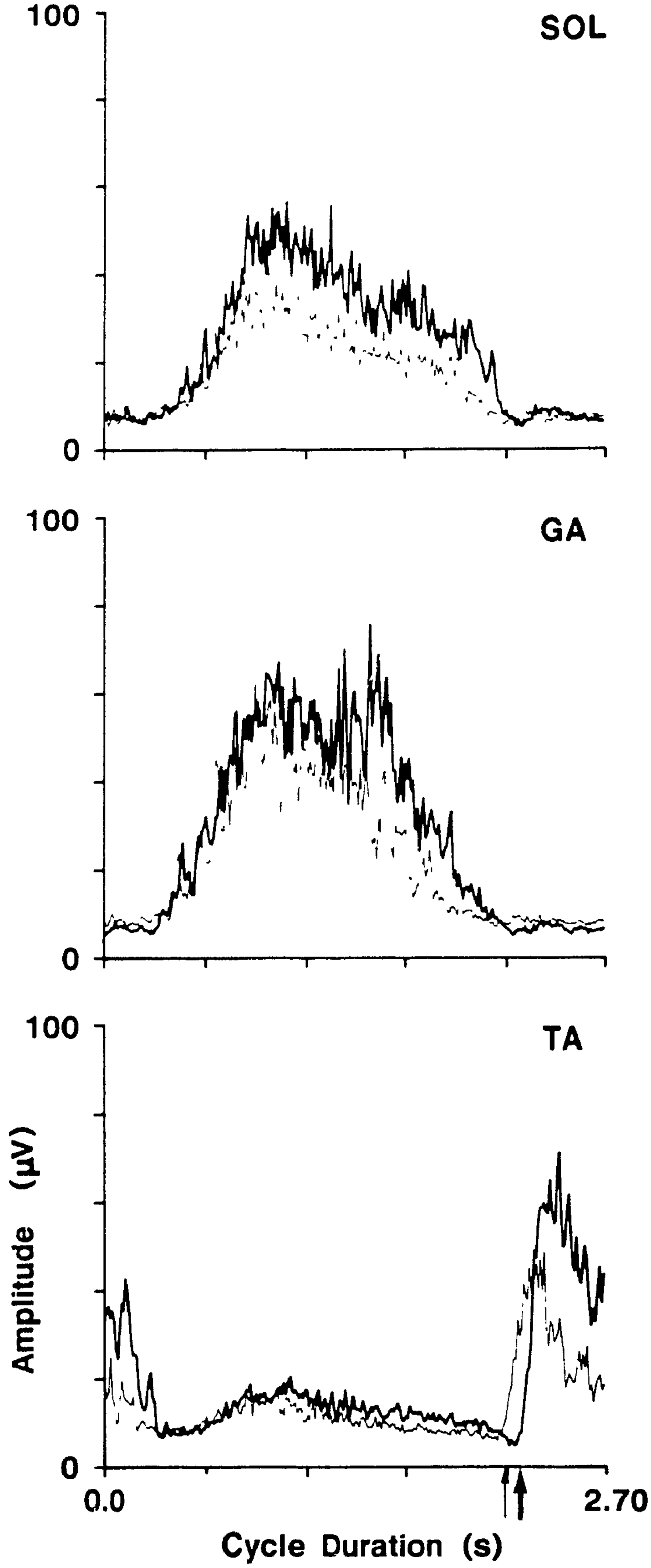
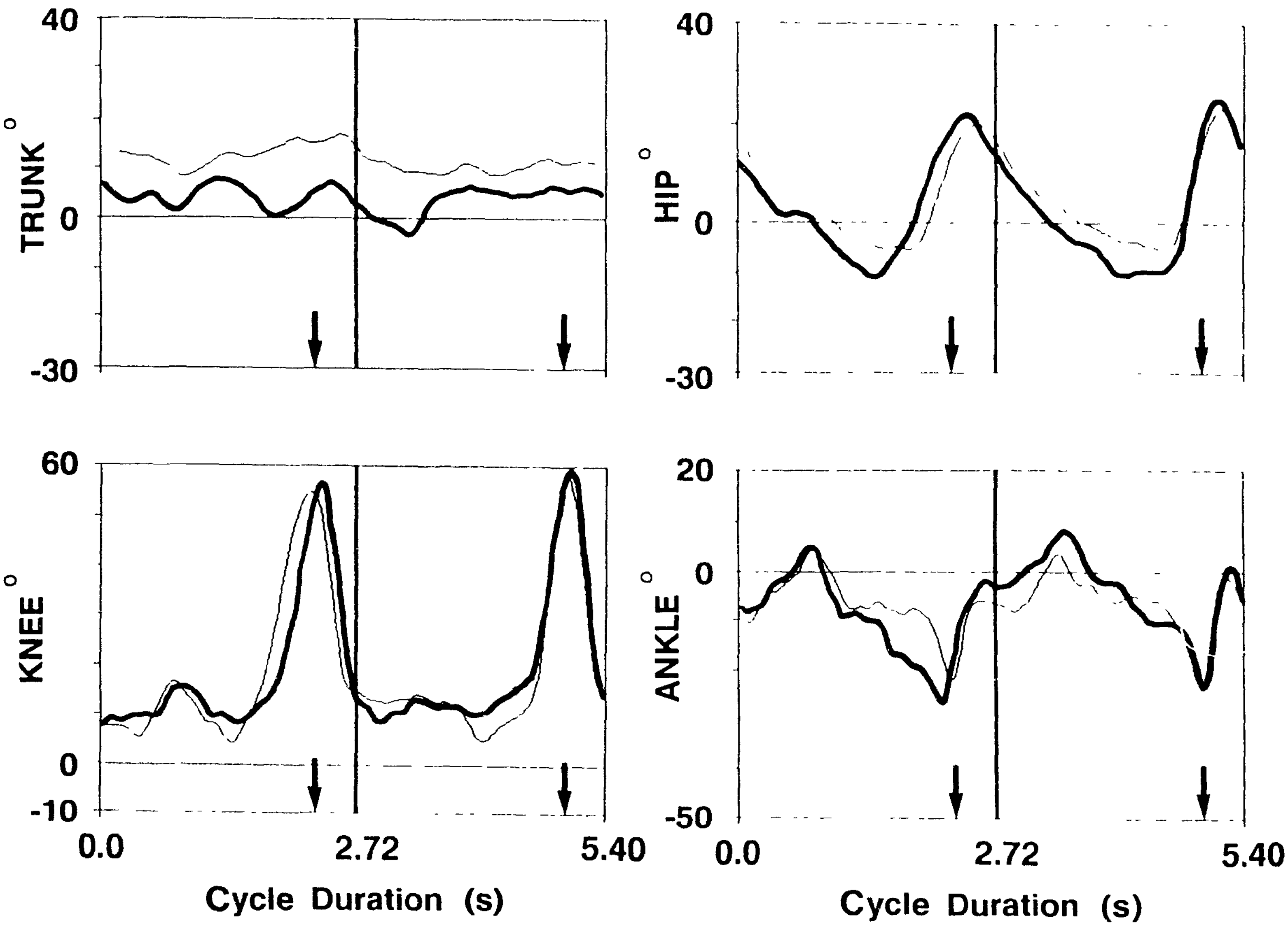

100
SOL
0
100
GA
0
100
TA
Amplitude (µV)
0
0.0
2.70
Cycle Duration (s)

Figure 4.13. Comparison of the sagittal angular excursions of two consecutive gait cycles (expressed in real time) for the trunk, hip, knee, and ankle of one spastic paretic patient (LR) under the two walking conditions, 0% and 40% BWS. The angular excursions recorded at 0% BWS are depicted by the heavy line and those recorded at 40% BWS are depicted by the light line. The solid vertical line denotes termination of the first gait cycle and the arrows denote the stance-swing transition, which in this patient is the same for the two conditions. The positive and negative angular displacements of the trunk indicate flexion and hyperextension, respectively. The positive and negative angular displacements at the hip and ankle indicate flexion/dorsiflexion and extension/plantarflexion, respectively. The positive angular displacement of the knee indicates the degree of flexion at that joint.



