Functional electrical stimulation assisted walking in spinal cord injured persons with an incomplete motor function loss: evaluation of the control and capacity

submitted by
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Preface

STRUCTURE OF THE DISSERTATION

This dissertation is meant to be a collection of papers that have a cohesive, unitary character making them a report of a single program of research on the functional recovery of walking by spinal cord injured persons with an incomplete motor function loss. According to the Guidelines for Thesis Preparation from McGill University's Faculty of Graduate Studies and Research the structure for the manuscript-based thesis can include, as part of the thesis, the text of one or more papers submitted, or to be submitted, for publication, or the clearly-duplicated text (not the reprints) of one or more published papers. These texts must conform to the Guidelines for Thesis Preparation and be more than a collection of manuscripts. If coauthored papers are included in a thesis the candidate must have made a substantial contribution to all papers included in the thesis. In addition, the candidate is required to make an explicit statement in the thesis as to who contributed to such work and to what extent. This statement should appear in a single section entitled "Contributions of Authors" as a preface to the thesis.

Following these guidelines, I have included in this dissertation four manuscripts that are either accepted for publication or submitted for review. I have also included a bridging section between the manuscripts concerning the effect of functional electrical stimulation assisted (FES) walking in spinal cord injured participants and the manuscripts concerning the adaptations during obstructed walking in able-bodied and spinal cord injured participants. Furthermore, I have included in the appendices three published manuscripts in which I had a substantial contribution. With the sections preceding and following the manuscripts, this dissertation is certainly more then a mere collection of manuscripts. To orient the readers a more specific scope of each manuscript follows.

The first section of this dissertation is comprised of two manuscripts on the effect of FES assisted walking in spinal cord injured participants. Both manuscripts are accepted for publication in the Scandinavian Journal of Rehabilitation Medicine and are presently in press. The specific scope of the first manuscript is to report the changes in maximal overground walking speed whereas the second manuscript reports the changes in kinematics and physiological cost of walking with training using FES assisted walking. The second section of the dissertation reports on the anticipatory locomotor adjustments during obstructed walking. The first manuscript of this section is under review by the Journal of Neuroscience. The scopes of this manuscript is to report on the linearity and scaling of the changes in toe trajectory as a function of obstacle height as well as the modulation of the causes underlying the toe trajectory adjustments. The second manuscript is under review by Gait and Posture. The scope of the manuscript is a comparison between able-bodied and spinal cord injured persons with an incomplete motor function on the kinematic strategy used during obstructed walking.

Abstracts and acknowledgment sections of all manuscripts are omitted from the text in this dissertation. A reference list is included for each section or manuscript. The reference scheme is standardised for the whole dissertation with roman numerals in between parentheses in the text and references in order of presentation for each section included in the reference list of each section.

CONTRIBUTIONS OF AUTHORS

All the manuscripts included in the main body of the dissertation are written by the candidate, in collaboration with Dr. Hugues Barbeau as well as with Dr. Bradford J. McFadyen for the two manuscripts included in the obstructed walking section. The order in the authorship of each paper represents the contribution of each to the manuscript with myself being the first author on all manuscripts. Dr. Hugues Barbeau is second author on the FES assisted walking manuscripts as well as for the second manuscript of the obstructed walking section. Dr. Bradford J. McFadyen is, respectively, second and third author on the manuscripts of the obstructed walking section. All the manuscripts are based on data collected and analysed by myself.

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results we gathered within the canadian multi-centre trial of FES assisted walking.
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During my doctoral studies there was two fellow PhD students that helped, supported and challenged me to help me grow as a scientist. Words always fail me when I want to express the gratitude I have for Dr. Kathleen Norman and Dr. André Pépin. The times we shared are filled with joy and sadness, relaxation and anxiety, as well as many sleepless nights before conferences. I may not have said it enough but your help during the experiments as well as during the training sessions were very appreciated. I will never forget the countless discussions, debates and lunches we spent discussing and criticising our ideas on the recovery of walking. Kathleen, your insights on rehabilitation helped me tremendously. For this and for all of your answers I would like to say a very special thank

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Abstract

With new developments in traumatology medicine, the proportion of spinal cord injured persons with an incomplete motor function loss (SCI-IMFL) is likely to continue to rise, making it increasingly important to develop treatment strategies and understand the factors that can enhance recovery of walking following spinal cord injury. The effect of long term use of functional electrical stimulation (FES) assisted walking by SCI-IMFL was evaluated. It was shown that FES assisted walking alone modified the locomotor angular pattern but this modification remained constant with training time. Furthermore, FES assisted walking alone had a minimal effect on the speed, spatio-temporal parameters, and physiological cost of walking but all parameters improved with time of training. To better understand the factors involved in the therapeutic effect of FES assisted walking, it is hypothesised that the evaluation of the locomotor behaviour should include new protocol that will necessitate an adaptation to the locomotor pattern like obstructed walking. In our studies of obstructed walking, we have shown that able-bodied participants clear unilateral obstacles of low to moderate height by scaling the increase of their toe trajectory to match the height of the obstacle. This increase in toe height was caused by the generation and modulation of a new knee flexor power generation burst at the initiation of the swing phase that was linearly related to the obstacle height. Even though SCI-IMFL adapt their walking pattern to clear an obstacle, it was shown that this adaptation is different from the one seen in able-bodied participants. SCI-IMFL have a greater increase in their knee flexion in comparison with able-bodied participants and seem to raise their greater trochanter height more than able-bodied participants for obstacles of low heights. One striking result is that none of the SCI-IMFL increased their hip flexion when clearing the obstacle.

Abrégé

La population de blessé médullaire ayant une incapacité motrice partielle (BMI) continuera probablement d'augmenter grâce à l'amélioration des connaissances en médecine de la traumatologie. Ce fait suggère un accroissement de la nécessité de développer de nouvelles stratégies de traitement pour la récupération de la marche chez les BMI. De plus, ces nouvelles stratégies de traitement doivent être basées sur une compréhension accrue des facteurs de récupération de la marche. La première partie de la thèse rapporte les changements concomitants à l'utilisation de la marche assistée par stimulation électrique fonctionnelle (SEF). Les résultats de cette étude démontrent que la marche assistée par SEF, seule, modifie le patron angulaire de la marche mais que ces changements restent sensiblement stables tout au long de la période d'entraînement. De plus, la marche assistée par SEF a un effet minimal sur la vitesse, les paramètres spatio-temporels et le coût énergétique de la marche. Cependant, tous ces paramètres s'améliorent avec la période d'entraînement. Ces études démontrent la nécessité de développer de nouveau protocole d'évaluation de la marche pour mieux comprendre les facteurs impliqués dans cet effet thérapeutique. Pour ce faire, nous avons étudié la marche obstruée. Nos études sur la marche obstruée démontrent que chez les participants normaux l'augmentation de la hauteur des orteils est identique à la hauteur de l'obstacle. De plus, cette augmentation de la hauteur des orteils est le produit de la génération et la modulation d'une nouvelle bouffée de génération de puissance par les muscles fléchisseurs du genou au moment de l'initiation de la phase d'oscillation. L'amplitude de cette nouvelle bouffée est reliée de façon linéaire à la hauteur de l'obstacle. Il est aussi démontré que les BMI adaptent leur patron de marche lorsqu'ils passent un obstacle. Cependant, cette adaptation est différente de celle des participants normaux. Lorsque les BMI passent un obstacle de basse hauteur, ils ont une augmentation de leur flexion du genou lors du passage de l'obstacle ainsi qu'une tendance à augmenter l'élévation de leur grand trochanter en comparaison avec les participants

normaux. Il est remarquable qu'aucun BMI augmente leur flexion de la hanche lorsqu'ils passent un obstacle.

Introduction

Spinal cord injury (SCI), by nature, has a very sudden impact on an individual, both in a physical sense as well as in an emotional and social sense (1). In the United States, 200 000 persons have a SCI (2). As reported, the causes of injury and demographics of those patients seem to be constant for the last 20 years (3). The incidence of SCI varies around the world, but where comprehensive data are available, the incidence is usually reported to be between 20 and 50 cases per million per year, approximately half of whom are under 30 years of age (4-11). In the United States, the annual incidence of SCI is around 10 000 (2). In Canada, the incidence of SCI is 35 cases per million per year and 900 new cases of SCI are reported per year to the Canadian Paraplegic Association (12). Of all new cases of SCI, a large proportion have some preservation or recovery of sensory and/or motor function caudal to the level of lesion. Such an injury is termed an incomplete lesion. In many areas of the world, incomplete injuries are now more common than clinically complete injuries (6, 7, 9, 13). Because of new developments in traumatology medicine, the proportion of spinal cord injured persons with an incomplete motor function loss (SCI-IMFL) is likely to continue to rise. There are several other potential new treatments for acute SCI (14, 15) that may continue to reduce the severity of neurological impairment and consequently increase the potential for walking ability.

In view of the above findings and trends in the epidemiology of SCI, it is becoming increasingly important to develop treatment strategies that can enhance recovery motor function, walking in particular, following SCI. In conjunction with providing treatment to enhance recovery of walking, rehabilitation specialists must consider the factors that may influence such recovery.

RATIONALE FOR THE WORK IN THE DISSERTATION

Functional electrical stimulation (FES) assisted walking has been used for the restoration and improvement of walking for SCI-IMFL for more then 15 years (16). Since then research has been focusing mainly in assessing the role and effect of such an orthosis (17-

20). However, even though the therapeutic effect of such a device on paretic muscle force has been shown in hemiplegics persons (21), and to a lesser extent in SCI-IMFL (22), only a preliminary study had anecdotical reports of the therapeutic effect of FES assisted walking in SCI-IMFL (18). Hence, it was necessary to undertake such a study to investigate the presence and extent of this therapeutic effect on the recovery of walking. To fully understand the functional recovery in the walking behaviour occurring with FES assisted walking, or any other intervention, it becomes obvious that it is necessary to evaluate the walking behaviour not only on a treadmill or a flat overground surface but also with new experimental tasks. The selection and necessity for a different task to evaluate the control of walking emerges from the literature on the theory of action (23). The task of obstructed walking is a new evaluative task that requires a modification of the ongoing locomotor behaviour. As shown in cat studies, obstructed walking requires many interactions between the supraspinal and spinal as well as afferent and motor systems (24). Hence using obstructed walking in the evaluation of the walking behaviour would complement an evaluation that would be using only overground walking, on a flat surface and at a comfortable speed. Furthermore, it would provide a greater understanding of the impairment following a SCI with an incomplete motor function loss by posing a realistic locomotor challenge to the participant.

OBJECTIVES OF THE DISSERTATION

The main objectives of the present work are twofold. The primary objective is to investigate changes in the walking behaviour that are concomitant with the use of functional electrical stimulation assisted walking. The second objective is to investigate the adaptations of the walking patterns of SCI-IMFL during obstructed walking.

In this dissertation the outcome measures for the first objective, namely the investigation of the changes in the walking behaviour with FES assisted walking, are three fold. The outcome measures used are the maximal overground walking speed, kinematics of walking at a comfortable speed over a flat overground surface and physiological cost of walking. These measures, although they do not completely cover all behavioural aspects of walking, represent important aspects in the functional recovery of walking.

To gain information with the goal of improving the evaluation of the functional recovery of the walking behaviour we investigated obstructed walking as a new evaluative task. The sub-objectives of the first part of this section were to investigate the linearity and scaling of the changes in toe trajectory as a function of obstacle height as well as the modulation of the causes underlying the toe trajectory adjustments in able-bodied participants. These objectives allowed us to understand how able-bodied participants adapt during obstructed walking as well as providing us with normative data. The sub-objective of the second part of this section is to investigate and compare the kinematic strategies of SCI-IMFL during obstructed walking.

SCHEME OF THE DISSERTATION

The dissertation has three sections: a review of literature, a section on the effect of FES assisted walking in SCI-IMFL, and a section on the adaptation of the locomotor pattern during obstructed walking of able-bodied participants and SCI-IMFL.

The first section reviews the literature pertaining first to the changes in the locomotor behaviour that are concomittant with a SCI. Secondly, a review of the literature on FES assisted walking and its effect is presented. The last section of the review of literature presents the evidence gathered on the adaptations of the locomotor pattern during obstructed walking.

The following section describes in two manuscripts a series of experiments on the longitudinal effect of FES assisted walking training in SCI-IMFL. Participants in the longitudinal study of FES assisted walking have a marked improvement in their walking behaviour, an indication that FES assisted walking is a powerful rehabilitative tool. Another particularly important result from the study is that the greater part of the effect of FES assisted walking comes from a therapeutic effect. These results have a strong clinical importance by showing a different purpose for the use of FES assisted walking as well as by showing that even in chronic SCI-IMFL there is a potential for substantial improvement of their walking behaviour.

In the third section we present in one bridging text and two manuscripts a series of experiments on obstructed walking in able-bodied participants and SCI-IMFL. The bridging section presents a model of the evaluation of the functional recovery of walking. The two manuscripts report evidence that able-bodied participants have a linear increase of their anticipatory locomotor adjustments that was scaled to the obstacle of low to moderate heights. This adaptation to obstacle height was done by generating and linearly scaling a new knee flexor power generation burst occurring at foot-off. In the second manuscript of that section, evidence is presented that SCI-IMFL adapt their walking behaviour differently then able-bodied participants even though they also use an increased knee flexion to go over obstacles. These results can be used to further develop an evaluative walking task that could be used in clinical trials related to different interventions.

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Review of literature

The literature pertinent to this thesis is reviewed in three sections: changes in the walking behaviour following SCI, the utilisation of FES assisted walking in the rehabilitation of walking for SCI-IMFL, the need for new protocols in the evaluation of the control of walking, and the last section reviews the literature on obstructed walking.

CHANGES IN WALKING BEHAVIOUR FOLLOWING SCI

Spinal cord injury, and the inactivity that follows, triggers and is associated with, numerous modifications of the musculo-skeletal, circulatory systems and central nervous system that will influence the walking behaviour. Examples of the changes in the musculo-skeletal system are the increased stiffness of the passive components of the ankle joint (1,2), the increased fatigability and modification of the biochemical properties of motor units (3-6), and the higher incidence of osteoporosis (7-12). Furthermore, depending on the level of the injury there is a modification of the adaptations of the circulatory system during exercise (13-20).

Spinal cord injury is also associated with changes within the central nervous system that include problems of muscle activation such as weakness (21-31), hyperactive spinal reflexes (32-36), and loss of sensory function (37, 38). It is believed that all of these modifications lead to postural problems related to bearing weight, maintaining balance and developing propulsion.

Biomechanical correlates of walking in able-bodied participants

Movement emerges from the interactions of the organism with the environment and its organisation will not only be adapted to the demands of the tasks, but also to the biomechanical and morphological restrictions of the moving system (39). The nervous, musculo-skeletal and energy supplying (ergonic) systems will interact to produce a muscular moment around a joint (contractile activity of the muscles). But other sources of input to the moments at the joints depend on gravity and interactions between the connected segments. Interactions between the segments can be calculated in two ways:

the calculations of the torques generated by the different sources (40), or the calculation of the energy absorption, generation, and transfer of the different segments included in the multiarticular system (41). Hence, the moments or energy needed for behaviours such as walking does not need to be totally generated by the contractile activity, but emerges from the summation of all the interactions occurring between the environment and the person making the relationship between the kinematics and nervous impulses an unequivocal one (42).

The gait cycle can be grossly divided in two majors phases: the stance and swing phase (see Fig. 1). However, such a description fails to really discriminate the phases functionally. In a more functional description of the walking cycle the stance phase is divided into the initial contact, the loading response, midstance, terminal stance, and pre-swing, whereas the swing phase can be divided in the initial swing, midswing and the terminal swing (43).

The initial contact sub-phase can be used to represent the onset of the walking cycle. The functional objective of this sub-phase is to position the foot correctly as it comes in contact with the floor. The loading response occurs between foot-contact and the contralateral foot-off which, in normal participants, represent from 0 to 10% of the total cycle time. This is the first period of double-limb support and the purpose of this phase is the absorption of the energy generated by the impact of the lower limb on the ground. During this phase there is an absorption of energy by the ankle dorsiflexors and knee extensors (41). Midstance occurs from the contralateral leg foot-off to ipsilateral heel rise (10-30% of the gait cycle). The purpose of this sub-phase is to maintain the vertical trajectory of the centre of gravity. It is in this sub-phase that the centre of mass reaches its zenith and that forward velocity is at a minimum. In this phase there is an absorption of energy both by the knee and hip extensors (41). The terminal stance sub-phase occurs from heel rise to foot contact on the opposite side (30-50% of the gait cycle). In this phase, the centre of mass of the body moves in front of the base of support and accelerates by falling forward toward the unsupported side. This phase is characterised by a generation of energy from the ankle plantarflexors and knee extensors (41). The pre-swing phase (50-60% of the gait cycle) is the second period of double-limb support and corresponds to the loading response on the contralateral side. The goal of this sub-phase is to prepare the limb for the swing phase. In this phase, there is an absorption of energy from the knee extensors and a generation of energy from the hip flexors to initiate the swing phase (41).

The initial swing occurs form 60 to 70% of the gait cycle and the purposes of this sub-phase are to maintain a suitable distance between the foot and the ground as well as generate the energy necessary for the completion of the swing. This is done by a generation of energy from the hip flexors and ankle dorsiflexors (41). The midswing sub-phase occurs from maximum ipsilateral knee flexion to contralateral heel rise (70-85% of the gait cycle). Its purpose is in the control of the toe trajectory to maintain a proper distance with the floor. The terminal swing sub-phase happens from the contralateral heel rise to foot contact (85 to 100% of the gait cycle). The purpose of this sub-phase is to decelerate the leg and position the foot properly for floor contact mainly by an absorption of energy by the knee flexors (41). Furthermore, it has been shown that in the swing phase the general muscle torques are counteracting the motion dependent torques (44, 45). It should be noted that those motion dependent torques are related to the velocity of the segments (45).

Neural correlates of walking in able-bodied participants

The generalized muscle torque is controlled by the nervous system through the activation of the muscles by motoneurons. The motoneuron activation is the summation of all the inhibitory and excitatory influences into a final common motor pathway (46). As such during locomotion the motoneurons are influenced by sensory afferences, spinal and supraspinal systems. It has been well established for invertebrates and vertebrates, even in mammals, that the spinal cord can generate a basic locomotor-like pattern (47). However, such a capacity has not been proven for primates and humans. Historically, unsuccessful attempts to initiate locomotion have been reported for primates with an acute or chronic spinalisation (48) and chronic paraplegics with a complete motor function loss (49). Because of the methodological setup of those studies it was not possible to reject the hypothesis that the primate spinal cord centres involved in the generation of the locomotor-like pattern would need a higher level of tonic facilitation than cats. However, a study of fictive locomotion in marmoset monkeys, using higher concentrations of clonidine then a previous study (48), show the possibility of the spinal cord of some primates to generate a locomotor-like pattern (50). In humans, the possibility that the spinal cord could generate a locomotor-like pattern is debatable. Evidence suggesting that such a network is present are accumulating. First, it has been shown that the late flexion reflex occurring after stimulation of the flexor reflex afferents in spinalised cats and attributed to the interneurons comprised in the central pattern generator (CPG) for walking have a human analogue in paraplegics with a complete lesion (51). Secondly, an epidural electrical stimulation of the spinal cord has been shown to produce locomotor-like electromyography (52-54). Thirdly, locomotor-like electromyography activity has been seen in patients that are presumed to have a complete spinal cord lesion (55). New studies of locomotion training using body weight support and manual assistance of spinal cord injured participants with a complete motor function loss have also shown the possibility of the spinal cord in conjunction with sensory afferences to generate a locomotor-like movement (56). But it should be noted that none of the patterns elicited so far could be used functionally.

The role of afferences from skin, muscle and joint receptors in enhancing and resetting the locomotor activity of cats by acting directly or through the CPG for locomotion has been studied extensively (57-61). In humans, several studies have attempted to evaluate the contribution of muscle afferent feedback to the locomotor electromyographic activity during walking by measuring stretch reflex activity in different phases of the walking cycle (62-64) or by unloading the ankle extensors (65). From those studies it can be concluded that afferent feedback contributes importantly to the locomotor electromyographic activity of the ankle extensors.

As seen above, there is evidence that the generation of a locomotor-like pattern can be generated by a spinalised cat preparation, and to a lesser extent by humans. Even though a spinalised cat preparation can generate a locomotor-like pattern, such a preparation still has some locomotor deficits thereby showing the need for some supraspinal inputs structures to locomote (66). The output of the CPG can be modulated at the level of the interneurons included in the CPG or by changing the weighted sum on the motoneurons from the CPG output and descending command. The role of supraspinal structures during locomotion of cats range from the initiation of walking to adaptations to environmental constraints (66). The localisation of the brainstem nuclei that initiate locomotion are specific to the locomotor task but are all situated in the upper brainstem within the mesencephalic locomotor region (67). In conjunction with the monoaminergic pathways it would play a role in the enabling and triggering of the locomotion (68). Other structures within the brainstem, pons and cerebellar structures plays a major role in the modulation of the locomotor behaviour. Stimulation of the tegmental fields of the pons show the possibility that these structures would modulate the postural tone (69). Whereas the reticular formation would play an important role in the postural adjustments necessary during locomotion as deducted from their evoked responses to stimulation during walking (70-74), the modulation of their firing during locomotion (70, 75, 76) and their anatomical branching (77). Based on the evoked response to stimulation of the Deiter's nucleus and the action of the cerebellum, it can be deducted that the vestibular system and the vestibulospinal pathways play an important role in the regulation of the level of activity in the extensors of the ipsilateral limb and the adaptations for the maintenance of equilibrium (66, 71, 78). The cerebellum can also influence locomotion by its action on the red nucleus and rubrospinal tract (66). However, the rubrospinal tract may not exist in humans (79). Finally, the role of the motor cortex in the regulation of the trajectory of the distal segment has been established by the evoked responses to motor cortex stimulation (80, 81), the locomotor deficits resulting from the ablation of the motor cortex (82-84) and modification of their firing pattern during spatially constraint tasks (85-87).

Different models of the organization of locomotor control have been put forward. Historically, Grillner and his collaborators (47, 88, 89), based on the studies of Orlovsky (70, 78, 90), modeled the influence of the supraspinal structures on the flexors and extensors motoneurons. However, new experimental evidence suggests that supraspinal inputs converge onto a core locomotor pattern generator in the spinal cord (for a review see 91). Such a convergence is plausible since many studies have found that the CPG could be influenced by descending inputs (71, 92, 93). The organisation of the supraspinal inputs during discrete movement has been investigated in primates and humans (94-95). However, the role of supraspinal inputs in the locomotor behaviour of humans has been scarcely investigated. The few reports available show a very strong link between the tibialis anterior muscle and the motor cortex during walking (96-98).

Walking in SCI-IMFL

Biomechanical descriptions of walking by SCI-IMFL are scarce, but caution should be exercised when comparing the results because of the heterogeneity of the samples included in those studies. It has been reported that the pattern of SCI-IMFL walking on a treadmill, in comparison to able-bodied participants at matched treadmill speed, can be characterised by a shorter cycle duration and stride length in association (99, 100). Furthermore, it was also reported that the total knee excursion was reduced, foot contact occurred with a flexed knee, the ankle was more dorsiflexed and that the ankle did not plantarflex during the stance phase (99, 100). However, in a case study of a SCI-IMFL walking overground it was shown that cycle duration was increased and that foot contact was with an increased knee flexion and with a plantarflexed ankle (101). In comparison to the changes in the kinetic or kinematic patterns, the changes in the electromyographic patterns have been a little bit more thoroughly investigated (99-104). Briefly, changes in muscle activation are seen in an increased tibialis anterior muscle activation during swing (102). The triceps surae amplitude was slightly reduced (102) with a broadened and flattened electromyographic profile (99, 101, 103) and an activation early in the stance phase (99, 102-105). Muscles acting on the knee also show some changes in there electromyographic pattern with a prolonged muscle activation (103, 104) that seems to be related to the activity of muscles acting on the ankle joint (106).

It can be seen that SCI-IMFL do have problems in most of the sub-phases of the gait cycle. Weaknesses of the knee extensors would certainly modify the gait cycle during the loading and midstance sub-phases (99, 103). Weakness of the ankle plantarflexors would correspond to changes in the terminal stance and pre-swing sub-phases (107). The initial and midswing subphases would be affected by changes in the ankle plantarflexors passive and reflex stiffness as well as weakness of the ankle dorsiflexors (102, 103). Finally, the terminal swing phase would be affected by a decrease in angular velocity generated by the weakness of the hip flexors since intersegmental dynamics are dependent on the velocity of the movement (45). Furthermore, another factor is the energy requirement necessary for ambulation. Although wheelchair propulsion approximates the energy requirement of normal walking (108), the energy cost of walking by SCI-IMFL is higher than for speed-matched walking by able-bodied participants (109).

FES ASSISTED WALKING FOR SCI-IMFL

Multiple modalities have been tried to improve walking in SCI-IMFL. Those modalities range from changes in the internal constraints like pharmacological (for a review see 110) or surgical modalities (111) to the use of modalities modifying the constraints like ambulatory assistive devices (112) or mechanical orthoses (for a review see 113, 114). Another kind of modality has been the use of active orthoses using functional electrical stimulation (FES) assisted walking. The following section will review the literature on the use of FES assisted walking for SCI-IMFL.

There is a long history for the therapeutical use of electrical stimulation. Electrical activity from torpedo fishes was used by Greek and Roman physicians before the time of Jesus-Christ (115). Furthermore, the feasibility of contracting paralysed muscles with electrical stimulation was already shown in the eighteenth century (116). More than 30 years ago functional electrical stimulation (FES) was developed as an orthotic system to prevent "foot drop" in hemiplegic subjects or to be used for SCI patients (117,118). Since then FES has been used to restore a variety of movements including walking (for a review see 119- 121) and was first reported for SCI-IMFL subjects in 1989 (122). The goal of FES is to obtain an immediate contraction of the skeletal muscles that will lead to a functional movement. As reviewed in the previous section, the gait of SCI-IMFL is impaired to some extent by weaknesses in the knee and hip extensors as well as ankle dorsiflexors. In relation to wide spectrum of motor impairment following SCI there are a multitude of systems for FES assisted walking. The simplest system uses a stimulation of the common peroneal nerve through surface electrodes in conjunction with a footswitch (117). However, systems using either four or six channels (120) could also be used. It is interesting to note that new peroneal nerve stimulator systems, using implanted stimulators and natural sensing, are still being developed even 30 years after the initial use of a common peroneal nerve surface stimulator (123-125).

Even though FES assisted walking has been available for more then three decades, it has not been widely used in rehabilitation for a number of technical reasons. The stimulators were bulky, unreliable, prone to breakage and expensive (109). Recent studies of simple system for FES-assisted walking (common peroneal nerve stimulator) showed an increase of the walking speed (less then 0.1 $m*s^{-1}$) and a reduction of oxygen consumption during gait (109, 126, 127).

The combination of simple FES system with locomotor training show the usefulness of such devices to improve walking. Remarkably, walking speed of SCI-IMFL using FES assisted walking is sometimes increased even when the stimulator is temporarily turned off. This retained increase of walking speed without FES, called the therapeutic effect, has been reported anecdotally for hemiplegic patients (117, 128) and in SCI-IMFL (129). Changes in their assistive ambulatory devices as well as changes in the energy efficiency of the walking behaviour were also shown. The relative contributions of locomotor training and FES-assisted walking in the improvement of walking speed still need to be determined.

THE NEED FOR NEW PROTOCOLS IN THE EVALUATION OF THE CONTROL OF WALKING

Many models of motor control are found in the literature. From those many models three different classes can be established: the reflex model (46), the hierarchical model (Jackson, see 130) and the systems model (42). Each model has underlying assumptions regarding the goal of the nervous system in the control of movement. Each class of model, with their underlying assumptions, produces different procedures of motor rehabilitation and evaluative tasks. The influence of the reflex and hierarchical models in the evaluative tasks is seen in the testing of reflexes of newborns and individuals with motor control problems (131-133) as well as in the scoring of the Fugl-Meyer scale (134). However, such evaluations are not suited for the comprehension of behavioural tasks (135). The theory of action (136-144) includes the assumptions of the systems model and provides a precise theoretical and operational framework for the analysis of functional tasks. Based on this operational framework a distinction between the terms coordination, control and skill has been suggested (145): 1) coordination is the function that constrain the potentially free variables into a behavioural unit, 2) control is the parameterisation of this function, 3) skill is the optimal parameterisation of this function.

The assessment of coordination with these operational definitions can be done with the use of angle-angle plots (146-148). However, the numerous degrees of freedom involved during walking cannot be expressed intelligibly by such a technique. A multivariate statistical methods like principal component analysis (149) with a distortion analysis can be used to analyse walking (150). Such an analysis can reveal the coordination of multisegmented synergies (coordinative function) involved in the accomplishment of the task.

Control of walking behaviour as defined by the parameterisation of the walking coordination function cannot be evaluated by simply asking the person to walk on a flat surface at comfortable speed. As such, it requires the use of tasks that will necessitate an adaptation in the walking coordination function that will allow the quantification of the parameters included in the walking coordination function.

Assessing the skill of a coordination function necessitates the comparison of the coordination function to another coordination function that has been optimised in relation to an arbitrary criterion. For example, the assessment of the skill of walking of a SCI-IMFL can be made by comparing the walking coordination function produced by this individual with a walking coordination function optimised for minimum energy expenditure.

The assessment of behaviours based on these operational definitions have been mostly studied with the patterns of hand trajectories. One of many models that can explain the patterns of hand trajectory, and the underlying coordination, control and optimisation process, is the maximum-smoothness theory (151; reviewed in 152). The predictions of the maximum-smoothness theory have not been explicitly tested during locomotion, but some experimental results fit the predictions. Trajectory of the toes during the swing phase is used as an example since planning of the foot trajectory has been proposed as one of the three sub-tasks necessary for safe walking (153). During obstructed locomotion, the path of the toes is curved and has a single peak curvature for the first part of the swing phase (154). The prediction of invariance of shape trajectories under translation and amplitude scaling is met since the placement or translation of the obstacle (154) or multiple height of obstacles (155) does not change the shape of the curve. These experimental results fit the predictions of planned movement with one accuracy point but an analysis of the tangential velocity to the obstacle is still lacking to establish if the movement is planned for maximum smoothness. Nevertheless, from the literature review it can be speculated that studying the adaptations required to clear an obstacle during walking can be used as a tool to assess the control (changes in the coordination function) and skill of the walking behaviour.

ADAPTATIONS TO THE LOCOMOTOR PATTERN DURING OBSTRUCTED WALKING

Obstructed walking by humans has been studied using a spectrum of obstacle configurations. Bilateral obstacles have been placed in the late swing phase of the leading leg (156-162) or in the early swing phase of the trailing leg (163, 164) and unilateral obstacles have been placed at different points in the swing phase (154, 155, 159).

A study of the time required for the processing of the adaptations needed to clear a bilateral obstacle show that those adaptations can be established in less than one step (159). Changes in the temporal parameters of walking were an increase time of contralateral stance phase which changes from a mean of 677 ms during unobstructed walking to a mean of 705 ms for an obstacle of 80 mm (159). Furthermore, the swing phase of the ipsilateral leg increased from a mean of 547 ms to a mean of 597 ms while clearing the 80 mm obstacle (159). In a unilateral obstacle configuration, an increased percentage of the ipsilateral swing phase is also shown (154). The effect of obstacles on the walking speed is still unclear with one study showing no effect (156) and another showing a decreased walking speed as a function of obstacle height (157). Other temporal-spatial parameters did not show any significant difference between the unobstructed walking pattern and the obstructed pattern. The variables investigated were stride length (154, 156), distance between toes and obstacle before clearance, distance between heel and obstacle after clearance and step width (156).

Studies of the leading leg reported that toe clearance from the obstacle remained constant (157) or increased (156, 160) with obstacle height. One study showing an increased toe clearance over the obstacle characterised this relationship as hypermetric and nonlinear (156). As seen by the report toe clearance was on average 64 mm for a 25 mm obstacle but increased to 119 mm for a 152 mm obstacle (156). No studies reported the relationship of toe clearance as a function of obstacle height in a unilateral obstacle configuration. However, one study reported that the elevation of the toes during mid-swing increased from 6.8 mm during unobstructed walking to more than 150% of the height of the obstacle when the obstacle height was 10% of the lower limb length (155).

This adaptation of the trajectory of the toes during obstructed walking can be attributed to either an increased ipsilateral limb flexion or increased height of the ipsilateral pelvis. In bilateral obstacles, it was reported that the two components have a different relative contribution to the overall increase in toe clearance, with 20% of toe clearance due to ipsilateral pelvis elevation and 80% due to ipsilateral limb flexion (160).

The increased ipsilateral pelvis elevation could be attributed to an increased contralateral vertical impulse (159). It should be noted that the contralateral soleus also increased its activity during the stance phase (159). However, studies reporting the kinematic and dynamic data important to the understanding of the strategy used to elevate the pelvis in order to go over the different obstacles is still lacking even though two studies looked at external joint moments of the trailing limb (163, 164). In contrast, a different strategy may be used for unilateral obstacles placed at the mid-swing location. It can be postulated that the increased vertical hip position may be absent since the vertical braking impulse of the contralateral leg was not changed for such an obstacle placement (159).

The strategy used to increase the limb flexion while going over an obstacle has been more thoroughly investigated. It is reported that the hip, knee and ankle increase their flexion while going over the obstacle (155, 159). The amount of maximal flexion of the hip and knee is increased respectively up to twice and 1.5 times the values measured during unobstructed walking and is related to obstacle height (155). Furthermore, the relative timing of this maximal flexion is modified in relation to the obstacle height. As an example, this is seen for the knee joint with a maximal knee flexion occurring at 70% of the normalized walking cycle during unobstructed walking and at 80% when clearing an obstacle of moderate height (155). The ankle is modified slightly by reducing the amount of dorsiflexion during the preceding stance phase with an increase in dorsiflexion during the swing phase (155). This increased flexion has been attributed to a reorganisation of the energy generation toward a knee flexor strategy with the generation of a new power burst at foot-off which has been called an anticipatory locomotor adjustment (155). More precisely, it was shown that during the push-off, more precisely around toe-off, preceding the obstacle clearance there is a decrease in the amount of hip flexor moment and hip flexor power. At the knee, the dynamics are dependent on the obstacle height. For the lower obstacle (14 mm) there is a decrease in the extensor moment with a diminution of the extensor power. However, for the higher obstacle (10% leg length) there is a generation of flexor moment at the knee with the emergence of a flexor power burst (155). This strategy is robust since a study of obstacles placed at different position during the swing phase repeated those experimental results (154). The muscle activity of the ipsilateral leg did follow this strategy with an increased activity in the semitendinosus and biceps femoris (155, 159). However, the changes in the electromyographic activity seems to be dependent on the type of obstacle encountered with a reduction in the pre-swing phase for a bilateral obstacle (160) and an increased activity for a unilateral obstacle (155, 159). Concordantly, it has been shown, in bilateral obstacles, that changes in the angular kinematics of the hip are the product of intersegmental dynamics since rotational energy applied at the hip joint decreased with obstacle height (165). This modification of the generation and absorption of power at the knee and hip occurring at the start of swing changes the dynamic strategy from a "hip-pull" strategy during unobstructed walking to a "knee flexion" strategy for higher obstacles.

