THE EFFECT OF BODY WEIGHT SUPPORT ON THE LOCOMOTOR PATTERN OF SPASTIC PARAPARETIC SUBJECTS WALKING ON A TREADMILL

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ABSTRACT

This study investigated the effects of providing body weight support (**BWS**) on the gait pattern of 15 incomplete spinal cord lesioned subjects. Electromyographic (**EMG**), joint angular displacement and temporal distance parameters were simultaneously recorded as subjects walked on a treadmill while 0% and 40% of their body weight was mechanically supported by an overhead harness. The effects of 0% and 40% BWS while walking with and without parallel bars and at different treadmill speeds was further investigated in a subgroup of 8 subjects.

In general, 40% BWS led to a decrease in prolonged EMG activity of proximal muscles, a decrease in premature activation of distal muscles as well as a decrease in clonus. Such changes in EMG activity were evident especially during more demanding external conditions such as walking without parallel bars or at faster treadmill speeds. A general decrease in EMG mean burst amplitude for lower limb muscles was also noted. Sagittal angular displacement profiles revealed a straighter trunk and knee alignment during initial and midstance. Forty percent BWS appeared to facilitate gait by allowing subjects to walk for longer periods and to walk at faster treadmill speeds. Increases in stride length, single limb support time and decreases in total double support time were also noted with 40% BWS at comparable treadmill speeds. The results this study suggest that BWS facilitates the expression of a more normal gait pattern.

ABREGE

Cette étude a pour but d'étudier l'effet d'un support de poids du corps sur le patron de marche chez quinze sujets ayant une lésion incomplète de la moelle épinière. L'activité électromyographique (EMG) des muscles des membres inférieurs, les données cinématiques et les paramètres du cycle de marche furent enregistrés simultanément pendant la marche sur tapis roulant 0% (poids total) et 40% du poids du corps supporté par un système mécanique avec harnais. Nous avons également évalué l'effet du poids du corps (0% et 40%) pendant la marche, avec ou sans barres parallèlles et à différentes vitesses chez 8 des 15 sujets.

En général, 40% du poids du corps produit une diminution dans la prolongation de la bouffée d'EMG des muscles proximaux, une diminution de l'activité prématurée des muscles distaux et une diminution du clonus. Ces changements dans le profil d'activation EMG furent notés particulièrement lors des conditions plus exigantes pendant la marche sans barres parallèlles à des vitesses plus élevées. Une diminution de l'amplitude maximale de la bouffée d'EMG des muscles des membres inférieurs fut observée. De plus, les données cinématiques ont révélé un meilleur alignement du tronc et du genou au début et au milieu de la phase d'appui. Une augmentation de la longueur du pas, de la durée de la phase de simple appui et une diminution de la période de double appui furent également observées à 40% du poids du corps supporté pour des vitesse comparables. Les sujets pouvaient marcher pour des périodes plus longues et à des vitesses comfortable plus élevées. Les résultats de cette étude suggèrent que cette nouvelle strategie d'entraînement locomoteur facilite une expression de marche plus normale.

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STATEMENT OF AUTHORSHIP

I certify that I am the primary author of all manuscripts contained in this thesis. I also claim full responsibility for the content and style of all texts included.

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CHAPTER ONE: INTRODUCTION

One of the primary goals in the rehabilitation of the neurological patient is to reeducate gait. Patients who have sustained an incomplete spinal cord lesion usually present with marked deficits in their ability to walk. The presence of spasticity, paresis, disturbances in muscle activation patterns, abnormal reflex activity and postural instability leads to significant gait abnormalities (Knutsson, 1980; Conrad et al, 1985; Benecke and Conrad, 1986; Barbeau et al, 1988). Spastic paretic patients are also characterized by their inability to adequately bear weight through their affected lower extremities during the loading phase of gait (Knutsson, 1983; Barbeau et al, 1988).

The gait disturbances present often make spinal cord lesioned patients poor functional walkers and in severe cases only a non functional level of ambulation is attained following rehabilitation (Young et al, 1982; Burke et al, 1985). Many patients will rely on their wheelchairs for mobility. Conventional treatment remains rather conservative consisting of strengthening the trunk and paretic limbs, bracing of the lower extremities and the usage of ambulatory aids to allow the patient to move his body forwards (Guttman, 1976; Peterson, 1985). A trial of antispastic drugs will frequently be used to decrease the spasticity which may be interfering with the locomotor pattern (Knutsson, 1983; Wainberg et al, 1986). Advances in functional electrical stimulation allow simulation of a primitive reciprocal gait through stimulation of paralyzed muscles (Badj et al, 1983). Nevertheless, the numerous problems encountered with this technique preclude it as a practical solution at the present time.

Based on the spinal animal model (Barbeau et al, 1987), a new gait training strategy, consisting of providing body weight support (BWS) during treadmill locomotion has been proposed to retrain gait following a spinal cord lesion (Finch and Barbeau, 1985; Barbeau et al, 1988). This alternative approach consists of symmetrically supporting a percentage of the patient's body weight, centrally at the trunk, by an overhead harness. The percentage BWS would be progressively decreased while retraining gait until the patient could ambulate while supporting his full body weight on his lower extremities. Decreasing the load on the lower extemities may facilitate interactive gait training resulting in a more normal gait pattern following training. Before implementing this approach as a gait training strategy for spinal cord injured patients, it is important to evaluate the effects of BWS on the different gait parameters in this patient population. The results of this study will determine the applicability of BWS during locomotor training in the spinal cord injured population.

1. OBJECTIVES OF THE STUDY

1.1 Main Objective:

The main objective of this research project is to investigate the effects of supporting a percentage of body weight; 0% (full weight bearing) and 40%, on the locomotor pattern of spastic, incomplete spinal cord lesioned subjects during treadmill locomotion.

1.2 Specific Objectives:

1. To evaluate the effects of 0% and 40% body weight support (**BWS**) on the electromyographic (**EMG**) profiles of prime flexor and extensor lower extremity muscles which are active during locomotion.

2. To determine the effects of 0% and 40% BWS on the sagittal angular displacement profiles at the trunk, hip, knee, and ankle during treadmill locomotion.

3. To investigate the influence of 0% and 40% BWS on the temporal distance parameters of treadmill locomotion. The specific temporal distance parameters evaluated and compared at each BWS level will be cycle duration, single limb support time, percentage stance, percentage double support time, stride length, and maximal comfortable treadmill speed.

4. To contrast the effects of external parameters such as walking with and without parallel bars, and increasing treadmill speeds on the EMG activity and sagittal angular displacement profiles of lower limb muscles during treadmill locomotion. The effects of 40% BWS on the above external parameters will be described.

CHAPTER TWO: LITERATURE REVIEW

2.1 Epidemiology

2.1.1 Incidence

A lesion to the spinal cord can result from either a traumatic episode (e.g. motor vehicle accident, diving injury, etc.) or a non-traumatic syndrome (e.g. multiple sclerosis, tumor, etc.). The incidence of spinal cord lesions in Canada and elsewhere has been increasing with an overwhelming majority resulting from traumatic incidents and affecting the younger population between the ages of 16 and 30.

Young et al (1982) reported 2769 incidences of paraplegia and 3110 incidences of quadriplegia over an eight year period in 17 Regional Model Spinal Cord Centers located throughout the United States. In 1988, 444 new cases of paraplegics and quadriplegics were reported to the Canadian Paraplegic Association (1988). Within the same year, 72 new cases of spinal cord lesioned individuals were registered at the Quebec Paraplegic Association (1988), increasing the total to 3357 registrants for the province of Quebec alone.

2.1.2 Ambulatory Status

The degree of disability resulting from a spinal cord injury depends on the level and the extent of the lesion. In a study by Young et al (1982) 20.3% of the paraplegic subjects and 16.4% of the quadriplegic subjects regained the ability to walk (a minimum of 50 yards) following the injury. In an epidemiological study, Burke et al (1985) reported that 63% of 262 spinal cord injured patients discharged from a hospital center were able to walk. Thirty-nine percent of these patients were functional walkers while 24% of them walked for exercise only. Neither of these studies described the gait pattern nor the types of walking aids used by the patients.

Hussey and Stauffer (1973) studied 164 spinal cord injured, ambulatory patients (ranging from independent community ambulators to wheelchair dependent ambulators) and correlated their level of ambulation to their motor and sensory examination. The authors defined 'motor power' as a key determinant in the level of functional ambulation achieved. Patients at the higher levels of ambulation generally had good pelvic control, active hip flexion and some voluntary contraction in their quadriceps muscles. Among the good walkers, the presence of spasticity decreased the level of ambulation from community to household ambulation. The need to brace the lower extremities, especially with bilateral long leg braces, was also a limiting factor in achieving the higher levels of ambulation.

2.2 Spastic Paraparetic Gait

Among incomplete spinal cord lesion patients, a considerable degree of gait variation exists, depending on the level and the extent of the lesion. The residual motor and sensory deficits coupled with the presence of spasticity, abnormal reflexes and impaired balance contribute to the gait deviations seen.

As there is relatively little documentation which specifically describes spastic paraparetic gait in terms of its electromyographic (EMG), kinematic and temporal distance parameters, it is necessary to also rely on literature based on spastic gait due to other pathologies. The following will summarize the findings related to gait patterns for various spastic neurological syndromes with emphasis placed on spastic paraparetic gait.

2.2.1 EMG Activity

Pioneering work by Marks and Hirschberg (1958) provided a qualitative description of the EMG activity in a group of hemiplegic patients walking overground. They reported an overall decrease in activity of the affected lower limb muscles. Peat et al (1976), following the analysis of EMG activity in 20 hemiplegic subjects during independent ambulation, reported a loss of the normal muscle phasing typically seen in lower limb muscles during gait. Instead, peak muscle activity occured concurrently at midstance during the support phase in the four muscles investigated.

Knutsson and Richards (1979) identified three types of abnormal muscle activation patterns which contributed to the gait disturbances seen in a group of hemiplegic patients. The type I pattern consisted of premature activation of the calf muscles during stance accompanied by a decrease in peak activity. The type II pattern was characterized by a significant decrease of EMG activity in 2 or more muscles. The type III pattern was described as abnormal co-activation of several lower limb muscles usually elicited during the stance phase while the limb was being loaded.

Hirschberg and Nathanson (1952) studied the EMG activity of lower extremity muscles during overground locomotion in 36 spastic paraparetic subjects due to multiple sclerosis. The authors reported a large inter-subjects variability in EMG pattern, with a large interlimb variability within the same subject. Their findings were consistent with those of other authors in that they reported prolonged activation of EMG activity and co-activation of agonist/antagonist muscles occuring during stance.

Abnormal activation patterns of lower limb muscles during ambulation in spastic paraparesis were briefly described by Knutsson (1980). Spastic reflexes, paresis and co-activation abnormalities similar to those seen in spastic hemiparesis (Knutsson and Richards, 1979) were primarily identified. However, the combination of these disturbances were more complex than those seen in the hemiplegic subjects. Of the subjects investigated, five demonstrated spastic reflexes, three demonstrated a combination of spastic reflexes and paresis while one subject demonstrated a combination of spastic reflexes and co-activation. Four subjects showed co-activation of 4 out of 6 lower extremity muscles while paresis was the dominant problem in the remaining three subjects. However, the gait abnormalities resulting from these altered patterns were not described in terms of kinematic and temporal distance parameters.

Knutsson (1983) also identified prolonged activation of weight supporting muscles, quadriceps and abductors, during the stance phase in response to loading and stretch activation. He defined this phenomenon as 'crutch spasticity' present to increase limb stability during the loading phase.

Conrad et al (1985) and Benecke and Conrad (1986) conducted studies that specifically evaluated the effect of spasticity on paraspastic gait. Ten subjects with upper motorneuron lesions, resulting in a paraspastic gait (symmetrical involvement) with minimal paresis and sensory disturbances, walked on a treadmill at 2 km/h. Abnormalities in the EMG recruitment patterns were revealed, which were characterized by early onset of activity with prolonged activation and delayed muscle relaxation. Once the muscles were recruited there was no increase in the level of

EMG activity, leading to a loss of the dynamic peaks of activity typically seen in healthy subjects at strategic points in the gait cycle. This was especially evident in the ankle and knee flexor muscles. Quadriceps showed a decrease in peak activity occuring shortly after heel strike with an occasional second burst appearing between the end of stance and midswing. A decrease in peak activity was seen in the tibialis anterior muscle at heel strike with an increase in amplitude in the second burst appearing at the beginning of swing. The results of these studies revealed a link between the onset of EMG activity and muscle length. This was seen in the gastrocnemius, biceps femoris and quadriceps muscles as they were the most active while in a lengthened position.

It may be important to consider variables other than spasticity underlying neurologically impaired gait. It has been postulated (Dietz et al, 1981; Dietz and Berger; 1983; Berger et al, 1984) that the increased tension recorded in the gastrocnemius during stance phase is not due to hyperactive reflex activity alone, but also to accompanying changes in the muscle's viscoelastic properties. This was concluded following the observation that tension in gastrocnemius muscle continued to increase during the stance phase without a corresponding increase in the muscle's EMG activity.

In summary, studies of EMG activity in neurological gait reveal one or more of the following characteristics: a general decrease in EMG activity, an early onset of activity, prolonged duration of the burst and co-activation of agonist/antagonist muscles.

2.2.2 Sagittal Angular Displacement Profiles

Abnormal sagittal angular displacement profiles of lower limbs joints in spastic gait can result from the altered patterns of muscle activation. Different underlying mechanisms can be responsible for the same joint movement abnormality (Knutsson and Richards, 1979). For example, knee hyperextension during stance can result from premature activation of calf muscles which pulls the lower leg back as the body moves forward; or it can occur in cases of paresis as a compensatory mechanism to increase limb stability (Knutsson and Richards, 1979).

Sagittal angular displacement patterns characteristic to paraspastic gait have been briefly described by Conrad et al (1983, 1985). These include an overall decrease in the amplitude of knee angular displacement in conjunction with a significant degree of knee flexion at heel strike. During the stance phase the knee progressively extended although it never reached full extension. The ankle joint was held in excessive dorsiflexion throughout the gait cycle.

2.2.3 Temporal Distance Parameters

In spastic syndromes, like those reported by Conrad et al (1985), the stance/swing ratio does not differ significantly from that of normal subjects. In general, a spastic gait produces a shorter cycle duration with an increase in the frequency of stepping. Frequently in spastic paretic gait there is an asymmetrical involvement of the lower limbs similar to that seen in hemiplegia. In such cases the less involved lower extremity plays a major compensatory role. The involved side will show a prolonged swing time and a decrease in stance time compatible with the patients' inability to bear weight on the more affected extremity (Brandstater et al, 1983;

Knutsson and Richards, 1979).

Conrad et al (1983, 1985) propose that some of the disturbances of gait seen in neurological syndromes are in part due to the presence of 'protective gait mechanisms'. When comparing spastic subjects walking on a treadmill with and without parallel bars, a decrease in step duration, an increase in footflat contact time and double limb support time were observed. These changes in temporal distance parameters serve to increase stability during times when unexpected disturbances in balance may occur.

Neurological patients walk at speeds that are considerably lower than the comfortable walking speed of healthy subjects (Conrad et al, 1985; Shiavi et al, 1987; Barbeau et al, 1988). Shiavi et al (1987) compared the stride parameters of 12 hemiplegic subjects to those of healthy subjects walking at very low speeds (average: 0.55 ms^{-1}). The authors reported that speed and percentage single limb support time were less than normal, while stride time and percentage stance were greater than that of normals walking at very low speeds. The reasons for neurological patient's inability to walk at higher speeds remains to be elucidated. The presence of muscle hypertonia (Dietz, 1986), speed-dependant hyperactive stretch reflexes (Burke and Lance, 1973), and an inability to generate force at the higher speeds (Knutsson and Martensson, 1980) may be contributory factors.

2.3 Current Methods of Treatment

A patient's functional walking status following a spinal cord lesion depends on the level and completeness of the lesion, as well as the motor and sensory deficits that result from it. The usual therapeutic approach to retrain gait in this population remains conservative. Conventional treatment focuses mainly on strengthening and bracing of the paretic and spastic lower extremities. Pharmacological agents to decrease spasticity play a potential role in controlling abnormal reflex activity and improving gait. In recent years, the increasing use of functional electrical stimulation, has been the only significant change in treatment approach.

2.3.1 Conventional Treatment

Following an incomplete spinal cord lesion, the conventional treatment approach is to reeducate gait, beginning with strengthening trunk musculature and any lower extremity muscles where voluntary movement has been preserved (Peterson, 1985). As patients frequently present with excessive paralysis, proprioceptive problems and/or spasticity, bracing is employed to overcome these problems and allow the patient to ambulate. Once the patient has been strengthened and the appropriate orthotic apparatus (long or short leg braces) have been prescribed, gait retraining follows the same regular course as other neurological conditions. Guttman (1976) describes such a regimen. The patient practices pelvic mobility and control to increase balance and weight shifting while standing in the parallel bars preferably in front of a mirror. The gait is then broken up into its components and the patient practices postural stability and balance in stance and swing while trying to achieve proper foot placement. Depending on the patient's voluntary movement and level of lower extremity bracing a four-point gait, swing-to-gait, or swing-through gait is taught. Unfortunately the high energy requirement associated with these types of gait make them non-functional. In the long run, most patients use their wheelchair as a functional means of ambulating.

2.3.2 Drug Therapy

Pharmacological intervention for the treatment of spasticity is a common practice in the management of the spinal cord injured patient. Among the most popular antispasmodic drugs are diazepam, baclofen and dantrolene (Young and Delwaide, 1981a, 1981b). Diazepam acts postsynaptically to enhance the action of the inhibitory neurotransmitor GABA in the spinal cord and supraspinal structures, and thereby decreasing spinal reflex activity (Davidoff, 1978). Baclofen appears to act on presynaptic mechanisms and is effective by producing hyperpolarization of afferent fibers and a decrease in monosynaptic and polysynaptic reflexes (McDowell, 1981). Dantrolene acts directly on the muscle's contractile mechanism by depressing calcium release from the sarcoplasmic reticulum (McDowell, 1981), however the decrease in spasticity is most often accompanied by muscle weakness. Although these drugs may have a positive influence on spasticity, side effects such as nausea, light headedness, and drowsiness often accompany their use.

Little has been reported on the quantitative effects of antispasmodic drugs on the locomotor pattern in spinal cord injured pateints. Decreasing spasticity will not invariably result in an improved gait pattern, spastic reflexes present at rest do not necessarily interfere with the locomotor pattern (Knutsson, 1980). Knutsson (1983) reported that decreasing stretch reflexes with tizanidine would cause a deterioration in gait, by decreasing activation of weight supporting muscles (quadriceps) which require the stretch reflex during loading to prevent knee collapse. However, decreasing stretch reflexes in the calf muscles led to an improved gait pattern with greater dorsiflexion at the ankle. Cyproheptadine and clonidine have been reported to be effective in improving the ability to walk (Barbeau et al, 1982; Stewart et al, 1987). Quantitative improvements in locomotor parameters such as decreased cycle duration, improved foot-floor contact, increased joint excursions as well as higher walking speeds were reported with cyproheptadine (Wainberg et al, 1986) while improvements in the timing of muscle acivation have been reported with clonidine (Stewart et al, 1987).

2.3.3 Functional Electrical Stimulation (FES)

In recent years the use of FES as an alternative to lower limb bracing to restore gait following a spinal cord lesion has gained popularity. Coordinated stimulation of lower limb peripheral nerves and muscles can simulate a primitive reciprocal gait in patients with complete spinal cord lesions between the levels of T5 to T12 (Stojnik et al, 1979; Bajd et al, 1983; Braun et al; 1985). Before this type of gait training is commenced, lower extremity muscles are strengthened using electrical stimulation, and supported standing in parallel bars is achieved. FES to the knee extensors provides the patient with the ability to rise independently from the wheelchair with the use of arm supports (Badj et al, 1982).

The use of FES as an ambulatory aid presents numerous advantages. The energy required is less than for the swing-to or swing-through gait since patients are not required to lift their body weight. As well, the electrodes are easier to apply than orthoses, the resulting muscle activity provides blood flow to the extremities, it prevents muscle contractures, it decreases spasticity and provides the patient with a sense of well-being by allowing the use of the legs to stand and walk (Mizrahi et al, 1985). An added benefit of FES is that it allows weight bearing on the lower extremities which is necessary during the stance phase for proper ambulation (Stojnik et al, 1979). Mizrahi et al (1985) reports up to 80% weight bearing in standing and 41% to 61% weight bearing is reported by Braun et al (1985).

Regardless of these advantages, numerous problems continue to exist, making the use of FES to ambulate unfeasible to achieve functional ambulatory independence. Although the energy required is lower than that for ambulating with bilateral long leg braces, it remains 10x greater than for normal gait. This is reflected by the increased oxygen consumption, heart rate and muscle lactic acid levels (Isakov et al, 1985). Moreover, the walking speed attained with FES is very low, ranging from about 0.10 ms^{-1} to 0.17 ms^{-1} (Mizrahi et al, 1985; Braun et al; 1985), therefore not meeting the needs for outdoor ambulation. Additionally, the stimulated muscles fatigue, limiting the amount of time during which FES can be used (Vodovnik et al, 1981). Safety is another important element, as instances of the FES system malfunction have been reported while the patient is walking.

2.4 New Gait Training Strategy Proposed

In light of the literature presented above, it becomes apparent that the treatment strategies available for locomotor training following a spinal cord injury are largely inadequate. It is a common clinical finding that spinal cord injured patients have difficulty weight bearing (through their affected lower extremities) while progressing forwards. Parallel bars and walking aids are often used for this reason. One of the major limitations of gait training is that it has been performed under full weight bearing conditions. A new training strategy has been proposed for retraining neurological gait consisting of progressive weight bearing and treadmill stimulation (Finch and Barbeau, 1986; Barbeau et al, 1988). The strategy involves walking patients on a treadmill while a percentage of their body weight is supported by an overhead harness. This constitutes a dynamic, task-specific approach to retrain gait. The advantages and clinical implications of such a system appear to be numerous. Supporting a percentage of body weight while patients are walking may facilitate the expression of the locomotor pattern. The support and safety presented by the harness would decrease 'protective gait mechanisms' (Conrad et al, 1985). This strategy would train the three components of gait simultaneously; i.e. weight bearing, balance and stepping. It would allow for 'interactive locomotor training' (Rossignol et al, 1986), since instant correction of gait deviations by the therapist could be achieved as the patient walks on a treadmill. Locomotor training could be initiated earlier in the rehabilitation period, as patients could be provided with as much body weight support (**BWS**) as needed.

This proposed training strategy is largely based on the spinal animal model which will be presented in the following sections. The effects of BWS in infants and normal subjects will also be discussed.

2.4.1 Animal Studies

2.4.1.1 Infant Lesions

Cats which are spinalized during the neonatal period from 7 to 14 days following birth recover a locomotor pattern which remarkably resembles that of the intact cat (Grillner, 1973; Forssberg et al, 1980a; 1980b; Smith et al, 1982). Once the kittens reach maturity they are able to walk on the treadmill at different speeds while supporting their body weight on their hindlimbs (weight supported locomotion). The EMG activity of the prime muscles active during locomotion at the hip, knee and ankle are comparable to that of the intact cat with respect to both the on/off timing of the muscle burst in relation to the gait cycle, and the profile of the muscle burst. The extensor muscles are activated throughout the stance phase producing enough force to support the animal's weight.

2.4.1.2 Adult Lesions

Eidelberg et al (1980) studied the recovery of locomotion following spinal cord transection in the adult cat. Following two months of treadmill training, the cats were capable of producing stepping movements of the hindlimbs, although none of the cats were capable of weight-supported locomotion. During walking there was little intra and interlimb coordination, a large variability in the duration of the step cycle and its components, lack of knee and ankle coupling as well as dragging of the dorsal surface of the paw. Smith et al (1982) later recognized the importance of training in the recovery of locomotion in the adult spinal cat. Five cats were trained to walk on a treadmill over a 4 month period. Although the locomotor pattern was generally rated inferior to that of cats spinalized in the neonatal period, 3 of the 5 cats demonstrated excellent weight supported locomotion. There was normal recruitment of the soleus and gastrocnemius muscles. Kinematic abnormalites which persisted were an absence of the yield phase at the knee during stance as well as uncoupling of the knee and ankle.