40
TRUNK °
0
-30
40
HIP °
0
-30
60
KNEE °
0
-10
0.0
2.72
5.40
Cycle Duration (s)
20
ANKLE °
0
-50
0.0
2.72
5.40
Cycle Duration (s)

BWS, the H-reflex modulation, as shown in figure 4.11B, was changed, in that there was a peak level in the reflex in phase seven and then a gradual inhibition of the reflex until toe-off, after which it was slightly increased. However, with 40% BWS, the H-reflex was still not completely inhibited in the swing phase, as it is in normals In examining the EMG muscle burst profiles (see figure 4 12) it is evident that BWS caused a decrease in mean EMG amplitude in all of the muscles The TA muscle activation pattern under both conditions was near normal, and the timing of the muscle bursts in all three muscles remained unchanged with BWS

The angular excursions of the major joints of patient LR during two consecutive strides are shown in Figure 4.13. It can be seen that the jerky movements of the trunk under 0% BWS were replaced by a smoother oscillation, but in a more flexed position, with 40% BWS The near normal hip displacement profile at 0% BWS was altered very little at 40% BWS, except for a decrease in maximum extension. The only change in the knee with BWS was a better initial flexion and extension during the yielding phase BWS, and its effect on the knee joint, caused the ankle to be less dorsiflexed at heel strike and enabled it to achieve a more correct push-off, with a less abrupt plantarflexion, followed by a quick dorsiflexion This change in push-off resulted from the decreased hiking motion made with her good leg, that LR used to assist in clearing the foot of the more affected leg, during swing Qualitatively LR exhibited a smoother gait pattern at 40% BWS

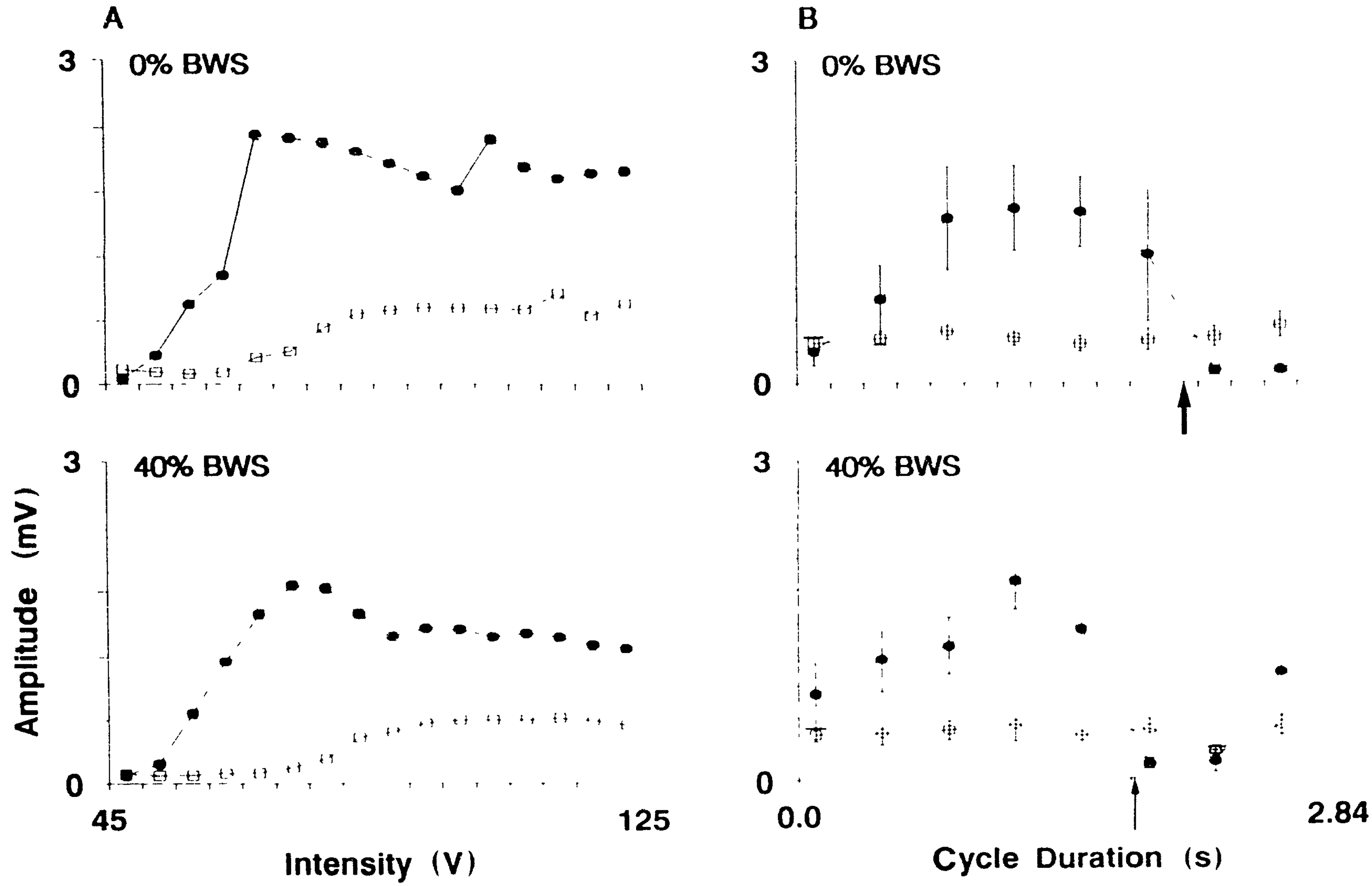
Figures 4.14, 4 15, and 4.16 display the results obtained from a more spastic patient (RM). His H- and M-wave recruitment curves, depicted in figure 4.14A, are similar to those of patient CH, but the H-reflex curve reached a plateau even earlier, once it had attained its peak amplitude. The M-wave did not increase much at all, under either condition, 0% or 40% BWS. With 40% BWS, the maximum H-reflex amplitude was lowered, as was the plateau of the H-reflex amplitude at higher stimulus intensities.

The modulation of the soleus H-reflex during one average step cycle under both conditions, 0% and 40% BWS, for subject RM is shown in figure 4.14B. At 0% BWS the H-reflex was modulated throughout the step cycle in a near normal manner. This was surprising because the ensemble average muscle profiles, as

Figure 4.14. Comparison of H-reflex data recorded from one spastic paretic patient (RM) under the two experimental conditions, 0% and 40% BWS.

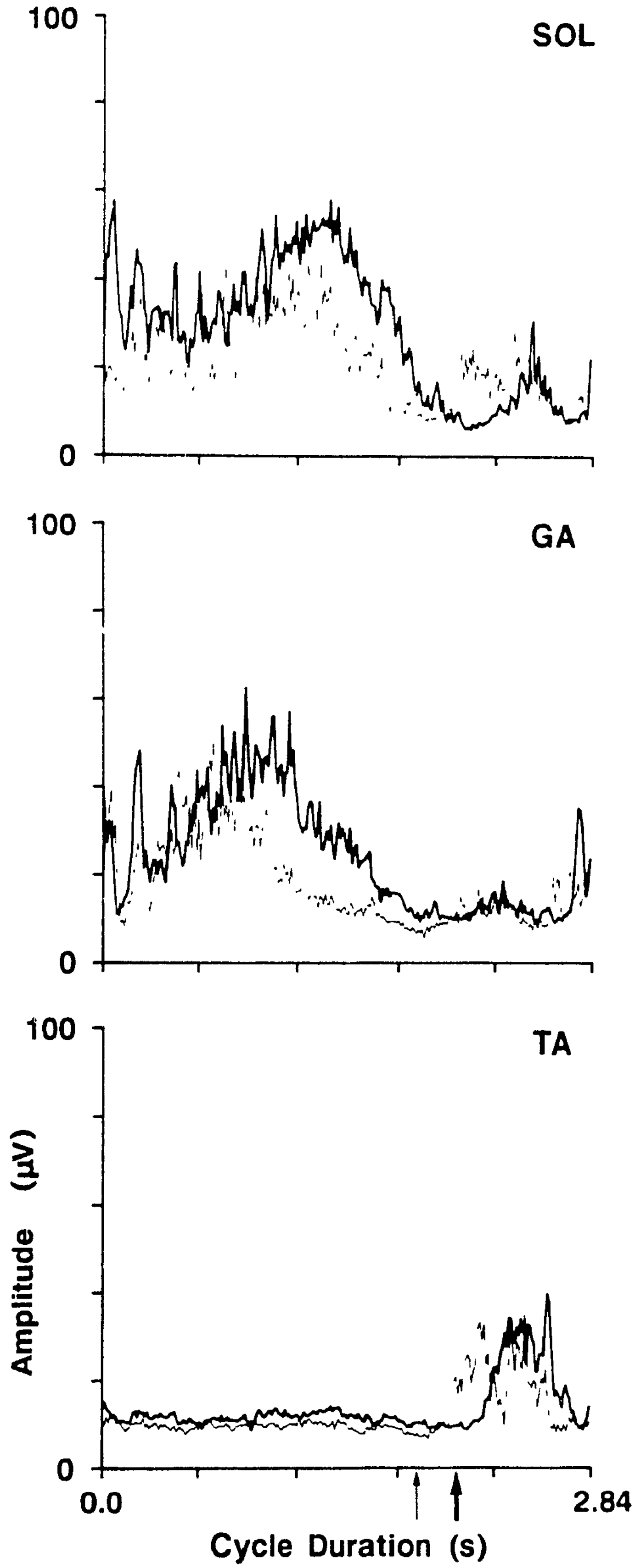
A: Soleus H-reflex (filled circles) and M-wave (open squares) recruitment curves, each point represents one stimulation.

B: The modulation pattern of the soleus H-reflex (filled circles, mean ± sd, n≥10) during an average gait cycle, with the M-wave (open squares, mean ± sd, n≥10) kept constant. The arrow denotes mean occurrence of swing in the average gait cycle: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS.