Models of obstructed walking by humans have also been presented in the literature (157, 166, 167). Chou and collaborators (157) tested the hypothesis that the adaptations to the foot trajectory was governed by the criterion of minimum mechanical energy. In their study they compared the predicted and measured modifications of the trajectory of the swing ankle during obstructed walking. Surprisingly, it was found that the adaptations during obstructed walking were not optimised for the reduction of mechanical energy. Anticipatory locomotor adjustments were also modeled based on a controller divided in a module that generates the desired movement goal and a feedforward control (166). In this model (157), the desired movement goal is generated by comparing the predicted endpoint of the movement and the static environmental characteristics through internal models. The variable extracted by using these internal models is the ratio between the height of the obstacle and the predicted, instantaneous, vertical position of the toe. Furthermore, the feedforward controller produces a set of weighting function through a front-end compensation based on the knowledge of the plant (muscle). Use of the interaction from a discrete movement generator and a neural rhythm generator for the anticipatory control of walking is presented in an more recent model (167). The neural rhythm generator had been presented elsewhere (168). In this model the discrete movement generator perceives the location of the obstacle and decides when to initiate adjustments of on-going walking patterns. Although not presented in the paper (167) some neural structures can be linked to the properties of this discrete movement generator. Mechanisms used in the recognition of the environment in regards of anticipatory locomotor adjustments are becoming more numerous (169, 170) and suggest the use of a different cortical pathway in the required transformations for the control of obstructed walking in contrast to the experiential perception of the visual world (171). As stated in the model (167), there is a decision process occurring between the recognisance of the obstacle and the implementation of the desired response. It can be proposed that such a decision process in the initiation of a movement involves connections from the association cortex to the motor cortex via cortico-cortical connections, connections through the basal ganglia or the lateral part of the cerebellum and dentate nuclei (172). These processing pathways would however be influenced by the sensory analysis and motor responses required (173). Lastly, the investigation of the implementation of the desired response during obstructed walking of cats has shown to involve motor cortex as well as the reticular formation (72, 85-87, 174-177).

It was discussed in this literature review that even though SCI-IMFL have a disability, the use of FES assisted walking could prove to be beneficial in the recovery of functional walking. But evidence is still lacking for this population on the both the effect of the orthosis on the walking pattern as well as on its therapeutic effect. Furthermore, the use of obstructed walking in the evaluation of the control of walking has also been discussed. But no studies have ever investigated the modulation of toe height, hip elevation, and knee flexor power generation as a function of obstacle height as well as the presence of adaptations during obstructed walking by SCI-IMFL. Hence the objectives of this thesis are three-fold: to investigate the changes in the walking behaviour of SCI-IMFL with the use of FES-assisted walking, to investigate the control of the foot trajectory during walking by able-bodied participants and SCI-IMFL.

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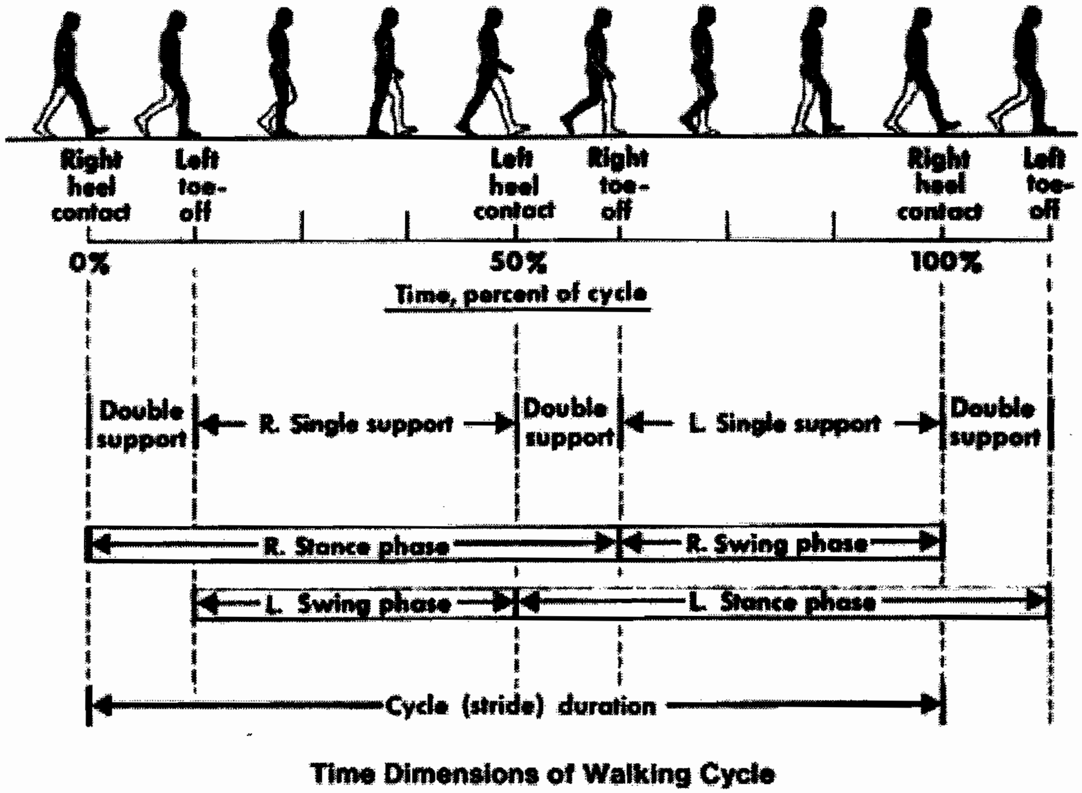

Figure 1. Temporal characterisitcs of the gait cycle (Adapated from Inman VT et al., 1981)

Functional electrical stimulation-assisted walking for persons with incomplete spinal injuries: Longitudinal changes in maximal overground walking speed

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INTRODUCTION

Until recently, a spinal cord injury usually resulted in the loss of all motor and sensory functions below the injury site. With improvement in the primary care, as well as other factors, a greater proportion of spinal cord injured persons retains some motor and/or sensory functions. The epidemiology of spinal cord injury has resulted in an increase in the percentage of spinal cord injured persons with an incomplete motor function loss (SCI-IMFL) (1). Restoration of walking is possible for a greater proportion of the spinal cord injured population because of this rise in SCI-IMFL percentage. Electrical stimulation of the common peroneal nerve, eliciting a motor response from the tibialis anterior muscle as well as a flexion withdrawal reflex, can be used like an active orthosis to assist the swing phase of walking.

Functional electrical stimulation (FES) was developed more than 30 years ago to prevent "foot drop" in hemiplegic (2) or SCI (3) persons. FES has been used to restore a variety of movements, including walking (4), and was first reported for SCI-IMFL in 1989 (5). Simple systems utilizing FES-assisted walking with common peroneal nerve stimulation were efficacious for the hemiparetic population (6, 7) but showed a minimal effect for SCI-IMFL (8, 9). Walking speed was increased on average by 0.08 m/s, independent of initial walking speed, whereas oxygen consumption during gait decreased by 3%. Furthermore, studies of gait modulation with the orthosis turned off showed a therapeutic effect in participants who had suffered a stroke or head trauma (for a review see (10)). Such an effect has also been reported for the SCI-IMFL population (11, 12), which showed a minimal effect during a 12-week FES-assisted walking programme (range 0.0 to 0.1 m/s) (12).

The objective of this study was to characterize the magnitude and time course of the changes in maximal overground walking speed (MOWS) resulting from the use of the FES-assisted walking for SCI-IMFL. Preliminary results of this study have been reported previously (13-15).

METHODOLOGY

Fourteen SCI-IMFL, with an average age of 33 years (range from 25 to 48.9) participated in the study and constituted a sample of convenience. The time interval between the injury and the start of the FES-assisted walking programme ranged from 1.8—19.1 years. Participants' lesions ranged in neurological level from the fifth cervical (C5) to the first lumbar dermatome (L1). According to the International Medical Society of Paraplegia (IMSOP) classification (16), the participants were in either in the C category (5/14) or D category (9/14). Furthermore, the participants spanned the range of functional categories from community (n = 4) to least community (n = 3) and household walkers (n = 6) (17). All participants required an ambulatory assistive device at the onset of the study. The ambulatory assistive devices used were either a walker (n=6), a forearm crutch (n=4) or a cane (n=4). Mobility was evaluated on a functional scale consisting of 13 items scored on a 7-point scale (maximum score: 84 points). The psychometric properties of the scale have been tested and some modifications made (18). The mobility score ranged from 45—83 points.

FES-assisted walking was accomplished by stimulating the common peroneal nerve to help initiate and accomplish the swing phase and quadriceps stimulation was provided to help maintain the extension of the knee during the stance phase (12). The common peroneal nerve and quadriceps stimulations were generated by one of three stimulators depending on the required number of channels and availability of the device. When stimulation of the quadriceps and the common peroneal nerve was required, the participants used a Quadstim stimulator (Biomotion, Inc.), which was capable of providing monophasic stimulation with a constant current. The parameters of stimulation were adjustable for the output current (0—150 mA), stimulus frequency (10—30 Hz) and stimulus pulse width (50—500 μs). When the participant required only one channel of stimulation, they were fitted with either a Unistim (Biomech Engineering Ltd) or MikroFES (Institut Jozef Stefan) device. The Unistim stimulator provided monophasic stimulation with a constant voltage output. The output voltage was adjustable from 0—100 V, whereas the stimulus frequency and pulse width were constant (23 Hz and 300 μs, respectively). The MikroFES stimulator also provided monophasic stimulation, with a constant voltage output that could be adjusted from 10 to 130 V. The frequency of stimulation was fixed at 25 Hz with a pulse width of 150 μs. The triggering of the common peroneal nerve stimulation with these devices could be made either with hand or foot switches that used either force-sensing resistors (Interlink, Inc.) or mechanical contacts. Details of the participants, ambulatory assistive device and channels of stimulation used in this study can be found in Table I.

Prior to the start of FES-assisted walking all the participants signed a consent form. Following a 4-week initiation program on the proper use of the FES stimulator, the participants were asked to use FES-assisted walking as much as possible in their activities of daily living.

To evaluate the MOWS, the participants were asked to walk as fast as possible on a 15-m walkway. The surface of the walkway was an industrial carpet with a high coefficient of friction. Participants were allowed to start walking before the starting line in order to accelerate and reach a steady speed before the start of the timing of the performance. The performance in the middle 10 m of the walkway was timed with a handheld stopwatch and the values transformed into an average MOWS. This outcome measure was shown to have good reliability for a sample of persons suffering from rheumatism (19). The participants were asked to perform under different randomized conditions: with the use of the FES orthosis, without orthosis, and with their various ambulatory assistive devices. Measures of the MOWS with or without the FES orthosis determined the combined and therapeutic values, respectively. The difference in MOWS between the combined and therapeutic values established the orthotic effect (20).

The time course of the changes in MOWS was fitted with two models, a linear model (Ymodel=A+BX) and a model of exponential association:

(Ymodel =Ystart+Yspan!(1-e-K!DELTA_X)

where Ystart = MOWS at the start of FES-assisted walking, Ymax = MOWS reached during the period of FES-assisted walking, Yspan=Ymax - Ystart, Xstart = week during which the FES-assisted walking was done using a particular ambulatory assistive device and DELTA_X = X - Xstart. Some variables of the equation were determined and kept constant (Ystart, Xstart) whereas others were calculated using a non-linear regression (Prism v3.0, GraphPad Software). Richardson's method was used to evaluate the derivatives during the fitting of the non-linear regression. The model with the lowest residual sum-of-squares was used to represent the experimental data.

RESULTS

Changes within the first year

When possible, measures of the MOWS were done repeatedly to establish a baseline measure of spontaneously occurring changes. No changes in MOWS were found in participants that were evaluated at least three times prior to the start of the FES-assisted walking study. This result indicates that no spontaneous recovery had occurred prior to the start of FES-assisted walking.

FES-assisted walking has a functional impact, as seen by the increase in functional mobility within the first year (*t*-test = 6.702, df: 13, $p < 0.000$). This increase in functional mobility is dependent on many factors, but changes according to which ambulatory assistive device is used during walking (Table I) and when walking speed increases. Table I also reports changes within the first year in the orthosis appropriate for each participant. Some participants (3/14) needed a simpler system of stimulation within the first year of FES-assisted walking.

Within the first year of FES-assisted walking, the combined MOWS increased on average by 0.26 m/s (standard deviation (SD): 0.24; Fig. 1A) and the therapeutic MOWS increased by 0.25 m/s (SD: 0.18; Fig. 1A). Both were significantly different from zero (t = 4.106, df=13, p=0.0012 and t = 4.821, df=13, p=0.0003, respectively). The increases in MOWS for the combined condition ranged from -0.02 to 0.79 m/s (Fig. 1B) and from 0.00 to 0.69 m/s for the therapeutic condition (Fig. 1C). The changes in the orthotic effect were not significant (t = 0.2624, df=13, p=0.7971), ranging from -0.22 to 0.39 m/s (Fig. 1D). These increases were normally distributed as tested by the Kolmogorov-Smirnov (KS) distance test (21) (KScombined: 0.2323, $p > 0.10$; KStherapeutic: 0.2267, $p > 0.10$; KSorthotic: 0.1504, $p > 0.10$). The values of the MOWS for all the conditions of each participant as well as their changes can be found in Table II. These increases in MOWS were correlated to the initial value of MOWS for both the combined ($r = 0.57$, $p = 0.033$) and therapeutic conditions ($r = 0.69$, $p = 0.007$), as shown in Fig. 2 (panel A). However, the initial orthotic effect was inversely correlated to the initial therapeutic walking speed as shown in panel B of Fig. 2 ($r = -0.73$, $p = 0.003$).

Longitudinal changes of MOWS

The increases in MOWS were not instantaneous, but followed a longitudinal progression for both combined and therapeutic conditions. Figure 3 shows the changes in MOWS for the participants in the combined condition. Changes in MOWS occurred for the three functional categories of walkers (A: Community walkers, B: Least community walkers, C and D: Household walkers) with no striking difference between categories in terms of the model that best described the longitudinal changes. The increase in MOWS in the combined condition was a percentage of the initial MOWS (71%, SD: 65). This idea is supported by the fact that the ratio of increase (R(C) from Table II) was not correlated to the initial MOWS ($r = -0.02$, $p = 0.944$). In addition, as reported above, the magnitude of change correlated to the initial walking speed, showing a greater increase for the participants that were community walkers at the onset of the study. Of the 14 participants, 2 had no changes in MOWS in the combined condition. Of the remaining 12 participants, the longitudinal changes of 4 were best described with a linear model, whereas those of the other 8 participants were best described with an exponential association model. This difference in the qualitative aspect of the increase in MOWS, as represented by the different models of change, demonstrates the variety of longitudinal changes. The values of the parameters of change (either K or B) for each participant can be found in Table III. Like the longitudinal changes in MOWS in the combined condition, the increases in the therapeutic condition were not instantaneous, but depended on the time since the start of FES- assisted walking. Furthermore, as seen in Fig. 4 and in the longitudinal changes in MOWS of the combined condition, the longitudinal changes were similar in many aspects between functional categories of walkers (A: Community walkers, B: Least community walkers, C and D: Household walkers). These changes consisted of a percentage of the initial MOWS (70%, SD: 107) as seen by the fact that the ratio of increase (R(T) from table II) was not correlated to the initial walking speed ($r = -0.35$, $p = 0.288$), whereas the absolute increase in MOWS was correlated. Figure 4 reports the longitudinal changes of the 12 participants for whom there was a change in therapeutic MOWS. Like the combined MOWS, the longitudinal changes for eight participants were best described with an exponential association model of increase, whereas the changes in four participants were best described using a linear model. Individual values of the parameter of change can be found in Table III.

Comparison of the parameters of changes (Table III) for the combined and therapeutic conditions, when the participants used the same ambulatory assistive device, shows that the parameters were similar for half of the participants (7/14) in the two conditions. Two had different parameters of changes (participants DT and MR) that remained similar qualitatively, whereas two participants, different models were necessary (participants MS and SM). Their changes were best described using a linear model for one condition and an exponential association model for the other. Three of the 14 participants could not be compared because in one or both conditions, combined or therapeutic, there were no changes with FES-assisted walking. Parameters of longitudinal changes for the participants that were able to use different ambulatory assistive devices were calculated. Comparison between devices revealed a difference for only one participant (MR) out of nine (Table III).

The orthotic effect varied according to both the participant and the time since the start of FES-assisted walking. Some participants (4/14) had an increase in the orthotic effect over time (Fig. 5A), others (4/14) decreased over time (Fig. 5B), and some participants (4/14) showed no change in the orthotic effect with time (Fig. 5C). For two participants the results were mixed: there was an increase in the orthotic effect for the first year, followed by a time-dependent decrease (Fig. 5D). It should be noted that the orthotic effect ranged widely, from -0.30 to 0.35 m/s.

In summary, longitudinal changes in MOWS showed a wide variety of responses. Most participants showed increases in both the combined and therapeutic conditions that were qualitatively and quantitatively similar for the different conditions and according to the different ambulatory assistive device used. However, the longitudinal parameters of change differed in magnitude for some participants, as did the qualitative models of change. It should also be noted that one participant exhibited changes in only the combined condition (RL) and showed no therapeutic effect, whereas another showed changes in the therapeutic condition (SH) alone, with no change in the combined condition.

DISCUSSION

The aim of this study was to characterize the magnitude and time course of the changes in MOWS using FES-assisted walking for SCI-IMFL. The main result is that SCI-IMFL who use an FES orthosis have improved mobility scores due to an increased MOWS and to the change in the type of ambulatory assistive device used. This increased is also seen during the same time course when the FES orthosis is turned off.

Unlike previous studies of the effect of simple systems of FES-assisted walking for SCI-IMFL (8,9,11), which showed a minimal effect on walking speed, this study showed an averaged increase of the walking speed three orders of magnitude higher than previously reported for the whole spectrum of SCI-IMFL. This greater increase can be explained by two factors. First, we used the MOWS instead of the comfortable walking speed used in other studies. Secondly, the duration of our study was longer. In previous studies, the time since the start of FES-assisted walking was either not taken into account (9) or was controlled with only a relatively short duration (8, 11). The models fitted to the experimental data in the present study reveal that time since the start of FES-assisted walking is a factor in the increase in MOWS. Even the effect of the orthosis was dependent on the time since the start of FES-assisted walking.

Most of the increase in MOWS in the combined condition stems from the therapeutic effect of FES-assisted walking, showing that plasticity in walking behaviour is possible for chronic stage SCI-IMFL. This therapeutic effect may be due to factors such as plasticity of the peripheral system and within the central nervous system. Activating muscles by electrical or voluntary means incurs changes in the properties of their fibres (22). No study so far has reported the effect of FES-assisted walking using common peroneal nerve stimulation on changes to the muscle fibre properties of either tibialis anterior or triceps surae. However, studies of electrical stimulation of the peroneal nerve have shown stimulus-dependent changes in muscle properties (23, 24). Our participants reported that lower leg muscles hypertrophied while using FES- assisted walking (probably from an increased muscle cross-sectional area related to the increased muscular activity).

Plasticity is characteristic of many sites within the nervous system. Changes in synaptic activity have been shown from connections in the monosynaptic stretch reflex (25) to cortical sensorimotor maps (26). More important for this study is the fact that electrical stimulation of the common peroneal nerve has been shown to reduce the amplitude of the

H-reflex in the soleus muscle (27). Furthermore, some participants reported fewer spasms and better bowel and bladder control. These changes in the activity-dependent feedback from all the receptors coupled with changes in the muscles could trigger a reorganisation of the planning and generation of walking behaviour.

Behavioural modifications can be seen in the increased strength of the lower limbs and changes in the output of the motor patterns. An increase in ankle dorsiflexion strength has been reported for hemiparetics that received common peroneal nerve stimulation while walking or sitting (28). Three types of responses to a programme of electrical stimulation of the quadriceps have been shown in SCI-IMFL: some participants experience increases in both their voluntary and electrically-activated strength, others only in their electrically-activated strength and some showed no changes in either condition (12). These three types of changes in lower limb strength are similar to the changes reported above for MOWS in our participants. Change in walking behaviour using FES-assisted walking have been reported in the population of persons who have experienced a cerebrovascular accident. One case study of a person who had experienced a cerebrovascular accident and was using FES-assisted walking showed an improvement in the control of weight acceptance on the paretic side, changes in the kinematic pattern of walking with a return to normal knee flexion at toe-off, and peak knee flexion and knee extension at heel strike (29).

An increase in walking speed fitting an exponential association model has also been seen in SCI persons with complete motor function loss who used FES-assisted walking (30), as well as in participants in a computer-assisted gait training program for hemiparetic persons (31). Because the changes were similar in all conditions, it is hypothesized that changes occurring using FES-assisted walking are systemic changes that could also be used in other motor activities.

Because of the heterogeneity of our sample, the results of this study have many functional implications relevant to the whole spectrum of SCI-IMFL. Improvements in MOWS were seen in the whole spectrum of SCI-IMFL with greater absolute effects for the participants that were walking faster at the onset of the study. Furthermore, the reported time courses of the changes in MOWS allow us to suggest that clinical trials on the effect of an FES orthosis for SCI-IMFL should use a period longer than 12 weeks of training and should evaluate the therapeutic effect of the orthosis. Due to discrepancies between the results of this study and those of previous studies (8, 9, 11) of the effect of FES-assisted walking in SCI-IMFL, we propose that evaluation of the changes in the walking behaviour occurring with an intervention should take into account at least two factors, such as the control and capacity of the walking behaviour (32). These factors can be evaluated using walking tasks that demand variation of the walking pattern from the usual comfortable speed on a flat, overground surface. Furthermore, this study shows that the FES orthosis has potential as a training tool in the rehabilitation of walking for SCI-IMFL, and that it will help in restoring walking behaviour.

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Table I. Characteristics of the participants. Abbreviations: IMSOP, International Medical Society of Paraplegia; AAD, ambulatory assistive device; C, cane; K, forearm crutches; W, walker; N/A, no AAD; Qr - Ql, stimulation of the right and left quadriceps; Pr - Pl, stimulation of the right and left common peroneal nerve; ✚', stimulation of the tibial nerve.

					FES orthosis (start)						FES orthosis (end of 1st year)					
Participants	Neurological Level of Lesion	IMSOP	Time Post Injury (years)	Age (years)	Mob. Score	AAD (init)	Qr	Ql	Pr	Pl	Mob. Score	AAD (end)	Qr	Ql	Pr	Pl
AC	T12-L1	P_D	12.3	34.0	80	1C			✚	✚	81	1C			✚	✚
DB	T8	P_D	6.6	36.0	74	2K			✚	✚	82	N/A			✚	✚
DT	T11	P_C	3.1	31.8	45	W			✚	✚	58	W			✚	✚
FG	C5-C6	T_D	3.7	48.9	62	2C			✚	✚	71	2C			✚	
JB	T10	P_D	2.5	37.8	78	2C			✚		81	2C			✚	
LR	T10	P_D	19.1	32.4	63	2K			✚	✚	73	2C			✚	✚
LS	T9-T12	P_C	9.9	36.3	60	W	✚	✚	✚	✚	64	2K	✚	✚	✚	✚
MA	C5-C6	T_C	6.1	36.3	54	W			✚	✚	61	2K			✚	✚
MR	C3-C6	T_D	6.3	26.8	83	1C				✚'	84	N/A				✚'
MS	C6	T_D	8.8	27.6	73	2K			✚	✚	76	2K			✚	
RL	C5	T_D	1.8	29.1	59	2K			✚		66	2K			✚	
RP	C5-C6	T_C	4.3	28.5	58	W	✚	✚	✚	✚	67	2K			✚	✚
SH	C7	P_C	6.1	30.8	64	W			✚	✚	70	W			✚	✚
SM	C5-C6	T_D	4.3	25.0	66	W				✚	75	2K				✚

Table II. Changes in the maximal overground walking speed occurring within the first year of FES-assisted walking. Abbreviations: Part, participant; AAD, ambulatory assistive device; C, cane; K, forearm crutches; W, walker; Init(C)-Init(T), initial walking speed in the combined and therapeutic condition; Init(O), initial orthotic effect; End(C)-End(T), maximal walking speed obtained within the first year in the combined and therapeutic condition; End(O) maximal orthotic effect; D(C)-D(T)-D(O), difference between the initial and maximal values in the combined and therapeutic condition as well as in the orthotic effect; R(C)-R(T) percentage of improvement during the first year.

		Combined Effect				Therapeutic Effect				Orthotic Effect			Comparison	
Part	AAD	Init(C) $m{*}s^{-1}$ (SD)	End(C) $m{*}s^{-1}$ (SD)	D(C) $m{*}s^{-1}$	R(C) %	Init(T) $m{*}s^{-1}$ (SD)	End(T) $m{*}s^{-1}$(SD)	D(T) $m{*}s^{-1}$	R(T) %	Init(O) $m{*}s^{-1}$	End(O) $m{*}s^{-1}$	D(O) $m{*}s^{-1}$	D(C)-D(T) $m{*}s^{-1}$	D(T)/D(C) %
AC	2C	1.11 (0.07)	1.32 (0.04)	0.19	19	1.11 (0.03)	1.28 (0.02)	0.17	15	0.00	0.04	0.04	0.02	89
DB	2K	0.97 (0.01)	1.76 (0.07)	0.79	81	1.02 (0.08)	1.71 (0.04)	0.69	68	-0.05	0.05	0.10	0.10	87
DT	W	0.19 (0.02)	0.39 (0.05)	0.20	105	0.05 (0.02)	0.24 (0.02)	0.19	380	0.14	0.15	0.01	0.01	95
FG	2C	0.55 (0.03)	0.67 (0.02)	0.12	22	0.48 (0.01)	0.70 (0.02)	0.22	46	0.07	-0.03	-0.10	-0.10	183
JB	2C	0.85 (0.03)	1.56 (0.04)	0.71	84	1.15 (0.06)	1.47 (0.02)	0.32	28	-0.30	0.09	0.39	0.39	45
LR	2K	0.57 (0.01)	0.82 (0.04)	0.25	44	0.57 (0.01)	0.82 (0.02)	0.25	44	0.00	0.00	0.00	0.00	100
LS	2K	0.13	0.22	0.09	69	0.00	0.20	0.20	NA	0.13	0.02	-0.11	-0.11	222
MA	2K	0.17 (0.01)	0.15 (0.00)	-0.02	-12	0.16 (0.00)	0.17 (0.00)	0.01	06	0.01	-0.02	-0.03	-0.03	-50
MR	1C	1.54 (0.02)	1.97 (0.08)	0.37	24	1.60 (0.01)	2.25 (0.13)	0.65	41	-0.06	-0.28	-0.22	-0.28	176
MS	2K	0.50 (0.03)	0.61 (0.01)	0.11	22	0.45 (0.07)	0.63 (0.02)	0.18	40	0.05	-0.02	-0.07	-0.07	164
RL	2K	0.18	0.43	0.25	139	0.10	0.10	0.00	00	0.08	0.33	0.25	0.25	00
RP	W	0.10	0.25 (0.02)	0.15	150	0.00	0.30 (0.01)	0.30	NA	0.10	-0.05	-0.15	-0.15	200
SH	W	0.13	0.17	0.04	31	0.00	0.19	0.19	NA	0.13	-0.02	-0.15	-0.15	475
SM	2K	0.18	0.58	0.40	222	0.18	0.37	0.19	106	0.00	0.21	0.21	0.21	48
Mean		0.51	0.78	0.26	71	0.49	0.74	0.25	70	0.02	0.03	0.01	0.01	131
SD		0.45	0.62	0.24	65	0.53	0.68	0.20	107	0.11	0.14	0.17	0.18	126
P				0.0012				0.0003				0.7971		0.0019

Table III. Parameters of the models of the longitudinal changes occurring with the use of FES assisted walking (combined and therapeutic effects for AAD). Abbreviations: Idem to table I except for 2K(a) which means the time adjusted function.

		Combined			Therapeutic		
Participant	AAD	Category	K or B	R^2	Category	K or B	R^2
AC	2C	Exp. Ass.	0.187 (0.067)	0.93	Exp. Ass.	0.238 (0.017)	0.99
DB	2K	Exp. Ass.	0.085 (0.032)	0.89	Exp. Ass.	0.100 (0.027)	0.93
	N/A		NA		Exp. Ass.	0.116 (0.039)	0.87
DT	W	Exp. Ass.	0.241 (0.075)	0.97	Exp. Ass.	0.056 (0.040)	0.92
FG	2C	Exp. Ass.	0.212 (0.105)	0.86	Exp. Ass.	0.705 (1.011)	0.81
	1C	Exp. Ass.	0.160 (0.013)	0.99	Exp. Ass.	0.580 (0.392)	0.96
JB	2C	Exp. Ass.	0.261 (0.051)	0.98	Exp. Ass.	0.181 (0.006)	0.99
	1C	Exp. Ass.	0.198 (0.091)	0.56	Exp. Ass.	0.217 (0.272)	0.40
LR	2K	Linear	0.021 (0.001)	0.99	Linear	0.020 (0.003)	0.97
LS	2K	Linear	0.004 (0.000)	0.98	Linear	0.006 (0.000)	0.99
MA	2K		NA			NA	
MR	1C	Exp. Ass.	0.229 (0.176)	0.83	Exp. Ass.	0.018 (0.028)	0.93
	N/A	Exp. Ass.	0.556 (0.335)	0.98	Linear	0.022 (0.003)	0.96
MS	2K		0.003 (0.002)	0.48	Exp. Ass.	0.500 (NA)	0.36
RL	2K	Exp. Ass.	2.992 (0.674)	0.96		NA	
RP	2K	Linear	0.001 (0.000)	0.67	Linear	0.001 (0.000)	0.81
SH	W		NA		Exp. Ass.	2.169 (0.834)	0.96
SM	2K	Exp. Ass.	0.023 (0.006)	0.92	Linear	0.002 (0.000)	0.81
	2K(a)	Linear	0.002 (0.000)	0.74	Linear	0.002 (0.001)	0.76
	2C	Linear	0.001(0.000)	0.90	Linear	0.001 (0.000)	0.88
	Mean (SD)	Exp. Ass.	0.215 (0.140)			0.271 (0.238)	
		Linear	0.005 (0.007)			0.008 (0.009)	

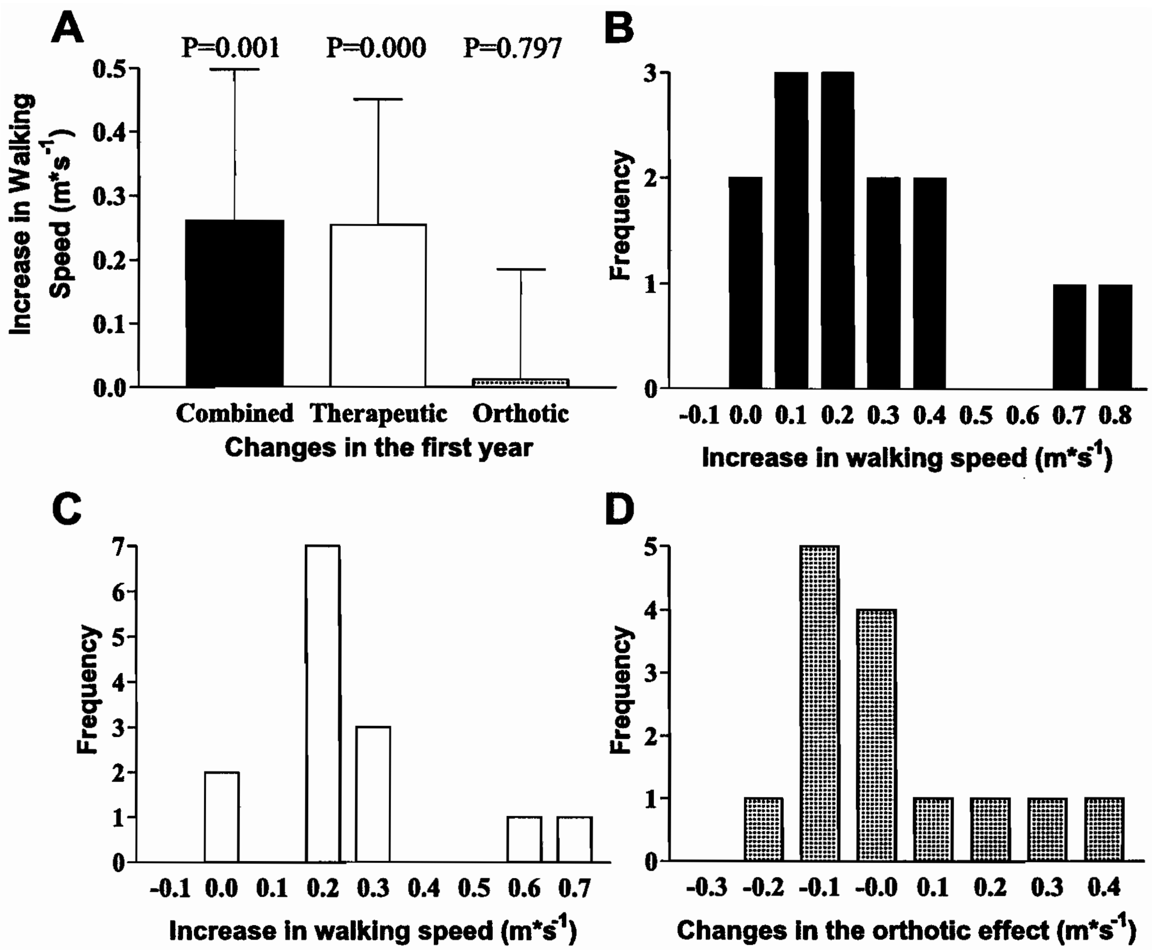

Fig. 1. Changes in the combined, therapeutic and orthotic parameters within the first year of using the FES-assisted walking. A. Average increases for the three parameters. B. Frequency distribution of the increases in maximal overground walking speed in the combined condition. C. Frequency distribution of the increases in maximal overground walking speed in the therapeutic condition. D. Frequency distribution of the changes in the orthotic effect. The error bars represent the standard deviation.

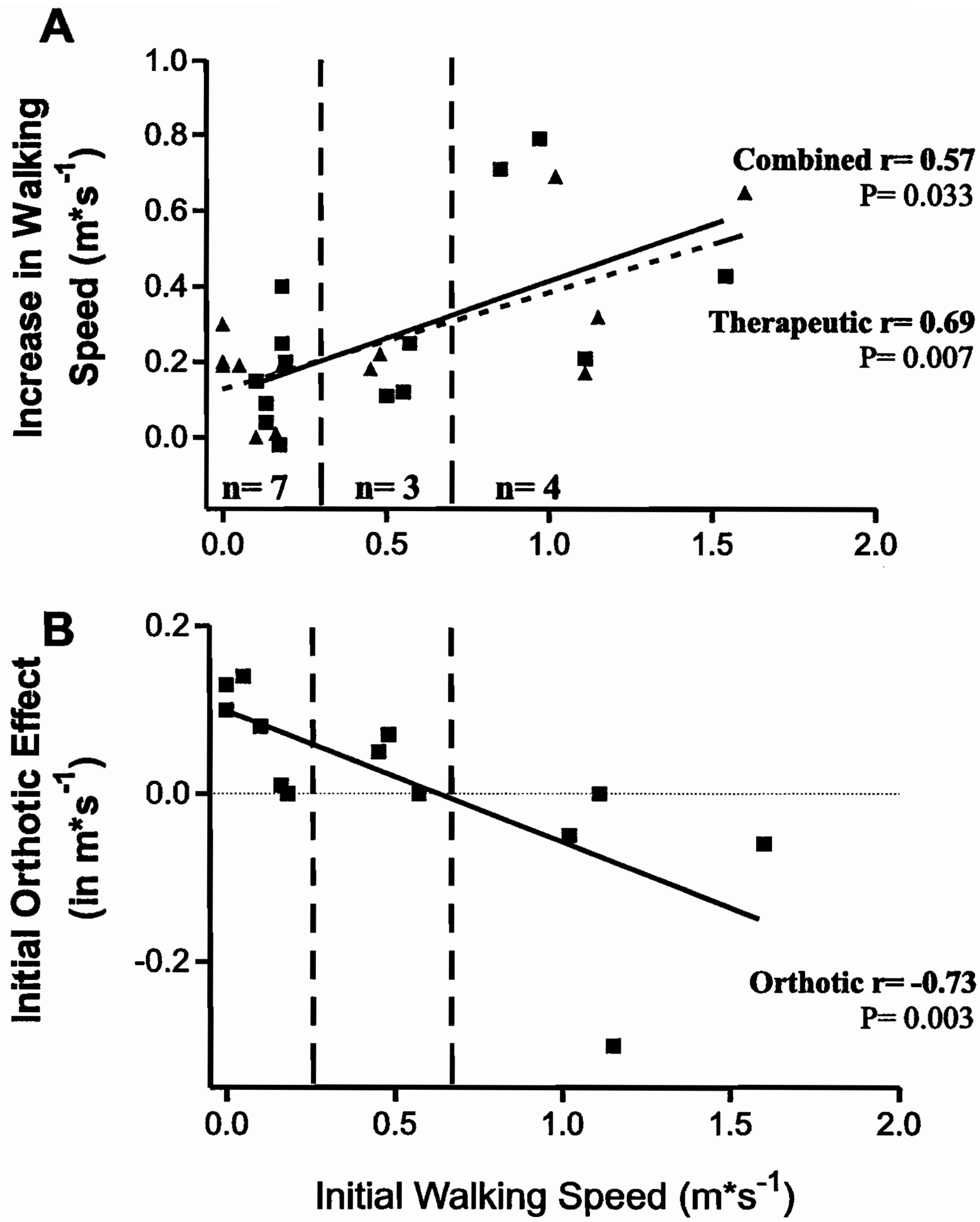

Fig. 2. Relations with the initial maximal overground walking speed. A. Relations between the increases found in the combined and therapeutic condition and their initial walking speed (combined condition: triangle, therapeutic: squares). B. Relation between the initial orthotic effect and the initial therapeutic maximal overground walking speed. The vertical lines represent the functional walking category (explanation in the text).