Barbeau et al (1987) reported a remarkable recovery of the locomotor pattern in the adult spinal cat. After 4 to 12 weeks of regular 'interactive locomotor training', all the cats demonstrated weight supported locomotion with proper foot placement with the sole of the foot on the treadmill. The EMG and kinematic qualities of the gait closely resembled that of the intact cat. By the end of the recovery period, many of the gait abnormalities which had been reported by other authors were not evident. The authors define 'interactive locomotor training' as a close interaction between the experimenter and the animal, allowing the animal to bear only the amount of weight it is capable of within a given period, thereby allowing for proper foot placement. As the gait pattern improved, the weight bearing was progressively increased until the animal reached full weight bearing. It appeared that both training and the type of training the animal was subjected to were equally important in the recovery of locomotion in the adult spinal cat. Progressively increasing the amount of weight borne by the animal seemed to favor the recovery of a near normal gait pattern.

2.4.2 Effects of BWS on Locomotion

2.4.2.1 Infant Studies

Infants learn how to walk from the 6th to the 12th month of life by supporting themselves as they walk in order to compensate for their lack of equilibrium (Forssberg, 1985; 1986; Berger, 1986). During this period the child walks full weight bearing and the EMG profiles of lower extremity muscles reveal co-activation of agonist/antagonist muscles (Forssberg, 1985). It is only later, after the child has already achieved independant ambulation, that reciprocal activation patterns of prime movers of the hip, knee and ankle begin to appear.

Kazai et al (1975) repeatedly analysed the EMG activity of lower extremity muscles in 2 infants during the period in which they learned how to walk. In the 7 month old child, walking with the support of a moveable support (infant wheeled walker), a reciprocal activation pattern in TA and GA would occasionally appear. The authors explained this as resulting from a of decreased load on the ankle joint although the actual load decreased was not quantified. When the same child's EMG were examined during crawling, at a time when there is no load on the ankle, a reciprocal firing pattern in these two muscles was clearly evident. In a 12 month old child on his first day of independent walking, synchronous activity of flexor and extensor lower limb muscles was recorded. When support was provided, a decrease in EMG amplitude and a more distinct EMG pattern for biceps femoris and rectus femoris was seen. On the 6th day of the child's free walking, supported walking revealed a reciprocal pattern of activation of tibialis anterior and gastrocnemius not seen during free walking.

It would therefore appear that decreasing the load on the lower limbs of infants during the initial walking period where co-activation is widely evident, will produce a more reciprocal activation of agonists and antagonists similar to that seen during mature gait.

2.4.2.2 Normal Subjects

The effects of 0, 30, 50 and 70% BWS on the EMG, kinematic and temporal distance parameters of gait has been investigated in 7 normal subjects walking on a treadmill with their body weight mechanically supported by an overhead harness

(Finch, 1986). It was found that with increasing levels of BWS, the comfortable walking speed progressively decreased. A significant decrease ($p < .01$) was found in the percentage stance and total double support time resulting in an increase in single limb support time with increasing body weight support. A decrease in total angular displacement at the hip and knee were also noted. The greatest decrease at the hip was seen at heel strike, foot flat, heel off, and maximum swing angle. The knee showed the greatest decrease at toe off, foot flat, and maximum swing angle.

The on/off timing of the muscle bursts in relation to the gait cycle for the muscles investigated was unchanged by increasing BWS, except for medial hamstrings which showed a delayed onset and a prolonged duration of activity. At all BWS levels, there was a significant decrease in mean burst amplitude for erector spinae and gastrocnemius and an increase for tibialis anterior. Vastus lateralis was unaffected by BWS while gluteus medius and medial hamstrings showed a significant decrease only at 70% BWS.

In summary, it appears that in normal sujbects BWS at levels below 70% modify gait parameters without producing an abnormal gait pattern.

2.5 Conclusion

Treatment approaches used at the present time to retrain gait in spinal cord injured patients present numerous limitations. Too frequently, a functional level of ambulation is not achieved and the patients rely on wheelchairs for mobility. Studies on the chronic adult spinal cat have shown that progressive weight bearing coupled with 'interactive locomotor' training plays a critical role in achieving a near normal gait pattern following spinalization. Since spinal cord injured patients experience difficulty in weight bearing through the affected lower extremities, this treatment strategy of progressively increasing the weight borne on the lower extremities as the patient is trained to walk appears applicable.

The effects of BWS have been investigated in normal subjects, however they have never been been studied in neurological patients. In order to validate this gait training strategy as a treatment approach for spastic paraparetics it is important to investigate the influence BWS has on this population's locomotor pattern.

CHAPTER THREE: METHODOLOGY

3.1 Population:

Fifteen male and female subjects, between 20 and 56 years of age, volunteered to participate in this research study. Seven subjects (table 3.1) participated in the preliminary study which investigated the effects of BWS on spastic paretic gait. Eight different subjects (table 3.2) participated in the main study designed to investigate the effects of parallel bars, speed, and body weight support on spastic paretic gait. Twelve of the subjects had sustained trauma-induced incomplete spinal cord lesions to the cervical or thoracic spine, two subjects suffered from non-familial progressive spastic paraparesis, while one subject had a surgically induced lesion at the level of T-10 following resection of a spinal tumor. All subjects were assessed by a neurologist to rule out other neurological disorders affecting the central nervous system or musculo-skeletal disorders of the lower extremities.

3.2 Orientation Session

Each subject participated in both an orientation session and an experimental session. During the orientation session the subjects were familiarized with the goals of the study, the experimental procedure, and the equipment which was utilized for data collection. Informed consent was also obtained from each subject for his participation in the study. The subjects were habituated to walking on the treadmill for one to five minutes, depending on their tolerance, at the BWS trials, 0% and 40%. The safety features incorporated in the treadmill and BWS system were explained to each subject before he walked on the treadmill. Following each walking trial, the

TABLE 3.1: Demographic data of the seven subjects participating in the preliminary study. C=cervical spine; T=thoracic spine; SP=spastic paraparesis.

DEMOGRAPHIC DATA

SUBJECT	SEX	AGE (YRS)	LESION LEVEL	CHRONICITY (YRS)
MP	M	27	C4-5	3.5
RP	M	41	C5-6	>1.0
MB	M	56	SP	7.0
RM	M	31	C-6	15.0
BM	M	23	C-4	>1.0
RL	M	56	T-11	1.0
SQ	M	24	T4-7	1.0

TABLE 3.2: Demographic data of the 8 subjects participating in the main study. C=cervical spine; T=thoracic spine; SP=spastic paraparesis.

DEMOGRAPHIC DATA

SUBJECT	SEX	AGE (YRS)	LESION LEVEL	CHRONICITY (YRS)
BP	M	32	SP	3.0
CF	M	23	C6-7	1.5
LR	F	22	T-10	1.5
RC	M	23	C6-7	4.0
LL	M	32	C7-T1	1.5
BG	F	26	C1-2	21.0
FS	M	20	T8-9	1.5
JS	M	42	T9-10	0.6

subject's blood pressure and pulse were recorded and compared to a resting baseline measure, obtained at the beginning of the session, to monitor effects on the cardiovascular system.

During the orientation session, a clinical neurological examination was performed on those subjects participating in the main study. The examination was comprised of sensation charting, muscle testing, spasticity evaluation (clonus test and tonic stretch reflex at the knee and ankle) and an overground ambulation profile to identify gait asymmetries and functional capacity (see Appendix).

3.3 Experimental Session:

The experimental session was held the day following the orientation session. Habituation to treadmill walking was followed by the application of electrodes, joint markers, and foot switches to the left or right lower extremity for the recording of gait parameters.

The subjects then walked on the treadmill at two randomly assigned BWS trials (0% and 40%), from which data were recorded. Experience with the BWS system revealed that providing more than 40% BWS resulted in loss of heel ground contact in some patients. Therefore 40% BWS was chosen as the level of BWS to be investigated. A 10-minute rest period was given between each BWS trial to prevent fatique. Blood pressure and pulse were monitored following each trial to control for undue stress on the patients.

In the preliminary study, the habituation trial was used to determine each subject's comfortable maximum treadmill speed at 0% BWS. This speed was kept constant for comparison of the two BWS trials (0% and 40%) in order to control for the confounding effect of speed on the gait parameters. Maximum comfortable walking speed at 40% BWS was also recorded to determine changes with BWS.

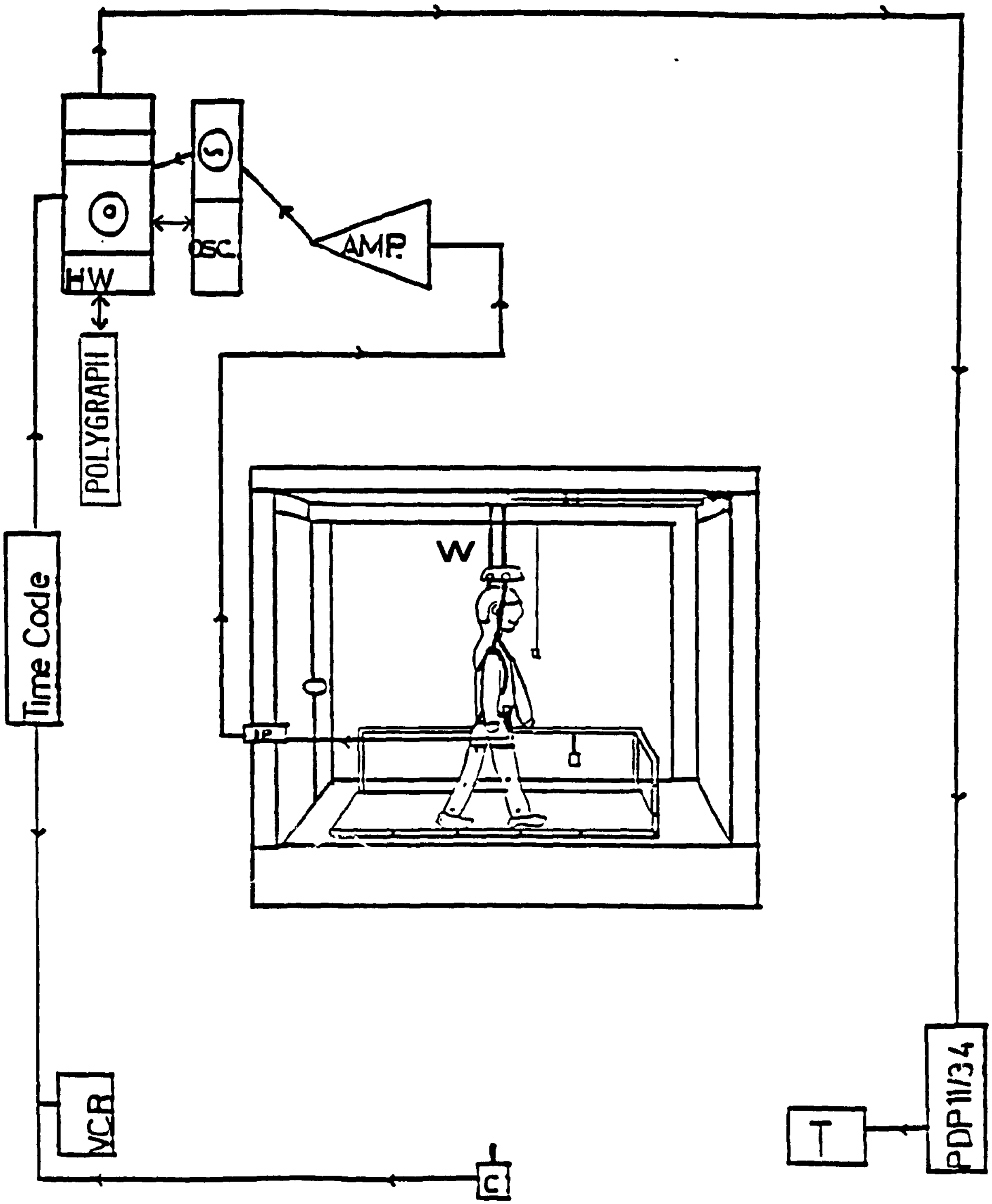
For subjects who participated in the main study, maximum comfortable treadmill speed at 0% BWS was determined in a similar fashion. Once the comfortable walking speed was identified it was determined whether the subjects could walk at higher speeds. Data were then collected for three pre-determined speed levels (minimal, comfortable and maximal) at 0% and 40% BWS while the subjects used parallel bar support. In addition, for those subjects who were able to walk without parallel bar support such a subtrial was added at each BWS and speed level.

3.4 Data Recording and Instrumentation:

Data for this research stuay were collected in the Human Gait Laboratory at McGill University's School of Physical and Occupational Therapy, which has been previously described in detail (Barbeau et al, 1987). Subjects walked on a motor driven treadmill while their body weight was supported by an overhead safety harness. Figure 3.1 shows the instrumentation of the laboratory.

Each subject walked on a motorized treadmill (W.E. Collins #101) while a percentage of BWS was provided. The BWS apparatus (figure 3.2) consisted of a custom designed harness (modified tyrolean lift) which mechanically supported the subject vertically over the treadmill. The harness comprised a pelvic band attached around the hips and two padded straps which pass between the legs to attach

FIGURE 3.1: Schematic flow diagram of data acquisition and processing. Data flows in the direction of the arrows. HW: Honeywell; OSC: Oscilloscope; VCR: Video Recorder; C: Camera; T: Transiac; AMP: EMG Differential Amplifier; P: Pre-amplifier; W: Weight Support System.


HW
OSC.
AMP.
POLYGRAPH
Time Code
W
IP
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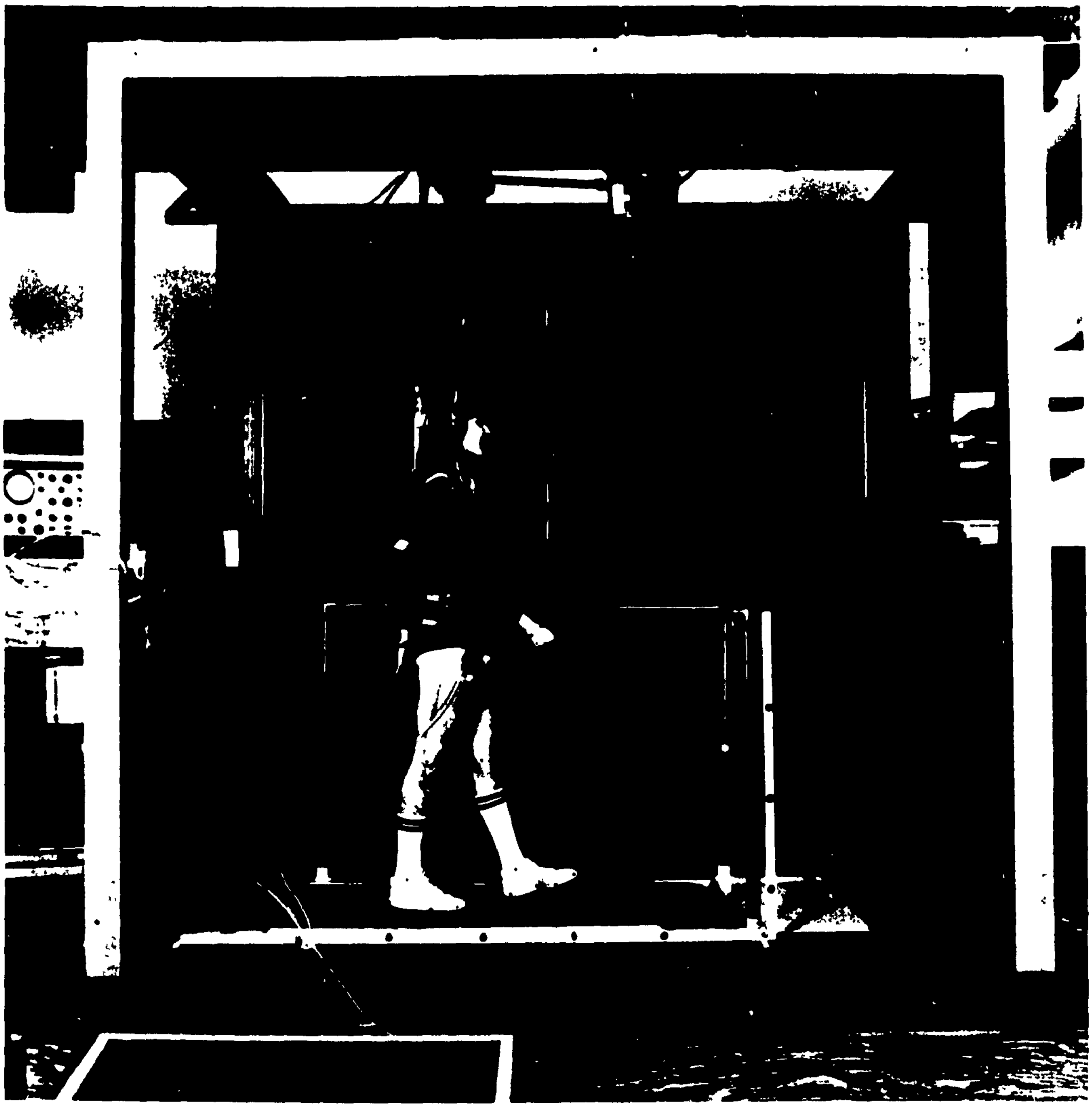
C
T
PDP11/34

FIGURE 3.2: A lateral view of the BWS system consisting of a custom designed harness, which supports the patient over a motorized treadmill.



anteriorly to the pelvic band. Two metal parallel bars were also available for support. The system provides balance, support and treadmill stimulation while supporting body weight. At the beginning of each trial, the BWS system was calibrated to each subject's weight by using a force transducer. While the subject was seated, with no tension on the harness, the weight support calibration was set to 0%. The subject was then lifted and suspended with his feet off the treadmill and the weight support calibration was set to 100%. The subject was subsequently lowered into a standing position with either 0% or 40% of body weight supported by the harness, depending on the trial.

While the subject walked on the treadmill, EMG activity of lower extremity muscles was recorded. Footswitches applied to the soles of the shoes defined the gait cycle and provided data for determining the temporal distance parameters. Videotape recordings allowed for the analysis of sagittal angular displacement data at the trunk, hip, knee, and ankle. In order to synchronize EMG with sagittal angular displacement data, a signal from the time code generator (Skotel:TCG-80) was simultaneously recorded on both the videotape and FM magnetic tape. The EMG and footswitch signals were also printed out on a Grass polygraph and an artifact-free sequence was chosen for analysis.

3.4.1 EMG:

The EMG activity in representative lower extremity muscles during treadmill walking was detected by bipolar surface silver/silver chloride electrodes (Meditrace Pellet Electrodes). EMG activity from the Gluteus Maximus (GX), Vastus lateralis (VL), Medial Hamstrings (MH), Tibialis Anterior (TA), Medial Gastrocnemius

(GA), and Soleus (SOL) was recorded from the right or left lower limb muscles. Electrodes were placed 2 cm apart (center to center), longitudinally over the belly of each muscle. Skin preparation prior to electrode placement involved shaving and rubbing with alcohol until the skin was pink in order to lower skin impedance. A ground electrode was placed over the bony surface of the right or left tibia. A point finder, which detects areas of lower skin resistance, was used to confirm the location of motor points. The surface electrodes were placed over these motor points.

Electrolyte gel was applied to the electrodes before placing them on the skin in order to improve conductance. The leads were then snapped onto the electrodes and fixated with adhesive strips over the electrodes. EMG signals were buffered and preamplified as they travelled through the leads which connected into a multichannel box. The signals travelled through a four meter cable to amplifiers, where they were amplified and bandpassed (10-1000Hz). An eight channel oscilloscope (Nihon Kohden V.C.-6800) displayed the EMG signals as well as the footswitch signals and weight support. Finally, these signals were recorded onto a 14 channel magnetic FM tape recorder (Honeywell #101) at 3.75 IPS (midband recording level 2500Hz). A short sequence of EMG activity was recorded during quiet sitting, prior to treadmill walking, to provide a reference for resting level EMG activity.

3.4.2 Foot Switches:

Temporal distance parameters of gait were determined via pressure sensitive footswitches (Tapeswitch Systems of America) placed under the heel, 5th metatarsal head and big toe of each subjects' shoes. Footswitches indicated heel contact, foot flat and toe off and defined the different components of the gait cycle such as double support time, single limb support time, cycle duration, etc.. Footswitch signals were synchronized with EMG data to normalize muscle activity to the gait cycle.

3.4.3 Video Recordings:

Sagittal angular displacement profiles of the trunk, hip, knee, and ankle were filmed by a rotatory shutter video camera (Sony:RSC-1010) placed central, perpendicular and at a distance of 4 meters from the treadmill. The exposure time was set at 0.002 sec. This allowed filming of the entire upper and lower body. The treadmill area was illuminated by two 1000 watts quartz lamps (Acme Lite Co. 71056). Reflective joint markers (half ping-pong balls filled with polyfoam and centered with a 1 cm dot) were placed at the shoulder, hip, knee and ankle as well as the heel, metatarsal and toe region of the lateral border of the right or left shoe. Whenever possible, the joint markers were attached directly to the skin. Additional markers were placed on a horizontal and vertical bar to be used as absolute coordinates for the video analysis. A video monitor (Panasonic : WV-5470) was used to view the trials which were recorded on 3/4 inch VHS video tapes (Videocasette Recorder Panasonic: NV- 9240) at a speed of 60 fields per second. A remote search controller (Panasonic NV-A505) allowed field by field viewing for measurement.

3.5 Data Analysis:

Following anti-aliasing filtering (low-pass cutoff at 450 Hz), the EMG signals were digitized at 1 KHz for offline computer analysis. A PDP 11/34 computer was used. The EMG signals were further full wave rectified and low pass filtered with a 3.0 Hz cut-off frequency to produce analog linear envelopes (Olney and Winter, 1985). The footswitch and EMG signals were displayed on a computer terminal (Transiac TRI 1024), and interactive computer programs were used to analyze the data collected. Placement of arrows on the foot switch signals allowed for the data to be synchronized to the normalized stride duration defined as the period from initial foot-floor contact (0%) to the subsequent foot-floor contact (100%). The within-subject ensemble average of 10 strides were used for each muscle as the represenatative profile for a given subject. Temporal distance parameters of gait such as cycle duration, single limb support time, % swing, % stance and % total double support time were also determined.

The sagittal angular displacements of the trunk, hip, knee, and ankle were manually measured from the monitor screen using a goniometer. Once the subject had reached a steady state while walking on the treadmill, one representative gait cycle for each subject during each experimental paradigm was analyzed. The joint angular displacements were measured at every 5% of the gait cycle. The trunk and hip angles were calculated with respect to a vertical line, with the neutral position in standing being taken as 0° displacement of the trunk and hip, flexion being positive and extension negative. Likewise, in calculating the knee and ankle angles, the neutral standing position, with the knee at full extension, and the shank axis perpendicular to the foot, was taken as 0°. Knee flexion and ankle dorsiflexion beyond neutral was taken as positive angular displacements, and ankle plantarflexion beyond neutral was taken as negative angular displacement.