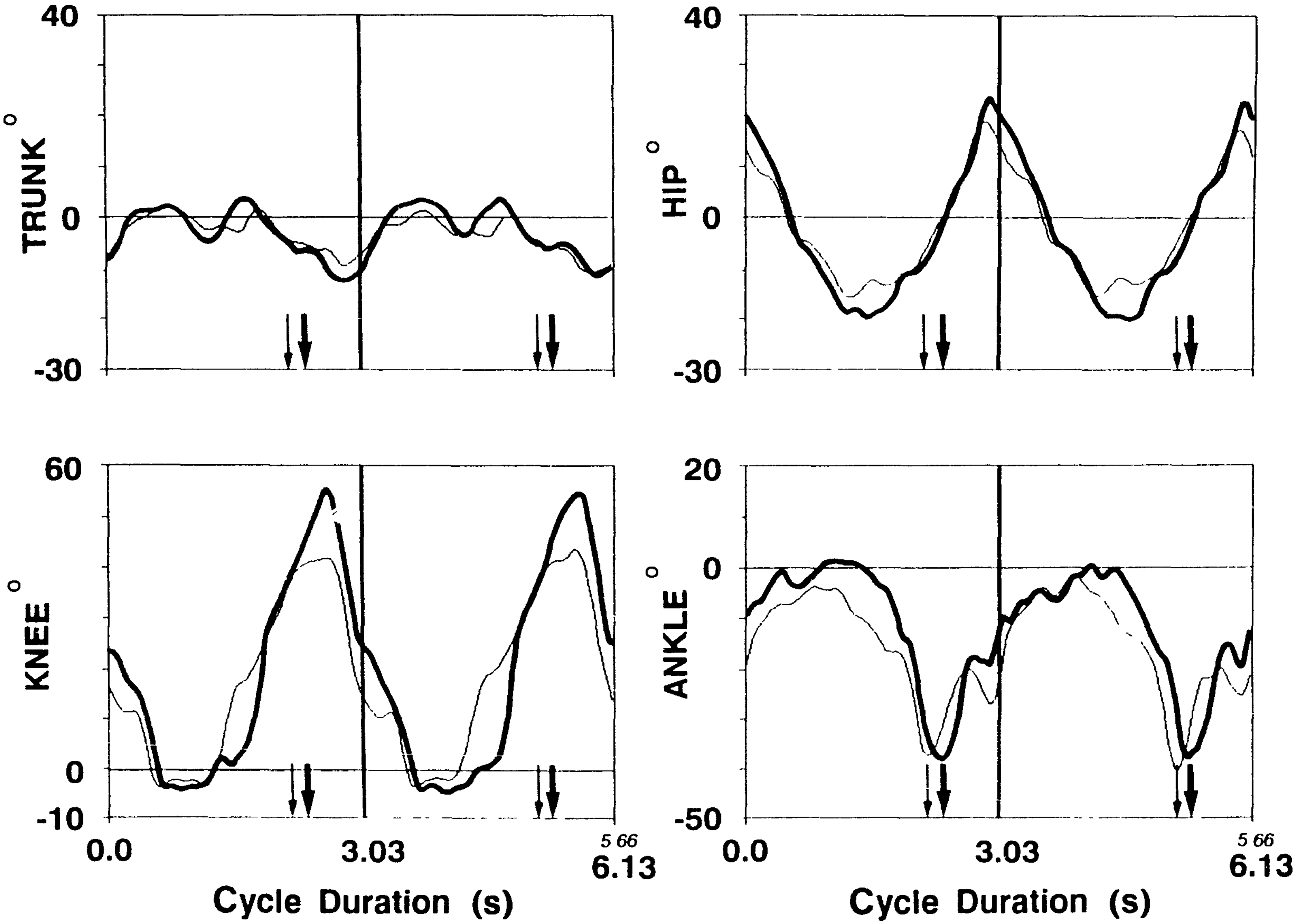
A
0% BWS
40% BWS
3
0
3
0
45
125
Intensity (V)
Amplitude (mV)
B
0% BWS
40% BWS
3
0
3
0
0.0
2.84
Cycle Duration (s)

Figure 4.15. Comparison of EMG mean ensemble averages (across 10 strides) of the soleus (SOL), medial gastrocnemius (GA), and tibialis anterior (TA) muscles of the tested leg of one spastic paretic patient (RM), normalized to the average gait cycle, under the two walking conditions, 0% and 40% BWS. The EMG activity recorded at 0% BWS is depicted by the heavy line and that recorded at 40% BWS is depicted by the light line. The arrows denote stance-swing transition: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS.



100
SOL
0
100
GA
0
100
TA
0
Amplitude (μV)
0.0
2.84
Cycle Duration (s)

Figure 4.16. Comparison of the sagittal angular excursions of two consecutive gait cycles (expressed in real time) for the trunk, hip, knee, and ankle of one spastic paretic patient (RM) under the two walking conditions, 0% and 40% BWS. The angular excursions recorded at 0% BWS are depicted by the heavy line and those recorded at 40% BWS are depicted by the light line. The heavy vertical line depicts termination of the first gait cycle, which is the same under the two conditions. The arrows denote the stance-swing transition: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS. The trunk and knee positive and negative angular displacements indicate flexion and hyperextension, respectively, at these joints. The positive and negative angular displacements at the hip indicate flexion and extension, respectively. The positive and negative angular displacements at the ankle indicate dorsiflexion and plantarflexion, respectively.



TRUNK °
40
0
-30
HIP °
40
0
-30
KNEE °
60
0
-10
ANKLE °
20
0
-50
0.0
3.03
5 66
6.13
Cycle Duration (s)
0.0
3.03
5 66
6.13
Cycle Duration (s)

displayed in figure 4.15, are quite abnormal. The 0% BWS soleus and GA muscle profiles (thick line) exhibited early stretch activation, clonus in early stance, and a burst of activity during swing. The TA was minimally active with only one burst of activity in swing that coincided with the activity in the soleus during swing. This co-activation was part of the reason foot-floor contact was made with the ball of the foot, rather than the heel. With 40% BWS, the H-reflex modulation was somewhat altered, as compared to that at 0% BWS (see figure 4.14B). The peak H-reflex amplitude was no longer shared by three phases, as at 0% BWS, but occurred definitely in phase seven. Inhibition of the reflex occurred earlier, as a result of the earlier initiation of swing under the 40% BWS condition. The H-reflex in phase 15, at end of swing, was not inhibited, and was related to the burst of soleus activity during swing, as is shown in figure 4.15 (thin line). BWS decreased the mean amplitude of the muscle bursts and reduced the amplitude of the early stretch activation and clonus present in the soleus and GA.

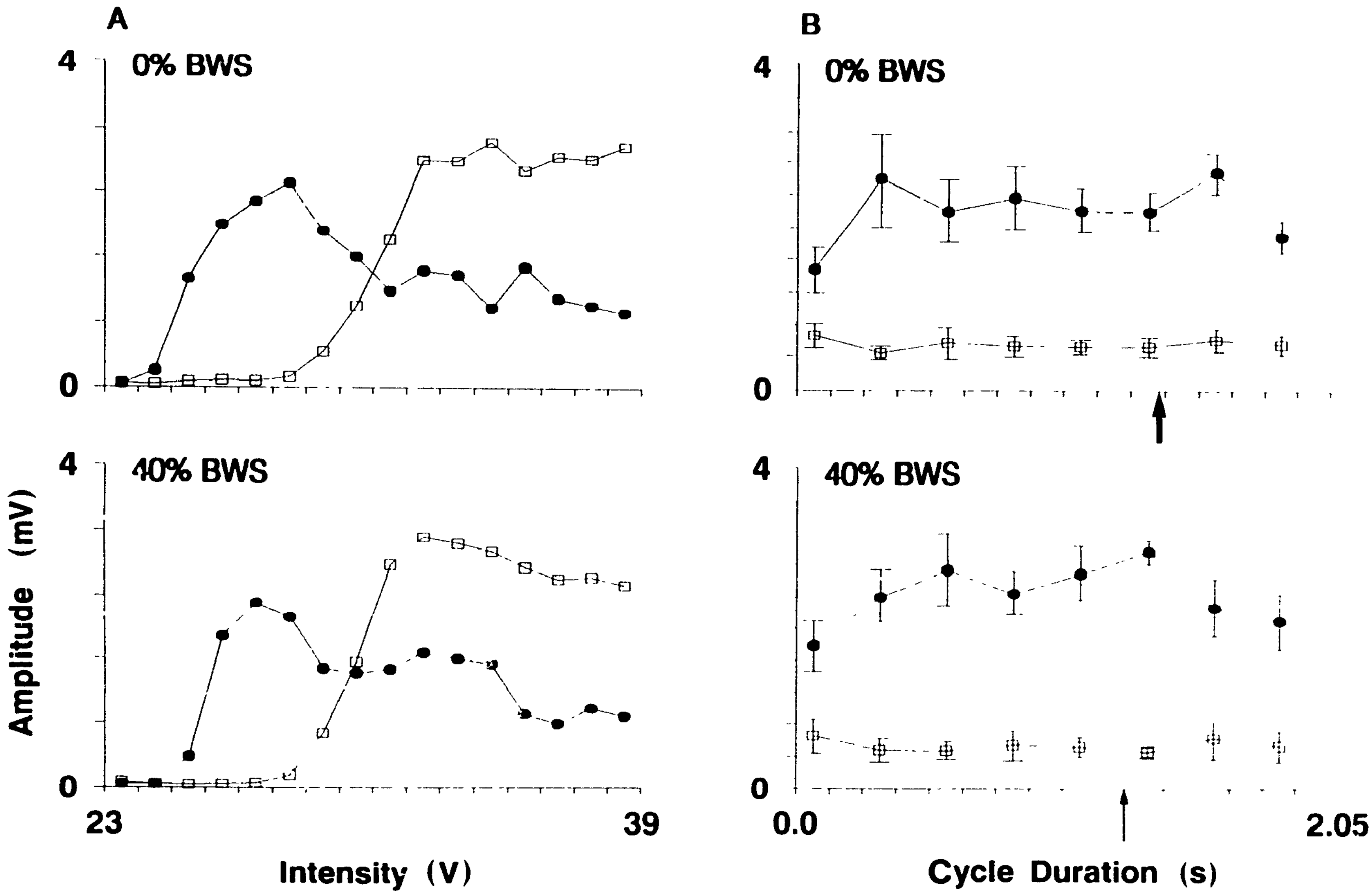
Kinematically RM did not show much change when 40% BWS was provided, as compared to the kinematic data measured under the 0% BWS condition. This was not surprising, as his kinematic data were quite similar to normal, and as it was shown in normals, joint angular displacements are not altered a great deal by the provision of 40% BWS. In figure 4.16 it can be seen that the degree of trunk extension prior to foot-floor contact, which was used to assist in propelling the swinging leg forward, was maintained even with 40% BWS. With BWS, the change at the hip joint was both a loss of a small degree of flexion at foot-floor contact and of extension in late stance. The effect of 40% BWS at the knee was to decrease flexion at foot-floor contact and to reduce the degree of flexion in swing; however, the hyperextension present at midstance remained. With BWS there was a less abrupt plantarflexion at the ankle at push-off, with a slight dorsiflexion prior to foot-floor contact, which never exceeded 0 degrees. The abnormal degree of plantarflexion during swing was due to the lack of foot-floor clearance, resulting in toe-drag.