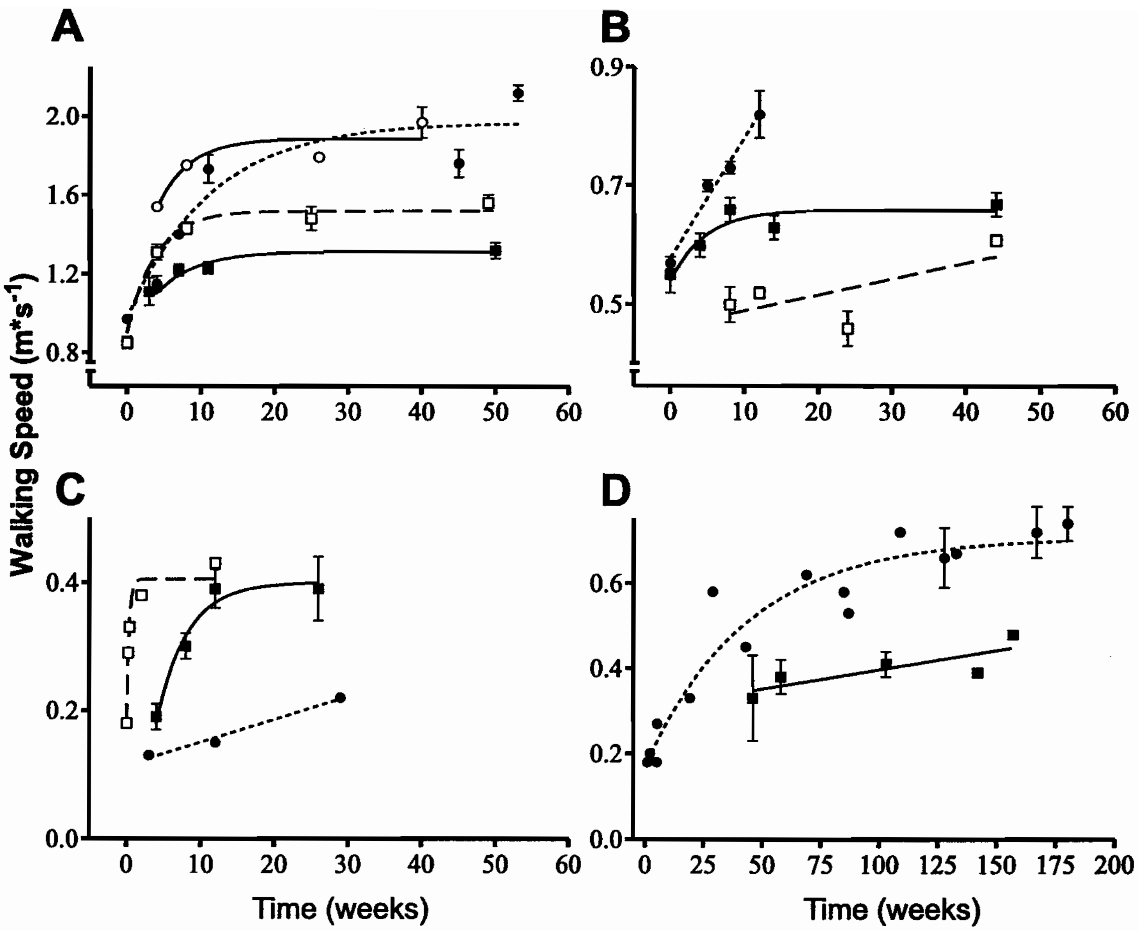

Fig. 3. Longitudinal changes of the combined maximal overground walking speed. The graphs represent the results of subjects that were respectively unlimited community (A: AC (filled squares, full line), DB (filled circles, short dotted line), JB (open squares, long dotted line), MR (open circles, full line)), least community (B: FG (filled squares, full line), LR (filled circles, short dotted line) , MS (open squares, long dotted line)) and household (C: DT (filled squares, full line), LS (filles circles, short dotted line), RL (open squares, long dotted line); D: RP (filled squares, full line), SM (filled circles, short dotted line)) walkers.

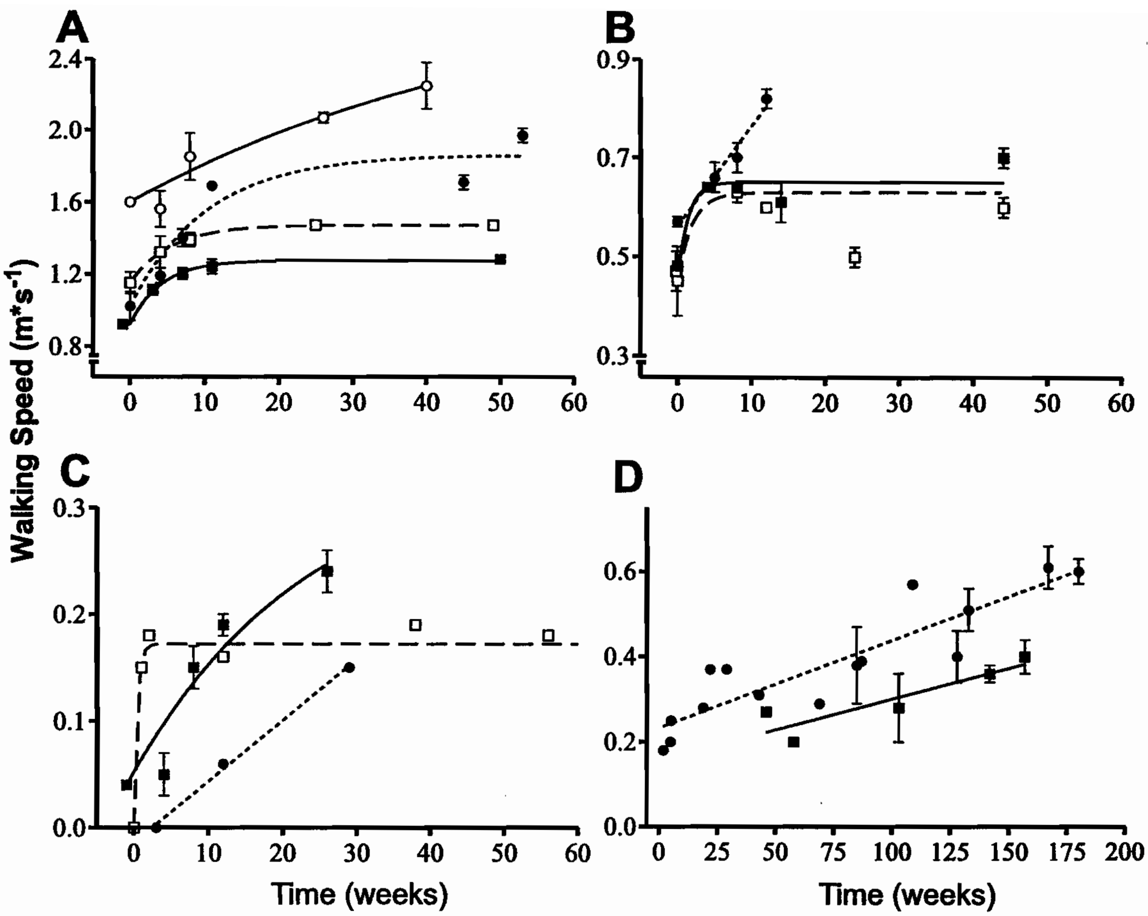

Fig. 4. Longitudinal changes of the therapeutic maximal overground walking speed. The graphs represent the results of subjects that were respectively unlimited community (A: AC (filled squares, full line), DB (filled circles, short dotted line), JB (open squares, long dotted line), MR (open circles, full line)), least community (B: FG (filled squares, full line), LR (filled circles, short dotted line) , MS (open squares, long dotted line)) and household (C: DT (filled squares, full line), LS (filles circles, short dotted line), SH (open squares, long dotted line); D: RP (filled squares, full line), SM (filled circles, short dotted line)) walkers.

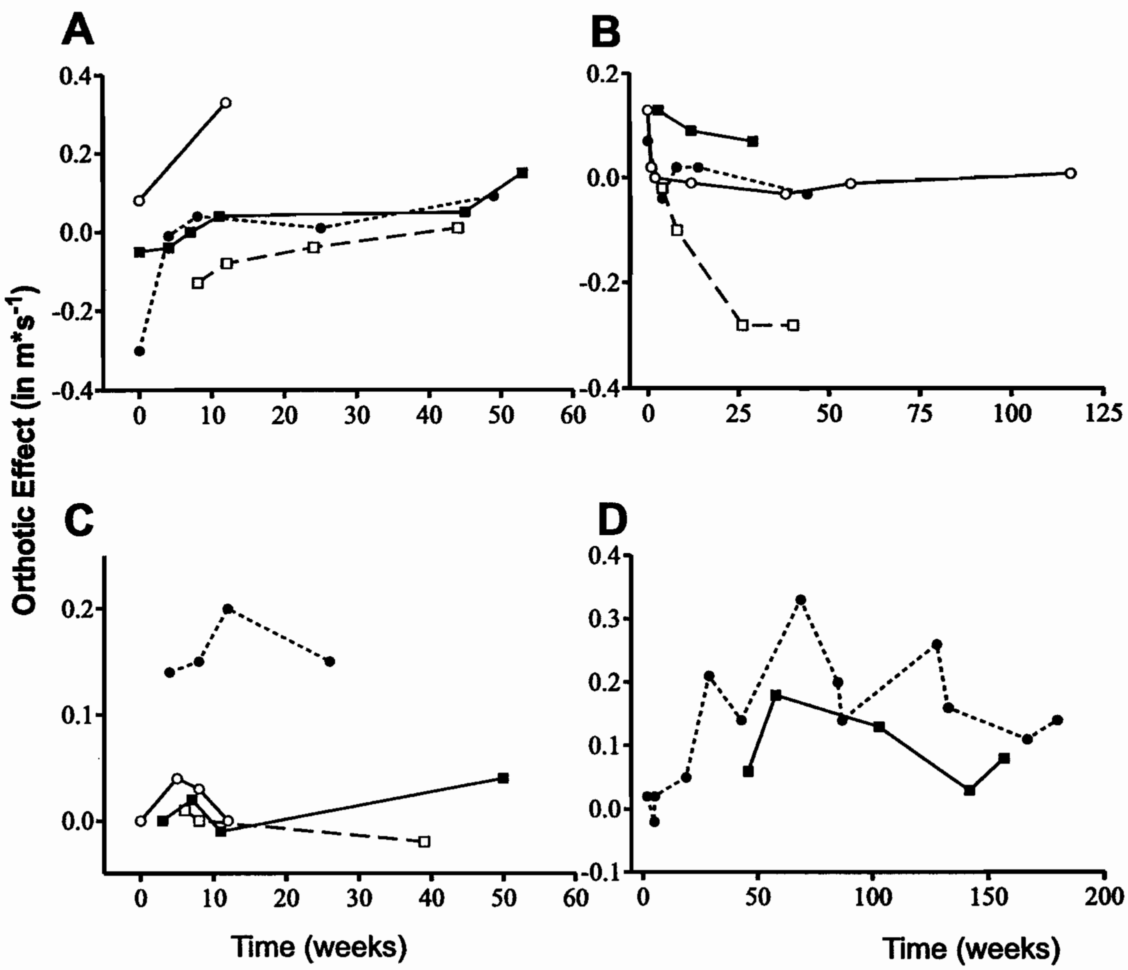

Fig. 5. Longitudinal changes of the orthodic effect. The graphs represent the results of subjects that learned to use the effect of the orthosis (A: DB (filled squares, full line), JB (filled cicrcles, short dotted line), MS (open squares, long dotted line), RL (open circles, full line)), used the orthosis less with time (B: LS (filled squares, full line), FG (filled circles, short dotted line), MR (open square, long dotted line), SH (open circles, full line)), did not change their use of the effect overtime (C: AC (filled squares, full line), DT (filled cicrcles, short dotted line), MA (open squares, long dotted line), LR (open circles, full line)) and learned to use the effect followed by a reduction of the effect after a certain time (D: RP (filled squares, full line), SM (filled circles, short dotted line)).

Functional electrical stimulation-assisted walking for persons with incomplete spinal injuries: changes in the kinematics and physiological cost of overground walking.
M. Ladouceur and H. Barbeau, Scandinavian Journal of Rehabilitation Medicine 2000;32:72-79.

INTRODUCTION

Increases in functional mobility for 12 out of 14 persons with spinal cord-injuries with incomplete motor function loss (SCI-IMFL) were shown in a previous study involving long-term use of functional electrical stimulation (FES) assisted walking (1). This increase was partly due to a change in the type of ambulatory assistive devices as well as time-dependent increase in the maximal overground walking speed whether the participant was using FES-assisted walking or not. The absolute increase in maximal walking speed with FES-assisted walking was negatively correlated with initial walking speed, and no differece was found between FES-assisted walking and non-assisted walking. Furthermore, changes were not instantaneous but were dependent on the time of start of FES-assisted walking.

Changes in walking speed, stride length and stride frequency with the use of FES-assisted walking have been reported for SCI-IMFL for electrically activated common peroneal (2,3) and tibial nerves (4). Stein and collaborators (3) reported an improvement of 0.08 m/s when using FES-assisted walking independently from their control walking speed. This improvement was not linked to changes in either stride length or frequency. Another study measuring the therapeutic effect of FES-assisted walking by SCI-IMFL showed no difference in walking speed or stride frequency but an increased stride length (5). This study also reported that three out of six participants increased their walking speed (5). This increase in walking speed could be explained by changes in stride length, stride frequency, or a combination of both factors (5). With the exception of one study using tibial nerve stimulation (4), joint angular kinematics was reported only as examples (3). No studies have reported kinematic changes occurring with the use of FES-assisted walking or have evaluated the changes during a long-term walking programme using FES-assisted walking.

Another factor in the evaluation of the functional and rehabilitation benefits of using FES-assisted walking is the energy requirement necessary for ambulation. Although wheelchair propulsion approximates the energy requirement of normal walking (6), the energy cost of walking by SCI-IMFL is higher than for speed-matched walking by able-bodied participants (3). When using FES-assisted walking, SCI-IMFL do not reduce their oxygen consumption (3), although a reduction in the physiological cost index (7) has been shown after 12 weeks of training (5).

The purpose of this study is twofold. First, changes in the kinematics of the lower limb are examined when the participant is using FES-assisted walking over a long period of training. Second, the purpose of this study is to report the changes in walking speed and the physiological cost of walking during a 5-minute walk when the participant is using FES-assisted walking over a long period of training.

METHODOLOGY

Fourteen SCI-IMFL participated in this study. All participants were also involved in a study of the changes in maximal overground walking speed with the use of FES-assisted walking (1). Relevant characteristics of the participants can be found in Table I.

Kinematics

The kinematic patterns in the sagittal plane and temporal variables were reconstructed from video recordings of the most affected side. The data were digitized and reconstructed using a Peak Performance Analysis system. Markers were placed on the lateral side of the 5th metatarsal, heel, lateral malleolus, knee joint axis, greater trochanter and acromion. The participants were asked to walk at their comfortable walking speed on a 5-m walkway. The video recordings were digitized and joint angle time-courses were calculated. The measurements followed the International Society of Biomechanics convention (8). The stride duration and stride length were extracted for each cycle from the heel marker trajectory. Hip, knee and ankle excursions were measured from the peak-to-peak values for each cycle from the joint angular excursions. Ankle angles at foot contact and during mid-swing were also extracted for each cycle from the ankle joint angular excursion. The results were compared using an analysis of variance for each parameter (repeated measure 2 X 2; stimulation X time; Systat, V5.0).

Physiological cost of walking

The physiological cost of walking, was measured with the physiological cost index (PCI). The index was established by dividing the difference between the heart rate during walking and during sitting with the walking speed measured in metres per minute. The result of this calculation is the physiological cost of walking in heart beats per metre walked. The recordings were made by asking the participants to sit for a period of 5 minutes, to stand for a period of 3 minutes and then to walk for a period of at least 5 minutes on a 15-m walkway. After the completion of the walking period, the participants were asked to sit again and to rest until their heart rate was back to the resting value found during sitting. Once the heart rate reached this level, the procedure was repeated. The use of FES-assisted walking was randomized. When possible, the participants were asked to repeat the procedures with the different ambulatory assistive devices they were able to use. Heart rate was recorded by a Polar Vantage XL heart rate monitor (Polar Electro Oy, Finland; see (9) for validity and stability). The values for sitting heart rate, walking heart rate, and walking speed represent the average of the last 30 s of recordings for the appropriate condition. Walking speed was calculated from times recorded by a manual stopwatch at each end of the 15-m walkway. A paired t-test was used to assess the effect of FES-assisted walking on walking speed and physiological cost of walking measures.

RESULTS

Kinematics

Spatio-temporal parameters

We analysed the spatio-temporal changes by using FES-assisted walking in 10 participants (Table I). The remaining four participants were not analysed because the initial data with the use of the FES-assisted walking were not collected or some of the participants stopped using this technique before a final kinematic evaluation could be performed.

Figure 1 shows examples of the changes in the spatio-temporal parameters found with a programme of FES-assisted walking (Figs. 1A-1C; one stride in each condition). A comparison of Figs. 1A and 1B shows the increased stride length (0.61 and 1.22 m) and reduced stride time (4.4 and 3.9 s) that occurred with long-term use (43 weeks) of FES-assisted walking even though the stimulator was turned off both times. The example also shows that using FES-assisted walking has a minor effect on the stride length (1.22 to 1.28 m) and stride time (3.9 to 3.2 s for this example, although the average in both conditions is 3.6 s) when the comparison is made for the same day (Figs. 1C versus 1B).

As shown in Fig. 2, training with FES-assisted walking during the first year increased walking speed by 0.10 m/s (Table II and Fig. 2A). This improvement in walking speed results from an average increase of 0.12 m in stride length and 0.04 Hz in stride frequency (Table II and Figs.2B and 2C, respectively). The increase in stride frequency is due to a decreased stance time of 0.22 s with only minor changes in the swing duration (Table II and Fig. 2D). Surprisingly, FES-assisted walking had only minor effects on any of the spatio-temporal parameters investigated in this study (Table II).

Longitudinal changes in the spatio-temporal parameters of one representative participant are shown in Fig. 2 (SM; Figs. 2E-H). It can be seen in this figure that changes in the spatio-temporal parameters are not instantaneous but are dependent on duration of the FES-assisted walking programme.

Joint angular parameters

Joint angular measurements were extracted for eight participants (Table I) but hip angular excursion was the only measurement extracted for one participant (LS). Figure 1 illustrates examples of the changes occurring with the use of FES-assisted walking as well as changes occurring with the training programme (Figs. 1D-F). One stride is shown for each condition. Hip angular excursion increases with FES-assisted walking but changes only minimally with time (Fig. 1D). Figure 1E shows the time-normalized joint angles of the knee. It can be seen that the angular excursion does not change with and without FES-assisted walking nor with time. However, the angular pattern is modified with FES-assisted walking, the knee being more flexed in the early stance phase and more extended in the later stance phase (from 0 to 30% and 50 to 80% of the normalized cycle time, respectively). Figure 1F represents the time-normalized joint angles of the ankle. Although the ankle joint angles during the stride does not change among the three conditions, there was an increased ankle dorsiflexion angle during the swing phase of gait when FES-assisted walking was used. The use of FES-assisted walking increased hip angular excursion by 3.2 deg (Table II and Fig. 3A). In addition, it increased ankle dorsiflexion during the swing phase by 10.9 deg and decreased ankle plantar/flexion at foot contact by 5.6 deg (Table II and Figs. 3B-C). In contrast, the knee and ankle angular excursions did not change (Table II). There were small, but not significant, changes in the joint angular parameters with the duration of the FES-assisted walking training programme (Table II).

Longitudinal changes in selected joint angular parameters of one representative participant are shown in Fig. 3 (SM; Figs. 3D-F). Unlike the longitudinal changes reported for the spatio-temporal parameters in Fig. 2, there is no clear increase in the joint angular parameters associated with the duration of the FES-assisted walking training programme. However in most of the evaluations there was a greater hip angle excursion, ankle angle at foot contact, and ankle dorsiflexion during swing with FES-assisted walking compared with the control condition (Figs. 3D-F, respectively).

Physiological cost of walking

The effect of FES-assisted walking on the physiological cost index and walking speed during the 5-minute walk was examined in 12 participants who had been using the programme for at least three months (Table I). Because initial measurement of the physiological cost of walking, as measured by the physiological cost index, was not acquired at the onset of FES-assisted walking for four participants the effect of time was not investigated. An example of the data recorded by the heart rate monitor can be found in Fig. 4A. The heart rate increased for participant RP from about 110 beats per minute when he was sitting to about 160 beats per minute when he was walking. Although, the changes in the heart rate remained constant, his walking speed doubled with the use of FES-assisted walking which diminished the physiological cost of walking by a factor of two. As seen in Fig. 4B, FES-assisted walking did not have any effect on either the physiological cost index or the walking speed during the 5-minute walk (Table II). Changes in the walking speed during the 5-minute walk with FES-assisted walking were correlated to the walking speed in the control condition ($r= -0.571$; $P= 0.052$) but not the physiological cost index ($r= 0.419$; $P= 0.175$).

This study shows that the use of FES-assisted walking does not change the walking speed or the physiological cost of walking. To evaluate the effect of training with FES-assisted walking on the physiological cost of walking we studied 9 participants longitudinally from onset of FES-assisted walking. When combined with time, we see five different types of responses: in three participants the physiological cost index remained constant and the walking speed increased, in three participants the physiological cost index decreased and the walking speed remained constant, in one participant the physiological cost index decreased and the walking speed increased (DT; Fig. 4C), in one participant the physiological cost index and walking speed increased, and in one participant the physiological cost index increased and the walking speed remained constant. These results show a positive effect for 8 out of the 9 participants who were evaluated for a duration of more than 3 months.

DISCUSSION

The aim of this study was to characterize the magnitude and time-course of changes in kinematic and physiological cost parameters of walking with FES-assisted walking for SCI-IMFL. The main results are that spatio-temporal parameters of walking are similar with and without FES-assisted walking but improve with the duration of the FES-assisted training programme and that both stride length and frequency are factors in the increased walking speed with time. In contrast, FES-assisted walking changes some joint angular parameters but training with FES-assisted walking does not alter, or only minimally, these parameters overtime. There was no difference in the physiological cost of walking nor walking speed when the participant used FES-assisted walking. However, the majority of participants that were followed longitudinally show some positive effects.

The minor differences observed in the walking speed with and without FES-assisted walking are similar to results found for maximal overground walking speed reported in a previous study (1). However, this is the first study showing that the increase in walking speed is related to an increase in both stride length and stride frequency and that changes in the stride frequency are related to changes in the duration of stance. Previous studies showed that the use of FES-assisted walking could be related to changes in both stride frequency and stride length (5,10) but trends could not be extracted because of the limited number of participants. This study is the first to report changes in joint angular parameters with use of a peroneal nerve stimulator for SCI-IMFL. In this study, there was no difference in the physiological cost of walking during FES-assisted walking. This is similar to results found by an earlier study using oxygen consumption (3). In addition, changes in the physiological cost of walking when the participants were not using FES-assisted walking have been reported previously (5). This study showed that these changes were progressive.

It is surprising to note that the changes in walking speed reported for the kinematic evaluation during the first year of FES-assisted walking are about half that reported for maximal overground walking speed (0.10 m/s in comparison to 0.25 m/s). Since there was no interaction factor between the use of FES-assisted walking and duration of the FES-assisted walking training programme, the improvement in walking speed is entirely due to the therapeutic effect of the FES assisted walking training programme. This result has also been showed in previous reports (1, 11). As discussed in the previous study, the improvement occurs because changes can take place at many sites within the nervous and muscular systems.

Because of the heterogeneity of our sample, the results of this study have many functional implications that can be relevant to the whole spectrum of SCI-IMFL. For example, this study shows that FES-assisted walking can be used as rehabilitation device that would enhance the recovery of walking. Furthermore, the results of this study show that even though the participants were classified as chronic SCI-IMFL it is still possible to improve their walking behaviour when an appropriate treatment modality is used.

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Table I. Characteristics of the participants. Abbreviations: IMSOP, International Medical Society of Paraplegia; AAD, ambulatory assistive device; C, cane; K, forearm crutches; W, walker; Qr - Ql, stimulation of the right and left quadriceps; Pr - Pl, stimulation of the right and left common peroneal nerve; ✚', stimulation of the tibial nerve.

								FES orthosis				Outcome variables analysed			
Participants	Neurological Level of Lesion	IMSOP	Time Post Injury (years)	Age (years)	Mob. Score (Init)	AAD (Init)	AAD (< 1 year)	Qr	Ql	Pr	Pl	Energy cost	Joint Angular	Spatio-temporal	Time for the < 1 year condition (weeks)
AC	T12-L1	P_D	12.3	34.0	80	1C	1C			✚	✚	✚	✚	✚	50
DB	T8	P_D	6.6	36.0	74	2K	2K			✚	✚	✚	✚	✚	45
DT	T11	P_C	3.1	31.8	45	W	W			✚	✚	✚			
FG	C5-C6	T_D	3.7	48.9	62	2C	2C			✚	✚	✚	✚	✚	44
JB	T10	P_D	2.5	37.8	78	1C	1C			✚		✚	✚	✚	49
LR	T10	P_D	19.1	32.4	63	2K	2K			✚	✚	✚			
LS	T9-T12	P_C	9.9	36.3	60	W	2K	✚	✚	✚	✚			✚	29
MA	C5-C6	T_C	6.1	36.3	54	W	2K			✚	✚	✚		✚	26
MR	C3-C6	T_D	6.3	26.8	83	1C	1C				✚'	✚	✚	✚	26
MS	C6	T_D	8.8	27.6	73	2K	2K			✚	✚	✚			
RL	C5	T_D	1.8	29.1	59	2K	2K			✚					
RP	C5-C6	T_C	4.3	28.5	58	W	2K	✚	✚	✚	✚	✚	✚	✚	46
SH	C7	P_C	6.1	30.8	64	W	W			✚	✚	✚	✚	✚	56
SM	C5-C6	T_D	4.3	25.0	66	W	2K				✚	✚	✚	✚	43

Table II. Statistical results of selected outcome variables. Abbreviations: F, results from an ANOVA; t, results from a student t-test; P, probability of type I errors.

	Initial- < 1 year		No-With FES	
	Result	**P**	**Result**	**P**
Spatio-temporal variables				
Walking speed	$F_{(1,9)} = 11.90$	0.007	$F_{(1,9)} = 0.00$	0.991
Stride length	$F_{(1,9)} = 4.43$	0.065	$F_{(1,9)} = 0.12$	0.737
Stride frequency	$F_{(1,9)} = 13.26$	0.005	$F_{(1,9)} = 0.40$	0.541
Stance duration	$F_{(1,9)} = 9.43$	0.013	$F_{(1,9)} = 0.24$	0.235
Swing duration	$F_{(1,9)} = 2.82$	0.127	$F_{(1,9)} = 3.22$	0.106
Joint angular parameters				
Hip angular excursion	$F_{(1,8)} = 1.68$	0.231	$F_{(1,7)} = 6.65$	0.033
Knee angular excursion	$F_{(1,7)} = 0.26$	0.625	$F_{(1,7)} = 0.73$	0.421
Ankle angular excursion	$F_{(1,7)} = 0.02$	0.883	$F_{(1,7)} = 0.94$	0.364
Ankle angle at foot contact	$F_{(1,7)} = 2.08$	0.193	$F_{(1,7)} = 8.54$	0.022
Ankle dorsiflexion angle during swing	$F_{(1,7)} = 0.31$	0.597	$F_{(1,7)} = 19.37$	0.003
Energy cost				
Walking speed			$t_{(11)} = 0.63$	0.543
PCI			$t_{(11)} = 1.18$	0.264

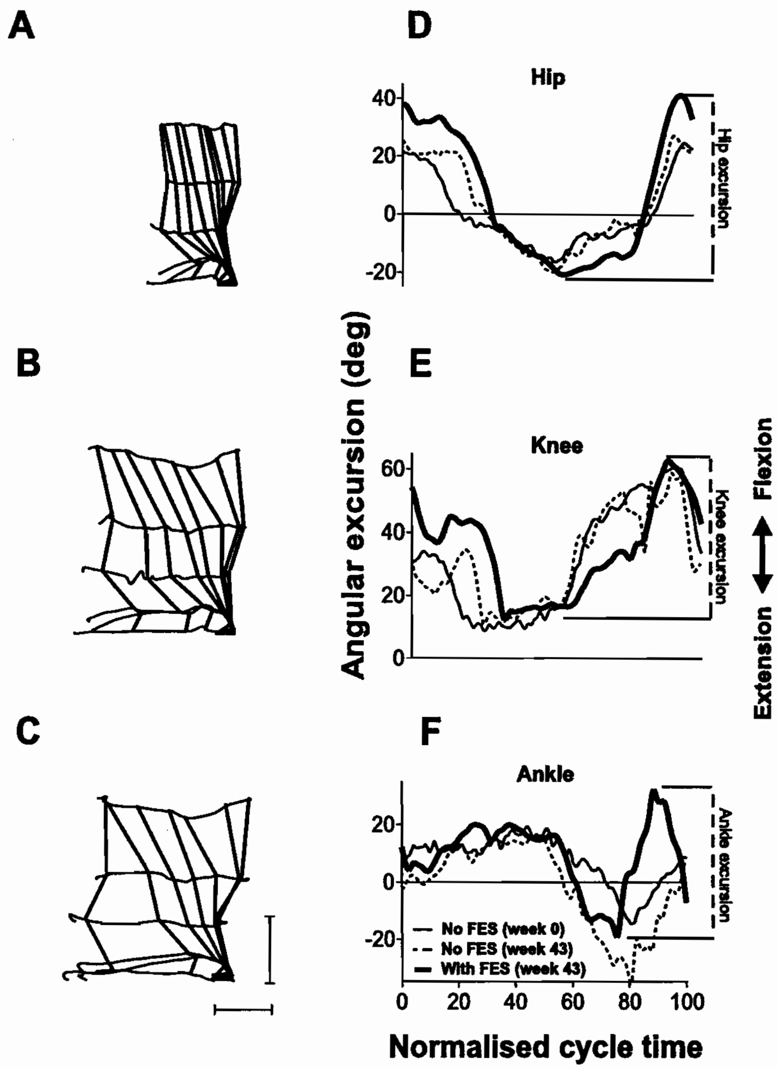

Fig. 1. Example of the changes in the kinematics with FES assisted walking. Panels A to C show the stick figure (every 0,5 s) and trajectories of retro-reflective markers (see methods) when SM walks without FES at the beginning of FES assisted walking (Panel A), without FES but after 43 weeks of FES assisted walking (Panel B) and with the FES assisted walking at the same evaluation session (Panel C). Panel D to F show the angular excursion of the hip (Panel D), knee (Panel E) and ankle (Panel F) of the affected lower limb in those trials. Calibration bars for panels A to C represent 0.5 m. The hip, knee and ankle angular excursions (Panel D to F) without FES assisted walking at onset of training, without FES assisted walking after 43 weeks of training, and with FES assisted walking after 43 weeks of training are represented by, respectively, thin, dotted and thick lines.

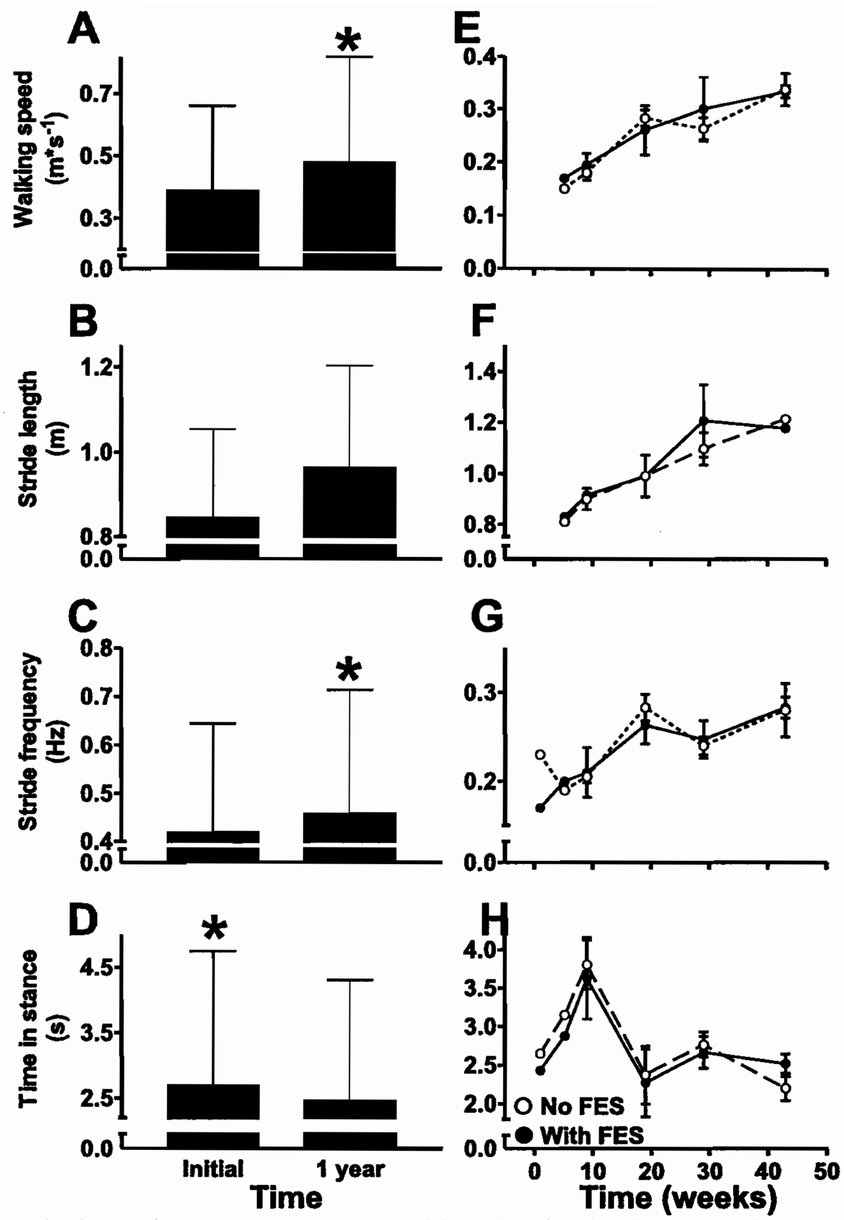

Fig. 2. Changes in the spatio-temporal parameters with FES assisted walking. Panel A to D show the significant changes occurring during the first year for walking speed (Panel A), stride length (Panel B), stride frequency (Panel C), and time in stance (Panel D). Examples of longitudinal changes in the walking speed (Panel E), stride length (Panel F), stride frequency (Panel G), and time in stance (Panel H) for 1 participant (SM) are also presented. Panel B, C, F and, G show that both stride length and frequency are changed when there is a modification of the walking speed. Panels C, D, G and H show that modification in the cycle time is related to modification of the stance time. Significant difference in the means are represented with an asterisk whereas the no and with FES assisted walking condition are represented, respectively, by open and filled circles.