CHAPTER FOUR: RESULTS AND DISCUSSION

4.1 THE EFFECTS OF BODY WEIGHT SUPPORT ON THE LOCOMOTOR PATTERN OF SPASTIC PARETIC PATIENTS

(Can J Neurol Sci 1989; 16:315-325)

4.2 THE EFFECTS OF PARALLEL BARS, SPEED, AND BODY WEIGHT SUPPORT ON THE MODULATION OF THE LOCOMOTOR PATTERN OF SPASTIC PARETIC GAIT

(to be submitted to: Journal of Neurology, Neurosurgery and Psychiatry)

4.1 The Effects of Body Weight Support on the Locomotor Pattern of Spastic Paretic Subjects

The effects of mechanically supporting a percentage of body weight on the gait pattern of spastic paretic subjects during treadmill locomotion was investigated. Electromyographic (EMG), joint angular displacement and temporal distance data were simultaneously recorded while 7 spastic paretic subjects walked at 0% and 40% body weight support (BWS) at their maximal comfortable treadmill speed. Forty percent BWS produced a general decrease in EMG mean burst amplitude for the lower limb muscles investigated, with instances of more appropriate EMG timing in relation to the gait cycle. The joint angular displacement data at 40% BWS revealed straighter trunk and knee alignment during the weight bearing phase especially at initial foot-floor contact and midstance. An increase in single limb support time and a decrease in percentage total double support time were evident at 40% BWS. An increase in stride length and maximum comfortable walking speed was also seen with BWS. The use of BWS during treadmill locomotion as a therapeutic approach to retrain gait in neurologically impaired patients is discussed.

THE EFFECTS OF BODY WEIGHT SUPPORT ON THE LOCOMOTOR PATTERN OF SPASTIC PARETIC PATIENTS

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Introduction:

Many of the gait deviations observed in neurologically impaired patients result from their inability to adequately bear weight through their affected lower extremities during the loading phase of the gait cycle (Carlsoo et al, 1974; Dickstein et al, 1984; Wall and Turnbull, 1986). Conventional treatment to retrain gait following a lesion to the central nervous system consists of retraining weight bearing, weight shifting and balance during isolated events of the gait cycle before incorporating these components into the dynamic locomotor task (Guttman, 1976; Johnson, 1976; Bobath, 1978). However, retraining gait under such static conditions appears limited as gait deviations are often seen to persist following such a treatment approach (Wall and Turnbull, 1986; Bogardh and Richards, 1981).

An alternative approach may be to support a percentage of the patient's body weight while retraining gait under dynamic conditions. As proposed by Finch and Barbeau (1985), this gait retraining strategy would consist of walking the patient on a treadmill at his maximum comfortable speed while a percentage of his body weight is supported centrally at the trunk by an overhead harness (Figure 4.1.1). The rationale for this retraining strategy is based on findings from the spinal animal model.

Recovery of locomotor function following a spinal cord transection was considered to be largely dependent on the age of the animal at the time of the lesion (Grillner, 1973; Forssberg et al, 1980a; Smith et al, 1982). Until recently, cats spinalized at maturity were described as poor functional walkers with major deficits in their gait pattern. Although they were capable of producing stepping movements with their hindlimbs, they were unable to support their body weight on their hindquarters up to 8 weeks post transection (Eidelberg et al, 1980). Recently, the importance of training in accelerating the recovery (Barbeau and Rossignol, 1987) and maximizing the quality of the locomotor pattern (Smith et al, 1982; Lovely et al, 1986; Rossignol et al, 1986) in the adult spinal cat has been recognized. Rossignol et al (1986) and Barbeau et al (1987) have shown that cats spinalized (T13) as adults could recover a near normal locomotor pattern following an 'interactive locomotor training' program. During interactive locomotor training, the animal was supported by the tail and allowed to bear only the amount of weight such that it could walk with proper foot placement (with sole of the foot) on the treadmill. Following a period of one to three months of this training regimen, the animal was capable of walking while completely supporting the weight of its hindquarters with proper foot placement. Moreover, the gait pattern was comparable in many aspects to that of the intact adult cat (Engberg and Lundberg, 1969). The authors concluded that interactive locomotor training is an important factor in the recovery of locomotion in the adult spinal cat.

Based on the above animal findings, and clinical observations of inadequate weight bearing among neurologically impaired patients, it has been proposed that supporting a percentage of body weight and progressively decreasing the support while retraining gait may be an effective approach. In order to validate this training strategy, it is important to study the influence that body weight support (BWS) has on neurologically impaired gait. Since spastic paretic subjects have difficulty coping with loading of the lower limbs (Knutsson, 1983; Barbeau et al, 1988), such patients have been chosen as the focus for this study. The effects of providing BWS on the electromyographic (EMG), joint angular displacement and temporal distance parameters of spastic paretic gait were investigated.

Methods:

Subjects:

The study was conducted in the Human Gait Laboratory which has previously been described (Barbeau et al, 1987). The participants in this study were 7 spastic paretic sujects ranging in age from 23 to 56 years (mean= 36.9 years). Six of the subjects had sustained trauma-induced incomplete spinal cord lesion to the cervical or thoracic spine while one subject suffered from non-familial progressive spastic paraparesis. The chronicity of the lesion was one or more years for all the subjects (Table 4.1.1).

The seven subjects were described as mildly, moderately, or severely spastic, on the basis of their maximum comfortable treadmill speed and a qualitative visual grading of spasticity during locomotion (i.e. presence of clonus, stiff lower limb movements with decreased angular excursions at 0% BWS). Two mildly spastic subjects (MP, RP) walked at 0.39 ms^{-1} and 0.43 ms^{-1}, while two moderately spastic subjects (MB, RM) walked at 0.30 ms^{-1}. The remaining three subjects (BM, RL, SQ) were severely spastic and could only walk at the minimal treadmill speed of

TABLE 4.1.1: Demographic data

Demographic data of the seven subjects participating in the study. C=cervical spine; T=thoracic spine; SP=spastic paraparesis.

SUBJECT	SEX	AGE (yrs)	MAXIMAL COMFORTABLE TREADMILL SPEED ms^{-1}	LESION LEVEL	CHRONICITY (yrs)
MP	M	27	0.43	C4-5	3.5
RP	M	41	0.39	C5-6	>1.0
MB	M	56	0.30	SP	7.0
RM	M	31	0.30	C6	15.0
BM	M	23	0.26	C4	>1.0
RL	M	56	0.26	T11	1.0
SQ	M	24	0.26	T4-7	1.0

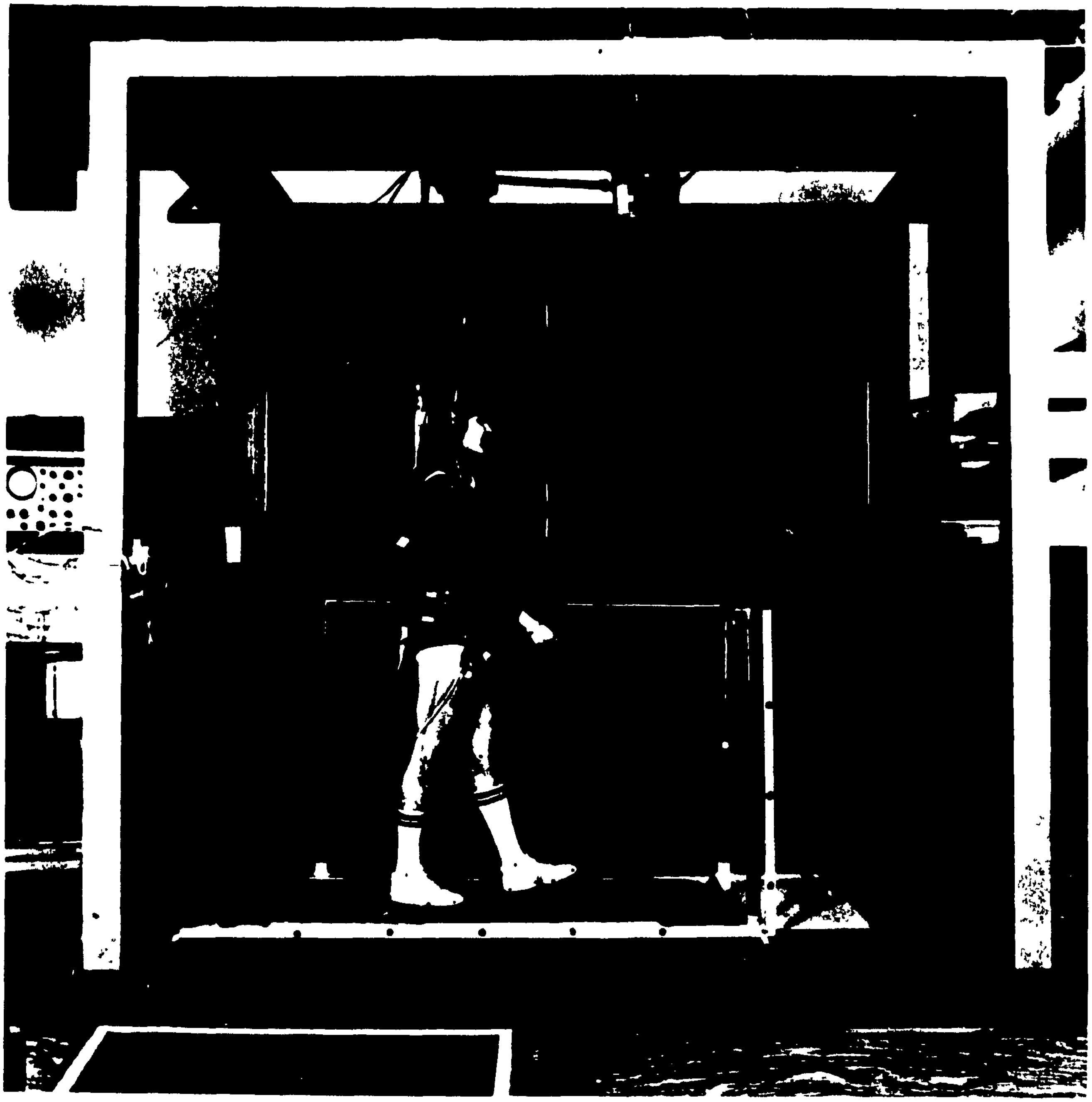
0.26 ms^{-1}. One (BM) of these three subjects was able to walk independently while the other two required manual assistance to advance the left lower extremity.

Body weight support:

The subjects walked on a treadmill while 0% (full weight bearing) and 40% BWS was provided. Previous experience with the BWS system revealed that providing levels of BWS higher than 40% resulted in a loss of heel-ground contact for some patients. Hence, 40% was chosen as the level of support to be investigated. The BWS apparatus (Figure 4.1.1) consisted of a custom designed harness which mechanically supported the patient vertically over the treadmill.

The harness consisted of a pelvic band attached around the hips and two padded straps which pass between the legs to attach anteriorly to the pelvic band. The percentage of BWS provided was calibrated using a force transducer. The force was normalized to each subject's weight (100%) and the sequence of % BWS provided was randomly assigned into two trials given within the same experimental session. Prior to data collection, each subject was habituated at 0% BWS for 1 to 5 minutes depending on his walking tolerance. During this trial, treadmill speed was slowly increased from the slowest speed of 0.26 ms^{-1} to the subjects' maximum comfortable walking speed. This speed was chosen for comparison of the two proceeding BWS trials in order to control for the confounding effect of speed on the gait parameters. Maximum comfortable walking speed at 40% BWS was also recorded to determine changes with BWS. A 10 minute rest period was given between each trial to prevent fatigue. Blood pressure and pulse were monitored

Figure 4.1.1: A lateral view of the BWS system, consisting of a custom designed harness, which supports the patient over a motorized treadmill.



following each trial to control for undue stress on the patients.

EMG and Footswitch Data:

EMG activity was recorded from the vastus lateralis (VL), medial hamstrings (MH), tibialis anterior (TA) and medial gastrocnemius (GA) of the right lower limb while the subject walked on the treadmill. Bipolar surface electrodes (2.5 cm center to center) were placed over the belly of each muscle following conventional skin preparation. The EMG signals were preamplified, differentially amplified and bandpassed (10-450 HZ). Footswitches placed under the heel, fifth metatarsal head, and big toe of each subject's shoes were used to detect heel strike, foot flat, and toe off, allowing for the determination of the temporal distance parameters. The EMG and footswitch signals were then recorded at 3.75 IPS on a 14 channel FM tape with a frequency response of 2500 HZ.

A sequence of artifact-free EMG signals was chosen for analysis. The EMG signals were digitized at 1 KHz for offline computer analysis. They were full-wave rectified and low-pass filtered with a 3.0 Hz cut-off frequency to produce analog linear envelopes. The EMG data were synchronized to the normalized stride duration defined as the period from the initial foot-floor contact (0%) to the subsequent foot-floor contact (100%). The within-subject ensemble average of 10 strides was used for each muscle as the representative profile for a given subject.

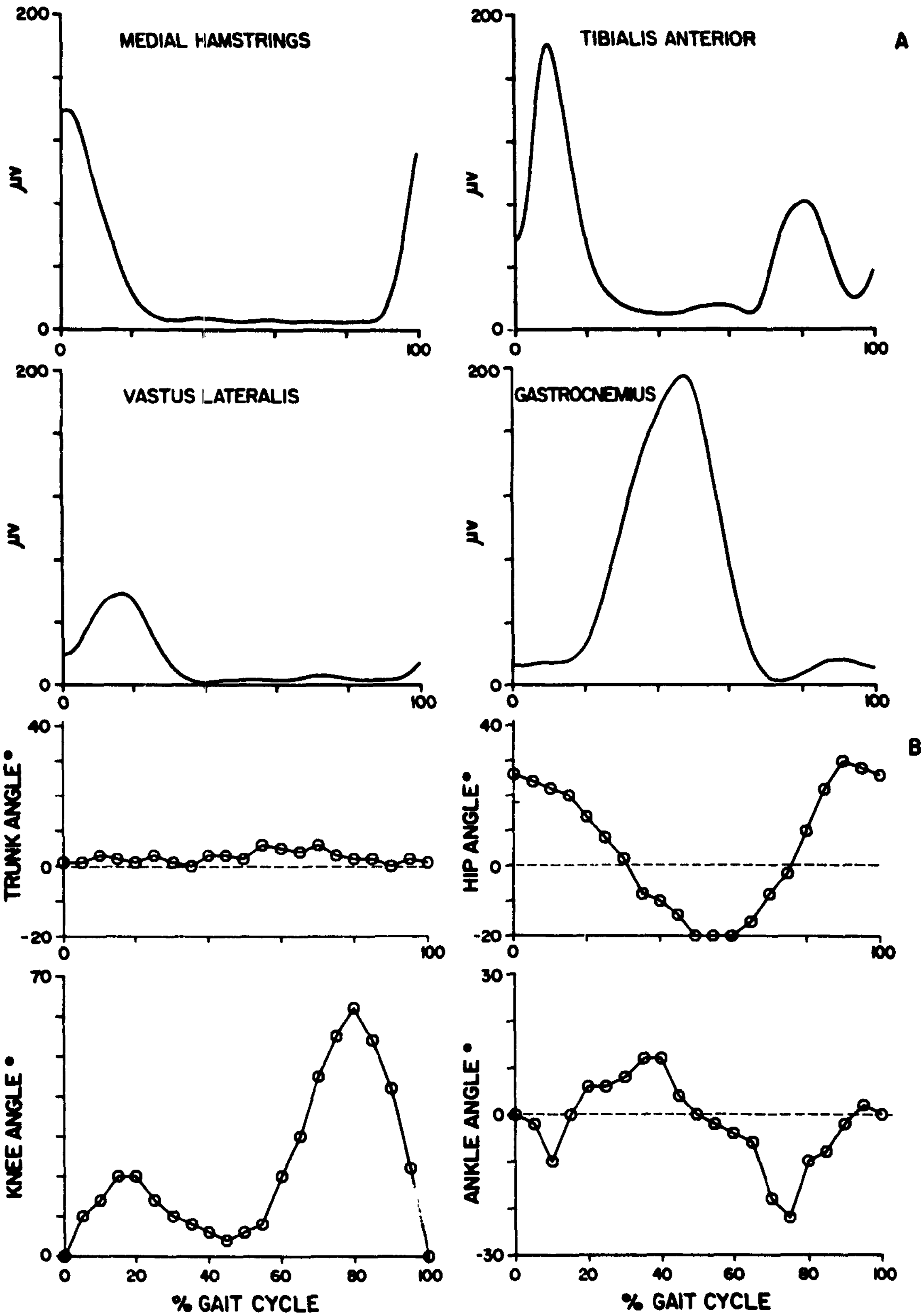
Joint Angular Displacement Data:

Joint angular displacement data were collected from the right lower limb. To do so, the subjects were videotaped as they walked on the treadmill using a shutter video camera. Reflective joint markers were placed at the shoulder, hip, knee, and ankle as well as the heel, fifth metatarsal head, and toe region of the lateral border of the right shoe. Additional markers were placed on a vertical and a horizontal bar to be used as absolute coordinates for the video analysis. The trials were recorded on a 3/4 inch video tape at a speed of 60 fields per second. A remote search controller was used for field by field viewing. The sagittal angular displacements were manually measured from the monitor screen using a goniometer. Once the subject had reached a steady state while walking on the treadmill, one representative gait cycle for each subject at each BWS was analyzed. The joint angular displacements were measured at every 2 or 5% of the gait cycle, depending on the stride duration. The trunk and hip angles were calculated with respect to a vertical line, with the neutral position in standing being taken as 0° displacement of the trunk and hip, flexion being positive, and extension negative. Likewise, in calculating the knee and ankle angles, the neutral standing position, with the knee at full extension, and the shank axis perpendicular to the foot, was taken as 0°. Knee flexion and ankle dorsiflexion beyond neutral was taken as positive angular displacements, and ankle plantarflexion beyond neutral was taken as negative angular displacement.

Normal Subject:

Data were collected in the same fashion for 10 normal subjects during treadmill walking at their maximum comfortable speed at 0% BWS. The profiles of EMG activity (averaged across 5 strides (Arsenault et al, 1986)) and the sagittal

Figure 4.1.2: A) EMG mean ensemble averages (across 5 strides) for MH, VL, TA and GA of a normal subject during treadmill walking at 1.36 ms^{-1} at 0% BWS and B) corresponding saggittal angular excursions of a representative cycle for the trunk, hip, knee and ankle.


200
MEDIAL HAMSTRINGS
μv
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TIBIALIS ANTERIOR
A
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VASTUS LATERALIS
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0
0
100
200
GASTROCNEMIUS
μv
0
0
100
40
TRUNK ANGLE °
0
-20
0
100
40
B
HIP ANGLE °
0
-20
0
100
70
KNEE ANGLE °
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40
60
80
100
% GAIT CYCLE
30
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0
-30
0
20
40
60
80
100
% GAIT CYCLE

angular displacements were similar to those reported previously by various authorsfor treadmill locomotion (Thorstensson et al, 1984; Murray et al, 1985; Arsenault et al, 1986). The EMG and kinematic data did not vary extensively between the 10 normal subjects, therefore permitting the illustration of one subject's gait profile (Figures 4.1.2A and B) to provide a template against which the pathological gait profiles can be compared.

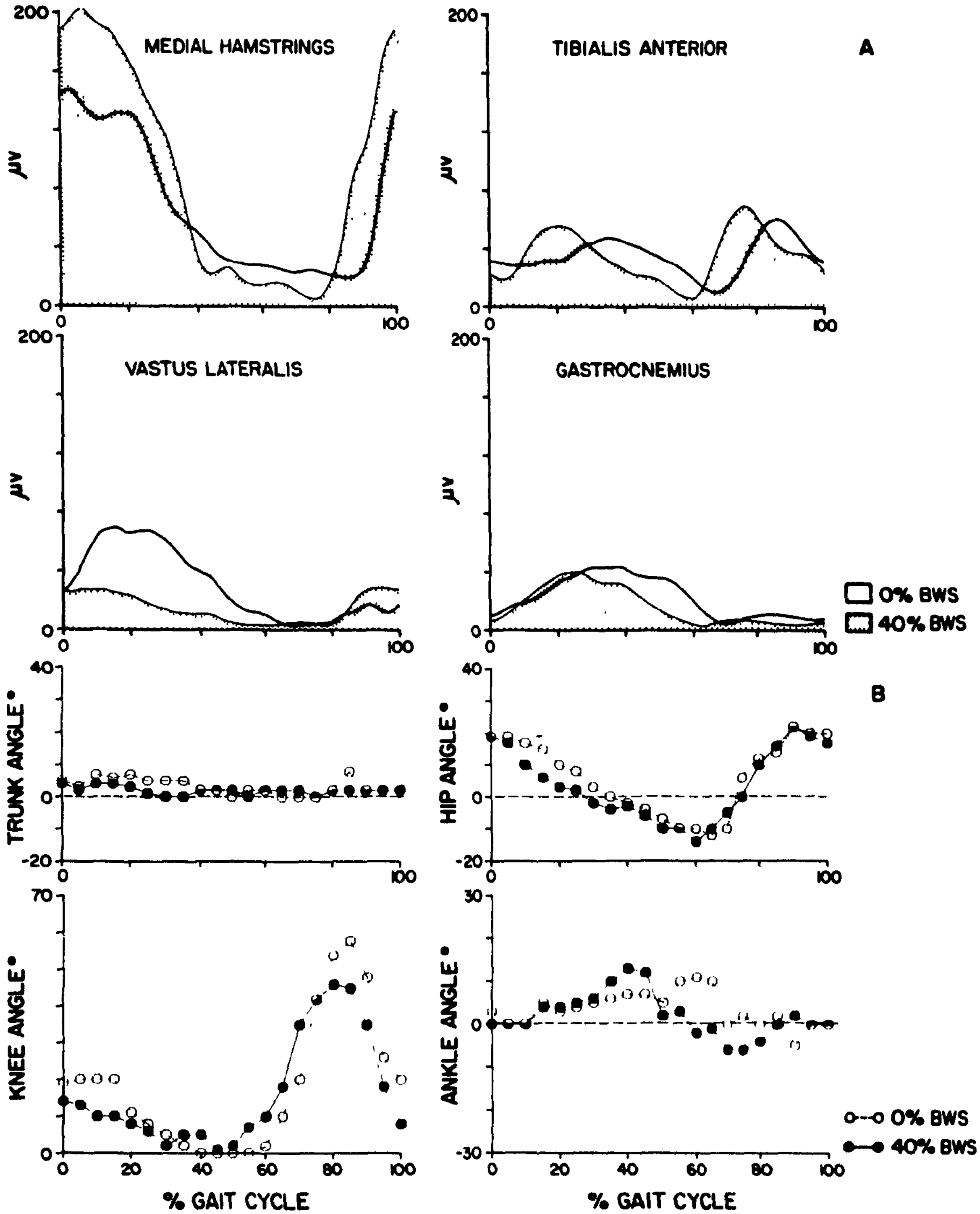
Results:

The EMG, joint angular dsiplacement, and temporal distance data collected for the spastic parapretic subjects displayed high variability between subjects, thus precluding pooling of the results for global analysis. Each subject was studied as a descriptive case study. Three of these cases will be described in detail.

Case 1: MP (Speed=0.43 ms^{-1})

The gait profiles described by EMG and joint angular displacement parameters of a mildly spastic subject (MP) walking at 0.43 ms^{-1} with 0% and 40% BWS are contrasted in Figures 4.1.3A and B. At 0% BWS this subject's ambulation profiles deviated from that seen in the normal (Figures 4.1.2A and B). At 40% BWS certain gait parameters were modified resulting in a more normal gait. Figure 4.1.3A shows an increase in amplitude for MH's EMG burst at 40% BWS as well as a decrease in EMG activity level during the muscle's silent period (between 40 and 80% of the gait cycle). The main effect of 40% BWS on VL was to produce a marked decrease in burst amplitude during the stance phase (Figure 3A). A more

FIGURE 4.1.3: A) Comparison of the MH, VL, TA and GA EMG mean ensemble average (across 10 strides) between 0% and 40% BWS trials in a mildly spastic patient (MP) during treadmill locomotion at 0.43 ms^{-1} and B) the corresponding saggittal angular excursions of a representative cycle for the trunk, hip, knee and ankle. Stance/swing transition at 0% BWS: 72.6% gait cycle; at 40% BWS: 55.4% gait cycle.