The data from the fourth spastic subject (AN) is displayed in figures 4.17, 4.18, and 4.19. The H- and M-wave recruitment curves, as displayed in figure 4.17A, are

Figure 4.17. Comparison of H-reflex data recorded from one spastic paretic patient (AN) under the two experimental conditions, 0% and 40% BWS.

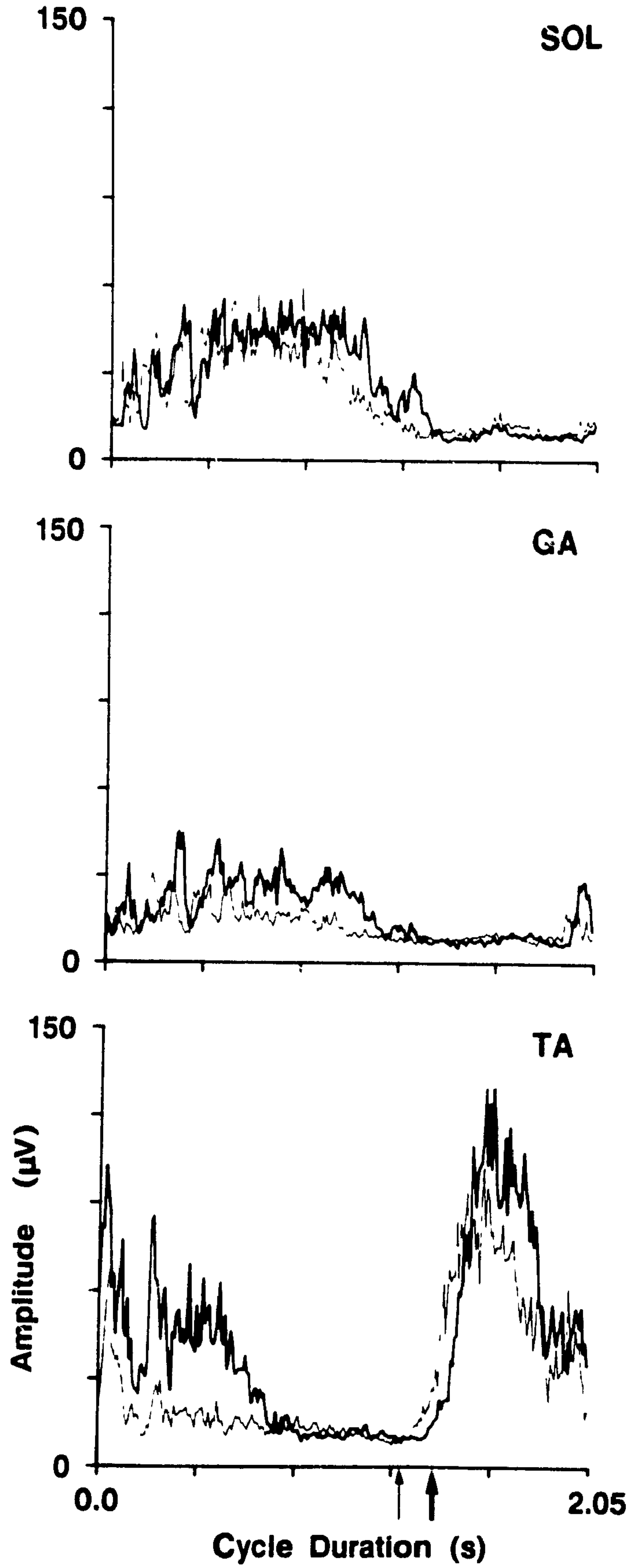
A: Soleus H-reflex (filled circles) and M-wave (open squares) recruitment curves, each point represents one stimulation.

B: The modulation pattern of the soleus H-reflex (filled circles, mean ± sd, $n \geq 10$) during an average gait cycle, with the M-wave (open squares, mean ± sd, $n \geq 10$) kept constant. The arrow denotes mean occurrence of swing in the average gait cycle: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS.



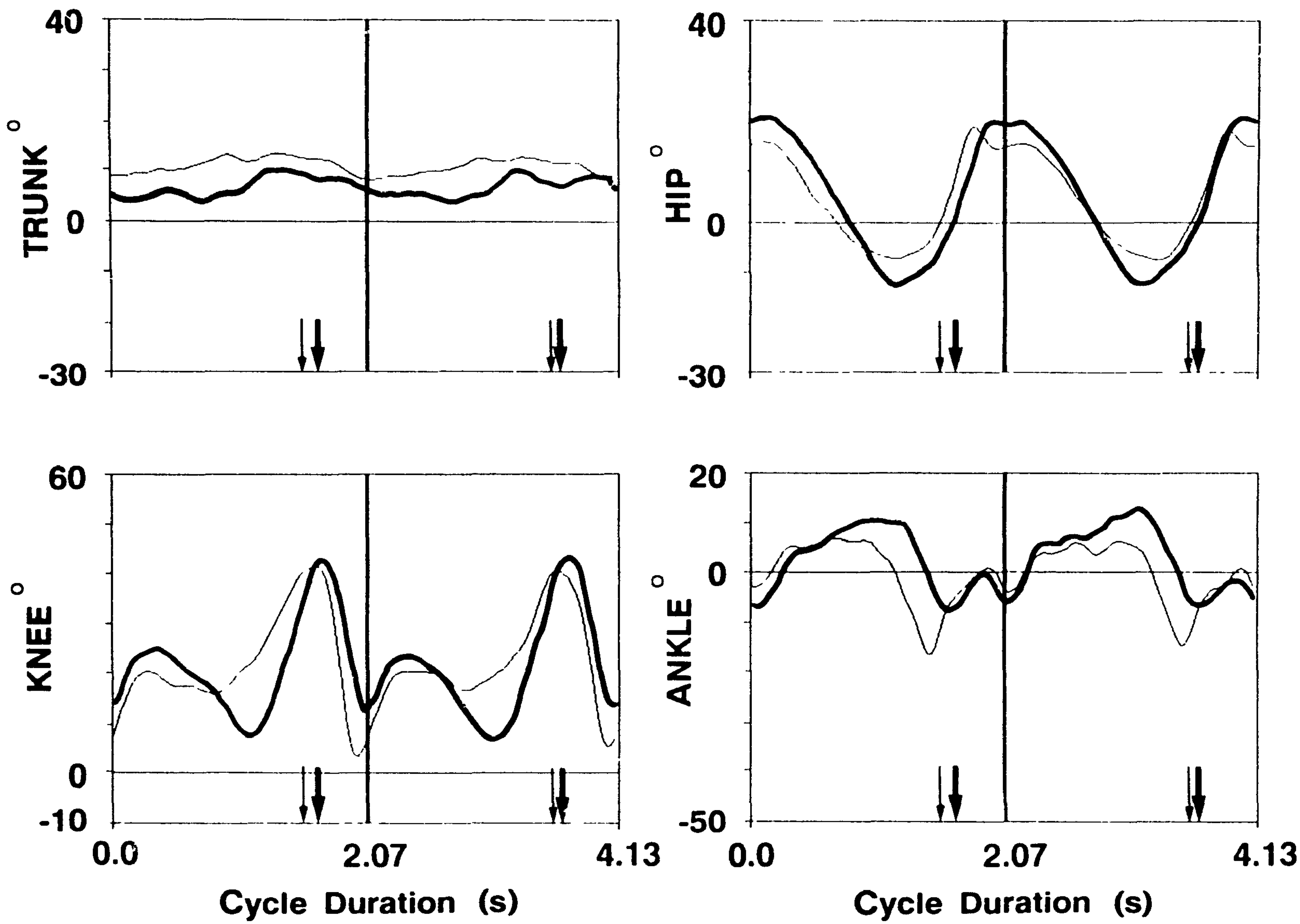
A
B
0% BWS
40% BWS
0% BWS
40% BWS
4
0
Amplitude (mV)
23
39
Intensity (V)
0.0
2.05
Cycle Duration (s)

Figure 4.18. Comparison of EMG mean ensemble averages (across 10 strides) of the soleus (SOL), medial gastrocnemius (GA), and tibialis anterior (TA) muscles of the tested leg of one spastic paretic patient (AN), normalized to the average gait cycle, under the two walking conditions, 0% and 40% BWS. The EMG activity recorded at 0% BWS is depicted by the heavy line and that recorded at 40% BWS is depicted by the light line. The arrows denote stance-swing transition: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS.



