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Locomotor and postural adaptations to inclined walking in healthy and spinal cord injured subjects

Submitted by
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May 2001

A thesis submitted to the Faculty of Graduate Studies and Research in partial fulfilment of the requirements of the degree doctorate in Rehabilitation Science

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PREFACE

Contributions of Authors

The present dissertation consists of a collection of three original research papers that have been written by the candidate, in collaboration with Dr. Joyce Fung and Dr. Hugues Barbeau. The first paper presented in the dissertation has been published in Experimental Brain Research whereas the second one has recently been accepted for publication in Gait & Posture. Concerning the third paper, it is in preparation and will be submitted for publication to Experimental Brain Research before the end of year 2001. The candidate has been the first author in each of the three papers. This typically means that the candidate's contribution was substantial and significant in every research paper. Dr. Joyce Fung and Dr. Hugues Barbeau were second and third authors respectively in each of the three papers.

ACKNOWLEDGMENTS

Throughout the course of my Ph.D. studies, I had the opportunity to receive great support from my two supervisors namely Dr. Hugues Barbeau and Dr. Joyce Fung. They both provided important feedbacks on the direction of the research project and on important notions to be included in the thesis. In addition to this and even more important, they gave me the opportunity to acquire a very solid scientific background which is very important (if not crucial) in my present career as a university professor. By their original ideas and great expertise in our field of research, they highly contributed to the development of my scientific formation. For all these reasons, I am grateful to Dr. Fung and Dr. Barbeau. Thank you

Je tiens à remercier ma conjointe Sophie Dubé qui, à maintes reprises, a agi comme guide et m'a aidé à percevoir différents obstacles et moments difficiles au cours de mes études doctorales d'un point de vue réaliste et positif. Par son écoute et ses réflexions éclairées et positives, j'ai toujours su trouver une source de réconfort et la motivation pour aller de l'avant. Merci Sophie!

Je tiens aussi à remercier mes parents, Lise Liboiron et Marcel Leroux, qui ont grandement contribué à mes succès académiques en me montrant, dès mon plus jeune âge, que j'étais capable de réussir ce que j'entreprenais la plupart du temps en y mettant l'effort et la volonté. J'y ai cru et, aujourd'hui, je ne retire que des bienfaits de cet enseignement.

Merci aussi à mes deux filles, Chloé et Léa-Maude qui, jour après jour, m'accueillent à mon retour du travail avec leurs plus beaux sourires remplis de joie et de bonheur.

ABSTRACT

This research project investigated locomotor and postural strategies to adapt to uphill and downhill treadmill walking in healthy and spinal cord injured (SCI) subjects. In the first experiment, subjects from both groups walked uphill at three different inclinations (5, 10 and 15°) while holding on to the handrails on the treadmill. The goal of this first experiment was to characterize changes in lower limb movements and muscle activity patterns required for up-slope walking. The results showed that the key mechanism when adapting to uphill walking in healthy subjects was to lift up the swinging limb by performing a simultaneous increase in hip and knee flexion of that limb. An increase in ankle dorsiflexion was also required to place the foot properly on the inclined surface. Modifications in lower limb movements were accompagnied by progressive increases in the peak amplitude of electromyographic activity of lower limb muscles as the treadmill grade went steeper. The most important increase was observed in the plantarflexor muscles during push off. This increase in muscle activity was necessary to propel the body forward and upward and counteract the resistance due to gravity in uphill conditions. In SCI subjects, a similar trend of adaptation was found only at the hip during uphill walking. Furthermore, the majority of lower limb muscles, including ankle plantarflexors, showed very limited adaptation during uphill walking. The limited changes observed in lower limb movements and muscle activity suggest that SCI subjects have to used different strategies than healthy subjects to adapt to uphill walking. These strategies could involve movements from the trunk and pelvis to compensate for lower limb deficits.

The second experiment specifically focused on the role of the trunk and pelvis in the strategies of adaptation to uphill and downhill treadmill walking in healthy and SCI subjects. Results from

healthy subjects showed that walking on an inclined surface required modifications in trunk and pelvic vertical alignment in the sagittal plane. Uphill walking induced a progressive forward inclination of both trunk and pelvis whereas the opposite trend was found in downhill conditions. These modifications in trunk and pelvic alignment are necessary to counteract the effects of gravity on forward momentum at different slope conditions. In contrast to healthy subjects, SCI subjects with minor deficits in lower limb kinematic patterns showed a greater forward inclination of both trunk and pelvis during level and inclined walking, especially in downhill conditions where both segments were maintained at similar inclination as in level gait. SCI subjects with major deficits revealed analogous (but more pronounced) feature in trunk and pelvic orientation during walking. The whole group of subjects with SCI seemed to adopt this anterior posture to benefit from greater stability during level and inclined walking. Results from frontal and transverse plane motions revealed that compensatory movements from trunk and pelvis were performed only by subjects with SCI showing severe mobility dysfunctions. These compensatory movements were maintained similar across treadmill grades. Results from subjects with SCI and healthy subjects did not show any significant changes in trunk and pelvic total angular excursions in the three planes of motion when comparing level and inclined walking. This absence of changes suggests that lower limbs are mainly responsible for generating or absorbing more energy during uphill and downhill walking in these two groups of subjects.

ABRÉGÉ

Le but de ce projet de recherche était de déterminer les stratégies locomotrices et posturales requises lors de la marche en plan incliné chez des sujets sains et des sujets ayant une blessure médullaire (BM). Lors de la première expérimentation, les sujets provenant des deux groupes marchaient sur un tapis roulant à trois niveaux de pente ascendante (5, 10 et 15 °) en maintenant un contact entre les mains et les barres de sécurité (main courante) du tapis roulant. Le but de cette expérimentation était d'identifier les patrons de mouvement des membres inférieurs ainsi que les patrons d'activité musculaire requis lors de la marche en pente ascendante. Les résultats de cette expérimentation ont démontré que l'adaptation à la marche en pente ascendante requiert une augmentation de la flexion de la hanche, du genou et de la cheville (flexion dorsale) lors de la phase de balancement. Ces modifications étaient assistées par une augmentation de l'amplitude maximale de l'activité électromyographique des muscles des membres inférieurs. L'augmentation la plus importante était produite au niveau des fléchisseurs plantaires durant la phase de poussée. Cette augmentation est requise pour propulser le corps vers l'avant et vers le haut et contrebalancer la résistance plus prononcée provenant de la force de gravité. Chez les sujets ayant une BM, une adaptation était présente uniquement à la hanche. De plus, la majorité des muscles des membres inférieurs, incluant les fléchisseurs plantaires, démontraient une adaptation limitée durant la marche en pente ascendante. Les changements limités provenant des patrons de mouvement et de l'activité musculaire suggèrent que les sujets ayant une BM utilisent des stratégies différentes que les sujets sains pour s'adapter à la marche en pente ascendante. Ces stratégies pourraient provenir du tronc et du bassin afin de pallier aux déficiences observées au niveau des membres inférieurs chez ces sujets.

La deuxième expérimentation avait pour but de déterminer le rôle du tronc et du bassin dans les stratégies d'adaptation à la marche en pentes ascendante et descendante chez les sujets sains et les sujets ayant une BM. Chez les sujets sains, les résultats ont démontré que la marche en plan incliné requiert des modifications au niveau de l'alignement vertical du tronc et du bassin dans le plan sagittal. La marche en pente ascendante entraînait une inclinaison progressive du tronc et du bassin vers l'avant tandis que la marche en pente descendante produisait l'effet opposé. Ces modifications de la part du tronc et du bassin semblent être requises pour contrebalancer l'effet de la force de gravité sur la propulsion du corps lors de la marche en plan incliné. Chez les sujets ayant une BM entraînant une déficience mineure du patron de mouvement des membres inférieurs durant la marche, l'inclinaison du tronc et du bassin vers l'avant était plus prononcée que chez les sujets sains. De plus, cette inclinaison vers l'avant était présente lors de la marche en pente descendante où le tronc et le bassin démontraient le même niveau d'inclinaison que lors de la marche à l'horizontal. Le même phénomène, mais plus prononcé, était présent chez les sujets ayant une BM entraînant une déficience majeure du patron de mouvement des membres inférieurs durant la marche. Cette posture anormale du tronc et du bassin vers l'avant semble être adoptée par les sujets ayant une BM dans le but de bénéficier d'une plus grande stabilité durant la marche en plan incliné et à l'horizontal. Les résultats provenant des mouvements produits dans les plans frontal et transverse ont démontré que les sujets ayant une BM entraînant une déficience majeure des membres inférieurs utilisent le tronc et le bassin comme mécanisme de compensation durant la marche. Les mouvements compensatoires produits par le tronc et le bassin durant la marche à l'horizontal étaient maintenus constants à tous les niveaux de pente. Les résultats des sujets sains et des sujets ayant une BM ont aussi démontré que les excursions angulaires totales du tronc et du bassin dans les trois plans de mouvement ne variaient pas durant la

marche en plan incliné. Cette absence de changement suggère que les membres inférieurs sont principalement responsables de générer ou d'absorber plus d'énergie durant la marche en pente ascendante ou descendante chez les sujets sains et les sujets ayant une BM.

INTRODUCTION

In everyday situation, the walking pattern has to be adapted to different conditions such as stair climbing, inclined surfaces, changes in walking speed and others. These adaptations have to be performed by changing the level of activity of the appropriate muscles as well as the pattern of lower limb movements. In addition to the lower limbs, each of the conditions mentioned above also require postural adaptations from upper body segments to maintain balance. For instance, when adapting to uphill walking, healthy subjects increase the flexion of lower limb joints during swing phases and adopt a forward inclination of the trunk and pelvis. These modifications are also accompanied by changes in the level of activity of lower limb muscles. When sensory and motor functions are intact, the appropriate locomotor and postural responses are generated to adapt rapidly to the new mechanical demand.

Following a spinal cord injury (SCI), sensory and motor functions are usually impaired and this may affect the capacity of the basic walking pattern to adapt to different environments. From animal studies, it was shown that complete spinal cats have limited capacities to adapt to increases in speed and inclined surfaces (Bélanger et al., 1988; Bélanger et al., 1989). In humans who sustain incomplete SCI, the adaptability of the locomotor pattern to changes in the mechanical demands could be greater than in the complete spinal animal since some descending tracts are preserved.

Inclined walking represents an excellent task to determine the capabilities of adaptation following SCI. During uphill walking, the body has to be moved forward and upward and counteracts the increased resistance due to gravity. This can only be achieved by increasing the forward propulsion of

the body and by modifying lower limb patterns of movements. Furthermore, an uphill surface tends to push the body passively backwards by gravity and, to counteract this effect, postural modifications have to be performed. On the other hand, walking down-slope produces the opposite effect of pushing the body forwards by gravity and active control is required to counteract this passive physical effect. Thus, the task of walking on inclined surfaces has been selected to evaluate both locomotor and postural capabilities of adaptation following SCI.

Adaptation to inclined walking has received relatively little attention in the literature and most of these studies were done in individuals with no walking deficits. Hence, very little is known about the adaptability of SCI persons to such changes. The main purpose of the present work is to determine the locomotor and postural adaptations to uphill and downhill walking in healthy and SCI subjects. Specifically, the kinematic patterns of lower limbs, pelvis and trunk as well as the electromyographic (EMG) activity of some lower limb muscles will be examined during inclined walking. Furthermore, by examing the relationship between lower limb movements and trunk and pelvic control during inclined walking, it will be possible to determine how can trunk and pelvic segments participate in the process of adaptation in healthy and SCI subjects.

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LITERATURE REVIEW

Introduction

The problem of adaptation in individuals with impaired locomotor function is complex and not well covered in the literature. Hence, very little in known about the contribution of the lower limbs and upper portions of the body to the adaptation of the gait pattern in individuals with SCI. The first part of this literature review describes the strategies of adaptation of the gait pattern to inclined walking in animals and humans. The second portion of this review demonstrates the importance of pelvic and trunk movements in the control of normal and pathological gaits. Adaptation of pelvic and trunk segments to changes in the mechanical demands such as speed and incline are reviewed.

Adaptation of the gait pattern to inclined walking

The effects of incline on the gait pattern have been addressed in only few studies. From animal studies, different mechanisms of adaptation of the basic motor patterns were proposed to be responsible for slope adaptation in cats and rats. On the other hand, human studies suggested that adaptation to inclined walking was achieved by changing the pattern of lower limb motion and by varying the level of excitation of the relevant flexor and extensor motor unit pools.

Gait pattern has been studied during inclined walking in normal cats (Pierotti et al., 1989; Smith and Carlson-Kuhta, 1995) and rats (Hutchison et al., 1989; Roy et al., 1991). From these studies, it was shown that the basic walking pattern observed in level gait is altered during inclined walking. Some authors (Smith and Carlson-Kuhta, 1995) found important alterations in the basic motor patterns of

the flexor muscles while others (Pierotti et al., 1989; Hutchison et al., 1989; Roy et al., 1991) showed that inclined walking influenced extensor muscles of the lower limbs.

Pierotti and colleagues (1989) examined the effect of different inclines (10 and 20%) on the EMG activity of cat ankle flexors and extensors during walking on a treadmill. When comparing uphill to level walking at the same speed, they showed an increase in integrated and mean EMG amplitude of the medial gastrocnemius. Similarly, the soleus EMG activity increased slightly during uphill walking. In contrast, relatively little change occurred in the EMG amplitude of the tibialis anterior in these walking conditions.

Roy and collaborators (1991) studied EMG activity of rat ankle flexors and extensors during level and uphill walking on a treadmill and found similar results as Pierotti et al. (1989). Comparison of uphill (10, 20 and 30%) and level walking at the same speed, showed an increase in the activation levels of the medial gastrocnemius but not in the soleus or tibialis anterior. The same results were obtained with increased treadmill speed. Difference between medial gastrocnemius and soleus was attributed to the morphologic components of those muscles. The medial gastrocnemius is mostly composed of fast-twitch fibers while soleus is a slow-twitch plantar flexor muscle. Authors suggested that due to the fact that soleus muscle is almost fully recruited even at slow speed of walking, this muscle seems to be unable to compensate for the additional demand required at increased level of treadmill slope. However, one of the limitation of this study is that the slopes used do not seem to be steep enough to produce major changes in the rat EMG patterns. The use of a steeper slope, such as 50% (27°), may have lead to an increase in the amplitude of the soleus muscle.

Smith and Carlson-Kuhta (1995) investigated EMG activity of some lower limb muscles during slope (25% to 100%) walking in the cat. When comparing uphill and downhill cycle duration, it was shown that downhill cycles tended to be shorter than uphill cycles for the same slope. In the downhill conditions, unexpected changes were found in both flexor and extensor muscles: During the stance phase, flexor muscles at the hip and ankle joints became active while activity of hip extensors disappeared. Further, EMG activity of vastus lateralis and lateral gastrocnemius was significantly reduced in duration during stance phase of downhill walking compared to uphill condition. According to Smith and Carlson-Kuhta (1995), the presence of flexor muscle activity during stance phase of downhill gait would be required to counteract the extensor moment produced by external forces.

Results obtained in the Smith and Carlson-Kuhta study differ from those of others (Hutchison et al., 1989; Pierotti et al; Roy et al., 1991) which showed alterations in the extensor muscles instead of flexors muscles. Divergence in these findings may be attributed to differences in methodological approaches. Smith and Carlson-Kuhta (1995) studied EMG burst duration during both uphill and downhill walking whereas others (Hutchison et al., 1989; Pierotti et al; Roy et al., 1991) investigated EMG amplitude in uphill walking only. Hence, alterations in the motor patterns of the flexor muscles reported in the Smith and Carlson-Kuhta study were found only in downhill walking. Furthermore, walking grades used in that study were much more pronounced than in the other studies. However, some results were in concordance with findings from studies presented earlier. Even if Smith and Carlson-Kuhta did not report it in their study, it can be clearly seen from their results that the lateral gastrocnemius is much more active in uphill (50%) than downhill (50%) walking. These results are in

agreement with those of others (Hutchison et al., 1989; Pierotti et al; Roy et al., 1991) which demonstrated that gastrocnemius activity rose in amplitude as the walking slope increased.

Adaptation of human gait to inclined walking is achieved by modifying several factors such as stride length, cadence, angular displacements and muscle activity. During downhill walking, stride length decreased significantly while it increased somewhat in uphill walking (Hirokawa, 1989; Kawamura et al., 1991; Wall et al., 1981). Inclined walking produced different effects on the cadence. Hence, it was shown that cadence decreased in uphill walking and remained relatively constant in downhill walking (Hirokawa, 1989; Kawamura et al., 1991). Further, the decreased cadence was accentuated as the uphill grade went steeper (Kawamura et al., 1991). Concerning the stance to swing ratio and the cycle duration, it was shown that they were relatively constant in inclined walking (Kawamura et al., 1991; Simonsen et al., 1995). The effects of incline upon the afore-mentioned spatio-temporal variables of gait were found to be similar during treadmill (Simonsen et al., 1995; Wall et al., 1981) or overground (Hirokawa, 1989; Kawamura et al., 1991) walking.

Changes in angular displacements of lower limbs have been reported during inclined (-20 to 20%) walking in healthy subjects (Wall et al., 1981). Wall et al. (1981) showed that during uphill walking, there was a significant increase in hip and knee flexion from mid-swing to early stance. In downhill walking, the hip flexion decreased from swing to early stance while there was an increase of the knee flexion in the stance phase. According to the authors, the increased knee flexion of the supporting limb observed in downhill walking, is performed to lower the centre of mass and decrease the impact of heel strike of the opposite limb.

Changes of normal patterns of muscle activity measured during inclined walking are associated with modifications of joint angular displacements discussed above. Gray and Basmajian (1968) performed EMG and kinematic analyses on healthy subjects during uphill (21.25% or 12°) and downhill (21.25%) walking at comfortable speed. They found that during uphill walking, the foot contact is made progressively more toward the fore foot. Modification in the initial foot contact found during uphill walking leads to a decrease of tibialis anterior activity at early stance phase. However, when considering the complete cycle, tibialis anterior was more active during walking downhill as compared to level. These results are in disagreement with Simonsen and colleagues (1995) who found no changes in tibialis anterior activity during uphill walking as compared to level. Difference in the results may be attributed to the treadmill slope. In the Simonsen et al. study, the treadmill slope was only 8% (4,8°). This incline may be insufficient to induce important modifications in the ankle joint displacements and lead to changes in muscle activity. When investigating the effects of incline upon the gait pattern, other studies (Dietz et al., 1981; Gray and Basmajian, 1968; Tokuhiro et al., 1985; Wall et al., 1981) always used steeper slope (from 17,6 to 21,25%) to examine modifications in angular joint displacements and muscle activity.

Simonsen and colleagues (1995) found important changes in the activation level of ankle extensor muscles during inclined walking. Comparison of uphill (8%) and level walking showed an increase in the mean EMG amplitude of the medial gastrocnemius and soleus. This increase was more pronounced in medial gastrocnemius muscle. In contrast, during downhill walking, both ankle plantarflexors showed a significant decrease in mean EMG amplitude as compared to level walking.

Inclined walking can also induce important changes in the timing of EMG activity. Tokuhiro et al. (1985) studied the effect of different inclines (0 to 21.25%) on the phasic EMG activity of lower limb muscles during walking in healthy subjects. When comparing uphill to level walking, tibialis anterior activity started earlier (at heel off) and contracted during the push off period while the phasic pattern of gastrocnemius activity did not change. In downhill walking, the phasic pattern of tibialis anterior activity was the same as in level walking whereas gastrocnemius showed premature activation (at heel contact). Proximal muscles of the lower limbs also showed adaptations to inclined walking. In uphill walking, the initial burst of rectus femoris activity was delayed while semitendinous showed a prolongation of stance-related activity as the slope increased. In downhill walking, rectus femoris activity ceased earlier (at toe-off) and the semitendinosus showed biphasic pattern with a new burst of activity that emerged during late stance. Tokuhiro et al. (1985) provided a detailed analysis of the changes of phasic EMG patterns during inclined walking. However, the major limitation of this study is related to the fact that they did not measure EMG amplitudes. Without this information, it is difficult to obtain a deep understanding of the roles and adaptations of these muscles during inclined walking. According to the results of Wall et al. (1981), it can be hypothesized that modification in the timing of proximal muscles (rectus femoris and semitendinosus) during the stance phase is associated with the increased knee flexion of the supporting limb in uphill and downhill walking.

The effects of inclined walking upon the gait pattern have also been investigated in spastic paretic subjects, although the results were very brief and incomplete. Dietz and colleagues (1981) reported that during uphill (10°) walking, the reciprocally organized pattern of activity between medial gastrocnemius and tibialis anterior seen in healthy subjects was not altered in the spastic paretic

subjects. Despite similar EMG profiles in the two groups, modifications were observed in ankle angular displacements of the spastic paretic subjects as compared to the healthy subjects. During uphill walking, spastic paretic subjects performed the initial foot contact with the ankle in a more plantarflexed position than healthy subjects. In some subjects, the initial contact with the ground was made with the ball of the foot. Further, even if the spastic paretic subjects showed a strong tibialis anterior activation during the swing phase, their ankle remained in a plantarflexed position and did not show adaptation to uphill walking. The major limitation of this study is related to the fact that the authors compared healthy and spastic paretic subjects during uphill walking only. Comparisons of level and uphill walking for both groups of subjects would help to provide a detailed analysis of the adaptation strategy of healthy and spastic paretic subjects during inclined walking.

Findings obtained from animal and human studies tend to show that adaptation to inclined walking is achieved by changing the pattern of lower limb motion and by varying the level of activation of the relevant flexor and extensor muscles. This assumption is, however, very limited and do not take into consideration the importance of upper portions of the body in adaptation strategies. Inclined walking may induce important adaptations in pelvic and trunk motions. Meanwhile, this issue was addressed in only one study (Wall et al., 1981) that looked exclusively at pelvic movement in transverse plane. Preliminary observations from our laboratory suggest important adaptations of the pelvis and trunk during inclined walking. It was observed that during uphill walking, the pelvis and the trunk leaned forward. Increase in pelvic and trunk flexion would be performed to move the center of gravity forward in order to counteract the effect of slope. Increase in pelvic and trunk rotations were also observed in SCI subjects during uphill walking. Accentuation of the rotations would serve to assist

the swinging leg in its forward movement. Further studies will be required to understand the contribution of the pelvis and trunk in the adaptation of the gait pattern to inclined walking.

Control of pelvic and trunk movements during normal gait

The control of the pelvis and trunk is important to achieve smooth locomotion and maintain body equilibrium during gait. The pelvis and trunk, which represent a large portion of the body mass, have to interact with the lower limbs to accomplish efficient gait. In the six primary determinants of gait identified by Saunders et al. (1953), two were related to pelvic movement: 1) rotation of the pelvis in the transverse plane; 2) tilt of the pelvis in the frontal plane. These pelvic movements are performed to decrease the displacement of the center of gravity in each plane and to smooth the gait.

During walking, the pelvis and trunk move in the three anatomical planes and three-dimensional analyses reveal that trunk motion is dependent upon that of the pelvis (Stokes et al., 1989; Thorstensson et al., 1984). In fact, the pelvis and trunk act together or against each other to maintain body equilibrium and achieve smooth locomotion. In the sagittal plane, the pelvis and trunk shows similar patterns of angular displacements during gait which consisted of two full oscillations per step cycle. The peak flexion of the pelvis and trunk is reached near each heel contact (during the double-support phase) and the maximum extension occurs during single limb support (Krebs et al. 1992; Stokes et al., 1989). The trunk and pelvic total range of motion (TRM) is very low and varied from 2 to 8 degrees in the sagittal plane (Krebs et al. 1992; Murray et al., 1971; Stokes et al., 1989; Thorstensson et al., 1982; Thurston, 1985; Winter 1991). Stokes et al. (1989) reported a time delay

between pelvic and trunk angular displacement. According to their results, the pelvis reaches flexion peak 5 to 10% (of the gait cycle) earlier than the trunk.

Frontal plane views show that the pelvis and trunk reach their maximum positions after toe-off and their movements are in opposite directions: the pelvis leans toward the swinging limb while the trunk moves toward the supporting limb (Krebs et al. 1992; Stokes et al., 1989). During the swing phase, a reversal occurs and the trunk starts to lean toward the swinging limb whereas the pelvis begins to move toward the opposite limb (Krebs et al. 1992; Stokes et al., 1989). The pelvic lateral tilt occurring at the beginning of the swing phase is performed to lower the vertical displacement of the center of gravity and decrease energy consumption during walking (Saunders et al., 1953). The TRM of the pelvis (9°) in the frontal plane is almost twice that of the trunk (5°) (Stokes et al., 1989).

Motions in the transverse plane reveal that the pelvis and trunk rotate in opposite directions during most of the gait cycle (Krebs et al; 1992; Stokes et al., 1989; Wagenaar and Beek, 1992). Pelvic rotation reaches a maximum amplitude at the end of the swing phase to lengthen the stride (Saunders et al., 1953). To counteract the rotational momentum of the body, the trunk rotates in the opposite direction to bring the ipsilateral shoulder posterior to the heel contact limb (Krebs et al., 1992). In moving out of phase with the pelvis, the trunk provides a damping effect on the pelvic rotation and contributes to smooth the gait (Strokes, 1989). As in the frontal plane, the TRM of the pelvis (9°) in the transverse plane is about twice that of the trunk (4,7°) (Stokes et al., 1989).

The function of muscles controlling trunk movements during walking has been addressed in very few studies (Carlson et al., 1988; Thorstensson et al., 1982; Waters and Morris, 1972; Winter, 1991). In a study investigating the function of specific muscle groups of the back and abdomen, Waters and Morris (1972) showed that bursts of EMG activity from both muscle groups were closely related to the periods of heel contact. However, the internus (OI) and externus (OE) portions of the oblique abdominal muscles showed phasic activity throughout the walking cycle. Function of the oblique muscles seems to be task dependent. Using different oscillatory movements of the pelvis and trunk to study the control of intrinsic trunk muscles in human, Oddson and Thorstensson (1990) showed that OI and OE muscles are responsible for pelvic rotations and involved in pelvic tilts, spine lateral bending and flexion-extension of the trunk. During walking, Winter (1991) showed that OE had a moderate level of activity with two distinct peaks, a higher one at ipsilateral heel contact and a smaller one at contralateral heel contact. According to MacKinnon and Winter (1993), following the right heel contact there is a left lateral bending moment (at L4 level) and the right OE is used to counteract this left lateral lean. The role of OE in normal gait appears to be a trunk stabilisator in the frontal plane rather than a rotator as seen previously in oscillatory movements of the trunk (Oddson and Thorstensson 1990).

Concerning trunk muscular moments, the principal components occur about the bilateral axis (sagittal plane) during walking (Cappozo, 1983). Further, the forces and moments exchanged in the trunk during walking are almost exclusively generated by the trunk extensor muscles, the mechanical action of the trunk flexors being negligible (Cappozo, 1983). The trunk extensor muscles would create a peak force during or just before heel contact to counteract the trunk flexion occurring at this period

of the gait cycle. According to Waters and Morris (1972), the erector spinae (ES) muscle group (multifidus and rotatores) is responsible to counteract the trunk flexion during the heel contact period. However, others (Carlson et al., 1988; Thorstensson et al., 1982) attributed different function to the ES muscles. Thorstensson et al. (1982) and Carlson et al. (1988) observed that ES muscle activity occurred during backward movement of the trunk and, then, stipulated that these muscles were not involved in the deceleration of the trunk flexion following heel contact. Divergence in the interpretation of the lumbar back muscle function during walking seems to be related to the differences in the methodology used in each study. To measure trunk angular displacement, Thorstensson et al. (1982) and Carlson et al. (1988) used the same procedure: the trunk movements were measured relative to the third lumbar vertebra only and not relative to a calibration structure (Krebs et al., 1992; Stokes et al., 1989) which can provide trunk angular displacement in regard of the room coordinates. Due to the fact that the third lumbar vertebra is moving in the three anatomical planes, it becomes difficult to calculate the true displacement of the trunk during walking. In regard of this frame of reference, Thorstensson et al. (1982) and Carlson et al. (1988) observed that the trunk was moving backward during heel contact period and claimed that ES muscles were not used to brake trunk flexion but to counteract movements in the frontal plane. This is inconsistent with other studies (Krebs et al., 1992; Stokes et al., 1989) which showed that peak flexion of the trunk is closely related to the heel contact period. In regard of the results obtained from different studies (Cappozo, 1983; Krebs et al., 1992; Stokes et al., 1989; Waters and Morris, 1972; Winter, 1991) it is clear that one important function of the ES muscles is to decelerate the trunk flexion at the swing to stance transition.

Other functions have been attributed to the ES muscle group. According to Waters and Morris (1972), the ES muscle group would stabilize the trunk in the frontal plane. Findings from Cappozo (1983) also support the possibility of ES involvement in the control of the trunk in frontal plane motions. Cappozo (1983) found that trunk moment components in the frontal plane are important and the magnitude represents about 66% of that of the sagittal plane component. Muscular moments about the frontal plane have to be exerted by the trunk lateral benders. Since ES can perform both extension and lateral bending of the trunk (Carlson et al., 1988), this muscle group could play a role in the control of the trunk in the frontal plane. ES and OE muscles would then act together to stabilize the trunk in the frontal plane during walking.

Control of pelvic and trunk movements during pathological gait

During normal gait, as previously seen, the main functions of the pelvis and trunk are to maintain body equilibrium in order to achieve smooth locomotion. During pathological gait, however, the pelvis and trunk could perform additional functions to compensate for deficits of lower limb movements. Murray and colleagues (1971) studied the walking patterns of 28 subjects with unilateral hip pain due to osteoarthrisis and avascular necrosis of the femoral head and found a relationship between restriction of the hip motion and abnormal movements of the pelvis and trunk. To compensate for limited hip motion, the subjects accentuated pelvic movements in the sagittal plane. The mean maximum pelvic motion was 11° in the group of subjects with limited hip motion and only 4° in the control group. Analyses of pelvic motion revealed that anterior tilting was accentuated during the stance phase to assist hip extension on the painful side while increase in posterior tilting

served to initiate the swing phase and help the swinging painful leg to move forward. Similar results were reported later in the same population of subjects (Thurston, 1985).

Motions in the transverse plane showed high level of variability in both pelvic and trunk rotations in subjects with hip pain (Murray et al., 1971; Thurston, 1985). The pattern of pelvic rotation varied from a marked reduction of pelvic movement to a slightly greater rotation during single-limb support on the painful side (Murray et al., 1971; Thurston, 1985). The average pelvic rotation was found to be slightly greater in subjects with hip pain (11°) than in the healthy subjects (9°) (Murray et al., 1971). Opposite results were found for the trunk where the average rotation was shown to be lower in subjects with hip pain as compared to the unaffected subjects (Murray et al., 1971). However, Thurston (1985) reported that some subjects with hip pain performed a sudden rotation of the lumbar spine during the swing phase on the painful side to help the swinging leg to move forward.

As previously shown for subjects with hip pain, subjects with hemiplegia showed high level of variability for pelvic and trunk motions in the transverse plane (Wagenaar and Beek, 1992). These authors demonstrated three types of adaptations: (1) excessive pelvic and trunk rotations, (2) lack of pelvic and trunk rotations, and (3) different timing of trunk rotation. The average of the TRM revealed that, contrary to the subjects with hip pain and to the control subjects where the pelvic rotation was much greater than the trunk rotation, the subjects with hemiplegia had greater trunk rotation as compared to the pelvic motions (Wagenaar and Beek, 1992). These results suggest that accentuated trunk rotation could be an important mechanism to compensate for walking deficits in subjects with hemiplegia.

Motions in the frontal plane also revealed the presence of compensatory mechanisms from the pelvis and trunk in subjects with hip pain during walking. Two studies (Murray et al. 1971; Thurston, 1985) showed that during the swing phase of the painful hip, the pelvis was abnormally elevated toward the swinging leg. This elevation compensates for limited hip motion and provides a toe-floor clearance similar to normal walking. An important lateral shift of the trunk toward the weight bearing side during stance on the painful side was also observed in the frontal plane (Murray et al., 1971). In moving the center of gravity closer to the support line, this shift would decrease forces required from the hip abductor muscles on the painful side (Murray et al., 1971).

As previously mentioned, the contribution of pelvic and trunk movements during gait in SCI subjects has never been reported in the literature. However, clinical observations (videotaped trials) taken from 7 SCI subjects suggest an important contribution of the pelvis and trunk during walking. Further, these compensatory mechanisms seem to be severity dependent. The more severely affected subjects elevated the pelvis ("hip hiking") on the unsupported side during the swing phase while the less severely affected subjects showed near normal excursion of the pelvis in the frontal plane. Excessive pelvic tilt seen in some SCI subjects during the swing phase was also reported in subjects with hip pain due to osteoarthrisis (Murray et al., 1971; Thurston, 1985). Elevation of the pelvis in SCI subjects appears to be associated with an accentuation of the pelvic and trunk rotations in the transverse plane. Lateral pelvic tilt would be used to counteract the lack of dorsiflexion seen SCI subjects during the swing phase and would provide adequate toe-floor clearance. On the other hand, accentuation of the pelvis and trunk rotations would served to compensate for push-off weakness and help the swinging limb to move forward during the swing phase.

Adaptation of pelvic and trunk movements to inclined walking

The effects of inclines upon pelvic movements have been studied in healthy subjects (Wall et al., 1981). However, this study only focused on pelvic rotations. It was found that pelvic rotations remained relatively constant through the different walking slopes (from -20 to 20%).