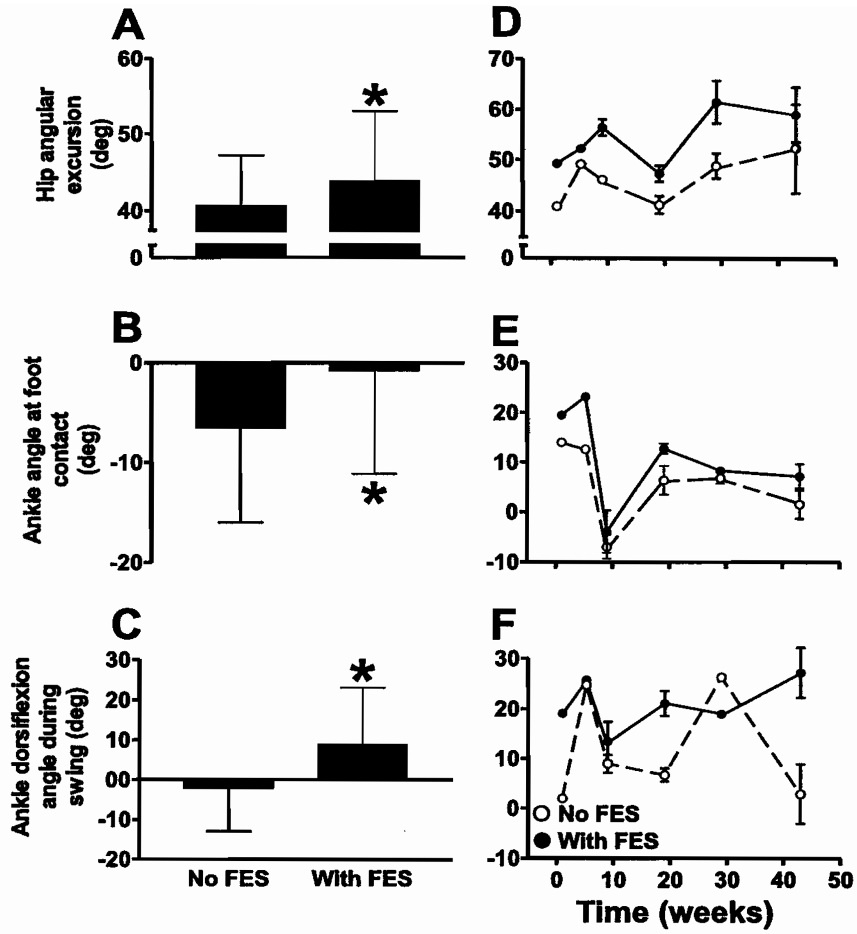

Fig. 3. Changes in the angular parameters with FES assisted walking. Panel A to C show the significant changes occurring with FES assisted walking for hip angular excursion (Panel A), ankle angle at foot contact (Panel B), and ankle dorsiflexion angle during swing (Panel C). Examples of longitudinal changes in the hip angular excursion (Panel D), ankle angle at foot contact (Panel E), and ankle dorsiflexion angle during swing (Panel F) for 1 participant (SM) are also presented. Panel D through F show that hip angular excursion, ankle angle at foot contact and ankle dorsiflexion angle during swing remain constant with long term use of FES assisted walking but differ between with and without FES assisted walking. Positive angular values represented in panel B, C, E and F represent dorsiflexion whereas negative values represent plantarflexion. Significant difference in the means are represented with an asterisk whereas the no and with FES assisted walking condition are represented, respectively, by open and filled circles.

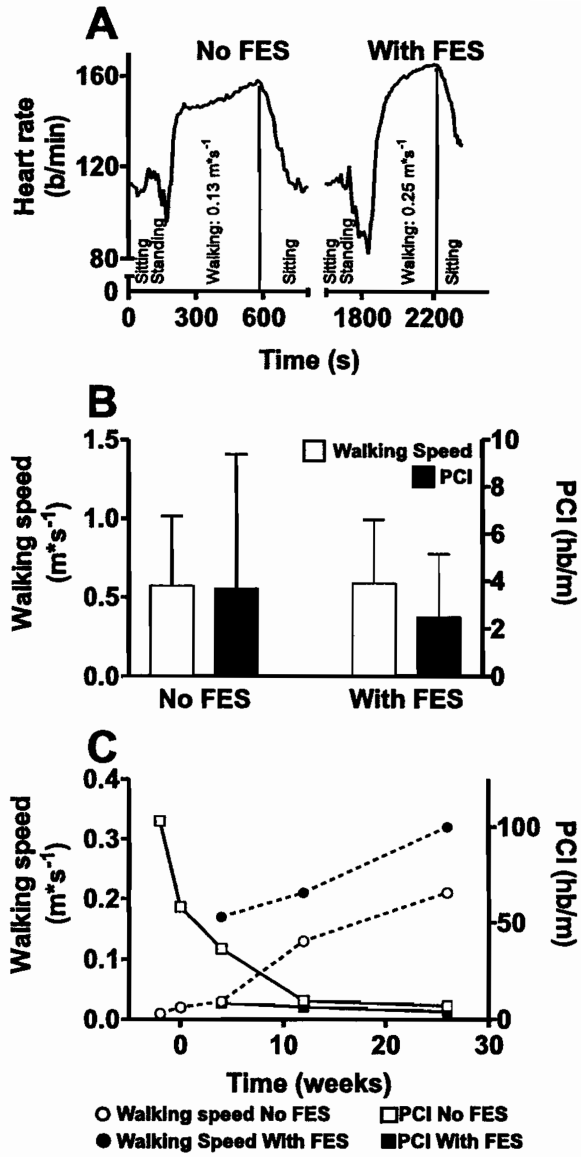

Fig. 4. Physiological cost of FES assisted walking. Panel A shows an example of the data used to calculate the PCI. Panel B shows the mean and standard deviation of the walking speed (clear bar and left axis) in the therapeutic (no FES) and combined condition (with FES) as well as for the PCI (filled bar and right axis). Panel C shows as an example of the longitudinal changes in the physiological cost of FES assisted walking the only participant (DT) that had a decrease in PCI combined with an increased walking speed. Significant difference in the means are represented with an asterisk. The no and with FES assisted walking condition are represented, respectively, by open and filled symbols whereas walking speed is represented by circles and PCI by squares.

Evaluation of the functional recovery of walking

The first two manuscripts of this thesis provide evidences that using different walking tasks will result in different perspectives on the recovery of the walking behaviour and on development of comprehensive treatment approaches. The sections that will follow give a rational for investigating new evaluative tasks and some experimental results on one of these new experimental tasks, namely obstructed walking.

A MODEL FOR THE EVALUATION OF WALKING BEHAVIOUR

The question of what is controlled during the walking behaviour has been investigated for many decades (1-4). An analysis of the kinematics and kinetics of human walking demonstrates that the walking behaviour can be segmented into four controlled sub-tasks: the generation and absorption of energy at key points in the cycle, the foot trajectory during swing, the support of body weight, and the balance of the head, arms and trunk (HAT) (5). These four sub-tasks are organised by the CNS through three interdependent systems of neurological structures (functional units): a functional unit that generates the basic locomotor pattern required to achieve propulsion, a second functional unit used for the maintenance of equilibrium, and a third unit used for the adaptation of the locomotor pattern to the behavioural goals of the persons as well as to the constraints imposed by the environment (6).

Fig. 2 represents a model of the evaluation of the walking behaviour. This model consists, in the centre, of the three functional units for the organisation of the walking behaviour. The inner pentagon illustrates the four sub-tasks that are controlled during the walking behaviour. The outer pentagon shows examples of the proposed locomotor tasks that would be used to evaluate the control and capacity of the individual. It is recognised that the four controlled variables are represented to a certain extent in all locomotor evaluative tasks. However, some locomotor evaluative tasks stress certain control variables more than others.

For example, the task of stepping while using a BWS system (7) is used to evaluate the ability of SCI-IMFL on the controlled variable of body weight support. The generation and absorption of energy at key points during the walking cycle can be evaluated using a task of modulation of the walking speed (8) or walking slope (9). The trajectory of the foot during the swing phase can be evaluated using a task of obstructed walking (10-12). In order to evaluate the maintenance of equilibrium, an indirect approach is the study of walking of SCI-IMFL either on the treadmill with and without parallel bars or with changes in overground walking aids (from walker to forearm crutches to single cane to no aids) (13). More direct approaches for the study of maintenance of equilibrium during walking have been used (14).

A MODEL OF RECOVERY OF WALKING

Functional recovery of walking, unlike the process of recovery, is defined as a cascade of transformations from one functional state to another rather than a continuous function. This cascade has been quantified in several scales including the modified functional walking scale (15-16). A combination of velocity of walking and knee extension strength was found to be the best discriminant between household and community walkers with a threshold of 0,42 $m*s^{-1}$ for the velocity variable.

Research and clinical observations of the recovery of walking in SCI-IMFL subjects have led us to propose a model of the process of recovery which is composed of at least two dimensions (control and capacity). The dimension of control was defined earlier and was shown in Fig. 2 (see section A model for the evaluation of walking behaviour). The notion of capacity on the other hand refers to the maximal value each variable can reach for each of the four sub-tasks. The control and capacity dimensions are found for each sub-task involved in the control of walking. For example, the possibility of changing the trajectory of the foot defines the control dimension, whereas the maximum height and length of swing measure the capacity on this sub-task.

This model of the process of recovery of the walking behaviour is illustrated in Fig. 3. The process follows a path of recovery characterised by periods of stability and changes

(17), of which four examples are shown. The upper leftmost example is of a severely disabled subject on a motorized treadmill with the support of a harness and parallel bars (or other fixed support for the upper limbs). Example 2 and 3 represents an individual walking on a horizontal, non-compliant, smooth surface using either a walker or cane for aid. Example 4 in the lower right part of the figure represents walking in a context that approaches a state of nearly full functional recovery of walking. The uneven surface represents the changes in surface height, inclination and compliance and the upright rectangles represent obstacles that may be encountered.

For the individuals in these examples, the demands of generation and absorption of mechanical energy, trajectory of foot, support of the body weight and balance of the upper body are increasing both in control and capacity. It should be noted that using BWS and a treadmill allows the evaluation of recovery of walking in subjects that would otherwise be considered unable to walk.

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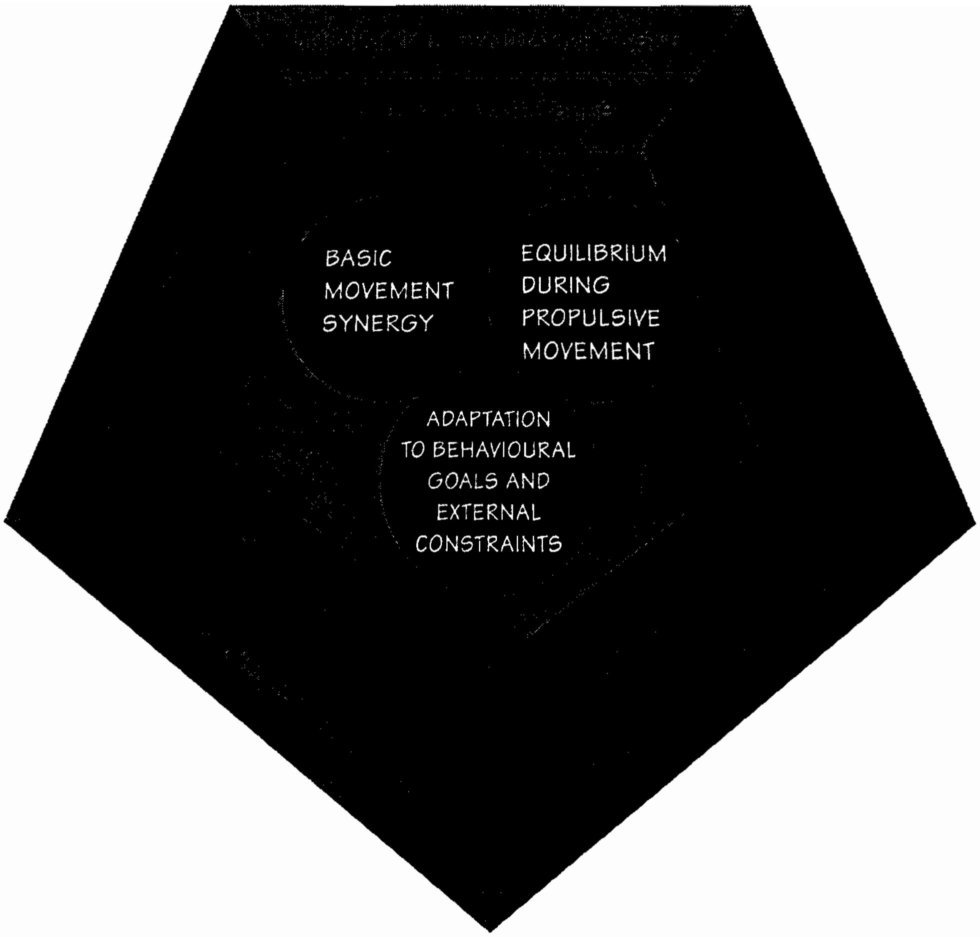

Figure 2. Model of the evaluation of the walking behaviour. The inner pentagon represents four controlled variables during walking whereas the outer pentagon shows examples of walking tasks that can be used to evaluate those control variables. The inner three circles represent the functional units of control used by the nervous system.

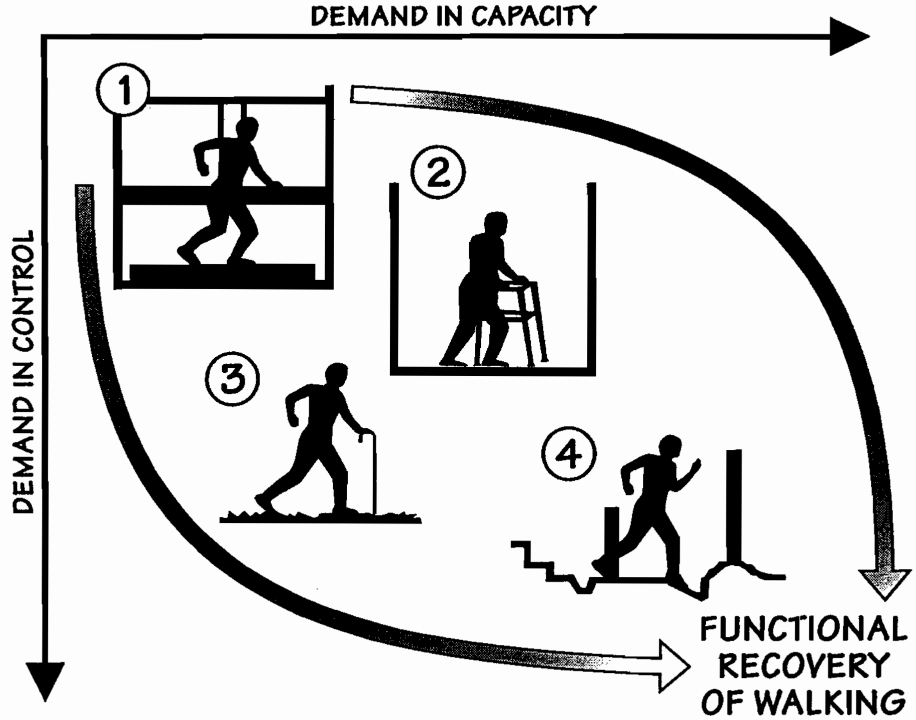

Figure 3. Model of the recovery process of walking. The four examples are explained in the text.

The regulation of toe height during the swing phase of human obstructed walking.

M. Ladouceur, B. J. McFadyen and H. Barbeau, Experimental Brain Research, submitted.

INTRODUCTION

Much is yet to be understood about the neural structures and mechanisms involved in the generation and control of human walking even though they have been studied since the late nineteenth century (see 1). However, it has been proposed that new experimental tasks requiring an adaptation in a spatially constrained environment, like walking on a ladder or stepping over an obstacle, can be used to investigate the mechanisms involved in the control of human walking (2).

Stepping over an obstacle when walking has been studied using a spectrum of obstacle configurations. The leading lower limb (3-7) as well as the trailing leg (8-9) in a bilateral obstacle configuration has been studied as well as unilateral obstacles placed at different points in the swing phase (6, 10, 11).

The relative contributions of increased hip elevation versus lower limb flexion for the modification of the foot trajectory for unilateral obstacles placed at the mid-swing location are still unknown. However, it can be speculated that the increased hip height may be absent since the vertical braking impulse of the contralateral lower limb, a factor involved in the elevation of the centre of mass, was not changed for such an obstacle placement (6). Furthermore, an increased flexion at all joints of the lower limb going over the obstacle was shown (11). This increased flexion has been attributed to a reorganisation of the energy generation toward a knee flexor strategy with the generation of a new power burst at foot-off which has been called an anticipatory locomotor adjustment (10, 11).

No study has investigated the modulation of toe height, hip elevation, and knee flexor power generation as a function of obstacle height.

The objective of this study was to investigate the control of the foot trajectory during walking by measuring:

1. The linearity and scaling of the relationship between changes in toe trajectory during the swing phase and obstacle height.

2. The modulation of the causes underlying the adjustment in the toe trajectory in relation to obstacle height.

MATERIALS AND METHODS

The experiments were conducted at the Département de Kinanthropologie, Université du Québec à Montréal and were approved by the ethics committee of that institution and the School of Physical and Occupational Therapy of McGill University. Five males (ages: 18-34 years; masses: 63.5-97.5 kg; heights: 1.75-1.85 m; lower limb lengths: 0.85-0.94 m), with normal or adjusted to normal visual acuity, walked along a tiled floor (8 m) in which an AMTI force platform was embedded.

The experimental tasks consisted of walking unobstructed (0 m obstacle) and with unilateral obstacles (width: 0.2 m; depth: 0.05 m) of six different heights (0.005, 0.030, 0.045, 0.060, 0.075 m and 10% lower limb length) placed in the pathway at the location of mid-swing that followed contact with the force platform. The obstacle was visible at all time. Participants were randomly presented with blocks of trials consisting of different obstacle heights. The participants were asked to walk at their preferred speed.

Markers were placed on the right side of the participants on the toe, head of the fifth metatarsal, heel, distally and proximally on the leg, distally and proximally on the thigh as well as on the acromial process. The position of the lateral malleolus, tibial plateau and greater trochanter were reconstructed from the markers placed on the leg and thigh by calculating the length and orientation of those segments following a calibration procedure. The kinematic patterns in the sagittal plane and temporal variables were calculated from coordinate marker data that were digitised for each field (i.e., 60 Hz) from video recordings by three cameras, then reconstructed and filtered (6 Hz, Butterworth, 4th order, low-pass, zero-lag filter) using a Peak Performance Analysis system. The force platform data was sampled (600 Hz) and subsequently filtered (25 Hz, Butterworth, 4th order, low-pass, zero-lag filter). The video recordings and force platform data were then synchronised using a signal that simultaneously marked the video recordings and the force plate data. The instant of foot contact and foot-off were established using the vertical ground reaction force when available, or the trajectory of the heel marker when forces were unavailable.

Both absolute segment and relative joint angles were calculated as indicated by Winter (13). Respective angular velocities and accelerations were derived and classical link segment calculations (14) were performed to estimate the net muscle moments of force at the ankle, knee and hip in the sagittal plane. Generalised muscle mechanical power was calculated as the product of the net muscle moment of force and the relative joint angular velocity. The net muscle moment of force provides an estimate of the contribution of the predominant muscle group about a joint. The muscle mechanical power can provide insight into the coordinated movement strategies underlying joint angular displacement by estimating the net instantaneous energetic states (generation or absorption of mechanical energy) of the muscles acting about the joint. Moment and power amplitudes were normalised to body mass to allow comparison across participants, and areas under selected power bursts were calculated using a trapezoidal technique. All time series plots were normalised to 100% of total stride-time beginning with heel contact on the force platform.

The spatio-temporal variables extracted were the walking speed, stride length, and the durations of stride, first double support, right single limb support, second double limb support, and right swing. The ratio of the swing to stride durations was also calculated. Furthermore, the apex position of the toes in the region of the forward edge (+/- 0.05 m) of the obstacle was found. The position of the greater trochanter and angle of the hip, knee and ankle joints at the time of this apex were also extracted. Four muscle power bursts occurring around the time of the initiation of swing, and associated with anticipatory locomotor adjustments (11), were analysed (labelling of the bursts can be found in Figures 1G to 1I): H3 is a hip flexor power generation burst, K3 consists of a knee extensor absorption power burst, A2 is an ankle plantarflexor generation burst, and K5 consists of a knee flexor power generation burst that is only reported during unobstructed walking.

A repeated measure analysis of variance was performed on all outcome variables between the unobstructed and obstructed experimental conditions. Significant findings ($p < 0.1$) were further analysed with a Dunnett's Multiple Comparison Test. Linear and non linear regression analyses were also carried out on all of the outcome variables to establish their relationship to obstacle height.

RESULTS

Figure 1 shows an example of the trajectory and time series plots recorded and calculated for three of the experimental conditions (unobstructed, 0.045 m and 10% obstacle heights). Figure 1A shows an example of the trajectory of the greater trochanter and toe markers. While the toes are raised in order to clear the obstacles, no noticeable changes can be seen in the trajectory of the greater trochanter. Figures 1B and 1C present, respectively, the vertical and horizontal time course of the toe marker and the time when the toes are over the forward edge is marked with a line for each experimental condition. It can be seen from these plots that the time course of the vertical component across conditions (Fig. 1B) of the toe trajectory is modified at around 60% of the gait cycle. However, no changes could be seen in the horizontal component of the trajectory (Fig. 1C). Figures 1D to 1F show examples of the time series plots of the angular kinematics for one cycle for three different experimental conditions. It can be seen that when going over the obstacle (marked with a line for each experimental condition) the lower limb is more flexed at the hip and knee with changes occurring at the ankle after the time at which the toe reached the apex. Figures 1G to 1I show an example of the time series plot of the muscle power for one cycle for three different experimental conditions for one participant. A small decrease in the K3 knee extensor muscle absorption burst as well as in the A2 ankle plantarflexor muscle power generation burst can be seen. More importantly, the emergence of a new knee flexor generation power burst (labelled K5) occurring at toe-off can be seen in Fig. 1H.

Toe height adaptations match increases in obstacle height

Toe height over the obstacle was increased in comparison to unobstructed walking for all obstacle heights ($F(5,20) = 52.16$, $p < 0.001$). It can be seen in Fig. 2A that there is a very large correlation between the changes in toe height and obstacle height. Furthermore, the slope of the linear regression of toe height in relation to obstacle height shows that the increase in toe height was well matched to the increase in obstacle height (slope = 1.09; 95% confidence interval (CI): 0.87—1.31). It is also interesting to note the presence of an offset (0.04; 95% CI: 0.03—0.05) representing the effect of the presence of an obstacle. The vertical position of the greater trochanter at the apex was not significantly different for any experimental condition ($F(5,20) = 1.28$, $p < 0.313$). Furthermore, the horizontal position of greater trochanter and toes in relation to the obstacle was not significantly different at the apex (greater trochanter: $F(5,20) = 2.07$, $p < 0.083$; toes: $F(5,20) = 1.21$, $p < 0.315$).

A significant increase in the proportion of swing can be seen for all obstacles higher than 0.005 m in comparison to the unobstructed experimental condition ($F(5,20) = 13.06$, $p < 0.001$). When the stride duration remains constant this increase in the proportion of swing can occur by either an increased swing time, a decreased stance duration or a combination of both factors. The duration of the right swing phase was significantly different from the unobstructed condition only for the 0.075m and 10% lower limb length obstacles ($F(5,20) = 6.83$, $p < 0.001$) a linear regression analysis showed a small relation between the duration of the right swing phase and obstacle height (not illustrated, Right swing phase duration=0.46*Obstacle height + 0.43; $R^2 = 0.13$; $p < 0.001$). Furthermore, this could be associated with a decrease in stance duration since the stride duration was not significantly changing in between the experimental conditions ($F(5,20) = 1.70$, $p = 0.181$).

Hip and knee joints are more flexed when going over an obstacle

Averages of the angle at the three lower limb joints for the different experimental conditions show that both the hip (for obstacles higher then 0.005 m) and knee (for all obstacles) angles are significantly increased (hip: $F(5,20) = 13.07$, $p < 0.001$; knee: $F(5,20) = 36.37$, $p < 0.001$) whereas the ankle angle was not significantly changed ($F(5,20) = 1.92$, $p = 0.136$). Figures 2B and 2C show, respectively, the large and very large correlation between the increase in hip and knee angles and obstacle height. It can also be noted that the slopes of both relationships are not different (95% CI for the slope, hip: 1.39—2.66, knee: 1.725—2.51). Furthermore, there was a small correlation between the changes in ankle angle and the obstacle height (not illustrated, Ankle angle changes = —0.75*Obstacle height + 0.05; $R^2 = 0.07$; $p < 0.02$).

The new knee flexor power burst at toe-off regulates toe height during the swing phase

The hip flexor generation muscle power burst (labelled H3), the knee extensor absorption muscle power burst (labelled K3), and the ankle plantarflexor generation muscle power burst (labelled A2) did not significantly change between the different experimental conditions (H3: $F(4,20) = 0.78$, $p = 0.557$; K3: $F(4,20) = 0.76$, $p = 0.565$; A2: $F(4,20) = 1.98$, $p = 0.146$) whereas the knee flexor generation power burst (labelled K5) showed a significant increase in comparison to unobstructed walking for obstacles higher then 0.030 m (K5: $F(4,20) = 28.22$, $p = 0.001$). Even though changes in the grouped averages of the H3 muscle power burst did not significantly change there was a negative regression slope for the small correlation between the area of H3 and toe height (not illustrated; H3 = —19.7* Toe height + 11.7; $R^2 = 0.08$; $p < 0.02$). Figure 3A and 3B show, respectively, the linear relationship between the emergence of the knee flexor power generation as a function of obstacle height (Fig. 3A, $R^2 = 0.54$; $p < 0.001$) and toe height at the apex (Fig. 3B, $R^2 = 0.60$; $p < 0.001$). Furthermore, the equations for the linear regression between the K5 muscle power burst and obstacle or toe heights greater show that the emergence of this new knee flexor generation muscle power is occurring for obstacles higher then 0.01 m or for toe heights higher then 0.05 m.

DISCUSSION

In our daily activities it is necessary to adapt our gait to environmental constraints. When faced with unilateral obstacles placed at the location of mid-swing the height of the toes is increased linearly in relation to obstacle height. Furthermore, the increase in toe height is equivalent to the increase in obstacle height. This increase in toe height seems to be completely dependent on the increased hip and knee flexions which are a product of the emergence of a knee flexor power generation burst at toe-off.

The linear relationship and scaling between toe height at the apex and obstacle height is in contrast to relationships reported in some studies for the leading lower limb for bilateral obstacles (3, 7). However, when obstacle heights are no higher then 10% of lower limb length the scaling reported by Patla & Rietdyk (7) is similar to the one found in this study. Furthermore, the increase in toe clearance due to the presence of an obstacle with no height has been found to be similar to a study of bilateral obstacles [95% CI: 0.021 to 0.032 m; (3)].

The lack of changes in the greater trochanter height is also different from the behaviour reported for the leading lower limb crossing bilateral obstacles (7). This discrepancy could be explained by the position of the obstacle. It was found (6) that when unilateral obstacles were placed in mid-swing the vertical braking impulse, one of the events associated with the changes in the greater trochanter height, did not change whereas for obstacles placed in late swing, a comparable position to the leading lower limb of a bilateral obstacle, an increase in the vertical braking impulse was shown (6).

As reported previously (10) there is an emergence of a new knee flexor power generation burst at toe-off. The present work shows a highly significant relationship between the emergence and the amplitude of this power burst with the elevation of the toes. In contrast to previous reports (10, 11), the reduction in both the hip flexor power generation (H3) as well as the knee extensor absorption power burst (K3) at toe off were not seen. One explanation may be the increased obstacle depth in this study [0.05 versus 0.02 m for (10) and (11)] since changes in depth from 0.067 to 0.268 m have been shown to change toe clearance as well as hip, knee and ankle angles in bilateral obstacles placed in the late swing phase (7).

Overall, it is interesting to note that even though the increased hip flexion when going over an obstacle is generated by intersegmental dynamics (10, 12) the increased toe height remains linearly related to K5. Such a result shows a very simple system of control in which the toe height is only regulated by the emergence and modulation of a knee flexor generation power burst at foot-off. It can be speculated that corticospinal pathways are most likely involved since they do not project to numerous muscles (15) and that the motor cortex has been shown to be involved in spatially constrained walking tasks like ladder or obstructed walking (16-20). Furthermore, it can be further speculated that it would not be necessary for the control signal arising from the motor cortex to project to the subsystem responsible for the generation of the basic locomotor pattern but directly to the knee flexor motorneuron pool.

In conclusion, it was shown that for mid-swing obstacles of relatively low to moderate heights, changes in foot trajectory are scaled to obstacle height, that mainly hip and knee flexion is utilised to modify toe height and that this adaptation is the result of the emergence and modulation of a knee flexor power generation. In the future it would be interesting to know when is limb elevation (hip hiking) involved in the modification of the toe trajectory.

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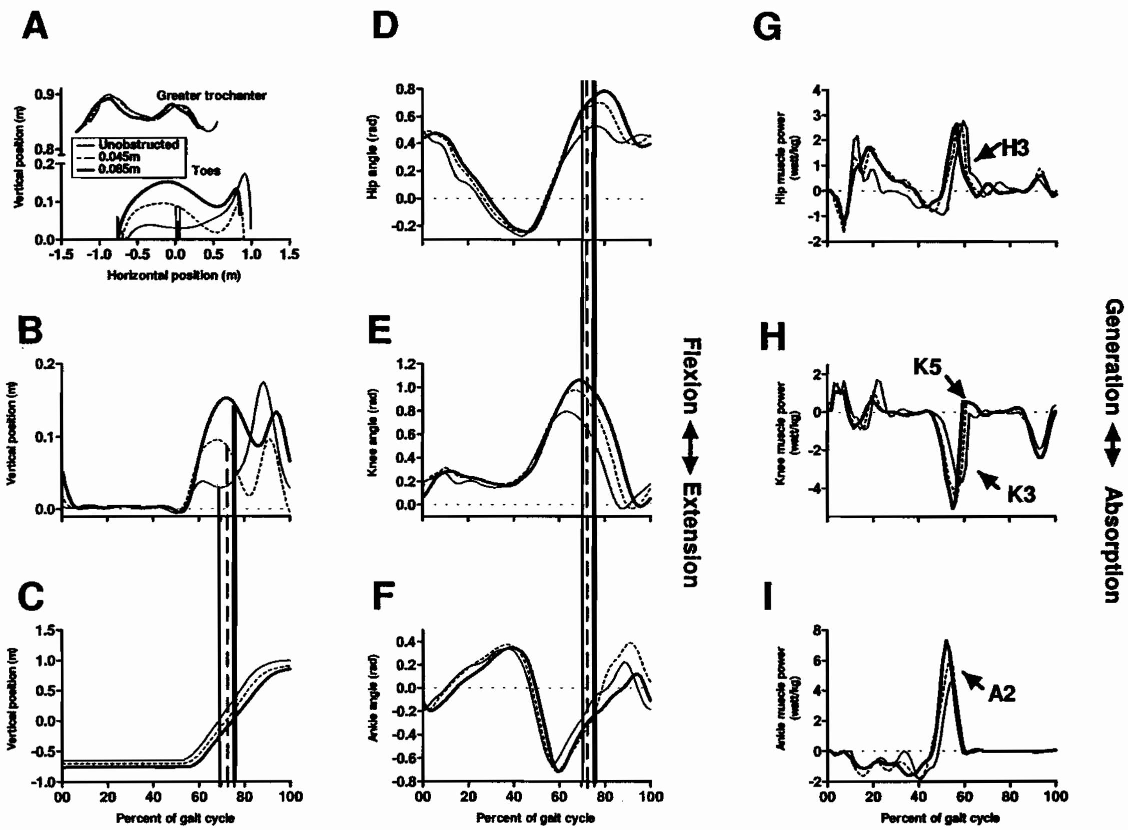


Figure 1. Example of the changes in the lower limb trajectory, angular kinematics and muscle power bursts during obstructed walking. kinematics for the unobstructed, 0.045 m and 0.085 m experimental conditions. Panel A shows an example of the changes in the trajectory of the greater trochanter and toe markers for one cycle of one participant (GV) for the three experimental conditions. The origin of the coordinate system was placed in reference to the forward edge of the obstacle. The 0.045m and 0.085m obstacles are represented, respectively, by clear and filled rectangles. Panels B and C represent, respectively, the time course of the vertical and horizontal position of the toe marker. Whereas panels D to F represent examples of the changes in the angular kinematics of the passing limb at the hip (Panel D), knee (Panel E) and ankle (Panel F). Panels G to I represent examples of the changes in the muscle power of the passing limb at the hip (Panel A), knee (Panel B) and ankle (Panel C) for three different experimental conditions. The experimental conditions are represented with different lines (thin line: unobstructed; dotted line: 0.045m; bold line: 0.085m)

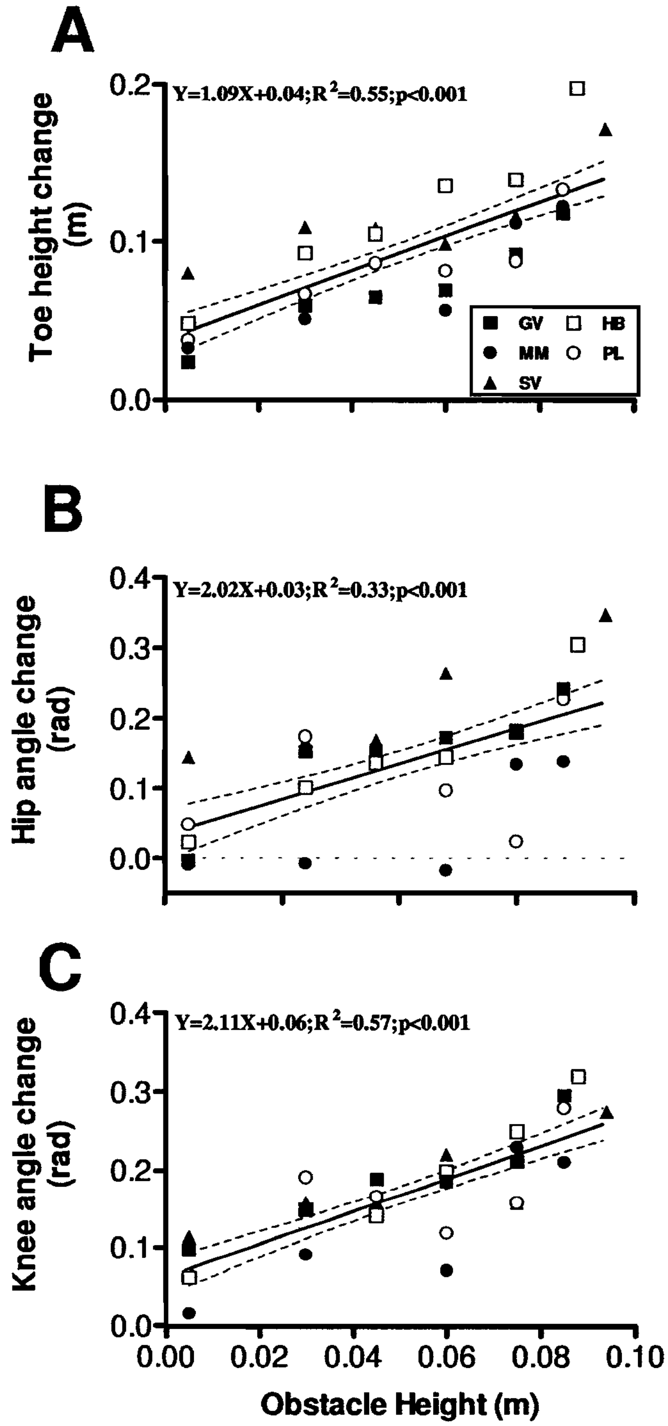
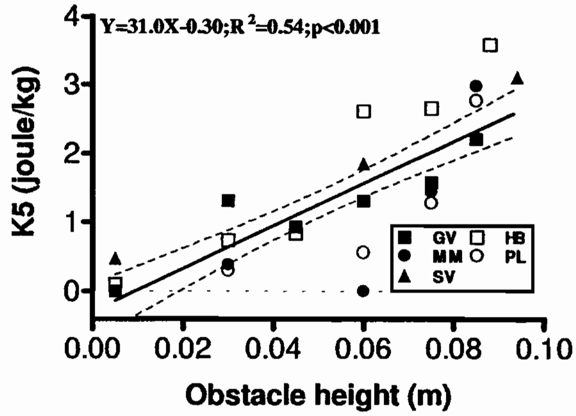


Figure 2. Changes in the kinematics during obstructed walking. Panel A presents the changes in toe height in relation to obstacle height whereas Panel B and C present, respectively, the changes in the angle of the hip and knee when the toes were at the apex in relation to obstacle height. For all panels the linear regression (solid line) and 95% confidence interval (dotted line) is also presented as well as the individual means of change for each participant.