MEDIAL HAMSTRINGS
TIBIALIS ANTERIOR
A
VASTUS LATERALIS
GASTROCNEMIUS
μV
0% BWS
40% BWS
B
TRUNK ANGLE °
HIP ANGLE °
KNEE ANGLE °
ANKLE ANGLE °
% GAIT CYCLE
0% BWS
40% BWS

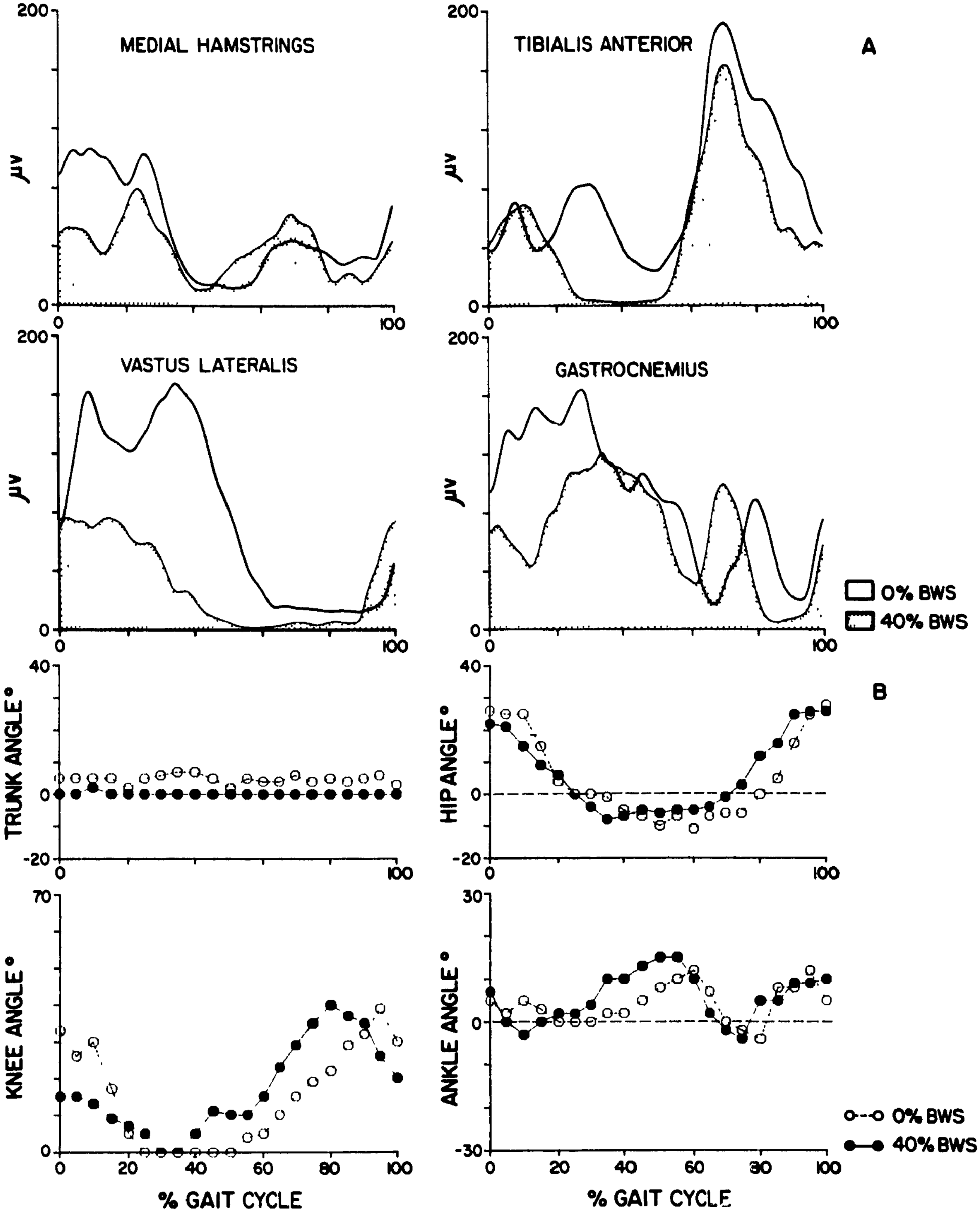
appropriate EMG profile was noted for TA at 40% BWS, with the initial burst occurring earlier in the stance phase (Figure 4.1.3A) as seen in the normal subject (Figure 4.1.2A). GA EMG activity subsided earlier in the gait cycle at 40% BWS, associated with a decrease in % stance (Figure 4.1.3A; Table 4.1.2).

The near normal trunk and hip joint angular displacement profiles at 0% BWS were not altered at 40% BWS. The more remarkable improvements at 40% BWS were seen at the knee. At heel strike there was a straighter knee alignment with gradual knee extension during midstance (Figure 4.1.3B). Qualitatively, subject MP demonstrated a smoother, less spastic gait at 40% BWS.

Case 2: MB (Speed=0.30 ms^{-1})

Figures 4.1.4A and B depict the EMG and joint angular displacement profile of a moderately spastic subject, MB. Figure 4.1.4A illustrates the prolonged activation during the entire stance phase of all 4 muscles at 0% BWS. At 40% BWS there was a decrease in amplitude in the initial MH burst, with a small increase in the second burst during the stance to swing transition. At 0% BWS, VL showed an initial burst of activity occurring within the first 10% of the gait cycle and a second burst appearing during single limb support at midstance while the limb was fully loaded. At 40% BWS a remarkable change in EMG profile was evident, with a decrease in burst duration resulting in proper timing relative to the gait cycle as that seen in the normal subject (Figure 4.1.2A). In TA the abnormal burst seen during midstance at 0% BWS was absent at 40% BWS. The prolonged burst of GA lasting throughout the stance phase was replaced by a profile showing enhanced activity at

FIGURE 4.1.4: A) Comparison of the MH, VL, TA and GA EMG mean ensemble average (across 10 strides) between 0% and 40% BWS trials in a moderately spastic patient (MB) during treadmill locomotion at 0.30 ms^{-1} and B) the corresponding saggittal angular excursions of a representative cycle for the trunk, hip, knee and ankle. Stance/swing transition at 0% BWS: 64.1% gait cycle; at 40% BWS: 62.4% gait cycle.


MEDIAL HAMSTRINGS
TIBIALIS ANTERIOR
A
VASTUS LATERALIS
GASTROCNEMIUS
μv
0% BWS
40% BWS
B
TRUNK ANGLE °
HIP ANGLE °
KNEE ANGLE °
ANKLE ANGLE °
% GAIT CYCLE
0% BWS
40% BWS

0, 40 and 70% of the gait cycle during 40% BWS. More appropriate timing between TA and GA was also noted with 40% BWS.

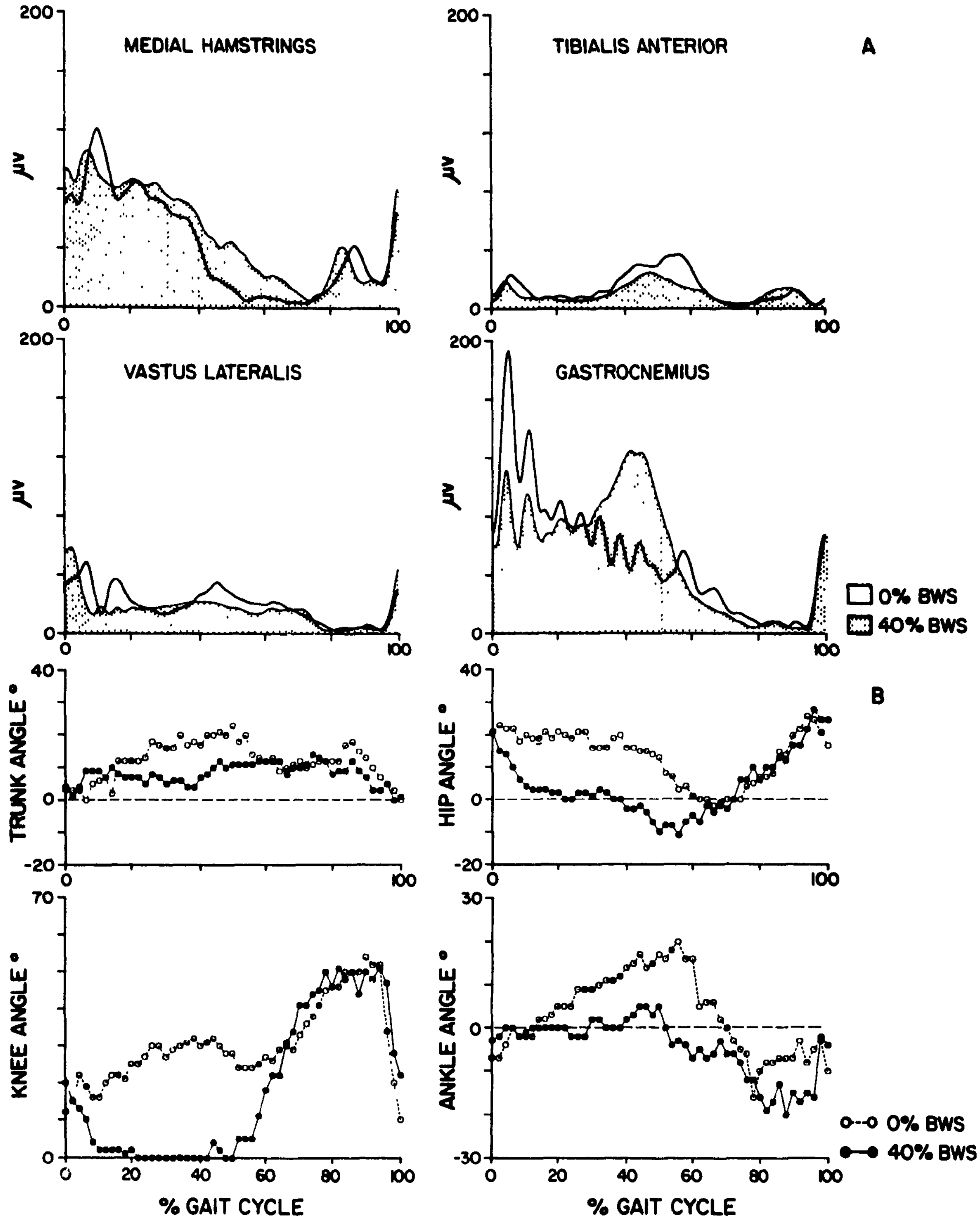
Figure 4.1.4B shows the sagittal angular displacement of the trunk, hip, knee and ankle throughout the gait cycle. There was a straighter trunk alignment at 40% BWS when compared to 0% (Figure 4.1.4B). The hip angular excursion pattern was near normal at 0% BWS and remained unchanged with 40% BWS (Figure 4.1.4B). At 0% BWS initial foot floor contact occurred with a flexed knee with subsequent knee extension (0°) being maintained throughout the loading phase. At 40% BWS, the knee angular excursion pattern approaches that of the normal subject (Figure 4.1.2B). A straighter alignment of the knee at heel strike was noted, with progressive extension until midstance, followed by flexion in late stance and midswing. The maximum swing angle of the knee occurred earlier in the gait cycle (75%). The straighter knee alignment at initial foot-floor contact seen at 40% BWS, allows the dorsiflexed foot to make initial contact with a heel strike rather than the entire sole of the foot as is seen at 0% BWS (Figure 4.1.4B). This change in the sagittal angular displacement of the ankle may be responsible for the more normal GA EMG profile. Following initial foot floor contact with the entire sole of the foot at 0% BWS there was premature stretching of GA as the body moved over an already stationary foot. At 40% BWS, following heel strike, the dorsiflexed ankle plantarflexed to place the sole of the foot on the ground and GA was shortened early in the stance phase. This delays the stretch on GA to later in the stance phase as observed in the normal subject. The burst in GA seen at heel strike at 40% BWS may be due to a stretch on the muscle from the knee being extended and the ankle dorsiflexed. The early recruitment of GA in late swing may also be centrally activated as has been reported in immature gait (Sutherland et al, 1980; Forssberg, 1985).

Case 3: BM (Speed=0.26 ms^{-1})

Figures 4.1.5A and B present the data of a severely spastic subject, BM, at 0% and 40% BWS. The most marked change in EMG activity was seen in GA where the sustained clonic activity during stance at 0% BWS was diminished and a burst appeared with proper timing in relation to the occurence of push-off during the gait cycle (Figure 4.1.5A). At 40% BWS, VL also showed the appearance of a small burst early in the stance phase. Minimal changes were noted for MH and TA with BWS.

The important changes in the joint angular displacement patterns seen at 40% BWS were a straighter trunk alignment during the weight bearing phase and the presence of hip extension (from -2° to -10°) in the latter part of stance and initial swing. At 0% BWS the knee was maintained in constant flexion throughout the gait cycle, forcing the ankle into excessive dorsiflexion during the loading phase. This may have been responsible for the sustained clonus seen throughout the stance phase. At 40% BWS, following initial foot-floor contact with a flexed knee, the knee progressively assumed an extended position during the weight bearing phase resulting in full knee extension during midstance. Associated with the overall decrease in trunk, hip, and knee flexion at 40% BWS, was a more normal ankle excursion

FIGURE 4.1.5: A) Comparison of the MH, VL, TA and GA EMG mean ensemble average (across 10 strides) between 0% and 40% BWS trials in a severely spastic patient (BM) during treadmill locomotion at 0.26 ms^{-1} and B) the corresponding saggittal angular excursions of a representative cycle for the trunk, hip, knee and ankle. Stance/swing transition at 0% BWS: 68.3% gait cycle; at 40% BWS: 72.2% gait cycle.


A
B
MEDIAL HAMSTRINGS
TIBIALIS ANTERIOR
VASTUS LATERALIS
GASTROCNEMIUS
µV
0% BWS
40% BWS
TRUNK ANGLE °
HIP ANGLE °
KNEE ANGLE °
ANKLE ANGLE °
% GAIT CYCLE
o--o 0% BWS
●—● 40% BWS

pattern with a decrease in excessive dorsiflexion during mid and late stance (Figure 4.1.5B).

Global Effects of the BWS on the 7 Subjects:

In general, there was a decrease in mean burst amplitude for most muscles in all seven subjects, with certain muscles showing a more appropiate EMG timing as evidenced by the examples presented above.

The changes in joint angular displacement patterns seen in the seven subjects revealed that four subjects walking with a flexed posture at 0% BWS demonstrated a straighter trunk alignment at 40% BWS. Four of the seven subjects having knee flexion during initial contact at 0% BWS showed greater knee extension at 40% BWS (19° to 14°; 28° to 18°; 35° to 22° and 35° to 15°). Two out of three subjects having excessive knee flexion during midstance at 0% BWS gained knee extension at 40% BWS (22° to 16° and 35° to 5°). Qualitatively, six of the seven subjects demonstrated smoother, freer movements of the lower limbs at 40% BWS.

Table 4.1.2 contains a summary of the temporal distance data collected on the seven subjects while walking on the treadmill at 0% and 40% BWS. There was an increase in cycle duration ranging from 5.6% to 20.6% for six subjects at 40% BWS. This resulted in an increase in stride length ranging from 13.7% to 42.9% in the same subjects. Single limb support time was also found to increase in six subjects with increases ranging from 3.3% to 64.3%. A decrease in percentage stance with BWS was also noted in 6 subjects. In two of these subjects (MP and RP) the prolonged stance phase approached normal values (55.4 and 63.1 respectively) at

TABLE 4.1.2: Temporal Distance Parameters

The average cycle duration (ms) ; single limb support time (SLST) (ms); % stance; % total double support time (TDST); and stride length (cm) recorded at 0% and 40% BWS at the same treadmill speed are represented. The maximum comfortable treadmill speed (ms^{-1}) at 0% and 40% BWS are also shown. The averages are calculated across 10 strides. The number in parenthesis represents 1 standard deviation.

SUBJECT	CYCLE DURATION ms 0%	CYCLE DURATION ms 40%	SLST ms 0%	SLST ms 40%	STANCE % 0%	STANCE % 40%	TDST % 0%	TDST % 40%	STRIDE LENGTH cm 0%	STRIDE LENGTH cm 40%	SPEED ms^{-1} 0%	SPEED ms^{-1} 40%
MP	1592 (84)	1789 (69)	395 (46)	557 (34)	72.6 (3.3)	55.4 (2.6)	45.3 (4.7)	22.1 (3.3)	29.0	33.0	0.43	0.62
RP	1080 (53)	1160 (52)	337 (45)	348 (41)	71.3 (2.7)	63.1 (2.8)	37.7 (3.3)	29.6 (4.1)	14.0	18.0	0.39	0.43
MB	1763 (58)	2126 (57)	614 (58)	739 (37)	64.1 (2.8)	62.4 (1.0)	27.3 (2.4)	26.0 (1.9)	25.0	30.0	0.30	0.36
RM	3074 (136)	2963 (90)	1205 (68)	1313 (78)	69.3 (1.8)	66.3 (1.9)	29.0 (1.8)	20.6 (2.3)	41.0	37.0	0.30	0.31
BM	3165 (146)	3341 (119)	384 (48)	631 (90)	68.3 (5.6)	72.2 (2.2)	55.0 (5.3)	52.1 (3.8)	39.0	50.0	0.26	0.32
RL	2230 (205)	2608 (143)	500 (135)	476 (51)	85.6 (3.3)	81.9 (3.6)	61.0 (7.4)	62.1 (5.1)	21.0	30.0	0.26	0.35
SQ	3596 (254)	4041 (333)	1127 (188)	1407 (259)	68.4 (6.9)	65.5 (7.2)	34.5 (6.9)	32.0 (6.7)	47.0	61.0	0.26	0.26

40% BWS. A further finding was a decrease in percentage total double support time (% TDST) in 6 subjects with the more marked decreases seen in the mildly and moderately spastic subjects (MP, RP and RM). Another important finding was the ability of the 5 subjects to walk at higher maximal comfortable walking speeds at 40% BWS. The increase in speed ranged from 10.3% to 44.2%.

All subjects subjectively reported it was easier and less fatiguing to walk at 40% BWS than at 0% BWS. This was reflected in their walking tolerance which increased at 40% BWS, resulting in walking trials which were up to twice as long as at 0% BWS (0% BWS=2.5 $\pm$ 1.4 min; 40% BWS=3.7 $\pm$ 2.0 min). All subjects also showed a less elevated heart rate (8% to 19%) following the 40% BWS trial than following the 0% BWS trail indicating greater cardio-pulmonary efficiency when walking with BWS.

Discussion:

EMG Activity:

The EMG profiles of lower limb muscles of spastic paretic subjects have been described as having early recruitment with prolonged activation during the stance phase (Knutsson and Richards, 1979; Conrad et al, 1985; Benecke and Conrad, 1986). The authors attributed these deviations of the EMG profiles to exaggerated stretch reflexes since most of the activity in the muscles coincided with instances of muscle lengthening during the gait cycle. Four of the subjects in this study revealed prolonged activation of all four lower extremity muscles being investigated throughout the stance phase. Such a pattern of muscle activation can be modified with BWS. As can be seen in MB's GA EMG profile, the prolonged activation seen at 0% BWS was replaced by a profile showing enhanced activity between 20 and 60% of the gait cycle required for push-off at 40% BWS. This is most probably due to the more normal angular excursion occurring at the knee and ankle which could decrease premature stretch on the GA during initial loading.

In the more severely spastic patient (BM), the enhanced stretch reflexes in GA produced sustained clonus during the entire stance phase at 0% BWS. At 40% BWS some clonus was still present, but it ceased in midstance, and a burst of activity in GA appeared between 20 and 60% of the gait cycle. This coincided with an increase in plantarflexion at the time of push-off. Decreasing the load on the lower extremities also resulted in a straighter trunk, hip, and knee which then allowed the ankle to be in a neutral position during midstance, thereby decreasing the stretch on the triceps surae. This may explain the decrease in clonus, thus possibly allowing the GA to be activated and produce a burst for push-off.

Prolonged muscle activation of proximal muscles in spastic paretic gait has also been described by Knutsson (1983). The author observed prolonged activation in quadriceps and abductor muscles during the stance phase and defined the phenomenon as 'crutch spasticity' resulting from a response to load and tonic stretch activation. An example of crutch spasticity can be seen in the VL for subject MB. In this subject the problem of prolonged activation was alleviated by decreasing the load on the lower extremities. At 40% BWS, VL has a more normal EMG profile, showing a burst of activity early in the stance phase with a definite silent period in late stance.

Joint Angular Displacement Patterns:

The more important changes in joint angular displacement patterns at 40% BWS were seen during the loading phases of gait. The straighter knee alignment during initial loading and midstance indicated that the subjects were able to bear weight on the lower extremities without assuming the flexed posture which is characteristic of spastic gait (Conrad et al, 1985). Therefore, decreasing the load led to more normal joint angular displacement profiles. The occurrence of hip extension in BM at 40% BWS is an important finding considering that hip extension plays an important role in the initiation of flexion during locomotion (Grillner and Rossignol, 1978). For this subject, the increase in hip extension resulted in an increase in stride length.

Temporal Distance Parameters:

The changes seen in the temporal distance parameters during weight supported locomotion provides further evidence that this strategy allows the patient to cope better with the loading of the lower extremities. The increase in cycle duration and accompanying increase in stride length, seen at 40% BWS, at a constant treadmill speed, indicates that the subjects are able to take longer steps. This may be a result of the subject's ability to bear weight on the affected lower extremity for longer periods of time therefore allowing the contralateral limb to take a longer step. Single limb support time (SLST) is a critical component of gait (Perry,

1969) requiring both the ability to balance and bear weight while the ipsilateral limb is loaded during the swing phase of the contralateral limb. Neurologically impaired gait is characterized by a shortened contralateral swing phase resulting in a shortened SLST of the affected lower limb due to an inability to adequately bear weight (Brandstater et al, 1983). Six of the spastic paretic subjects in this study showed an increase in SLST at 40% BWS. This indicated that with BWS the subjects are able to cope with the loading phase of gait and can bear weight on the limb for longer periods of time. At 40% BWS, a decrease in % TDST was also noted in the majority of subjects. This demonstrates that the subjects may be able to transfer their weight from one limb to the other with greater ease, requiring less support from both limbs during the loading phases.

Speed is a temporal parameter of gait associated with higher levels of lower limb motor recovery (Brandstater et al, 1983) and locomotor function (Mizrahi et al, 1982). Spastic paretic patients walk at speed considerably lower than normals (Conrad et al, 1985). Muscle hypertonia has been suggested as one of the causes for the spastic paretic subject's inability to walk at faster speeds (Dietz, 1986). The smoother, less spastic gait evidenced at 40% BWS was accompanied by an increase in comfortable walking speed. This suggests that decreasing load on the lower extremities facilitates locomotion, allowing spastic paretic subjects to walk faster. This may be due, in part, to the decreasing influence of exaggerated stretch reflexes with BWS, and may also be related to the increase in stride length, decrease in TDST, and increase in SLST which were observed.

New Gait Training Strategy:

It is a common clinical finding that neurologically impaired patients have difficulty weight bearing through their affected lower extremities during ambulation. One of the limitations of conventional gait retraining is that it is done under full weight bearing conditions, most of the time using parallel bars and walking aids to alleviate the load on the lower extremities. A dynamic and task specific approach consisting of progressive weight bearing during treadmill locomotion may be an effective strategy to retrain neurologically impaired gait. The clinical implications and advantages of such a strategy are numerous. The three components of gait: weight bearing, balance and stepping could be retrained simultaneously under dynamic conditions. Gait retraining could be initiated early in the rehabilitation period, providing as much BWS as needed to assume an upright position and allow for assisted or unassisted stepping of the lower limbs. This approach would enable 'interactive locomotor training' (Rossignol et al, 1986; Barbeau and Rossignol, 1987). As the patient walks on the treadmill with a reduced load on the lower extremities gait deviations can instantly be corrected and peripheral stimulation can be provided to facilitate muscle activation during the stance or swing phase.