Adaptation of pelvic and trunk movements to modifications in walking speed

Changes in walking speed lead to adaptations of pelvic and trunk movements (Waters and Morris, 1972; Stokes et al., 1989; Thorstensson et al., 1984; Wagenaar and Beek, 1992). As previously found for lower limb angular excursions (Nilsson et al., 1985), increase in walking speed also lead to larger amplitude of pelvic motions. Increase in pelvic range of motions was found to be more pronounced in the transverse plane (Stokes et al., 1989). Furthermore, it was shown that when speed increased beyond the comfortable walking speed, the TRM of pelvic rotation increased markedly and contributed to lengthening of stride (Stokes et al., 1989; Wagenaar and Beek, 1992).

At the trunk level, however, the range of motions was less speed-dependent. When increasing speed, the amplitude of trunk motions showed only a small increase in the frontal plane and no changes occurred in sagittal and transverse plane motions (Stokes et al., 1989). Trunk muscle activity also changed with speed. Increase in amplitude and duration of EMG activity in both anterior and posterior muscles of the trunk were reported with increasing speed (Waters and Morris, 1972; Thorstensson et al., 1982). During fast walking, as seen in comfortable walking, bursts of EMG activity of trunk muscles were associated with each heel contact (Waters and Morris, 1972; Thorstensson et al., 1982). According to the results obtained from trunk kinematics, where only

small changes occurred with increasing speed, it can be presumed that increase in EMG amplitude and duration of both anterior and posterior trunk muscles would serve to stabilize the trunk in the three plane of movements during fast-speed walking.

Introductory remarks on research papers

This thesis was comprised of three research papers. This first research project was specifically designed to assess the adaptability of the locomotor pattern to uphill walking following SCI. The objective of this study was to determine if SCI subjects could adapt to uphill walking and to compare this potential adaptation to the one observed in healthy subjects. The second research paper was designed to study the postural adaptation to uphill and downhill walking in healthy subjects. It was hypothesized that the pelvis and trunk would be tightly controlled during standing but not walking on slopes. It was expected that walking uphill would lead to a progressive, forward inclination of the trunk and pelvis with the increase in slope and vice versa during downhill walking. The third research paper was designed to study the postural adaptation to uphill and downhill walking in SCI subjects. The general hypothesis was that SCI subjects would use to a greater extent the pelvis and trunk segments to perform level and inclined walking as compared to healthy subjects.

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RESEARCH PAPER 1:

ADAPTATION OF THE WALKING PATTERN TO UPHILL WALKING IN NORMAL AND SPINAL CORD INJURED SUBJECTS

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ABSTRACT

Lower limb movements and muscle activity patterns were assessed from seven normal and seven ambulatory subjects with incomplete spinal cord injury (SCI) during level and uphill treadmill walking (5, 10 and 15°). Increasing the treadmill grade from 0 to 15° induced an increasingly flexed posture of the hip, knee and ankle during initial contact in all normal subjects, resulting in a larger excursion throughout stance. This adaptation process actually began in mid swing with a graded increase of hip flexion and ankle dorsiflexion as well as a gradual decrease of knee extension. In subjects with SCI, a similar trend was found at the hip joint for both swing and stance phases whereas the knee angle showed very limited changes and the ankle angle showed large variations with grade throughout the walking cycle. A distinct coordination pattern between the hip and knee was observed in normal subjects but not in subjects with SCI during level walking. The same coordination pattern was preserved in all normal subjects during uphill walking. The duration of electromyographic (EMG) activity of thigh muscles was progressively increased during uphill walking whereas no significant changes occurred in leg muscles. In subjects with SCI, EMG durations of both thigh and leg muscles, which were already active throughout stance during level walking, were not significantly affected by uphill walking. The peak amplitude of EMG activity of vastus lateralis, medial hamstrings, soleus, medial gastrocnemius and tibialis anterior was progressively increased during uphill walking in normal subjects. In subjects with SCI, the peak amplitude of EMG activity of the medial hamstrings was adapted in a similar fashion whereas the vastus lateralis, soleus and medial gastrocnemius showed very limited adaptation during uphill walking. We conclude that subjects with

SCI can adapt to uphill treadmill walking within certain limits, but they use different strategies to adapt to the changing locomotor demands.

Key words: human walking, uphill, spinal cord injury, adaptation, EMG, kinematics.

INTRODUCTION

The human locomotor pattern can be adapted easily and rapidly to modifications in external demands such as speed (Brandell, 1977; Murray et al., 1966; Nilsson et al. 1985) and slope (Brandell, 1977; Lange et al., 1996; Simonsen et al. 1995; Wall et al. 1981). This adaptation is normally achieved by changing the pattern of lower limb motions (Lange et al., 1996; Murray et al., 1966; Nilsson et al. 1985; Wall et al. 1981) and by varying the motor recruitment of the relevant flexor and extensor muscles (Brandell, 1977; Lange et al., 1996; Nilsson et al. 1985; Simonsen et al. 1995). Following a spinal cord injury, descending pathways are usually damaged affecting the adaptability of the locomotor pattern. Animal studies have shown that cats with complete spinal cord transection can recover locomotion on the treadmill and adapt to changes in speed (Barbeau and Rossignol 1987; Bélanger et al., 1989; Forssberg et al. 1980). However, the spinal cat has difficulty in adapting its walking pattern to changes in inclined planes (Bélanger et al., 1988). For example, contrary to the results obtained in the intact cat, only minor changes in the electromyographic (EMG) activity are observed during uphill (15°) and downhill (20°) walking in the spinal animal (Bélanger et al. 1988). Moreover, without manually supporting the spinal cat, the animal could only adapt to a few degrees of roll, as compared to at least 20° observed in the intact cat (Bélanger et al. 1988).

In humans, uphill walking is a demanding task that requires specific modifications in lower limb movements (Lange et al., 1996; Wall et al. 1981) and muscle activation (Brandell, 1977; Lange et al., 1996; Simonsen et al. 1995). As the walking grade increases, more propulsion has to be generated from the lower limbs (Brandell, 1977) and postural adjustments have to be performed to maintain equilibrium (Kawamura et al., 1991). Following spinal cord injury (SCI), the basic

locomotor pattern is altered (Barbeau et al., 1998) and the ability to adapt to uphill walking could be affected. However, it remains unknown if persons with SCI can adapt to uphill walking. This study is a first effort to assess the adaptability of the locomotor pattern to uphill walking following SCI.

METHODOLOGY

Subjects

Seven normal subjects and seven subjects with SCI (see Table 1) took part in the experiment. The SCI group included six males and one female who were free of any comorbid medical complications. The normal subjects were males and aged between 28 and 40 years. The study was approved by the ethics committee of the School of Physical and Occupational Therapy of McGill University and informed consent was obtained from all subjects prior to the experiment.

Walking assessment

The subjects walked on a custom designed treadmill (see Norman et al., 1995) at four different grades (0, 5, 10 and 15°). To standardize the walking conditions, each subject had to grasp the handrails during walking grades. During level gait, this procedure does not alter the sagittal plane kinematics of normal walking (Siler et al., 1997). The walking grades were presented in random order. Subjects with SCI performed all walking conditions at their self-selected comfortable walking speed while normal subjects walked at 0.4 and 1.0 m/s. Subjects habituated themselves to the different grades of treadmill walking before the experimental sessions began. Data were collected from the right side for normal subjects and from the more affected side for subjects with SCI. No

major asymmetry was found in any of the subjects with SCI. The more affected limb was chosen based on muscle strength testing.

Pressure sensitive footswitches were placed bilaterally on the soles of the subjects' shoes to signal the heel, fifth metatarsal, and hallux contacts. The footswitch signals were used to measure stance, swing and cycle durations. Using preamplified surface electrodes, EMG activities were recorded from the vastus lateralis (VL), medial hamstrings (MH), soleus (SOL), medial gastrocnemius (MG) and tibialis anterior (TA). The EMG signals were amplified, bandpass filtered at 10-1000 Hz and recorded on FM tape (Honeywell). The movements of the lower limb were recorded with two videocameras (Panasonic VHS VTRs) operating at a frequency of 60 Hz and located at about 90° from each other. Reflective markers were attached over the fifth metatarsal, the heel, the lateral malleolus, the tibial plateau, the greater trochanter of the femur and the greater tuberosity of the humerus. The 3-D coordinates of each marker were obtained from the two camera views. To synchronize EMG signal and joint angular displacement records, a SMPTE time code signal was recorded on both FM and videotapes.

Data analysis

Five consecutive walking cycles were selected for the analysis of the kinematic and EMG signals. The Peak Performance system was used to digitize the X, Y coordinates of the reflective markers, which were then used to calculate the relative angular excursions of the hip, knee and ankle joints in the sagittal plane. Kinematic analysis of the hip, knee and ankle joints was done in the sagittal plane, using conventional definitions (Winter, 1991). Briefly, the hip and knee angles were

calculated with respect to the proximal segment, with the neutral position in standing being taken as 0° displacement, flexion being positive, and extension negative. Likewise, in calculating the ankle angle, the neutral standing position with the leg axis perpendicular to the foot was taken as 0°. Ankle dorsiflexion beyond neutral was taken as positive angular displacement, and ankle plantar flexion beyond neutral was taken as negative angular displacement. The footswitch and EMG signals were digitized (1.2 KHz sampling) using a custom designed program. The EMG signals were later digitally bandpassed (20-400 Hz), full-wave rectified and smoothed using a moving window average of 38.4 ms (Data Pac II program from Run Technologies). Twenty-three points from each side of the center were included in the moving window average.

For each subject, five walking cycles were averaged to obtain individual means of angular excursions and EMG patterns across the different walking grades. The amplitude of the averaged EMG profile was normalized with respect to the peak activity of each muscle occurring during level (0°) walking. An interactive computer program was used to determine the onset and offset of EMG activity. A two-way ANOVA with repeated measures was used to test for differences in EMG peak amplitude, EMG burst duration and kinematic parameters between groups and across grades. Due to the fact that only 3 out of 7 subjects with SCI performed the 15° grade, this grade was not included in the statistical tests. When significant differences were found, pairwise comparisons were made using Student-Newman-Keuls tests. Statistical significance was set at $p = 0.05$.

RESULTS

Angular excursions of lower limbs

Figure 1 shows the angular excursions of the hip, knee and ankle in normal subjects and 2 subjects with SCI. For the normal subjects, each trace represents an average from all 7 subjects. The gray area represents the 95% confidence interval for the level condition. It can be seen that the variations in lower limb angular displacements are small between normal subjects and within the same subject with SCI. The normal subjects showed marked modulation in joint angular displacements during uphill walking. At the hip joint, the range of flexion increased during swing to lift the leg up the slope, resulting in a larger excursion during stance. At the knee joint, the range of extension decreased during late swing such that foot contact was made with increased knee flexion, and the increasingly flexed posture was followed by a progressively larger knee excursion during stance as the slope increased. For the ankle joint, the range of dorsiflexion increased during swing to lift the ankle up the slope, and remained elevated during stance. However, angular displacements of the hip and knee were invariant around stance-swing transition across all grades in every normal subject. This indicates that the hip and knee position at this period of the walking cycle was kept constant across grades. This point of invariance was also observed for the hip-knee coordination pattern in normal subjects (see phase 3 on Figure 3). The height of the greater trochanter during swing, as shown on top of Figure 1, showed only minor changes, as the excursion varied from 1.64 cm at 0° grade to 5.36 cm at 15° grade. In subjects with SCI, however, lower limb motions did not show consistent patterns of adaptation as in normal subjects. Rather, a spectrum of adaptation was found and ranged from a near normal pattern of adaptation to one in which only the hip angle was modified. The subjects with SCI who showed near normal walking patterns during level walking had a near normal adaptation to

uphill walking. Two subjects with SCI, who represent the opposite of the spectrum, are shown on Figure 1. Subject AN performed all uphill conditions whereas PO was unable to achieve the most elevated treadmill grade. Furthermore, as observed in normal subjects, the adaptation occurred at the hip, knee and ankle joints in AN during uphill walking. In contrast, PO showed little adaptation in all three lower limb joints except for a small increase in hip flexion during late swing and early stance. Instead, a different compensation mechanism involving hip hiking was used in all walking conditions, as shown by the vertical trajectory of the greater trochanter (Figure 1, top inset).

To determine lower limb movement adaptation to grade, joint angles at foot contact were averaged for normal subjects walking at slow (0.4 m/s) and comfortable (1.0 m/s) speeds as well as SCI subjects walking at self-selected speed (Figure 2). Significant main effects due to grade were found at the hip, knee and ankle joints ($p < 0.01$). Post-hoc comparisons revealed that hip, knee and ankle flexion at foot contact was significantly larger as grade increased in normal subjects walking at either speeds but not SCI subjects. A significant main effect due to group (normal vs SCI) was found only at the knee joint. Knee flexion at level walking was significantly larger ($p < 0.01$) for the SCI group.

Results from each individual subject with SCI revealed that a clear adaptation pattern occurred only at the hip (Figure 2B). As found in normal subjects, every subject with SCI showed a gradual increase in hip flexion as the treadmill grade increased. Contrary to the normal subjects, every subject with SCI made foot contact with the knee in flexion during level walking (Figure 2B). Furthermore, except for one subject (LT), the changes in flexion at the knee were limited or absent

when adapting to uphill walking. Ankle joint adaptation was most variable in subjects with SCI. Instead of showing a gradual increase in ankle dorsiflexion as in normal subjects, some subjects with SCI (AN, CP, LT) showed both increase and decrease in ankle dorsiflexion at increasing treadmill grades.

Temporal distance data

Figure 1 shows that the percentage of the step cycle devoted to stance for the two subjects with SCI was greater than that for the normal subjects. This increase in the proportion of stance duration is naturally observed at low walking speeds even in normal subjects (Nilsson et al., 1985). Walking at different grades did not affect temporal distance variables in normal subjects and in subjects with SCI. No significant changes were found for the stride length, cycle duration and stance-swing ratio across the different walking grades.

Intralimb coordination

Figure 3 illustrates hip-knee cyclographs of normal subjects and subjects with SCI across different grades. In normal subjects, only one hip-knee coordination pattern was found. Thus, all data were averaged and pooled together. During level walking, the cyclograph curve was characterized by a slanted "S" shape in stance and by a parabolic shape during swing in normal subjects. When the treadmill grade increased, the starting point (phase 1), representing hip-knee position at foot contact, was progressively shifted toward the right upper region, indicating an increase in hip flexion and a decrease in knee extension. Following foot contact, there was a knee flexion (yield) associated with weight acceptance (phase 2). This knee yield was preserved but

became less prominent during uphill walking. From early to mid stance (phase 2 to 3), the hip and knee extended simultaneously. In late stance (push off, phase 3), the knee flexed rapidly while the hip reversed from extension to flexion. This point of transition was maintained constant across the different grades. The peak of knee flexion, reached in mid swing (phase 4), was similar across different grades, although a progressive increase in hip flexion was noted.

Contrary to the normal subjects, many different joint coordination patterns were present even for level walking in subjects with SCI (Figure 3). For example, in AN the knee was relatively flexed at the onset of foot contact during level walking (as it was in all subjects with SCI), and exhibited a very limited yield. The peak of knee flexion, which occurred during mid swing in normal subjects (see phase 4), was reached at the end of stance (similar in PR and LT). Furthermore, while the hip flexed near the end of the swing phase, knee extension was abnormally reversed into flexion. In general, some normal adaptation features were preserved which included a progressive shift in the starting hip-knee position and in the peak of knee flexion. Abnormal features included the presence of knee hyperextension, thus altering the transition between mid stance and push off, as well as an increase in peak knee flexion during mid swing.

In PO and PR, substantial differences were found in the coupling between the hip and knee movements during level walking. The knee yield (phase 2) was much larger than normal revealing problems with weight acceptance. Between mid stance and push off (phase 3), the knee flexed while the hip continued to extend giving rise to a prolonged jerky transition. The parabolic curve usually observed in normal subjects in phase 4 was markedly flattened. Very limited changes were observed

as grade increased, except for the emergence of a more normal hip-knee coupling in swing phase. In four of the other subjects with SCI (PR, CP, RM, FS), the hip-knee coordination patterns were changed in a similar fashion to that observed in normal subjects, at least during the swing phase.

Muscular activity

Uphill walking induced marked modulation in the amplitude of EMG activity in normal subjects (Figures 4 and 5). As the treadmill grade increased from 0 to 15°, there was a progressive increase in the amplitude of all recorded muscles. In subjects with SCI, however, the amplitude of EMG activity was minimally affected by uphill walking in most muscles as shown in subjects AN and PO (Figure 4; same subjects selected as examples on Figure 1). In AN, the EMG amplitude of thigh muscles (VL and MH) also increased with grade, although to a lesser extent than in normal subjects, whereas the leg muscles (SOL, MG and TA) showed very limited changes in amplitude during uphill walking. In PO, the EMG activity of both thigh and leg muscles was minimally affected by uphill walking. Furthermore, a complete absence of activity was seen in TA. In general, for the plantarflexor muscles (SOL and MG), a prolongation of the EMG activity and an absence of peak of activity during the push off period were found in all subjects with SCI.

To determine the effect of uphill walking on lower limb muscle activation, peak amplitude of thigh muscles during stance phase and leg muscles during push off period (SOL and MG) and swing phase (TA) were averaged for normal subjects at slow (0.4 m/s) and comfortable (1.0 m/s) speeds as well as SCI subjects walking at self-selected speed (Figure 5). Significant main effects due to group (normal vs SCI, $p < 0.05$) and grade ($p < 0.01$)) were found in VL, SOL and MG but not in MH and

TA. Post-hoc tests revealed that significant differences were present only between 0 and 10° grades for these three muscles. For the VL, the net amplitude change from 0 to 10° grade was significantly larger ($p < 0.01$) in the normal group walking at 0.4 m/s than in SCI group whereas for the SOL and MG, the changes from 0 to 10° grade were significantly larger ($p < 0.01$) in the normal group walking at 1.0 m/s than in SCI group. Thus, significant adaptation was observed mainly in the extensors.

Uphill walking induced modulation in the timing and duration of EMG activity in normal subjects (Figure 6). Increasing the treadmill grade from 0 to 15° led to a progressive increase in the duration of thigh muscles. This increase in duration was mainly attributed to a shift in the offset of EMG activity toward the end of the stance phase. This shift was also observed in the SOL and MG, although less pronounced. For the TA (not shown on Figure 6), uphill walking did not alter either the timing or the duration of EMG activity. In subjects with SCI, all muscles showed prolonged activation. Limited adaptation in terms of changes in timing and duration of EMG activities were observed in uphill walking. Significant main effects due to group (normal vs SCI) were found in all muscles ($p < 0.05$) except TA whereas significant main effects due to grade were found only in the MH ($p < 0.01$). Post-hoc tests revealed that significant differences were present between 0 and 10° grades. The net amplitude change from 0 to 10° grade was significantly larger ($p < 0.01$) in normal than in SCI group (31.3% vs 7.3%).

DISCUSSION

This study showed that subjects with SCI could adapt to uphill walking, albeit with certain limitations. Only 3 out of 7 subjects with SCI performed every walking grade while the rest reached

only the 10° grade. However, the strategies of adaptation were different from those of normal subjects.

Adaptation of lower limb movements

In all normal subjects, the strategy to adapt to uphill walking was to increase hip flexion and ankle dorsiflexion as well as to decrease knee extension of the forward limb from mid swing to the foot contact. Similar findings have been reported during uphill treadmill walking (Lange et al., 1996; Wall et al. 1981). However, in the SCI group, similar adaptation was found only at the hip joint. Major limitations were seen at the knee and at the ankle joints. Both neural and biomechanical factors could give rise to the limitations.

Pronounced knee flexion during foot contact and throughout stance even during level walking has been observed in subjects with spastic paresis of multiple origins (Conrad et al., 1985) and in cats with lesions of the dorsolateral funiculi and dorsal columns (Jiang and Drew 1996). This pronounced knee flexion may have been caused, in part, by hyperactivity in ankle extensors during early stance and also by weakness of knee extensors, as suggested by Winter (1991). When the knee is already in a flexed position, it becomes a biomechanical constraint such that it is more difficult to adapt to the new treadmill grade. Likewise, the limited adaptation observed at the ankle joint might be due to weakness or absence of dorsiflexor activity (as seen in subject PO, Figure 4). Meanwhile, pronounced knee flexion during foot contact can give rise to increased ankle dorsiflexion during uphill stance. Consequently, increased triceps surae activation is needed to counteract the flexed posture and generate adequate push off, but this could not be achieved by subjects with SCI.

Normal subjects use the same hip-knee coordination pattern in uphill walking as in level walking. The major adaptation feature is a progressive shift of the starting position which involves a progressive increase in hip flexion and a simultaneous decrease in knee extension towards the end of swing (see Figure 3). Contrary to the normal subjects, many different hip-knee coordination patterns were found in the subjects with SCI even during level walking. However, as found in the normal subjects, these patterns were preserved in adapting to uphill walking, at least during the swing phase. Alterations in the coupling between the hip and knee joints during locomotion have also been reported in cats with partial lesions of the spinal cord (Jiang and Drew 1996). In the present study, changes in the timing and magnitude of hip and knee angular movements at specific phases lead to modifications in the coupling between these two joints, and result in a variety of coordination patterns in subjects with SCI.

Adaptation of motor recruitment pattern

Uphill walking induced a progressive increase in the duration and amplitude of the EMG activity of thigh muscles. The increase in duration was caused by a shift in the offset of MH and VL activities toward the end of the stance phase, as seen in both cats (Smith and Carlson-Kuhta, 1995) and humans (Tokuhiro et al., 1985) during uphill walking. Such a motor adaptation may be required to counteract the increased hip and knee flexion in uphill stance and bring the whole limb back to an extended position before initiating swing. While a large increase in the amplitude of the SOL and MG is required to generate more propulsion during push off in uphill walking, the offset of these plantarflexors is minimally affected in normal subjects. This is not an unexpected finding because the

offset of plantarflexor activity is directly related to the toe off timing and that event is invariant across grades.

In subjects with SCI, all muscles showed prolonged activation throughout stance without much adaptation to changes in walking grades. This prolonged activation indicates an overall exertion possibly due to impaired motor recruitment even during level walking, and becomes a limiting factor in adapting to uphill walking. Premature activities were observed in plantarflexors of subjects with SCI during all walking grades. Abnormal activity in plantarflexors at the end of the swing phase and in early stance was previously described in subjects with spastic paresis and subjects with SCI (Conrad et al., 1985; Fung and Barbeau, 1989; Knutsson, 1985). This abnormal activity could arise from stretch activation of the triceps surae, which may be a result of defective gating mechanisms of Ia afferents (Fung and Barbeau, 1994; Yang et al., 1991). The presence of cocontraction patterns of all lower limb muscles observed in the present study has previously been reported in subjects with SCI (Fung and Barbeau, 1989). Cocontraction may be used to counteract muscle weakness and stabilize lower limb joints throughout the stance phase in subjects with SCI.

The lack of voluntary control over the leg muscles in subjects with SCI is undoubtedly due to the loss of descending pathways following spinal cord injury. The absence of increase in the level of activation of plantarflexors likely results in a weak push off and compensatory mechanisms from proximal segments might be used to adapt to uphill walking. For instance, the hip flexors can be recruited to pull the swinging limb upward and forward during uphill walking. Compensatory mechanisms from hip muscles have previously been identified in some gait-related pathologies (Olney

et al. 1988; Winter and Sienko 1988). Pelvic elevation on the unsupported side during the swing phase can partially compensate for weakness of paralysis of ankle dorsiflexors, and helps to provide adequate toe floor clearance (as shown in subject PO, Figure 1). Other emerging adaptation strategies may include accentuation of pelvic and trunk rotations during walking, as reported in patients with hemiplegia (Wagenaar and Beek 1992) or unilateral hip pain (Thurston 1982), to compensate for push off weakness and assist foot swing during uphill walking.

Impact due to level and extent of spinal cord lesion

Our SCI group comprised of 3 subjects (LT, PR, RM) with cervical injuries, 2 subjects (CP, FS) with thoracic injuries and 1 subject (PO) with both cervical and thoracic injuries (Table 1). The etiology of the injury was traumatic in these 6 subjects. Results obtained from angular excursions of lower limbs and intralimb coordination during graded walking revealed that important differences were present within each sub-group of injury level but not between the sub-groups. For instance, RM and FS, who belong to different injury level sub-groups, show very similar hip-knee coordination patterns when adapting to uphill walking (Figure 3). In contrast, RM and LT, who belong to the same sub-group, showed very distinct coordination patterns during both level and uphill walking. Krawetz and Nance (1996) also found no significant difference in lower limb joint excursions during overground walking in a much larger group (n=27) of spinal cord injured subjects with lesions at cervical, thoracic and lumbar levels. Moreover, the injury level did not affect either the amplitude or the timing and duration of lower limb EMG activity during level and inclined walking.

The injury level did not seem to influence the overall performance of the subjects with SCI. For the 3 subjects with cervical injuries, one was able to perform the most elevated treadmill grade whereas the two others were unable to go beyond the 10° grade. None of the 2 subjects with thoracic injuries achieved the steeper grade. Furthermore, the 2 subjects with SCI (PR, CP) who achieved the fastest walking speed on the treadmill belong to two different sub-groups. This is in contrast with Krawetz and Nance (1996) who, in a larger scale study, reported a reduced walking speed for subjects with thoracic lesions as compared to subjects with cervical lesions. Other factors such as spasticity, hypertonia and muscle weakness could also alter the overall performance of walking, and, thus, influence its adaptability.

The extent of the spinal cord lesion can also have an impact on locomotor adaptability. Following partial spinal cord lesions involving the dorsal columns and the dorsolateral funiculi cats can recover the ability to perform treadmill walking at different inclinations (both roll and tilt) without any support or assistance (Jiang and Drew, 1992). However, permanent deficits were seen in the performance of anticipatory adjustments during locomotion such as stepping over obstacles or walking through a horizontal ladder (Jiang and Drew, 1992; Jiang and Drew, 1993). More recently, it has been shown that after ventral and ventrolateral spinal lesions leading to extensive damage to reticulospinal and vestibulospinal pathways, cats had major difficulties to adapt to inclined walking (Brustein and Rossignol, 1998). The deficits observed during level walking, which included problems to maintain lateral stability, inconsistent stepping and occasional stumbling, were accentuated during inclined walking. The lack of hindlimb muscle adaptation during uphill and downhill locomotion was attributed to the loss of reticulospinal axons (Brustein and Rossignol, 1998). In the present study, the

extent of the spinal cord lesions could not be determined. However, none of our subjects with SCI had any major asymmetrical impairment. Clinically, all subjects with SCI were classified as "mildly impaired" (D) using the ASIA impairment scale (Ditunno et al., 1994), with motor and sensory functions preserved below the neurological level. Active movement against gravity can be ellicited from the majority of key muscles below the lesion. However, the ability to adapt to uphill walking is not correlated with the actual ASIA motor and sensory scores.

In general, the treadmill walking speed seems to be a good indicator of the capability of locomotor adaptation in subjects with SCI. Subjects who had faster maximal treadmill speed were the ones who achieved the steeper grades, except for CP who felt uneasy walking up-slope (for fear of falling). On the other hand, no single muscle or kinematic pattern could predict the capability in adapting to the steepest grade. The study of other factors such as the kinetics of incline walking and the contribution of pelvic and trunk segments would likely help to discriminate between the capabilities of adaptation in subjects with SCI.

CONCLUSION

The results of this study showed that all normal subjects used the same kinematic and motor strategies to adapt to uphill walking. However, in subjects with SCI, a variety of strategies exists. Some of the lower limb deficits seen after an incomplete spinal cord injury led to alterations in the basic walking pattern and changed the kinematic and motor strategies in adapting to uphill walking. For instance, the lack of changes in knee flexion angle, the large variations in ankle angle and the limited changes in the peak amplitude of plantarflexor muscles during uphill walking were three major

abnormal features observed in the subjects with SCI. In general, the capability to adapt to the steepest grade of locomotion cannot be pre-determined on the basis of the level of the injury, the clinical motor and sensory scores, the kinematic pattern or the EMG profile. In contrast, the maximum speed achieved during level treadmill walking was always a good indicator of the patient's ability to walk on an inclined plane.

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Table 1: Profile of the spinal cord injured subjects

Subjects	Age	Cause and Level of injury	Time since injury	Comfortable treadmill speed (m/s)	Maximal treadmill speed (m/s)	Treadmill grades performed (°)
AN	24	Spastic paresis, non familial origin	> 14 years	0.40	0.80	0 to 15
CP	46	MVA*, T11	14 months	0.70	1.10	0 to 10
FS	25	MVA*, T8-T9	> 6 years	0.30	0.60	0 to 10
LT	20	Diving, C5	2.8 years	0.60	1.00	0 to 15
PO	16	MVA*, C7-T1	10 months	0.40	0.70	0 to 10
PR	20	MVA*, C6	> 5 years	0.80	1.30	0 to 15
RM	38	Diving, C6	> 20 years	0.30	0.60	0 to 10

*MVA = Motor vehicle accident

LEGENDS

Figure 1: Angular excursions of the hip, knee and ankle during uphill walking normalized to the gait cycle (from one foot contact -0% to the next -100%). The trajectory of the greater trochanter is shown on top. This figure shows the average of the 7 normal subjects and two subjects with SCI (each line represents the average of 5 cycles). The two subjects with SCI illustrate the two extremes in the spectrum of adaptability to graded walking. The gray area represents the 95% confidence interval of the level condition for the normal group and each of the two subjects with SCI.

Figure 2: Average hip, knee and ankle angles at foot contact during level and uphill walking for: (A) group of normal subjects (n = 7); (B) individual subjects with SCI. Gray line represents the grand average of the 7 subjects with SCI.

Figure 3: Cyclographs showing the coupling between the hip and knee joints during uphill walking. This figure shows the average of the 7 normal subjects and the 7 individual subjects with SCI rank-ordered by their comfortable walking speed. The solid lines represent the stance phase and the dotted lines represent the swing phase. The arrows indicate the direction of the movement. On the graph showing the normal subjects, ① indicates foot contact, ② corresponds to early stance, ③ represents the transition from mid stance to push off and ④ indicates mid swing. Displacement of the curves diagonally up and right indicates a simultaneous increase of knee and hip flexion.

Figure 4: EMG linear envelope of lower limb muslces during uphill walking normalized to the gait cycle (from one foot contact -0% to the next -100%). Amplitude of EMG activity was normalized and expressed as a percentage of the level condition. The average of the 7 normal subjects and two subjects with SCI (each line represents the average of 5 cycles) are presented. The gray area represents the 95% confidence interval of the level condition for the normal group and each of the two subjects with SCI.

Figure 5: Average peak amplitude (mean ± 1 SD) of thigh muscles during stance phase (A) and leg muscles during push off (B) for the normal group walking at 1.0 and 0.4 m/s and the SCI group walking at self-selected speed during uphill conditions. Amplitude of EMG activity was normalized and expressed as a percentage of the level condition.

Figure 6: Average timing and duration of EMG activity of thigh and leg muscles for the normal (A) and the SCI (B) groups during walking at 0, 5, 10 and 15°.