150
SOL
0
150
GA
0
150
TA
Amplitude (µV)
0
0.0
2.05
Cycle Duration (s)

Figure 4.19. Comparison of the sagittal angular excursions of two consecutive gait cycles (expressed in real time) for the trunk, hip, knee, and ankle of one spastic paretic patient (AN) under the two walking conditions, 0% and 40% BWS. The angular excursions recorded at 0% BWS are depicted by the heavy line and those recorded at 40% BWS are depicted by the light line. The heavy vertical line depicts termination of the first gait cycle, which is the same under the two conditions. The arrows denote the stance-swing transition: the heavy arrow that at 0% BWS, the light arrow that at 40% BWS. The positive angular displacement of the trunk indicates that it was maintained in a flexed position with respect to the vertical, during the two gait cycles. The positive and negative angular displacements of the hip indicate flexion and extension, respectively. The positive angular displacement of the knee indicates the degree of flexion at this joint. The positive and negative angular displacements at the ankle indicate dorsiflexion and plantarflexion, respectively.


40
TRUNK °
0
-30
40
HIP °
0
-30
60
KNEE °
0
-10
0.0
2.07
4.13
Cycle Duration (s)
20
ANKLE °
0
-50
0.0
2.07
4.13
Cycle Duration (s)

near normal; however, once again the lack of total inhibition of the H-reflex was evident. With 40% BWS there was little change in the recruitment curves except for a slight decrease in the peak H-reflex amplitude elicited.

The soleus H-reflex modulation during gait was pretty much absent; the H-reflex amplitude remained fairly stable in all phases of gait except for early stance (phase one), where it was somewhat decreased. BWS had little effect on the H-reflex amplitude modulation, except to cause a slight increase in amplitude in the stance phases. The EMG ensemble average muscle profiles are depicted in figure 4.18, and it can be seen that the soleus and GA EMG activity was smaller than normal in amplitude, with the presence of some clonus in early stance during FWB. The activity in the TA was quite strong and the first burst was abnormally long, leading to co-activation with the triceps surae. With BWS, the mean EMG amplitude was hardly decreased in the soleus muscle, but was somewhat decreased in the GA, and more so in the TA. In addition, the first TA muscle burst was decreased in duration, so that the co-activation with the soleus was replaced by an alternate pattern. However, clonus was still present in the soleus and GA with BWS.

The kinematic data of patient AN during walking is displayed in figure 4.19. With 40% BWS there was more flexion in the trunk angle and a reduction in the degree of extension at the hip joint. At the knee there was more extension at heel strike and less yield during early stance with BWS. Under both conditions, the knee remained more flexed than normal. At FWB, the ankle joint lacked a proper degree of plantarflexion at push-off, which was improved under the 40% BWS condition, more as a result of the presence of toe-drag in early swing. With BWS, there was slightly less dorsiflexion at midstance as well. In patient AN the lack of H-reflex modulation during the step cycle with 40% BWS was surprising, because of the fact that the EMG profiles and kinematic data were fairly close to normal.

CHAPTER 5: DISCUSSION

5.1 Technical Considerations

As has been reported in the literature (Crayton & King, 1981), the soleus H-reflex, elicited by electrical stimulation of the posterior tibial nerve, was found to be stable across time. This was evident in the data illustrated of the one representative normal subject, who was examined twice over a period of two months, under the two experimental conditions, standing and walking. This provides confidence that the H-reflex is a stable measure, which therefore can be used to evaluate the outcome of different therapeutic treatments or training strategies.

The consistent stability of the M-wave amplitude data, for each normal and SCI subject, was an important set of results, as the M-wave amplitude is a means of determining the effectiveness of the stimulus strength, and thus was successfully maintained as constant as possible between conditions and trials in this study. Therefore, changes in the H-reflex amplitude, as recorded in the subjects, cannot be accorded to changes in stimulus intensity.

5.2 Effect of BWS Upon Normal Subjects and Patients During Standing

Based on our results it is clear that BWS has very little effect in changing the amplitude of the H-reflex in the standing position of normal subjects or of patients, pooled as two separate groups. This is not surprising, as in the two standing conditions the ankle position was the same (90 degrees with respect to the lower leg) and the EMG activity in the muscles of the tested lower leg was not significantly different. The only difference between the two conditions was the degree of weight supported by the lower limbs. Thus, in this static postural task of standing, it can be stated that 40% BWS does not significantly modify the excitability of the soleus motoneuron pool, as measured by the H-reflex, in either the normal group or the patient group.

The amplitude of the soleus H-reflex can be modulated in the standing position, with changes in mean soleus EMG activity, as was shown in four of the normal subjects. The amplitude of the H-reflex increased with increasing levels of maximal isometric soleus contraction. We observed this trend, using the mean of ten stimulations at each level of contraction, which is similar to that reported by Capaday and Stein (1986), who used only one stimulation at each of four levels of soleus EMG contraction and allowed their subjects to stand on their toes, thus changing the ankle position, which in our study was kept constant. However, in contrast with what has been reported previously, the H-reflex amplitude was not always closely related to the EMG produced. This was evident in the first three levels of maximal contraction (0%, 20%, 40%), in which the H-reflex in some cases decreased in amplitude. ES was the only subject whose H-reflex amplitude dropped with increased background EMG activity, the reason for which is not immediately apparent. Therefore, the relative contribution of ankle position and degree of soleus EMG activity upon the H-reflex modulation needs to be further investigated.

5.3 Effect of BWS on Normals During Walking

That the H-reflex amplitude modulation is task specific, as hypothesized by Stein and Capaday (1988), is clear from the difference in its modulation during an active task such as walking. The soleus H-reflex modulation pattern, as measured in the normal subjects, was similar to the results that have been previously published (Capaday & Stein, 1986; Crenna & Frigo, 1987; Garrett et al, 1984). This normal H-reflex amplitude modulation during the step cycle is characterized by a decrease in H-reflex amplitude during early stance, followed by a progressive increase in reflex amplitude throughout stance, with a peak amplitude at the push-off phase and inhibition of the H-reflex throughout swing. It is interesting to note that, in most cases, despite a significantly higher ($p<0.05$) level of soleus EMG activity in early stance (phase 1), as compared to the soleus EMG activity in quiet standing, the H-reflex is much smaller in amplitude. Thus, despite a slight difference in soleus EMG activity between standing and the early stance phase in walking, the H-reflex amplitude is substantially different in the two tasks, and therefore cannot be attributed to the difference in soleus EMG activity alone. The possible neural mechanisms for this marked reduction of the H-reflex amplitude during the early stance phase of the gait cycle include: (i) the presence of presynaptic gating of the

Ia afferents, as hypothesized by Capaday and Stein (1987a), (ii) the activation of fast Ib afferents which would gate the autogenetic excitation from the triceps surae, as it is stretched at heel contact (Morin et al, 1982), and (iii) reciprocal inhibition of the soleus by the TA, as the TA contracts to gently lower the foot after heel contact.

During later stance the H-reflex was found to be maximally facilitated, which could be explained by one or more of the following: the near-maximal motoneuron depolarization caused by the descending central command, increased firing of the muscle spindles, and oligosynaptic inputs originating from cutaneous receptors and stretch of intrinsic foot muscles (Burke et al, 1984; Crenna & Frigo, 1987). Then in swing, the inhibition of the H-reflex may be explained by descending facilitation of Renshaw cell activity exerted on the soleus motoneurons, which is facilitated during the TA contraction to dorsiflex the foot, and would thus counteract the Ia effects of stretch of the soleus during the TA contraction (Katz & Pierrot-Deseilligny, 1984; Pierrot-Deseilligny et al, 1977). In addition, the soleus motor units are being de-recruited at the end of stance and there may be reciprocal inhibition exerted on the soleus motoneuron pool, as a result of the TA contraction, that would add to the inhibition of the H-reflex during swing.