Validation of this novel gait training strategy is in its preliminary stages. The effects of BWS need to be investigated in a larger group of subjects stratified according to the degree of spasticity to determine which subjects would respond more favorably to BWS. The effects of supporting varying levels of BWS from 0% to 40% also need to be investigated. This would help to define the criteria by which

BWS should be decreased during training. One important issue that needs to be resolved is whether the more normal gait pattern elicited with BWS can be retained and carried over to full weight bearing conditions following a training regimen incorporating BWS where weight bearing through the lower extremities progressively increased. Preliminary studies suggest that interactive gait training with BWS was important in optimizing the locomotor pattern and acheiving full weight bearing in 2 spastic paretic subjects (Fung et al, 1988).

Conclusion:

BWS during treadmill locomotion in spastic paretic subjects appears to alleviate some of the problems of early stretch and prolonged muscle activation encountered due to their inability to cope with loading under full weight bearing conditions. Decreasing the load on the lower extremities allowed for more normal timing of EMG activity. The straighter trunk and knee alignment during the loading phase, accompanied by a decrease in TDST and an increase in SLST, stride length and speed suggests that BWS facilitates the expression of the locomotor pattern. Consequently it is proposed that the use of BWS during gait training presents potential as a therapeutic approach to retrain gait in neurologically impaired patients.

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4.2 The Effects of Parallel Bars, Speed and Body Weight Support on the Modulation of the Locomotor Pattern of Spastic Paretic Gait

The effects of walking with and without parallel bars, increasing speed, and providing 40% body weight support (BWS) on the gait pattern of spastic paretic subjects during treadmill locomotion were investigated. In asymmetrically involved subjects, walking without parallel bars led to a more symmetrical gait pattern with decreased compensation of the less involved side. This was accompanied by changes in electromyographic (EMG) and sagittal angular displacement profiles which favoured a more normal swing phase. When symmetrically involved subjects walked without parallel bars, increases in EMG activity, with prolonged activation during the stance phase were noted especially in the distal muscles. Increasing treadmill speed increased or elicited clonus in some subjects and only caused a small increase in EMG amplitude, while not eliciting abnormal reflex activity, in other subjects. Providing 40% BWS led to a decrease in clonus which resulted during more demanding external conditions such as walking without parallel bars or at higher speeds. When excessive clonus was present in the distal muscles of severely impaired subjects, BWS had minimal effect on gait parameters. Forty percent BWS facilitated gait when walking without parallel bars therefore decreasing compensatory gait patterns seen with parallel bars. Implications for gait training are discussed.

THE EFFECTS OF PARALLEL BARS, SPEED AND BODY WEIGHT SUPPORT ON THE MODULATION OF THE LOCOMOTOR PATTERN OF SPASTIC PARETIC GAIT

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Introduction:

Incomplete spinal cord lesions in man commonly result in disturbances of the locomotor pattern. Altered recruitment patterns with premature muscle activation, as well as prolonged electromyographic (EMG) activity and delayed muscle relaxation have been identified in paraspastic gait (Conrad et al, 1985). The pattern may also be coupled with flattening of the EMG profiles, characterized by diminished or abolished dynamic peaks of EMG activity. Knutsson (1980) described 3 types of abnormal activation patterns, consisting of i) early stretch activation of distal muscles, ii) paresis and iii) abnormal co-activation of agonist and antagonist lower limb muscles, in response to loading. One or more of these impaired activation patterns could be identified to interfere with gait following a spinal cord lesion. The complexity of abnormal activation patterns identified following a spinal cord lesion makes it difficult to establish a universal gait training strategy.

External factors such as parallel bars, speed and body weight support can influence the locomotor pattern during gait training. The use of parallel bars during gait training appears controversial. Conrad et al (1983; 1985) report that the decreased stability incurred during treadmill locomotion without parallel bars leads to a deterioration in the gait pattern of paraspastic patients. In contrast, conventional gait training for neurological patients discourages the use of parallel bars as it is believed to lead to an asymmetrical gait with compensation of the less involved side (Bobath, 1978).

An additional concern while retraining gait is the characteristic slow walking speeds evident among spastic patients (Barbeau el al, 1988; Visintin et al, 1989).

Gait training aims at increasing the walking speed, attempting to make the gait more efficient. However, little quantitative data exists to elucidate the effects of walking speed on a spastic paretic gait pattern.

A new gait training strategy which provides body weight support (**BWS**) during treadmill locomotion has been proposed to retrain gait following a spinal cord lesion. BWS has been shown to facilitate gait and elicit a more normal gait pattern with respect to EMG, sagittal angular displacement patterns and temporal distance parameters in a group of spastic paretic subjects during treadmill locomotion at their comfortable speed (Visintin and Barbeau, 1989). In order to establish a comprehensive treatment approach using BWS, its effects in combination with other external factors such as parallel bars and speed remain to be investigated.

It is the aim of this study to quantitatively and qualitatively describe the effects of parallel bars, speed, and BWS on the locomotor pattern of spastic paretic subjects. The influence of BWS while walking with and without parallel bars and at various treadmill speeds will be addressed.

Methods:

The study was conducted in the Human Gait Laboratory which has previously been described in detail (Barbeau et al, 1987). Eight spastic paretic subjects, ranging in age from 22 to 42 years (mean = 27.5 years), participated in this study. Seven subjects had sustained a traumatic incomplete spinal cord lesion to the cervical or thoracic spine, one subject suffered from non-familial progressive spastic paraparesis and one subject had a surgically induced lesion at the level of T-10 following resection of a spinal tumor. Each subject was capable of independent overground locomotion with or without the use of external aids. The chronicity of the lesion ranged from 7 months to 21 years. Demographic data and the details of a neurological clinical evaluation, including reflex activity, muscle strength, sensation and an overground ambulation profile for each subject are summarized in Table 4.2.1.

Body Weight Support:

The subjects walked on a treadmill while 0% (full weight bearing) and 40% BWS was provided. Previous experience with the BWS system revealed that providing levels of BWS higher than 40% resulted in loss of heel-ground contact for some patients. Hence, 40% BWS was the maximum level of BWS provided. The BWS apparatus consisted of a custom-designed harness which mechanically supported the patient vertically over the treadmill. The harness consisted of a pelvic band attached around the hips and two padded straps which pass between the legs to attach anteriorly to the pelvic band. The percentage of BWS provided was calibrated using a force transducer. The force was normalized to each subject's weight (100%) and the sequence of % BWS provided was randomly assigned into two trials given within the same experimental session. Prior to data collection, each subject was habituated at 0% BWS for 1 to 5 minutes depending on his walking tolerance. During this trial, treadmill speed was slowly increased from 0 ms^{-1} to each subject's comfortable walking speed. Once the comfortable walking speed was identified, it was determined whether the subjects could walk at more than one treadmill speed. All subjects, except CF, were able to walk at a minimal and

TABLE I: DEMOGRAPHIC DATA

A description of subjects participating in the study with regards to sex, age, level of lesion and chronicity are tabulated. The results of a clinical neurological examination are presented as well. The following grades were used:

muscle tone: + minimal response to passive movement

++ moderate response to passive movement

+++ severe response to passive movement

ankle clonus: + 1 to 2 beats of clonus

++ unsustained clonus

+++ sustained clonus

muscle strength: only those muscles with a grade less than 3/5 (active movement against gravity) are identified

abbreviations: SP:spastic paraparesis; C:cervical spine; T:thoracic spine; bil:bilateral; (r):right; (l):left; ext:extension; abd:abduction; flex: flexion; ↓:decrease

SUBJECT SEX AGE	LEVEL OF LESION	CHRONICITY (YEARS)	MUSCLE TONE		ANKLE CLONUS	MUSCLE STRENGTH	KINESTHESIA POSITION SENSE	SYMMETRICAL INVOLVEMENT	COMFORTABLE TREADMILL SPEED (ms^{-1})	MAXIMAL TREADMILL SPEED (ms^{-1})	OVERGROUND WALKING AIDS
			KNEE	ANKLE							
BP M 32	SP	3.0	+++bil	+++bil	+(r) ++(l)	all muscles > 3	(l) ↓ hip/knee/ ankle (r) ↓ knee/ankle	yes	0.40	0.60	bil foot drop braces
CF M 23	C6-7	1.5	+++(r) ++(l)	+++(r) ++(l)	+++bil	hip ext 2+bil abd 2+bil knee flex 1+(r) ankle TA 0(r) SOL 1(r)	unable to test due to marked extensor tone	(r>l)	0.08	0.08	(l) long leg brace; canadian crutches
LR F 22	T10	1.5	+++(r)	+++(r)	+++(r) ++(l)	hip flex 2(r) abd 2+(r) knee flex 1(r) ankle TA 0(r)	(r) intact (l) ↓ hip and knee absent ankle	(l>r)	0.10	0.30	(r) foot drop brace; 1 cane
RC M 23	C6-7	4.0	++(l) +++(r)	++(l) +++(r)	+++bil	Hip abd 2 bil	(r) ↓ hip/knee/ankle	yes	0.40	0.40	2 canes
LL M 32	C7-T1	1.5	+++bil	+++bil	+++bil	hip flex 2(r),2+(l) ext 1+ bil abd 0(r),2+(l) knee flex 2+(r) ankle TA 2+(r) SOL 1(r),1+(l)	(l) ↓ ankle (r) ↓ hip/ankle	yes	0.15	0.25	walker
BG F 26	C1-2	21.0	++(l)	++(l)	++bil	all muscles > 3	intact bil	yes	0.43	not tested	no aids
FS M 20	T8-9	1.5	+++bil	+++bil		hip ext 2(r),2+(l) abd 2+ bil	(l) ankle absent	yes	0.15	0.15	2 canes
JS M 42	T9-10	0.6	++(r) ++(l)	+++bil	+++bil	hip flex 2 bil ext 2 bil knee flex 1+(r),1(l) ankle SOL 2+(r),0(l) TA 2(l)	(l) ↓ ankle	(l>r)	0.10	0.15	(l) foot drop brace walker

comfortable treadmill speed. Four subjects (JS, LR, BP, LL) were able to walk at a maximal treadmill speed. Data were then collected for pre-determined speeds (minimal, comfortable and maximal speeds) during subtrials at 0% and 40% BWS with parallel bars. In addition, for those subjects who were able to walk without parallel bars at minimal or comfortable speeds (subjects: CF, LR, BG, BP), data were collected during these instances at 0% and 40% BWS. Of the 8 subjects, only 1 subject (JS) required a short foot drop brace on his left ankle to walk on the treadmill during data collection.

A 10-minute rest period was given between each BWS trial to prevent fatique. Blood pressure and pulse were monitored following each trial to control for undue stress on the subjects.

EMG and Footswitch Data:

EMG activity was recorded from the gluteus maximus (GM), vastus lateralis (VL), medial hamstrings (MH), tibialis anterior (TA), medial gastrocnemius (GA) lateral soleus (SOL) of either the left (subjects: BG, RC, LL, FS, BP) or right (subjects: CF, LR, JS) lower limb while the subject walked on the treadmill. When subjects presented with asymmetrical involvement of the lower extremities (i.e. one limb showing minimal spasticity and weakness), data were collected from the more involved lower limb. When the degree of involvement of the lower limbs was largely symmetrical, data were then collected from the limb that showed a greater degree of clinical spasticity. Bipolar surface electrodes (2.5 cm centre to centre) were placed over the belly of each muscle following conventional skin preparation. The EMG signals were preamplified, differentially amplified and bandpassed (10-450 Hz).

Footswitches placed under the heel, fifth metatarsal head and big toe of each subject's shoes were used to detect heel strike, foot flat and toe off, allowing for the determination of the temporal distance paramters.The EMG and footswitch signals were then recorded at 3.75 IPS on a 14 channel FM tape with a frequency response of 2500 Hz.

The EMG signals of the 6 muscles along with the footswitch signals were then printed out on polygragh and an artifact-free sequence of 10 or more consecutive steps was chosen to represent each experimental paradigm for each subject.

Joint Angular Displacement Data:

Joint angular displacement data were collected from the same limb as the EMG recordings for each subject. The subjects were videotaped as they walked on the treadmill using a shutter video camera. Reflective joint markers were placed at the shoulder, hip, knee and ankle as well the heel, fifth metatarsal head and the toe region of the lateral border of the shoe. Additional markers were placed on a vertical and a horizontal bar to be used as absolute coordinates for the video analysis. The trials were recorded on a 3/4 inch videotape at a speed of 60 fields per second. A remote search controller was used for field by field viewing. The sagittal angular displacements were manually measured from the monitor screen using a goniometer. Once the subjects had reached a steady state while walking on the treadmill, one representative gait cycle for each subject during each experimental paradigm was analyzed. The joint angular displacements were measured at every 5% of the gait cycle. The trunk and hip angles were calculated with respect to a vertical line, with the neutral position in standing being taken as 0° displacement of the trunk and hip, flexion being positive, and extension negative. Likewise, in calculating the knee and ankle angles, the neutral standing position, with the knee at full extension and the shank axis perpendicular to the foot, was taken as 0°. Knee flexion and ankle dorsiflexion beyond neutral was taken as positive angular displacements, and ankle plantarflexion beyond neutral was taken as negative angular displacement.

Results:

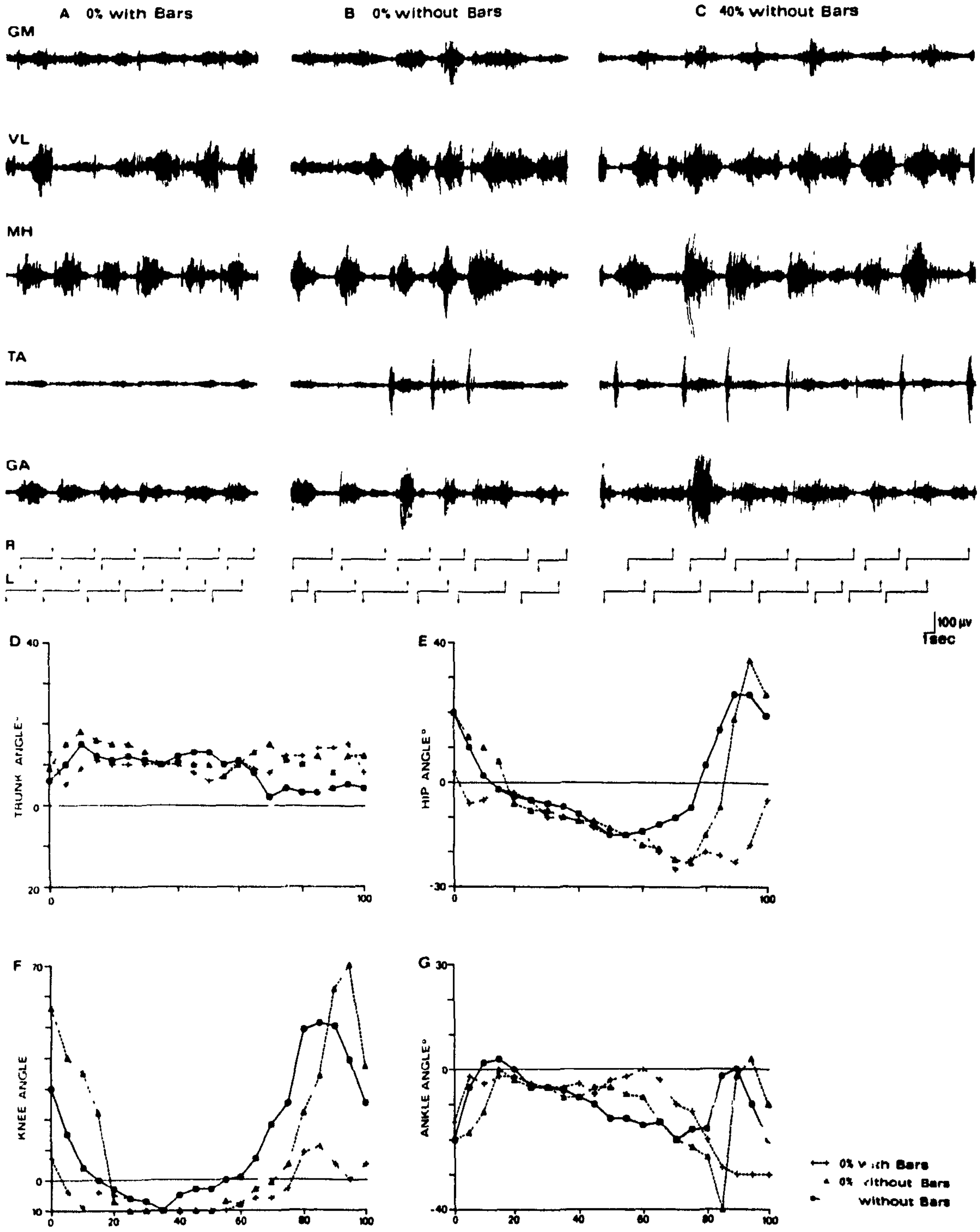
In this section the individual effects of parallel bars, treadmill speed, and BWS on spastic paretic gait will be presented. The effects of providing 40% BWS with and without parallel bars, and at different treadmill speeds will also be addressed.

Effects of Parallel Bar Support:

Removing parallel bar support during treadmill locomotion at comfortable speeds, with 0% BWS, had either a positive or negative influence on gait parameters depending on whether the subjects were presented with an asymmetrical or symmetrical gait pattern.

Figures 4.2.1A and B contrast the EMG patterns of a subject (CF) who had a more impaired involvement of the right lower limb, while he walked with and without parallel bars at 0% BWS at his comfortable treadmill speed of 0.08 ms^{-1}. Without parallel bars, there was persistent activation of GM and VL during the entire gait cycle with activity peaking in mid to latter stance. MH was active from late swing to midstance and a second small burst of activity was seen in early swing. Low level tonic activity appeared in TA during stance. GA showed continuous

Figure 4.2.1: The right lower limb EMG activity of subject CF walking on the treadmill, at a speed of 0.08 ms^{-1}, at A) 0% BWS with parallel bars, B) 0% BWS without parallel bars, and C) 40% BWS without parallel bars. The downward arrows indicate foot-floor contact, while the upward arrows indicate toe-off, with the solid line depicting stance duration and the space denoting swing duration, for both right (R) and left (L) lower limbs. In B, the second, third and fourth steps of the right lower limb represent the 3 steps the subject takes without parallel bars. In B and C, note the burst of activity in TA during the swing phase. In C, note the more definite silent period for GM and VL during early swing. The corresponding sagittal angular displacement patterns of a representative cycle for the D) trunk, E) hip, F) knee and G) ankle are also illustrated. Note the increase in hip and knee flexion and ankle dorsiflexion during the swing phase when walking without parallel bars.

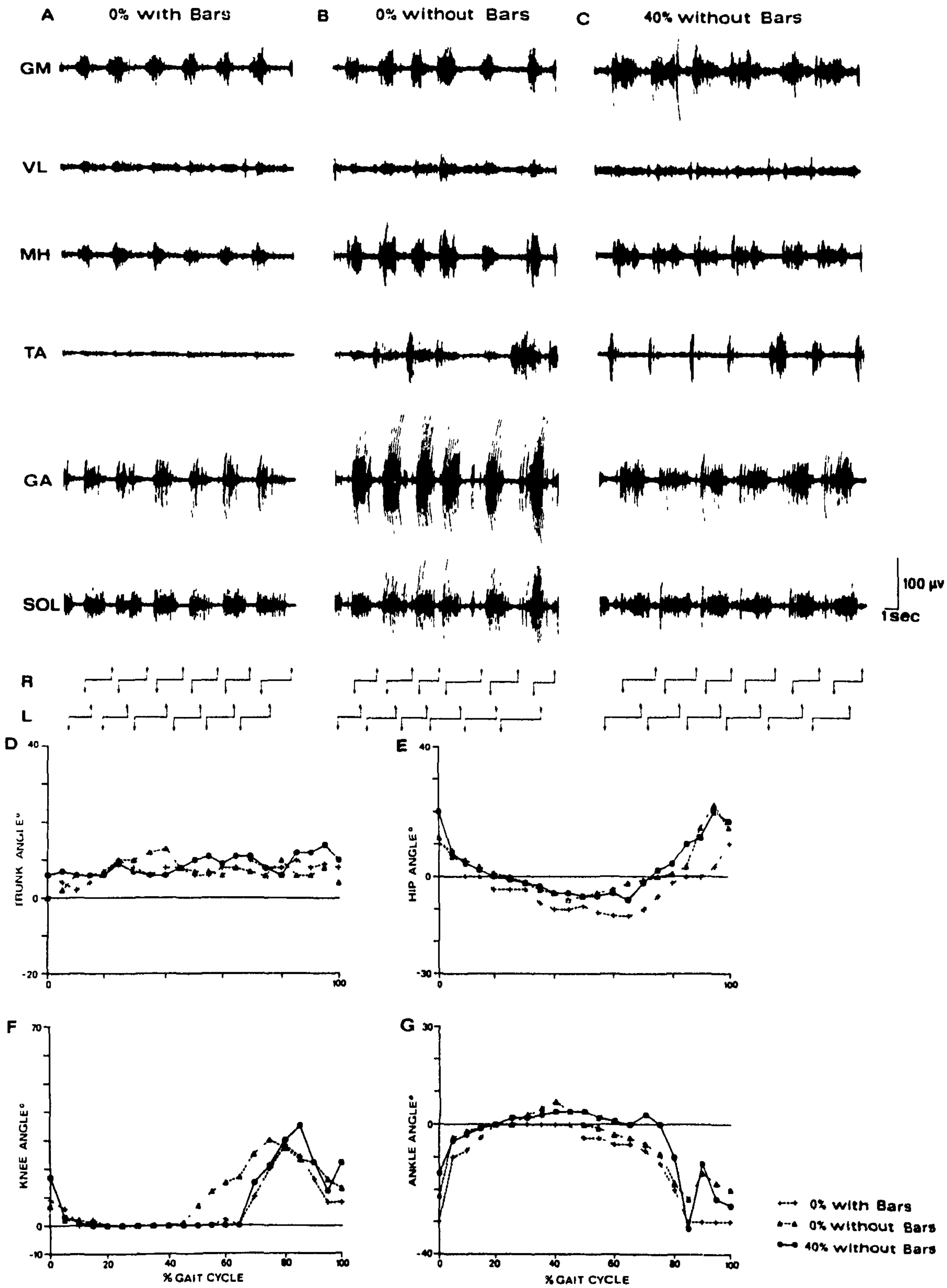

A 0% with Bars
B 0% without Bars
C 40% without Bars
GM
VL
MH
TA
GA
R
L
100 μv
1 sec
D
TRUNK ANGLE°
E
HIP ANGLE°
F
KNEE ANGLE
G
ANKLE ANGLE°
% GAIT CYCLE
0% with Bars
0% without Bars
without Bars

activity which initiated in the swing phase and continued for the entire stance phase. During the same experimental conditions, the subject was able to take only 3 laborious steps without parallel bars (Figure 4.2.1B). The most evident EMG changes were seen during the swing phase. A burst of activity appeared in TA, resulting in ankle dorsiflexion (Figure 4.2.1G); and a decrease in GM and VL activity was seen accompanied by hip and knee flexion (Figures 4.2.1E and F).

This subject's asymmetrical gait pattern consisted of walking on his toes during left stance in order to swing the right leg through while holding it almost fully extended. The sagittal angular displacement patterns, plotted in Figures 4.2.1D to G, for the right lower extremity, revealed no hip or ankle flexion with knee flexion reaching a maximum of (11°) during the swing phase. When the subject released the parallel bars, a marked increase in hip and knee flexion as well as ankle dorsiflexion emerged. Although walking without parallel bars was very difficult for this subject, it facilitated a more normal swing phase by not allowing him to compensate with his less involved side, thereby placing a maximum demand on the more affected lower extremity.