A



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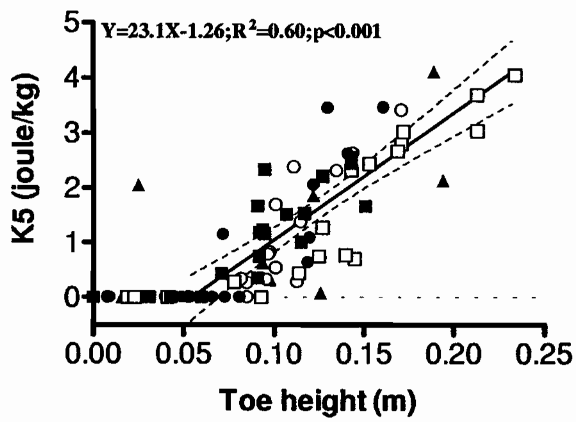

Figure 3. Changes in the muscle kinetics during obstructed walking. Panels A and B present the changes in the knee flexor power generation burst (K5) in relation to obstacle (Panel E) or toe height (Panel F). For all panels the linear regression (solid line) and 95% confidence interval (dotted line) is also presented as well as the individual means of change for each participant.

Kinematic adaptations of spinal cord injured persons during obstructed walking.

M. Ladouceur, H. Barbeau and B. J. McFadyen, Gait and Posture, submitted

INTRODUCTION

Great strides have been accomplished in the last years toward an improved treatment strategy after a spinal cord injury (1-5). These improvements have increased the proportion of injuries resulting in an incomplete motor function loss that provide one with the possibility of regaining some walking capability (6-9). Much still needs to be investigated about the control of human walking and even more so for the neural structures and mechanisms involved in walking by spinal cord injured persons. It was proposed that obstructed walking can be used to investigate the mechanisms involved in the control of the foot trajectory during walking (10). Obstructed walking has been shown to involve unilateral or bilateral strategies that are dependent on the position, height, width, deepness of the obstacle (11-20). For unilateral obstacles of small heights (<10% leg length) placed at the location of mid-swing it was shown that modulation of the toe trajectory was related to an increased knee and hip flexion when the limb was such that the toes were at the forward edge of the obstacle (21). Furthermore, because the adaptations in the locomotor pattern appeared in a very specific muscle group, namely the knee flexor group, it was speculated that these adaptations may involve mainly the corticospinal pathways. No study has ever characterised the anticipatory locomotor adjustments of spinal cord injured persons with an incomplete motor function loss (SCI-IMFL). The modification of the behaviour to clear obstacles in SCI-IMFL may also show the importance of the interaction between the supraspinal and spinal neural structures in the control of the anticipatory locomotor adjustment.

The objectives of this study were:

1. To characterise the kinematic anticipatory locomotor adjustments used by SCI-IMFL by measuring the changes in greater trochanter height, hip, knee, and ankle angle when the toes were at the forward edge of the obstacle.

2. To compare the anticipatory locomotor adjustments of SCI-IMFL and able-bodied participants.

MATERIALS AND METHODS

The experiments were conducted at the School of Physical and Occupational Therapy, McGill University and Département de Kinanthropologie, Université du Québec à Montréal. The study was approved by the ethics committee of both institutions and informed consent was obtained from all participants prior to the evaluation session.

Six SCI-IMFL, with more than one year post-injury and an impairment of C or D on the American Spinal Injury Association scale (22), and five able-bodied participants walked along either a tiled or hardwood floor (more then 5 m). The visual acuity of the participants were normal or corrected to normal by wearing glasses or contact lenses. The characteristics of the SCI-IMFL are reported in Table I.

The experimental tasks consisted of walking unobstructed (0 m) and with obstacles (width: 0.2 m; depth: 0.05 m) of two different heights (0.005 m and 0.030 m) placed in the pathway at the location of mid-swing of the stride taken in the middle of the walkway. Participants were presented with blocks of trials consisting of different obstacle heights in a randomised order. Three successful trials were collected for each obstacle height. The participants were asked to walk at their preferred walking speed. SCI-IMFL used their usual ambulatory assistive device.

Markers were placed on the most affected side of SCI-IMFL and on the right side for able-bodied participants. For SCI-IMFL, the location of the markers were the fifth metatarsal, heel, lateral malleolus, knee joint, greater trochanter and acromional process. For the able-bodied participants markers were placed on the toe, head of the fifth metatarsal, heel, distally and proximally on the shank, distally and proximally on the thigh as well as on the acromional process. The position of the lateral malleolus, knee joint and greater trochanter were reconstructed from the markers placed on the shank and thigh by calculating the length and orientation of those segments following a calibration procedure. The kinematic patterns in the sagittal plane and temporal variables were calculated from coordinate marker data that were digitised for each field (i.e., 60 Hz) from video recordings by three cameras, then reconstructed and filtered (6 Hz, 4^{th} order, Butterworth, low-pass, zero-lag filter) using a Peak Performance Analysis system. The moment of foot contact and foot-off were established using the trajectories of the foot markers. The relative joint angles were calculated as indicated by Winter (23). The resulting time-series plots were normalized for the duration of the stance and swing phases in order to be comparable to the relative duration found during unobstructed walking by young adult subjects (respectively 60 and 40%).

The spatio-temporal variables extracted were the walking speed, stride and swing duration. The greater trochanter height, hip, knee, and ankle angles when the foot was over the forward edge of the obstacle were also extracted. An analysis of variance (repeated on the obstacle factor) was performed on all outcome variables. Significant findings ($p < 0.1$) were further analysed with a Bonferroni Post-hoc test. All participants could accomplish all experimental tasks with the exception of one SCI-IMFL (LS) who was not capable of walking over the 0.030 m obstacle. The data from this participant were excluded from the statistical analyses.

RESULTS

In comparison with able-bodied participants, SCI-IMFL had a slower walking speed (range: 0.15-0.44 vs 1.17-1.41 $m*s^{-1}$), a longer stride duration (range: 2.10-4.54 vs 1.16-1.31 s) and a greater stance to stride ratio (range: 0.70-0.84 vs 0.62-0.66).

SCI-IMFL uses a spectrum of kinematic anticipatory locomotor adjustments

Three examples of the results are shown in figure 1. These examples represent one able-bodied participant (AP) and two representative examples (FS and LS) of the spectrum of adaptations found in the group of SCI-IMFL.

The adaptation to the obstacle by AP can be seen by the changes in the trajectory of the fifth metatarsal which increases in height during the obstructed conditions (dashed and thick lines). This modification is concomitant with a change in the trajectory of the greater trochanter. It should be noted, however, that modification of the trajectory of the greater trochanter is similar for the 0.005 and 0.030 m obstacles. Changes in the angular excursions of the lower limb joints occur for both obstacle conditions with an increased knee and hip flexion when passing the obstacle. The timing of the changes in the angular excursion was different for the two articulations. The changes in the knee angular excursion always preceding changes at the hip joint. It is worth noting that the fifth metatarsal trajectory offset (0.02 m) at toe off for the 0.030 m obstacle was a result of the increased hip and knee flexion occurring at the start of the swing phase.

Participant FS raised the foot, as seen by the increased height of the fifth metatarsal, in order to clear the obstacles. This adaptation of the fifth metatarsal trajectory is related to an increased knee flexion for both obstacle conditions with no changes in the hip angular excursion and a slight decrease in the dorsiflexion of the ankle. There were changes in the greater trochanter height only for the 0.03 m obstacle. On the opposite end of the spectrum of adaptations, LS was capable of clearing only the 0.005 m obstacle. LS did not adapt the trajectory of the fifth metatarsal during the obstructed condition showing that the unobstructed foot trajectory was sufficient to clear such a low height obstacle. However, there were changes in the locomotor pattern with an elevation of the greater trochanter and, as seen in Fig. 1, a decrease in knee flexion at the initiation of the swing phase. Therefore, the adaptation of the walking behaviour is different between the two SCI-IMFL with one person (LS) adapting the behaviour by mostly increasing the height of the greater trochanter whereas the adaptation to the obstacle by the second SCI-IMFL (FS) is done by increasing knee flexion.

Fig. 2A compares the kinematic anticipatory locomotor adjustments of SCI-IMFL and able-bodied participants. None of the SCI-IMFL had a strategy located within the bounds of the 95% confidence interval of able-bodied participants. As stated previously, only one SCI-IMFL (stars in Fig. 2A, LS) could not accomplish both obstacle conditions. Of the remaining five SCI-IMFL, one participant (open squares in Fig. 2A, RM) decreased his knee flexion angle with the higher obstacle and increased the greater trochanter height. Three of the four SCI-IMFL that increased their knee flexion decreased their greater trochanter height. Only one SCI-IMFL (diamonds in Fig. 2A, FS) showed the trend found in able-bodied participants by increasing the height of the greater trochanter and knee flexion angle when going over an obstacle of greater height.

SCI-IMFL flexed their lower limb differently when going over an obstacle

Fig. 2B shows the change in greater trochanter height for the two obstacle conditions for both the SCI-IMFL and able-bodied groups. It can be seen with the 95% confidence interval that height of the greater trochanter is not increased when going over obstacles. Furthermore, the effect of obstacle heights ($F(1,8) = 0.002$, $p = 0.964$) and groups ($F(1,8) = 0.008$, $p = 0.931$) showed no statistically significant differences. These results show that both groups, for these locomotor tasks, adapt their walking behaviour by increasing the flexion of the lower limb mainly at the knee.

Fig. 2C through Fig. 2E show the changes in the lower limb flexion at the forward edge of the obstacle. Changes in the hip angle are reported in Fig. 2C for both groups and both obstacle heights. It can be seen that values for SCI-IMFL were smaller then for able-bodied participants ($F(1,8) = 12.92$, $p = 0.007$). Furthermore, an interaction effect is present ($F(1,8) = 28.98$, $p < 0.001$) as seen by the increased hip flexion for the 0.030 m obstacle in able-bodied participants in comparison to a decreased hip flexion for SCI-IMFL. However, the changes in the hip flexion with an increased obstacle height was not significantly different ($F(1,8) = 0.526$, $p = 0.489$). Fig. 2D shows the changes in the knee flexion angle with the different obstacle conditions and experimental groups. It can be seen that in both groups there was an increased knee flexion for the higher obstacle ($F(1,8) = 10.08$, $p = 0.013$). The difference in results between the two groups was not statistically different ($F(1,8) = 0.743$, $p = 0.414$). It should be noted that the knee flexion means for both obstacle heights are higher for the group of SCI-IMFL with a substantial variability as seen by the 95% confidence interval. Changes in ankle flexion are reported in Fig. 2E for all groups and obstacle heights. SCI-IMFL were more plantarflexed in comparison to able-bodied participants ($F(1,8) = 4.328$, $p = 0.071$) but ankle angle did not change with increasing obstacle height ($F(1,8) = 2.803$, $p = 0.133$).

An exploratory investigation of the effect of the ambulatory assistive device on the kinematic anticipatory locomotor adjustment for the 0.005 m obstacle is reported in Fig. 2F. Even though the 95% confidence intervals are overlapping participants using a walker increased their greater trochanter in order to go over the obstacle. This is in contrary to the strategy used by SCI-IMFL using forearm crutches or able-bodied participants.

DISCUSSION

Rehabilitation strategies designed to improve the recuperation of the walking behaviour also need to be evaluated for the adaptations to environmental constraints if they are to be of any functional significance. SCI-IMFL that are faced with unilateral obstacles placed at the location of mid-swing do adapt their walking pattern. This adaptation is different from the one seen in able-bodied participants. There were two major differences. First, SCI-IMFL only increased their knee flexion during obstructed walking in comparison with the increase in flexion at the hip and knee joint by able-bodied participants. Secondly, SCI-IMFL seem to increase their greater trochanter height more than able-bodied participants for obstacles of low heights. One striking result is that none of the SCI-IMFL increased their hip flexion when clearing the obstacle. In able-bodied participants, that increased hip flexion when going over an obstacle is the resultant of the emergence and modulation of a knee flexor power generation at toe-off acting through intersegmental dynamics (16,19,21). Such dynamics are speed-dependent (24) and since the group of SCI-IMFL did not walk as fast as able-bodied participants it could be speculated that these dynamic interactions were minimal. Although there was a wide spectrum of adaptation strategies, the SCI-IMFL favoured an increase in knee flexion. Such a strategy is different from the one favoured by below-knee amputees who primarily increase their hip flexion power

(25). Perhaps SCI-IMFL do not increase their hip flexion because of weakness in the hip flexor muscles or that such a strategy was not used because it would bring the foot closer to the obstacle increasing the probability of touching it.

The result of the present study provides some evidence that the selection of an appropriate response made by the participants was based on multiple factors like the ambulatory assistive device used and walking speed. It can also be speculated that strength of the different muscle groups and/or level of spasticity would also influence the strategy that will be used for the adaptation to obstacle clearance. This underlies the importance of the initial conditions as well as the limits and dynamics of the system in the choice of the strategy used during obstructed walking. Evidence that anticipatory locomotor adjustments are visually triggered central commands (26), in opposition to a visually triggered reflex responses as suggested by some authors (27), are provided by the numerous factors that can influence the selection of the strategy used during obstructed walking as shown by the variability in the obstructed walking behaviour of our participants.

Elevation of the greater trochanter was shown either not to be used, or only a minor contributing factor if used at all, in the kinematic strategy of adaptation to obstacles both by able-bodied participants and SCI-IMFL. However, it was shown that clearance off a 0.005 m obstacle for some SCI-IMFL involved an increase in greater trochanter height that may be dependent on the ambulatory assistive device used. This increase in the height of the greater trochanter can be attributed to actions of the upper extremities and contralateral leg. It would be worthwhile to further investigate the factors involved in the selection of a strategy involving the raising of the greater trochanter height in order to clear obstacles.

This study shows that SCI-IMFL can adapt their walking behaviour, although using a wide range of strategies, to obstacles placed in their walking path. New evaluative tasks, such as obstructed walking, can be used to investigate the control of the walking behaviour in order to understand the changes occurring after an impairment, as well as developed to be used as an outcome measure in clinical trials of different treatment strategies. Further studies with an increased sample size using other ambulatory assistive devices as well as the recording of the kinetics changes are needed to understand better the modifications occurring in the walking behaviour and its control after a spinal cord injury.

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Table I. Relevant characteristics of the spinal cord injured participants. ASIA: Impairment scale of the American Spinal Injury Association; AAD: Ambulatory Assistive Device; W: Walker; K: Forearm crutche

					Walking Speed ($m*s^{-1}$)		
SCI-IMFL	Neurological Level of lesion	ASIA	AAD	Passing leg	0m	0.005m	0.030m
LS	T9-T12	C	W	Left	0.15(0.02)	0.15(0.02)	N/A
RP	C5-C6	C	W	Left	0.17(0.01)	0.18(0.02)	0.23(0.01)
SM	C5-C6	C	2K	Left	0.21(0.02)	0.21(0.02)	0.17(0.02)
FS	T9	D	2K	Right	0.31(0.04)	0.31(0.02)	0.30(0.01)
AN	Undefined	D	2K	Left	0.36(0.02)	0.35(0.02)	0.34(0.02)
RM	C6	D	2K	Left	0.44(0.03)	0.36(0.02)	0.34(0.01)

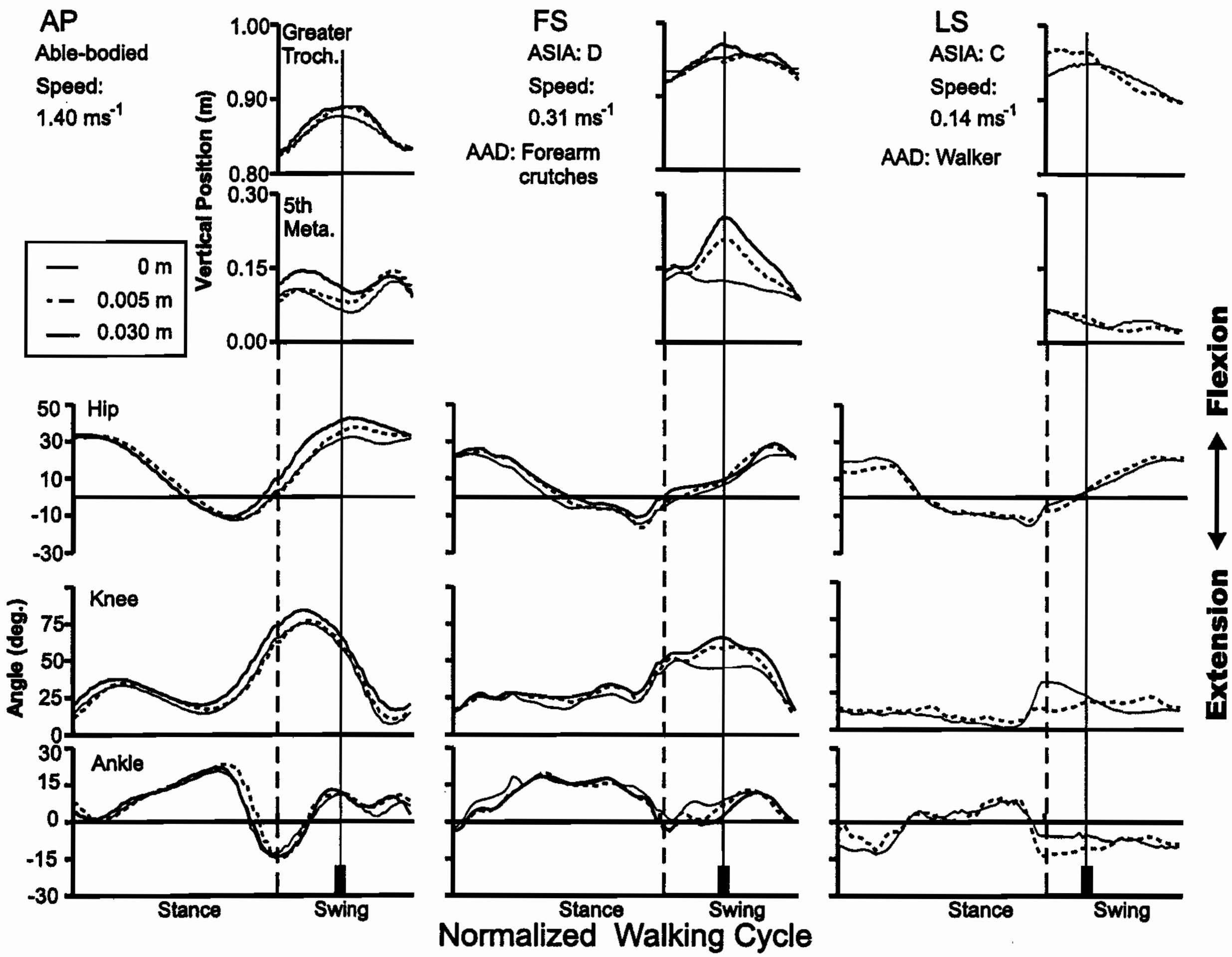

Figure 1. Normalised kinematics during obstructed walking. This figure represents the normalised kinematics of the greater trochanter and fifth metatarsal as well as the angular excursions of the hip, knee, and ankle of the passing leg for the three obstacle conditions (0 m: thin line, 0.005 m: dashed line, 0.030 m: thick line) . This figure shows one able-bodied participant (AP) and two representative examples of the spectrum of adaptations found in spinal cord injured persons with an incomplete motor function loss. The walking speeds and type of ambulatory assistive devices (AAD) are indicated for each example. The time when the toes are over the forward edge of the obstacle is represented by a solid rectangle.

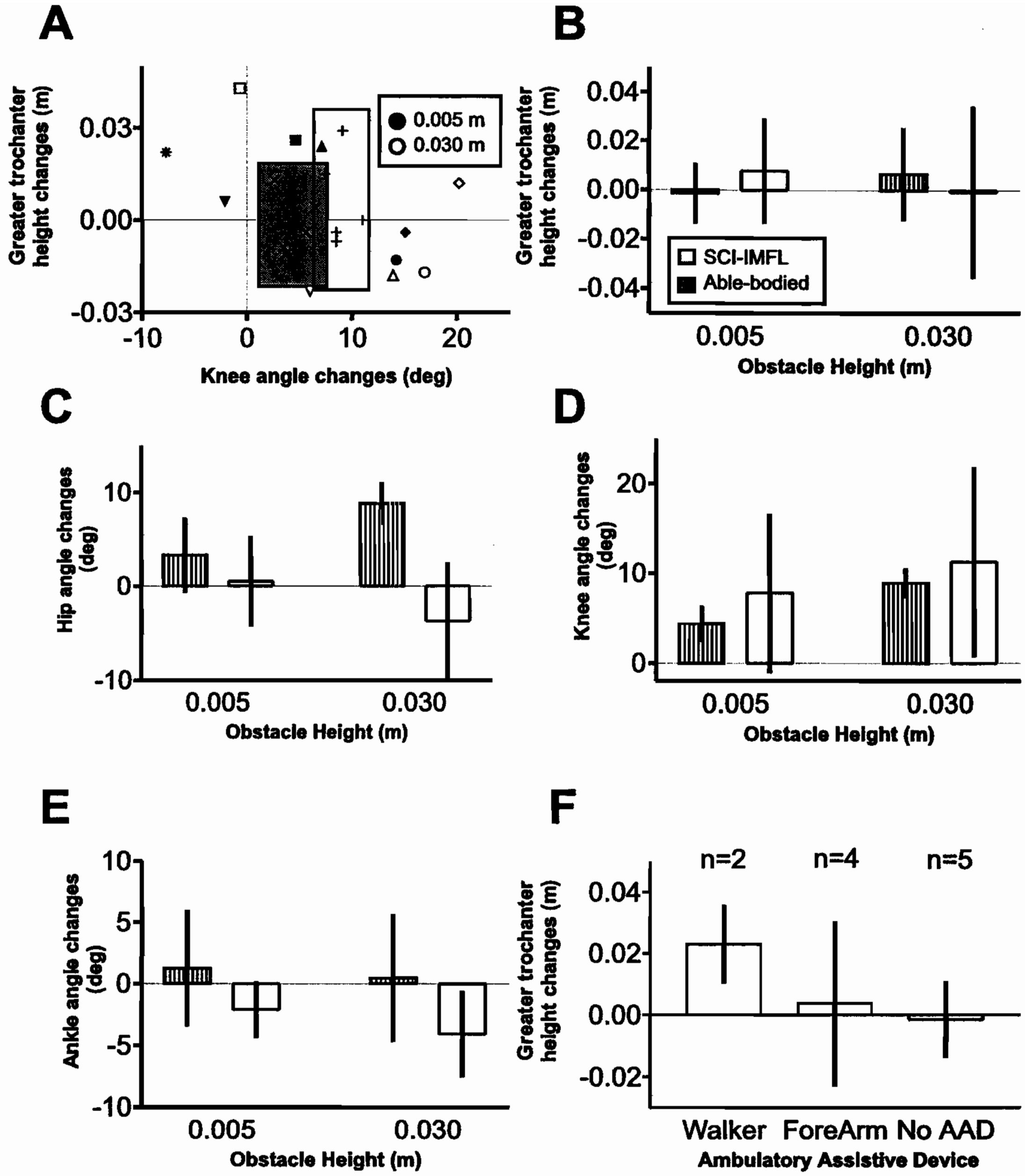

Figure 2. Kinematic anticipatory locomotor adjustments. Panel A shows the changes in greater trochanter height as a function of knee angle changes with two obstacle heights for able-bodied participants (0.005 m: x, 0.030 m: cross; 95% confidence interval 0.005 m: filled rectangle, 0.030 m: open rectangle) and spinal cord injured persons with an incomplete motor function loss (LS: star, RP: upward triangle, SM: downward triangle, FS: diamond, AN: circle, RM: square; 0.005 m: filled, 0.030 m: open). Panels B through E represent respectively the mean and 95% confidence interval of the changes in the greater trochanter height (B), hip angle (C), knee angle (D) and ankle angle (E) for spinal cord injured persons with an incomplete motor function loss (clear bar) and able-bodied participants (dotted bar). Panel F represents the mean and 95% confidence interval of the increase in greater trochanter height for the 0.005 m obstacle as a function of ambulatory assistive device used.

Summary and conclusions

Functional recovery of walking is a goal for many SCI-IMFL, as well as for the interventions of many health professionals involved in their rehabilitation. Evidence presented in the present thesis on the effect of FES assisted walking gave a rationale for the use of such an orthosis in the rehabilitation of walking for this population, even for SCI-IMFL that were in a chronic stage. It was shown that an orthotic effect was present for SCI-IMFL that walk slower than 0.6 $m*s^{-1}$. Remarkably, the absolute therapeutic effect of using FES assisted walking was greatest for faster SCI-IMFL walkers even though the orthotic effect was, at best, minimal. Furthermore, it was shown that FES assisted walking (orthotic effect) changed the angular kinematic pattern of the stimulated lower limb by dorsiflexing the ankle during mid-swing, reducing the plantarflexion angle at foot contact and increasing the hip angular excursion. As seen in the maximal overground walking evaluations, FES assisted walking did not immediately increase walking speed, stride length, stride frequency or reduce the physiological cost of walking. However, all those parameters improved in a functionally important manner over time (longitudinal study). From these results, it can be speculated that FES could not only be used as an orthosis, but also as a therapeutic modality.

In the second section of the thesis, evidence was shown that able-bodied participants, faced with an obstacle of low to moderate height, have an increase in the height of their toe trajectory that matches the height of the obstacle. Furthermore, it was shown that this adaptation of the toe trajectory was the product of an increased lower limb flexion that was caused by the sole generation and modulation of a new knee flexor power generation burst at the initiation of the swing phase. It was also shown that SCI-IMFL, faced with obstacles of low height, used a different strategy in comparison with able-bodied participants. Even though the strategy was different, SCI-IMFL still chose to primarily use an increased knee flexion to go over the obstacle. It was speculated that this difference in the strategy to clear the obstacle was seen because SCI-IMFL could not generate and exploit the intersegmental dynamics to increase their hip flexion because of their reduced walking speed.

Future studies should investigate the factors associated with the therapeutic effect. On the behavioural aspect, factors associated with limitations in maximal walking speeds should be investigated. In a case report, it was shown that training with FES assisted walking was associated with an increased maximal frequency of stepping (1). Changes in the nervous and musculo-skeletal system caused by long term FES assisted walking should also be studied. Preliminary evidence show a modification of the ankle plantarflexor stiffness dynamics associated with long term use of FES assisted walking (2). In that study (2) it was shown that in some SCI-IMFL using FES assisted walking there was a decrease in the reflex and passive stiffness gains as well as an increase in the intrinsic stiffness gain. This long term modification of the stretch reflex gain could be the product of a repetitive conditioning from the action of a peroneal nerve group II inhibitory pathway projecting to the ankle plantarflexor motorneuron pool (3-4). Another question that needs to be answered is the apparent discrepancy between the modifications in the spatio-temporal parameters and changes in the kinematics of the lower limb. This question arises from the fact that changes in the kinematics of walking are present earlier than any changes in the walking speed. This evidence is the basis for the speculation that those changes in the angular pattern become a permissive situation that needs to be incorporated into the walking pattern with training. Another argument for this speculation can be found in the trajectory of the foot reported in Fig. 1C of the second manuscript on FES assisted walking. When the participant uses FES assisted walking, the foot is brought back before it contacts the ground. This could be explained by a lack of control of the intersegmental dynamics contributing to the late part of swing phase (5). A second explanation is that the dynamic projection of the centre of gravity may not follow the improvement in the forward advancement of the foot resulting in a backward force that would cause a disequilibrium. Further investigations in the control of the foot trajectory and projection of the centre of mass during swing could be use to resolve this question.

Obstructed walking is a relatively new experimental protocol and much still needs to be investigated in the range of behaviours that are used in this locomotor adaptation as well as in the neural structures implicated in this task. Evidence of the variability in the kinematic strategies used to go over obstacles presented in this thesis is coherent with the hypothesis that anticipatory locomotor adjustments are visually triggered central commands (6) in opposition to the suggestion by some authors (7) that anticipatory locomotor adjustments are visually triggered reflex responses. As such it would be very interesting to investigate the role of the neural structures involved in the initiation of movement in the selection and implementation of the adaptations during obstructed walking. Furthermore, it is becoming imperative, in light of the differences in strategy used by SCI-IMFL and able-bodied participants, to test the effect of walking speed on the strategy used to clear an obstacle during walking.

This dissertation is pertinent to the field of rehabilitation science. Fig. 4 presents a schematic view of the large scope spanned by the rehabilitation science field. Briefly, this figure reflects the fact that rehabilitation methods used in clinical practice should be based on scientific evidence of effectiveness and efficacy. One aspect of rehabilitation science (clinical trial axis in Fig. 4) is the gathering of different levels of evidence used in the design of clinical practice guidelines (8) by clinical trials with different objectives (9). It should be noted that, as seen in this schematic drawing with the shaded polygon titled Motor Control, therapeutic modalities tested in clinical trials should be based on evidence gathered in neuroscience (Motor Control). Another aspect of the rehabilitation sciences is shown in Fig. 4 with the axis concerning the development of evaluative tools for the different levels of clinical trials. As shown in Fig. 4 the development of new evaluation tools should also be based on basic and applied neurophysiological (Motor Control) evidence.

This thesis reports work that has been done in both areas (development of therapeutic modalities and evaluative tools) of rehabilitation science. Our studies on FES assisted walking in SCI-IMFL contributed to knowledge in rehabilitation sciences by providing the first long-term clinical trial of the changes in walking behaviour concomitant with the use of FES assisted walking. These studies will also be useful in the planning of phase III clinical trials on the efficacy of such a therapeutic modality by providing a basis for the selection of the duration of the treatment and the selection of outcome variables. Whereas the evidence on obstructed walking will help in designing a new experimental task that could be useful as an outcome measure for the gathering of evidence in phase I and II clinical trials with limited number of participants, as presented in Fig. 4.

In conclusion, recovery of walking is becoming possible for an increasing number of SCI persons. New modalities of treatment are becoming available for this population but most still need to be evaluated for their efficacy. In this thesis, I provided evidence that FES assisted walking could be used as a therapeutic modality even in SCI-IMFL that were in a chronic stage. Furthermore, I also showed that new experimental locomotor tasks are needed to improve our understanding of the changes that occur on the locomotor behaviour with any therapeutic modalities and reported on the first steps in the development of such an experimental task in the second part of this thesis. It is then possible to envision that by using more efficiently new as well as readily available therapeutic modalities, care and recovery of SCI-IMFL could be improved.

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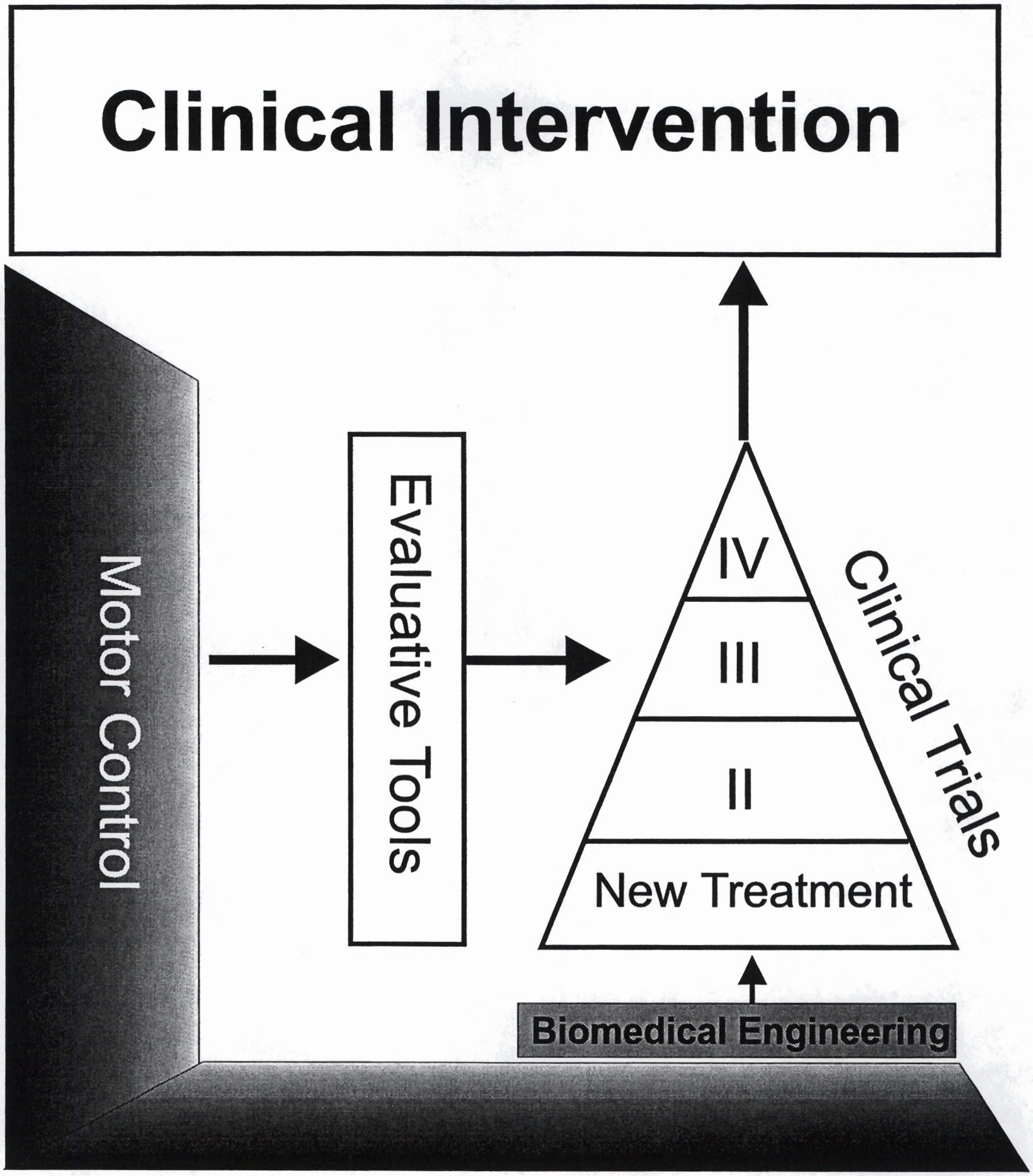


Fig. 4. Schematic view of the scope of rehabilitation science. Explanations are given in the text.