It was found that providing 40% BWS during walking had no significant effect upon the H-reflex modulation pattern in the normal subjects, except for a slight decrease in H-reflex amplitude in the push-off phase. This may be attributed to the decrease in soleus EMG activity in later stance, as a result of the smaller push-off force required, since the load to be moved upwards was 40% less. The ensemble average EMG profiles of the soleus, GA and TA under the FWB condition were comparable to the typical patterns reported in the literature (Dubo et al, 1976; Knutsson & Richards, 1979; Winter, 1987). Providing BWS had the effect of decreasing the mean burst of activity of these three muscles, without altering the timing of the muscular activity, which confirms previous results reported in normal subjects (Finch, 1986). In addition, the normal angular displacements at the four joints studied, as recorded in the representative normal subject walking at FWB, which are similar to those found in the literature (Murray et al, 1966; Winter, 1987), were minimally modified by BWS. BWS had the effect of reducing the maximum extension and/or flexion that occurred at these joints, to a small degree, due to the combination of weight support and harness restraints.

5.4 The H-reflex Modulation and Locomotor Pattern in Patients During Walking

The patients studied in this research project represented a somewhat diverse group of incomplete spinal cord injured subjects, but they all showed an abnormal H-reflex amplitude modulation during gait. During walking, the EMG activity recorded in the lower leg for each patient revealed that in all cases the reciprocal contraction of the soleus and TA muscles was maintained to a large degree. In some patients (eg. RM) the first TA burst was absent, and foot-floor contact was made with the ball of the foot. As described in the literature (Benecke & Conrad, 1986; Conrad et al, 1985; Dietz et al, 1981) spastic gait is characterized by: (i) knee and ankle joint flexion during foot-floor contact, such that often the ball makes contact with the ground first, (ii) more dorsiflexion of the foot than normal throughout stance which causes the triceps surae to be stretched, and (iii) prolongation of the EMG activity of the lower limb muscles, such as soleus and GA, with flattening of the amplitude profile, attributed in part to exaggerated stretch reflexes during muscle lengthening. These characteristics were present to some degree in all of the patients studied here, depending on the severity of the deficits caused by the spinal cord injury. One finding in common in all patients was a decreased, but prolonged, soleus EMG activity, with premature recruitment and delayed muscle relaxation, similar to what Conrad et al (1985) reported in their spastic subjects. This was in contrast with normals, who exhibited a shorter period of soleus muscle activation with a gradual increase in activity followed by peak activity at push-off.

During walking, the patients pooled as a group showed an abnormally elevated H-reflex amplitude in both the early stance phase and the swing phase, with its amplitude in the push-off phase being slightly elevated, as compared to the normal amplitude. Even when the mean M-wave amplitude is similar between normals and patients, as was made possible by selecting a subset of four of the normals whose mean M-response was comparable to the patient group, this cannot explain the lack of phasic H-reflex modulation observed in the patients.

The soleus EMG activity alone is not closely related to the H-reflex amplitude modulation in patients. For example, during FWB walking, RM had early stretch activation of the soleus followed by clonus; but surprisingly the H-reflex amplitude was reduced in early stance, as it is in normals, who have very little soleus EMG activity in this phase. Again in swing, RM's H-reflex was suppressed, despite a burst of activity in the soleus. On the other hand, patient AN, who had much less soleus EMG activity in early stance at FWB, along with some clonus, had an abnormally elevated H-reflex amplitude. This high H-reflex amplitude was present in AN throughout the step cycle, even during swing when the soleus muscle was silent. Therefore, in the spastic paretic subjects studied here, there may be a disruption in the normal gating mechanism of the Ia afferent input, at the level of the soleus motoneuron pool.

This data confirms the results obtained in an extensive study, (Yang et al, 1990), of 21 spastic paretic patients that we conducted in collaboration with Dr. Stein's laboratory in Edmonton. In these 21 patients we found varying levels of H-reflex modulation during walking; an example of each was presented in the results section of this study to illustrate the spectrum of H-reflex modulation patterns. Again, a minimal relationship between the H-reflex amplitude and the EMG activity was observed in the patients. The mechanisms responsible for these hyperactive reflexes are not completely known, but may be due to one or a combination of the following: (i) a decrease in presynaptic inhibition of Ia afferent inputs (Burke & Ashby, 1972; Iles & Roberts, 1986), (ii) denervation supersensitivity, following injury to the spinal cord (Barbeau et al, 1981; Nygren et al, 1974), (iii) alteration in reciprocal inhibition (Ashby & Wiens, 1989) or lack of reciprocal inhibition due to weakness of the TA muscle, or (iv) a lack of descending facilitation of Renshaw cell inhibition upon soleus motoneurons, as is postulated to occur in intact humans (Katz & Pierrot-Deseilligny, 1984).

Another factor which contributes to the lack of H-reflex modulation, during gait in these patients, is the abnormal position of the lower limb, and more particularly of the ankle, in the different phases. It has been determined in previous studies that passive dorsiflexion of the ankle produced a reduction in the amplitude of the H-reflex; conversely passive plantarflexion produced an elevation of the H-reflex amplitude (Gerilovsky et al, 1986; Robinson et al, 1982). Thus, the excitability of the soleus motoneuron pool is markedly affected by the position of the ankle joint and the resulting stretch of the soleus muscle. Robinson et al (1982) postulated that the alteration in the H-reflex amplitude, produced by differing ankle positions, was due to the intramuscular receptors, either the muscle spindles or golgi tendon organs. However, these authors also mentioned that presynaptic inhibition may be involved in the diminishment of the H-reflex during muscle stretch. Of course, during walking, the limb position is related to the EMG activity occurring in that limb's muscles, and thus the relative importance of either upon the H-reflex modulation is difficult to distinguish. However, the joint angular displacements, and the proprioceptive information it conveys, may be more important in its effect upon the motoneuron pool than has been previously thought. Preliminary results in our lab show that the static position of the lower limb, specifically positions which reproduce foot-floor contact, midstance, and midswing as they occur during walking, can modulate the H-reflex amplitude in patients in whom there was no reflex modulation during gait. In one severely spastic patient, the H-reflex was inhibited to 50% of its control value when the tested lower limb was held in the swing position, which was never observed at this phase during walking (unpublished results). Therefore, the importance of the changing joint angles upon the H-reflex modulation during walking requires further investigation.

5.6 Effect of BWS on Patients During Walking

That the mechanisms responsible for the phase dependent modulation of the H-reflex remain somewhat intact, in individual SCI patients, is evident in view of the fact that it was present to some degree in five of the patients studied here. In the patients in which there was some modulation of the H-reflex during gait, 40% BWS had the effect of modifying this modulation, such that it became more normal. In the patients pooled as a group it was found that there was a significant interaction effect between the phase in the gait cycle and the level of BWS given, upon the H-reflex amplitude. This interaction was due predominantly to the decrease in the mean H-reflex amplitude at the push-off phase under the 40% BWS condition, which was, most likely, a result of the decrease in peak amplitude of the soleus muscle. In addition, BWS caused a decrease in the H-reflex amplitude during midstance and a more normal inhibition of the H-reflex in swing, with the exception of phase 15 just prior to heel strike, in which the H-reflex was elevated (eg.LR, RM). That BWS had the effect of reducing EMG amplitude in these lower leg muscles and of inducing more approriate EMG activity timing in relation to the step cycle, is consistent with what is reported in the literature (Visintin & Barbeau, 1989; Visintin & Barbeau, 1990). Thus, in these two patients, who already had a near normal kinematic pattern during walking, 40% BWS had a minimal effect, as observed in the normal subjects, except for a small change in H-reflex modulation, which could be a result of a possible decrease in the reflex gain at the motoneuron level.

In two patients (CH, SH) who had little or no H reflex amplitude modulation during locomotion. the provision of BWS also caused a decrease in the gain of the H-reflex such that the amplitude was lower, and in one patient (AN) caused a slight increase in the reflex amplitude. However, in these three patients, BWS was not able to induce a phasic modulation of the H-reflex, despite the effect BWS had in decreasing the EMG mean burst amplitude, improving the EMG activity timing, and improving the joint angular displacements. Worth noting is the fact that two of these patients (CH, AN), who lacked H-reflex modulation, had the greatest improvement in their ankle angular excursions with BWS. Therefore, in agreement with what Yang et al (1990) proposed, it is unclear whether in these patients the mechanisms that produce phasic modulation of the H-reflex are intact, but hidden by the high reflex gain, or whether the mechanisms that produce the H-reflex modulation are indeed impaired.

Certainly BWS does have a significant effect upon kinematic and temporal distance data. This second component which has not yet been dealt with, should be mentioned. In the patient group the cycle duration was two to five times greater than was the case in the normals ($p < 0.05$, indicating difficulty in coping with faster walking speeds. The percentage of double support time was two times greater in patients, but was brought into a more normal range with 40% BWS. BWS had a significant effect of reducing the percentage of stance during the gait cycle and of increasing the percentage of swing. In four of the eight patients, the prolonged stance phase measured at FWB approached normal values at 40% BWS. These results confirm those reported by Visintin and Barbeau (1989), who examined the effects of different levels of BWS upon the gait patterns of seven spastic paretic subjects.

CHAPTER 6: CONCLUSIONS

This research project has investigated the modulation of the H-reflex amplitude in normal and spinal cord injured subjects, during full weight bearing standing and walking, as well as the effects of providing 40% BWS upon the H-reflex amplitude under these same conditions.

In this chapter, the overall conclusions of this study will be presented, along with the limitations of the study and an outline of future directions for related research.