Similar findings were observed in subject LR (Figures 4.2.2A and B), who also displayed an asymmetrical gait pattern by compensating with her less involved right side as seen in subject CF. As the subject ambulated while holding on to the parallel bars, at her comfortable speed of 0.10 ms^{-1} at 0% BWS, there was minimal tonic activity in TA. When walking without parallel bars, a burst of activity became evident in TA during the swing phase with clonus during the stance phase. This corresponded to a more dorsiflexed ankle during the swing phase, as revealed by the

Figure 4.2.2: The right lower limb EMG activity of subject LR walking on the treadmill, at a speed of 0.10 ms^{-1}, at A) 0% BWS with parallel bars, B) 0% BWS without parallel bars, and C) 40% BWS without parallel bars. The downward arrows indicate foot-floor contact, while the upward arrows indicate toe-off, with the solid line depicting stance duration and the space denoting swing duration, for both right (R) and left (L) lower limbs. In B and C, note the burst of activity in TA during the swing phase. In C, note the decrease in clonus in the distal muscles. The corresponding sagittal angular displacement patterns of a representative cycle for the D) trunk, E) hip, F) knee and G) ankle are also illustrated. Note the increase in ankle dorsiflexion during the swing phase when walking without parallel bars.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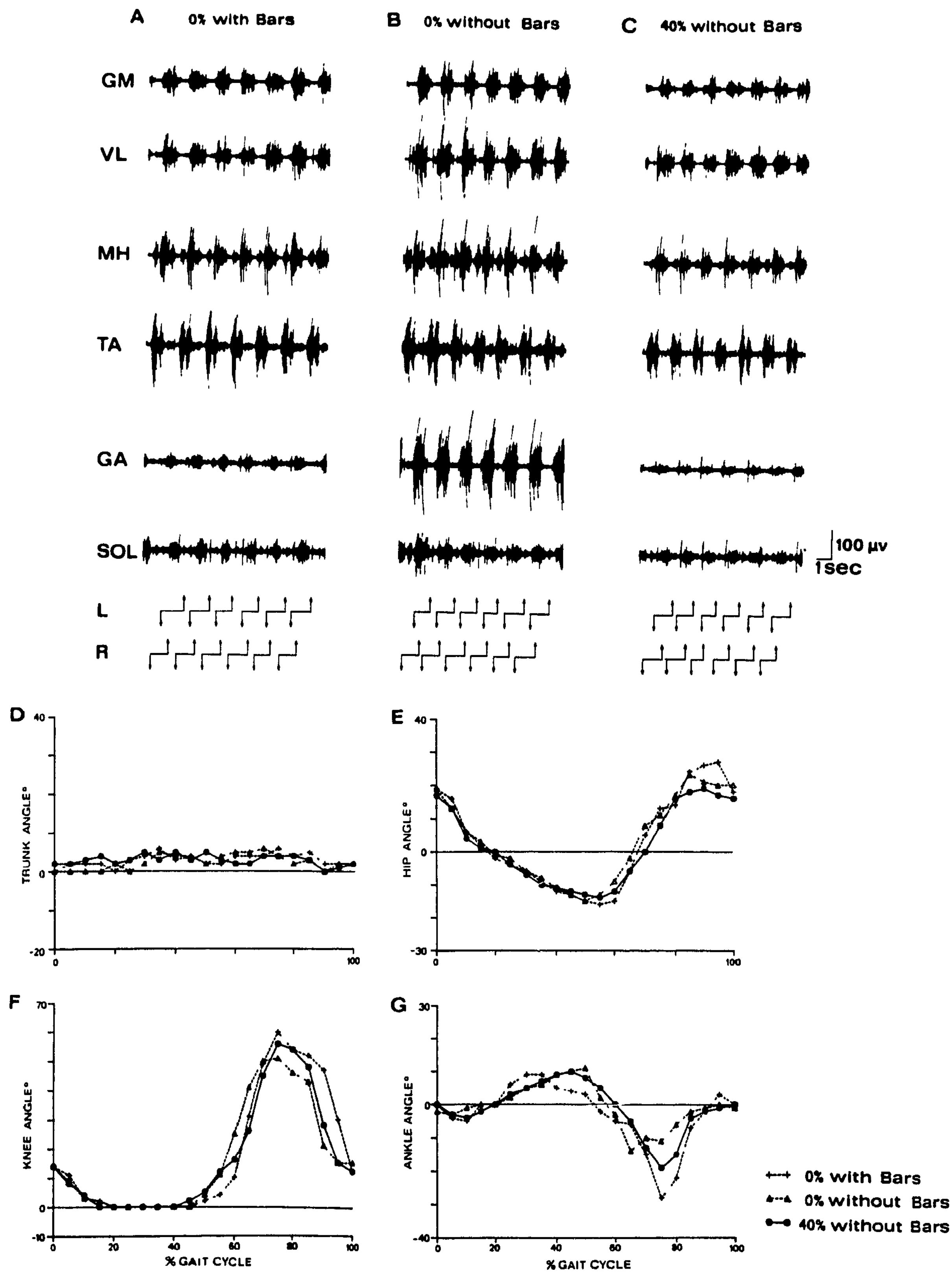
ankle's sagittal angular displacement pattern plotted in Figure 4.2.2G. The problem of increased clonus appearing in the distal muscles, as the subject walked without parallel bars, could be improved by providing 40% BWS (discussed in BWS section).

In subjects with a symmetrical gait pattern, walking without parallel bars produced a broadening and/or increase in amplitude of the EMG activity in lower limb muscles. An example is represented in Figures 4.2.3A and B for subject BG walking at her comfortable treadmill speed of 0.43 ms^{-1} at 0% BWS, with and without parallel bars. Except for prolonged VL activation, this subject's lower limb EMG patterns illustrated in Figure 4.2.3A deviate little from those seen in normal subjects (Hirschberg and Nathanson, 1952; Fung and Barbeau, 1987). When the subject walked without parallel bars, an increase in VL activity was noted especially at foot-floor contact. TA demonstrated an increase in tonic activity during stance. A broadening of both the GA and SOL bursts were noted with early activation at foot-floor contact. This was accompanied by a large increase in EMG amplitude especially evident in GA. Walking without parallel bars resulted in minimal changes in lower limb angular displacement profiles in this subject (Figures 4.2.3D to G). There was a slight decrease in maximum swing angle at the hip and knee as well as a decrease in plantarflexion at the ankle during swing.

In summary, both subjects in this study with an asymmetrical, compensatory gait pattern exhibited changes in their EMG and kinematic profiles which favoured a more normal gait pattern when parallel bars were removed at 0% BWS. On the contrary, subjects with a symmetrical gait pattern and equal involvement of the lower limbs, demonstrated a deterioration in the gait pattern similar to that which has

Figure 4.2.3: The left lower limb EMG activity of subject BG walking on the treadmill, at a speed of 0.43 ms^{-1}, at A) 0% BWS with parallel bars, B) 0% BWS without parallel bars, and C) 40% BWS without parallel bars. The downward arrows indicate foot-floor contact, while the upward arrows indicate toe-off, with the solid line depicting stance duration and the space denoting swing duration, for both left (L) and right (R) lower limbs. In B, note the increase in VL activity as well as the broadening of activity in TA, GA and SOL during stance. The corresponding sagittal angular displacement patterns of a representative cycle for the D) trunk, E) hip, F) knee and G) ankle are also illustrated.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previously been described (Conrad et al, 1983).

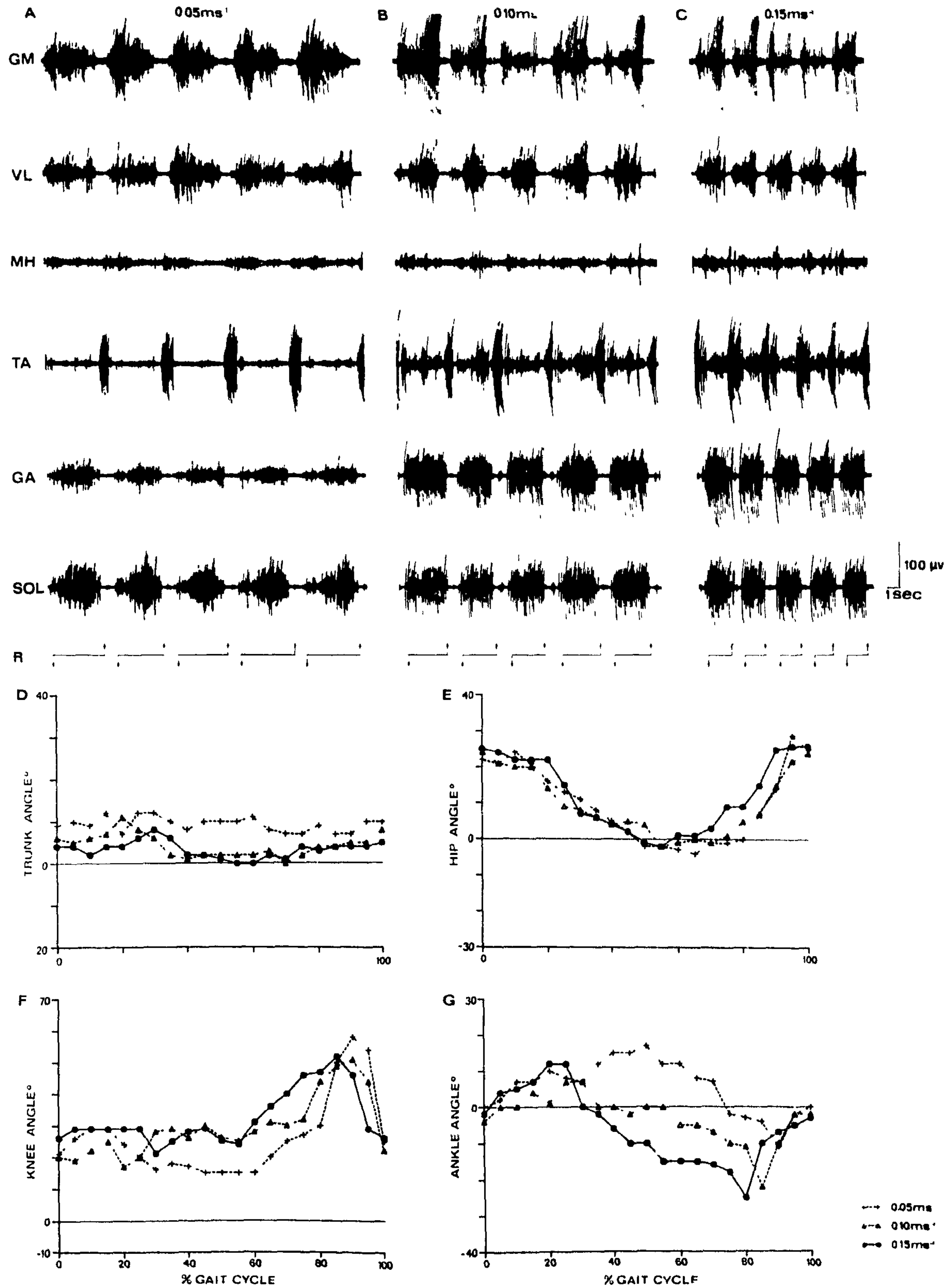
Effects of Speed:

Increasing treadmill speed resulted in an increase in clonus of the distal muscles for subjects JS, FS, and LR. In subject BP clonus, which was not present at the slowest speed, was elicited at the higher speeds. An example is illustrated in Figures 4.2.4A to G for subject JS with increases in treadmill speed from 0.05 ms^{-1} to 0.15 ms^{-1}. At the lowest speed of 0.05 ms^{-1} (Figure 4.2.4A), co-activation of the extensor muscles was noted. There was persistent activity in GM and VL throughout stance. Minimal activity was evident in MH. A burst of activity during the swing phase was noted in TA. GA was characterized by low tonic activity during stance with the onset of low amplitude clonus shortly after initial foot-floor contact. SOL also showed clonus during that period, followed by a burst of activity continuing until terminal stance.

Increasing the treadmill speed to 0.10 ms^{-1} and 0.15 ms^{-1} (Figures 4.2.4B and C) resulted in sustained clonus of all muscles during the entire stance phase and, in some instances, during early swing. This resulted in a laboured gait, especially at the highest speed, with increased difficulty walking and clonic oscillations visible at the ankle. It is interesting that regardless of the presence of clonus at 0.10 ms^{-1}, the subject identified this as his comfortable treadmill speed.

Figure 4.2.4D revealed a straighter trunk alignment at the higher speeds for subject JS. Minimal changes in the hip angular excursion patterns were evident (Figure 4.2.4E). At the higher speeds an increase in knee flexion during mid to terminal stance was noted, with a small decrease in maximum swing angle

Figure 4.2.4: The right lower limb EMG activity of subject JS walking on the treadmill, with parallel bars, at A) 0.05 ms^{-1}, B) 0.10 ms^{-1}, and C) 0.15 ms^{-1}. The downward arrows indicate foot-floor contact, while the upward arrows indicate toe-off, with the solid line depicting stance duration and the space denoting swing duration, for the right (R) lower limb. Note the presence of sustained clonus at the higher treadmill speeds. The corresponding sagittal angular displacement patterns of a representative cycle for the D) trunk, E) hip, F) knee and G) ankle are also illustrated.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0.10ms⁻¹
0.15ms⁻¹

BWS at Higher Speeds:

The effects of BWS at speeds higher than the comfortable treadmill speed was investigated in 4 subjects. In LR and BP increases in walking speed gave rise to an increase in clonus in the distal muscles which subequently showed a decrease in amplitude with 40% BWS at comparable speeds. While subject LL walked at his fastest speed (0.25 ms^{-1}) with 40% BWS, a decrease in the amplitude of all muscles was noted when compared to 0% BWS. In the last subject JS, no effects of 40% BWS at his fastest speed (0.15 ms^{-1}) were noted, as he was presented with general irradiating clonus in all muscles even with minimal effort. In general, all subjects reported it was easier to walk at 40% BWS at the faster speeds, and qualitatively their walking pattern appeared smoother and less spastic.

BWS at Comfortable Treadmill Speed with Parallel Bars:

One of the effects of 40% BWS, while subjects walked at their comfortable speed with parallel bars, was a decrease in EMG amplitude. This was especially evident in subjects who presented with a mild disturbance in the locomotor pattern. An example is illustrated in Figures 4.2.7A and B for subject RC. At 0% BWS, aside from prolonged VL activation and the absence of a TA burst during initial stance, the EMG timing of the other muscles did not deviate markedly from that of a healthy subject (Hirschberg and Nathanson, 1952; Fung and Barbeau, 1987). VL was active just prior to foot-floor contact with activity continuing into latter stance. The effects of providing 40% BWS were a decrease in burst amplitude in GM, GA, and to a lesser degree in SOL. Other subjects who showed similar trends in one or more muscles were CF, BG, FS, LL, and BP. In 4 of these 5 subjects the decrease in

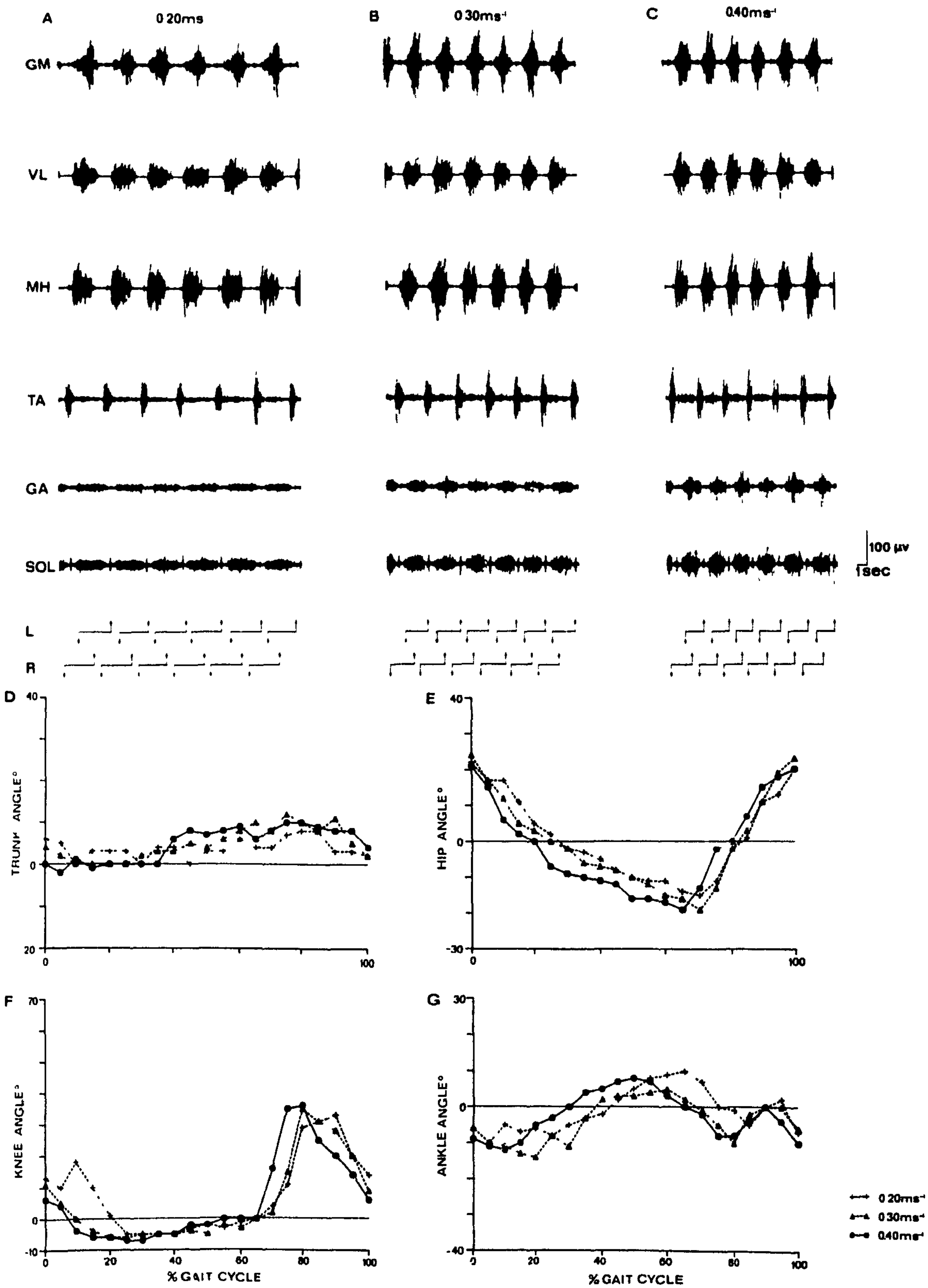
(Figure 4.2.4F). Changes in the ankle kinematic profiles were the occurrence of plantarflexion during midstance as the subject went up on his toes to facilitate swinging the left leg through (Figure 4.2.4G). An increase in plantarflexion at push-off was also noted.

In subjects who showed minimal abnormal reflex activity in the distal muscles during locomotion (LL, RC, BG), increases in speed had a minimal effect on EMG timing and amplitude. This is exemplified in Figures 4.2.5A to C for subject RC during increases in treadmill speed from 0.20 ms^{-1} to 0.40 ms^{1}. At 0.20 ms^{-1}, (Figure 4.2.5A) GM and VL were active from late swing to latter stance. MH showed prolonged EMG pattern with activity initiating in late swing and ceasing in midstance. TA displayed a burst during swing to dorsiflex the ankle and minimal activity during stance. Low burst of tonic activity throughout stance was noted in both GA and SOL.

The effects of increasing treadmill speed to 0.30 ms^{-1} and 0.40 ms^{-1} were to produce a small increase in EMG amplitude for GM, TA, GA and SOL. The subject's fastest walking speed at 0% BWS was 0.40 ms^{-1}. The small burst amplitude of GA and SOL at this speed suggests that the subject may have been unable to generate the force required to walk at faster speeds. The mean peak activity averaged over 10 cycles was 45.07 μV for GA and 64.6 μV for SOL at 0.40 ms^{-1}.

The sagittal angular displacement profiles plotted in Figures 4.2.5D to G revealed that increasing the walking speed had minimal effect on the trunk or hip kinematics. The highest speed resulted in a loss of yield at the knee as well as a decrease in knee flexion of (from 15° to 5°) at initial foot-floor contact, following

Figure 4.2.5: The left lower limb EMG activity of subject RC walking on the treadmill, with parallel bars, at A) 0.20 ms^{-1}, B) 0.30 ms^{-1}, and C) 0.40 ms^{-1}. The downward arrows indicate foot-floor contact, while the upward arrows indicate toe-off, with the solid line depicting stance duration and the space denoting swing duration, for both left (L) and right (R) lower limbs. Note the small increase in EMG amplitude in GM, TA, GA and SOL at the higher treadmill speeds. The corresponding sagittal angular displacement patterns of a representative cycle for the D) trunk, E) hip, F) knee and G) ankle are also illustrated.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0 30ms-1
0.40ms-1

which the knee was kept hyperextended to increase stability during stance. A small increase in ankle plantarflexion was noted during initial swing at the higher speeds. Qualitatively, the gait appeared more laboured at the higher speeds.

In summary, an increase in clonus was evident in the distal muscles at the higher speeds in 4 of the 7 spastic paretic subjects able to walk at more than one treadmill speed. In the other 3 subjects, small increases in burst amplitude of lower limb muscles were noted at the higher speeds. A point of interest was that clonus had been elicited in these 3 subjects when tested at rest during the neurological examination, yet it was not present even at their highest speed while walking on the treadmill. It could be that the walking speed was still too slow to elicit clonus, or that the hyperactive stretch reflexes present at rest were not present during locomotion and therefore did not interfere with the gait pattern.

Effects of BWS:

BWS Without Parallel Bars:

The effects of 40% BWS during treadmill locomotion was to facilitate gait and this was most evident during the more demanding external conditions of walking without parallel bars, especially in severely involved subjects who experienced difficulty during these experimental constraints at 0% BWS.

The gait parameters of a severely impaired subject (CF), walking at 0% and 40% BWS without parallel bars, are represented in Figures 4.2.1B to G. Under these experimental constraints, the subject experienced great difficulty walking at 0% BWS. As he released the parallel bars, his legs dragged at the end of the treadmill following which he was able to initiate and complete only 3 laborious steps. During these steps, both GM and VL showed prolonged EMG activity throughout the stance phase with minimal activity during the swing phase (Figure 4.2.1B). MH activity was initiated in mid to late swing and continued until mid stance. A burst of activity was seen in TA during the swing phase resulting in ankle dorsiflexion for foot-floor clearance. GA was persistently active throughout the stance phase with clonus frequently present.

When 40% BWS was provided (Figure 4.2.1C) a much smoother, less strenuous gait resulted, thereby allowing the subject to take a greater number of steps (a minimum of 10 steps per trial). The EMG patterns revealed a more definite silent period for GM and VL during early swing, accompanied by hip and knee flexion. Minimal effects of 40% BWS were seen in MH and TA. GA showed prolonged activity during stance.

Figures 4.2.1D to G contrast the sagittal angular displacement profiles of the trunk, hip, knee, and ankle at 0% and 40% BWS. At 0% BWS the trunk is maintained in flexion throughout the gait cycle. At 40% BWS, trunk extension appeared during the latter half of the gait cycle as the limb was preparing for swing. A decrease in maximum hip extension was noted (from -23° to -15°) during late stance as well as a decrease in hip flexion (from 35° to 25°) just prior to foot-floor contact at 40% BWS. However, the general trajectory of the hip remained unchanged. At 40% BWS the knee was considerably straighter at foot-floor contact (30°) when compared to 0% BWS (56°). The most remarkable change seen at the ankle with 40% BWS was an earlier initiation of dorsiflexion in late stance.