Appendix I. Definition of certain terms

Spinal cord injured persons with an incomplete motor function loss:

A Spinal Cord Injury (SCI) consist of a damage to the spinal cord that results in a loss of function such as mobility or feeling. The neurological level of the injury is determined by the most caudal segment with normal function. The spinal cord injury is classified as a complete injury (ASIA impairment of A) when there is no sensation and motor function at the S4-S5 dermatome level. The other four classes (ASIA B, C, D, and E) are based on the amount of residual motor function below the neurological level of the injury with a classification of B when there is no motor function, a classification of C when the majority of the key muscle cannot produce an active movement against gravity, a classification of D when the majority of the key muscle can produce an active movement against gravity, and a classification of E when all key muscles can produce movement against full resistance. Classes C, D, and E comprise the group of spinal cord injured persons with an incomplete motor function loss. The form to establish the neurological classification of the spinal cord injury can be found at the end of this appendix.

Muscle power bursts during walking

Muscle power is calculated by multiplying the calculated joint muscle moment by the angular velocity at the joint. When the product is positive, the summation of the muscle action around the joint is said to generate angular power, whereas when the product is negative the muscles acting around the joint are absorbing angular power. Walking is a complex behaviour in which muscles generate and absorb energy at different points in the gait cycle. More specifically, four muscle power bursts acting around the joints of the lower limb were investigated in this these.

The A2 muscle power burst is a generation of energy by the ankle plantarflexors to accelerate the centre of mass forward and occurs during the late stance phase (push-off). The K3 muscle power burst is an absorption of energy by the knee extensors occurring around the time of toe-off. The H3 muscle power burst is a generation of energy by the hip flexors occurring also in the pre- and initial swing phase. The purpose of this muscle power burst is to initiate the swing phase of the lower limb. The K5 muscle power burst is a generation of energy by the knee flexors occurring around the moment of toe-off. This muscle burst has been observed when walking over obstacles.

STANDARD NEUROLOGICAL CLASSIFICATION OF SPINAL CORD INJURY

MOTOR

KEY MUSCLES

	R	L	
C2			
C3			
C4			
C5	☐	☐	Elbow Flexors
C6	☐	☐	Wrist Extensors
C7	☐	☐	Elbow Extensors
C8	☐	☐	Finger Flexors (distal phalanx of middle finger)
T1	☐	☐	Finger Abductors (little finger)
T2			
T3			
T4			
T5			
T6			
T7			
T8			
T9			
T10			
T11			
T12			
L1			
L2	☐	☐	Hip Flexors
L3	☐	☐	Knee Extensors
L4	☐	☐	Ankle Dorsiflexors
L5	☐	☐	Long Toe Extensors
S1	☐	☐	Ankle Plantar Flexors
S2			
S3			
S4-5			☐ Voluntary anal contradction (Yes/No)

0 = total paralysis
1 = palpable or visible contraction
2 = active movement, gravity elliminated
3 = active movement, against gravity
4 = active movement, against some resistance
5 = active movement, against full resistance
NT = not testable

TOTALS ☐ + ☐ = ☐ **MOTOR SCORE**

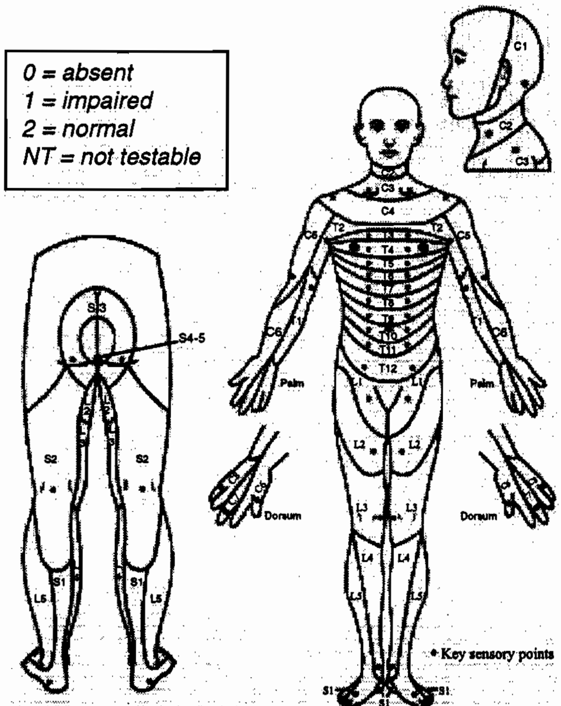
(MAXIMUM) (50) (50) (100)

SENSORY

KEY SENSORY POINTS

	LIGHT TOUCH R	LIGHT TOUCH L	PIN PRICK R	PIN PRICK L
C2	☐	☐	☐	☐
C3	☐	☐	☐	☐
C4	☐	☐	☐	☐
C5	☐	☐	☐	☐
C6	☐	☐	☐	☐
C7	☐	☐	☐	☐
C8	☐	☐	☐	☐
T1	☐	☐	☐	☐
T2	☐	☐	☐	☐
T3	☐	☐	☐	☐
T4	☐	☐	☐	☐
T5	☐	☐	☐	☐
T6	☐	☐	☐	☐
T7	☐	☐	☐	☐
T8	☐	☐	☐	☐
T9	☐	☐	☐	☐
T10	☐	☐	☐	☐
T11	☐	☐	☐	☐
T12	☐	☐	☐	☐
L1	☐	☐	☐	☐
L2	☐	☐	☐	☐
L3	☐	☐	☐	☐
L4	☐	☐	☐	☐
L5	☐	☐	☐	☐
S1	☐	☐	☐	☐
S2	☐	☐	☐	☐
S3	☐	☐	☐	☐
S4-5	☐	☐	☐	☐

0 = absent
1 = impaired
2 = normal
NT = not testable



☐ Any anal sensation (Yes/No)

TOTALS ☐ + ☐   ☐ + ☐

(MAXIMUM) (56) (56) (56) (56)

☐ **PIN PRICK SCORE** (max: 112)

☐ **LIGHT TOUCH SCORE** (max: 112)

NEUROLGICAL LEVEL *the most caudal segment with normal function*		R	L	**COMPLETE OR INCOMPLETE** ☐ *Incomplete = Any sensory or motor function in S4-S5*	**ZONE OF PARTIAL PRESERVATION** *Partially innervated segments*		R	L
	SENSORY	☐	☐			SENSORY	☐	☐
	MOTOR	☐	☐	**ASIA IMPAIRMENT SCALE** ☐		MOTOR	☐	☐

Version 4p
GHC 1996

Functional Independence Measure (FIM)

LEVELS		
	7 Complete Independence (Timely, Safely) 6 Modified Independence (Device)	No Helper
	Modified Dependence 5 Supervision 4 Minimal Assist (Subject = 75%+) 3 Moderate Assist (Subject = 50%+) **Complete Dependence** 2 Maximal Assist (Subject = 25%+) 1 Total Assist (Subject = 0%+)	Helper

	ADMIT	DISCH
Self Care		
A. Eating	☐	☐
B. Grooming	☐	☐
C. Bathing	☐	☐
D. Dressing-Upper Body	☐	☐
E. Dressing- Lower Body	☐	☐
F. Toileting	☐	☐
Sphincter Control		
G. Bladder Management	☐	☐
H. Bowel Management	☐	☐
Mobility		
Transfer:		
I. Bed, Chair, Wheelchair	☐	☐
J. Toilet	☐	☐
K. Tub, Shower	☐	☐
Locomotion	W ☐ C ☐	W ☐ C ☐
L. Walk/Wheelchair	☐	☐
M. Stairs	☐	☐
Communication	A ☐ V ☐ V ☐ N ☐	A ☐ V ☐ V ☐ N ☐
N. Comprehension	☐	☐
O. Expression	☐	☐
Social Cognition		
P. Social Interaction	☐	☐
Q. Problem Solving	☐	☐
R. Memory	☐	☐
Total FIM	☐	☐

Leave no blanks; enter 1 if patient not testable due to risk.

ASIA IMPAIRMENT SCALE

☐ **A = Complete:** No motor or sensory function is preserved in the sacral segments S4-S5.

☐ **B = Incomplete:** Sensory but not motor function is preserved below the neurological level and includes the sacral segements S4-S5

☐ **C = Incomplete:** Motor function is preserved below the neurological level, and more than half of key muscles below the newurological level have a muscle grade less than 3.

☐ **D = Incomplete:** Motor function is preserved below the neurological level, and at least half of key muscles below the neurological level have a muscle grade of 3 or more.

☐ **E = Normal:** motor and sensory function is normal

CLINICAL SYNDROMES

☐ Central Cord
☐ Brown-Sequard
☐ Anterior Cord
☐ Conus Medullaris
☐ Cauda Equina

Appendix II. Functional mobility scale

Seaby L. Mobility: a functional domain for clinical assessment and program evaluation in physiotherapy [Dissertation]. McGill University, 1987.
Seaby L, Torrance G. Reliability of a physiotherapy functional assessment used in a rehabilitation setting. Physiother Can 1989;41:264-270.

The score of each item is based on the following description. The scores are then summed to provide the total functional mobility score. The maximal score is 84.

Item #1. Rolls side to side in bed

1- Dependent

2- 1 person assistance (with/without device)

4- Rolls, needs help for final position (getting comfortable)

5- Independent with device

6- Independent (awkward, more effort)

7- Normal

Item #2. Gets to a sitting position from lying in bed

1- Fully dependent

2- Needs assistance

4- Needs supervision

5- Independent with device

6- Independent (awkward, slow, more effort)

7- Fully independent

Item #3. Sitting balance

1- Not able to sit unsupported

2- No displacement

3- Minimal displacement

5- Moderate displacement

6- Maximal displacement

7- Normal

Item #4. Performance of transfer

1- 2 or more people (with/without device)

2- 1 person with a device

3- 1 person assistance

4- Supervision (with/without device)

5- Assistive device

6- Function (awkward, slow) on level surface, excluding floor to chair

7- Independent, including to and from floor

Item #5. Performance of ambulation (amount of assistance)

1- Non-functional ambulation

2- Dependent for continuous physical assistance

3- Dependent for intermittent physical assistance

4- Dependent for supervision

5- Independent, level surfaces only, assistance with environmental barriers

6- Independent including environmental barriers

7- Normal

Item #6. Performance of ambulation - endurance

1- Not walking

2- < 10 metres

3- > 10 metres

5- > 100 metres

7- > 500 metres

Item #7. Performance of ambulation - aids

1- Parallel bars

2- Walker (including rollator)

3- 2 aids

5- 1 aid (except straight cane)

6- Straight cane

7- No aids

Item #8. Performance of ambulation - velocity

1- 0 m/s

2- < 0.1 m/s

4- < 0.3 m/s

5- < 0.6 m/s

6- < 0.9 m/s

7- > 0.9 m/s

Item #9. Performance of wheelchair mobility (primary mode)

1- Dependent

2- Functional with manual assistance and assistive device (e.g. projection rims)

3- 30 metres but unable to consistently avoid objects

4- Functional with supervision (with/without assistive device)

5- Independent indoors

6- Independent outdoors except curbs and grass

7- Independent outdoors including curbs and grass

Item #10. Ability to handle environmental barriers

1- Dependent

2- Assistance (manual and a device)

4- Supervision (with/without assistive device)

5- Independent in home except stairs

6- Independent in home with stairs

7- Independent in community

Item #11. Arm function (right)

1- Unable to actively move any part of the arm

2- Some active movement - nothing useful

4- Grossly assistive arm movements

5- Partial movement, useful but awkward

6- Functional including fine movements (buttons)

7- Normal

Item #12. Arm function (left)

1- Unable to actively move any part of the arm

2- Some active movement - nothing useful

4- Grossly assistive arm movements

5- Partial movement, useful but awkward

6- Functional including fine movements (buttons)

7- Normal

Appendix III. Recovery of walking after spinal cord injury

M. Ladouceur, A. Pépin, K.E. Norman and H. Barbeau. ***Advances in Neurology*** **1997;72:249-255.**

INTRODUCTION

Spinal cord injury has an estimated incidence that varies greatly across developed countries. The range is from 2.1 to 123.6 per million of population (1-6). Recent epidemiologic studies revealed that the spinal cord injured (SCI) patients are younger, arrive sooner at the hospital, and have less severe cord injuries (7). These recent findings suggest that recovery of behavior such as walking is becoming a possibility for an increasing proportion of these patients. It is increasingly pertinent to study the recovery of walking of SCI subjects who have an incomplete sensory and/or motor function loss (SCI-IMFL). In the past decade, our laboratory has been investigating new therapeutic approaches to enhance the recovery of walking in SCI-IMFL (8).

In results from animal studies, after complete spinal cord transection in which all the descending systems have degenerated, monoaminergic agents that act on the receptors below the transection site were shown to initiate and modulate the locomotor pattern as well as modify spinal reflex activity (9-11). Based on these findings, as well as on preliminary trials (12-14), separate clinical trials have established the effectiveness of certain medications that act on monoaminergic receptors.

In two double-blind crossover studies comparing clonidine with placebo and cyproheptadine with placebo, it was found that with SCI-IMFL patients who were incapable of walking overground, treatment with both drugs greatly improved walking ability, with a more normal muscle activation pattern (15,16). These effects are consistent with findings of reduced excitability of spinal reflexes with clonidine and cyproheptadine in chronic spinalized cats (10,17). These two experimental medications (clonidine and cyproheptadine) are now being compared with each other in the same SCI-IMFL patients and systematically compared with a conventional antispastic medication (baclofen). Preliminary results are somewhat consistent with the previous studies showing a greater effect of cyproheptadine, clonidine, or both in severely SCI-IMFL patients (18). The results from these studies suggest that, as in the cat, noradrenergic and serotonergic drugs have a powerful effect on the modulation of locomotor patterns (19). In summary, results from a small sample showed that clonidine and cyproheptadine were associated with the initiation of locomotor pattern in severely impaired SCI-IMFL, whereas the drugs had minimal effects in most patients who were already capable of overground walking, but this needs further investigation.

To further the study of recovery of walking, Fung and collaborators (20) have investigated the combined effects of clonidine and cyproheptadine together with a locomotor training program using a body weight support (BWS) device and treadmill walking exercise in two chronic SCI-IMFL patients. These patients were incapable of walking overground and had completed standard rehabilitation. The results demonstrated that 6 weeks of locomotor training can further improve the locomotor pattern, as reflected by the kinematic changes, the improvement in the patient's ability to walk with full weight bearing, and the increase in walking speed. These findings suggest that the interactive locomotor training using BWS could be a powerful approach to rehabilitation of walking, after the locomotor pattern has been expressed with medications. This approach is in the process of validation in SCI-IMFL patients.

We have been studying the effect of functional electrical stimulation (FES) used as an orthosis on the recovery of walking of SCI-IMFL patients. Preliminary results (21,22) show that the effects of the FES orthosis are minimal initially, but that with a training program the maximal walking speed of most patients increased markedly. This increase in walking speed is concurrent with a change in walking aids and an increase in endurance. A kinematic analysis of walking showed that the increased walking speed was due to an increase in both stride length and stride frequency. It was also shown that for most patients the energy efficiency of walking is increased with the use of the FES orthosis.

In conclusion, results so far demonstrate clearly that new treatment approaches such as medication alone or combined with locomotor training using BWS and/or FES could enhance the recovery of walking in chronic SCI-IFML patients. The experiments on the use of different therapeutic strategies either to initiate or modulate the locomotor behavior has raised questions regarding the definition and evaluation of the functional recovery of walking.

A MODEL OF RECOVERY OF WALKING

Walking is a behavior that can be analyzed from several perspectives. From a neurophysiologic perspective, we can consider the three levels of neural control mechanisms put forth by Forssberg (23), namely: (a) a basic locomotor pattern required to achieve propulsion, (b) the maintenance of equilibrium, and (c) the adaptation of the locomotor pattern to the behavioral goals of the persons as well as to the constraints imposed by the environment. Some of the constraints imposed by the environment are attentional demand, initiation and termination of locomotion as and when needed, changes in direction, changes in speed, and obstacle management including stairs (clearance or avoidance). From a biomechanical perspective, we can consider the tasks within walking that need to be controlled, as put forth by Winter (24): the generation and absorption of energy at key points in the cycle, the foot trajectory during swing, the support of body weight, and the balance of the head, arms, and trunk.

These two perspectives, among others, contribute to our understanding of walking. However, they do not address how the recovery of walking occurs after spinal cord injury. The process of recovery of walking is a continuous one wherein incremental changes occur in numerous factors such as speed and endurance. Functional recovery of walking is not continuous but is rather a series of transformations from one functional state to another. These states are characterized by distinct ranges of walking abilities. The functional walking categories of Perry and collaborators (25), among other scales, reflect this characterization of functional states. Our studies of the recovery of walking in SCI-

IMFL patients have led us to propose a model of the process of recovery that comprises at least two dimensions (control and capacity). The dimension of control was explained above, whereas the walking capacity is evaluated by the maximum walking speed and distance, the walking aids used, and the management of the environment. In addition, our model of recovery is not seen as a linear process. Figure 25-1 illustrates the model and some examples of partial recovery of walking.

The upper leftmost example is of a severely disabled patient on a motorized treadmill with the support of a harness and parallel bars (or other fixed support for the upper limbs). For the individual in this example, the demands of generation and absorption of mechanical energy as well as support of the body weight and balance of the upper body are less than those of overground walking. Consequences of poor control are much less severe than in unsupported overground walking. Thus, the individual may be able to express the basic pattern of locomotion — a reciprocal stepping motion of the legs — but only in this situation. The walking behavior expressed in this first example represents, from our perspective, a beginning of the process toward functional recovery. The harness and treadmill provide an opportunity to evaluate and train severely disabled patients whose limited walking behavior would otherwise not be revealed. Further discussion of the use of treadmill and harness for the evaluation can be found elsewhere (26).

Examples 2 and 3 in Fig. 25-1 represent an individual walking on a horizontal, noncompliant, smooth surface using either a walker or cane for aid. It is important to note that the same individual may be simultaneously able to perform locomotion in both these situations. However, it should be noted that the capacity of those patients depends on the demands of control of the task. Example 4 in the lower right part of the figure represents walking in a context that approaches a state of nearly full functional recovery of walking. The uneven surface represents the changes in surface height, inclination, and compliance and the upright rectangles represent obstacles that may be encountered. With greater recovery of control, an individual is able to make adaptations in the locomotor pattern:

adapting toe clearance for obstacles, adapting stride length and/or cadence for speed changes, adapting foot placement and postural orientation for direction changes, and multiple adaptations for changes in inclination, avoidance of moving hazards, and achievement of personal goals such as specific destinations. Similar to the other states (examples 1-3), an increase in capacity can be achieved, such that, for example, an individual becomes capable of walking faster and for longer periods of time on slopes or uneven terrain.

Figure 25-2 shows experimental data from two chronic SCI-IMFL patients that illustrate the extent of recovery in the first three examples within our proposed model. These patients underwent experimental trials of therapeutic intervention in our laboratory. The top panel shows the progression of walking speed of a patient (DT) who first participated in the experimental trial of medication on the walking behavior. These trials were followed with the initiation of an FES-assisted walking training program. The FES orthosis consists of the stimulation of both common peroneal nerves and is triggered by handswitches. Two point arise from this panel. The first point is that DT was incapable of stepping independently before the first drug. The second point is that he remained incapable of overground walking until well after he completed the drug study. These results represent the functional situation illustrated in example 1 from Fig. 25-1. It should be noted that the overground walking speed subsequently surpassed that on the treadmill, showing the highly non-linear trend of the process of recovery as put forward by our model.

The bottom panel represents the progression of maximal walking speed of a patient (SM) who first participated in a 12-week BWS training program with no changes in either his maximal overground or treadmill speed. Forty weeks after the end of this therapeutic intervention, the walking speed had decreased and a program of FES-assisted walking was initiated. He used a single channel of stimulation over the left common peroneal nerve triggered by a handswitch. It can be seen that there was an increase in maximal walking speed over time. Concurrently, walking with different aids could be achieved. The participant used a walker at the start of the training period, then started using forearm crutches after 2 weeks of FES training and was able to use two canes after 30 weeks of training. The finding that maximal walking speed was lower with canes than with forearm crutches shows an example of the dependence of capacity on the constraint of the evaluation situation which is found in Fig. 25-1 (examples 2, 3).

IMPLICATIONS OF THE MODEL FOR THE EVALUATION OF WALKING BEHAVIOR

Evaluation of the walking behavior should be based on a theoretical framework. Our theoretical framework incorporates the concepts of Winter (24) and Forssberg (23) in particular. To evaluate neural control mechanisms for locomotion such as stated by Forssberg (23), our laboratory, among others, has developed tasks that enable us to differentiate among the different mechanisms of control.

The task of stepping with a BWS system on a motorized treadmill capable of very low speeds is used to evaluate the basic locomotor generating mechanisms. This system may be the only means of studying walking behavior in severely disabled patients, particularly those incapable of walking overground.

To evaluate the maintenance of equilibrium, an indirect approach used in our laboratory is to study patients walking either on the treadmill with and without parallel bars or with changes in overground walking aids (from walker to forearm crutches to single cane to no aids) (27). Other laboratories have used more direct approaches for the study of maintenance of equilibrium (28).

The third controlling mechanism is responsible for the adaptation of walking as a function of the individual's behavioral goals, as well as the constraints imposed by the environment. As stated earlier, such constraints include attentional demand, initiation and termination of walking as and when needed, change of direction or speed, and the capacity to pass over and around obstacles including stairs. Attentional demand of walking in SCI-IMFL patients has been evaluated to characterize their level of automaticity. For secure walking, it is imperative to be able to start and stop when required by the environmental context. To assess these capacities, we evaluated gait initiation and gait termination using both a predictive and reactive paradigm. We also correlated the changes in frequency of stepping with changes in treadmill speed. The capacity to adapt to obstacles has been evaluated by asking SCI-IMFL patients to go over known obstacles. Preliminary findings of studies in progress show some of the adaptations of walking of which SCI-IMFL patients are capable.

CONCLUSION

The process of recovery of walking encompasses increases in both control and capacity. We suggest that therapeutic intervention or evaluation of recovery requires attention to both. Furthermore, we have evidence that further recovery can often be achieved in individuals who were previously thought to have achieved as much recovery as was likely to ever occur (e.g., chronic SCI).

The implications for research are multiple. First, we need to recognize that interventions (pharmacologic, orthotic, surgical, etc.) that restore some aspect of behavior do not necessarily produce functional recovery in themselves. In many cases, the changes in walking behavior were reported after a period of time. Factors that could account for, and need to be investigated, are time, training, and the intervention as a permissive situation. Second, we need to understand better the process of recovery of control by expanding our repertoire of "controlled laboratory conditions" to include more complex environmental and mechanical demands.

Furthermore, we need to recognize that patients in a given diagnostic category (e.g., SCI-IMFL) will have a range of walking abilities. The probability of success of an intervention is clearly related to the integrity of the structures upon which the intervention depends. The success of FES, for example, depends on the integrity of the peripheral nervous system and associated reflex loops; the success of certain drug interventions depends on the state of the descending monoaminergic systems. However, the walking abilities before the intervention will determine what walking tasks should be measured in order to, most likely, show the effects of interventions. Patients may each have different abilities with respect to walking speed, need for aids, or mechanical demands to which they can adapt. Trials of therapeutic interventions need to account for these differences in their designs to avoid falsely negative conclusions about the efficacy of any intervention. Finally, animal studies have provided a theoretical foundation for some exciting work in the development of new rehabilitation strategies to enhance recovery of walking of SCI-IMFL patients. We believe that an enhancement of the walking abilities after SCI will be achieved by greater understanding of the process of recovery in different animal models, including humans. The greater understanding will arise from a more systematic evaluation of walking behavior after different types of lesions or in response to different interventions (29-33). We also believe that a systematic evaluation of walking for the assessment of interventions such as medications and FES will facilitate comprehensive and integrated treatment for the enhancement of recovery of walking following spinal cord injury.

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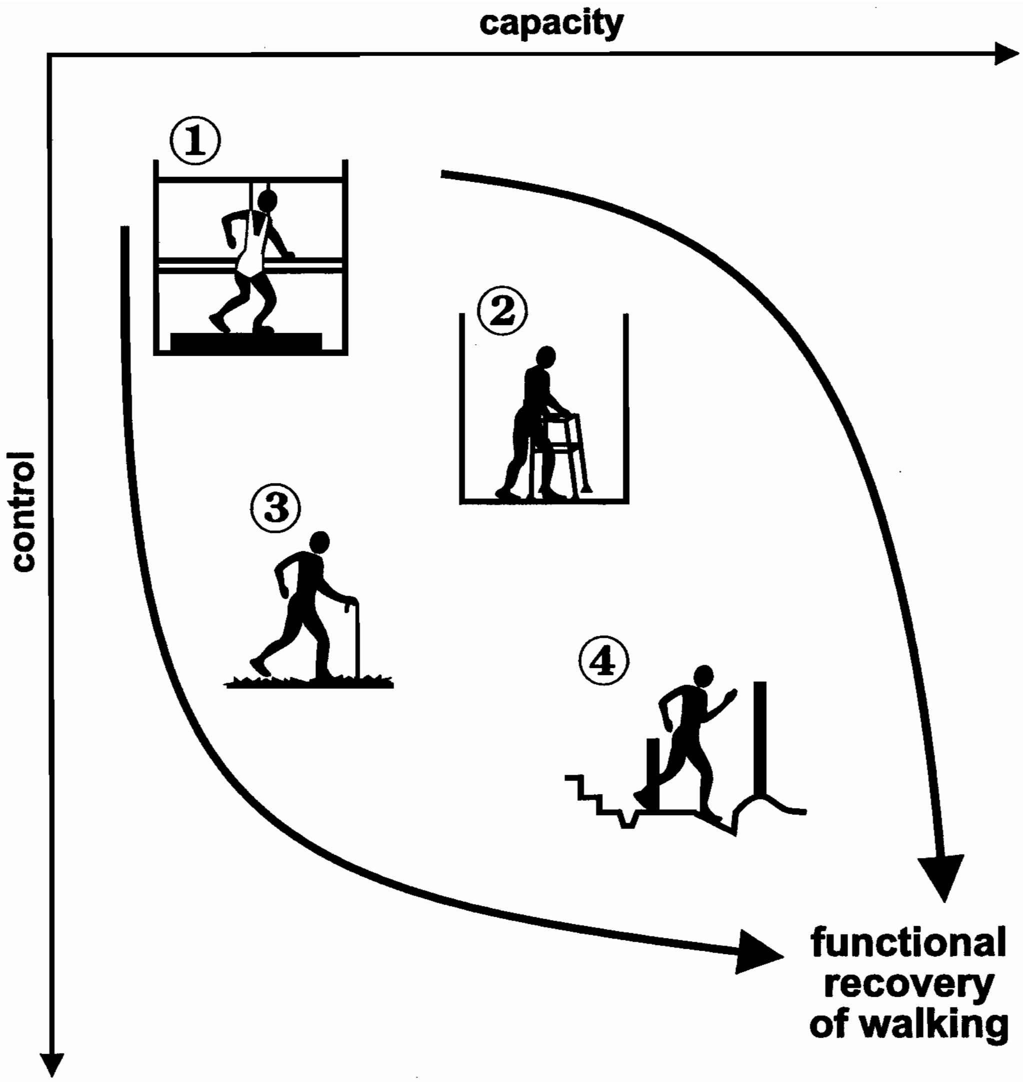

FIG. 25-1. Model of the recovery process of walking. The four examples are explained in the text.

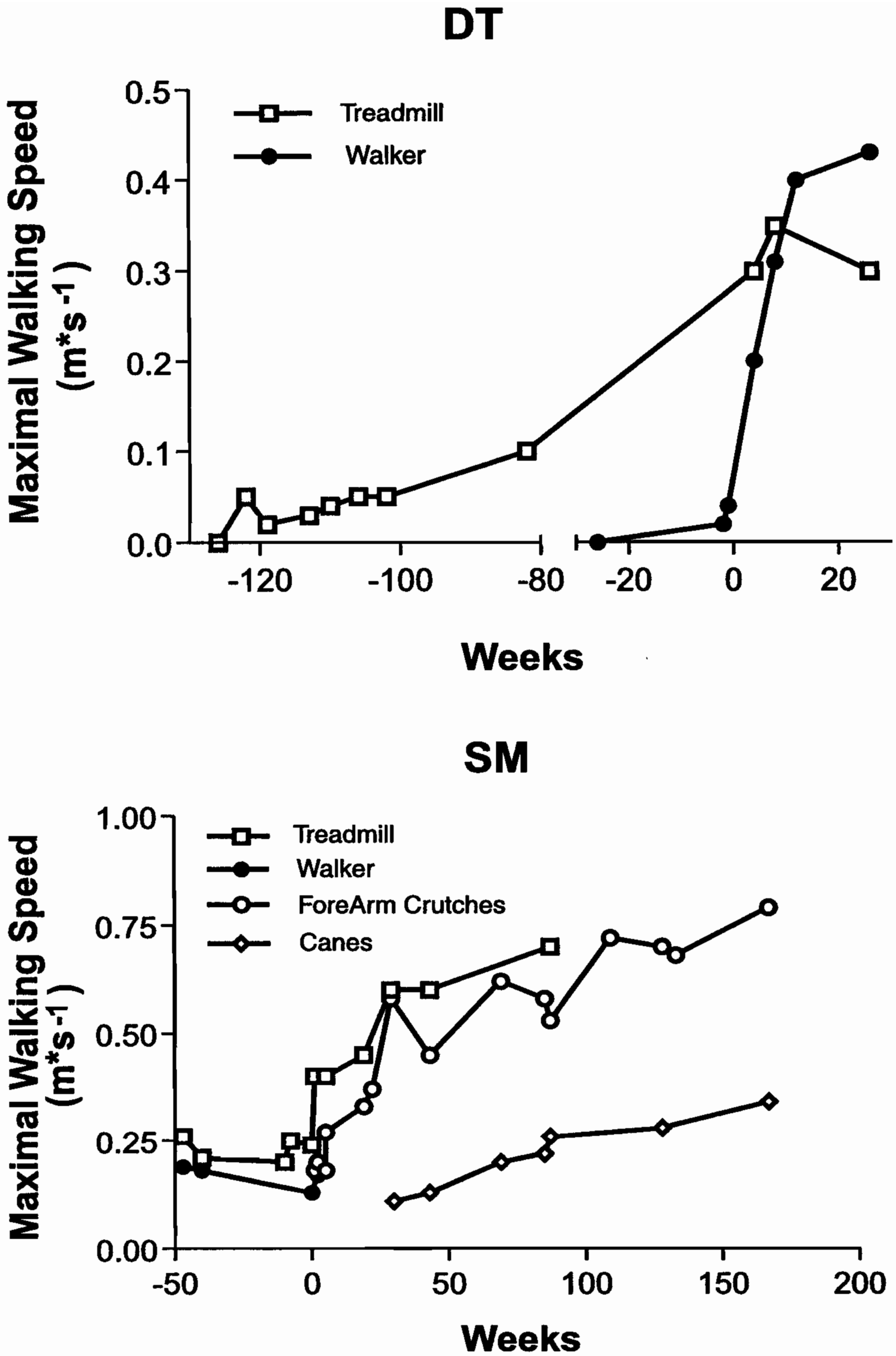

FIG. 25-2. Maximal walking speed data are shown from two chronic SCI-IMFL patients who underwent experimental trials of therapeutic intervention. The abscissa represents weeks of training. The 0 value represents the start of the FES training program. Top panel: progression of walking speed of a patient (DT) who first participated in an experimental trial of medication on walking behavior. The first value is taken before the onset of any intervention. These trials were followed with the initiation of an FES-assisted walking training program. The broken line corresponds to a 60-week period without walking evaluation. Bottom panel: Progression of maximal walking speed of a patient (SM) who first participated in an 8-week BWS training program followed by the onset of the FES training program.

Appendix IV. Walking following spinal cord injury: control and recovery.

H Barbeau, M Ladouceur, KE Norman, A Pépin, A Leroux. ***Archives of Physical Medicine and Rehabilitation*** **1999;80:225-235.**

INTRODUCTION

The incidence of spinal cord injury (SCI) varies around the world, but where comprehensive data are available, the incidence is usually reported to be between 20 and 50 cases per million per year, approximately half of whom are under 30 years of age (1-8). Of all new cases of spinal cord injury, a large proportion have some preservation or recovery of sensory and/or motor function caudal to the level of injury. Such an injury is termed an incomplete injury. In many areas of the world, incomplete injuries are now more common than clinically complete injuries (3, 4, 6). The proportion of patients with SCI who recover some ability to walk is unknown but, from data regarding outcomes on the Frankel scale (9), it may be estimated that one quarter to one third patients with SCI in rehabilitation regain some ability to walk by the time of discharge.

The proportion of SCI patients with incomplete motor function loss (SCI-IMFL) is likely to continue to increase. Trials of methylprednisolone sodium succinate (MPSS) and of GM-1 ganglioside have shown that administration of these drugs to subjects with acute and subacute SCI, respectively, leads to a reduced neurologic impairment in the long-term (10-12). There are several other potential new treatments for acute SCI (13-14) that may continue to reduce the severity of neurologic impairment and, consequently, increase the potential for walking ability.

In view of the above findings and trends in the epidemiology of SCI, it is becoming increasingly important to develop treatment strategies that can enhance recovery of walking after SCI. In conjunction with providing treatment to enhance recovery of walking, rehabilitation specialists must consider factors that may influence such recovery. As illustrated schematically in figure 1, the factors influencing recovery include problems of muscle activation such as weakness and dyscoordination, as well as postural problems related to bearing weight, maintaining balance, and developing propulsion. In addition, problems related to loss of sensory function and hyperactive spinal reflexes may interfere with coordination, balance and other factors important in walking.

WALKING DISORDERS IN SPINAL CORD INJURED SUBJECTS

A first striking difference observed in most SCI-IMFL subjects and in most cases of spastic paresis from different neurological disorders is a reduced walking speed, as compared to normal subjects (15-19). However, the walking speed in SCI-IMFL subjects ranges from "not able to walk" in very severe cases to "normal walking speed" in the least severe cases.

A second important difference observed in SCI-IMFL subjects is a longer cycle duration and shorter stride length. The increase in cycle duration is from an increase in both stance and swing durations. This augmentation of the duration parameters should be seen, in part, as a consequence of the low walking speed of SCI-IMFL subjects rather than a characteristic of walking in this population (20). Indeed, in normal subjects walking at low speed, the cycle, stance and swing durations were increased and the stride length decreased as compared with the values obtained at normal comfortable speed (20-21). Some of the characteristics in SCI-IMFL walking patterns as compared with normal subjects walking at the same speed (0.3 m/sec) are summarized here. Conrad and colleagues (17, 22) found that subjects with spastic paresis from various causes had shorter cycle duration than normal subjects at .55 m/sec. However, contrary results have also been published (20).

Angular displacements of lower limb joints in the sagittal plane in spastic paretic subjects reveal important changes as compared with normal subjects (17,22). Hip joint angular excursion was greater in the spastic paretic subjects than in the normal subjects. All spastic paretic subjects made foot contact with the knee in a flexed position, and for some the knee remained flexed throughout stance. At the ankle joint, the spastic paretic subjects tended to have a more dorsiflexed position at foot contact and the ankle remained more dorsiflexed than normal even at push-off in late stance. During swing, a greater variability of the ankle joint position was noted. It could remain more dorsiflexed or, in contrast, it could remain plantarflexed due to a foot drag (17, 22); similar results have also been reported in SCI-IMFL (20).

Electromyographic (EMG) recordings of lower limb muscles during walking reveal alterations in the timing and the amplitude of the activity in SCI-IMFL subjects (23). Coactivation of antagonist muscles at proximal and distal joints is often reported in spastic paretic and SCI-IMFL gait (23-24). Abnormal activation of triceps surae muscle in late swing or early stance accompanied with clonus is commonly seen, especially in subjects with marked spasticity (17, 22-24).