6.1 Research Project Conclusions

First of all, it was determined that in normals and patients the H-reflex was modulated differently in the two conditions studied here, standing and walking. This modulation cannot be explained simply by the difference in soleus EMG activity alone, as a minimal relationship between the H-reflex amplitude and the soleus EMG activity was observed.

Secondly, the phasic modulation of the soleus H-reflex in normal subjects during walking, confirmed the results of previous studies. However, the mechanisms that cause this modulation during gait are still under investigation.

Thirdly, it was determined that SCI patients have an abnormally elevated H-reflex amplitude which, in most cases, is not phasically modulated during the gait cycle, as it is in normal subjects.

Fourthly, the therapeutic intervention of providing 40% BWS to these patients during locomotion was found to significantly improve their locomotor pattern, as determined by changes in EMG activity, kinematic data, and by subjective report. What is interesting is the fact that this improvement in the locomotor pattern, which was measurable in terms of functional outcome, was not reflected clearly by the modulation of the H-reflex, which indirectly measures the excitability level of the soleus motoneuron pool. Therefore, this study cannot determine whether the elevated H-reflex amplitude in these patients, during early stance and swing, was due to an abnormal gating mechanism or abnormal reflex gain. An understanding of these mechanisms is important to further validate the use of BWS as a gait retraining strategy for SCI patients.

6.2 Limitations of the Study

- The small sample of patients makes it difficult to derive conclusions that would be applicable to a larger population of SCI patients.

- The effects of fatigue upon the H-reflex amplitude were not completely controlled for, because, even with adequate rest periods, some patients were quite tired by the end of the experiment.

- Only one leg was studied in each subject, which limits the observations made to that one side and does not allow for comparisons between a more or less affected side, such as in asymmetrical gait.

6.3 Future Directions for Related Research

One important question that requires further investigation concerns the effect of ankle position on the H-reflex amplitude. This is, in fact, ongoing research in our lab, particularly in terms of H-reflex modulation produced by static positions which mimic those encountered during gait: foot-floor contact, midstance, and midswing. This would help to distinguish between the relative influence of ankle position upon the H-reflex during walking.

Further research into the relationship between lower leg EMG activity, kinematics, and H-reflex modulation during gait in patients is required to understand the mechanisms involved in the different abnormal gait patterns and how these locomotor patterns are improved with BWS.

As the H-reflex is considered a reliable measure of the neuronal excitability, it is important to study the effects that the administration of antispastic drugs, such as cyproheptadine, clonidine, and baclofen, have upon the H-reflex amplitude modulation in patients. It has already been established that these antispastic drugs produce a marked improvement in the locomotor pattern of SCI patients, and that SCI patients exhibit an abnormal H-reflex modulation during gait. Therefore it would be important to determine the effect of these drugs upon the H-reflex modulation pattern, in order to understand their mechanism of action.

Finally, it would be important to determine the effects that locomotor training, using BWS, has upon the H-reflex modulation pattern and upon measures of the locomotor pattern.

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APPENDICES

Figure 1A. Comparison of the temporal parameters of gait (mean ± sd, n=8-10 strides): cycle duration (CD), percentage of gait cycle spent in stance (% STA) and in swing (% SWI), and percentage of total double support time (% TDST), recorded for each normal subject (n=8) under the two walking conditions: walking with 0% BWS and walking with 40% BWS.

TEMPORAL DISTANCE PARAMETERS RECORDED FOR NORMAL SUBJECTS

0% BWS

Subject/	Cycle Duration $\bar{X}$	Cycle Duration SD	% Stance $\bar{X}$	% Stance SD	% Swing $\bar{X}$	% Swing SD	% TDST $\bar{X}$	% TDST SD
ES	1.146	.031	64.8	1.6	35.2	1.6	28.0	3.0
JF	1.073	.031	64.8	1.2	35.2	1.2	26.2	2.3
HB	1.358	.021	67.9	0.5	32.1	0.5	32.7	0.9
MB	1.350	.024	62.1	1.8	37.9	1.8	23.4	2.7
KN	1.245	.024	62.4	2.0	37.6	2.0	23.8	2.9
JB	1.290	.000	61.6	0.0	38.4	0.0	20.9	0.0
AP	1.277	.026	64.1	2.0	35.9	2.0	22.5	2.3
NP	1.455	.053	65.6	2.4	34.4	2.4	29.4	5.9
$\bar{X}$ =	1.274	.122	64.2	2.1	35.8	2.1	25.9	3.97

40% BWS

Subject/	Cycle Duration $\bar{X}$	Cycle Duration SD	% Stance $\bar{X}$	% Stance SD	% Swing $\bar{X}$	% Swing SD	% TDST $\bar{X}$	% TDST SD
ES	1.152	.029	60.2	2.0	39.8	2.0	16.3	3.3
JF	1.403	.021	68.8	0.6	31.2	0.6	34.5	1.8
HB	1.307	.029	66.1	1.3	33.9	1.3	29.8	2.5
MB	1.335	.028	59.4	1.7	40.6	1.7	17.0	3.5
KN	1.238	.038	59.2	1.7	40.8	1.7	17.4	3.1
JB	1.305	.077	57.5	1.9	42.5	1.9	13.3	2.9
AP	1.298	.021	59.8	2.0	40.2	2.0	14.8	3.3
NP	1.455	.042	60.8	1.4	39.2	1.4	19.0	2.7
$\bar{X}$ =	1.312	.093	61.5	3.9	38.5	3.9	20.3	7.6

Figure 1B. Comparison of the temporal parameters of gait (mean ± sd, n=8-10 strides): cycle duration (CD), percentage of gait cycle spent in stance (% STA) and in swing (% SWI), and percentage of total double support time (% TDST), recorded for each patient (n=8) under the two walking conditions: walking with 0% BWS and walking with 40% BWS.

TEMPORAL DISTANCE PARAMETERS RECORDED FOR PATIENTS

0% BWS

Subject/	Cycle Duration $\bar{X}$	Cycle Duration SD	% Stance $\bar{X}$	% Stance SD	% Swing $\bar{X}$	% Swing SD	% TDST $\bar{X}$	% TDST SD
CH	3.288	.406	84.1	1.5	15.9	1.5	56.1	4.6
LR	2.694	.110	82.5	2.1	17.5	2.1	53.3	4.4
RM	3.036	.060	72.6	1.1	27.4	1.1	38.5	1.6
AN	2.010	.049	72.0	2.4	28.0	2.4	44.3	2.3
MB	2.016	.034	72.2	1.6	27.8	1.6	43.3	2.5
BP	1.656	.091	71.6	1.6	28.4	1.6	44.7	4.0
RG	2.562	.101	76.0	1.6	24.0	1.6	30.8	4.6
SH	6.588	.273	67.2	2.2	32.8	2.2	42.2	4.2
$\bar{X}$ =	2.981	1.56	74.8	5.8	25.2	5.8	44.2	7.9

40% BWS

Subject/	Cycle Duration $\bar{X}$	Cycle Duration SD	% Stance $\bar{X}$	% Stance SD	% Swing $\bar{X}$	% Swing SD	% TDST $\bar{X}$	% TDST SD
CH	3.330	.366	71.6	2.8	28.4	2.8	41.7	6.9
LR	2.676	.077	79.9	2.4	20.1	2.4	47.8	4.8
RM	2.616	.131	63.6	2.3	36.4	2.3	30.0	3.4
AN	2.076	.052	61.7	4.5	38.3	4.5	23.3	4.6
MB	2.01	.028	70.9	1.6	29.1	1.6	39.8	2.1
BP	1.734	.042	61.7	1.6	38.3	1.6	24.4	3.4
RG	3.128	.094	71.0	4.3	29.0	4.3	35.7	3.4
SH	6.470	.859	66.1	4.5	33.9	4.5	28.1	11.5
$\bar{X}$ =	3.005	1.50	68.3	6.2	31.7	6.2	33.8	8.8

McGill University **A - 2A**

School of Physical and Occupational Therapy

Consent to Participate in a Research Study on the Influence of Body Weight Support on Soleus H-Reflex Modulation During Standing and Walking

I, ____________________________, consent to participate in a research study on the influence of body weight support (BWS) on soleus H-reflex modulation during standing and walking.

Purpose and Design of the Study

Spinal reflexes are important elements in the control of posture and locomotion, the disruption of which can lead to spasticity and impairment in the locomotor pattern. The purpose of this experiment is to investigate the influence of suspension in a BWS harness on soleus H-reflex modulation during standing and walking in normal and spinal cord injured (SCI) subjects.

The H-reflex is obtained by delivering an electrical stimulus to the back of the knee. The electrical signals of different lower limb muscles will be recorded using disposable electromyographic (EMG) electrodes, after the skin has been shaved and abraded.

I am expected to come for one orientation session lasting ½ hour and one experimental session lasting approximately 4 hours, including time spent on skin preparation. In the orientation session, the H-reflex recruitment curve will be obtained and I will become familiar with the electrical stimulation. In the experimental session I will be suspended in a BWS harness above a treadmill belt with 0% and 40% of my body weight supported mechanically. In the standing condition, I will receive two sets of 10 stimulations every 10 seconds under the two BWS conditions. In the walking condition the speed of the treadmill belt will be gradually increased from 0.1 m/s to the comfortable walking speed and I will receive eight sets of 10 stimulations for each of two trials (one at 0% BWS and one at 40% BWS). Adequate rest will be given between trials. During walking I am equipped with an emergency switch with which I can stop the treadmill at any time.