Figure 1

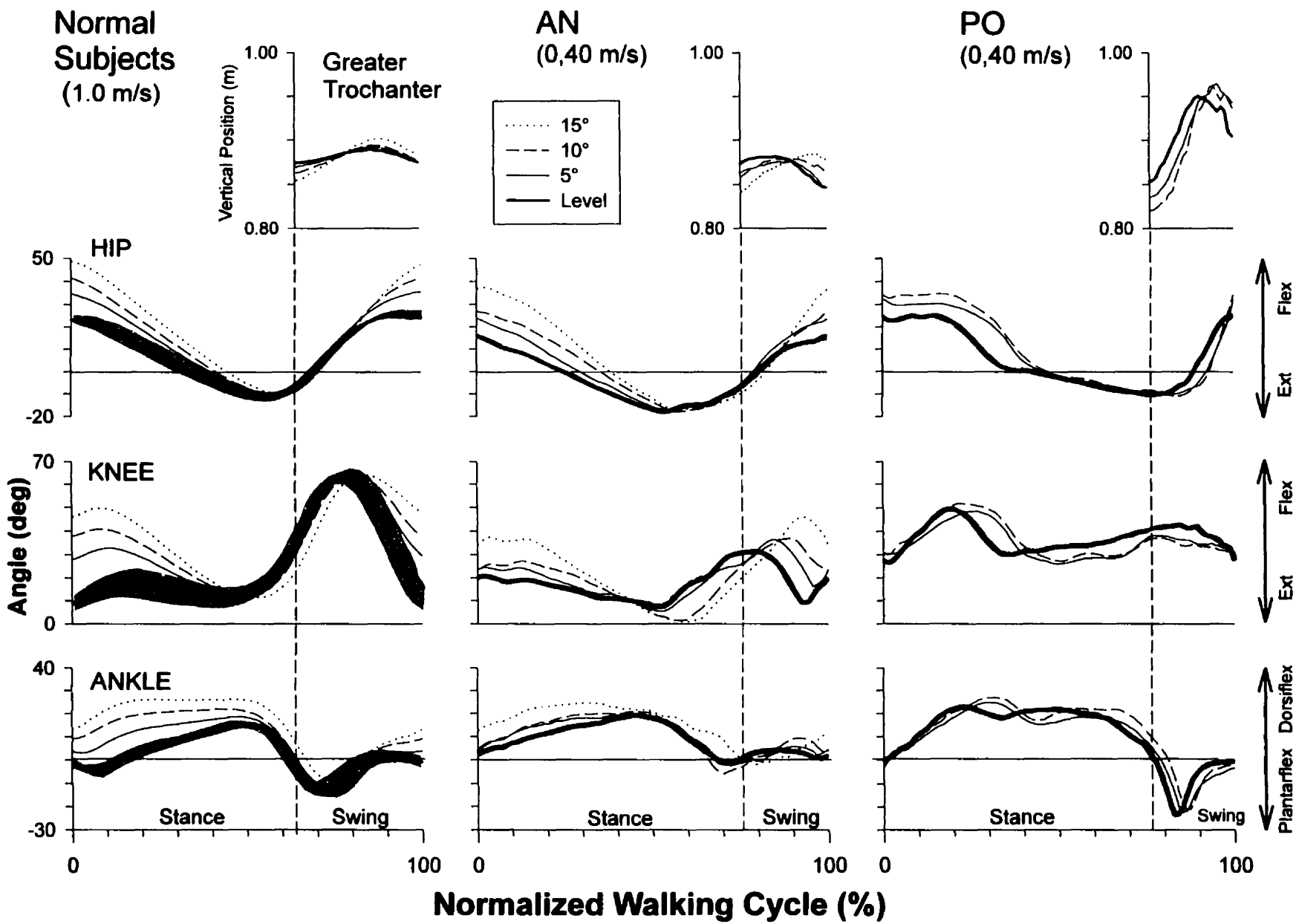

Normal Subjects (1.0 m/s)
AN (0,40 m/s)
PO (0,40 m/s)
Greater Trochanter
Vertical Position (m)
1.00
0.80
15°
10°
5°
Level
HIP
KNEE
ANKLE
Angle (deg)
50
-20
70
0
40
-30
Flex
Ext
Dorsiflex
Plantarflex
Stance
Swing
0
100
Normalized Walking Cycle (%)

Figure 2

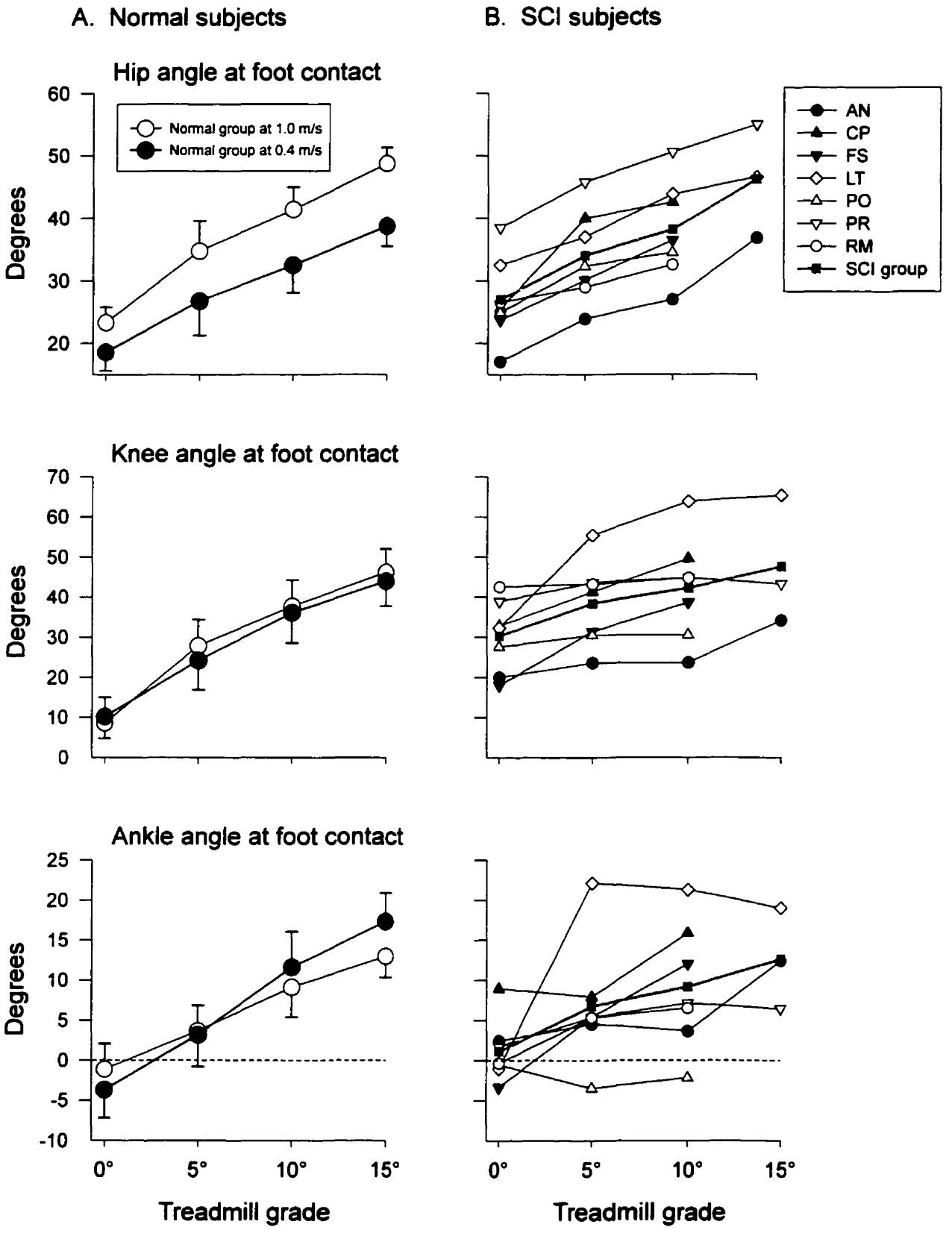

A. Normal subjects
B. SCI subjects
Hip angle at foot contact
Normal group at 1.0 m/s
Normal group at 0.4 m/s
AN
CP
FS
LT
PO
PR
RM
SCI group
Knee angle at foot contact
Ankle angle at foot contact
Degrees
0°
5°
10°
15°
Treadmill grade

Figure 3

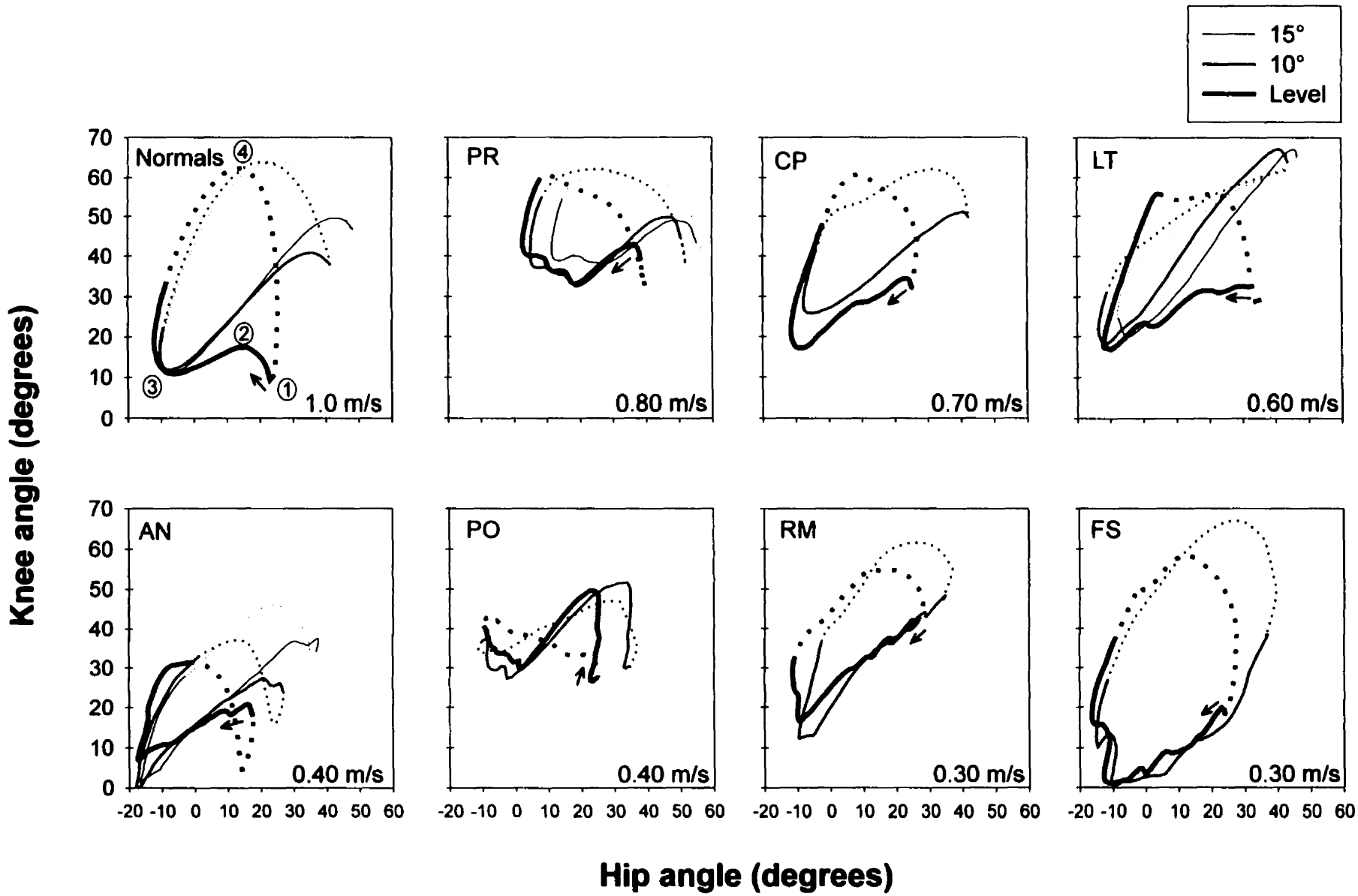

15°
10°
Level
Normals
PR
CP
LT
AN
PO
RM
FS
1.0 m/s
0.80 m/s
0.70 m/s
0.60 m/s
0.40 m/s
0.40 m/s
0.30 m/s
0.30 m/s
Knee angle (degrees)
Hip angle (degrees)

Figure 4

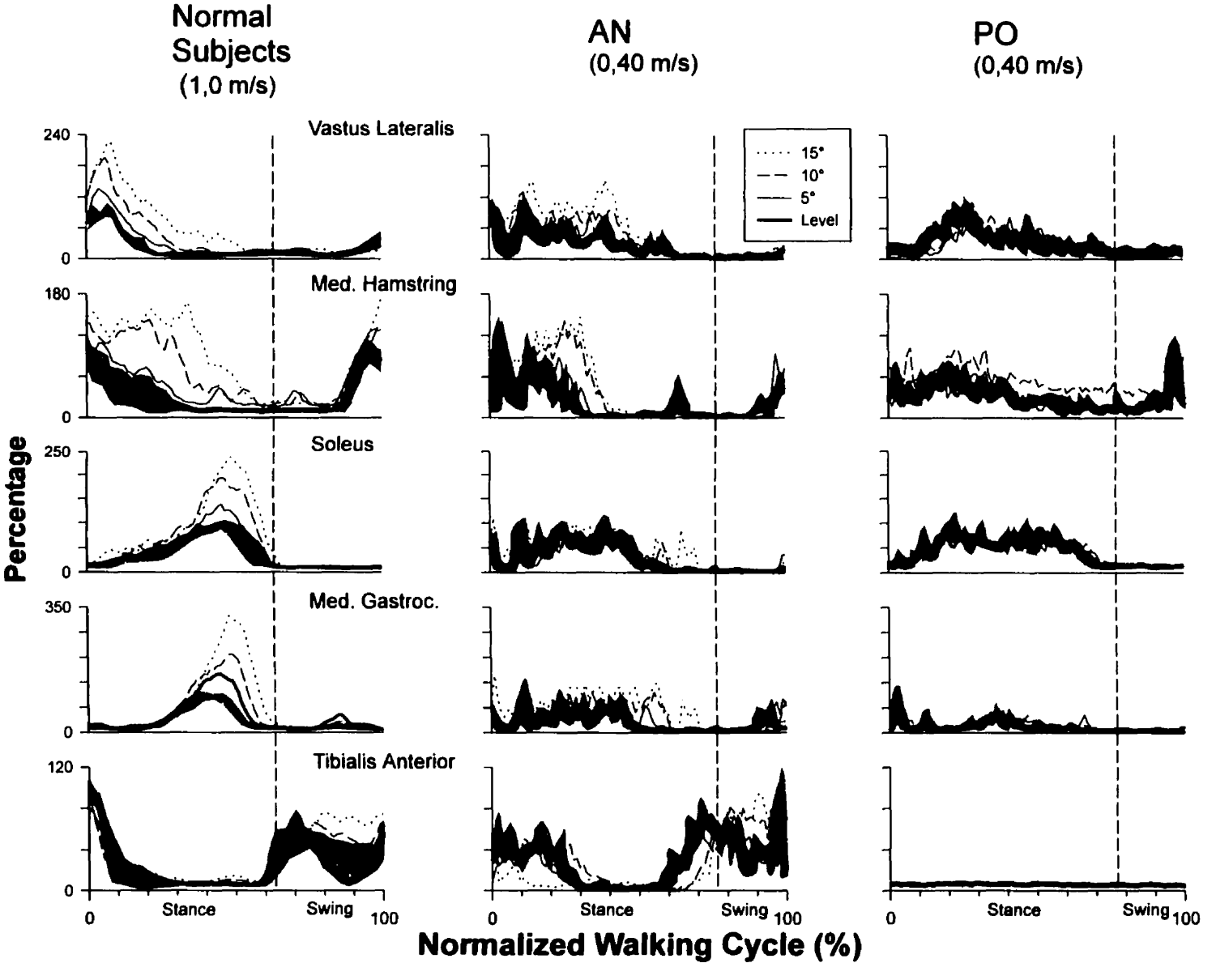

Normal Subjects (1,0 m/s)
AN (0,40 m/s)
PO (0,40 m/s)
Vastus Lateralis
Med. Hamstring
Soleus
Med. Gastroc.
Tibialis Anterior
15°
10°
5°
Level
240
0
180
0
250
0
350
0
120
0
Percentage
0
Stance
Swing
100
Normalized Walking Cycle (%)

Figure 5

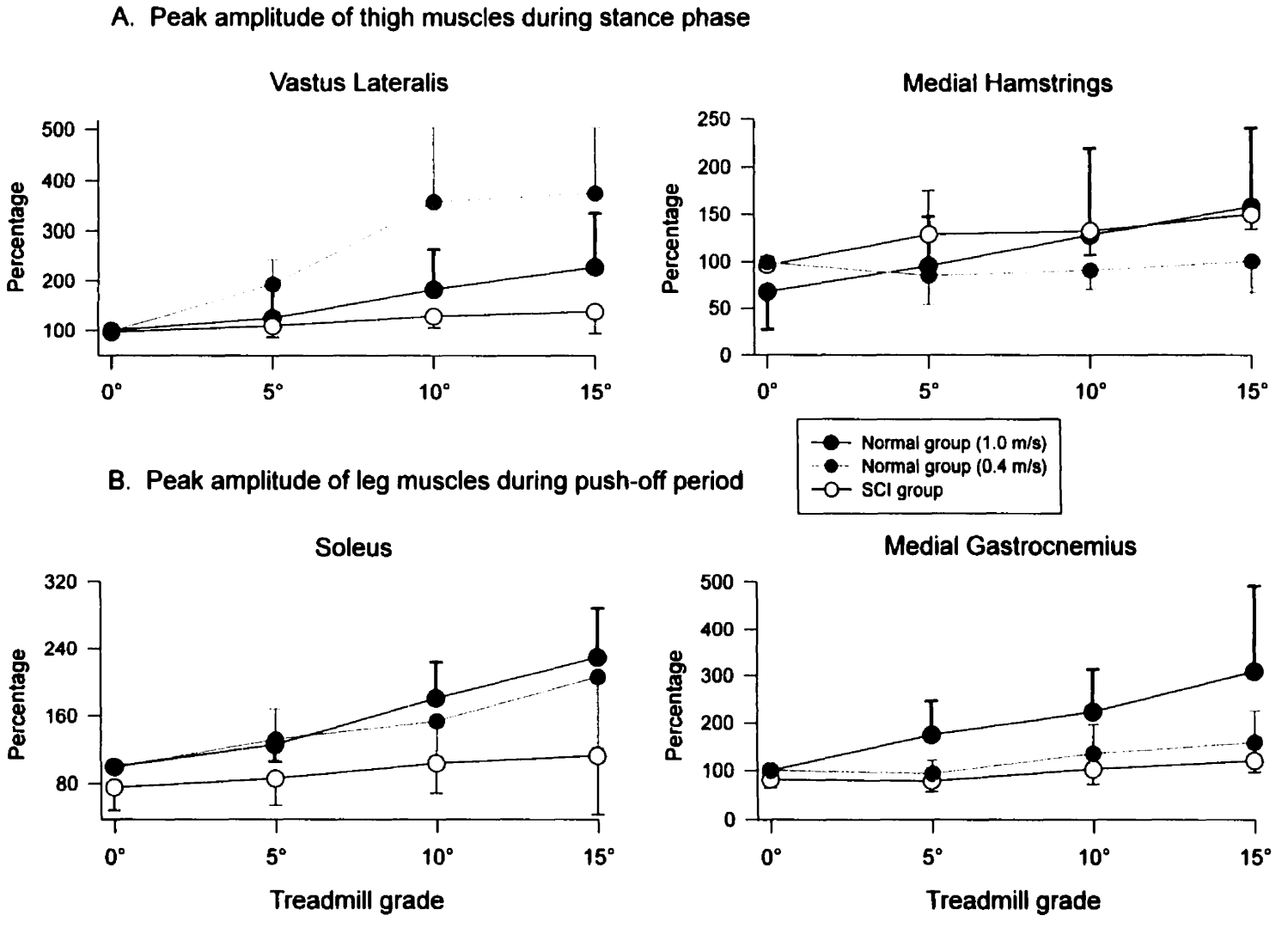
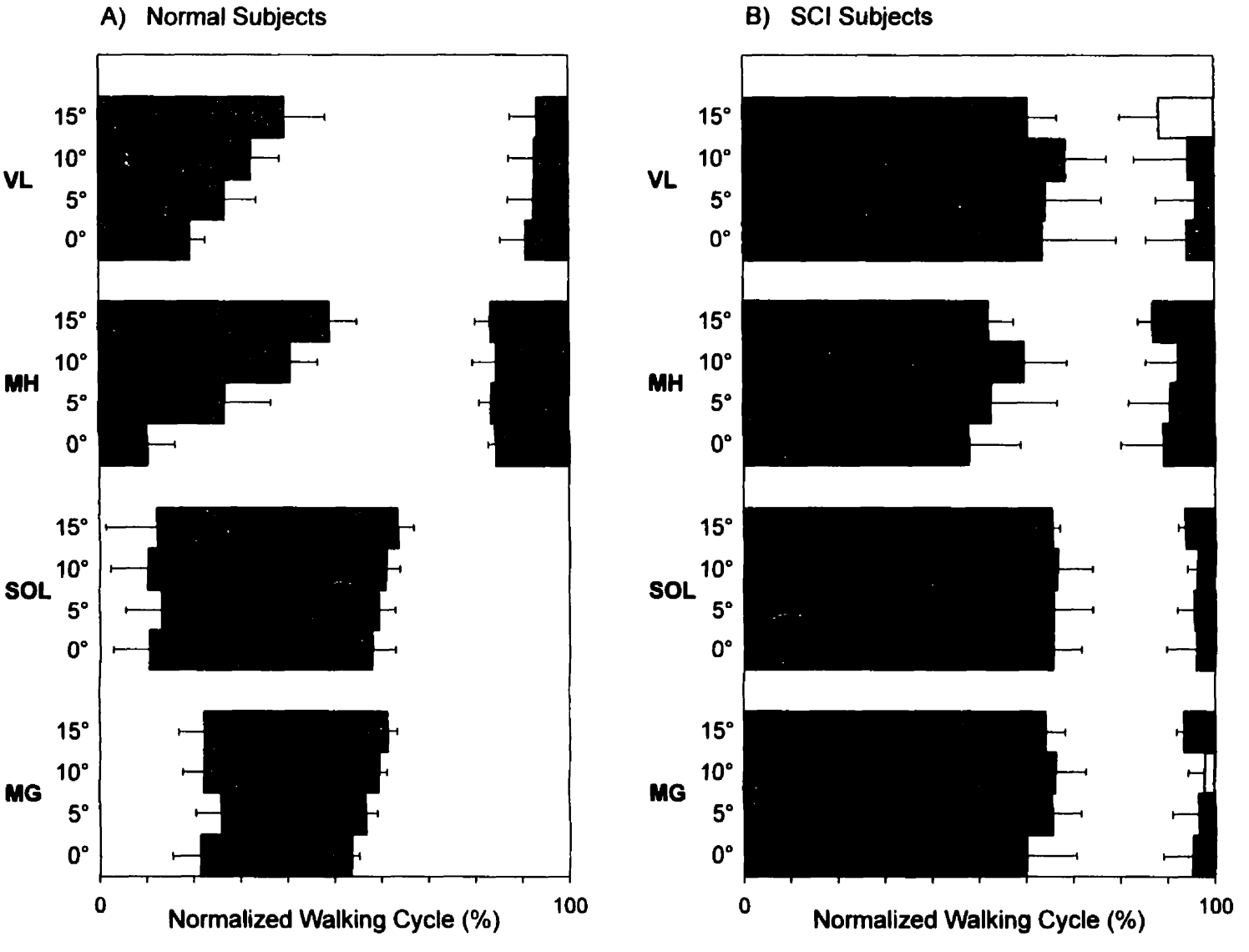

A. Peak amplitude of thigh muscles during stance phase
Vastus Lateralis
Medial Hamstrings
Percentage
500
400
300
200
100
250
150
50
0
0°
5°
10°
15°
Normal group (1.0 m/s)
Normal group (0.4 m/s)
SCI group
B. Peak amplitude of leg muscles during push-off period
Soleus
Medial Gastrocnemius
320
240
160
80
Treadmill grade

Figure 6


A) Normal Subjects
B) SCI Subjects
VL
MH
SOL
MG
15°
10°
5°
0°
0
100
Normalized Walking Cycle (%)

RESEARCH PAPER 2:

POSTURAL ADAPTATION TO WALKING ON INCLINED SURFACES: I. NORMAL STRATEGIES

Authors: Alain Leroux, M.Sc., Joyce Fung, Ph.D. and Hugues Barbeau, Ph.D.

Submitted to: Journal of Physiology

Category: Original research paper

Note: This work was presented in part at the 28th annual meeting of the Society for Neuroscience in 1998.

ABSTRACT

This study investigated the postural strategies to adapt to uphill and downhill treadmill inclinations (0, 5 and 10%) during walking and standing in 8 healthy subjects. Movements from trunk, pelvis and lower limbs were assessed while subjects walked at their comfortable speed. Increasing the treadmill grade from 0 to 10% induced an increasingly flexed posture of the hip, knee and ankle at initial foot contact as well as a progressive forward inclination of pelvis and trunk. These postural changes were accompanied by a progressive decrease in pelvic lateral drop toward the swinging limb and a gradual increase in stride length as the uphill slope became steeper. Decreasing the treadmill grade from 0 to −10% lead to a decreasingly flexed posture of the hip at initial foot contact as well as an increase in knee flexion during weight acceptance and late stance. These changes were accompanied by a gradual decrease in stride length, a progressive backward inclination of trunk and pelvis and an increase in pelvic lateral drop toward the swinging limb as downhill slope became steeper. Changes in trunk and pelvic postural alignment in the sagittal plane might be used to facilitate power generation or absorption in adapting to slope changes during walking. During quiet standing, however, the trunk and pelvis remained aligned with respect to earth's vertical at any surface inclination. These results showed that postural adaptations are task-specific and the control requirements are different between standing and walking on an inclined surface.

INTRODUCTION

The control of the pelvis and trunk is important to achieve smooth locomotion and maintain body equilibrium during gait. The pelvis and trunk, which represent a large portion of the body mass, have to interact with the lower limbs to accomplish efficient gait. In the six primary determinants of gait identified by Saunders et al. (1953), two were related to pelvic movement: 1) rotation of the pelvis in the transverse plane and 2) tilt of the pelvis in the frontal plane. These pelvic movements are performed to reduce the displacement of the center of gravity and to smooth the gait. During walking, the pelvis and trunk move in the three anatomical planes and three-dimensional analyses reveal very complex patterns of motions between pelvis and trunk (Stokes et al., 1989). In fact, the pelvis and trunk act together or against each other depending upon the period of the walking cycle to maintain body equilibrium and achieve smooth locomotion. For instance, it has previously been shown that the pelvis and trunk rotate in opposite directions during most of the gait cycle (Krebs et al; 1992; Stokes et al., 1989; Wagenaar and Beek, 1992). In moving out of phase with the pelvis, the trunk provides a damping effect on the pelvic rotation and contributes to a smooth gait pattern (Stokes, 1989).

The human locomotor pattern has been shown to be highly adaptable to different environment such as stair climbing (Andriacchi et al., 1980), changes in walking speed (Nilsson et al., 1985) or slope (Leroux et al., 1999; Wall et al., 1981). Previous studies have shown that gait adaptation on inclined surfaces is achieved by changing the pattern of lower limb motions (Lange et al., 1996; Leroux et al., 1999; Wall et al. 1981) and by modifying the level of activation of the relevant flexor and extensor muscles (Brandell, 1977; Lange et al., 1996; Leroux et al., 1999; Simonsen et al. 1995). This assumption is, however, very limited and does not take into consideration the importance of the trunk

and pelvis in the strategies of adaptation. This issue was addressed in only one study (Wall et al., 1981) that examined exclusively pelvic movements in the transverse plane. How postural adjustments from the pelvis and trunk are utilized to maintain balance when walking up and down slopes are yet to be determined. The main purpose of this study is to investigate postural adaptation to uphill and downhill walking in healthy subjects. Specifically, the three-dimensional kinematic patterns of the lower limbs, pelvis and trunk are examined during inclined treadmill walking. We hypothesize that the pelvis and trunk will be tightly controlled during standing but not walking on slopes. We expect a progressive, forward inclination of the trunk and pelvis during uphill walking with the increase in slope and vice versa during downhill walking. This adaptation process is necessary to move the center of gravity forward and backward, respectively, during up-slope and down-slope walking in order to counteract the effect of gravity. We further hypothesize that the total excursion (peak-to-peak range of motion) of the pelvis and trunk will remain constant during inclined walking, as the movements contribute more to balance maintenance than to changes in body propulsion during inclined walking.

METHODS

Subjects

Eight subjects who were free of any neurological or orthopaedic disorders took part in the experiment. This group of subjects, who were aged between 25 and 52 years (average: 34 ± 9.4 years), included 2 females and 6 males. Every subject signed an informed consent document and the study was approved by the ethics committee of the Jewish Rehabilitation Hospital (Laval, Québec, Canada).

Walking assessment

The subjects walked on a treadmill at five different grades (from –10% to 10%) without any assistance. Each subject had to wear a safety overhead harness throughout the course of the experiment. Several practice trials were used to familiarize the subjects with the different grades of treadmill walking and to determine the subjects' comfortable walking speed during inclined walking. During the practice trials and the experimental sessions, the subjects walked with elbows flexed at 90° to insure that each reflective marker could be detected by the cameras. All individuals were required to complete 15 walking trials at each treadmill grade. To control the effect of fatigue upon the gait pattern, the five walking grades were presented in random order. The subjects performed every walking grade at two different speeds: 1) their self-selected comfortable speed which ranged from 0.76 to 1.34 m/s (average: 1.03 ± 0.19 m/s) and 2) a very slow speed matched to that of a subject with incomplete spinal cord injury, ranging from 0.04 to 0.67 m/s (average: 0.38 ± 0.27 m/s) (paper in preparation). All subjects also performed quiet stance trials in which they were asked to stand for 5 seconds at each treadmill grade.

Data recording

Pressure sensitive footswitches (piezoresistive insoles) were inserted into subjects' shoes to signal the heel, fifth metatarsal, and hallux contacts. Signals from the footswitches were used to measure stance, swing and cycle durations. The footswitch signals were acquired at a sampling frequency of 1000 Hz per channel. The movements of the trunk, pelvis and lower limbs were acquired with four high-resolution cameras (BTS-ELITE system at 50 Hz). Reflective markers were placed on selected anatomical landmarks as shown in Figure 1A.

Data analysis

Five qualitatively representative gait cycles at every treadmill grade were selected for the analysis of the spatio-temporal variables as well as the kinematic data. Spatio-temporal variables analysed were stride length, cycle duration, percentage stance and percentage swing. The ELITE plus system was used to track all the reflective markers and compute their X, Y and Z coordinates, which were used to calculate the angular excursions of the trunk and pelvis in sagittal, transverse and frontal planes. Kinematic analysis of the hip, knee and ankle joints was done in the sagittal plane only. Segment and joint angles were calculated as shown on the kinematic model (see Figure 1B), with the directions displayed indicating positive angles from neutral and the reversed for negative angles. Segment angles from the trunk and pelvis during walking were normalized from the standing position of the level condition recorded during static trials (the values obtained in the different walking grades were subtracted to those obtained in quiet stance of the level condition). The hip and knee angles were calculated with respect to the vertical line, with the neutral position in standing being taken as 0° displacement. For the ankle angle calculation, the neutral standing position with the leg axis perpendicular to the foot was taken as 0°.

Critical events of the gait cycle characterizing the adaptation process to inclined walking have been chosen to average joint and segment angles across subjects. For lower limbs, hip and ankle angle at foot contact as well as knee angle during weight acceptance (peak value from 5 to 25% of the gait cycle) and late stance (peak value from 40 to 60% of the gait cycle) were averaged across subjects. For the trunk and pelvis, angular value during right single limb support (peak value from 20 to 50% of the gait cycle) were averaged across subjects and reported as trunk and pelvic inclination during walking. This time window represents the period where trunk and pelvic segments reach their peak forward inclination during walking.

Curve fitting using linear and non-linear regressions was performed only for $r^2 > 0.70$. Statistical analyses for all the variables were done in Systat (SPSS) using a two-way ANOVA with repeated measures. When significant differences were found ($p < 0.05$), pairwise comparisons were made using the Tukey test.

RESULTS

Sagittal plane kinematics

Figure 2 illustrates changes in lower limb joints in the sagittal plane during inclined walking for one representative subject. As previously described (Leroux et al., 1999), walking up-slope required an increased flexion from the three lower limb joints from mid-swing to early stance. This increase in flexion augmented progressively as the slope went steeper. Furthermore, the increase in hip flexion at the end of the swing phase might have led to the increase in stride length observed in uphill walking (see Figure 3). The main features associated with downhill walking were observed at the hip and knee joints. The hip joint showed a progressive decrease in flexion from mid-swing to early stance as the

slope went down. At the knee joint, there was a progressive flexion in early stance during weight acceptance and also in late stance (± 50% of the gait cycle) as the slope went down. At the ankle, no important changes were observed in downhill conditions.

The effects of treadmill grades on lower limb joint kinematics during critical events of the gait cycle for the group (n= 8) are shown in Figure 2-B. At the hip, a curvilinear relationship was found between treadmill grades and hip angulation at foot contact. This nonlinear relation ($r^2 = 0.99$) was characterized by a steeper slope in uphill walking as compared to downhill condition. A curvilinear relation ($r^2 = 0.99$) showing the same trend was also found at the ankle joint. At the knee, an increased flexion during weight acceptance was observed for extreme grades (-10 and 10%) whereas the least amount of flexion was measured in level condition during the same critical event. These changes lead to a "V" shape curve showing positive slopes when considering the level condition (0%) as the starting point (Figure 2-B). During late stance, knee angulation also showed important modifications. The relationship between treadmill grades and knee flexion was best described by an exponential function ($r^2 = 0.99$), such that the knee was more flexed in the extreme downhill condition and remained stable for the other conditions.

Increasing the treadmill slope from 0 to 10% uphill led to progressive increases in stride length whereas decreasing the treadmill slope from 0 to -10% downhill resulted in a progressive decrease of the latter variable (Figure 3-A). Statistical analyses revealed significant differences ($p<0.05$) between uphill and downhill conditions and between the extreme slope conditions (-10% and 10%) and the level condition. Considering the fact that the walking speed was maintained constant across treadmill grades, the increase in stride length had a consequence of increasing the cycle duration in these

conditions (Table). Opposite effects were observed for downhill conditions. However, the proportion of stance and swing phases relative to the cycle duration was not affected by the change in slope (Table).

The relationship between stride length and hip angle at heel contact across treadmill grades is shown in Figure 3-B. The non linear relationship between the two variables, which was best expressed by an exponential function ($r^2 = 0.97$), revealed two distinct adaptation features for downhill and uphill grades. In uphill walking, the important increase in hip flexion was directly associated with increases in both stride length and treadmill grade. Thus, an increase in hip flexion was used to adapt to changes in treadmill inclination and to generate a greater stride length. In downhill walking, however, the decrease in stride length was not associated with any substantial decrease in hip flexion (Figure 3-B). The increased flexion of the controlateral knee joint during late stance in downhill walking (see Figure 2-B) could mainly account for the decrease in stride length as discussed in the next section.

Figure 4 (A and B) illustrates trunk and pelvic motions in the sagittal plane during inclined walking for one typical subject. Both trunk and pelvis showed similar patterns of adaptation during inclined walking. Increasing the treadmill slope from 0 to 10% uphill led to a progressive forward inclination of trunk and pelvic segments whereas decreasing the treadmill slope from 0 to −10% downhill led to a progressive backward inclination of these two segments as compared to the level condition. Furthermore, the basic kinematic patterns for both the trunk and pelvis observed during level walking were not modified during inclined walking. Hence, a different absolute orientation was adopted for the trunk and pelvis during walking at different slopes, but the movements relative to the gait cycle remained the same.