Hyperactive reflexes and hypertonia are often associated with joint stiffness encountered in spastic paresis resulting from SCI (25). The contribution of the stretch reflex and the nonreflex component to the stiffness of the ankle extensors in normal and spastic subjects has been studied (26-28). Spastic subjects showed an increase of the reflex gain and an absence of inhibition during the swing phase as compared with normal subjects. Further, spastic subjects showed a small increase of the nonreflex torque during gait. These findings suggest that increases in the reflex gain and in the nonreflex torque could both contribute to the stiffness of the ankle joint seen in spastic subjects during gait. Hence, alterations in central mechanisms and changes in intrinsic properties of the muscle fibers could be responsible for the increased stiffness. Other studies also reported higher gain of soleus H-reflex in spastic SCI-IMFL subjects during walking as compared with normal subjects (29). Reduction of presynaptic inhibition of Ia terminals (30-31) and defective reciprocal inhibition (32) have been suggested as possible mechanisms contributing to hyperactive reflexes in SCI subjects.

These hyperactive spinal reflexes were emphasized in early definitions of spasticity along with the finding of increased resistance to passive movement (33-35). Recent discussions of spasticity have emphasized the relation between hyperactive reflexes and other movement disorders including some of the components presented in figure 1 (36-37). Most treatments for spasticity, however, have been designed to address only the hyperactive spinal reflexes or resistance to passive stretch although it has been reported that some medications may also change dystonia (38-39). Their effects on simple voluntary movements (40) and complex motor behaviors have been evaluated more rarely. The relation between treatment for spasticity and for movement, including walking, is increasingly important because of the high prevalence of antispastic treatment in the SCI population. An epidemiological study of spasticity among new SCI patients reported that one quarter had received medical or surgical antispastic treatment before being discharged from rehabilitation. Furthermore, those patients with Frankel B or C severity of injury were almost twice as likely to have received antispastic treatment than the others (41). Data from the United State Spinal Cord Injury Model Systems (US-SCIMS) database showed that almost one third of patients received some form of antispastic treatment before discharge and almost half had received treatment for spasticity within the first year after injury. Among persons with tetraplegia in the US-SCIMS database, two thirds of those with Frankel B or C severity and more then one third of those with Frankel D severity had received antispastic treatment within a year of injury (42). It has been suggested that most of the treatment is pharmacologic in nature (42).

There have been many pharmacologic strategies developed to address the problem of spasticity in various neurologic disorders (fig. 2). Some, e.g. topical anesthetics (43-44), dantrolene sodium (45), and botulinum toxin injections (46-47), have been directed at interrupting peripheral elements of the reflex. Most of the pharmacologic treatments for spasticity, however, are directed at the central nervous system (CNS). Their basis of antispastic effect is their action at several different receptor sites within the spinal cord. Many also have effects at other central receptor sites as well as at peripheral sites, possibly contributing to the antispastic effect and certainly giving rise to some of the side effects. The history of antispastic treatment includes the study of several gammaaminobutyric acid (GABA)-ergic, and glycinergic drugs, as well as noradrenergic, serotonergic and other categories of drugs (48-73).

GABAergic drugs have been widely used for several decades to treat spasticity. They include diazepam, a $GABA_A$ agonist (48), progabide, an agonist at both $GABA_A$ and $GABA_B$ receptors (49), and baclofen. Baclofen is a $GABA_B$ agonist that has been found to be exceptionally useful for treating spasticity and been referred to as the drug of choice for spasticity from spinal cord disease, at least partly because its side-effect profile is relatively benign. For individuals whose spasticity cannot be well controlled with oral baclofen, intrathecal means of drug delivery have been explored (50-61). Several glycinergic drugs have also been studied, including glycine itself (62) and one of its precursors, threonine (63). The use of intrathecal morphine relies on the common association between pain and spasticity (64).

Some of the other centrally acting drugs used to reduce spasticity have been directed at receptors normally stimulated by descending systems such as the noradrenergic and serotonergic systems. Noradrenergic blockers such as chlorpromazine, an alpha-adrenergic blocker, and propranolol, a beta-adrenergic blocker, have received attention because of their ability to reduce gamma motor activity but they have not generally found to be effective at reducing spasticity (48, 65). Propranolol has been found to be effective at reducing clonus (66), although this may arise because of its serotonergic blocking properties.

In contrast, alpha-adrenergic agonists such as tizanidine and clonidine have been more successful at reducing spasticity. Tizanidine, an alpha-2-adrenergic agonist and imidazoline derivative (67) has been shown to be clearly more effective than placebo (39) and comparable to baclofen (68) at reducing spasticity. Clonidine, like tizanidine, is also an alpha-2-adrenergic agonist, and the two drugs have some biochemical similarities (67). Clonidine has also been shown to be a clinically useful antispastic drug (69-71). Among serotonergic drugs, cyproheptadine, a serotonergic antagonist, was able to markedly reduce clonus and flexor spasms (72) and led to decreases in spasticity that were not different from those of clonidine and baclofen (73).

Thus, several neurotransmitters have been implicated in pharmacologic strategies for managing spasticity. However, there has been little consideration of whether antispastic drugs are likely to be associated with improved walking, except for a few studies (to be discussed below). To address this issue, it is important to understand the recovery process and the factors that can influence this process.

PHARMACOLOGIC APPROACHES TO ENHANCE LOCOMOTION IN SCI SUBJECTS

Noradrenergic Drugs

After complete spinal cord transection, all of the descending systems degenerate and their neurotransmitters progressively disappear. Monoaminergic agents that act on receptors caudal to the transection have been shown both to initiate and to modulate the locomotor pattern as well as to modify spinal reflexes.

The alpha-2 noradrenergic agonist, clonidine, has been used to induce locomotion in acute spinal cats (74-76). When clonidine or tizanidine was given during the first week after spinalization (when the spinal cord is transected), a well-coordinated locomotor pattern and full-weight support of the hindquarters could be observed. In contrast, injection of dopamine or serotonergic agonists could not initiate locomotion during the same period (77).

Chronic spinal cord injured cats may achieve a stable locomotor pattern if they are appropriately trained (77) (see section on locomotor training approaches to enhance locomotion in SCI subjects), and this finding has permitted the exploration of several pharmacologic interventions in the modulation of an established locomotor pattern. The administration of clonidine or tizanidine led to increases of the maximal walking speed, the step cycle duration for the same treadmill speed, excursions of the hip, knee, and ankle, and the duration of the EMG bursts (75,78). The EMG records from chronically implanted electrodes in these cats showed a marked increase in burst durations, particularly of the flexor muscles (75, 78). In addition, the threshold current needed to elicit a small flexion reflex through wires implanted in the dorsum of the paw was two to three times higher after clonidine administration. These effects could be partially reversed by yohimbine, a noradrenergic antagonist, confirming that it is through noradrenergic receptors that clonidine exerts a modulatory effect on the locomotor pattern (75,78). When clonidine was administered to chronic spinal cord injured cats that had developed a poor locomotor pattern from to a lack of training, the result was improved locomotor pattern and decreased cutaneous reflex excitability (79).

Based on the aforementioned animal findings, we believe that a rationale for locomotor pharmacotherapy can be developed for SCI-IMFL subjects. From an historical point of view, the first systematic investigation of the effect of antispastic drug therapy on walking used baclofen and tizanidine. The effects of tizanidine and baclofen were compared with those of placebo in subjects with spasticity and paresis from various causes (80). Few statistically significant differences were found in mean ankle and knee angles, and all represented a mean change of 1.5 degree or less (80).

In a placebo-controlled study of the effects of clonidine on the walking pattern of SCI subjects, subjects with clinically complete SCI were supported and assisted in stepping and their ability did not change with clonidine (81). These results have recently been replicated (82). Among SCI-IMFL subjects, the most severely disabled subject showed a marked improvement in walking ability, as revealed by EMG and kinematic patterns (81). A more normal activation pattern of the tibialis anterior and soleus muscles have also been observed in two case reports (83,84).

Recently, an abstract was published on the acute effect of an intrathecal injection of a bolus of clonidine on the overground walking pattern of SCI-IMFL subjects (85). Within a half-hour of the clonidine injection, three SCI-IMFL subjects doubled their maximal overground walking speed. The speed gains were maintained or increased over successive evaluations during the subsequent 6 hours. These gains were associated with increased stride length. During the same time period, the flexor reflex amplitude and the lower limb Ashworth Scale scores were reduced, whereas the H-reflex amplitude remained unchanged (85). In the same subjects, the changes in walking speed, flexion reflexes and spasticity score were minimal on days when placebo injections were given (85). The reduction of signs of spasticity concurrent with the increases of walking speed suggests that intrathecal clonidine may be useful in improving gait in SCI-IMFL subjects (85). Furthermore, these results are consistent with findings of changes in locomotor pattern and reduced excitability of spinal reflexes with clonidine in chronic spinalized cats. As presented in figure 1, there is a relationship between spasticity and walking behavior, but the nature of this relationship needs further investigation.

Serotonergic Drugs

Serotonergic drugs have also been found to have an important modulatory effect on the locomotor pattern of chronic spinal cord injured cats. Administration of serotonergic agonists or precursors led to a marked increase in EMG amplitude of both flexor and extensor muscles and produced movements that were more brisk with a larger angular excursion at the hip, knee and ankle joints (86). The brisk movements during walking resembled, in many respects, the clonus and spasms seen in SCI-IMFL subjects. In addition, serotonergic drugs led to increased flexor reflex responses to an electrical stimulation of the paw. Administration of cyproheptadine, a serotonergic antagonist, was able to partly reverse the effects of serotonergic agonists and precursor on both walking and reflexes (87).

Cyproheptadine has also been studied for its effect on walking pattern in SCI-IMFL (18, 87). In severely disabled subjects, cyproheptadine considerably decreased ankle clonus and spontaneous spasms. These results are consistent with the reduced excitability of spinal reflexes observed in chronic spinalized cats (86). Two of these SCI-IMFL subjects could walk during cyproheptadine therapy without the harness support (see section on locomotor training combined with body weight support) which previously had been necessary. In these subjects, higher walking speeds were achieved and were associated with changes in EMG timing and with marked improvement of joint angular displacement. In contrast, in other subjects who were functionally ambulatory at entry to the study and presented minimal signs of spasticity, very small changes in speed, and EMG and kinematic patterns were observed (87).

GABAergic Drugs

Although there have been numerous reports of baclofen's effects in SCI subjects, especially since the intrathecal delivery mode has come into widespread use, there has been little study of its effects on walking behavior; these are mostly limited to anecdotal reports about intrathecal baclofen's effects on walking (40, 53, 88-95). There is evidence from animal studies that baclofen reduces both reflexes and locomotion (96-98). For example, an increasing dosage of baclofen in chronic spinal cord injured cats led to a decreased flexor reflex as well as paw drag and reduced ability to support the weight of their hindquarters during treadmill locomotion. Higher doses of baclofen led to cessation of locomotion that was partly reversed by administration of a GABA antagonist, bicuculline (96). The effects of baclofen on walking behavior in SCI-IMFL subjects clearly need further investigation.

In summary, the findings presented above demonstrate the potential of pharmacotherapy to enhance the recovery of locomotion in SCI-IMFL subjects. The drugs' effects need to be investigated alone and in combination with each other. Further basic studies are needed to address a number of questions, including the effects of other classes of drugs in spinal cord injured animals (99), and the effects of drugs on the locomotor behavior in animals with a partial lesion (100).

TRAINING APPROACHES TO ENHANCE WALKING IN SCI-IMFL SUBJECTS

Recovery of locomotion after spinal cord transection was considered to be largely dependent on the age of the animal at the time of the injury (101-103). Until the early

1980s, cats whose cords were transected (spinalized) at maturity were described as poor functional walkers with major deficits in their gait pattern. Although they could produce stepping movements with their hindlimbs, they were described as being unable to support their body weight on their hindquarters up to 8 weeks after transection (103-104). This concept was reexamined, however, and the importance of training for accelerating the recovery and maximizing the quality of the locomotor pattern in the adult spinal cord injured cat has been recognized. Cats spinalized as adults can, in fact, recover significant locomotor capability after an "interactive locomotor training" program (75, 77, 105). During interactive locomotor training, the animal is supported by the tail and thus allowed to bear only a portion of its weight, so that it can walk with proper paw placement on the treadmill, while its forelimbs stand on a platform (105). The proportion of weight supported is adjusted according to the animal's locomotor abilities, and this training regimen is carried out on a daily basis. After a period of 3 to 4 weeks of training, the animal is capable of walking with its hindlimbs at different treadmill speeds while completely supporting the weight of its hindquarters with proper paw placement (105). Moreover, the cat's gait pattern is comparable in many respects to that recorded before spinalization (106) as well as that of the intact adult cat (107). Poor locomotor performance can be attained in untrained or stance-trained cats when compared with trained cats (108-109). Thus, interactive locomotor training is an essential factor in the recovery of locomotion in the adult spinal cat.

Locomotor Training Combined With Body Weight Support

To extend this concept to humans, locomotor training using a treadmill and body-weight support (BWS) was introduced in SCI-IMFL subjects (116). BWS reduces the load borne by the lower limbs and can be provided by different means: pneumatic (110), pulley (111-112), spring (113), and robotic systems (114). Supporting a percentage of the body weight (up to 40%) was associated with an increase in comfortable walking speed, single-limb support time, stride length, and endurance (115). The kinematic pattern revealed a straighter trunk and greater knee extension during the stance phase. Furthermore, there was a more normal timing of EMG activity throughout the cycle. There are many clinical implications and advantages of such locomotor training strategies. The different components of gait may be retrained simultaneously under dynamic conditions. Gait retraining may be initiated early in the rehabilitation period, with BWS provided as needed for the subject to assume an upright position and to allow assisted or unassisted stepping by the lower limbs. As patients walk on the treadmill with a reduced load on their lower extremities thus reducing the equilibrium demands, gait deviations may be alleviated. The use of BWS can also be combined with other strategies.

Recent clinical studies suggest that such interactive locomotor training combined with BWS is important in optimizing locomotor patterns and for achieving full weight bearing in SCI-IMFL (115, 116). Effects of locomotor training using BWS are presently being investigated by several groups, in patients with various neurologic disorders of varying severity (115-119).

Locomotor Training Combined With Functional Electrical Stimulation-Assisted Walking

Functional electrical stimulation (FES) was developed more than 30 years ago (120) as an orthotic system (replacing conventional braces with motor responses elicited by stimulating the common peroneal nerve during walking) to prevent "foot drop" in hemiplegic subjects or to be used for SCI patients (121). FES has been used to restore a variety of movements, including walking (122), and was first reported for SCI-IMFL subjects in 1989 (123). This approach, however, has not been widely used in rehabilitation for a number of technical reasons. The stimulators were bulky, unreliable, prone to breakage and expensive (124). Recent studies of simple system for FES-assisted walking (a common peroneal nerve stimulator) showed an increase of the walking speed (less than 0.1 m/sec) and a reduction of oxygen consumption during gait (124-126).

The combination of simple FES system with locomotor training show the usefulness of such devices to improve walking. A gradual improvement (up to 500%) of the maximal walking speed during extended periods of time (up to 3 years) was reported in a longitudinal study (127). But more remarkably, walking speed of the SCI-IMFL subjects was increased even when the stimulator was temporarily turned off. This retained increase of walking speed without FES, called the therapeutic effect, has been reported anecdotally for hemiplegic patients (120) and in SCI-IMFL subjects (127-129). Changes in their assistive ambulatory devices as well as changes in the ergonical efficency of the walking behavior were also shown. The relative contributions of locomotor training and FES-assisted walking in improveming walking speed still need to be determined.

Locomotor Training Combined With Biofeedback

No studies on the effect of locomotor training combined with biofeedback for SCI-IMFL has been published. However, studies using hemiplegic subjects have shown that using either EMG biofeedback (130) or ankle positional feedback (131) along with foot positional feedback (132) can improve the recovery of the walking behavior. However, ankle positional biofeedback seems to be more effective (131). Furthermore, when biofeedback is combined with FES-assisted walking it results in further increases in walking speed as well as improvement in knee and ankle kinematics and reduction in cycle time and asymmetries in stance phases (133). It should be noted that EMG biofeedback in SCI-IMFL has been attempted to improve the recovery of arm function, specifically activity in triceps brachii (134).

Locomotor Training Using Passive Mechanical Orthoses

Passive mechanical orthoses are used to restore locomotion in the SCI population, and the types of orthoses used are well reviewed elsewhere (135). No study, however, reports any therapeutic effect (as defined earlier) with the use of such orthoses.

Locomotor Training Combined With Drugs

The results described above suggest that there is some degree of plasticity in the locomotor circuitry, which evolves as a function of time after spinalization in the adult cat. To determine whether this evolution can be changed by pharmacologic agents, daily injections of clonidine, which can initiate locomotion for a period of up to 6 hours, were given to adult spinal cord injured cats before their daily training in treadmill locomotion (136). After 6 to 11 days of such intense training, the spinal cats could walk without clonidine and maintain this locomotor performance for several weeks (136).

The combination of noradrenergic drugs and interactive locomotor training have proven thus far to be powerful in accelerating the recovery of locomotion in spinal cord injured cats. In light of the above results, the combined effects of clonidine and cyproheptadine together with a locomotor training program using BWS and treadmill-walking exercise have been investigated in two chronic SCI-IMFL subjects (137). They had completed standard rehabilitation but were unable to walk when they entered the study (137). Both subjects were able to initiate walking on the treadmill while taking these drugs, and 6 weeks of locomotor training with continued drug therapy further improved their locomotor capabilities. The subjects became able to walk with full weight-bearing on the treadmill, and some limited overground walking was achieved with walking aids. On the treadmill, they showed a higher walking speed associated with a greater stride length and angular excursion at the hip, knee, and ankle. These findings suggest that interactive locomotor training using BWS may provide a powerful approach to neurologic rehabilitation, after the locomotor pattern has been expressed with medications.

The experiments presented above on the use of different therapeutic strategies either to initiate or improve the locomotor behavior of SCI-IMFL subjects have raised several issues regarding the evaluation and the definition of the functional recovery of walking. One of these is the discrepancy between perceived improvement from therapy and measured changes in walking behavior; this issue may be adressed with more comprehensive locomotor evaluative tasks. In the following sections, we present two conceptual models that integrate the experimental results on the evaluation of the locomotor behavior and its functional recovery.

A MODEL FOR THE EVALUATING WALKING BEHAVIOR

The question of what is controlled during walking behavior has been investigated for many decades (138-141). An analysis of the kinematics and kinetics of human walking demonstrates that the walking behavior can be segmented into four controlled subtasks: the generation and absorption of energy at key points in the cycle; the foot trajectory during swing; the support of body weight; and the balance of the head, arms and trunk (142). These four subtasks are organized by the CNS through three interdependent systems of neurologic structures (functional units): a functional unit that generates the basic locomotor pattern required to achieve propulsion; a second functional unit used to maintain equilibrium; and a third unit used for to adapt the locomotor pattern to the behavioral goals of the persons as well as to the constraints imposed by the environment (143).

Figure 3 represents a model of the evaluation of the walking behavior. This model consists, in the centre, of the three functional units for organizing walking behavior. The inner pentagon illustrates the four subtasks that are controlled during the walking behavior (the control variables). The outer pentagon shows examples of the proposed locomotor tasks that would be used to evaluate the control and capacity of the individual (locomotor evaluative tasks). It is recognized that the four controlled variables are represented to a certain extent in all locomotor evaluative tasks. However, some locomotor evaluative tasks stress certain control variables more than others.

For example, the task of stepping while using a BWS system (112) is used to evaluate the ability of SCI-IMFL on the controlled variable of BWS. The generation and absorption of energy at key points during the walking cycle can be evaluated using a task of modulation of the walking speed (144) or walking slope (145). The trajectory of the foot during the swing phase can be evaluated using a task of obstructed walking (146-148). To evaluate the maintenance of equilibrium, an indirect approach is to study how an SCI-IMFL subject walks either on the treadmill with and without parallel bars or with differing overground walking aids (changing from walker to forearm crutches to single cane to no aids) (149). More direct approaches for the study of maintenance of equilibrium during walking have been used (150).

A MODEL OF RECOVERY OF WALKING

Functional recovery of walking, as opposed to the process of recovery, is defined as a cascade of transformations from one functional state to another rather than a continuous function. This cascade has been quantified in several scales, including the modified functional walking scale (151-152). A combination of velocity of walking and knee extension strength was found to be the best discriminant between household and community walkers with a threshold of .42m/sec for the velocity variable.

Research and clinical observations of the recovery of walking in SCI-IMFL subjects have led us to propose a model of the process of recovery composed of at least two dimensions, control and capacity. The dimension of *control* was defined earlier and is shown in figure 3. The notion of *capacity*, on the other hand, refers to the maximal value each variable can reach for each of the four subtasks. The control and capacity dimensions are found for each subtask involved in controlling walking. For example, the possibility of changing the trajectory of the foot defines the control dimension, whereas the maximum height and length of swing measure the capacity on this subtask.

This model of the process of recovery of the walking behavior is illustrated in figure 4. The process follows a path of recovery characterized by periods of stability and change (127), four examples of which are shown. The first example is of a severely disabled subject on a motorized treadmill with the support of a harness and parallel bars (or other fixed support for the upper limbs). Examples 2 and 3 represents an individual walking on a horizontal, non-compliant, smooth surface using either a walker or cane for aid. Example 4 represents walking in a context that approaches a state of nearly full functional recovery of walking. The uneven surface represents the changes in surface height, inclination, and compliance, and the upright rectangles represent obstacles that may be encountered.

For the individuals in these examples, the demands of generation and absorption of mechanical energy, trajectory of the foot, support of the body weight and balance of the upper body are increasing both in control and capacity. It should be noted that using BWS and a treadmill allows for evaluation of recovery of walking in subjects that would otherwise be considered unable to walk.

STRATEGIES FOR LOCOMOTOR RECOVERY

Figure 5 represents a model of enhancement of the locomotor recovery after SCI and is based on the studies of locomotor recovery described earlier. The vertical and horizontal axes represent the process of locomotor recovery and time, respectively. Two functions are illustrated in this figure. The first function, the behavioral effect, consists of the functional recovery of the individual while using different interventions. The second function is called the therapeutic or residual effect and consists of changes in the functional recovery when all the interventions (eg, FES, drugs) are withdrawn. The difference between the two functions is called the intervention effect. In this model, changes in the functional recovery can be induced by pharmacologic, orthotic (passive and active) interventions, supportive devices (BWS), and ambulatory assistive devices, or any combination thereof. Any intervention can be coupled with an interactive locomotor training regimen to increase the slope of the functional recovery functions. Examples of such changes have been reported for locomotor training combined with drugs or with FES-assisted walking. An example of the changes in maximal overground walking speed after the proposed model has been reported (145). The initiation of a locomotor training program with FES-assisted walking improved the maximal overground walking speed. The initial improvement by using the simple FES-assisted walking system alone was less then 10%. Combined with a year of locomotor training, the maximal overground walking speed was improved by more than 400% when the stimulator was used and more than 200% when the stimulator was temporarily turned off.

The proportion of persons with SCI who have a potential to recover their walking skills is increasing. It was shown in this review that multiple treatment strategies can be used to enhance the recovery of walking in SCI-IMFL subjects both by providing drugs that may initiate or modulate the locomotor pattern, or by providing external orthoses that help them begin to walk (153). It was also emphasized that an evaluation of walking behavior should incorporate many tasks to measure the different controlled variables. Optimized rehabilitation strategies may be elaborated from recent findings regarding the importance of targeting the whole behavior and incorporating training with other interventions. Such strategies may be developed for persons with SCI or other sensorimotor disorders at different times from the onset of impairment.

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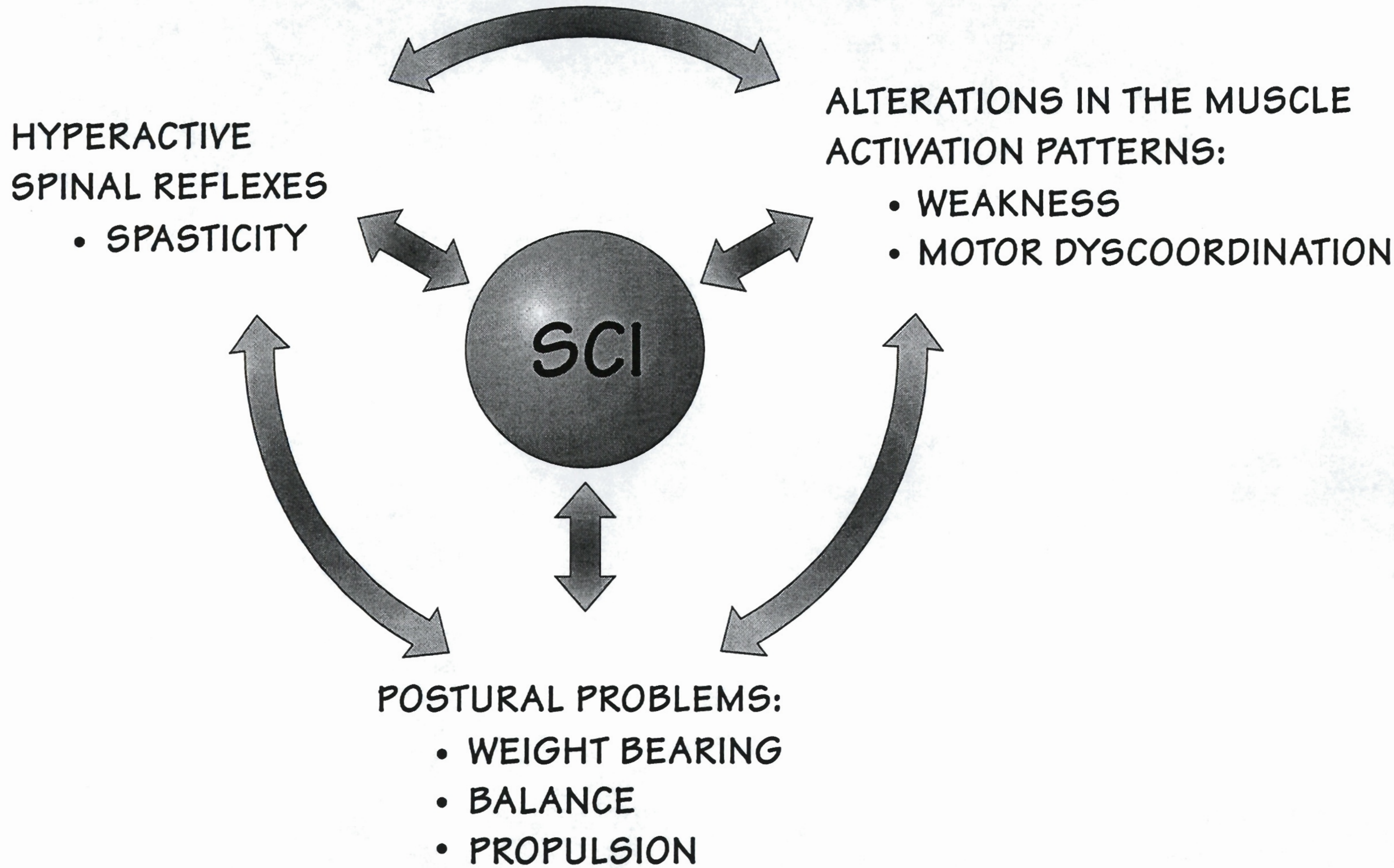

Fig 1. Factors involved in the walking disorders and recovery after SCI.

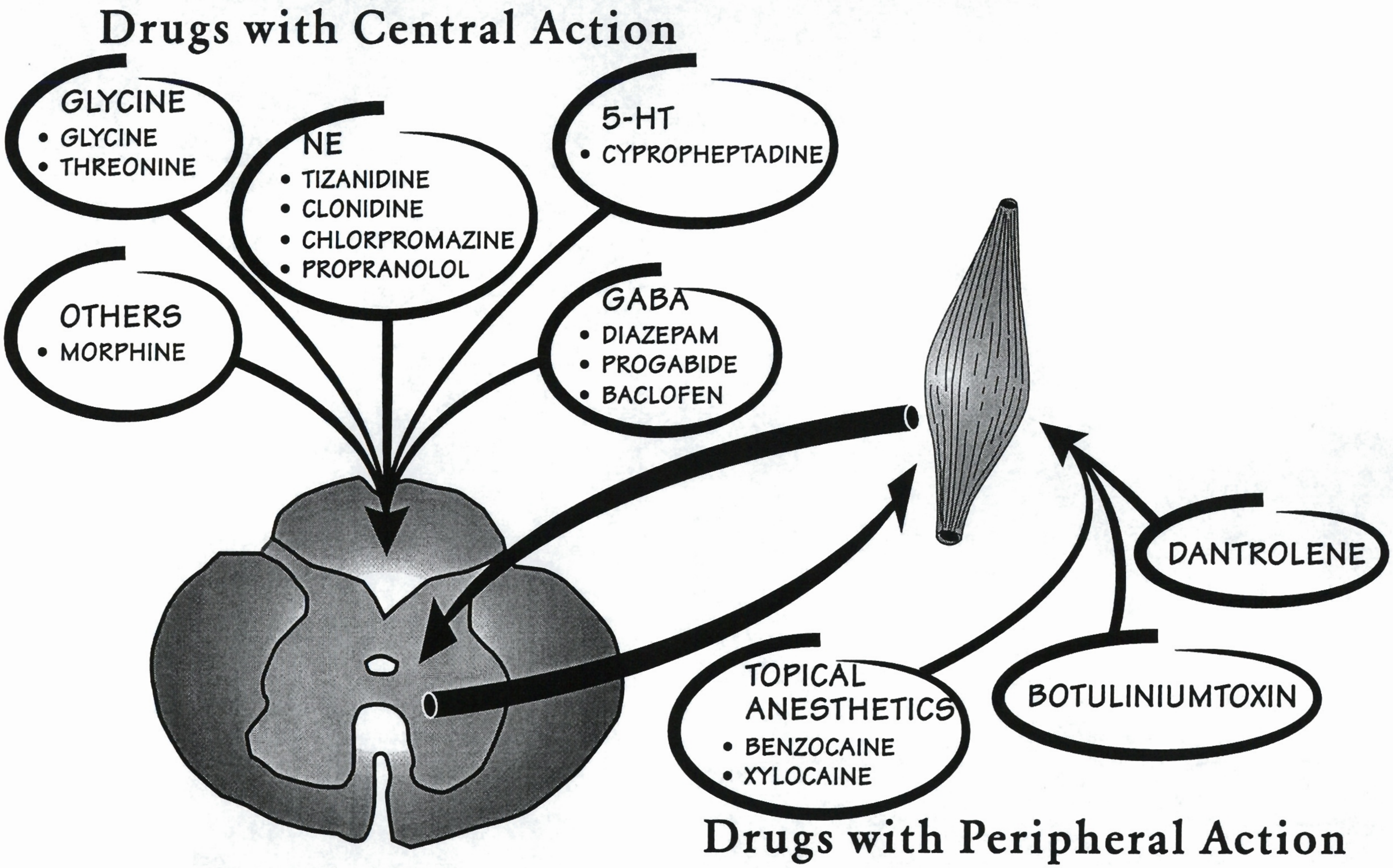

Fig 2. Pharmacologic strategies to address spasticity after SCI.

Fig 3. Model of the evaluation of the walking behavior. The inner pentagon represents four controlled variables during walking, whereas the outer pentagon shows examples of walking tasks that can be used to evaluate those control variables. The inner three circles represent the functional units of control used by the nervous system. HAT, head, arms, and trunk.

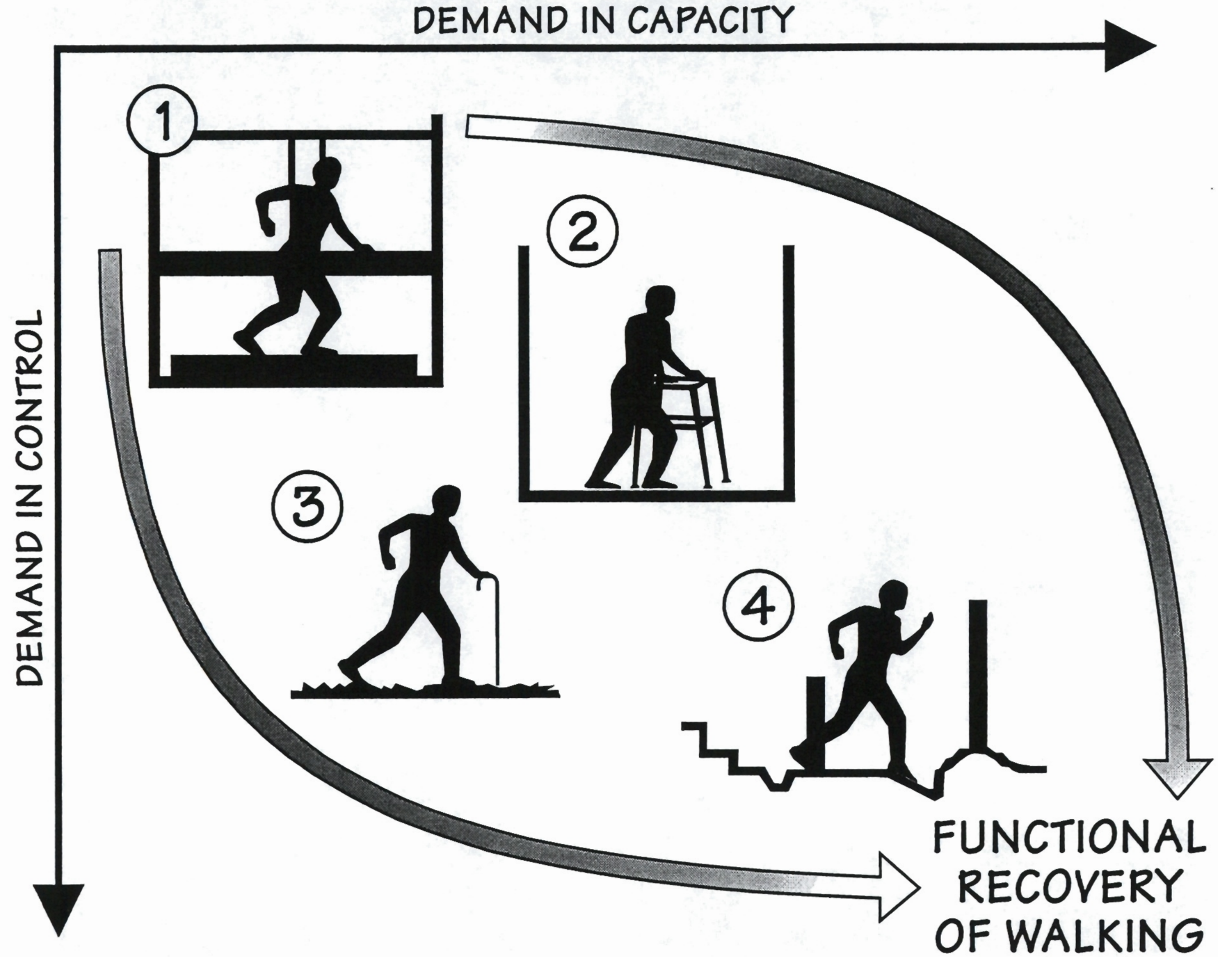

Fig 4. Model of the functional recovery process of walking. The four examples are explained in the text.