Disadvantages of Participation in the Study

A minor disadvantage is the discomfort experienced by being supported in the BWS harness while walking. There may be slight discomfort experienced as a result of the electrical stimulation. There is a possibility of slight skin irritation at the electrode sites.

Advantages of Participation in the Study

Although there are no personal benefits to be gained from participating in this study, the results from this research will help in our understanding of: (i) the role BWS plays in improving the locomotor pattern of spinal cord injured patients, and (ii) the use of BWS as a gait training strategy.

Inquiries Concerning the Study

I understand that any questions I may have about this study will be answered by Mr. Robert Blunt (398-6664) or Dr. Hugues Barbeau (398-4519).

Withdrawal from the Study

I understand that my participation in this research project is voluntary, and that I may withdraw at any time without prejudice to myself.

Dated, the ________ day of ________________, 19_____.

Signed by ______________________________________.

Witnessed by __________________________________.

I ____________________________, certify that I have explained to the above mentioned subject the nature of the research study and the known risks involved in participating in the study, and that the subject has the option of withdrawing from the study at any time.

I have assured him/her that although results of the study will be published , his/her identity will be held in confidence.

Signed ____________________________.

Dated ____________________________.

Université McGill
École de Physiothérapie et d'Ergothérapie

Formule de Consentement Clinique sur l'Effet d'une Diminution du Poids du Corps sur la Modulation du Réflexe-H dans la Soléaire Durant la Position Debout et Durant la Marche

Je soussigné(e), ______________________________, consens à participer à un projet de recherche sur l'effet d'une diminution du poids de corps sur la modulation du réflexe-H évoqué dans la soléaire durant la position debout et durant la marche.

But et Protocole de l'Étude

Le but de cette étude est de permettre d'une part de mieux comprendre l'effet d'une diminution du poids de corps sur la modulation du réflexe-H, et d'autre part de comparer ces données à celles obtenues chez des patients ayant une lésion de la moelle épinière.

Le réflexe-H sera mesuré en stimulant électriquement la partie postérieure du genou. Suivant une préparation adéquate de la peau (rasage et nettoyage), l'activité électromyographique (EMG) de différents muscles d'un membre inférieur sera enregistrée à partir d'électrode de surface.

Je participerai à une session d'orientation, qui durera une demi heure, et une session expérimentale, qui durera environ 4 heures, incluant le temps de préparation. Les deux sessions seront effectuées à l'école de physiothérapie et d'ergothérapie de l'université McGill.

Dans la session d'orientation la courbe de recrutement réflexe-H sera obtenue et je vais me familiariser aux stimulations électriques. Dans la session expérimentale, je serai suspendu dans un harnais au-dessus d'un tapis roulant avec 0% et 40% du poids de corps supporté par un système mécanique. Pendant la position debout, je recevrai deux séries de dix stimulations pour chaque condition de support de poids. Au début de chaque essai de marche, le tapis roulant sera augmenté graduellement de 0.1 m/s pour atteindre une vitesse confortable. Pendant la marche, je serai suspendu dans le harnais avec 0% et 40% du poids de corps supporté par la système mécanique et je recevrai huit groupes de dix stimulations pour chaque de deux essais (un à 0% de support et un à 40% de support). Après chaque essai, il y aura une période de repos. Toutes les mesures de sécurité (bouton de controle de vitesse et bouton d'arrêt) m'ont été clairement expliquées.

Désavantages de la Participation à cette Étude

Un léger inconvénient de cette étude est la suspension dans le système de support de poids pendant la marche. La stimulation électrique peut être légèrement inconfortable. Une légère irritation de la peau sous l'électrode peut être également possible.

Avantages de la Participation à cette Étude

Il n'y a aucun avantage direct à participer à cette étude. Nous espérons que les informations pourront nous servir à mieux comprendre les mécanismes qui améliorent le patron locomoteur chez les patients ayant une lésion de la moelle épinière. Il est possible que ces informations puissent nous aider à développer une nouvelle stratégie de réadaptation pour cette population cible.

Renseignements Concernant cette Étude

Je comprends que toutes informations supplémentaires que je voudrais obtenir concernant cette étude me seront fournies sur le demande en contactant M. Robert Blunt (398-6664) ou Dr. Hugues Barbeau (398-4519).

Retrait de l'Étude

Je comprends que ma participation à cette étude est tout à fait volontaire et que je peux la retirer à tout moment sans qu'il me soit porté préjudice.

Daté, le ________ jour de ____________________, 19____.

Signé par ____.

Témoin ____.

Je, ____________________________, soussigné, certifie que j'ai expliqué au sujet mentionné plus haut, la nature de l'étude projetée et que celui-ci peut se retirer de l'étude à tout moment.

Signé par ______________________________.

Je comprends que certaines informations peuvent faire l'objet de publications scientifiques, cependant mon anonymat sera respecté en tout temps.

Signé par ______________________________.

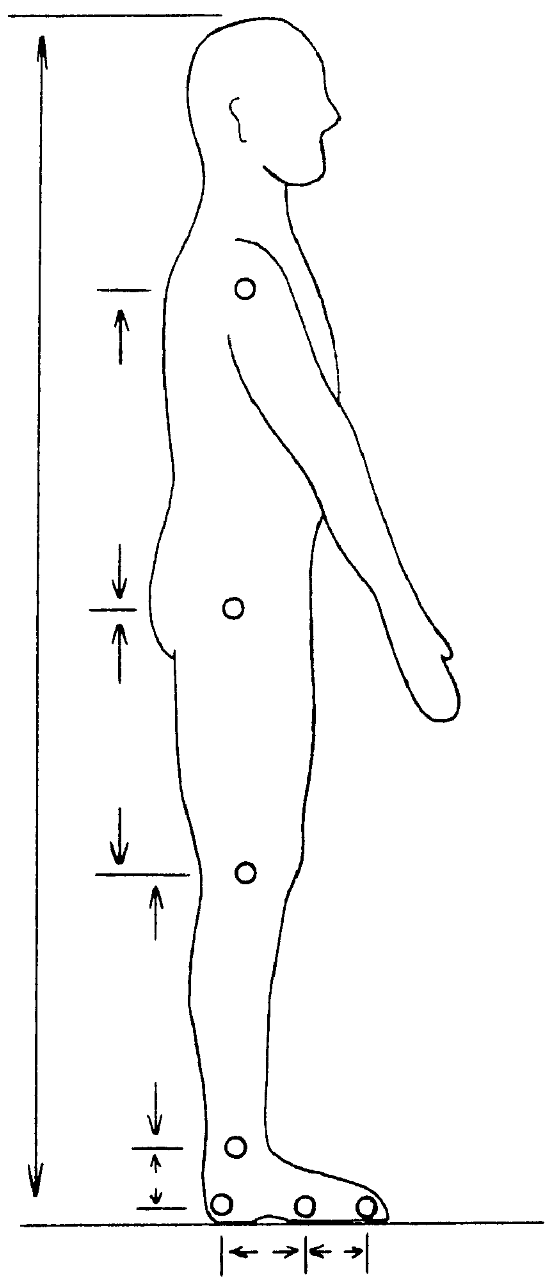
Date __________________________________.

Anatomical Location of Joint Markers and Anthropometric Data

Code No.:___ Subject/Patient Anthropometric Data Sheet

Name: ________________ Date:________

Height:____ Weight:____ Age:___ Sex:___



Anatomical Location of Markers

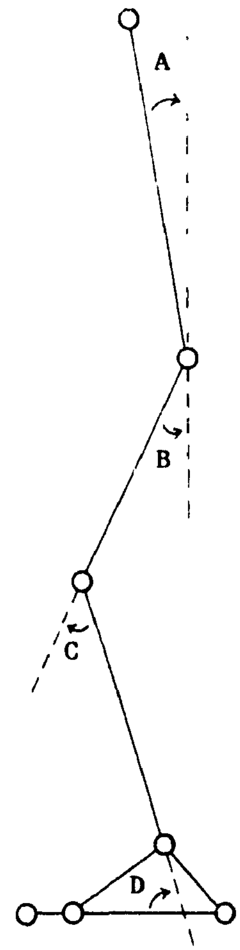
Shoulder	- just distal to the acromion process
Hip	- greater trochanter
Knee	- lateral femoral epicondyle (about 2 cm above knee line)
Ankle	- lateral malleolus of fibula
Heel	- about 2 cm above ground in line with rear of shoe
Meta-tarsal	- 5th metatarsal phalangeal joint
Toe	- about 2 cm above sole in line with front of shoe

A: Trunk Angle

B: Hip Angle

C: Knee Angle

D: Ankle Angle



Stick figure representing the angles that were calculated for each of the joints analysed.

Angular displacements at the trunk and hip are indicated as follows:
zero indicates vertical position of the trunk or thigh and flexion is indicated by positive displacement.

Knee joint displacement is zero when the leg is fully extended and positive when flexed.

Ankle displacement is zero when the foot is at 90 degrees with respect to the leg; dorsiflexion is positive.