Figure 4 shows the effect of treadmill grade on the sagittal orientation of the trunk (Fig 4C) and pelvis (Fig 4D). There were progressive changes from one treadmill grade to the next during walking. As shown in Figure 4 (C and D), these changes were linearly related to the treadmill grades. The linear relations were strong for both trunk ($r^2 = 0.98$) and pelvis ($r^2 = 0.99$), indicating that treadmill grades could account for almost the entire variance in trunk and pelvic orientations. During quiet stance conditions, however, trunk and pelvic segments were always maintained in the same position throughout the different treadmill grades. Thus, in the latter condition, no relationship could be found between treadmill grades and trunk and pelvic inclinations.

Figure 4 shows the effect of inclined walking on the total angular excursion (TAE) of trunk (Figure 4E) and pelvis (Figure 4F) in the sagittal plane for the group. Walking up-slope or down-slope did not affect the TAE in both trunk and pelvis. The movements were very small (2 to 3° for the trunk and 6 to 7° for the pelvis) and remained constant across the treadmill grades measured in the present study.

Frontal plane kinematics

Results obtained from the frontal plane revealed great variations in trunk movement patterns within and between subjects during inclined walking. Results from one representative subject are shown in Figure 5-A. Kinematic patterns revealed that this subject walked with a slight leaning of the trunk (toward left) during level and uphill conditions as expressed by the downward displacement of the curve. This left leaning disappeared in downhill conditions and a neutral position of the trunk was adopted. In terms of patterns of movement, inclined walking minimally affected the basic trunk kinematic pattern observed in level gait. When considering the whole group of subjects, no trend of adaptation could be found across treadmill grades. Furthermore, trunk TAEs were very limited (3 to

5° only) and remained relatively constant across walking grades. As opposed to the trunk, pelvic TAEs were affected by inclined walking in the frontal plane. The main changes were found during swing phases which represent the periods where the pelvis dropped toward the swinging leg. This lateral drop was more pronounced in downhill walking and reduced in uphill walking as compared to level gait. Modifications in pelvic lateral tilt across walking grades were present in every subject and gave rise to changes in pelvic TAE (Figure 5-D). These changes were directly related to the treadmill grade as expressed by the strong linear relationship ($r^2 = 0.97$) between pelvic lateral tilt and treadmill grades. When examining the relationship between trunk and pelvic segments, it can be seen that movements were performed in opposite directions in every phase of the gait cycle during both level and inclined walking (Figure 5, A-B). This relationship will be further examined in the cyclographs showing the different coordination patterns (Figure 7).

Transverse plane kinematics

The effects of inclined walking on trunk rotation as measured by shoulder girdle alignment and pelvic angular displacement in the transverse plane for one representative subject are shown in Figure 6 (A and B). As previously seen for the frontal plane, treadmill grades did not change the basic kinematic patterns of shoulder and pelvic girdle movements observed during level walking. Furthermore, the TAEs of the two girdles, which represent the maximal amount of rotations, were also not affected (Figure 6, C and D).

Pelvic-trunk coordination pattern

Figure 7 illustrates pelvic-trunk coordination during level and inclined walking in sagittal, frontal and transverse planes. Individual subjects all showed the same pelvic-trunk coordination patterns in each

of the plane of motions. Thus, all data were averaged and pooled together. In general, a diagonal curve showing a positive slope indicates that the pelvic and trunk segments move in synchrony and in the same direction. On the other hand, a diagonal curve with a negative slope indicates that the movements, although synchronous, occur in opposite directions. A square or rectangular pattern parallel to axes indicates that the two segments never move simultaneously throughout the gait cycle.

In the sagittal plane, the pelvic-trunk coordination pattern reveals that movements were greater for the pelvis than for the trunk (Figure 7-A, left panel). The relative coordination pattern observed in level did not change with inclined walking (Figure 7-A, right panel). However, as previously shown in Figure 4, both trunk and pelvis showed a shift in their absolute angular orientation during inclined walking. The strategy involves a forward inclination of the trunk and pelvis during uphill walking and backward inclination during downhill walking. Furthermore, the linearity of parts of the loops displayed in the cyclographs reveal a large degree of synchrony in trunk and pelvic movements.

In the frontal plane, pelvic and trunk movements occurred in opposite directions during most of the gait cycle as expressed by the negative slope of the diagonal curve in Figure 7-B (left panel). Furthermore, trunk and pelvic maximal excursions were reached at the same period of the gait cycle but in opposite directions. The basic pelvis-trunk coordination pattern observed in level walking was influenced by the change in slope. The cyclographs showed widening loops indicating less tightly coupled motions between the trunk and pelvis, especially during downhill walking, with large pelvic motions.

In the transverse plane, shoulder and pelvic girdle movements were not performed in sychrony during most of the gait cycle (Figure 7-C, left panel). Especially during uphill walking, a rectangular cyclograph was observed, indicating that one segment was stabilized while another moved, with the shoulder girdle moving mostly during the period of single limb support (swing) and the pelvic girdle moving from the late period of single limb support to the mid period of double limb support. A more synchronous coordination was observed during downhill walking, when the pelvic and shoulder girdles rotate in the same direction from the late period of single limb support to the end of the double limb support period. The absence of synchronicity was only found from the beginning to the mid period of single limb support when rotation occurred almost exclusively at the shoulder girdle.

DISCUSSION

This study showed that the main postural requirement when adapting to inclined walking is to change trunk and pelvic orientation relative to earth's vertical in the sagittal plane while maintaining the same patterns of movement across the different treadmill inclinations. Trunk and pelvic alignment showed a forward inclination during uphill walking and a backward inclination during downhill walking as compared to level walking. These modifications were not seen in quiet stance conditions where trunk and pelvic segments remained vertically oriented across different slopes. These results suggest that postural adaptations are task-specific and the control requirements are different between dynamic and static conditions.

Task dependent control of trunk and pelvic orientation

Results obtained from static condition showed that trunk and pelvic orientation remained aligned with

respect to earth's vertical for any treadmill grade. Our results further showed a tight control of the trunk and pelvis as expressed by the very limited variations in the trunk (± 0.7°) and pelvic (± 2.34°) angles across the different slopes. This tight control in stance across slopes suggests that human subjects can perceive and orient the trunk and pelvis with respect to gravity at a high level of accuracy. Psychophysical studies have shown that humans can perceive trunk orientation in space very accurately for lateral flexion (less than 3 degrees) and rotation (0.9 degrees) even in the absence of vestibular inputs (Jacobs et al., 1985; Taylor and Mccloskey, 1990). Based on findings from subjects with spinal cord injury, proprioceptive information from lumbosacral area and internal organs can also be used to perceive trunk orientation (Mittelstaedt and Fricke, 1988).

In addition to inclined surface, the importance of trunk orientation in the maintenance of stance posture has been shown in various tasks (Fung and Macpherson, 1995; Gurfinkel et al., 1981; Lacquaniti et al., 1984). When applying surface perturbations (platform tilt) during stance posture, Gurfinkel et al. (1981) found that human subjects always maintained the trunk oriented with respect to earth's vertical. In contrast, quadrupeds maintain trunk orientation parallel to the support surface (Fung and Macpherson, 1995, 1999; Lacquaniti et al., 1984). When varying interpaw distances in stance, cats used the limbs as levers and maintained a constant horizontal orientation of the trunk even after a complete spinal cord transection (Fung and Macpherson, 1999). Thus, trunk stabilization in space in both humans and cats appear to be a major controlled variable in postural orientation and equilibrium. This tight control may be explained by the fact that the greatest mass of the body is contained in the trunk. Furthermore, trunk vertical alignment in space could also act as an egocentric frame of reference in determining upper and lower limb positions. In a study on leg elevation in trained and untrained dancers, it was found that both groups of subjects used the trunk rather than the

external environment to calculate leg movements (Mouchnino et al., 1993).

The control mechanisms underlying the maintenance of balance are very distinct for walking and standing. Instead of keeping the body's center of gravity (C of G) safely within the base of support as in standing, human locomotion required a forward propulsion of the body that can move the C of G outside the area of support (Winter et al., 1993). Results from the present study show that every subject leans the trunk forward during level walking. These results are in agreement with previous findings on trunk angulation during level walking (Thorstensson et al., 1984; Vogt and Banzer, 1999). This forward leaning may be used by the CNS to assist in propelling the body forward. By inclining the trunk forward, subjects would move the center of gravity slightly ahead of the center of foot and the force of gravity would tend to accelerate the body forward relative to the center of support. This is in agreement with Hirokazu and Yamamuro (1987) who showed that the C of G has to be moved slightly ahead of the body during walking. Thus, trunk inclination during walking can assist the lower limbs in generating forward momentum.

As previously seen for the trunk, the orientation of the pelvis showed a different control for standing and walking conditions. In contrast to the standing condition where the pelvis was stabilized in a neutral position, the movements were much greater in the three anatomical planes during level walking. The main function of the pelvis during human walking is to interact with lower limbs to reduce the displacement of the C of G and to smooth the gait (Saunders et al., 1953). In the present study, changes in pelvic orientation in the sagittal plane were mainly performed to assist hip movements during swing phase. As illustrated in Figure 4, the two peaks of backward tilt corresponded to the period where the leg was lifted up. Thus, changes in pelvic orientation in the

sagittal plane seem to counteract the effect of gravity on hip movements and contribute to smoothing the gait.

Postural adaptations to surface changes

We hypothesized that there were significant postural modifications from the trunk and pelvis during uphill walking because the body tends to be pulled passively backwards by gravity and, to counteract this effect, an active forward inclination in the trunk and pelvis would be required. On the other hand, walking down-slope would produce the opposite effect of pushing the body forwards by gravity and active control is required to counteract this passive physical effect. There is no forward momentum generated during quiet stance and thus the postural control during inclined standing is very different, in which the trunk and pelvis simply orient to the gravitational force field (earth's vertical). Inputs from proprioceptive, visual and vestibular systems all contribute to the postural orientation and equilibrium when standing upright on an inclined surface. When examing the relative contribution of each of these systems, we might expect an important role from the proprioceptive system due to changes in ankle angulation across slopes. Treadmill slopes elicit stretches in the ankle muscles (gastrocnemius and soleus for uphill and tibialis anterior for downhill slopes) and this proprioceptive input could override the information coming from the two other systems. However, to avoid inappropriate muscle contraction that could perturb quiet stance, we believe that the reflex gain has to be reduced during such a standing task. On the other hand, proprioceptive inputs from the trunk and pelvis might be more important during standing on an inclined surface. Jakobs et al. (1985) showed that when human subjects are moved slowly from side to side in the frontal plane, they can center their trunk with respect to the pelvis with a high level of accuracy in standing position with eyes closed and in supine position with no contribution from visual and vestibular systems. Furthermore, in

response to a 5° lateral pelvic tilt in the standing position, subjects aligned their trunk with respect to earth's vertical instead of using the pelvic tilt as a new frame of reference. These data suggest that proprioceptive input from the trunk could play an important role.

We believe that the increase in trunk inclination during uphill walking is performed to assist lower limbs generating more momentum to counteract the resistance due to gravity. Evidence is provided in Figure 8 which clearly shows that the level of trunk inclinaton is directly related to the walking speed during inclined walking. In uphill grades, the trunk requires a greater degree of forward inclination at faster speeds. The need for generating this greater momentum is also reflected by a progressive increase in stride length as the treadmill slope becomes steeper in uphill walking. During downhill walking, the body is moved forward and downward by the mechanical effect of the slope. To counteract these external forces, lower limb joints (especially the knee) have to absorb more energy and brake the forward momentum of the body (Redfern and DiPasquale, 1997). This decrease in body momentum is reflected by a progressive decrease in stride length as the treadmill slope becomes steeper in downhill conditions. We believe that backward movements of the trunk and pelvis are needed to move the center of gravity backward to decrease the forward momentum assisted by gravity during downhill walking. Evidently, the trunk was less inclined during downhill walking when the speed was slow (Figure 8).

A direct relationship between trunk inclination and stride length has also been observed during level walking at fast speeds (Murray et al., 1966; Thorstensson et al., 1984). Murray et al. (1966) showed that both stride length and trunk forward inclination were much more pronounced when subjects performed fast walking (2.2 m/s) as compared to free walking (1.5 m/s). Comparable results from

Thortensson et al. (1984) study revealed a direct relationship between walking speed and trunk angulation. Increasing the walking speed from 1 to 2.5 m/s lead to a progressive increase in trunk forward inclination (from 6 to 13°) that was very similar to the results reported in the present study. In these two locomotor tasks (slope walking and fast walking) trunk forward inclination seems to be the best strategy in terms of generating a greater forward propulsion hence the most efficient gait pattern.

Adaptations versus compensations

Previous studies showed that inclined walking required important adaptations from lower limbs pattern of movements (Leroux et al., 1999; Wall et al., 1981). Results from the present study showed that the adaptation process also includes modifications in trunk and pelvic orientation in sagittal (as discussed above) and frontal planes. As shown in Figure 2, the key mechanism when adapting to uphill walking is to lift up the swinging leg by performing a simultaneous increase in hip and knee flexion of that limb. This period of the gait cycle corresponds to the period where the pelvic tilt toward the swinging limb was decreased (see Figure 5). By decreasing this lateral tilt, the pelvis would permit a greater lift up of the swinging limb. During downhill walking, the key mechanisms of adaptation are to decrease hip flexion of the swinging limb and increase knee flexion of the controlateral limb during the same period of the gait cycle. The increased flexion of the supporting knee is used to decrease: 1) the vertical displacement of the center of mass; 2) the impact force at heel contact; 3) the stride length (Wall et al., 1981). We believe that the increase in the lateral drop of the pelvis toward the swinging limb found during downhill walking would also contribute to the decrease of the three factors mentioned above. Thus, as previously proposed, lateral pelvic tilts would assist lower limbs movements when adapting to both uphill and downhill walking.

We hypothesed that trunk and pelvic movements contribute more to balance maintenance than to changes in body propulsion during uphill and downhill walking. This hypothesis is in agreement with our results showing no significant changes in TAEs in the three anatomical planes across walking grades. Furthermore, the consistency in trunk and pelvic kinematic patterns across grades in all planes suggest a tight control from the nervous system. We believe that trunk and pelvic functions during level and inclined walking could be modified following neuromuscular and orthopedic disorders affecting lower limb movements. For instance, accentuations of pelvic and trunk rotations during walking were found in subjects with hemiplegia (Wagenaar and Beek; 1992) and unilateral hip pain (Thurston, 1985). Leroux et al. (1999) also reported an accentuation in pelvic lateral tilt on the unsupported side during the swing phase in one subject with spinal cord injury. These accentuations in trunk and pelvic movements were all performed to compensate for lower limb deficits during level walking. Compensatory movements from the trunk and pelvis could even be more pronounced and play an important role for the maintenance of balance when performing uphill and downhill walking in subject with lower limb deficits. This question is currently being examined in subjects with incomplete spinal cord injury (paper in preparation).

CONCLUSION

Our results showed that when standing on an inclined plane, the main strategy was to maintain the center of gravity within the base of support and to do so subjects oriented the trunk and pelvis with respect to earth's vertical across treadmill slopes. Thus, in the maintenance of standing posture on an inclined plane, the vertical orientation of the trunk and pelvic is the main controlled variable. During walking, however, trunk inclination is necessary to move the center of gravity ahead of the base of support to assist the forward propulsion of the body. Modifications in trunk and pelvic postural

alignment during inclined walking are necessary to counteract the effects of gravity on forward momentum at different slope conditions. In summary, results of this study showed that postural adaptations are task-specific and the control requirements are different between standing and walking conditions on an inclined surface.

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Table. Temporal measurements for level and inclined walking (average values from the 8 subjects)

	% Grade				
	-10	-5	0	5	10
Cycle duration (s)	1.13 ± 0.13	1.17 ± 0.15	1.21 ± 0.15	1.24 ± 0.16	1.25 ± 0.18
Stance, % of cycle	62.70 ± 3.20	64.02 ± 3.51	63.04 ± 4.38	62.99 ± 4.33	63.65 ± 3.29
Swing, % of cycle	37.30 ± 3.20	35.98 ± 3.51	36.96 ± 4.38	37.01 ± 4.33	36.35 ± 3.29

LEGENDS

Figure 1: A, Lateral and posterior views of the kinematic model. To model upper and lower limbs movements, reflective markers were placed bilaterally on the glenohumeral joint, posterior superior iliac spine (PSIS), anterior superior iliac spine (ASIS), greater trochanter, lateral femur epicondyle, lateral malleolus, calcaneus, fifth metatarsal as well as along the spine on the seventh cerebral vertebra (C_7) and the third lumbar vertebra (L_3). B, Segment and joint angle computations are illustrated for the sagittal plane (left) as well as frontal and transverse planes (right). For every segment and joint angles, the arrow direction indicates positive angles from neutral.

Figure 2: Angular excursions of the hip, knee and ankle during uphill and downhill walking in the sagittal plane. A, results from one representative subject normalized to the gait cycle (from one foot contact, 0% to the next, 100%). B, average hip, knee and ankle angles (mean ± 1 SD) for the group at critical events of the gait cycle.

Figure 3: A, average stride length (mean ± 1 SD) for the group during uphill and downhill walking. B, relationship between stride length and hip angle at foot contact across treadmill grades. Each point represents the group average (mean ± 1 SD).

Figure 4: Angular excursions of the trunk and pelvis during uphill and downhill walking in the sagittal plane. A-B, angular displacement of the trunk (A) and pelvis (B) normalized to the gait cycle (from one foot contact, 0% to the next, 100%) from one representative subject. Arrows at the bottom of the graphics are used to indicate periods of heel contact (downward direction) and toe-off (upward direction) as well as events related to the right (bold character) or left (regular character) lower limbs.

The same convention is used for all figures. C-D, average inclination (mean ± 1 SD) of the trunk (C) and pelvis (D) for the group during single limb support in standing and walking conditions across walking grades. E-F, average total angular excursion (mean ± 1 SD) of the trunk (E) and pelvis (F) for the group during uphill and downhill walking.

Figure 5: Angular excursions of the trunk and pelvis during uphill and downhill walking in the frontal plane. A-B, angular displacement of the trunk (A) and pelvis (B) normalized to the gait cycle (from one foot contact, 0% to the next, 100%) from one representative subject. C-D, average total angular excursion (mean ± 1 SD) of the trunk (C) and pelvis (D) for the group during uphill and downhill walking.

Figure 6: Angular excursions of the shoulder girdle and pelvis during uphill and downhill walking in the transverse plane. A-B, angular displacement of the shoulder girdle (A) and pelvis (B) normalized to the gait cycle (from one foot contact, 0% to the next, 100%) from one representative subject. C-D, average total angular excursion (mean ± 1 SD) of the shoulder girdle (C) and pelvis (D) for the group during uphill and downhill walking.

Figure 7: Cyclographs showing the coordination between the pelvis and trunk during level (left panels) and inclined (right panels) walking in the sagittal (A), frontal (B) and transverse (C) planes. This figure shows the average of the 8 normal subjects. The dotted lines represent the period of double limb support and the solid lines represent the period of single limb support for the right (R) and left (L) limbs. The arrows indicate the right heel contact event as well as the direction of the movement for the level condition. Displacement of the curves diagonally up and right indicates that

pelvic and trunk movements move in synchrony and in the same direction whereas a diagonal curve with a negative slope reveals that movements are also performed in synchrony but in opposite directions.

Figure 8: Average inclination of the trunk (mean ± 1 SD) during single limb support across walking grades for the sub-group of subjects walking at comfortable (1.03 m/s), medium (0.62 m/s) and slow (0.13 m/s) speed.

Figure 1

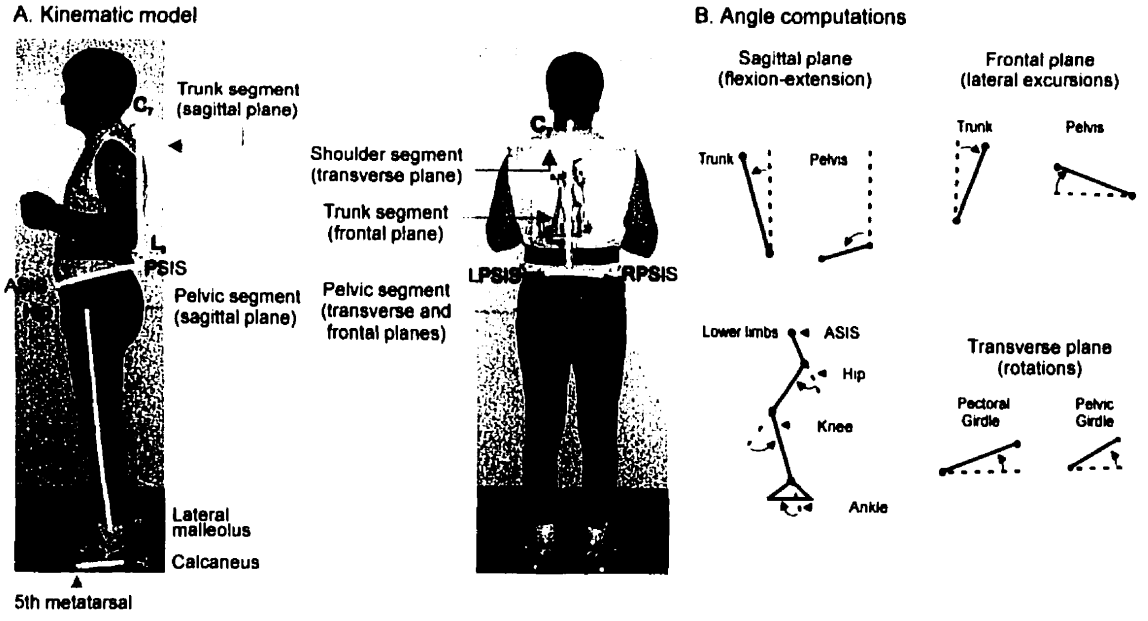

A. Kinematic model
Trunk segment (sagittal plane)
Shoulder segment (transverse plane)
Trunk segment (frontal plane)
Pelvic segment (transverse and frontal planes)
Pelvic segment (sagittal plane)
LPSIS
RPSIS
Lateral malleolus
Calcaneus
5th metatarsal
B. Angle computations
Sagittal plane (flexion-extension)
Frontal plane (lateral excursions)
Trunk
Pelvis
Lower limbs
ASIS
Hip
Knee
Ankle
Transverse plane (rotations)
Pectoral Girdle
Pelvic Girdle

FIGURE 2

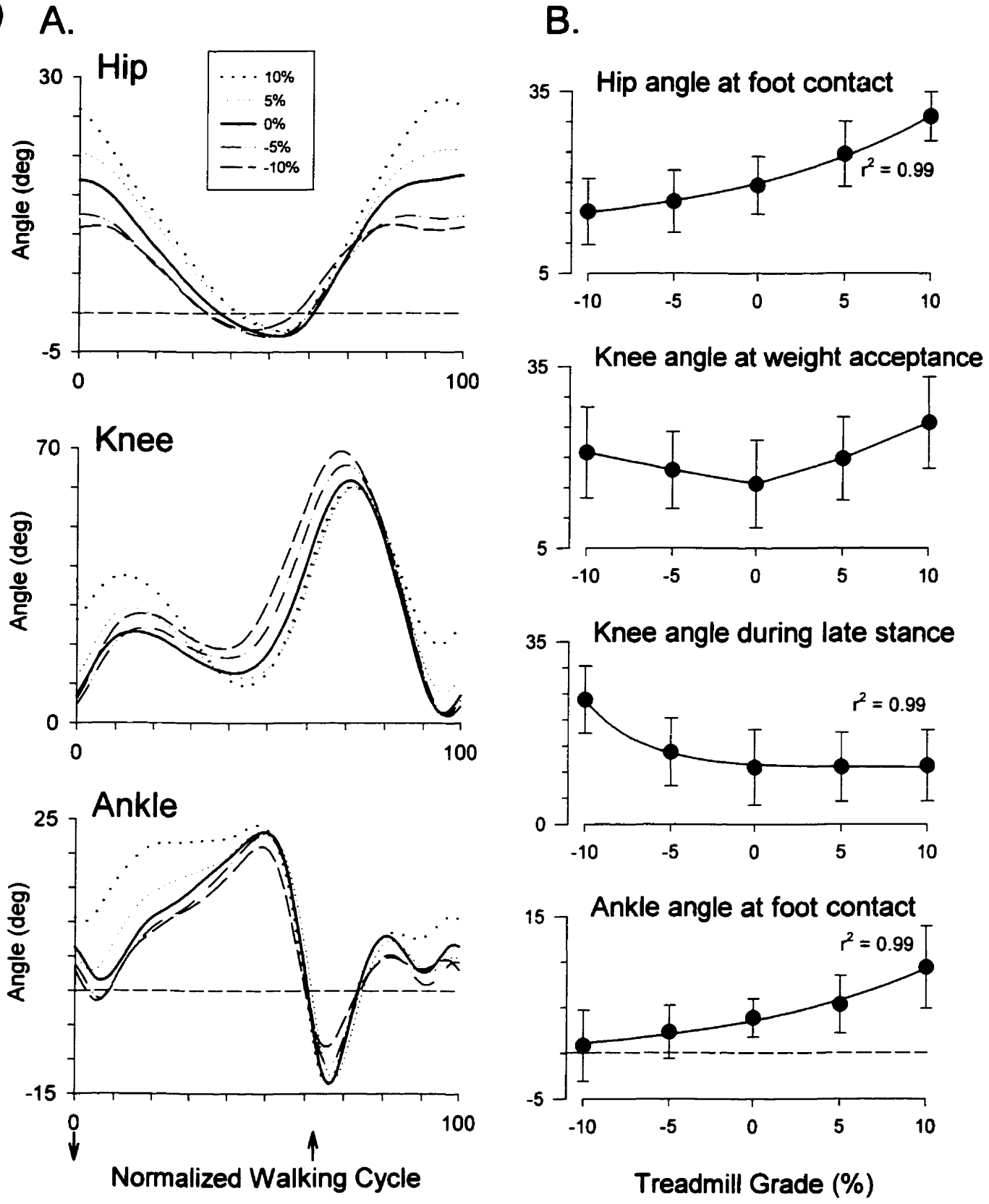

A.
Hip
10%
5%
0%
-5%
-10%
30
-5
0
100
Angle (deg)
Knee
70
0
0
100
Angle (deg)
Ankle
25
-15
0
100
Angle (deg)
Normalized Walking Cycle
B.
Hip angle at foot contact
35
5
r^2 = 0.99
-10
-5
0
5
10
Knee angle at weight acceptance
35
5
-10
-5
0
5
10
Knee angle during late stance
35
0
r^2 = 0.99
-10
-5
0
5
10
Ankle angle at foot contact
15
-5
r^2 = 0.99
-10
-5
0
5
10
Treadmill Grade (%)

FIGURE 3

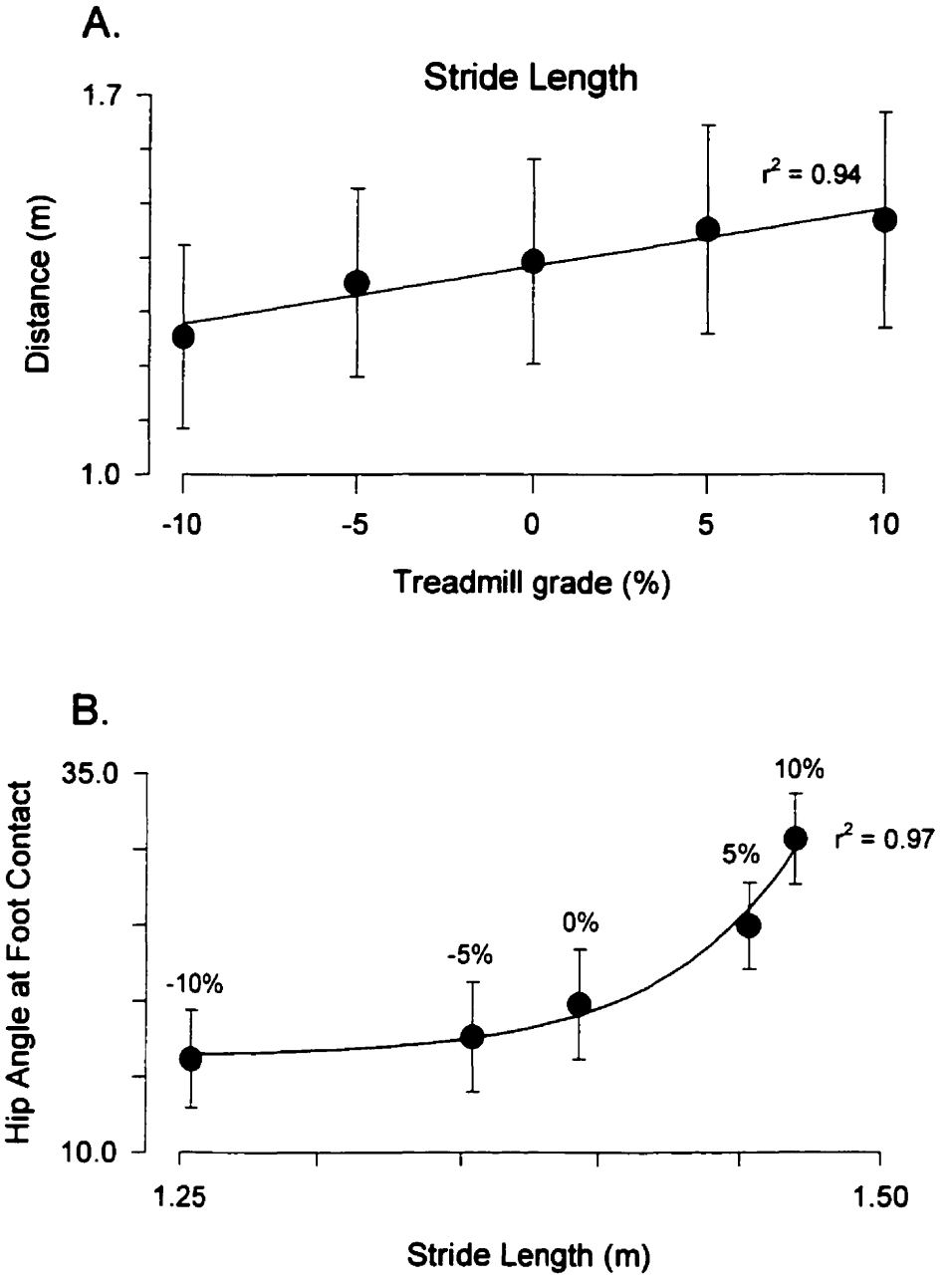
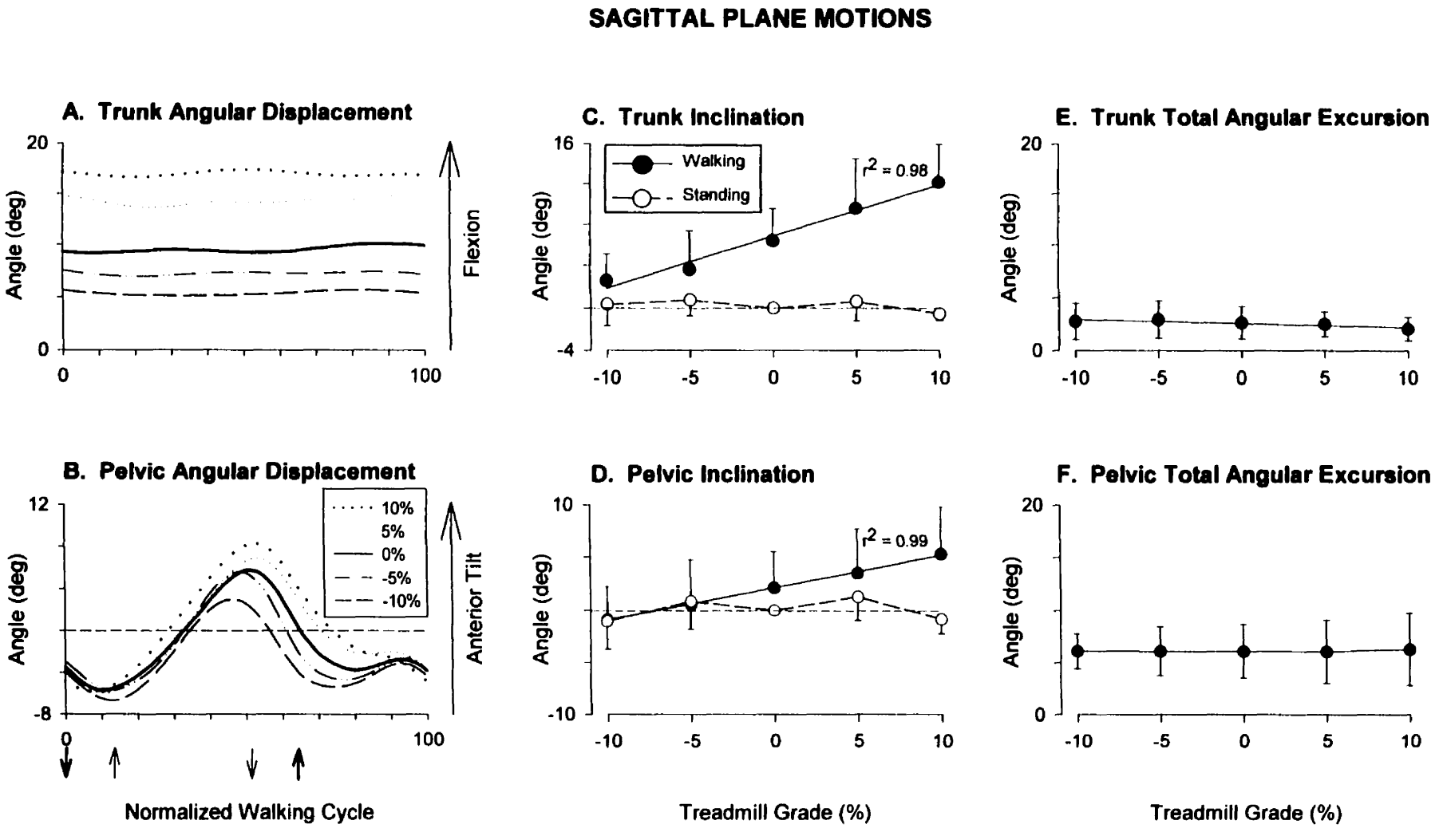