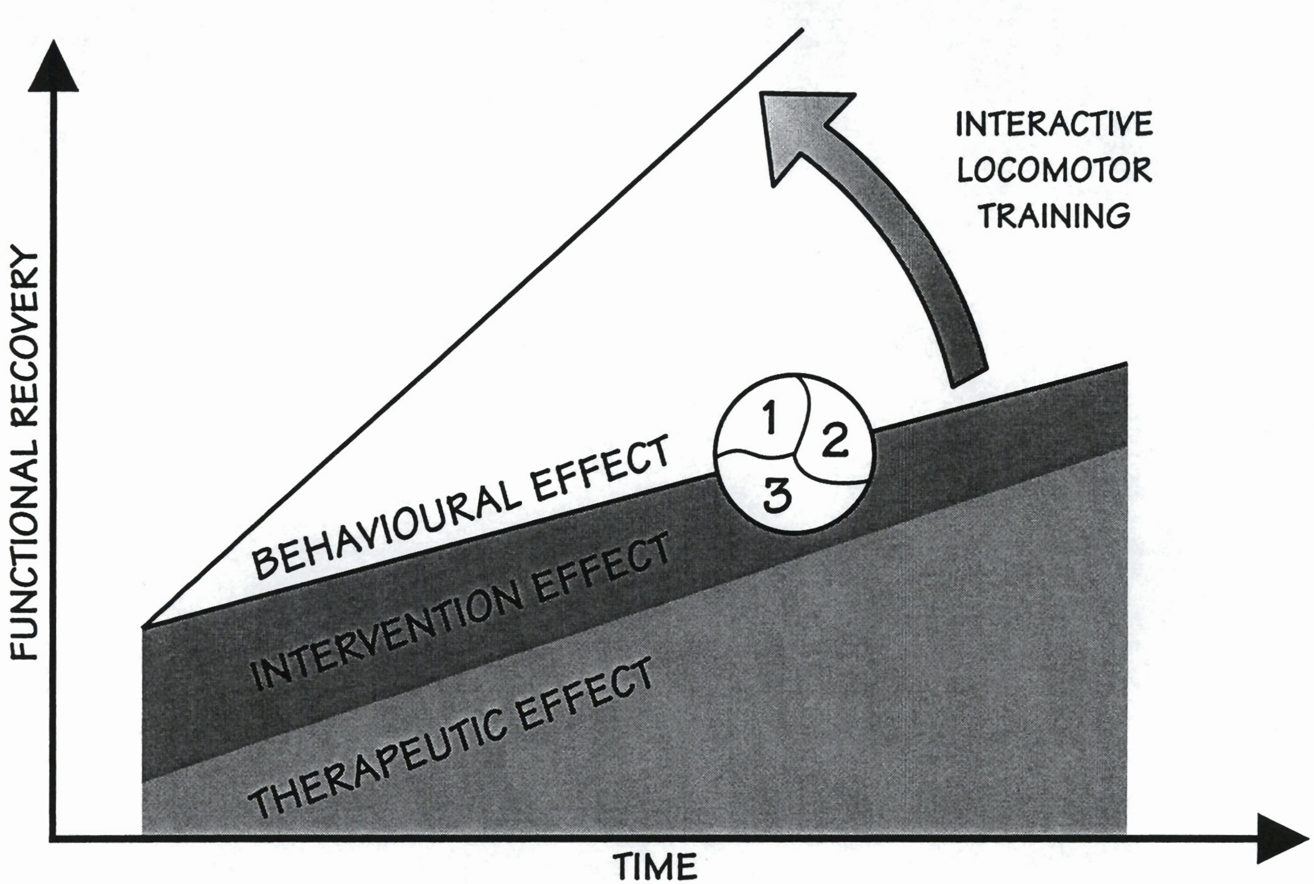

Fig 5. Model of enhancement of locomotor recovery after SCI. The vertical and horizontal axes represent, respectively, the process of locomotor recovery and time; the three functions are explained in the text. Intervention: 1, pharmacology; 2, active/passive assistive devices; 3, surgical procedures; 0, combination. Behavioral = therapeutic + intervention.

Appendix V. Multicenter Evaluation of Electrical Stimulation Systems for Walking

M Wieler, RB Stein, M Ladouceur, M Whittaker, AW Smith, S Naaman, H Barbeau, J Bugaresti, E Aimone, *Archives of Physical Medicine and Rehabilitation* 1999;80:495-500.

INTRODUCTION

More then 30 years ago, Liberson and colleagues (1) introduced single-channel electrical stimulation in stroke patients to prevent foot drop (ie, the foot drops or drags on the ground during the swing phase of walking because of lack of voluntary ankle dorsiflexion). The technique is now generally known as functional electrical stimulation (FES), because stimulation replaces or assists a functional movement that is lost after injury to or diseases of the central nervous system. FES can be distinguished from therapeutic electrical stimulation, which is applied to strengthen muscles weakened, for example, by disuse. Clinical use of FES is increasing; it is currently used to restore and augment respiration, bladder, bowel, and sexual function, hand grasp, and standing and walking (2). Many applications require surgical implantation of electrodes; others are non-invasive, with electrodes applied to the skin's surface. Several studies have reviewed the use of FES in the original application, foot drop (3-5), including one study of more than 1500 hemiparetic patients (4). FES has also been applied to persons who have a spinal cord injury (SCI) or have lost supraspinal control of alpha-motoneurons for other reasons. At least four channels of stimulation are required for walking after complete SCI (6) and more channels of stimulation may be needed for greater speed and better quality of gait (i.e., trunk stabilization) or other functions such as climbing stairs (7). Systems with more than 4 to 6 channels of surface electrodes are complex enough that they have only been tested on a research basis.

Graupe and Kohn (8) applied a commercially available, Parastep[a] to more than 100 SCI subjects. Solomonow and associates (9) tested FES with a reciprocal gait orthosis (RGO) on 70 SCI subjects. Some subjects did very well, but most do not use such a system in place of a wheelchair. Energy consumption is high with FES alone and subjects tire easily. Bracing can reduce the energy cost somewhat and can improve endurance (10-12), but is often bulky and inconvenient to use for long periods.

Simpler systems suffice for SCI subjects where some motor function remains (i.e., the injury is incomplete) (13-14). In a previous study 10 subjects with incomplete SCI showed a significant increase in walking speed (15). The benefits were sufficient to initiate a multicenter trial across Canada. The trial was undertaken because (1) the earlier study had a small sample, (2) FES systems were untested in other clinical centers with less experience using FES, (3) the long-term effects on function were unknown, (4) further insight was needed into the reasons why specific FES systems were accepted or rejected.

SCIs occur at a variety of levels and with varying degrees of sensory and motor preservation. Since our rationale was to determine which types of patients would benefit most and who would use these systems in daily life, we purposely accepted people with a wide range of initial walking speeds. Also, since some subjects were followed for several years, it was not pratical to enroll a very large number of people in the study. Therefore, each subject was tested in each session with and without FES. Subjects, in effect, served as their own controls since therapeutic effects from various sources (eg, electrical stimulation, training and increased amount of walking) could be separated from the specific benefits of FES.

Overall, the goals of the present study were to (1) evaluate simple, surface FES systems for walking, (2) study their acceptance by subjects with an incomplete SCI, compared to subjects with cerebral impairment caused by stroke or head injury, (3) assess limitations to widespread acceptance of the FES devices and (4) develop improved systems that might overcome these limitations. Brief accounts of some results have been presented previously (16-17).

METHODS

Subject Population

Forty subjects used FES-assisted walking for at least 3 months and many have continued for several years. Seven other subjects began the training, but discontinued participation before enough data were obtained for analysis. Of the 40 subjects that continued, 31 had an incomplete SCI. Eight others had cerebral impairment from stroke and 1 from head injury; these 9, however, presented clinically with motor deficits similar to the SCI subjects. The distribution of lesion levels (fig. 1A) is similar to that found in the general SCI population (18), excluding lumbosacral levels of injury to ensure that leg muscles of interest were neither denervated nor receiving normal descending input. The numbers of subjects from the four centers were: Edmonton, 14; Montreal, 14; Vancouver, 7; and Toronto, 5.

All subjects were older than 17 yrs and gave informed consent to participate in the study. They were assessed clinically for range of motion (active and passive), sensation, and voluntary muscle stregth (measured manually). Patients were excluded if they had symptomatic cardiovascular diseases, extreme spasticity, or problems with pressure sores. Adequate cognitive ability was required for subjects to give informed consent, as approved by the human ethics committees at the participating institutions. Although there was some overlap in the distribution of ages at the start of FES the cerebral impairment group was much older, 57 +/- 4 yrs (mean +/- SE), than the SCI group, 36 +/- 2 yrs (fig 1B). The cerebral impairment group was also seen sooner after injury, 3 +/- 1 yrs, than was the SCI group, 6 +/- 1 yrs (fig 1C). All but two subjects entered the study more than a year after injury, so neurological recovery had stabilized and was not expected to change significantly.

All subjects had difficulty walking because of muscle paresis/paralysis following injury to the spinal cord or brain. All could stand and, with one exception, could walk to some extent without FES. There were also a wide range of residual motor ability in the upper extremity of subjects. Figure 1D shows the distribution of initial walking speeds without FES; these are average values for all measurements, typically taken over the first month or two after beginning the program (3.4 +/- 0.5wks). Subjects used a variety of principal walking aids when they entered the program: 13 used a walker, 12 used crutches, 12 used cane(s), and 3 did not use any walking aids. The majority used no bracing, but 3 subjects used a knee-ankle-foot orthosis (KAFO) and 5 used an ankle-foot orthosis (AFO).

Stimulation

Stimulation was with 1 to 4 channels of stimulation, using either the Unistim[b] or WalkAide[c] (1 channel) devices or the Quadstim[b] (4 channel) device. The Unistim and Quadstim devices were based on designs developed at the Jozef Stefan Institute, Belgrade, Slovenia (19), but were produced in Edmonton with some improvements. The Unistim device employed a hand switch to turn stimulation on and off. With the Quadstim device, flexors were stimulated when a hand switch was pressed and extensors were stimulated when it was not. When stimulation of both legs was needed, two hand switches were provided. The WalkAide, designed to overcome problems associated with earlier devices, featured an inbuilt sensor that measures the tilt of the shank with respect to gravity (20). Either the tilt sensor or traditional foot and hand switches can be used to control stimulation. The stimulus is turned on near the end of the stance phase, when the lower leg was tilted back (behind the body), and turned off near the beginning of the next stance phase when the lower leg is tilted forward (in front of the body). Nine subjects used prototypes of the WalkAide that were developed at the University of Alberta. All devices were approved by the Canadian Standards Association.

Subjects presenting mainly with foot drop had stimulation applied to the common peroneal nerve. If the resultant ankle dorsiflexion was insufficient, then the stimulus was increased in some subjects to elicit a flexor reflex and activate hip and knee flexors. A few subjects also had stimulation to the hamstrings to increase knee flexion. If subjects showed knee or ankle instability during stance, stimulation of the quadriceps muscles or the tibial nerve was added. For instability around the hip/pelvis, gluteus medius could be stimulated. Subjects used Multiweek self-adhesive, surface electrodes[d] or the equivalent for stimulation. Stimulation of the quadriceps muscles or other large muscles (gluteals, hamstrings) was with rectangular electrodes (3.5cm x 8.5cm), but the active electrode over the common peroneal nerve was circular (3cm diameter). The type of stimulation used for each subject was left to the judgement of clinicians at each of the centers. No stimulation-induced complications requiring medical attention, such as burns, falls, or fractures, occurred at any of the centers. A few subjects showed minor skin irritation transiently.

Gait Analysis

Gait was analysed using a video camera (plus additional systems in some centers), while subjects walked rapidly, but safely, with and without FES. Subjects using more than one type of walking aid or brace were studied with each assistive device. Typically, four trials of >5 m of walking were interleaved to minimize the effects of fatigue. Subjects began walking before the beginning of the walkway and continued beyond the walkway. Time was superimposed on the video. The time of crossing the 0 and 5m marks was measured from the video and used to compute velocity. Stride length and cycle time were measured from all complete steps within the 5m walkway.

Availability for gait analysis varied with family and employment situations and distance from the participating centers (some lived >500km away). Some subjects of particular interest were recorded many times over more than 3 years to study the full time course of the changes. For general comparisons, initial and final values were used. Where several sessions were available after a stable final speed was reached, results were averaged (51 +/ - 5wks). Averaging increases accuracy, but may underestimate the overall changes, since substantial improvement can take place within the averaged periods (fig 2). The trials could not be done in a blinded fashion, since most subjects felt the stimulus clearly and it was obvious to the person making the gait measurements whether foot drop, for example, was present or absent.

Walking speed (m/sec), stride length (m) and cycle time (sec) were measured over all 5m segments for the more affected leg. Changes over time were fitted with a curve of the form y = a + b exp (ct), where parameters a, b, and c were chosen using a nonlinear algorithm[e].

RESULTS

Speed

Dramatic, long-term changes were observed in a subject (fig 2A), who used a single channel FES system stimulating the common peroneal nerve for over 3 years. His walking speed was initially less than 0.2m/sec without FES and showed little increase when he first tried FES 4.5yrs after his C5/C6 injury. He continued to improve toward an asymptote near 0.8m/sec, a four-fold increase. After a few weeks he switched from a walker to Canadian crutches and abandoned the use of a wheelchair almost completely. Later, he tried canes, which he likes, although his speed is slower. His speed also increased without FES, presumably because of the increased strength from stimulating the dorsiflexor muscles and increased conditioning and coordination from walking more. Originally, he took nearly a minute to walk 10m; he improved to cover this distance in less than 15sec. Five of 40 subjects progressed qualitatively in their normal mode of locomotion (from wheelchair to crutches, crutches to canes, etc.). Figure 2B shows a more typical subject who continued with the same walking aid. He increased his speed both with and without FES over a period of several months.

The total change in walking speeds for all subjects (fig 3) is the final value *with* FES minus the initial value *without* FES. However, some changes may be due to the training and attention that subjects received by participating, rather than a direct effect of FES. This training effect could be assessed by comparing the initial and final speeds *without* FES. The values were quite variable, so subjects were divided into five groups (quintiles) based on their initial speed without FES (fig 3). Subjects typically increased their speed *

Walking speed could be increased by changes in stride length, cycle duration or both factors. Repeated measures of these variables with and without FES were available for 31 of the 40 subjects (Fig. 6). Stride length increased over 20% ($p < 0.01$) with no significant change in cycle time. Other factors have also been analysed. There was a trend depending on the type of walking aid subjects used, but the relative benefits were 0.1 to 0.2m/sec (.14 +/- .03m/sec [mean +/- SE]) from an initial speed of .46 +/- .06m/sec. The training effect accounted for more than half of the increase, .10 +/- .03m/sec, but was not evenly distributed among the quintiles. For example, the fastest group (>1m/sec) showed the largest absolute increase in speed, but the increase resulted entirely from training. Thus, relative rather than absolute changes in speed are more important to analyse. Four values (initial and final walking speed, with and without FES) were averaged to obtain a mean walking speed and values were then expressed relative to this mean value. Overall results and values for the SCI and cerebral impairment subjects separately show an initial increase occurred with FES and a further increase occurred over time (fig 4). The total increase (45%) was observed in all four centers (range, 23% to 67%). SCI subjects' walking speed increased by 55% compared to 19% for those with cerebral impairment. The total change was highly significant ($p < 0.01$, Student's paired t-test) for the entire population and for the SCI and crebral impairment subjects separately. The training effect was also highly significant ($p < 0.01$) in the total and the SCI groups, but not the cerebral impairment group. Finally, the initial effect of FES was significant ($p < 0.05$) in the total and the SCI groups. Results were confirmed using a repeated measures analysis of variance.

Cerebral impairment subjects walked significantly faster than SCI subjects (.75m/sec, compared to .46 m/sec) and the difference between the groups could be an effect of speed, rather than age or other factors. To test this possibility, subjects were again divided into five quintiles (fig 5). The percent increase in speed was nearly 70% for the slowest group (mean walking speed, .12 m/sec) and only 20% for the fastest group (1.26m/sec). The training effect was about 20% for all groups, so the fastest walkers showed no specific benefit of FES. At the average walking speed for cerebral impairment subjects, an increase of 19% with a training effect of 8% is consistent with increase seen in the SCI population. Walking speed could be increased by changes in stride length, cycle duration, or both factors. Repeated measures of these variables with and without FES were available for 31 of the 40 subjects (fig 6). Stride length increased over 20% ($p < 0.01$) with no significant change in cycle time. Other factors have also been analyzed. There was a trend depending on the type of walking aid subjects used, but the relative benefits were confounded by the systematic differences in speed with different walking aids. Subjects who used multichannel rather than single-channel systems benefitted more with little difference in average speed, but the trend was not statistically significant.

Acceptance

At the end of the test period subjects were given the option of continuing with FES; 23 subjects continued to use FES on a regular basis, so acceptance was good. The other subjects discontinued use for a variety of reasons: unrelated medical problems (7), lack of time (4), perceived lack of progress (4), FES no longer needed because of improvements (1), or other (1). Table 1 shows the results of a short questionnaire that was answered by 30 subjects. The least positive answers were to the statement, "I can use the device easily on a regular basis": 7% could *not* use the devices easily on a regular basis and 20% were noncommittal; 73% could use the devices easily. Over 90% agreed with the statement that they could walk better using FES. All responded that the device helped them to do important things and that they would like to continue using FES for walking.

DISCUSSION

This study is the first long-term, multicenter trial of FES for subjects with incomplete SCI. The results were encouraging with a 45% increase in walking speed, over 20% initially with the use of FES and a further 20% during the study. The subjects' SCIs or cerebral impairment occurred on average 5.4yrs before they entered the study, so significant, spontaneous increases in walking speed would not have been expected. Biofeedback has been found to facilitate stroke rehabilitation when combined with FES (21). Thus, participation and the added attention may have contributed to the improvement, as well as the effects of muscle strengthening and general conditioning from walking. However, performance was assessed with and without FES in nearly every session, so the effects of training and involvement can be clearly separated from the specific effects of the stimulation. Also, the training effects would not have occurred without the device enabling them to walk better and farther.

Several subjects improved enough to switch from one aid to another (i.e., from a walker to canes) or to use walking, rather than a wheelchair, as the preferred mode of locomotion. Many subjects had orthoses, such as an AFO, but rejected them for a variety of reasons (eg, discomfort, difficulty fitting into normal shoes, poor appearance). An AFO may offer some of the advantages provided by FES, but FES was found to be more acceptable to many subjects.

SCI versus cerebral injury. We included some cerebral impairement subjects, because simple FES systems have been applied more widely to stroke subjects and therefore provided a good comparison to the results from persons with incomplete SCI. The two groups were quite different in respect to age and several other characteristics (fig 1). Cerebral impairment subjects increased their speed to a lesser extent (19%) then SCI subjects (55%), but the differences could be secondary to differences in walking speed. Simple FES systems were of most benefit to subjects who walked very slowly (fig 5) and our sample of stroke patients walked faster than the SCI subjects. Granat (5) also reported that walking speed improved with single-channel stimulation for subjects whose initial speeds were between 0.1 and 0.6m/sec. The reasons for the smaller improvement in subjects who walked faster initially is presumably that they already have good control over many muscle groups. Adding stimulation to one or a few muscle groups cannot substantially improve their walking speed above the improvement produced by compensating a missing movement for one over which they have voluntary control. Thus, with training those who walked fast initially were able to walk faster, but there was not specific added benefit of FES.

Acceptance. Overall, the responses to the questionnaire were very positive. A large majority responded that their walking was improved and that FES helped them to do things that were important to them. This was true for some subjects who showed no objective increases in speed. Granat (5) found improvements in foot inversion and the symmetry of gait in a number of subjects. These parameters were not systematically measured, but could contribute to the feeling of walking better in the absence of improved speed. Many subjects reported that using FES allowed them to move their legs more easily, to do household chores, and to transfer with less effort, and that it enhanced their sense of well-being. Such changes were not captured by our objective measures. Some subjects also reported that they got less tired walking, which could correlate with the somewhat lower oxygen consumption reported elsewhere (3,15). A few subjects were tested over longer periods of walking (15min) and often showed much larger improvements than would be predicted by the increase in speed over a short runway. These issues need further study.

Limitations. A few subjects disagreed with the statement that the device was easy to use on a regular basis. Difficulties included a problem in finding the sites for surface stimulation, particularly over the common peroneal nerve. Problems were also reported with leads and wires connecting switches and electrodes to the stimulator. No major equipment problems occurred, and minor items, such as broken lead or a malfunctioning switch, could be repaired easily or exchanged.

Further development. Based on the subjects' suggestions an improved single-channel stimulator was designed, which has an inbuilt sensor that measures the tilt of the shank with respect to gravity (20). The sensor signal can be used to turn the stimulator on and off, in addition to the traditional hand or foot switches, even in subjects who are walking barefoot. The stimulator, sensor and electrodes all fit in a comfortable, breathable garment contoured to fit snugly over the tibia. Positioning the electrodes with this device is much more automatic and quicker, even for a person with only one functional arm. Initial reactions to the University-built prototypes have been very positive and a commercial version is under development (22).

CONCLUSION

Subjects with walking speed deficits caused by SCI or cerebral damage who walk at less than 1m/sec can benefit from FES. The good acceptance of the current generation of devices and the development of more advanced, user-friendly devices, suggests that FES should be applied much more in coming years to treat gait disorders.

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Suppliers

a. Sigmedics, One Northfield Plaza, Suite 410, Northfield, IL 60003.

b. Biomotion Ltd., #1503, 10010 119th Street, Edmonton, AB, Canada T5K 1Y8.

c. Neuromotion, Suite 401, 11044 82nd Avenue, Edmonton, AB, CAnada T6G 0T2.

d. Chattanooga Corporation, 4717 Adams Road, PO Box 489, Chattanooga, TN 37343.

e. leastsq, matlab; Mathworks, Inc., 24 Prime Park Way, natick, MA 01760

f. SigmaStat; Jandel Scientific Software, 2591 Kerner Boulevard, San Rafael, CA 94901.

Table 1: Questionnaire Results

Statement	Response Categories 1	2	3	4	5
I can use the device easily on a regular basis.	53	20	20	0	7
I can walk better using FES than without stimulation.	70	20	10	0	0
The device helps me to do things that are important to me.	73	27	0	0	0
I would like to continue using FES for walking.	97	3	0	0	0

Response range, 1 (total agreement) to 5 (total disagreement). Values are reported as percentage of subjects queried (n = 30).

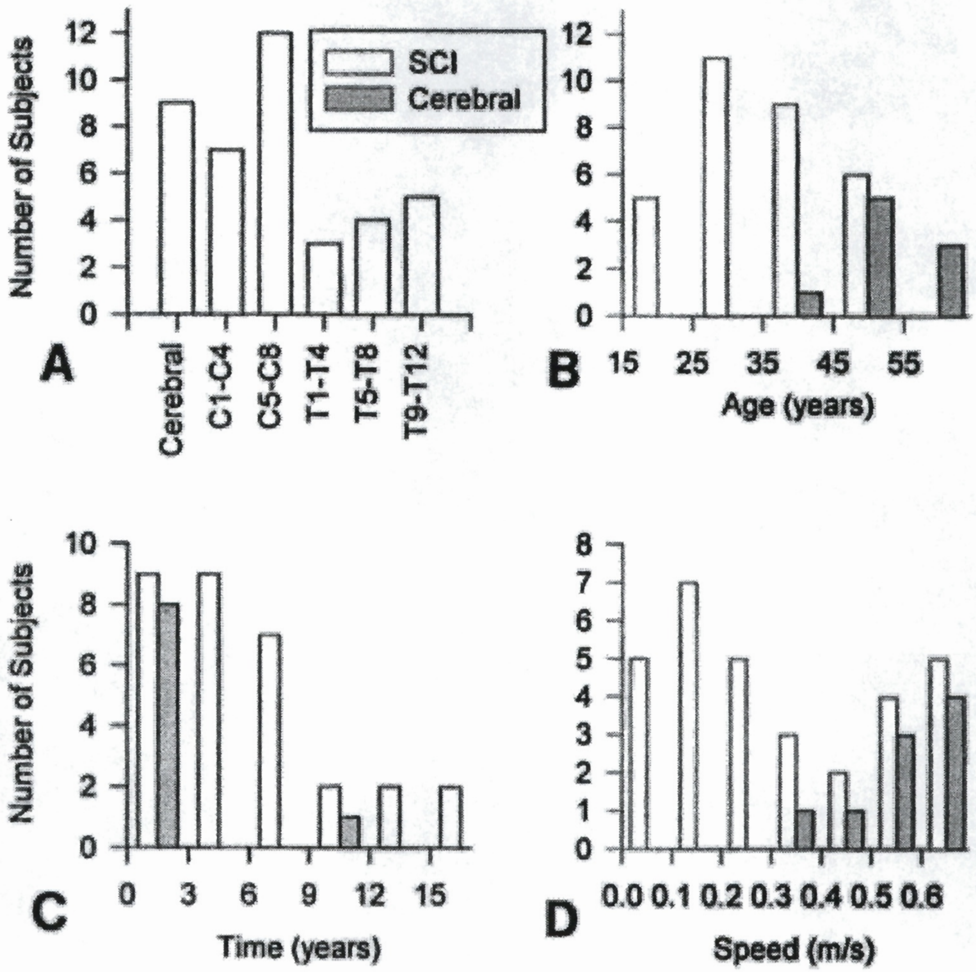

Fig 1. Distribution of (A) the level of the lesion, (B) the ages of the subjects at the start of the evaluation, (C) the time from the SCI or cerebral impairment, and (D) the initial walking speeds of the subjects.

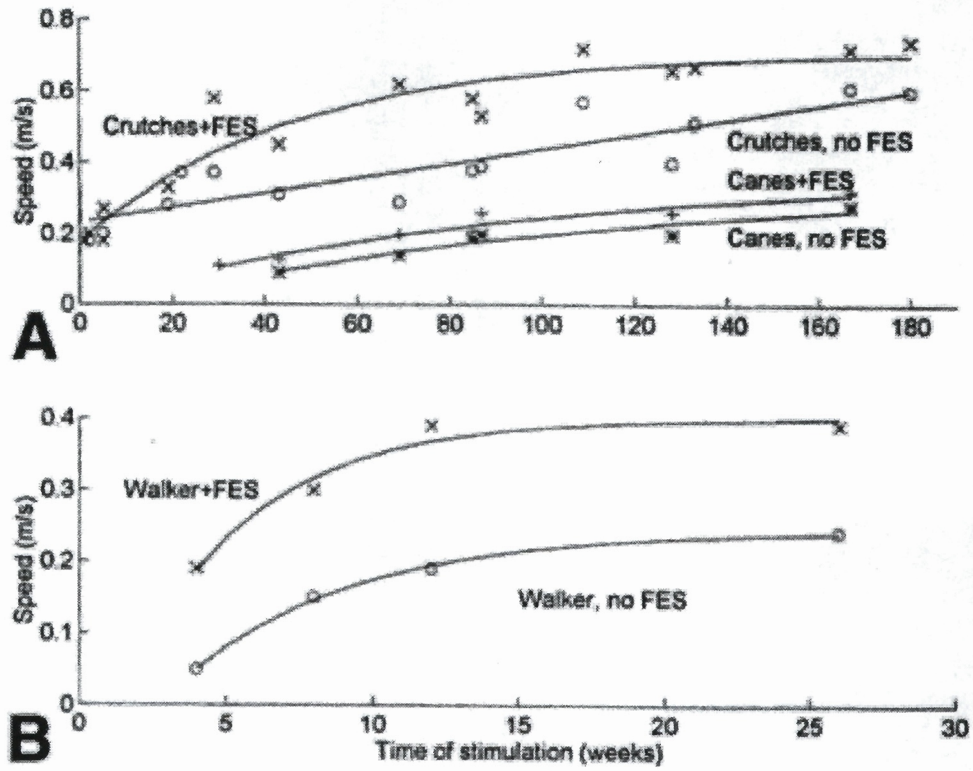

Fig 2. Increased walking speed of two SCI subjects using (A) single-channel or (B) two channels of FES. With single-channel FES, speed increased for more than 3 years, whether using crutches or canes. With two-channel FES an asymptotic speed was reached in a few months. Exponential curves have been fitted, as described in Methods.

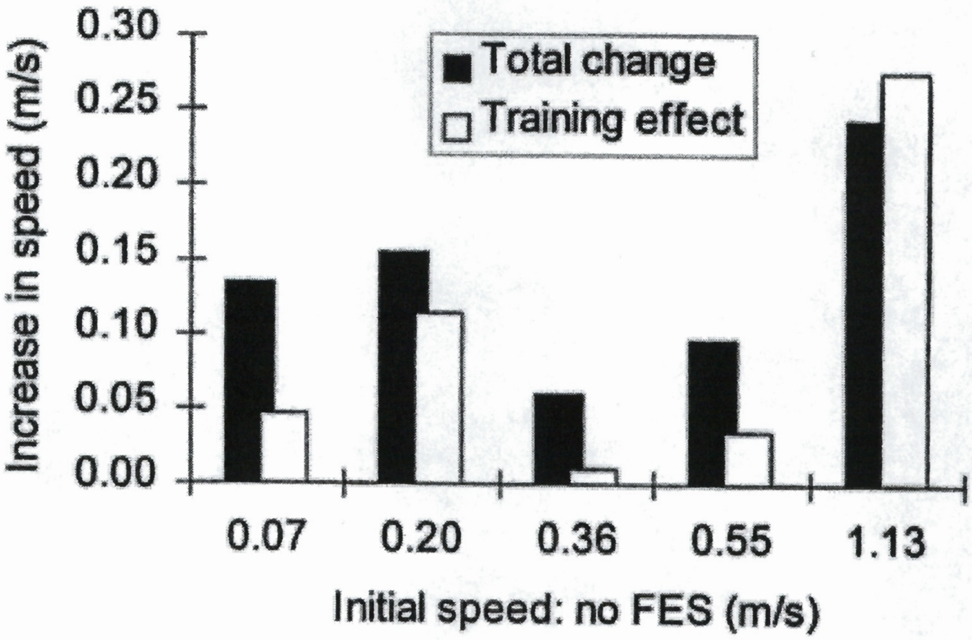

Fig 3. Increase in walking speed as a function of the initial speed with no FES. Subjects were divided into five groups, based on their initial speed. The total change was the difference between the initial value without FES and the final value with FES. The training effect was the difference between the initial and final values without using FES. Most subjects increased their speed by 0.1 to 0.2m/sec, but the entire increase in the fastest group resulted from training.

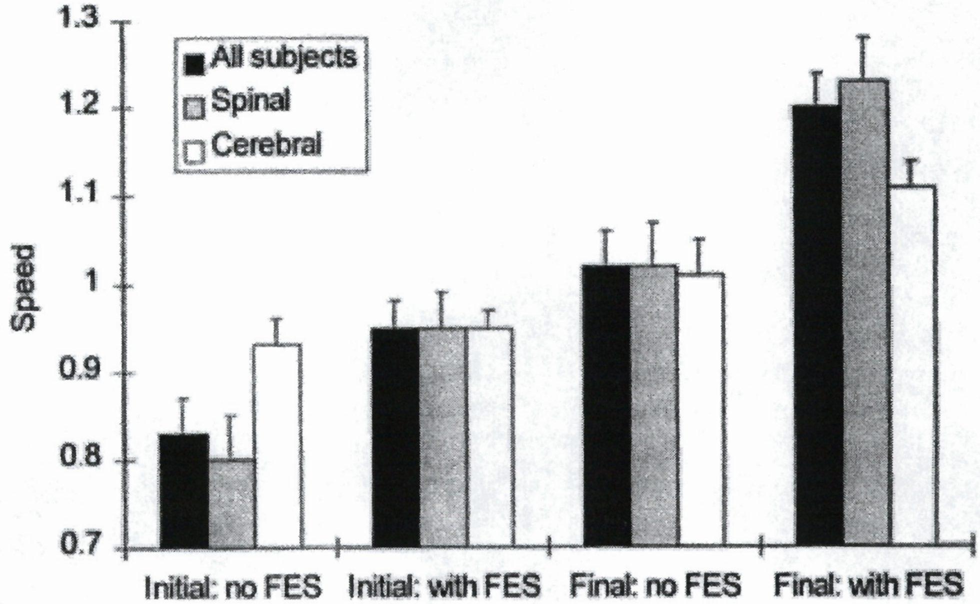

Fig 4. Changes in subjects' walking speed relative to their mean speed (a value of 1). Data are plotted for all subjects and those with spinal or cerebral injury. For each group four bars show the improvements from the initial to the final walking speeds without and with FES. Standard error of the means are also indicated.

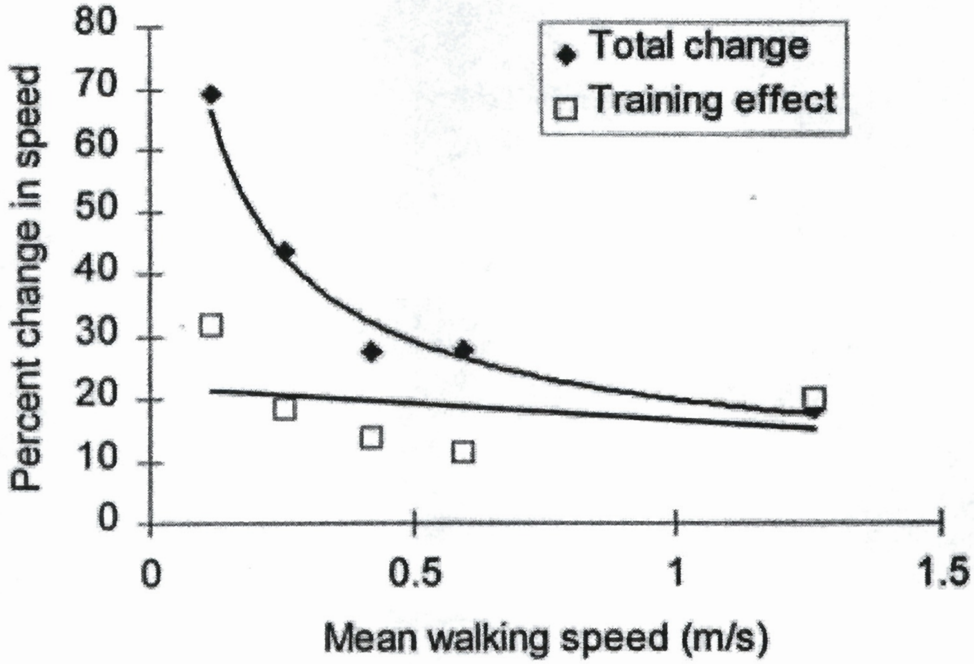

Fig 5. Effect of FES and training on speed. The total increase in speed, relative to mean values for each subject, was greatest for those walking at the slowest speeds (<0.3m/sec). The increase for subjects who walked at high speeds was all attributable to a training effect, which produced about a 20% increase for all 5 groups (8 subjects per group) divided according to their mean walking speed.

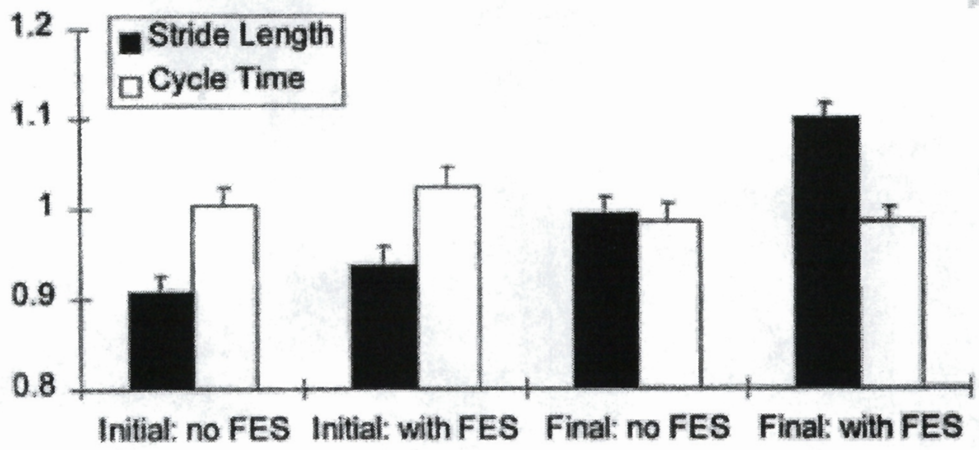

Fig 6. Relative stride lengths and cycle times. The increase in walking speed can mainly be attributed to an increase in stride length, rather then a decrease in the time for a complete step cycle. Values are again shown relative to the mean for each subject, as in figure 4.