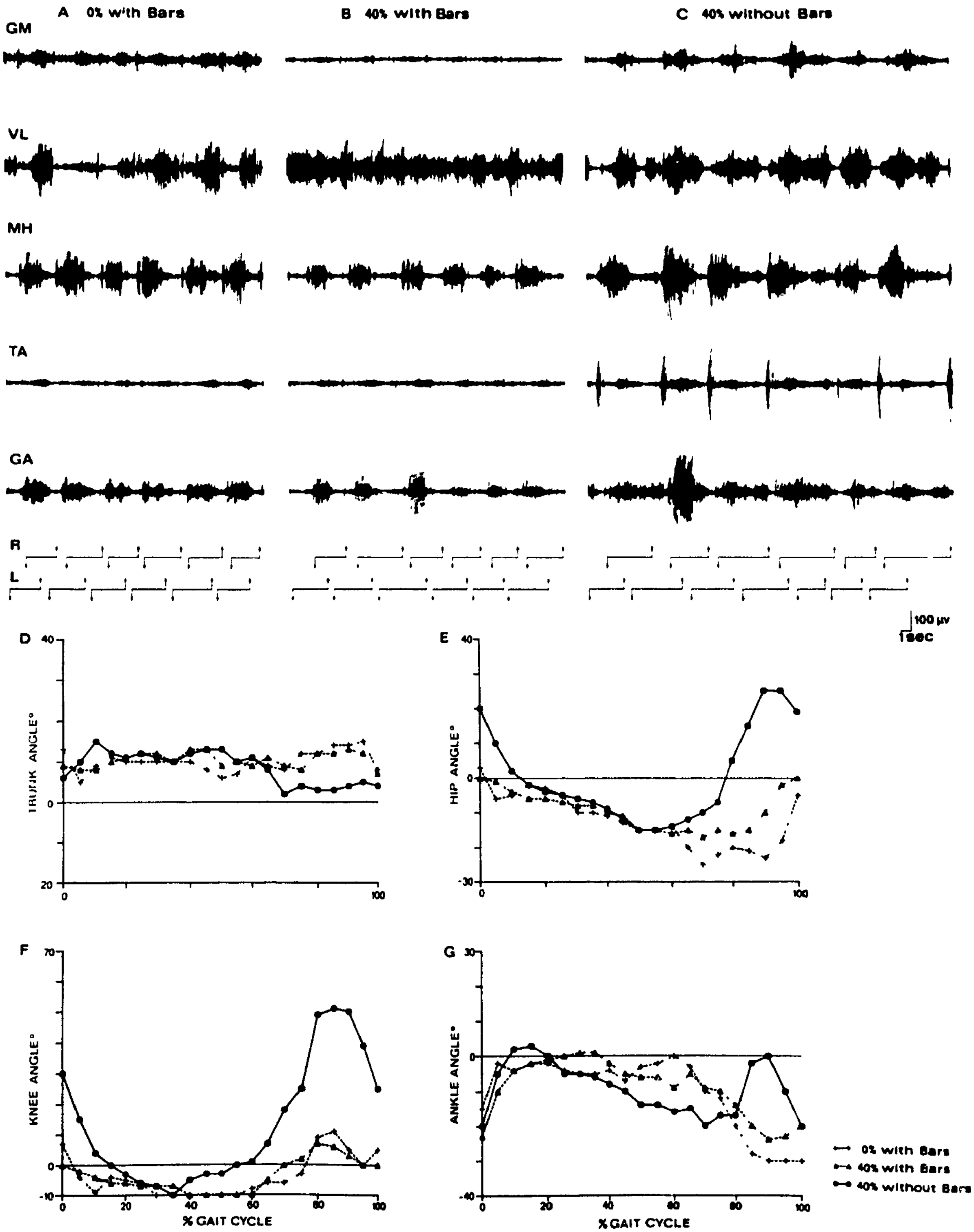
Facilitation of gait with BWS was also noted in 2 other subjects (LL and FS) who were unable to walk without parallel bars at 0% BWS. Both these subjects were able to walk without parallel bars at 40% BWS.

Providing 40% BWS led to a decrease in the amount of clonus and abnormal reflex activity present in the distal muscles when walking without parallel bars as illustrated in Figures 4.2.2B to G for subject LR, at her comfortable treadmill speed of 0.10 ms^{-1}. Although this subject was capable of walking without the parallel bars at 0% BWS, her gait was laboured, accentuated by excessive abnormal reflex activity in the distal muscles. The effect of 40% BWS on the proximal muscles, GM, VL, and MH, was a decrease in EMG amplitude. The most significant changes were evident in GA and SOL, where a marked decrease in clonus was seen. The irradiation of clonus, from GA and SOL seen in TA at 0% BWS, was abolished at 40% BWS. The decrease in clonus in both GA and SOL was accompanied by the appearance of an EMG burst during stance at 40% BWS.

Figures 4.2.2D to G revealed that 40% BWS did not result in significant changes in the sagittal angular displacement profiles at the trunk, hip, knee, and ankle. Walking without parallel bars at 0.10 ms^{-1} was difficult for this subject, however, providing 40% BWS enhanced the locomotor performance resulting in a smoother, less spastic gait pattern.

In these subjects who walked with an asymmetrical, compensatory gait pattern (CF and LR) 40% BWS appeared to have the most positive effects on the gait parameters when parallel bars were removed during treadmill locomotion. Figures 4.2.6A to G illustrate the effects of 40% BWS with and without parallel bars. As

Figure 4.2.6: The right lower limb EMG activity of subject CF walking on the treadmill, at a speed of 0.08 ms^{-1}, at A) 0% BWS with parallel bars, B) 40% BWS with parallel bars, and C) 40% BWS without parallel bars. The downward arrows indicate foot-floor contact, while the upward arrows indicate toe-off, with the solid line depicting stance duration and the space denoting swing duration, for both right (R) and left (L) lower limbs. In C, note the more phasic EMG activity in VL and the appearance of a burst of activity in TA during swing. The corresponding sagittal angular displacement patterns of a representative cycle for the D) trunk, E) hip, F) knee and G) ankle are also illustrated. Note the presence of hip and knee flexion and ankle dorsiflexion at 40% BWS without parallel bars.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subject CF walked with parallel bars at 0% BWS he compenstated with his less involved side, producing an abnormal gait pattern with minimal hip, knee and ankle flexion during the swing phase (Figures 4.2.6A and 4.2.6D to G). When 40% BWS was provided, while he walked with parallel bars, minimal changes were noted (Figures 4.2.6B and 4.2.6D to G) as the subject continued to compensate, and the abnormal gait pattern persisted. However, when 40% BWS was provided without parallel bars, a more normal gait pattern emerged, characterized by an appropriate swing phase with respect to EMG activity and sagittal angular displacement patterns (Figures 4.2.6C to G). Although parallel bars can provide support during gait training, there is a clear advantage to using BWS instead. Supporting a percentage of body weight, while retraining gait, appears to discourage the development of a compensatory, asymmetrical gait pattern and facilitates a more normal gait pattern.

In symmetrically involved subjects, providing 40% BWS during treadmill locomotion without parallel bars resulted in more normal EMG profiles. This is exemplified in Figures 4.2.3B and C for subject BG walking at her comfortable speed of 0.43 ms^{-1} at 0% and 40% BWS. A decrease in EMG amplitude of all muscles as well as later recruitment of GA and SOL was noted with 40% BWS. In Figure 4.2.3C the EMG profiles closely resembled those at 0% BWS with parallel bars, illustrated in Figure 4.2.3A. Minimal changes were noted in the sagittal angular displacement patterns when contrasting 0% and 40% BWS in this subject (Figures 4.2.3D to G). A similar trend was noted in subject BP at his minimal treadmill speed of 0.20 ms^{-1}.

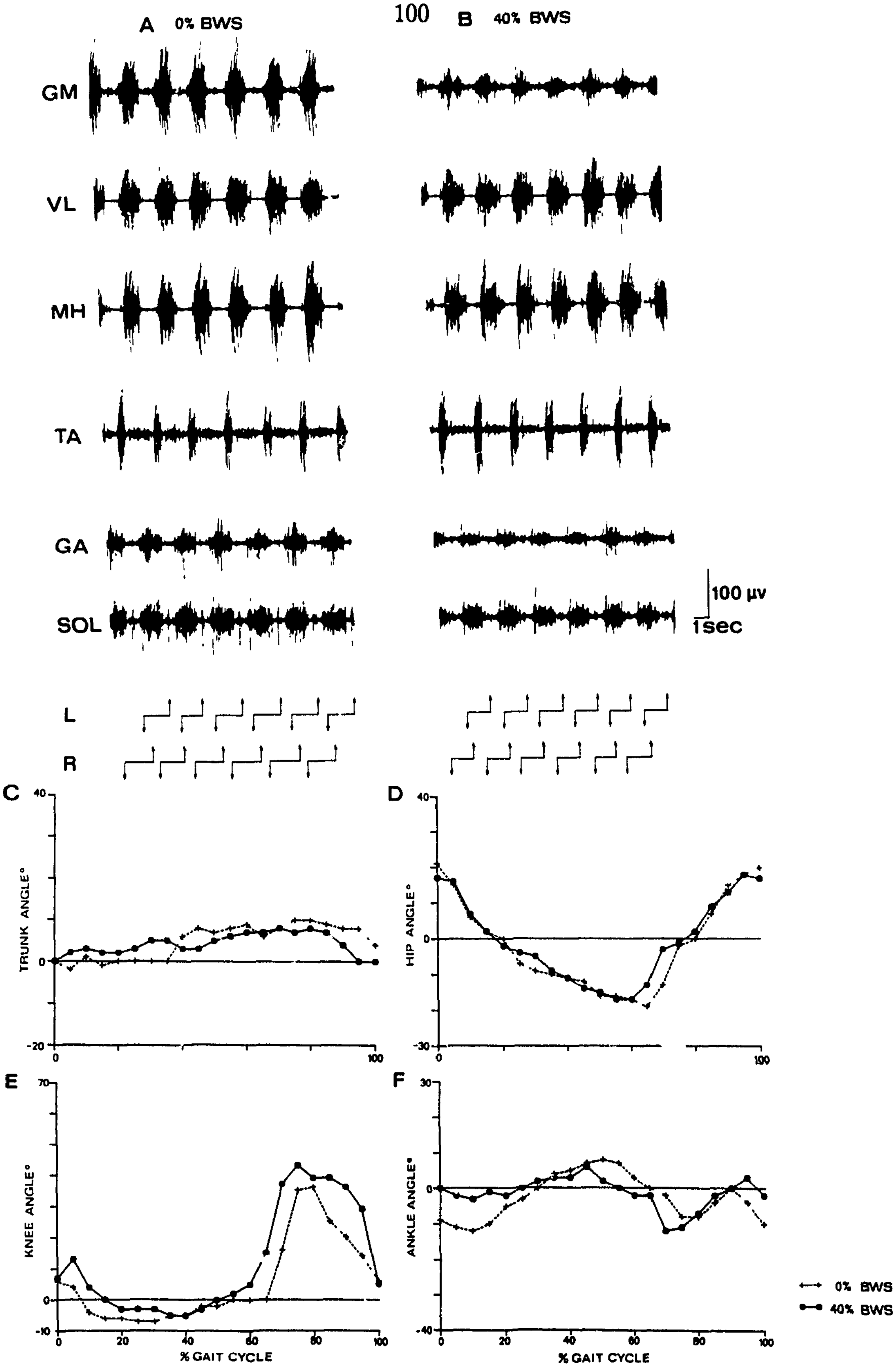
BWS at Higher Speeds:

The effects of BWS at speeds higher than the comfortable treadmill speed was investigated in 4 subjects. In LR and BP increases in walking speed gave rise to an increase in clonus in the distal muscles which subequently showed a decrease in amplitude with 40% BWS at comparable speeds. While subject LL walked at his fastest speed (0.25 ms^{-1}) with 40% BWS, a decrease in the amplitude of all muscles was noted when compared to 0% BWS. In the last subject JS, no effects of 40% BWS at his fastest speed (0.15 ms^{-1}) were noted, as he was presented with general irradiating clonus in all muscles even with minimal effort. In general, all subjects reported it was easier to walk at 40% BWS at the faster speeds, and qualitatively their walking pattern appeared smoother and less spastic.

BWS at Comfortable Treadmill Speed with Parallel Bars:

One of the effects of 40% BWS, while subjects walked at their comfortable speed with parallel bars, was a decrease in EMG amplitude. This was especially evident in subjects who presented with a mild disturbance in the locomotor pattern. An example is illustrated in Figures 4.2.7A and B for subject RC. At 0% BWS, aside from prolonged VL activation and the absence of a TA burst during initial stance, the EMG timing of the other muscles did not deviate markedly from that of a healthy subject (Hirschberg and Nathanson, 1952; Fung and Barbeau, 1987). VL was active just prior to foot-floor contact with activity continuing into latter stance. The effects of providing 40% BWS were a decrease in burst amplitude in GM, GA, and to a lesser degree in SOL. Other subjects who showed similar trends in one or more muscles were CF, BG, FS, LL, and BP. In 4 of these 5 subjects the decrease in

Figure 4.2.7: The left lower limb EMG activity of subject RC walking on the treadmill, at a speed of 0.40 ms^{-1}, with parallel bars, at A) 0% BWS and B) 40% BWS. The downward arrows indicate foot-floor contact, while the upward arrows indicate toe-off, with the solid line depicting stance duration and the space denoting the swing duration, for both left (L) and right (R) lower limbs. In B, note the decrease in EMG amplitude in GM, GA and SOL. The corresponding sagittal angular displacement patterns of a representative cycle for the D) trunk, E) hip, F) knee and G) ankle are also illustrated.



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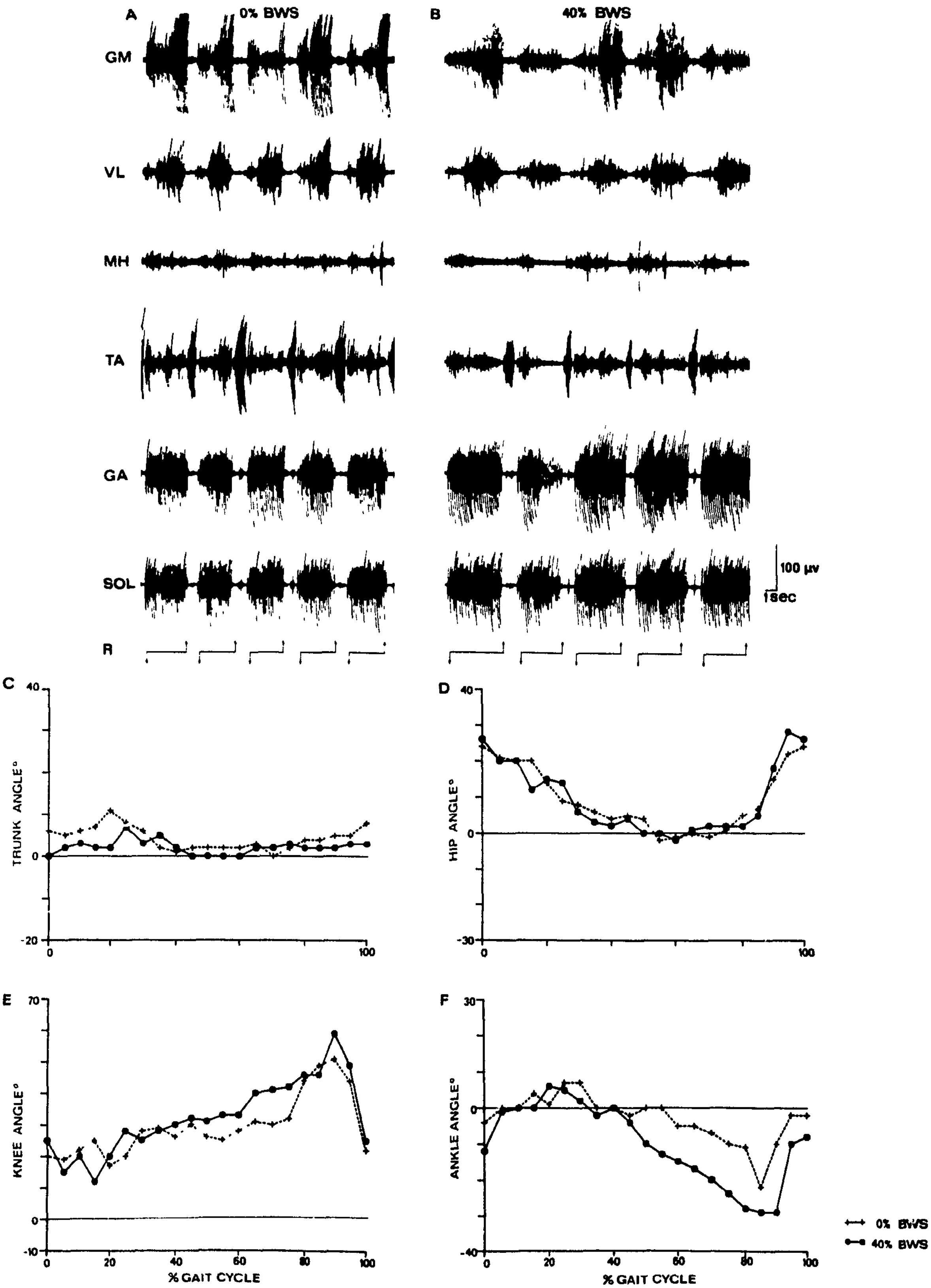
0% BWS
40% BWS

EMG amplitude was seen in the distal muscles.

The effects of 40% BWS on the sagittal angular displacements (Figures 4.2.7C to F) in this subject were minimal. A yield was noted at the knee during initial foot-floor contact with BWS (Figure 4.2.7E) as well as a small increase in maximum swing angle. The ankle was held in a neutral position (0°) at 40% BWS during foot-floor contact as opposed to the plantarflexed position (-9°) seen at 0% BWS.

In a severely spastic subject (JS) who was presented with marked clonus while walking with parallel bars at his comfortable treadmill speed (0.10 ms^{-1}), providing 40% BWS had negligible effects on the lower limb EMG patterns as illustrated in Figures 4.2.8A and B. At 0% BWS this subject's EMG patterns revealed the presence of clonus in all muscles during the entire stance phase. GM showed activity in stance with marked clonus in late stance continuing into swing. VL showed a burst of activity which was present from mid to late stance. Some activity was seen in TA during swing for foot-floor clearance. When 40% BWS was provided (Figure 4.2.8B), no significant changes in the EMG profiles were noted. Sustained clonus during stance continued to be present especially in the distal muscles. Minimal changes were noted in the sagittal angular displacement profiles with 40% BWS (Figures 4.2.8C to F).

Figure 4.2.8: The right lower limb EMG activity of subject JS walking on the treadmill, at a speed of 0.10 ms^{-1}, with parallel bars, at A) 0% BWS and B) 40% BWS. The downward arrows indicate foot-floor contact, while the upward arrows indicate toe-off, with the solid line depicting stance duration and the space denoting swing duration, for the right (R) lower limb. The corresponding sagittal angular displacement patterns of a representative cycle for the D) trunk, E) hip, F) knee and G) ankle are also illustrated.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Discussion:

Parallel bars:

Ambulating without parallel bars has been deemed by some investigators to accentuate gait disturbances in spastic syndromes (Conrad et al, 1983; 1985; Benecke and Conrad, 1986). In this study, removing parallel bars in symmetrically involved spastic paretic subjects resulted in an increase and prolongation of EMG activity during stance. Conrad et al (1983) attributed comparable EMG changes, in a group of paraspastic subjects, to non-specific protective gait mechanisms present during instances of emotional stress or anxiety when the subject's stability is threatened. However, during the present experiment, subjects were securely supported in the harness even though no BWS was provided (under full weight bearing conditions) while they walked without parallel bars. All subjects reported that the harness provided a feeling of safety which illiminated any fear of falling. Regardless of this, a deterioration in the gait pattern characterized by an increase and broadening of the EMG activity was noted in subjects with symmetrical gait involvement. These results suggest that parallel bars provide lateral stability which compensate for the decreased postural reactions observed in spastic syndromes (Benecke and Conrad, 1986) while walking on a moving surface. The changes in EMG patterns especially noted in the distal muscles, closely resemble those observed in an immature gait when postural reactions are not yet fully developped (Sutherland et al, 1980; Forssberg, 1985; Berger, 1986). Providing 40% BWS for such patients resulted in more normal EMG profiles and it can be suggested that BWS increases postural stability. This is supported by a previous study which reports that 40% BWS results in a decrease in percentage total double support time and an increase in single limb support time (Visintin and Barbeau, 1989).

An interesting phenomenon observed was the more symmetrical gait pattern which emerged in asymmetrically involved subjects when they walked without holding on to parallel bars at 0% BWS. Removing parallel bars produced a more normal swing phase by eliciting EMG activity in TA accompanied by ankle dorsiflexion. The hip, knee, and ankle flexion noted in subject CF had not emerged under conditions where the subject was allowed to compensate. The more normal gait pattern only emerged with an increased demand of the locomotor task, necessitating changes in the motor output allowing the subject to advance the limb. In rehabilitating gait and other functional tasks, it becomes important to manipulate the environment and provide external conditions which favor the response being sought. Tasks must be kept difficult yet attainable to enhance functional recovery (Bach-Y-Rita, 1983). In the case of the two subjects in this study (CF, LR), walking without parallel bars was facilitated by providing 40% BWS. Providing the appropriate environmental conditions is critical during early gait training in order to maximize the locomotor potential while preventing compensatory gait deviations.

Speed:

Variations in EMG activity, angular displacement profiles, and temporal distance parameters as a function of walking speed in healthy subjects have been extensively reported in the literature (Kirtley et al, 1985; Yang and Winters, 1985; Frigo et al, 1986; Shiavi et al, 1987a). This is in contrast with a lack of objective information which exists quantifying the effects of increased walking speed on gait parameters in spinal cord injured patients. Dietz (1986) alluded to muscle hypertonia as the main reason for the spastic paretic subjects' inability to walk at faster speeds.

In this study, the effects of increasing walking speed from a minimal to a maximal treadmill speed was the appearance or increase in clonus in 4 subjects. Burke and Lance (1973) have reported that the amplitude of stretch reflexes are directly proportional to the velocity of stretch. Hence, it is likely that speed of walking could influence the amount of clonus during locomotion. Providing BWS could result in a decrease in clonus in 3 of the 4 subjects.

In three other subjects, increasing walking speed resulted in only a small increase in muscle EMG amplitude. This is contrary to the increase in EMG amplitude reported in healthy subjects with increases in walking speed (Yang and Winter, 1985). The low amplitude in EMG activity suggests that these spastic paretic subjects may have been unable to produce the force required to walk at faster speeds. Knutsson (1980) reported a marked decrease in the ability of spastic subjects to develop force during fast movements in a task involving isolated flexion and extension of the knee at a preset speed. However, it is difficult to draw inferences on the spastic paretic subjects' inability to walk at faster speeds from such studies, as the experimental task varies significantly from that of locomotion.

A point of interest in these 3 subjects was that clonus had been elicited in all of them when tested at rest, yet it was not present in the distal muscles even at their maximal walking speed. It could be that the walking speed was too slow to elicit clonus, or that the hyperactive stretch reflexes elicited at rest are modulated during functional tasks such as locomotion and are therefore not necessarily elicited during gait.