A.
Stride Length
1.7
1.0
Distance (m)
r^2 = 0.94
-10
-5
0
5
10
Treadmill grade (%)
B.
35.0
10.0
Hip Angle at Foot Contact
-10%
-5%
0%
5%
10%
r^2 = 0.97
1.25
1.50
Stride Length (m)

FIGURE 4


SAGITTAL PLANE MOTIONS
A. Trunk Angular Displacement
Angle (deg)
20
0
0
100
Flexion
B. Pelvic Angular Displacement
Angle (deg)
12
-8
0
100
10%
5%
0%
-5%
-10%
Anterior Tilt
Normalized Walking Cycle
C. Trunk Inclination
Angle (deg)
16
-4
Walking
Standing
$r^2 = 0.98$
-10
-5
0
5
10
D. Pelvic Inclination
Angle (deg)
10
-10
$r^2 = 0.99$
-10
-5
0
5
10
Treadmill Grade (%)
E. Trunk Total Angular Excursion
Angle (deg)
20
0
-10
-5
0
5
10
F. Pelvic Total Angular Excursion
Angle (deg)
20
0
-10
-5
0
5
10
Treadmill Grade (%)

FRONTAL PLANE MOTIONS

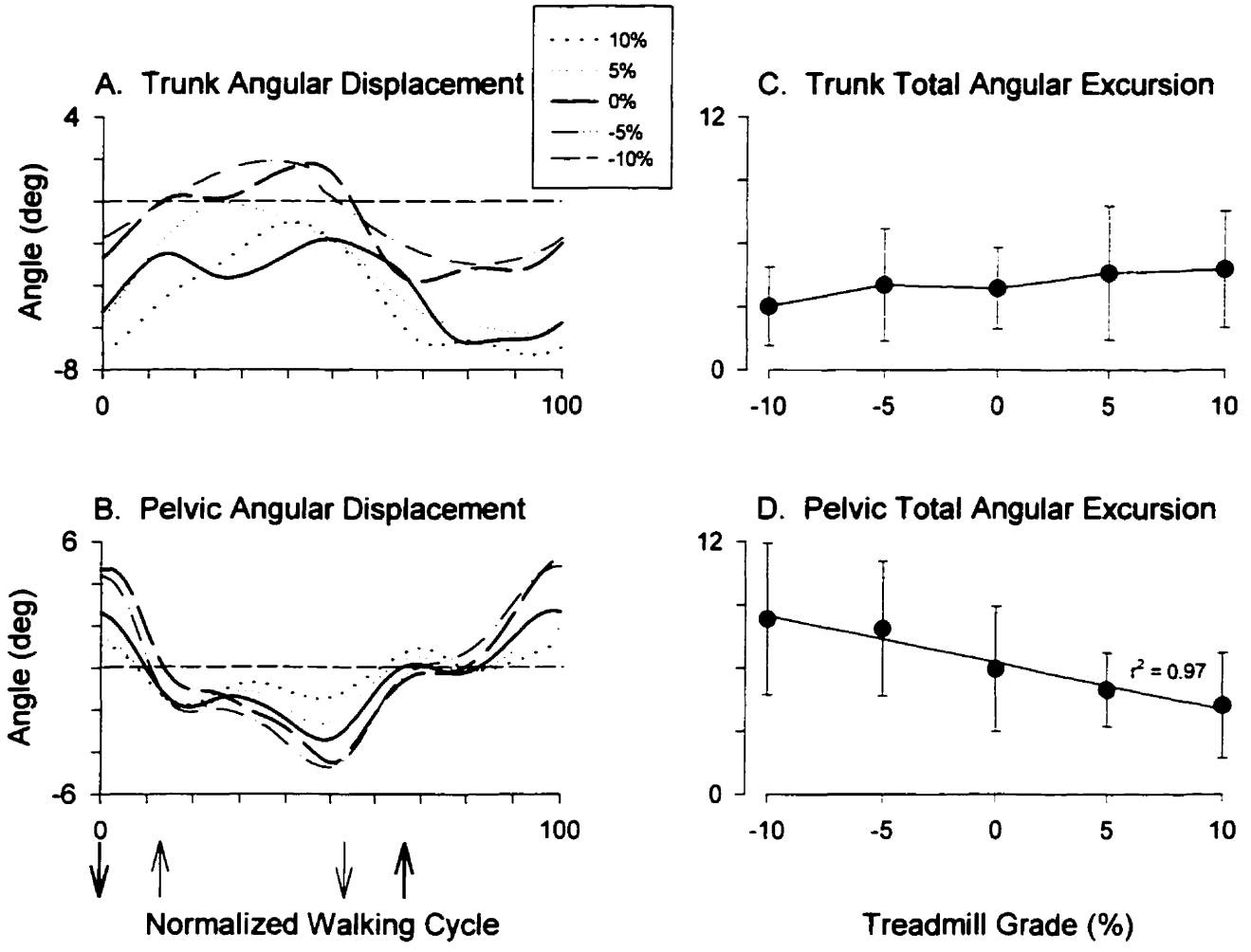


FIGURE 6

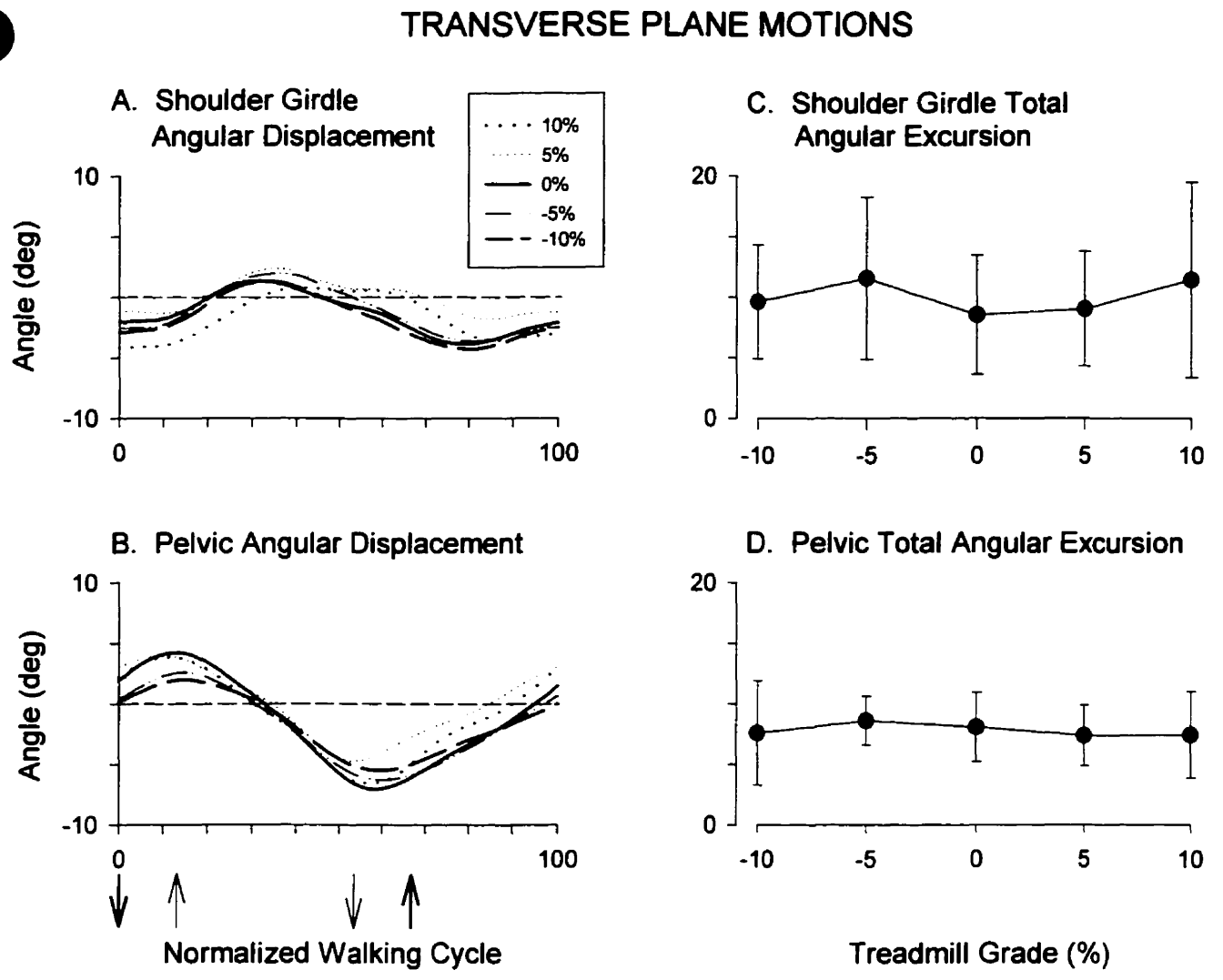

TRANSVERSE PLANE MOTIONS
A. Shoulder Girdle Angular Displacement
10%
5%
0%
-5%
-10%
Angle (deg)
10
-10
0
100
C. Shoulder Girdle Total Angular Excursion
20
0
-10
-5
0
5
10
B. Pelvic Angular Displacement
Angle (deg)
10
-10
0
100
Normalized Walking Cycle
D. Pelvic Total Angular Excursion
20
0
-10
-5
0
5
10
Treadmill Grade (%)

FIGURE 7

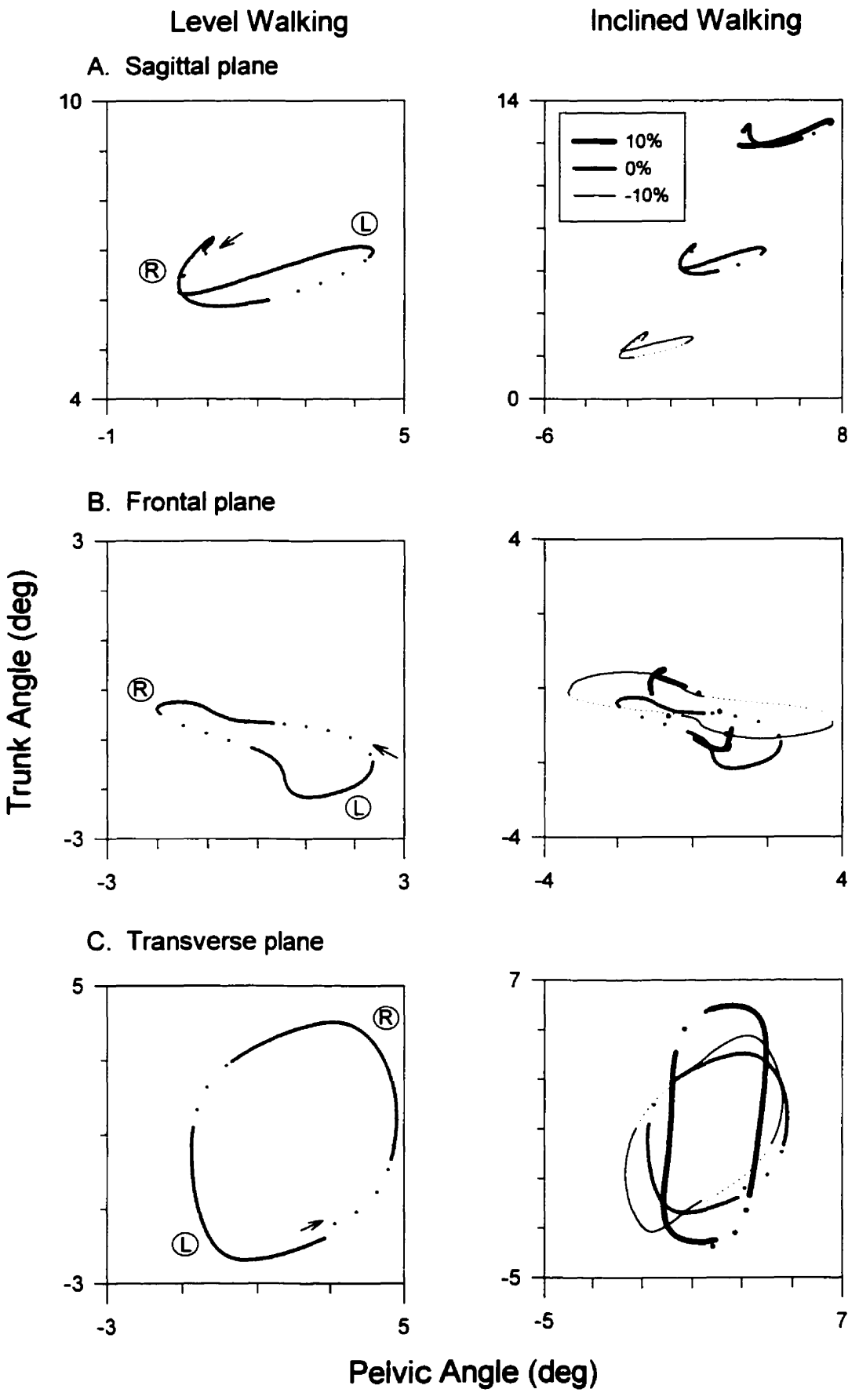
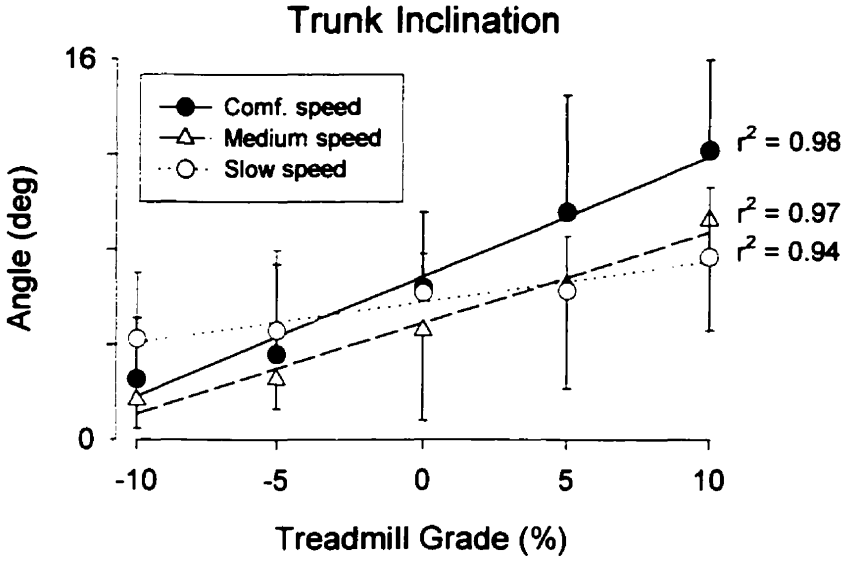

Level Walking
Inclined Walking
A. Sagittal plane
10
4
-1
5
R
L
14
0
-6
8
10%
0%
-10%
B. Frontal plane
3
-3
-3
3
4
-4
-4
4
C. Transverse plane
5
-3
-3
5
7
-5
-5
7
Trunk Angle (deg)
Pelvic Angle (deg)

FIGURE 8


Trunk Inclination
16
0
Angle (deg)
Comf. speed
Medium speed
Slow speed
r² = 0.98
r² = 0.97
r² = 0.94
-10
-5
0
5
10
Treadmill Grade (%)

RESEARCH PAPER 3:

POSTURAL ADAPTATION TO WALKING ON INCLINED SURFACES: II. STRATEGIES FOLLOWING SPINAL CORD INJURY

Authors: Alain Leroux, M.Sc., Joyce Fung, Ph.D. and Hugues Barbeau, Ph.D.

Submitted to: Experimental Brain Research

Category: Original research paper

Note: This work was presented in part at the 28th annual meeting of the Society For Neuroscience in 1998.

ABSTRACT

Trunk, pelvic and lower limb movements were assessed from 8 healthy subjects and 8 subjects with spinal cord injury (SCI) during level and inclined treadmill walking (at -10, -5, 0, 5 and 10% grades). Subjects with SCI walked at their comfortable speed (ranging from 0.04 to 0.67 m/s) while each healthy subject walked at a low speed matching that of a subject with SCI. Results showed that, following SCI, trunk and pelvic movements were affected by gait speed and the extent of lower limb deficits during walking. Subjects with SCI showing moderate mobility dysfunctions (SCI subgroup 1; n = 5) were the ones who walked faster and exhibited near normal patterns of trunk and pelvic movements in the three planes of motions during level and inclined walking. However, when comparing these subjects with the matched healthy subjects, they always showed a greater forward inclination of both trunk and pelvis during level and inclined walking, especially in downhill conditions where both segments were maintained at about the same level as in level gait. The same abnormal feature (but more pronounced) was found in subjects with SCI showing severe mobility dysfunctions (SCI subgroup 2; n = 3). All subjects with SCI seemed to adopt this anterior posture to achieve greater stability during level and inclined walking. Results from frontal and transverse plane motions revealed that compensatory movements from trunk and pelvic segments were performed by SCI subgroup 2 only. These compensatory movements were maintained similar across treadmill grades. The SCI subgroup 1 and healthy subjects did not show any significant changes for trunk and pelvic total angular excursions in the three planes of motions when comparing level and inclined walking. In contrast, SCI subgroup 2 showed marked changes in trunk and pelvic angular excursions in all three planes of motion during uphill and downhill walking. These modifications suggest a possible

involvement of the axial segments in inclined walking when the lower extremity functions are severely impaired due to SCI.

INTRODUCTION

The problem of gait adaptation in individuals with impaired locomotor function is complex and not well studied in the literature. Hence, very little is known about the contribution of the lower limbs and upper portions of the body to the adaptation of the gait pattern in subjects with spinal cord injury (SCI). During normal gait, the main functions of the pelvis and trunk are to maintain body equilibrium and achieve smooth locomotion (Saunders et al., 1953; Stokes et al., 1989). With gait pathologies, however, the pelvis and trunk could be used to compensate for lower limb deficits.

Results from a previous study (Leroux et al., 1999) showed that subjects with SCI had problems adapting their locomotor pattern to uphill walking. Contrary to the healthy subjects who showed a clear pattern of adaptation at the hip, knee and ankle joints, subjects with SCI used mainly the hip when adapting to uphill walking. Furthermore, while the plantarflexor muscles showed the greatest increase of activity in healthy subjects during uphill walking, it was minimally altered in all subjects with SCI. The lack of increase in the level of activation of plantarflexors in subjects with SCI likely leads to a weak push-off. Thus, compensatory mechanisms from axial and proximal muscles might be needed when adapting to inclined walking. Hence, abnormal movements of the pelvis and trunk may be used to compensate for the lower extremity deficits in subjects with SCI. Furthermore, these compensatory mechanisms may depend on the severity of sensori-motor impairment. The more impaired subjects might use the pelvis and trunk to a greater extent to compensate for lower limb deficits and adapt to the different inclines. For instance, it has been previously shown that subjects

with unilateral hip pain due to osteoarthritis and avascular necrosis used larger pelvic and trunk movements to compensate for limited hip motion (Murray et al., 1971; Thurston, 1985). These subjects showed an abnormal lateral elevation of the pelvis during swing phase to compensate for limited hip flexion and provide a toe-floor clearance similar to normal walking. Excessive rotations from the pelvis and trunk were also reported in subjects with hip pain (Thurston, 1985) and in those with hemiplegia due to stroke (Wagenaar and Beek, 1992). These studies showed that abnormal movements from trunk and pelvis can be used to overcome lower limb deficits in pathological gait.

The main purpose of the present study is to determine the postural adaptations to inclined walking in subjects with SCI. The contribution of the trunk, pelvis and lower limbs is examined during level and inclined treadmill walking. The general hypothesis is that subjects with SCI will use to a greater extent pelvic and trunk movements to perform inclined walking as compared to healthy subjects. Three specific hypotheses are tested in this study: 1- In subjects with SCI and healthy subjects, the forward inclination of the pelvis and trunk in the sagittal plane will increase with increasing treadmill grade (uphill walking) and decrease with decreasing grade (downhill walking). This sagittal plane adaptation would be necessary to move the center of gravity forward and backward, respectively, during up-slope and down-slope walking in order to counteract the effect of gravity. 2- Subjects with SCI will increase their trunk and pelvic excursions as the treadmill slope increases in uphill walking whereas it will remain constant in healthy subjects. 3- Greater deficits in lower limb kinematic patterns in subjects with SCI will lead to higher trunk and pelvic excursions in sagittal, frontal and transverse planes.

METHODS

Subjects

Eight subjects with an incomplete SCI (two females and 6 males) and eight age and sex matched healthy subjects took part in the experiment. All subjects with SCI had an impairment level of "D" (mildly impaired) according to the ASIA scale (Ditunno et al., 1994), with some motor and sensory functions preserved below the neurological level (see Table). To be included in the study, every subject with SCI had to fulfill the following criteria: (1) Stable locomotor pattern over time (i.e. last twelve months); (2) Able to perform level and inclined treadmill walking without any assistance; (3) Stable dose of medications, if any. Every subject signed an informed consent document and the study was approved by the ethics committee of the Jewish Rehabilitation Hospital (Laval, Québec, Canada).

Walking assessment

The subjects walked on a treadmill at five different grades (from −10% to 10%) without any assistance. To prevent falls, each subject had to wear a safety overhead harness throughout the course of the experiment. The overhead harness was not used as a body-weight support device and did not interfere with the movements of the subjects. Several practice trials were used to familiarize the subjects with the different grades of treadmill walking and to determine the subjects' comfortable walking speed during inclined walking. In order to control for the effect of fatigue upon the gait pattern, the five walking grades were presented in random order. Subjects with SCI had to perform walking trials at their self-selected comfortable speed which was held constant for level and inclined conditions. This comfortable speed ranged from 0.04 to 0.67 m/s (average: 0.38 ± 0.27 m/s) for the group. Healthy subjects were selected to match the sex and age of each subject with SCI. Each healthy subject was required to walk at the same low speed as their counterpart subjects with SCI.

Subjects from both groups were also asked to perform static trials in which they had to maintain a standing position (5 seconds) at each treadmill grade.

Data recording

Pressure sensitive footswitches (piezoresistive insoles) were placed into subjects' shoes to signal the heel, fifth metatarsal, and hallux contacts. Signals from the footswitches were used to measure stance, swing and cycle durations. The footswitch signals were acquired at a sampling frequency of 1000 Hz per channel. The movements of the trunk, pelvis and lower limbs were acquired with four high-resolution cameras (BTS-ELITE system at 50 Hz) located on three posts positioned at 110° from each other. To model upper and lower limbs movements, reflective markers were placed bilaterally on the glenohumeral joint, posterior superior iliac spine (PSIS), anterior superior iliac spine (ASIS), greater trochanter, lateral femur epicondyle, lateral malleolus, calcaneus, fifth metatarsal as well as along the spine on the seventh cerebral vertebra (C_7) and the third lumbar vertebra (L_3).

Data analysis

Five qualitatively representative gait cycles at every treadmill grade were selected for the analysis of the spatio-temporal variables as well as the kinematic data. The spatio-temporal variables analysed were stride length, cycle duration as well as stance and swing phases reported in percentage of cycle duration. Kinematic analysis of the hip, knee and ankle joints was done in the sagittal plane. Segment and relative joint angles were calculated as shown on the kinematic model (see Figure 1), with the directions displayed indicating positive angles from neutral and the reversed for negative angles. Segment angles from the trunk and pelvis during walking were normalized from the standing position of the level condition recorded during static trials (the values obtained in the different walking grades

were subtracted to those obtained in quiet stance of the level condition). The hip and knee angles were calculated with respect to the proximal segment, with the neutral position in standing being taken as 0° displacement. For the ankle angle calculation, the neutral standing position with the leg axis perpendicular to the foot was taken as 0°.

Critical events of the gait cycle characterizing the adaptation process to inclined walking have been chosen to average joint and segment angles across subjects. For the lower limbs, the hip and ankle angles at foot contact as well as the knee angle during weight acceptance (peak value between 5 and 25% of the gait cycle) and late stance (peak value between 40 and 60% of the gait cycle) were averaged across subjects. For the trunk and pelvis, angular value during right single limb support (peak value from 20 to 50% of the gait cycle) were averaged across subjects and reported as trunk and pelvic inclination during walking. Curve fitting using linear regressions and polynomials were performed only when $r^2 > 0.70$. Statistical analyses for all the variables were done in Systat (SPSS) using a two-way ANOVA with repeated measures. When significant differences were found ($p < 0.05$), pairwise comparisons were made using the Tukey test.

RESULTS

Spatio-temporal variables

The effects of treadmill grade on spatio-temporal variables of gait are illustrated in Figure 2. When comparing subjects with SCI with healthy subjects, no significant differences could be found for the variables measured (Figure 2). However, subjects with SCI had a shorter cycle duration and a greater percentage of the gait cycle devoted to the stance phase at every treadmill grade as compared to

healthy subjects. The absence of significant differences between the two groups seems to be related to the variability in walking speed that ranged from 0.04 to 0.67 m/s.

Step lengths on right and left sides during inclined walking are shown in Figure 2, C-D. Healthy subjects showed an increase in step length during uphill walking and a decrease in downhill conditions on the right side only. On the left side, however, a decrease in step length was observed in downhill conditions whereas limited changes occur in uphill walking as expressed by a plateau between 0 and 10% grades. In subjects with SCI, different relationships were found between treadmill grade and step length for the right and left sides. Only three subjects (S1, S2 and S5) showed a bilateral pattern of changes that somewhat ressembled to the pattern found in healthy subjects. All the others showed either asymmetrical changes between right and left sides across walking grades (subjects S3, S4 and S7) or non-conventional patterns of changes (subjects S6 and S8). For instance, greater step length during level walking only was found in subject S6 (right side) and subject S8 (bilaterally)

Figures 3 to 6 show kinematic patterns from trunk and pelvic segments as well as lower limb joints during inclined walking. Instead of averaging the angular displacements for the entire group of subjects with SCI, these subjects were classified into 2 subgroups (see Table) based on their walking speeds. Subjects from subgroup 1 all showed minor deficits in lower limb kinematic patterns that resulted in moderate mobility dysfunctions during level and inclined walking. In contrast, the 3 subjects from subgroup 2 had major deficits in lower extremities that led to severe mobility dysfunctions as reflected by their very low walking speed (see Table). These subjects also showed great difficulty controlling the knee (hyperextension or hyperflexion) and the trunk (pronounced forward inclination) during level and inclined walking.

Slope walking with moderate speed deficits (subgroup 1)

Sagittal plane motions from trunk, pelvis and lower limbs during inclined walking in one subject with SCI (subject S3) belonging to the subgroup 1 are illustrated in Figure 3-A. Even if the walking speed of subject S3 (0.63 m/s) was lower than the comfortable speed of healthy subjects, the trunk, pelvic as well as lower limb joint patterns of movements were very similar to those of healthy subjects. The trends of adaptation to uphill and downhill walking were the same as those observed previously in healthy subjects walking at comfortable speed (Leroux et al., 2001). Increasing the treadmill grade from 0 to 10% led to an increasingly flexed posture of the hip, knee and ankle at foot contact as well as a progressive forward inclination of pelvis and trunk. On the other hand, decreasing the treadmill grade from 0 to –10% led to a decreasingly flexed posture of the hip at foot contact as well as an increase in knee flexion during weight acceptance and late stance. These changes were accompanied by a gradual backward inclination of trunk as the treadmill grade went steeper. Subgroup averages from the five subjects with SCI as well as the five matched healthy subjects are shown in Figure 3 (B, C and D). As previously seen in healthy subjects (Leroux et al., 2001), the different treadmill grades did not have any effect on trunk and pelvic inclinations during static condition. Both trunk and pelvis were always maintained in a constant vertical position across grades. The relationship between treadmill grade and trunk inclination during walking was, however, somewhat different between subjects with SCI and healthy subjects. Healthy subjects showed a strong linear relationship (r^2 = 0.95) whereas a curvilinear function (r^2 = 0.91) best described the relationship between trunk inclination and treadmill grade in subjects with SCI. This non linear relationship was characterized by a much steeper slope in uphill walking as compared to downhill conditions. The relationship between pelvic inclination and treadmill grade also revealed linear (r^2 = 0.79) and curvilinear (r^2 = 0.91) functions for healthy subjects and subjects with SCI respectively. When considering trunk and pelvic

postures at any treadmill grade, it could also be seen that subjects with SCI always maintained a greater forward inclination of these two segments as compared to healthy subjects. Walking up-slope or down-slope did not change significantly the total angular excursion (TAE) of trunk and pelvic movements in both groups of subjects (Figure 3-C). The only difference between healthy subjects and subjects with SCI was found in the TAE across grades. For any treadmill grade, subjects with SCI always showed a slightly greater excursion from both trunk and pelvis as compared to healthy subjects.

The effects of treadmill grade on lower limb joint angulation during critical events of the gait cycle are shown in Figure 3-D. For each of these joints, the adaptation to inclined walking was very similar between the subgroup of subjects with SCI and the matched healthy subjects. Furthermore, these patterns of adaptation closely resembled to those previously described for healthy subjects walking at comfortable speed (Leroux et al., 2001). The only difference in the relationship between treadmill grade and lower limb joint angulation was observed at the ankle. A strong linear function ($r^2 = 0.97$) was found in healthy subjects whereas a curvilinear function ($r^2 = 0.99$) best expressed the relationship between ankle angulation and treadmill grade in subjects with SCI (Figure 3-D). This difference was mainly attributed to the flattened of the curve observed in downhill conditions in subjects with SCI. Another difference between the two groups of subjects was observed at the knee joint. Even if the relationship between treadmill grade and knee angulation (at weight acceptance and late stance) was very similar between the two groups, subjects with SCI always showed a greater knee flexion at any treadmill grade.

Trunk and pelvic motions in frontal and transverse planes during inclined walking are illustrated in Figure 4. Frontal plane motions from one subject with SCI belonging to the subgroup 1 (Figure 4-A) revealed that lateral trunk displacement was minimally changed during treadmill inclined walking. Furthermore, the patterns of movements were similar to those observed in the matched healthy subject. The lack of relationship between treadmill grade and trunk angular displacement in the frontal plane was observed in every subject with SCI composing this subgroup of subjects. This absence of grade effect was also reflected by minimal changes in trunk TAE across grades in both subgroups of subjects (Figure 4-C). For the pelvis, the lateral drop (toward the swinging limb) occurring during swing phases was appropriately modified (decreased in uphill and increased in downhill) in only one subject with SCI (Figure 4-B). The other subjects composing this subgroup showed either no changes or an increase of pelvic lateral drop in both uphill and downhill walking as compared to level walking. On the other hand, three out of five subjects from the healthy subgroup showed appropriate changes in pelvic tilt and the other two subjects exhibited no changes across walking grades. When considering the whole subgroup of healthy subjects, increasing the treadmill grade from −10% downhill to 10% uphill led to a progressive decrease in pelvic excursion in the matched healthy subjects. However, the magnitude of change was very limited (3°) when comparing the extreme slope conditions.

Trunk and pelvic rotations as measured by shoulder and pelvic girdle angles in the transverse plane were minimally affected by treadmill grade during walking in subjects with SCI from the subgroup 1 and the matched healthy subjects (Figure 4, E-F). These patterns of movements could vary from one grade to the other but there were no consistent changes across treadmill grades as shown in one subject with SCI (Figure 4, E-F). The only difference that could be observed between subjects with

SCI and matched healthy subjects was not related to the effect of treadmill grade but to the basic kinematic pattern. The structure of the pattern was similar in every subject (one positive and one negative peak) but the timing of the reversal points could vary (Figure 4, E-F). The TAEs of the trunk and the pelvis did not show any significant changes during inclined walking in both subgroups of subjects (Figure 4, G-H). However, as seen for the sagittal plane, subjects with SCI always showed a slightly greater excursion for any treadmill grade.

Slope walking with severe gait deficits (subgroup 2)

One selected subject with SCI (subject S6) walking at low speed (0.18 m/s) during level, uphill and downhill conditions is shown in Figures 5 and 6. This subject was more affected in the right lower limb as reflected by differences in step lengths (see Figure 2) and in basic lower limb kinematic patterns. When comparing left and right lower limbs, the left hip and knee showed greater angular excursions, especially from late stance to heel contact (Figure 5, D-E). Furthermore, lower limb kinematic patterns from the left side were more similar to those of the matched healthy subject. During inclined walking, left and right lower limb joints of subject S6 showed some degree of adaptation at the ankle joint (Figure 5, D-E). On the other hand, the changes were very limited at the hip joints while the two knees showed hyperextension throughout the stance phase and, thus, could not adapt to uphill and downhill walking.

Abnormal adaptation profiles were also found in trunk and pelvic kinematic patterns during level and inclined walking in the sagittal plane. Even if kinematic patterns were similar between subject S6 and the matched healthy subject, both trunk and pelvis showed a much more pronounced forward inclination during walking at any treadmill grade in subject S6 (Figure 5-A). Both subjects showed

minimal changes across treadmill grades during walking (Figure 5-B). The main difference was observed at the trunk where subject S6 exhibited a much greater forward inclination at any grade during walking. During static condition, subject S6 maintained the trunk and pelvis vertically aligned at every treadmill grade (Figure 5-B). Treadmill grade did not lead to major changes in trunk and pelvic TAEs either during walking (Figure 5-C). Both subjects showed minor changes across grades and pelvic angular excursion was greater at any grade in subject S6 as compared to the matched healthy subject.

Frontal and transverse plane motions revealed major differences between subject S6 and the matched healthy subject but these differences were not related to the treadmill grade effect. In fact, the two subjects showed distinct trunk and pelvic kinematic patterns that were only minimally affected by treadmill grade. As shown in Figure 6 (A-B and E-F), the walking pattern of subject S6 was characterized by larger lateral displacement and greater rotation from the trunk and pelvis as compared to the matched healthy subject. During inclined walking, trunk and pelvic patterns of subject S6 could somewhat vary from one grade to the other but no significant trends could be detected in frontal and transverse planes. Furthermore, treadmill grade did not lead to significant changes in trunk and pelvic TAEs during walking (Figure 6, C-D and G-H). However, subject S6 always showed much greater excursions in the two planes at any treadmill grade as compared to the matched healthy subject.

Pelvic kinematic patterns in the frontal plane showed that the two peaks occurring shortly after each toe-off (at 35 and 85% of the gait cycle for left and right toe-off respectively) were uneven in subject S6 (Figure 6-B). For any treadmill grade, the peaks occurring after right toe-off were always greater

than those occurring after left toe-off. In fact, the pelvis showed an increased angulation of about five degrees (all grades confounded) when comparing peaks related to right and left toe-off. This reveals that the pelvis had to be elevated on the right side to assist the right leg swinging through.

Deficits in lower limb movements as well as trunk and pelvic control during level and inclined walking were also found in subjects S7 and S8 from subgroup 2 (not shown in the figures). Subject S7 walked at a very slow speed (0.09 m/s) and had the shortest step lengths of the entire group of subjects with SCI. This subject showed a decrease in right hip excursion and knee hyperextension on the left side during walking. Kinematic patterns from the trunk and pelvis in the sagittal and frontal planes showed very similar results as in subject S6 during level and inclined walking. The trunk and pelvis exhibited a pronounced forward inclination in every treadmill grade in the sagittal plane while greater lateral excursions from these two segments were found in the frontal plane as compared to the matched healthy subjects. Furthermore, these lateral leanings were related to the toe-off periods as seen in subject S6. In the transverse plane, on the other hand, the trunk and the pelvis showed very limited rotations (lower than the matched healthy subject) in level and inclined walking.

Abnormalities in trunk and pelvic kinematic patterns were also found in subject S8. This subject walked at a very slow speed (0.04 m/s) and almost 90% of the entire cycle was devoted to the stance phase. Lower limb kinematic patterns revealed greater deficits on the left side as expressed by very limited excursions at the hip, knee and ankle joints and by a complete absence of adaptation during inclined walking. Kinematic patterns from the trunk and pelvis in the sagittal plane showed that these two segments performed sudden backward movements at the initiation of the left swing phase whereas a pronounced forward inclination of the pelvis was achieved after right toe-off during level

and uphill walking. For the 10% downhill condition, the trunk and pelvis performed forward movements of great amplitudes from left toe-off to the right heel contact and reached forward inclinations of 43° and 23° respectively. These pronounced forward inclinations often led to a loss of balance during downhill walking. Results from frontal plane showed a pronounced trunk leaning toward the swinging limb that also led to a loss of stability during level and inclined walking. These lateral movements were accentuated in the downhill conditions. In the transverse plane, sudden rotations of large amplitudes were performed from the trunk and the pelvis to assist the right leg swinging through during level and uphill walking.

Coordination of the pelvic and trunk segments in slope walking

Pelvic-trunk coordination patterns showing the average of the healthy subjects walking at self-selected comfortable speed (1.03 m/s) revealed that movements in the sagittal plane were greater for the pelvis than for the trunk (Figure 7-A). The same coordination pattern observed in level was preserved during inclined walking. However, as previously shown in Figure 3, both trunk and pelvis showed a forward inclination during uphill walking and a backward inclination during downhill walking as indicated by shifts towards upper right and lower left, respectively. On the other hand, for lower walking speed (0.18 m/s), the cyclographs of one healthy subject (this subject walked at the same speed as subject S6 in Figure 7-C) showed widening loops indicating less tightly coupled motions between the pelvis and trunk, especially during downhill walking, with larger trunk motions (Figure 7-B). Pelvic-trunk coordination pattern of subject S6 greatly differed from those found in the healthy subjects at matched and comfortable speeds (Figure 7-C). The difference was mainly attributed to the very large excursions from the trunk as illustrated by a spiral shape observed in level and inclined walking. During inclined walking, the same coordination pattern was preserved (with

minor modifications) but the patterns were not shifted appropriately. Instead of showing a progressive increase in pelvic and trunk forward inclination as the treadmill grade increased from –10 to 10%, subject S6 showed a greater degree of trunk forward inclination in downhill condition as indicated by the upper shift of the curve.

In the frontal plane, pelvic and trunk movements occurred in opposite directions during most of the gait cycle in the healthy subjects walking at comfortable speed (Figure 7-A). The basic pelvis-trunk coordination pattern observed in level walking was influenced by the change in treadmill slope. The cyclographs showed widening loops indicating less tightly coupled motions between the trunk and pelvis, especially during downhill walking, with large pelvic motions. This pelvic-trunk coordination pattern was greatly affected by the decrease in walking speed as expressed by the change in the general orientation of the curves in the matched healthy subject (Figure 7-B). At low walking speed, the cyclograph curves were oriented toward the ordinate indicating greater amplitude of movements from the trunk. During inclined walking, the cyclographs also showed greater excursions in both trunk and pelvic segments as expressed by the increased surface area of this curve. In contrast to the matched healthy subject, pelvic-trunk coordination pattern of subject S6 in the frontal plane showed some degree of coupling as illustrated by diagonal curves with positive slope (Figure 7-C). This indicated that trunk and pelvic movements were performed in the same direction during periods of double limb support during level and inclined walking. During downhill walking, however, the cyclographs showed widening loops indicating less tightly coupled movements between the trunk and pelvis, with large pelvic motions.

In the transverse plane, both walking inclinations and speeds led to modifications in pelvic-trunk

coordination patterns in the healthy subjects walking at comfortable speed (Figure 7-A). Results showed that trunk and pelvic rotations were not performed in synchrony during most of the gait cycle in level walking (Figure 7-A). Furthermore, during uphill walking, a rectangular cyclograph parallel to axes was observed, indicating that the two segments never moved simultaneously. A more synchronous coordination was observed during downhill walking, when the trunk and the pelvis rotated in the same direction from the late period of single limb support to the end of the double limb support period. As opposed to comfortable speed, the tightness of the loops displayed in the cyclographs of the matched healthy subject revealed a large degree of synchrony in trunk and pelvic movements, especially in level and downhill walking (Figure 7-B). A similar trend, though more pronounced as illustrated by greater tightness of the loops, was observed in subject S6 during both level and inclined walking (Figure 7-C).

Trunk and pelvic total angular excursions averaged across walking grades for the healthy subjects and for the two subgroups of subjects with SCI are illustrated in Figure 8. Healthy subjects always showed lower excursions from the trunk and the pelvis for each plane as compared to the two subgroups of subjects with SCI. When comparing these two subgroups, trunk and pelvic excursions were greater in subgroup 2 in every plane.

DISCUSSION

Results from the present study showed that, following spinal cord injury, trunk and pelvic movements are affected by the extent of lower limb deficits during walking. Subjects with SCI showing moderate mobility dysfunctions were the ones who exhibited near normal trunk and pelvic movements in the three planes of motions. However, these subjects showed irregularities in their trunk and pelvic

inclinations (sagittal plane) during inclined walking. When comparing results of subjects belonging to subgroup 1 with their matched healthy subjects, subjects with SCI always maintained a greater forward inclination of both trunk and pelvis during level and inclined walking, especially in downhill conditions where both segments were maintained at about the same level as in level gait. These results did not confirm our first hypothesis that the forward inclination of the pelvis and trunk would decrease with decreasing grade in downhill walking. What is the mechanism behind this postural modification observed in these subjects with SCI? It is tempting to associate this increase in forward inclination with the increased knee flexion observed in this subgroup of subjects. It has been previously shown that increased knee flexion during walking was associated with a forward inclination of the trunk (Grasso et al., 2000). The averaged data obtained from the subgroup 1 also showed an increase in knee flexion during stance that could be associated with trunk and pelvic forward inclination. However, knee flexion and trunk and pelvic inclination were poorly correlated for the individual subjects. As an example, subject S5, who walked with a pronounced flexion of both knees (from 20 to 30°) during stance, showed a normal inclination of the trunk and the pelvis throughout the gait cycle. Thus, we believe that the forward inclination of trunk and pelvis is not the direct consequence of the increased knee flexion in this subgroup of subjects. Another explanation could be related to the maintenance of balance during walking. The increase in forward inclination could be a strategy to compensate for a certain degree of instability due to lower limb deficits in these subjects with SCI. By moving their center of gravity more anteriorly over the base of support, these subjects would benefit from a greater stability. The same strategy of inclining the trunk forward during walking has been reported for subjects with balance deficits due to vestibulopathy (Krebs et al., 1992). Results obtained from two subjects belonging to subgroup 2 (subjects S6 and S8) showing major deficits in lower limb movements during walking also support this assumption on the role of the

trunk after spinal cord injury. These two subjects were the ones who had the greatest difficulty to maintain balance when performing walking trials. Furthermore, in everyday situation, they required the use of walking aids for ambulation (see Table). The great difficulty of performing walking trials from subjects S6 and S8 without any assitive devices was reflected by several losses of balance, especially in downhill conditions. This walking unstability was associated with an important forward inclination of the trunk during level and inclined walking. In fact, subjects S6 and S8 showed the greatest forward inclination of the whole group of subjects with SCI. Furthermore, this forward inclination was more pronounced in downhill walking which was considered by the subjects to be the most difficult condition to perform.

Interaction between upper and lower limb movements during walking

To compensate for lower limb deficits, it was expected that trunk and pelvic movements would increase with increasing treadmill slope during uphill walking in subjects with SCI. To be more specific in identifying compensatory movements from trunk and pelvis that could contribute to the forward momentum of the body, it is important to focus on the period corresponding to the stance to swing transition. This is a critical period where the body organizes itself to generate enough momentum to move forward. In the present study, all subjects belonging to subgroup 2 presented with push-off weakness. This weakness was reflected, in part, by a decrease in hip flexion during swing on the more affected side. All three subjects showed compensatory movements from trunk and/or pelvis in all three planes of movements to overcome lower limb deficits during walking. However, these compensatory movements were not increased during uphill walking. Sagittal plane motions analyses revealed that subject S8 performed sudden backward inclination of the trunk that resulted in a forward thrust of the pelvis at the initiation of swing on the more affected side. This

sudden forward thrust seems to be performed to pull the hip forward and assist the flexion of this joint. A similar strategy involving the pelvis has been found in subjects with unilateral hip pain due to osteoarthritis (Murray et al., 1971; Thurston, 1985). Subjects with hip pain accentuated posterior tilting of the pelvis at the initiation of the swing phase to help the swinging painful leg to move forward and counteract restricted hip movements (Murray et al., 1971; Thurston, 1985). Subject S8 from the present study also performed large trunk and pelvic rotations during right swing phase. These rotations are more likely performed to assist the right leg swinging through. Thurston (1985) also reported large rotations of the lumbar spine during the swing phase of the affected side in subjects with unilateral osteoarthritis of the hip joint. These rotations were performed to assist the painful swinging leg to move forward (Thurston, 1985). Thus, as found in subjects with osteoarthritis of the hip joint, subject S8 used compensatory movements from the trunk and pelvis to compensate for lower limb deficits during walking. Frontal plane analyses revealed that the other two subjects of subgroup 2 (subjects S6 and S7) performed greater pelvic elevation at the initiation of swing on the more affected side. This pelvic tilt seems to be performed to compensate for push off weakness and provide an adequate toe-floor clearance during swing of the more affected limb. Similar lateral tilt of the pelvis has been reported in subjects with unilateral osteoarthritis of the hip joint and was attributed to the limited motion of the hip observed in these subjects (Murray et al., 1971; Thurston, 1985).

Our results do not confirm the second hypothesis that accentuation in trunk and pelvic movements would increase with increasing treadmill grade during uphill walking. In fact, once these subjects with SCI adopted a locomotor strategy including trunk and pelvic segments, they did not modify it throughout walking grades. This means that they did not increase or decrease trunk and pelvic range

of motion to generate more propulsion or absorb more energy during uphill and downhill walking respectively.

Results of the present study confirm our third hypothesis that greater deficits in lower limb kinematic patterns in subjects with SCI led to higher excursions in trunk and pelvic motions in all three planes (see Figure 8). This increase in trunk and pelvic range of motion can be attributed to at least two factors: 1) compensatory movements to overcome lower limb deficits (as discussed previously) and 2) difficulty to maintain the trunk stable during specific phases of the gait cycle. The second factor was well illustrated in subjects S6 and S7 when examining trunk lateral movements in the frontal plane. These two subjects showed pronounced leaning of the trunk toward the supporting limb following each heel contact. We believe that these pronounced lateral movements are associated with deficits in lower limb patterns during walking. Subjects S6 and S7 walked with knee hyperextension on one side (subject S7) or both sides (subject S6). The lack of knee absorption following each heel contact did not provide the normal damping effect that contribute to smooth the normal walking pattern (Saunders et al., 1953). Larger lateral excursions from the trunk following each heel contact might have been the consequence of the stiff knees observed in these subjects with SCI.

Relationship between trunk and pelvic movement

Cyclographs from one subject with SCI (subject S6) showed that the relationship between trunk and pelvic movements was changed during level and inclined walking following SCI. These changes were mainly attributed to excessive amplitude of movements at the trunk (see Figure 7, sagittal plane) as well as modifications in the sequence of trunk-pelvic movements. The coupling found between trunk and pelvic movements in the frontal plane resulted from both excessive trunk leaning toward the

supporting limb after heel contact and upward tilt of the pelvis toward the swinging leg. These abnormal movements, peformed simultaneously, were mainly caused by bilateral knee hyperextensions as discussed above. In the transverse plane, the tight coupling found between trunk and pelvic movements during level and inclined walking in subject S6 can be explained by the very slow walking speed of this subject. The fact that the matched healthy subject also walking at very slow speed behaved the same way as subject S6 strongly favors the walking speed as the main factor responsible for these modifications in trunk-pelvic relationship. According to our results, this tight coupling mainly resulted from changes in the trunk sequence of movement. At very low speed, this segment rotated in synchrony and in the same direction as the pelvis in both subject S6 and the healthy subject. This change might be explained by the fact that for low walking speeds, the damping action of the trunk, which is usually provided by rotations in opposition to the pelvis (Stokes et al., 1989), is no longer required. Furthermore, by rotating together with the pelvis, the trunk can contribute to the generation of the forward momentum during walking at slow speed.

Relationship between locomotor adaptability and ASIA score

The ASIA impairment scale (Ditunno et al., 1994) was used to assess motor and sensory functions following SCI. In the present study, subjects with SCI all showed a loss of voluntary motor function as well as a decrease of sensory function (see ASIA scores in Table). However, both motor and sensory scores were not correlated to the ability to adapt to inclined walking. As an example, subject S6, who showed inappropriate adaptation in trunk and pelvic segments as well as limited adaptation in all three joints during inclined walking (see Figure 5), had greater motor and sensory scores than subject S3 who showed near normal adaptation to inclined walking (see Figures 3 and 4).

The treadmill walking speed, on the other hand, seems to be a good indicator of the ability to adapt to inclined walking following SCI. Subjects belonging to subgroup 1 had faster comfortable treadmill speed and exhibited near normal patterns of adaptation during inclined walking. In contrast, subjects belonging to subgroup 2 had a very low walking speed and showed very limited changes in kinematic patterns during inclined walking.

Effect of spinal cord lesion on postural control

In the present discussion, a lot of emphasis has been placed on biomechanical factors when discussing alterations in trunk and pelvic kinematic patterns during level and inclined walking. However, it is also important to consider the direct effect of the spinal cord lesion on trunk and pelvic control. It has been shown that, following surface translations, automatic postural responses were impaired in subjects with SCI (Gross et al., 1998). These impaired responses were characterized by inappropriate segment alignments and delayed or absent muscle activities (Gross et al., 1998). These results indicate that intact sensory and motor spinal pathways are needed for appropriate automatic responses in humans. In the present study, as stated above, subjects with SCI showed a loss of motor and sensory functions. Thus, automatic postural adjustments required when performing level and inclined walking might be impaired and lead to inappropriate patterns of trunk and pelvic movements following SCI. For instance, the sudden trunk flexion observed in subject S8 after left heel contact in level and inclined walking might be the result of delayed or absent EMG responses from trunk extensors (mainly the erector spinae). Accentuation in trunk lateral movements after each heel contact in subjects S6 and S7 could also be related to inappropriate activation of muscles responsible for trunk stabilization. The use of intra-muscular recording (to counteract high skin impedance) from trunk

muscles would likely provide useful information on the effect of spinal cord lesion on postural control during level and inclined walking.

CONCLUSION

Results from this study showed that postural adaptation to inclined walking is impaired following SCI. We believe that trunk and pelvic forward inclination observed in subjects with SCI at any treadmill grade is performed to provide greater stability during walking. This forward inclination would in fact counteract decreased balance skills due to lower limb deficits and possible alterations in automatic postural responses after SCI. Compensatory movements from trunk and pelvis performed by subjects with SCI during walking did not vary across treadmill grade. However, subjects with SCI showing severe mobility dysfunctions had marked changes in trunk and pelvic angular excursions in all three planes of motion during uphill and downhill walking. These modifications suggest a possible involvement of the axial segments in inclined walking when the lower extremity functions are severely impaired due to SCI.

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Table: Profile of subjects with spinal cord injury

Subjects	Age	Cause and Level of Injury	Time since Injury (years)	Walking Aids	Treadmill Walking Speed (m/s)	ASIA Score[1] Motor R (/50)	Motor L (/50)	Sensory R (/112)	Sensory L (/112)
SUBGROUP 1									
S1	42	Fall, T4-T5, T7-T8	3	None	0.67	48	46	104	105
S2	39	SDA[1], T8	10	None	0.63	49	49	78	78
S3	39	MVA[2], C2-C3	2	None	0.63	45	36	71	66
S4	50	MVA[2], T11	6	One cane	0.54	40	47	96	98
S5	35	MVA[2], C6-C7	3	None	0.22	41	46	94	67
SUBGROUP 2									
S6	45	SDA[1], T2-T10	6	One cane	0.18	46	48	106	91
S7	61	Virus, C5-C6	3	Forearm crutches	0.09	42	41	88	108
S8	31	MVA[2], C5-C6	9	Forearm crutches	0.04	48	33	101	108

[1]All subjects had an impairment level of "D" (mildly impaired)
[2]SDA = Scuba diving accident
[3]MVA = Motor vehicle accident

FIGURE LEGENDS

Figure 1: Segment and joint angle computations are illustrated for the sagittal plane (left) as well as frontal and transverse planes (right). For every segment and joint angles, the arrow direction indicates positive angles from neutral.

Figure 2: A-B, Average cycle duration (mean ± 1 SD) as well as stance and swing in percent of cycle duration (mean ± 1 SD) during level and inclined walking in subjects with SCI and healthy subjects. C-D, Right and left step lengths for each individual subject with SCI as well as for the group of healthy subjects. Each symbol represents a subject with SCI and the dotted lines indicate right and left step length averages for the group of healthy subjects.

Figure 3: A, Angular displacements of the trunk, pelvis, hip, knee and ankle normalized to the gait cycle (from one foot contact, 0% to the next, 100%) during level, uphill and downhill walking in the sagittal plane. The results are shown for one subject with SCI (subject S3) belonging to subgroup 1 and the matched healthy subject during level walking only. Arrows at the bottom of the graphics are used to indicate periods of heel contact (downward direction) and toe-off (upward direction) as well as events related to the right (bold character) or left (regular character) lower limbs. The same convention is used for all figures. B, Average inclination (mean ± 1 SD) of the trunk and pelvis for subgroup 1 and the matched healthy subjects during single limb support in standing and walking conditions across walking grades. C, Average total angular excursion (mean ± 1 SD) of the trunk and pelvis for subgroup 1 and the matched healthy subjects across walking grades. D, Average hip, knee

and ankle angles (mean ± 1 SD) for subgroup 1 and the matched healthy subjects at critical events of the gait cycle.

Figure 4: Angular displacements of the trunk and pelvis normalized to the gait cycle (from one foot contact, 0% to the next, 100%) during level, uphill and downhill walking in frontal (A-B) and transverse (E-F) planes. The results are shown for one subject with SCI belonging to subgroup 1 (subject S3) and the matched healthy subject during level walking only. The average total angular excursion (mean ± 1 SD) of the trunk and pelvis for subgroup 1 and the matched healthy subjects across walking grades are shown in frontal (C-D) and transverse (G-H) planes.

Figure 5: A, Angular displacements of the trunk and pelvis normalized to the gait cycle (from one foot contact, 0% to the next, 100%) during level, uphill and downhill walking in the sagittal plane. The results are shown for one subject with SCI belonging to subgroup 2 (subject S6) and the matched healthy subject during level walking only. B, Average inclination (mean ± 1 SD) of the trunk and pelvis for subject S6 and the matched healthy subjects during single limb support in standing and walking conditions across walking grades. C, Average total angular excursion (mean ± 1 SD) of the trunk and pelvis for subject S6 and the matched healthy subjects across walking grades. D-E, Angular displacements of the hip, knee and ankle normalized to the gait cycle (from one foot contact, 0% to the next, 100%) during level, uphill and downhill walking in the sagittal plane. The results are shown for subject S6 and the matched healthy subject during level walking only.

Figure 6: Angular displacements of the trunk and pelvis normalized to the gait cycle (from one foot

contact, 0% to the next, 100%) during level, uphill and downhill walking in frontal (A-B) and transverse (E-F) planes. The results are shown for one subject with SCI belonging to subgroup 2 (subject S6) and the matched healthy subject during level walking only. The average total angular excursion (mean ± 1 SD) of the trunk and pelvis for subject S6 and the matched healthy subjects across walking grades are shown in frontal (C-D) and transverse (G-H) planes.

Figure 7: Cyclographs showing the coordination between pelvic and trunk movements during level, uphill and downhill walking in all three planes. The results are shown for the average of the 8 normal subjects walking at comfortable speed (1.03 m/s) (A) as well as for one subject with SCI (subject S6) (C) and the matched healthy subject (B). The dotted lines represent the period of double limb support and the solid lines represent the period of single limb support for the right (R) and left (L) limbs. The arrows indicate the right heel contact event as well as the direction of the movement for the level condition. For each of the three anatomical planes, a diagonal curve showing a positive slope reveals that the pelvic and trunk segments move in synchrony and in the same direction whereas a negative slope indicates synchronous movements in opposite directions. A square or rectangular pattern parallel to axes indicates that the two segments never move simultaneously throughout the gait cycle.

Figure 8: Total angular excursions of the trunk and pelvis for healthy subjects as well as subgroups 1 (n = 5) and 2 (n = 3) in all three planes of motions. For each plane, the data from the five treadmill grades (from −10 to 10%) were averaged together and plotted as a mean value for each subgroup.

Figure 1

Angle computations

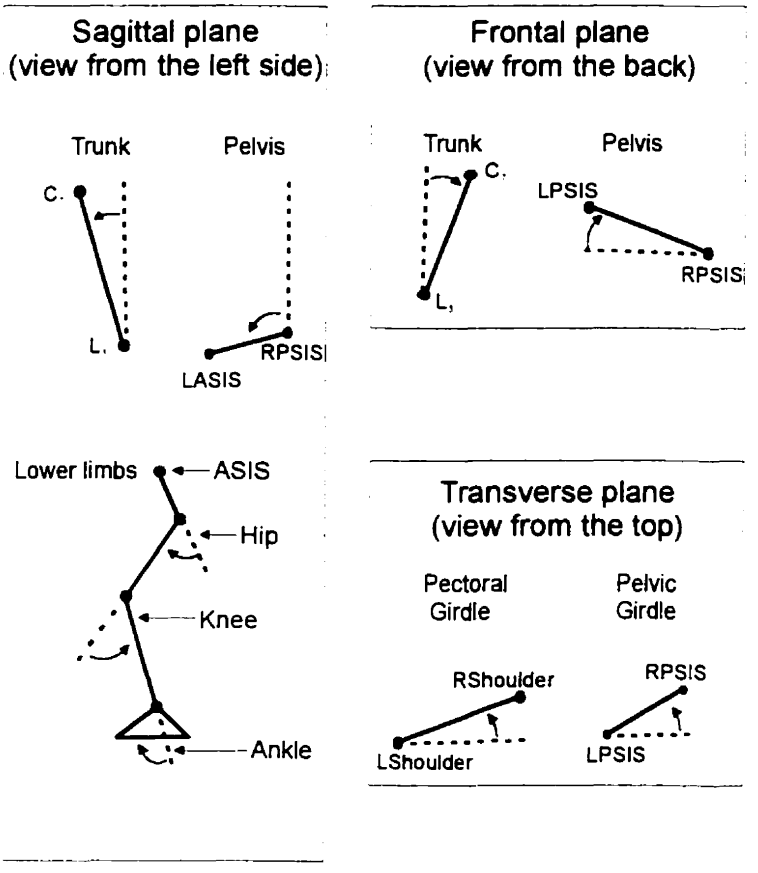


FIGURE 2

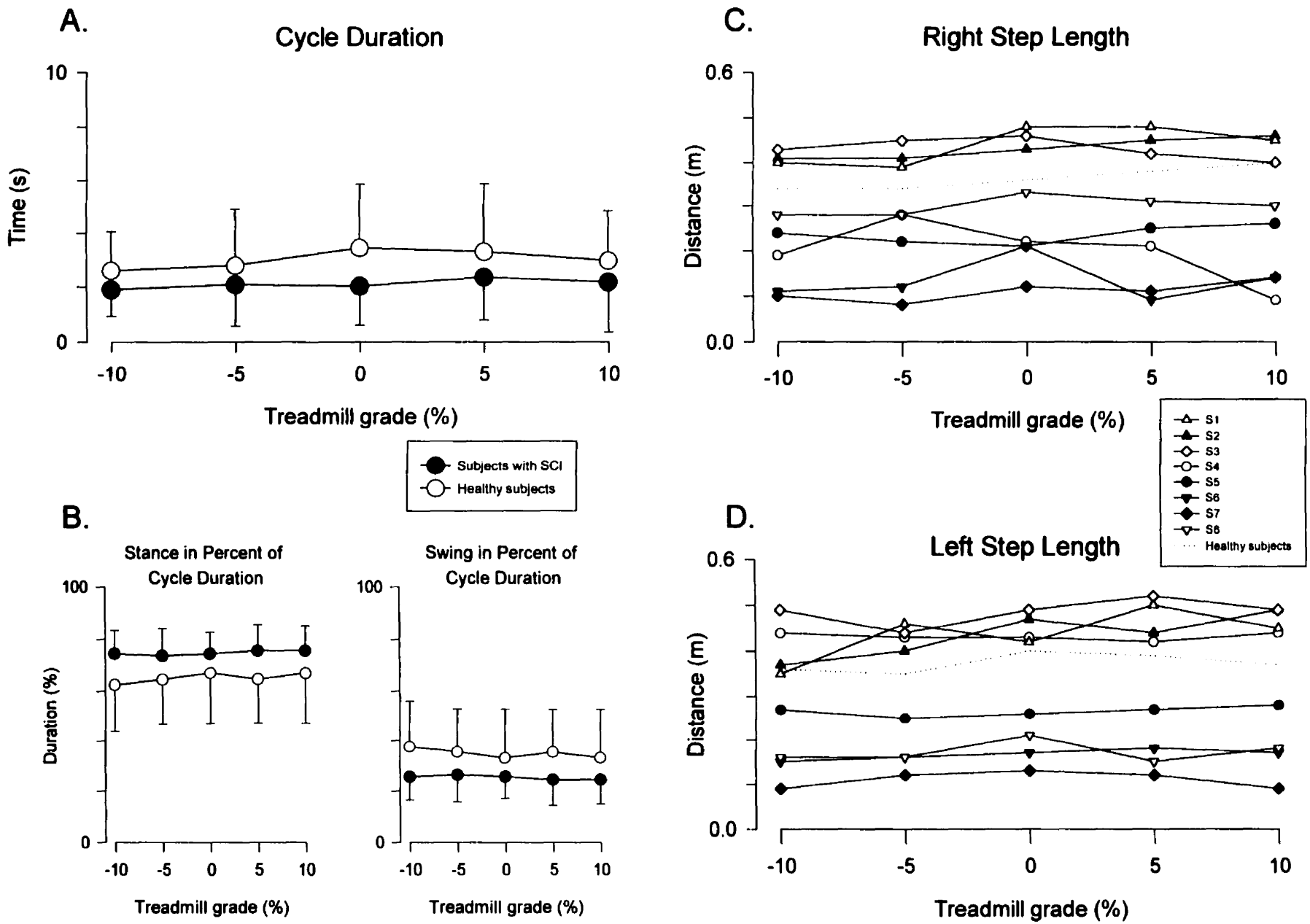

A.
Cycle Duration
Time (s)
10
0
-10
-5
0
5
10
Treadmill grade (%)
Subjects with SCI
Healthy subjects
B.
Stance in Percent of Cycle Duration
Swing in Percent of Cycle Duration
Duration (%)
100
0
Treadmill grade (%)
C.
Right Step Length
Distance (m)
0.6
0.0
Treadmill grade (%)
D.
Left Step Length
S1
S2
S3
S4
S5
S6
S7
S8
Healthy subjects

FIGURE 3

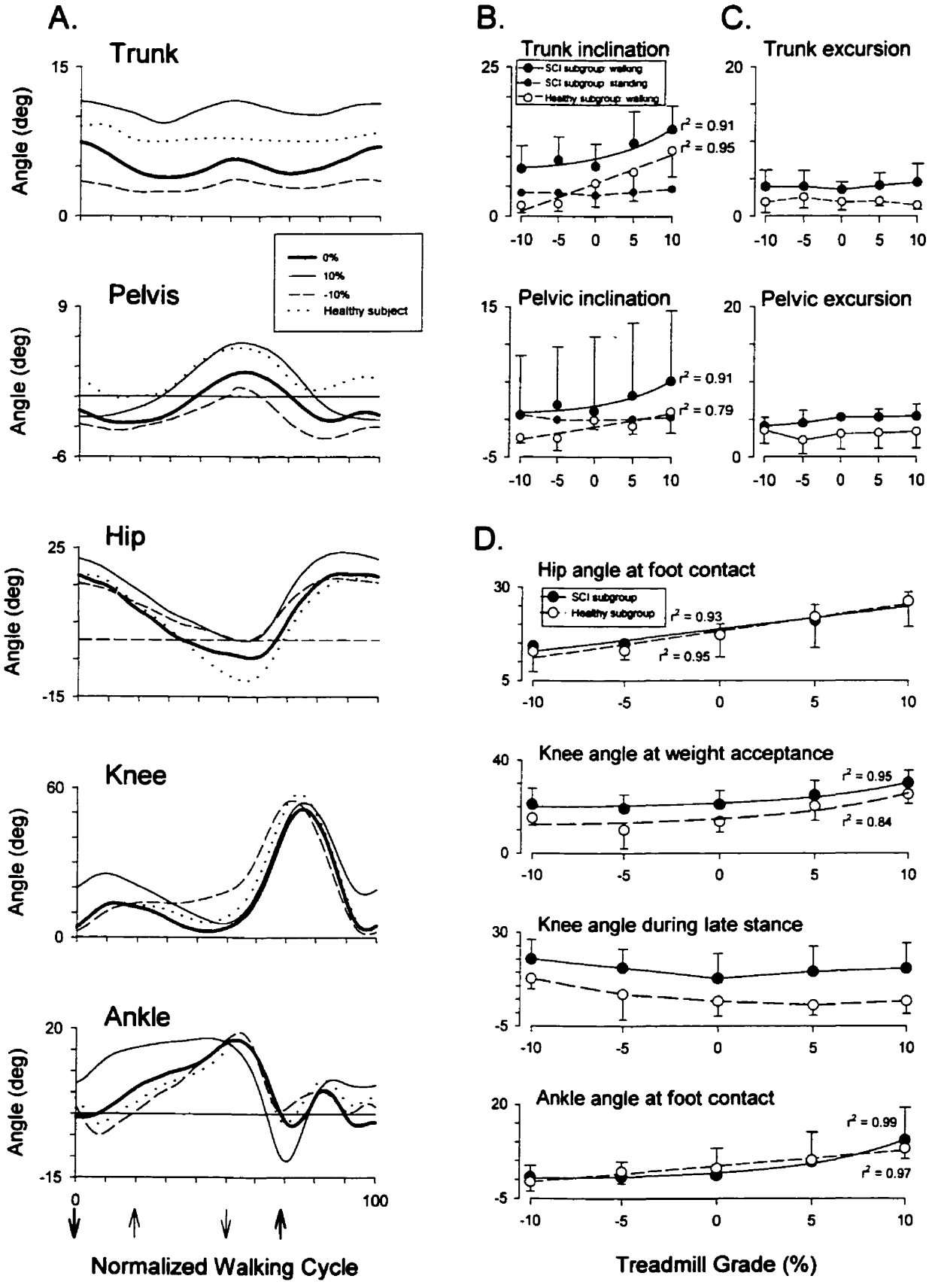
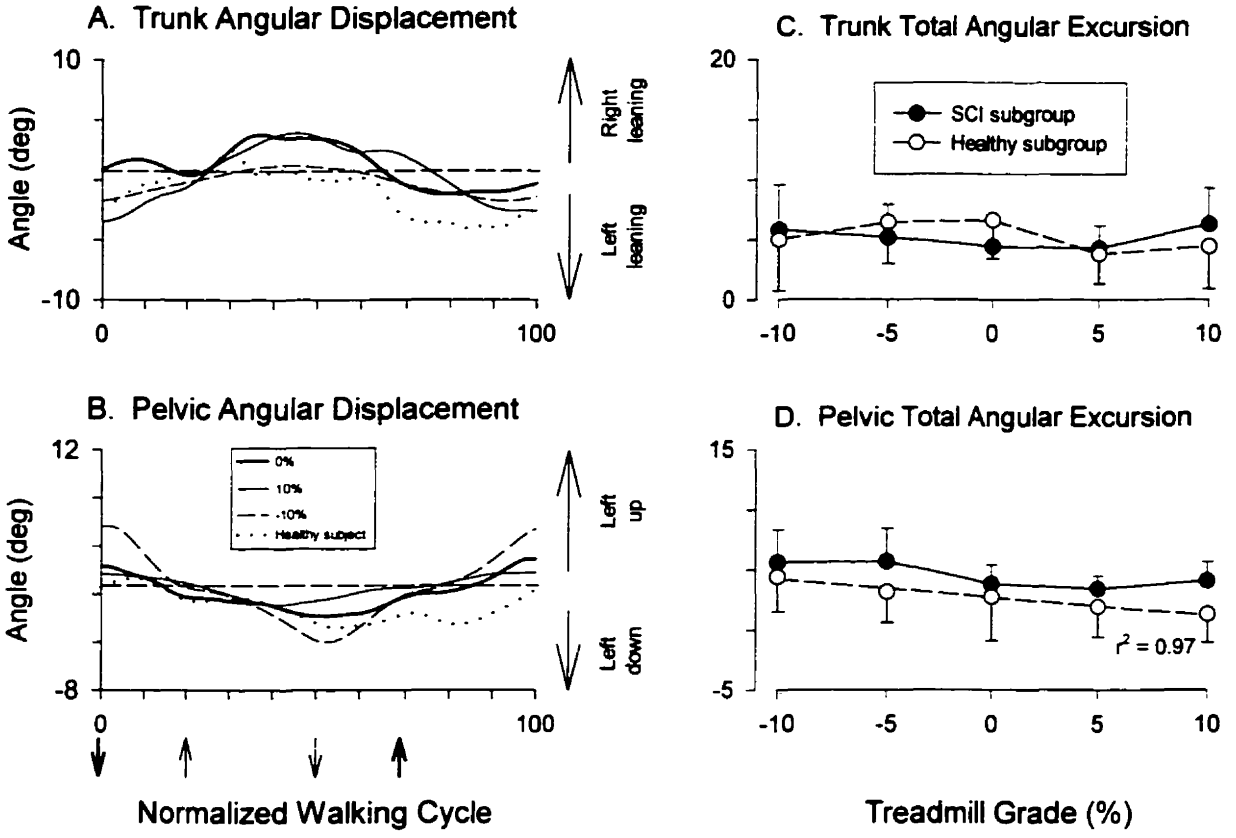