Spastic paretic subjects are known to walk at speeds considerably lower than that of normal subjects (Barbeau et al, 1988, Visintin and Barbeau, 1989). In the present group of subjects, the comfortable treadmill speeds ranged from 0.08 ms^{-1} to 0.40 ms^{-1}, and the highest maximal treadmill speed was 0.60 ms^{-1} for subject BP. At such low walking speeds it becomes imperative to distinguish gait characteristics caused by the pathology and those which are a result of slow walking speeds (Andriacchi et al, 1977; Longhurst, 1980). Until recently (Shiavi et al, 1987b), pathological gait profiles were compared to standardized normative data derived from healthy subjects walking at comfortable speeds (Peat et al, 1976; Knutsson and Richards, 1979; Conrad et al; 1985). Shiavi et al (1987a) investigated the EMG profiles of healthy subjects walking at very low speeds (lowest range: 0.34 ms-1) in order to provide a template for comparison with pathological gait while controlling the speed effect. They reported that MH muscle was biphasic at the very low walking speeds, with most subjects displaying a burst of activity at the stance-swing transition which was abolished as the speed increased. This type of MH EMG profile has been observed in this as well as other studies (Conrad et al, 1985; Visintin and Barbeau, 1989) and may be a result of low walking speed and not abnormal motor programming. Shiavi et al (1987a) also reported that gait parameters became more variable at the very low walking speeds and that muscles responded to individual movement requirements. Subjects described their gait as becoming less automatic. The low walking speeds may partially explain the increased variability in gait parameters observed among spastic paretic subjects when compared to normal subjects (Knutsson; 1985). Frigo et al (1986) studied lower limb angular displacements patterns in normal subjects at different walking speeds and reported the frequent absence of a yield at the knee during initial stance at lower speeds. This is a finding frequently observed in spastic paretic gait (Visintin et Barbeau, 1989), and the subjects' low walking speeds may partly be responsible for this.

In general, subjects in this study exhibited a low EMG amplitude for most muscles while walking at their comfortable treadmill speeds. Knutsson (1985) described one of the abnormal activation patterns as a marked decrease in EMG activity during gait in certain muscles, although the subjects were able to generate force in these muscles during other tasks. The decreases in EMG activity noted among spastic paretic subjects may partly be resulting from their low walking speeds as well as central paresis.

BWS:

Providing 40% BWS during treadmill locomotion facilitated gait in asymmetrically involved subjects walking without parallel bars. Subjects were able to take a larger number of steps with greater ease, making gait training without parallel bars more feasible. Retraining gait with BWS, while increasing the demand on the locomotor system by removing parallel bars, is worth considering in the early stages of gait training. The approach appears imperative in facilitating a more symmetrical gait pattern while discouraging gait asymmetries from developing. Although both parallel bars and BWS can be considered as a form of support during gait training, there appears to be definite differences between the two. Parallel bars appear to yield a more compensatory gait pattern, probably by facilitating weight transference to the less involved side in asymmetrically involved subjects. BWS encourages a more symmetrical gait pattern by supporting a percentage of body weight centrally, and discouraging compensation with the less involved extremity. The differences between these two types of supports need to be elucidated in a further study.

BWS had the effect of decreasing clonus in some subjects at the higher walking speeds, but not abolishing it completely. This suggests that speed has a greater influence on stretch-triggered clonus than BWS has. BWS had little effect on decreasing clonus in subjects who showed excessive clonus while walking at their comfortable treadmill speed. Such subjects could be candidates for antispasmodic drug therapy, such that abnormal reflex activity could be reduced prior to gait training, in order that the locomotor potential can be optimized (Barbeau et al, 1982; 1988; Wainberg and Barbeau, 1987).

One of the effects of 40% BWS observed in the present study, is a decrease in EMG amplitude when compared to 0% BWS at comparable speeds. This may explain the higher comfortable walking speeds reported at 40% BWS in a previous study (Visintin and Barbeau, 1989). Since one of the goals of rehabilitation is to increase walking speed, in order to render the gait more efficient, it appears plausible to retrain gait at higher walking speeds with BWS and progressively decrease the weight support during the training regimen. Further studies on BWS and its carry over effects with time are needed, although preliminary reports appear favourable (Fung et al, 1988).

In general, all participants subjectively reported that it was easier to walk at 40% BWS than 0% BWS during most experimental paradigms. Qualitatively, the gait appeared smoother, and less strenuous, suggesting a facilitatory role of BWS during locomotion. Supporting a percentage of body weight may increase the ease with which patients are able to correct their gait abnormalities during different events in the gait cycle further facilitating interactive locomotor training.

Implications for Gait Training:

The complexity of the disturbed locomotor pattern, coupled with postural instability and abnormal reflex activity, following a spinal cord lesion, make it difficult to establish a universal training strategy to reeducate gait. A clear understanding of the gait deficits and their causes is required in individual cases in order to propose a comprehensive gait training approach (Grimm, 1983; Knutsson, 1985; Benecke and Conrad, 1986). Likewise, an understanding of the influence of external parameters such as parallel bars, speed and BWS are needed in order to incorporate them into a training strategy.

Although removal of parallel bars has been thought to cause a deterioration in the gait pattern, it has been demonstrated here that in patients with an asymmetrical gait pattern, removing parallel bars will decrease the opportunity for compensation thus allowing for the expression of a more normal gait pattern. If the locomotor task is a difficult one under such conditions, BWS can be incorporated in the training regimen to facilitate gait. In symmetrically involved patients parallel bars can be removed and postural stability increased with BWS while retraining gait. Such patients appear to be candidates for gait training with BWS achieving the goal of increasing speed, increasing postural stability, while eliciting more normal EMG and angular displacement profiles of the lower extremities.

Increasing walking speed has been shown not to necessarily cause a deterioration in gait in some subjects. Those presenting with minimal abnormal reflex activity during gait showed just a small increase in EMG amplitude although they were not able to walk at speeds much beyond that of their comfortable level (LL: 0.15 ms^{-1} to 0.25 ms^{-1}; BP: 0.40 ms^{-1} to 0.60 ms^{-1}). Such subjects would probably benefit from gait training with BWS at speeds higher than their comfortable walking speed (Visintin and Barbeau, 1989). In subjects where increased speed led to an increase in abnormal reflex activity antispasmodic drug therapy would be indicated prior to gait training.

It becomes clear from the above results that locomotor retraining following a spinal cord injury requires a comprehensive, interactive approach in order to maximize the locomotor potential. An understanding of the underlying abnormal motor programs and reflex activity, coupled with an understanding of the influence of external parameters during locomotion, are essential to acheive this goal.

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CHAPTER FIVE: CONCLUSIONS

The results of this research project demonstrate that BWS can facilitate gait in spastic paretic subjects following an incomplete spinal cord lesion. BWS had a positive effect in subjects with a spastic gait characterized by prolonged activation of proximal muscles defined as 'crutch spasticity', resulting from an inability to cope with load during stance. Providing BWS facilitated weight acceptance during the stance phase and elicited more normal EMG profiles of proximal muscles. Spastic subjects who showed premature activation of distal muscles during stance responded to BWS by showing a decrease in early onset of activity. During more demanding external conditions, such as walking without parallel bars, early recruitment of distal muscles was seen in symmetrically involved subjects. Such subjects also benefitted from BWS as a decrease in premature muscle activation was noted.

BWS facilitated gait in asymmetrically involved subjects while walking without parallel bars. Subjects were able to walk with greater ease and take a greater number of steps. In comparison to parallel bars, which may be considered as an alternative type of support, BWS may elicit a more normal symmetrical gait pattern as it does not allow for compensation with the less involved side to occur. This is an important consideration during early gait training, when a symmetrical gait, with equal weight bearing, is sought.

BWS also led to a decrease in abnormal reflex activity such as clonus in some subjects when walking with parallel bars at a comfortable treadmill speed or during more demanding external conditions such as that without parallel bars or at faster treadmill speeds. In some instances of severe clonus in the distal muscles during stance, BWS appear to have minimal effects. In such cases, disturbance in the internal spinal program may result in a low threshold for reflex activity. Hyperactivity in the reflex pathways would have to be modified by pharmacological intervention in order to determine the effects of external factors such as BWS.

The changes evident in the temporal distance parameters suggest that BWS facilitates weight bearing on the affected limbs. An increase in single limb support time and a decrease in percentage total double support time were evident when subjects walked at 40% BWS. This suggests that the subjects were able to bear weight on their limbs for longer periods of time and they were able to decrease the amount of time they spent supporting themselves on both limbs during the gait cycle. This also suggests that there is an increase in postural stability with BWS. Increases in cycle duration, stride length and speed were also noted, suggesting a less spastic gait with freer limb movements and an increased ability to shift weight from one limb to another.

Changes in sagittal angular displacement profile seen with BWS also suggest an increased ability to cope with weight acceptance during the stance phase. The more remarkable changes were a straighter knee alignment during initial loading and at midstance. A straighter trunk alignment with BWS was also evident.

In summary, it appears that BWS may allow the expression of a more normal gait pattern in spastic paretic subjects presented with prolonged muscle activation of the proximal muscles, early recruitment of the distal muscles and abnormal reflex activity, such as clonus, in the distal muscles. Spastic subjects who have with difficulties bearing weight during loading, or walking with a typical flexed posture, may also respond positively to BWS. The advantage of providing BWS during gait training is that it can enhance locomotor performance during more demanding external conditions such as walking without parallel bars and at faster treadmill speeds. This allows for the manipulation of environmental conditions in order to maximize the locomotor potential; such as removing parallel bars to minimize gait asymmetries and compensation; or increase walking speeds to facilitate gait training.

Limitations of the Study

Given the heterogeneity of the subjects participating in this study and the large variability in gait parameters, pooling of the data was precluded. Each subject was studied as a case study, and grouping of subjects showing similar trends was attempted, followed by a quantitative and descriptive report of the results.

EMG and sagittal angular displacement profiles of a healthy subject, walking at a comfortable treadmill speed, were used as templates for comparison of the spastic paretic subjects' profiles. Since spastic paretic subjects walked at considerably lower speeds, the effects of speed per se on the EMG and sagittal angular displacement profiles may have been overlooked.

Future Directions for related Research Projects

It would be important to determine if similar effects of BWS would be observed during overground locomotion. This would be relevant to the clinical setting, where gait training is usually performed overground. Thus a BWS system within an adjustable and moveable framework may be seen more applicable. In adjunct, differences in treadmill and overground locomotion in spastic paretic patients should be identified. The feasibility of treadmill gait training, with and without BWS, and the carry over effects to overground locomotion should be adressed.

The effects of using BWS for gait training with other neurological conditions, such as cerebral vascular accidents, cerebral palsy, multiple sclerosis, etc. must also be determined, as well as the relative effectiveness of BWS in acute versus chronic lesions.

It would be important to determine the effects of other levels of BWS, such as 10%, 20%, and 30% BWS, on the locomotor pattern of spastic paretic subjects. This would provide some indication of how subjects respond to different levels of BWS providing some understanding on how to progressively decrease BWS during treadmill training.

A randomized clinical trial comparing the effects of conventional gait training to that of BWS, coupled with treadmill stimulation, is required to determine the effectiveness of this new training strategy. It would be important to determine if spastic paretic subjects achieve higher levels of functional ambulation and less gait deviations following this treatment approach when compared to conventional treatment.

Finally, the underlying neurophysiological mechanisms which allow for the expression of a more normal gait pattern with BWS remain to be elucidated. The effects of BWS on modulation of spinal reflexes during locomotion would provide some insight into the possible mechanisms facilitating gait. All these issues are currently being investigated.

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APPENDICES:

A-1 Clinical evaluation on clonus, TSR, overground temporal distance factors

A-2 Anthropometric chart

A-3 Lower extremity muscle chart

A-4 Body sensation chart

A-5 Consent form

CLINICAL EVALUATION

PATIENT NAME: ______________________

DATE: ______________________

COMMENTS: ______________________

ASSESSMENT RESULTS:

A. CLONUS

	R	L
no response (0):		
one to two beats (1):		
more than two beats (2):		
sustained response (3):		

B. TONIC STRETCH REFLEX

	K. FLEX.		K. EXT.	
	R.	L.	R.	L.
normal response (0):				
minimal response (1):				
moderate response (2):				
strong response (3):				
hyperactive response (4):				

C) OVERGROUND LOCOMOTION

velocity (cm/sec): ______________________

cadence (steps/min): ______________________

stride length (cm): ______________________

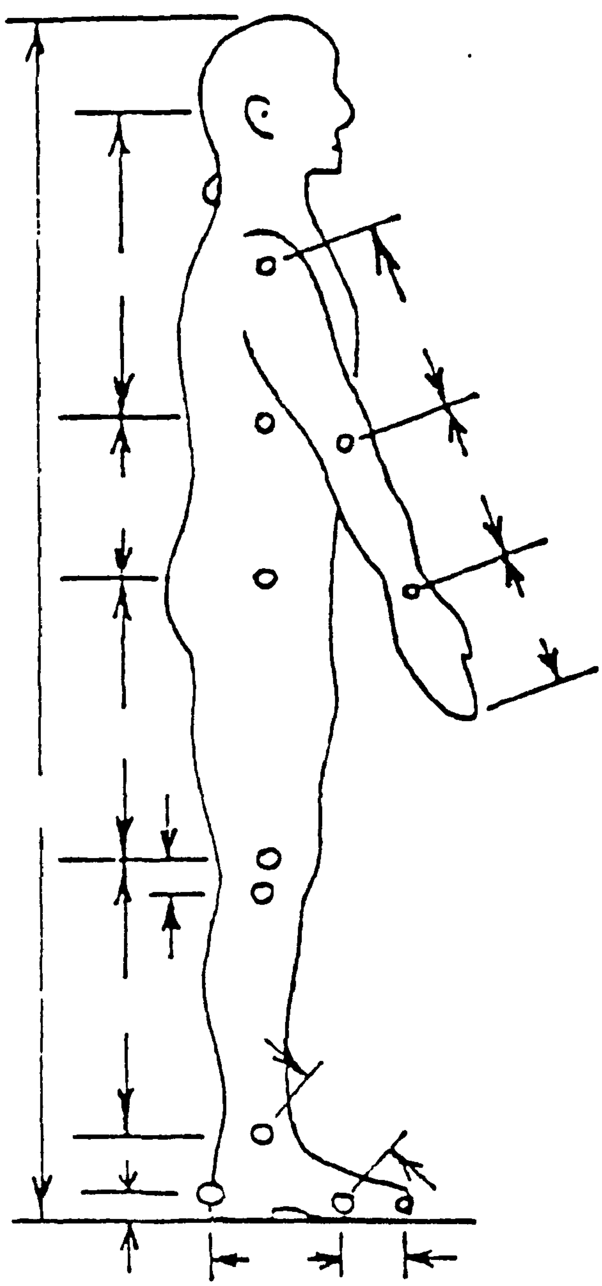
D) LEVEL OF SPASTICITY (VAS): ______________________

ANATOMICAL LOCATION OF MARKERS AND

ANTHROPOMETRIC DATA

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

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



Anatomical Location of Markers

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

Hip - greater trochanter

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

Ankle - lateral malleolus of fibula

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

Meta - 5th metatarsal phalangeal joint

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

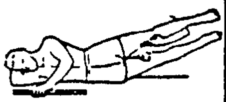
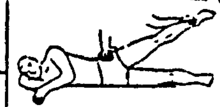
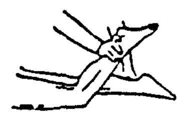
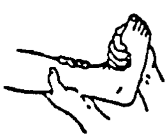
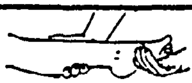
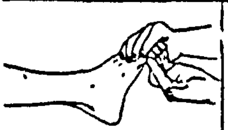
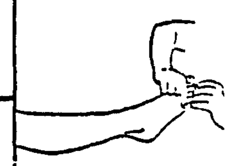
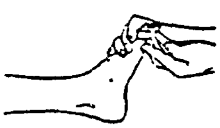
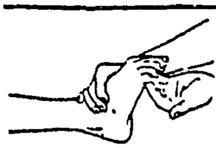
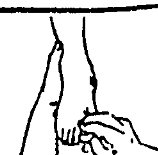
CHART FOR ANALYSIS OF MUSCLE IMBALANCE

LOWER EXTREMITY

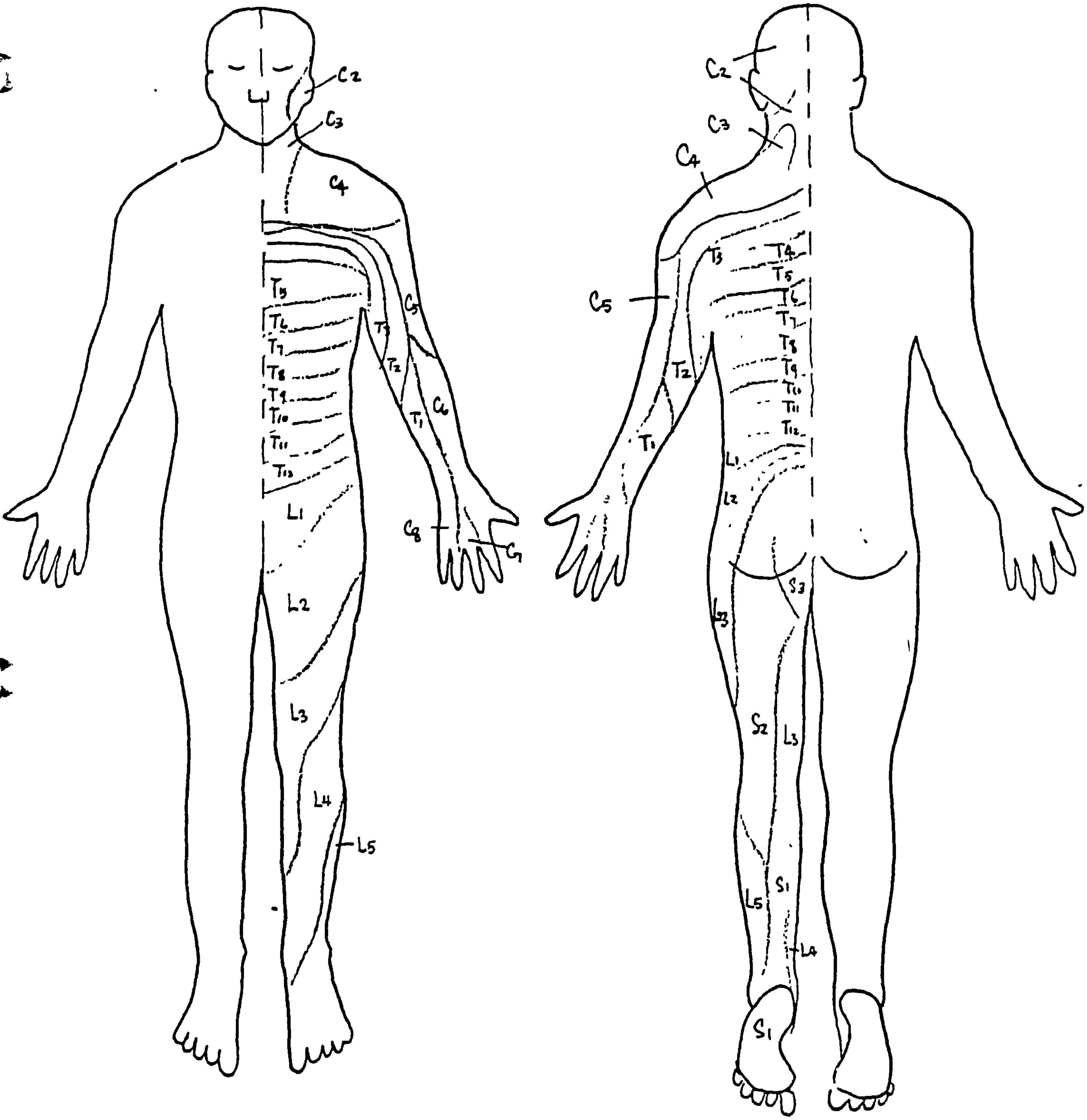
A - 3

Name:...*Date: 1st. Ex.-*....................*2nd. Ex -*......................

ıgnosis:..*Onset:*....................*Exam. of*..............*extremit*

	ILIOPSOAS SARTORIUS TENSOR FAS. LAT. RECTUS FEMORIS } HIP FLEXORS					GLUTEUS MAXIMUS	
	HIP ADDUCTORS					GLUTEUS MEDIUS	
						GLUTEUS MINIMUS	
						TENSOR FASCIAE LATAE	
	HIP LATERAL ROTATORS					HIP MEDIAL ROTATORS	
	QUADRICEPS					MEDIAL HAMSTRINGS LATERAL	
	TIBIALIS ANTERIOR					SOLEUS	
						GASTROCNEMIUS & SOLEUS	
						PERONEUS LONGUS & BREVIS	
	TIBIALIS POSTERIOR					PERONEUS TERTIUS	
	FLEXOR DIGITORUM LONGUS 1 2 3 4					1 2 3 4 DISTAL INTER-PHALANGEAL JOINT EXTENSORS	
	FLEXOR DIGITORUM BREVIS 1 2 3 4					1 2 3 4 PROXIMAL INTER-PHALANGEAL JOINT EXTENSORS	
	LUMBRICALES & INTEROSSEI 1 2 3 4					1 2 3 4 EXT. DIGITORUM LONGUS & BREVIS	
	FLEXOR HALLUCIS LONGUS					EXTENSOR HALLUCIS LONGUS & BREVIS	
	FLEXOR HALLUCIS BREVIS						
	ABDUCTOR HALLUCIS					ADDUCTOR HALLUCIS	

BODY SENSATION CHART



Anaesthesia/Analgesia

Proprioceptive loss

Hyperalgesia

LABORATOIRE DE LOCOMOTION HUMAINE

SCHOOL OF PHYSICAL AND OCCUPATIONAL THERAPY

McGILL UNIVERSITY

FORMULE DE CONSENTEMENT CLINIQUE

La nature et le but du projet de recherche m'ont clairement ete expliques. Ils consistent a evaluer l'effet d'une diminution du poids du corps, par un systeme de suspension avec harnais, sur la fonction locomotrice chez les personnes ayant une lesion de la moelle epiniere. Un index de spasticite durant la marche sera aussi developper pour evaluer les resultats d'autre etudes concernant la spasticite.

Je comprends que je devrais me presenter pour 3 session d'evaluation au laboratoire de locomotion de l'Ecole de Physiotherapie et d'Ergotherapie de l'Universite McGill durant une periode de 2 semaines.

Je comprends que je serai suspendu dans un harnais au-dessus d'un tapis roulant avec 0%, 20%, 40% et 60% du poids du corps supporter par un system mecanique durant differents essaie. Au debut de chaque essaie la vitesse du tapis roulant sera de .28 m/s et elle sera augmenter, si necessaire, pour atteindre une vitesse comfortable. Chaque evaluation consiste de 7 essaie de locomotion sur tapis roulant avec ou sans suspension d'une dure de 2 a 3 minutes chaque. Apres chaque essaie il y aura une periode de repos d'au moins 10 minutes.

Avant et apres chaque periode de marche, ma pression sanguine ainsi que mon poulx cardique seront mesures. Mes mouvements seront enregistres sur systeme video tandis que l'activite electromyographique de mes muscles sera enregistree a chacune des sessions. Des points de reference seront places au genou, a la cheville, et a la hanche de chaqune de mes jambes. L'activite electromyographique de 10 muscles des membres inferieurs sera enregistree par des electrodes de surface.

L'evaluation de la marche sur le sol sera evaluee d'apres les empreintes laissees par les semelles des chaussures, lorsque je marcherais sur un trajet recouvert de papier d'aluminium. Ces empreintes seront analysees plus tard. Des evaluations de force, de reflexes et de sensibilite seront faites pour les membres inferieures. Les sessions d'evaluation dureront au moins 3 heures.

Les couts de transport des participants seront defrayes par l'ecole de physiotherapie et d'ergotherapie. Les resultats de cette etude pourront paraitre dans les revues scientifiques ou serviront pour fin educationnelle. L'anonymat de chaque participant est assuree en tout temps.

Je comprend egalement que je peux me retirer a n'importe quel temps de l'etude sans prejudice.

Je, soussigne(e) ______________________________, comprends les procedures relatif a cette etude. Je consens a participer a cette etude.

Date le ________ jour de ____________ 198_.

Signe ____________________________________

Temoins __________________________________

Je soussigne(e) certifie avoir bien explique au (a la) participant(e) ci-haut nomme(e), la nature de l'etude et le fait qu'il (elle) a le droit de se retirer de l'etude n'importe quand. De plus, les informations seront gardees confidentielles.

Signature ________________________________