A.
Trunk
Pelvis
Hip
Knee
Ankle
Angle (deg)
0%
10%
-10%
Healthy subject
Normalized Walking Cycle
B.
Trunk inclination
SCI subgroup walking
SCI subgroup standing
Healthy subgroup walking
r² = 0.91
r² = 0.95
Pelvic inclination
r² = 0.91
r² = 0.79
C.
Trunk excursion
Pelvic excursion
D.
Hip angle at foot contact
SCI subgroup
Healthy subgroup
r² = 0.93
r² = 0.95
Knee angle at weight acceptance
r² = 0.95
r² = 0.84
Knee angle during late stance
Ankle angle at foot contact
r² = 0.99
r² = 0.97
Treadmill Grade (%)

FIGURE 4

FRONTAL PLANE MOTIONS



TRANSVERSE PLANE MOTIONS

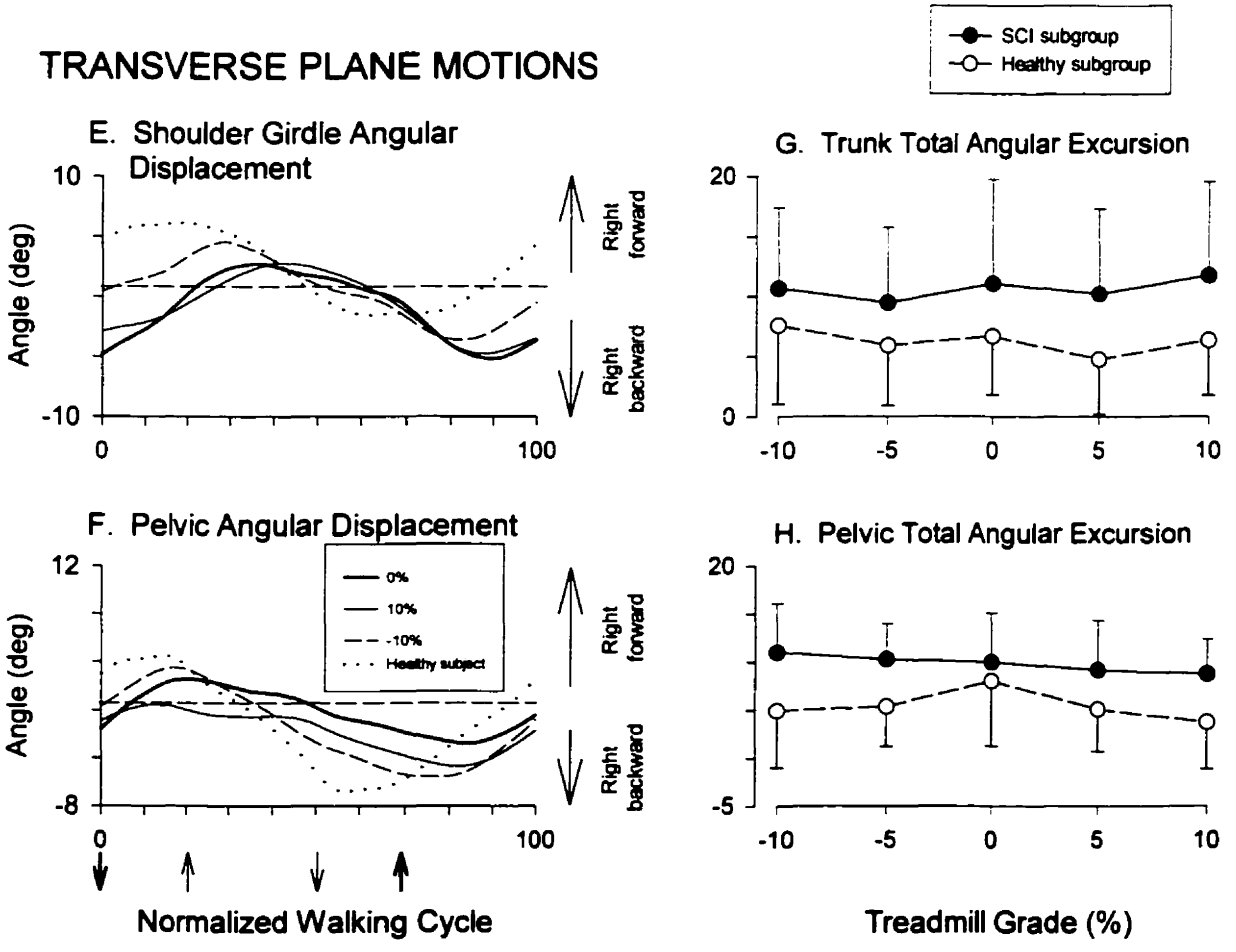


FIGURE 5

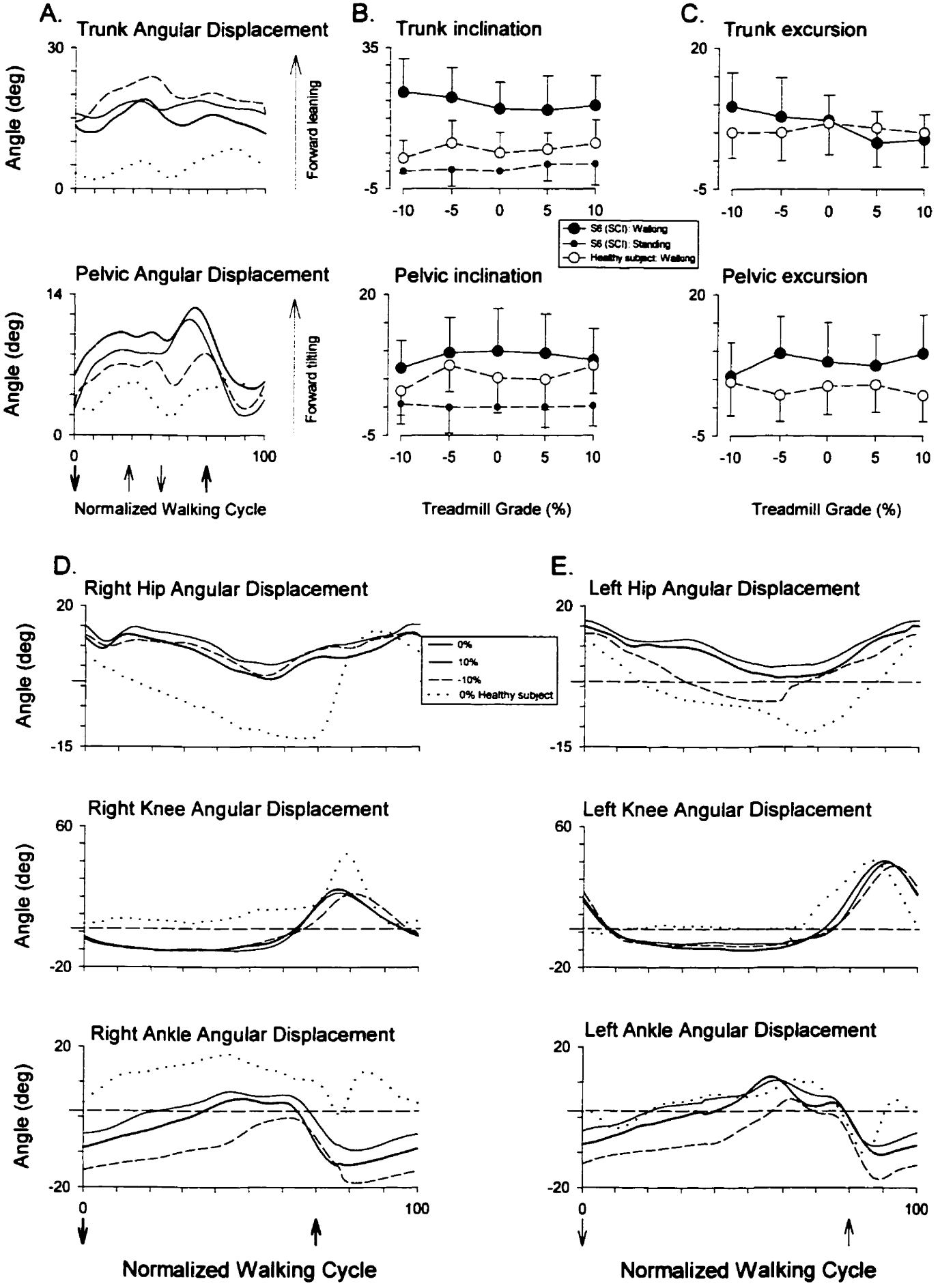
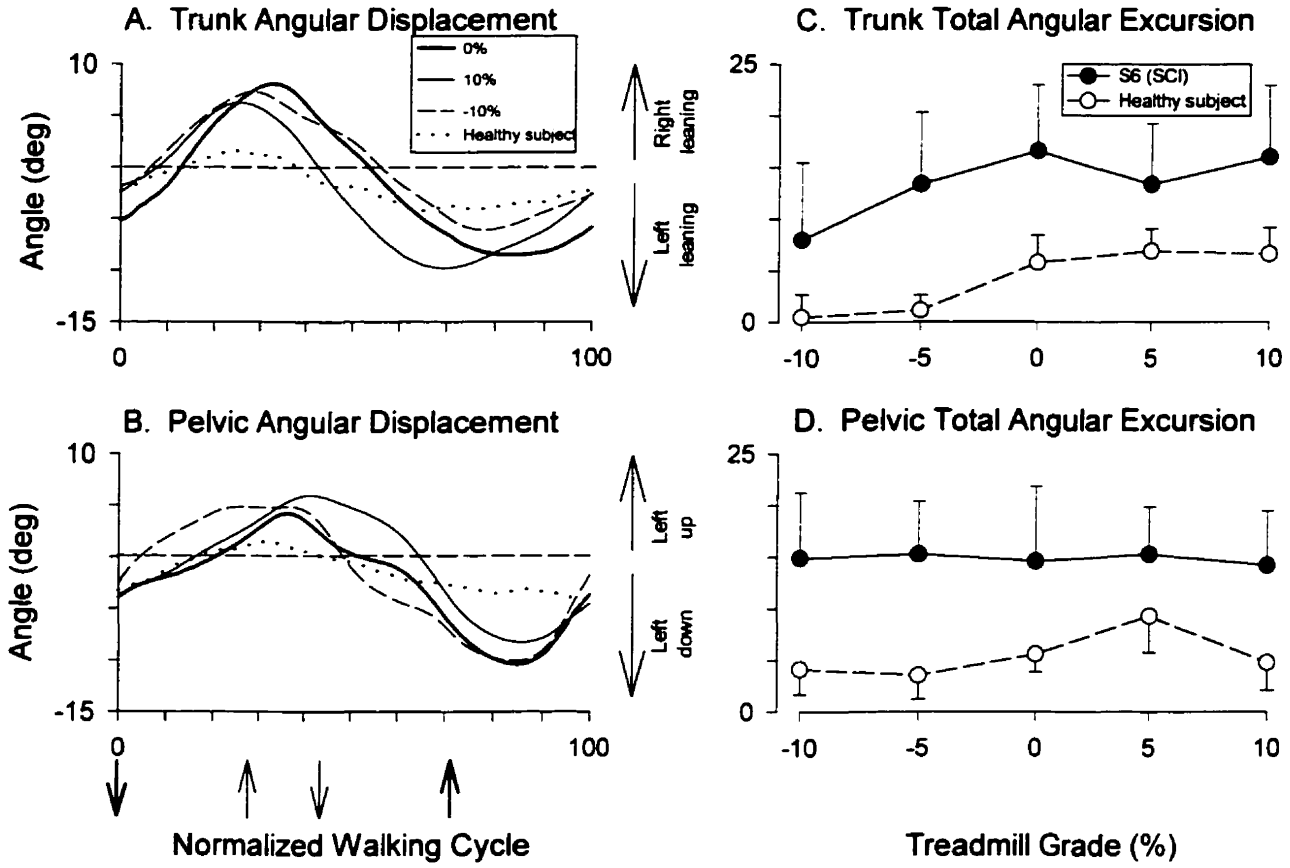

A.
Trunk Angular Displacement
Angle (deg)
30
0
Forward leaning
B.
Trunk inclination
35
-5
-10
-5
0
5
10
C.
Trunk excursion
20
-5
-10
-5
0
5
10
S6 (SCI): Walking
S6 (SCI): Standing
Healthy subject: Walking
Pelvic Angular Displacement
Angle (deg)
14
0
Forward tilting
0
100
Normalized Walking Cycle
Pelvic inclination
20
-5
-10
-5
0
5
10
Treadmill Grade (%)
Pelvic excursion
20
-5
-10
-5
0
5
10
Treadmill Grade (%)
D.
Right Hip Angular Displacement
Angle (deg)
20
-15
0%
10%
-10%
0% Healthy subject
E.
Left Hip Angular Displacement
20
-15
Right Knee Angular Displacement
Angle (deg)
60
-20
Left Knee Angular Displacement
60
-20
Right Ankle Angular Displacement
Angle (deg)
20
-20
0
100
Normalized Walking Cycle
Left Ankle Angular Displacement
20
-20
0
100
Normalized Walking Cycle

FIGURE 6

FRONTAL PLANE MOTIONS



TRANSVERSE PLANE MOTIONS

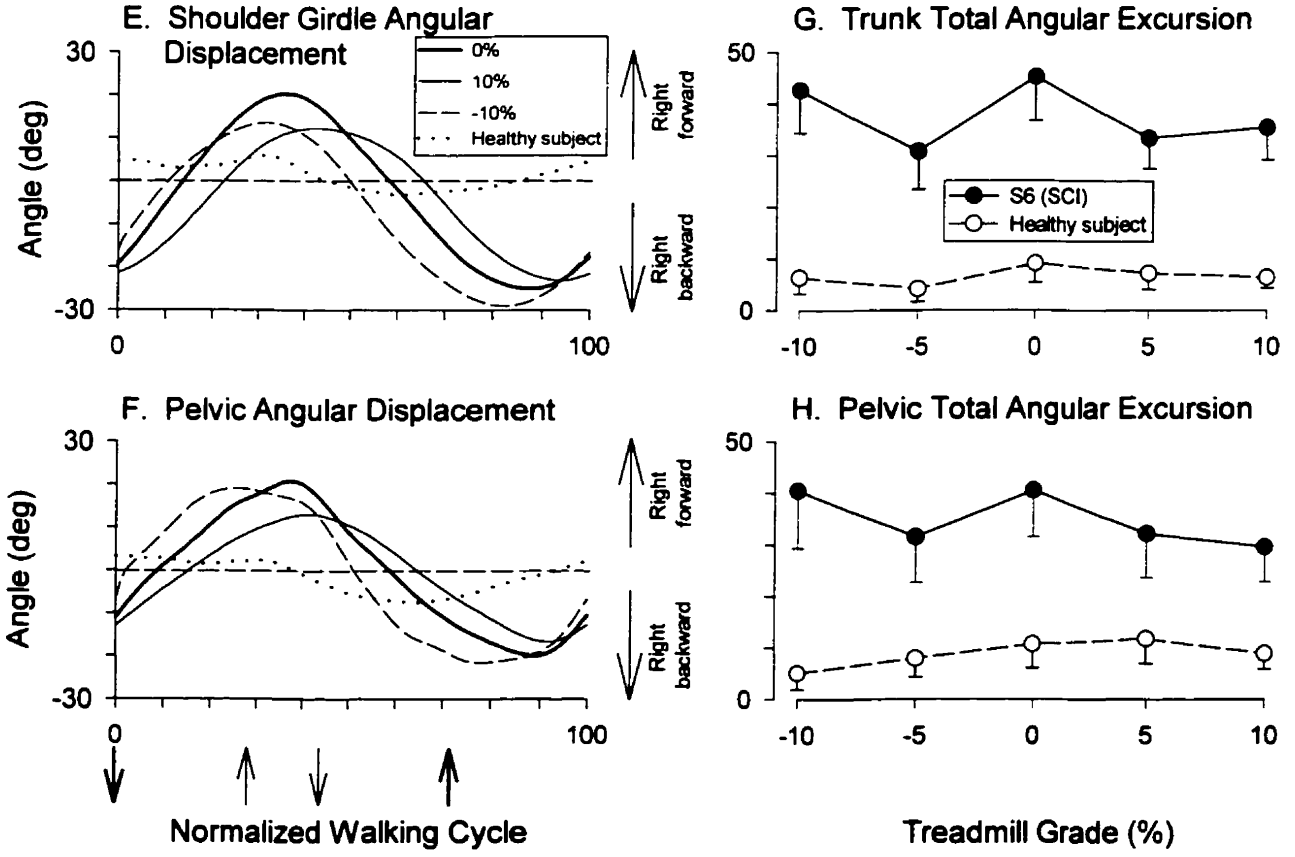


FIGURE 7

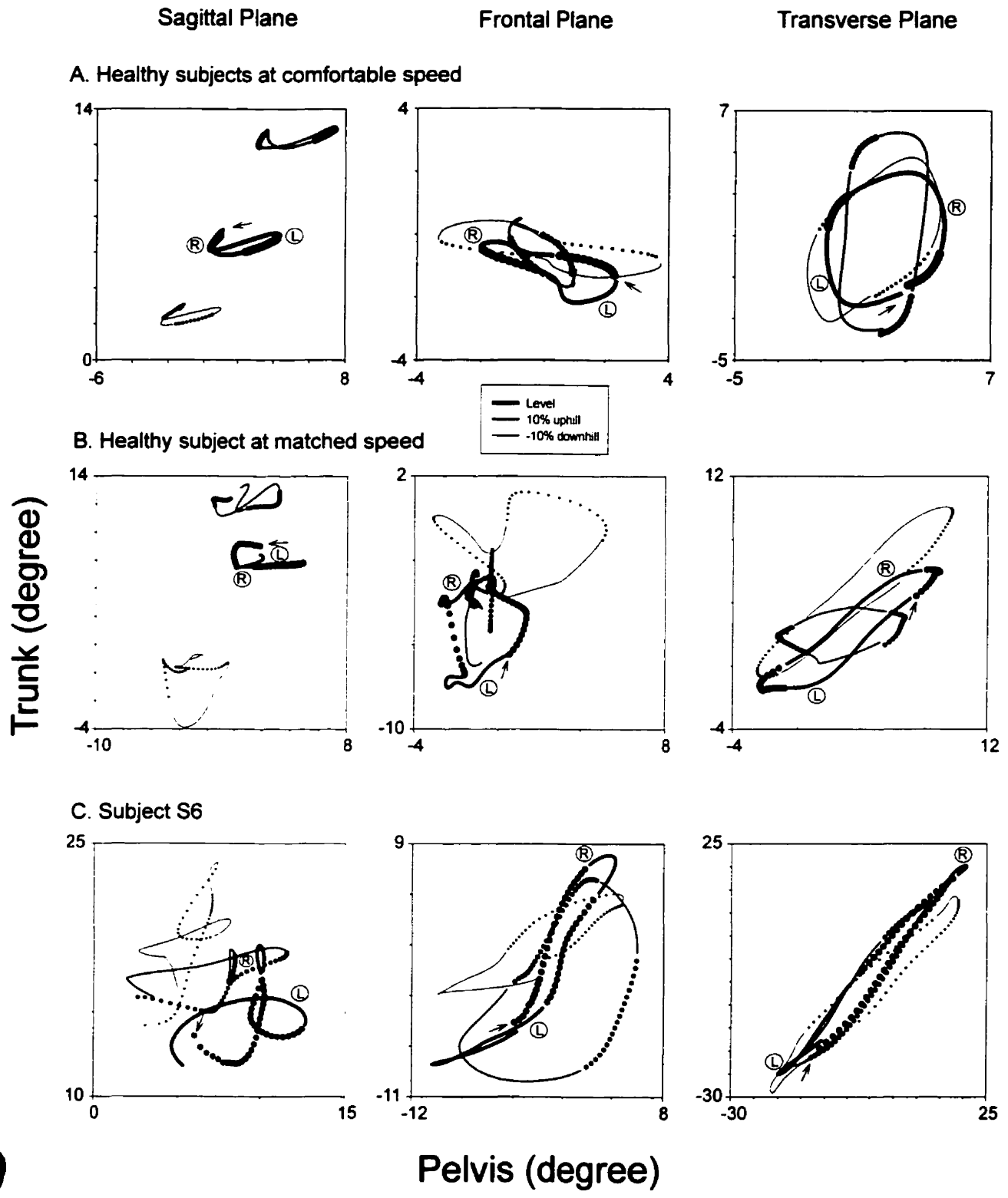
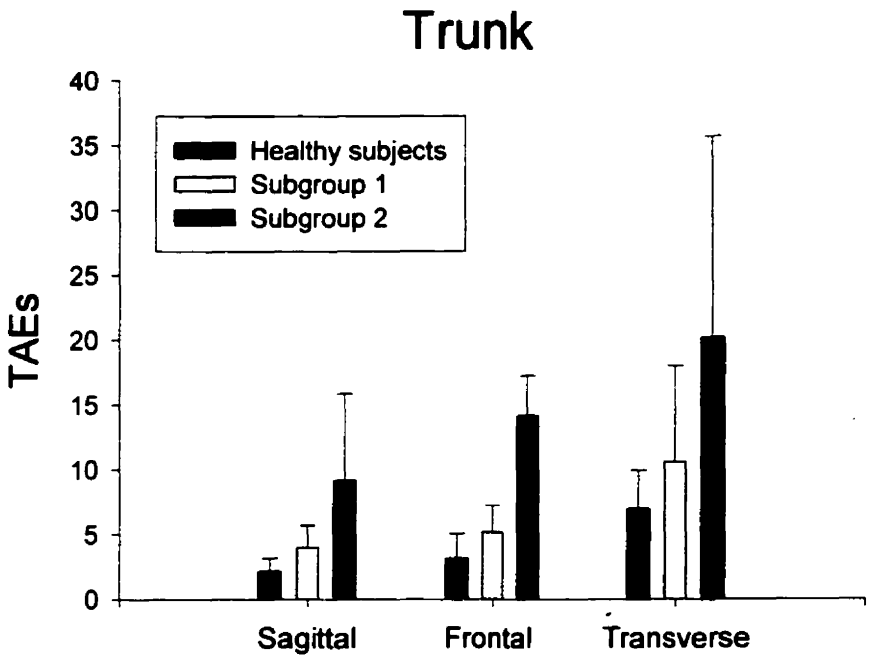
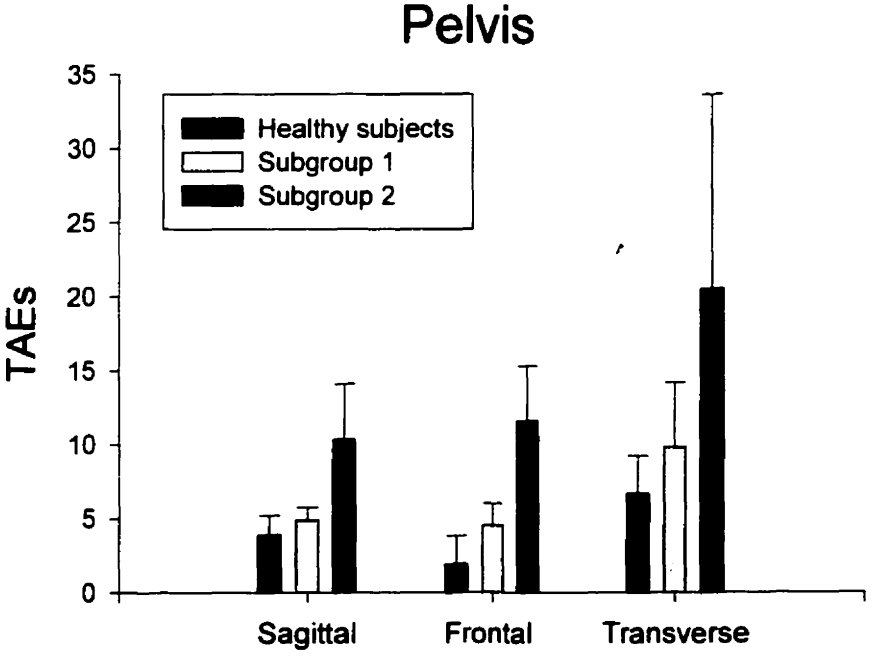

Sagittal Plane
Frontal Plane
Transverse Plane
A. Healthy subjects at comfortable speed
B. Healthy subject at matched speed
C. Subject S6
Level
10% uphill
-10% downhill
Trunk (degree)
Pelvis (degree)
14
0
-6
8
4
-4
-4
4
7
-5
-5
7
14
-4
-10
8
2
-10
-4
8
12
-4
-4
12
25
10
0
15
9
-11
-12
8
25
-30
-30
25
R
L

Figure 8


Trunk
TAEs
40
35
30
25
20
15
10
5
0
Healthy subjects
Subgroup 1
Subgroup 2
Sagittal
Frontal
Transverse


Pelvis
TAEs
35
30
25
20
15
10
5
0
Healthy subjects
Subgroup 1
Subgroup 2
Sagittal
Frontal
Transverse

FUNCTIONAL IMPLICATIONS RELATED TO WALKING ON INCLINED SURFACES AFTER SPINAL CORD INJURY

Every SCI subject who participated in the present research project reported greater difficulty to perform downhill walking as compared to uphill walking. Furthermore, the losses of balance during walking occurred almost exclusively in downhill grades. To have a better understanding of the reasons underlying such a difference in the level of difficulty between uphill and downhill walking grades, it is important to analyse the task requirements for each condition.

Uphill walking: task requirements and functional adaptations

During uphill walking, the body is moved forward and upward. To counteract the increased resistance due to gravity, the forward propulsion of the body has to be increased. As discussed previously in Paper 1, this increase in propulsion seems to be mainly achieved by plantarflexor muscles as reflected by an important increase in the amplitude of this muscle group during push-off. Furthermore, this increase in body's forward propulsion, which is reflected by an increase in stride length, is assisted by a forward inclination of the trunk and pelvis. In SCI subjects, the strategies of adaptation directly depends on the severity of deficits in lower limb kinematic patterns. SCI subjects showing minor deficits (see subgroup 1 from Paper 3) used a similar strategy as in healthy subjects which include modifications in all three lower limb joints as well as a forward posture of the trunk and pelvis. Furthermore, these subjects show an increase in the level of plantarflexor activity to produce a dynamic push-off in uphill condition (not illustrated in the Figures). On the other hand, in SCI subjects showing major deficits in lower limb kinematic patterns, this increase is usually not achieved and the same level of plantarflexor activity is maintained constant across walking grades.

Furthermore, even though these subjects show a forward inclination of the trunk and pelvis as in healthy subjects, there is no consistent increases in stride length that could reflect an increase in body's forward propulsion. Thus, SCI subjects with major deficits to lower limbs have to use different strategies than healthy subjects to adapt to uphill walking. These strategies involve mainly lower limb joints since no compensatory movements from the trunk and pelvis have been found when adapting to uphill walking (see Paper 3). The absence of increase in the level of activation of plantarflexors likely results in a weak push-off and compensatory mechanisms from proximal segments might be used to adapt to uphill walking. Results obtained from the present studies strongly favor the hip as the key joint when adapting to uphill walking in these subjects. This joint was the only one that always showed an increase in flexion from mid-swing to the stance phase during uphill slopes in every SCI subject with major deficits to lower limbs. Furthermore, an increase in the level of activation of hip flexors at the initiation of the swing phase during uphill walking was detected in some of these subjects (not illustrated in the figures). The hip would then be recruited to pull the swinging limb upward and forward during uphill walking. Instead of using a push-off strategy as in healthy subjects, these SCI subjects would mainly used a pull-off strategy from the hip joint. Similar compensatory mechanisms from hip muscles have also been identified in other gait-related pathologies (Olney et al. 1988; Winter and Sienko 1988).

Downhill walking: task requirements and functional adaptations

During downhill walking, the force of gravity tends to accelerate the body forward and downward. To counteract these external forces, the action of the knee joint is especially important at two periods of the gait cycle. Following the heel contact, the knee has to perform a greater flexion to increase

energy absorption during weight acceptance and brake the forward momentum of the body (Redfern and DiPasquale, 1997). An increase in knee flexion is also performed in late stance at the time where the controlateral limb is in preparation for the foot contact period. The purpose of this increased flexion of the knee is to to decrease the impact force of the controlateral limb at heel contact by lowering the vertical displacement of the center of gravity (Wall et al., 1981). This decrease in body momentum is also reflected by a decrease in stride length in downhill conditions. In SCI subjects showing major deficits in lower limb kinematic patterns (see subgroup 2 from Paper 3) the problems encounter during downhill walking can be more likely related to the difficulty of controlling the knee joint. These subjects showed either knee hyperextension or hyperflexion throughout the stance phase in downhill walking. The presence of knee hyperextension in late stance does not allow the center of gravity to drop which likely results in an increase in the forward momentum of the body at controlateral heel contact. However, this increased momentum cannot be reduced by increasing knee flexion during weight acceptance since this joint remains in an extended position throughout stance. Thus, this lack of knee flexion after heel contact do not provide the normal damping effect on the gait pattern. Furthermore, this lack of energy absorption from lower limbs can result in greater lateral displacements of the body after each heel contact and create instability in downhill condition. On the other hand, the presence of knee hyperflexion following heel contact reveals great difficulty to absorb energy and break the forward momentum of the body. This difficulty to support body weight can also lead to instability and even falls during downwhill grades.

The forward inclination of the trunk observed in SCI subjects across walking grades is likely associated with the greater difficulty to maintain balance from these subjects in downhill walking.

During downhill walking, the body is moved forward and downward by the mechanical effect of the slope. The forward posture adopted by SCI subjects contribute to increase this effect by moving the center of gravity even more anteriorly with respect to the area of support. This forward posture would more likely contribute to decrease the stability of the body and increase its forward momentum during downhill walking.

In summary, muscle weaknesses usually seen after SCI do not seem to be a limiting factor to perform up-slope walking (up to 15 degrees). Even if the main muscles responsible for such adaptation are usually impaired, SCI subjects use other muscle groups (mainly the hip muscles) to succesfully perform the task. The postural component of this task does not seem to be a limiting factor since they adopt a forward posture of the trunk and pelvis in every walking grade. On the other hand, two main factors would explain the greater difficulty from SCI subjects to perform downhill walking. The first factor is related to the difficulty from the knee joint to brake the increased downward momentum of the body produced by downhill slopes whereas the second factor is related to the inappropriate forward posture of the trunk that could decrease body's stability during downhill walking.

New perspectives on the role of posture in locomotion

The results presented in this dissertation bring new insights about the interaction between posture locomotion. Traditionally, these two fields of research have been studied separately and very few studies have examined the impact of postural modifications in locomotion (Grasso et al., 2000; Trank et al., 1996). The task of walking up-slope and down-slope is directly associated with postural modifications from the trunk and pelvis. Furthermore, as discussed in Paper 2, these changes might

facilitate power generation or absorption in adapting to slope changes during walking. According to these results, the purpose of such postural adaptations would be to facilitate lower limbs generating the proper momentum during inclined walking. Following SCI, however, the role of posture is changed and the trunk and pelvic alignment does not seem to assist lower limbs as in healthy subjects, especially during downhill walking. The trunk and pelvic forward posture observed in SCI subjects during both level and inclined walking would in fact be used to provide greater stability by moving the center of gravity forward and downward in function of the area of support.

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SUMMARY AND CONCLUSION

The purpose of this research thesis was to determine the strategies of locomotor and postural adaptations to inclined walking in healthy and SCI subjects. The results from healthy subjects reveal that walking on inclined surfaces requires adaptations from lower limb, pelvic and trunk movements in the sagittal plane. The key strategy of adaptation to uphill walking is to adopt a flexed posture of the hip, knee and ankle at initial foot contact as well as a forward inclination of the trunk and pelvis. An important increase in the amplitude of plantarflexor muscles is further requires to propel the body forward and upward and couteract the resistance due to gravity. In contrast, downhill walking requires a decreasingly flexed posture of the hip at initial foot contact as well as an increase in knee flexion during weight acceptance and late stance. These changes were accompanied by a progressive backward inclination of trunk and pelvis. Modifications in trunk and pelvic postural alignment in the sagittal plane seems to be used to facilitate power generation or absorption in adapting to slope changes during walking in healthy subjects. Following SCI, trunk and pelvic postural alignment do not seem to be used to assist lower limbs generating or absorbing more energy during inclined walking. The anterior posture of the trunk and pelvis, observed in every walking grade, would instead be adopted to provide greater stability during walking. Abnormal movements from the trunk and pelvis during walking were observed in subjects with SCI showing major deficits in lower limb kinematic patterns. These trunk and pelvic movements were either used to compensate for specific deficits in lower limbs or in reaction to such deficits. These abnormal movements were, however, maintained constant across walking grades suggesting that movements from the trunk and pelvis were not directly involved when adapting to inclined surfaces in subjects with SCI. Thus, lower limbs

would mainly be responsible for increasing or decreasing body's forward momentum during uphill and downhill walking respectively in subjects with SCI.

APPENDIX I

REPRINT OF RESEARCH PAPER 1 PUBLISHED IN:

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Alain Leroux · Joyce Fung · Hugues Barbeau

Adaptation of the walking pattern to uphill walking in normal and spinal-cord injured subjects

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Abstract Lower-limb movements and muscle-activity patterns were assessed from seven normal and seven ambulatory subjects with incomplete spinal-cord injury (SCI) during level and uphill treadmill walking (5, 10 and 15°). Increasing the treadmill grade from 0° to 15° induced an increasingly flexed posture of the hip, knee and ankle during initial contact in all normal subjects, resulting in a larger excursion throughout stance. This adaptation process actually began in mid-swing with a graded increase in hip flexion and ankle dorsiflexion as well as a gradual decrease in knee extension. In SCI subjects, a similar trend was found at the hip joint for both swing and stance phases, whereas the knee angle showed very limited changes and the ankle angle showed large variations with grade throughout the walking cycle. A distinct coordination pattern between the hip and knee was observed in normal subjects, but not in SCI subjects during level walking. The same coordination pattern was preserved in all normal subjects and in five of seven SCI subjects during uphill walking. The duration of electromyographic (EMG) activity of thigh muscles was progressively increased during uphill walking, whereas no significant changes occurred in leg muscles. In SCI subjects, EMG durations of both thigh and leg muscles, which were already active throughout stance during level walking, were not significantly affected by uphill walking. The peak amplitude of EMG activity of the vastus lateralis, medial hamstrings, soleus, medial gastrocnemius and tibialis anterior was progressively increased during uphill walking in normal subjects. In SCI subjects, the peak amplitude of EMG activity of the medial hamstrings was adapted in a similar fashion, whereas the vastus lateralis, soleus and medial gastrocnemius showed very limited adaptation during uphill walking. We conclude that SCI subjects can adapt to uphill treadmill walking within certain limits, but they use different strategies to adapt to the changing locomotor demands.

Key words Human walking · Uphill · Spinal cord injury · Adaptation · EMG · Kinematics

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Introduction

The human locomotor pattern can be easily and rapidly adapted to modifications in external demands, such as speed (Brandell 1977; Murray et al. 1966; Nilsson et al. 1985) and slope (Brandell 1977; Lange et al. 1996; Simonsen et al. 1995; Wall et al. 1981). This adaptation is normally achieved by changing the pattern of lower-limb motions (Lange et al. 1996; Murray et al. 1966; Nilsson et al. 1985; Wall et al. 1981) and by varying the motor recruitment of the relevant flexor and extensor muscles (Brandell 1977; Lange et al. 1996; Nilsson et al. 1985; Simonsen et al. 1995). Following a spinal cord injury, descending pathways are usually damaged, affecting the adaptability of the locomotor pattern. Animal studies have shown that cats with complete spinal cord transection can recover locomotion on the treadmill and can adapt to changes in speed (Barbeau and Rossignol 1987; Bélanger et al. 1989; Forssberg et al. 1980). However, the spinal cat has difficulty in adapting its walking pattern to changes in inclined planes (Bélanger et al. 1988). For example, contrary to the results obtained in the intact cat, only minor changes in the electromyographic (EMG) activity are observed during uphill (15°) and downhill (20°) walking in the spinal animal (Bélanger et al. 1988). Moreover, without manually supporting the spinal cat, the animal could only adapt to a few degrees of roll, as compared with at least 20° observed in the intact cat (Bélanger et al. 1988).

In humans, uphill walking is a demanding task that requires specific modifications in lower-limb movements (Lange et al. 1996; Wall et al. 1981) and muscle activation (Brandell 1977; Lange et al. 1996; Simonsen et al. 1995). As the walking grade increases, more propulsion has to be generated from the lower limbs (Brandell 1977), and postural adjustments must be performed to maintain equilibrium (Kawamura et al. 1991). Following spinal cord injury (SCI), the basic locomotor pattern is altered (Barbeau et al. 1998), and the ability to adapt to uphill walking could be affected. However, it remains unknown if persons with SCI can adapt to uphill walking. This study is a first effort to assess the adaptability of the locomotor pattern to uphill walking following SCI.

Materials and methods

Subjects

Seven normal and seven SCI subjects (see Table 1) took part in the experiment. The SCI group included six males and one female who were free of any comorbid medical complications. The normal subjects were males and aged between 28 and 40 years. The study was approved by the ethical committee of the School of Physical and Occupational Therapy of McGill University and informed consent was obtained from all subjects prior to the experiment.

Walking assessment

The subjects walked on a custom designed treadmill (see Norman et al. 1995) at four different grades (0, 5, 10 and 15°). To standardize the walking conditions, each subject had to grasp the horizontal handrails during walking grades. During level gait, this procedure does not alter the sagittal-plane kinematics of normal walking (Siler et al. 1997). The walking grades were presented in random order. SCI subjects performed all walking conditions at their self-selected comfortable walking speed, while normal subjects walked at 0.4 and 1.0 m/s. Subjects habituated themselves to the different grades of treadmill walking before the experimental sessions began. Data were collected from the right side in normal subjects and from the more affected side in SCI subjects. No major asymmetry was found in any of the SCI subjects. The more affected limb was chosen based on muscle-strength testing.

Pressure sensitive footswitches were placed bilaterally on the soles of the subjects' shoes to signal the heel, fifth metatarsal and big toe contacts. The footswitch signals were used to measure stance, swing and cycle durations. Using preamplified surface electrodes, EMG activities were recorded from the vastus lateralis (VL), medial hamstrings (MH), soleus (SOL), medial gastrocnemius (MG) and tibialis anterior (TA). The EMG signals were amplified, bandpass filtered at 10–1000 Hz and recorded on FM tape (Honeywell). The movements of the lower limb were recorded with two videocameras (Panasonic VHS VTRs), operating at a frequency of 60 Hz and located about 90° from each other. Reflective markers were attached over the fifth metatarsal, the heel, the lateral malleolus, the tibial plateau, the greater trochanter of the femur and the greater tuberosity of the humerus. The 3-D coordinates of each marker were obtained from the two camera views. To synchronize EMG signal and joint angular displacement records, a time code signal was recorded on both FM and videotapes.

Data analysis

Five consecutive walking cycles were selected for the analysis of the kinematic and EMG signals. The Peak Performance system was used to digitize the X and Y coordinates of the reflective markers, which were then used to calculate the angular excursions of the hip, knee and ankle joints in the sagittal plane. Kinematic analysis of the hip, knee and ankle joints was performed in the sagittal plane, using conventional definitions (Winter 1991). Briefly, the hip and knee angles were calculated with respect to the vertical line, with the neutral position in standing being taken as 0° displacement, flexion being positive, and extension negative. Likewise, in calculating the ankle angle, the neutral standing position with the leg axis perpendicular to the foot was taken as 0°. Ankle dorsiflexion beyond neutral was taken as positive angular displacement, and ankle plantar flexion beyond neutral was taken as negative angular displacement. The footswitch and EMG signals were digitized (1.2 kHz sampling) using a custom designed program. The EMG signals were later digitally bandpassed (20–400 Hz), full-wave rectified and smoothed using a moving window average of 38.4 ms (Data Pac II program from Run Technologies). Twenty-three points from each side of the center were included in the moving window average.

For each subject, five walking cycles were averaged to obtain individual means of angular excursions and EMG patterns across the different walking grades. The amplitude of the averaged EMG profile was normalized with respect to the peak activity of each muscle occurring during level (0°) walking. An interactive computer program was used to determine the onset and offset of EMG activity. A two-way ANOVA with repeated measures was used to test for differences in EMG peak amplitude, EMG burst duration and kinematic parameters between groups and across grades. Due to the fact that only three of the seven SCI subjects performed the 15° grade, this grade was not included in the statistical tests. When significant differences were found, pairwise comparisons were made using Student-Newman-Keuls tests. Statistical significance was set at $p=0.05$.

Table 1 Profile of the spinal-cord injured subjects. (*MVA*= Motor vehicle accident)

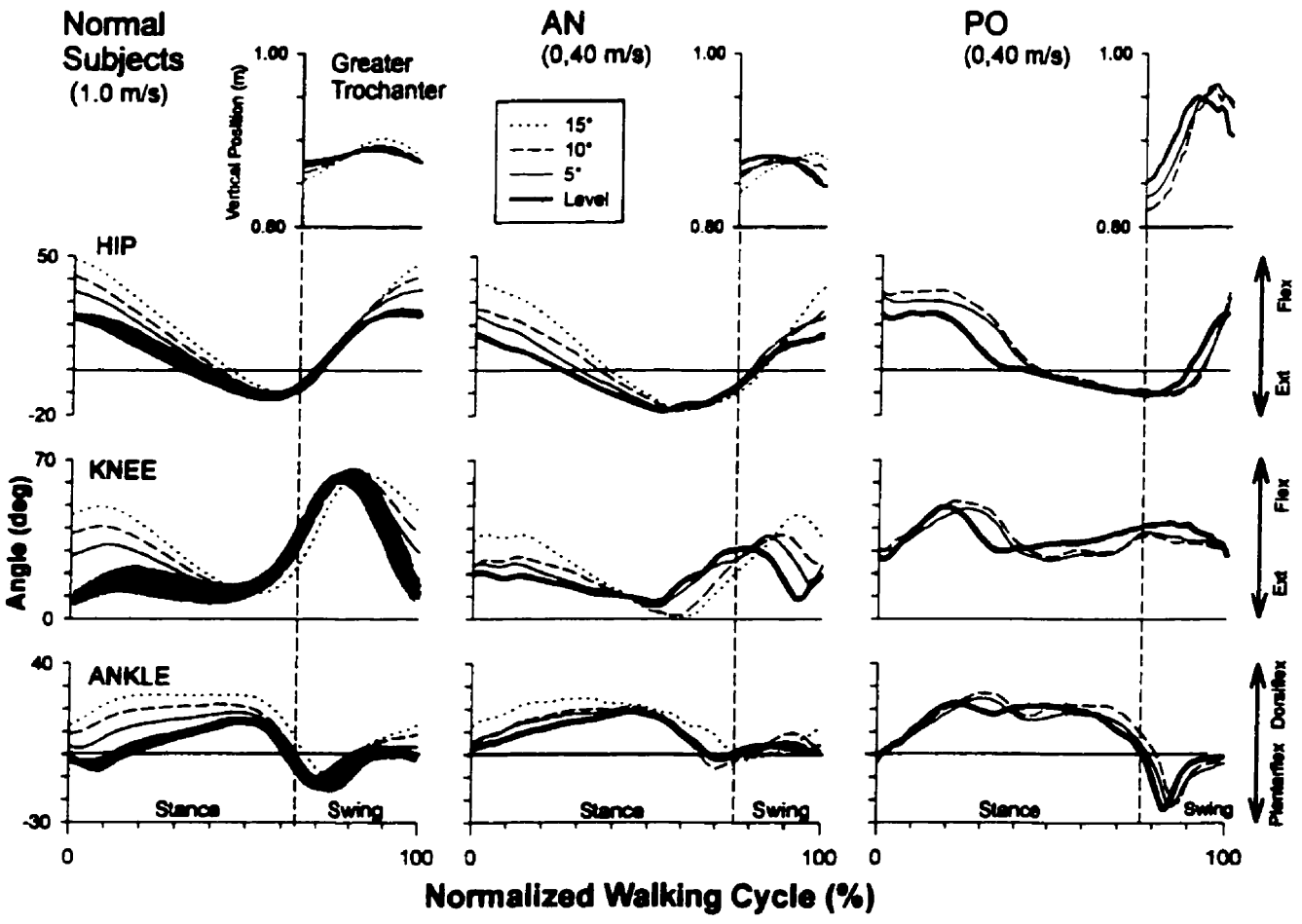
Subjects	Age	Cause and level of injury	Time since injury	Comfortable treadmill speed (m/s)	Maximal treadmill speed (m/s)	Treadmill grades performed (°)
AN	24	Spastic paresis, non-familial origin	>14 years	0.40	0.80	0–15
CP	46	MVA, T11	14 months	0.70	1.10	0–10
FS	25	MVA, T8-T9	>6 years	0.30	0.60	0–10
LT	20	Diving, C5	2.8 years	0.60	1.00	0–15
PO	16	MVA, C7-T1	10 months	0.40	0.70	0–10
PR	20	MVA, C6	>5 years	0.80	1.30	0–15
RM	38	Diving, C6	>20 years	0.30	0.60	0–10

Results

Angular excursions of lower limbs

Figure 1 shows the angular excursions of the hip, knee and ankle in normal subjects and two SCI subjects. For the normal subjects, each trace represents an average from all seven subjects. The gray area represents the 95% confidence interval for the level condition. It can be seen that the variations in lower-limb angular displacements are small between normal subjects and within the same SCI subject. The normal subjects showed marked modulation in joint angular displacements during uphill walking. At the hip joint, the range of flexion increased during swing to lift the leg up the slope, resulting in a larger excursion during stance. At the knee joint, the range of extension decreased during late swing, such that foot contact was made with increased knee flexion, and the increasingly flexed posture was followed by a progressively larger knee excursion during stance as the slope increased. For the ankle joint, the range of dorsiflexion increased during swing to lift the ankle up the slope and remained elevated during stance. However, angular displacements of the hip and knee were invariant around stance-swing transition across all grades in every normal subject. This indicates that the hip and knee position at this period of the walking cycle was kept constant across grades. This point of invariance was also observed for the hip-knee coordination pattern in normal subjects (see phase 3 on Fig. 3). The height of the greater trochanter during swing, as shown on top of Fig. 1, showed only minor changes as the excursion varied from 1.64 cm at 0° grade to 5.36 cm at 15° grade. In SCI subjects, however, lower-limb motions did not show consistent patterns of adaptation as in normal subjects. Rather, a spectrum of adaptation was found and ranged from a near normal pattern of adaptation to one in which only the hip angle was modified. The SCI subjects who showed near normal walking patterns during level walking had a near normal adaptation to uphill walking. Two SCI subjects, who represent the opposite of the spectrum, are shown in Fig. 1. Subject AN performed all uphill conditions, whereas PO was unable to achieve the most elevated treadmill grade. Furthermore, as observed

Fig. 1 Angular excursions of the hip, knee and ankle during uphill walking, normalized to the gait cycle (from one foot contact, *0%*, to the next, *100%*). The trajectory of the greater trochanter is shown at the top. This figure shows the average of the seven normal subjects and two SCI subjects (each *line* represents the average of five cycles). The two SCI subjects illustrate the two extremes in the spectrum of adaptability to graded walking. The *gray area* represents the 95% confidence interval of the level condition for the normal group and each of the two SCI subjects



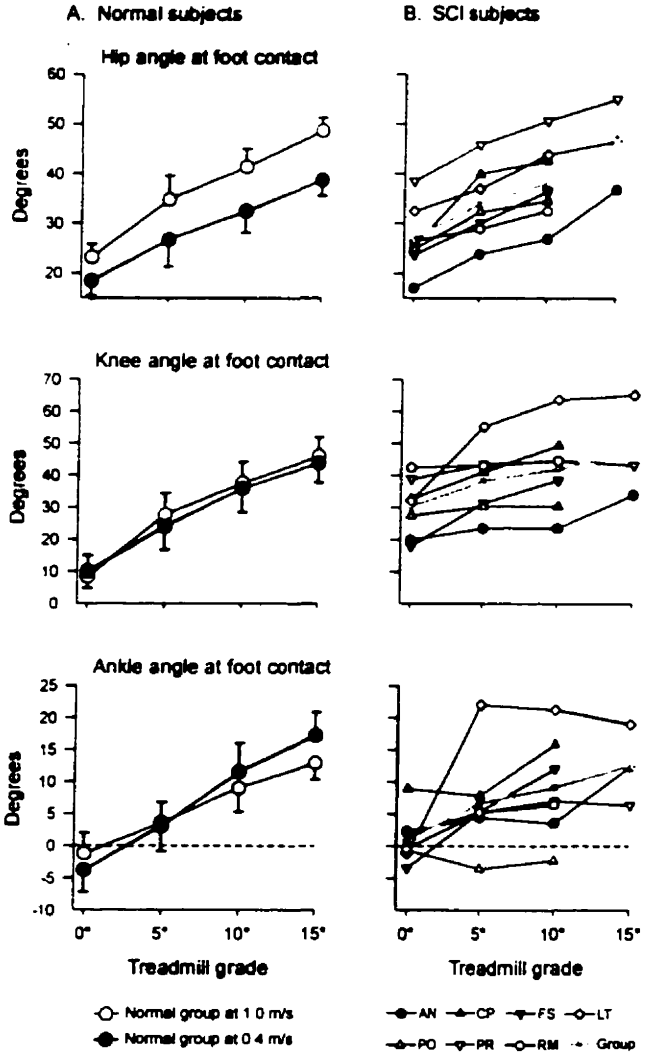


Fig. 2A, B Average hip, knee and ankle angles at foot contact during level and uphill walking for the group of normal subjects (n=7) (**A**) and the individual SCI subjects (**B**). *Gray line* represents the grand average of the seven SCI subjects. (*Initials* Individual subjects)

in normal subjects, the adaptation occurred at the hip, knee and ankle joints in AN during uphill walking. In contrast, PO showed little adaptation in all three lower-limb joints except for a small increase in hip flexion during late swing and early stance. Instead, a different compensation mechanism involving hip hiking was used in all walking conditions, as shown by the vertical trajectory of the greater trochanter (Fig. 1, top inset).

To determine lower-limb movement adaptation to grade, joint angles at foot contact were averaged for normal subjects walking at slow (0.4 m/s) and comfortable (1.0 m/s) speeds as well as SCI subjects walking at self-selected speed (Fig. 2). Significant main effects due to grade were found at the hip, knee and ankle joints (p<0.01). Post-hoc comparisons revealed that hip, knee and ankle flexion at foot contact became significantly larger in normal subjects walking at either speeds as grade increased, but not in SCI subjects. A significant main effect due to group (normal vs. SCI) was found only at the knee joint. Knee flexion at level walking was significantly larger (p<0.01) for the SCI group.

Results from each individual SCI subject revealed that a clear adaptation pattern occurred only at the hip (Fig. 2B). As found in normal subjects, every SCI subject showed a gradual increase in hip flexion as the treadmill grade increased. Contrary to the normal subjects, every SCI subject made foot contact with the knee in flexion during level walking (Fig. 2B). Furthermore, except for one subject (LT), the changes in flexion at the knee were limited or absent when adapting to uphill walking. Ankle-joint adaptation was most variable in SCI subjects. Instead of showing a gradual increase in ankle dorsiflexion as in normal subjects, some SCI subjects (AN, CP, LT) showed both an increase and decrease in ankle dorsiflexion at increasing treadmill grades.

Temporal distance data

Figure 1 shows that the percentage of the step cycle devoted to stance for the two SCI subjects was greater than that for the normal subjects. This increase in the proportion of stance duration is naturally observed at low walking speeds even in normal subjects (Nilsson et al. 1985). Walking at different grades did not affect temporal distance variables in normal subjects and in SCI subjects. No significant changes were found for the stride length, cycle duration and stance-swing ratio across the different walking grades.

Intralimb coordination

Figure 3 illustrates hip-knee cyclographs of normal and SCI subjects across different grades. In normal subjects, only one hip-knee coordination pattern was found. Thus, all data were averaged and pooled together. During level walking, the cyclograph curve was characterized by a slanted "S" shape in stance and by a parabolic shape during swing in normal subjects. When the treadmill grade increased, the starting point (phase 1), representing hip-knee position at foot contact, was progressively shifted toward the right upper region, indicating an increase in hip flexion and a decrease in knee extension. Following foot contact, there was a knee flexion (yield) associated with weight acceptance (phase 2). This knee yield was preserved, but became less prominent during uphill walking. From early to mid stance (phase 2 to 3), the hip and knee extended simultaneously. In late stance (push off, phase 3), the knee flexed rapidly while the hip reversed from extension to flexion. This point of transition was maintained constant across the different grades. The peak of knee flexion, reached in mid swing (phase 4), was similar across different grades, although a progressive increase in hip flexion was noted.

Contrary to the normal subjects, many different joint coordination patterns were present even for level walk-

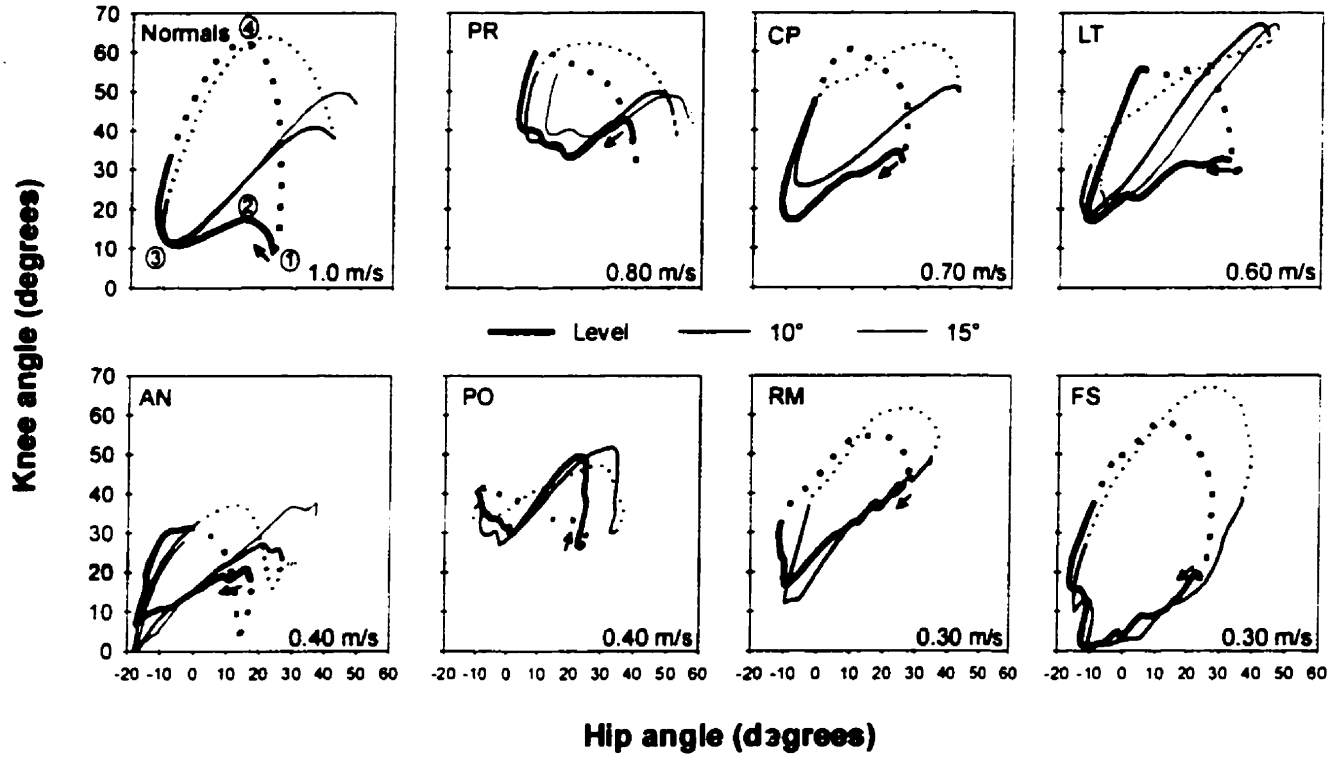


Fig. 3 Cyclographs showing the coupling between the hip and knee joints during uphill walking. This figure shows the average of the seven normal subjects (*normals*) and the seven individual SCI subjects (*initials*) rank-ordered by their comfortable walking speed. The *solid lines* represent the stance phase, and the *dotted lines* represent the swing phase. The *arrows* indicate the direction of the movement. On the graph showing the normal subjects, *1* indicates foot contact, *2* corresponds to early stance, *3* represents the transition from mid-swing to push-off and *4* indicates mid-swing. Displacement of the curves diagonally up and right indicates a simultaneous increase of knee and hip flexion

ing in SCI subjects (Fig. 3). For example, in AN the knee was relatively flexed at the onset of foot contact during level walking (as it was in all SCI subjects), and exhibited a very limited yield. The peak of knee flexion, which occurred during mid-swing in normal subjects (see phase 4), was reached at the end of stance (similar in PR and LT). Furthermore, while the hip flexed near the end of the swing phase, knee extension was abnormally reversed into flexion. In general, some normal adaptation features were preserved, which included a progressive shift in the starting hip-knee position and in the peak of knee flexion. Abnormal features included the presence of knee hyperextension, thus altering the transition between mid-stance and push off, as well as an increase in peak knee flexion during mid swing.

In PO, substantial differences were found in the coupling between the hip and knee movements during level walking. The knee yield (phase 2) was much larger than normal, revealing problems with weight acceptance. Between mid-stance and push off (phase 3), the knee flexed while the hip continued to extend, giving rise to a prolonged, jerky transition. The parabolic curve usually observed in normal subjects in phase 4 was markedly flattened. Very limited changes were observed as grade increased, except for the emergence of a more normal hip-knee coupling in swing phase. In four of the other SCI subjects (PR, CP, RM, FS), the hip-knee coordination patterns were changed in a fashion similar to that observed in normal subjects, at least during the swing phase.

Muscular activity

Uphill walking induced marked modulation in the amplitude of EMG activity in normal subjects (Figs. 4 and 5). As the treadmill grade increased from 0 to 15°, there was a progressive increase in the amplitude of all recorded muscles. In SCI subjects, however, the amplitude of EMG activity was minimally affected by uphill walking in most muscles, as shown in subjects AN and PO (Fig. 4; the same subjects selected as examples in Fig. 1). In AN, the EMG amplitude of thigh muscles (VL and MH) also increased with grade, although to a lesser extent than in normal subjects, whereas the leg muscles (SOL, MG and TA) showed very limited changes in amplitude during uphill walking. In PO, the EMG activity of both thigh and leg muscles was minimally affected by uphill walking. Furthermore, a complete absence of activity was seen in TA. In general, for the plantarflexor muscles (SOL and MG), a prolongation of the EMG activity and an absence of peak of activity during the push-off period were found in all SCI subjects.

To determine the effect of uphill walking on lower-limb muscle activation, the peak amplitude of thigh mus-

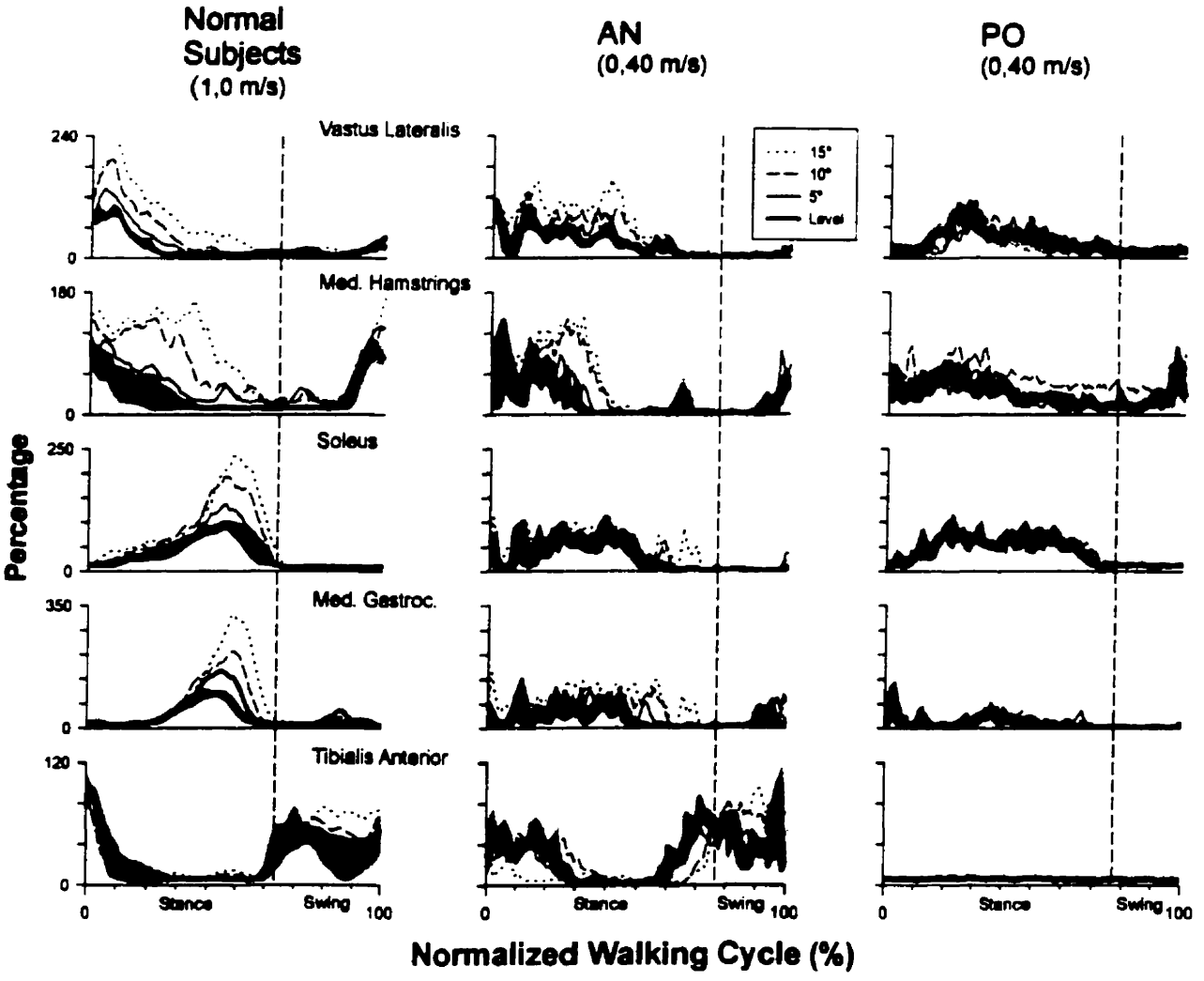


Fig. 4 Electromyographic (EMG) linear envelope of lower-limb muscles during uphill walking, normalized to the gait cycle (from one foot contact, *0%*, to the next, *100%*). Amplitude of EMG activity was normalized and expressed as a percentage of the level condition. The average of the seven normal subjects and two SCI subjects (*initials*) (each *line* represents the average of five cycles) are presented. The *gray area* represents the 95% confidence interval of the level condition for the normal group and each of the two SCI subjects

cles during stance phase and leg muscles during push-off period (SOL and MG) and swing phase (TA) were averaged for normal subjects at slow (0.4 m/s) and comfortable (1.0 m/s) speeds as well as for SCI subjects walking at a self-selected speed (Fig. 5). Significant main effects due to group (normal vs. SCI, $p<0.05$) and grade ($p<0.01$) were found in VL, SOL and MG, but not in MH and TA. Post-hoc tests revealed that significant differences were present only between 0 and 10° grades for these three muscles. For the VL, the net amplitude change from 0 to 10° grade was significantly larger ($p<0.01$) in the normal group walking at 0.4 m/s than in SCI group, whereas, for the SOL and MG, the changes from 0 to 10° grade were significantly larger ($p<0.01$) in the normal group walking at 1.0 m/s than in the SCI group. Thus, significant adaptation was mainly observed in the extensors.

Uphill walking induced modulation in the timing and duration of EMG activity in normal subjects (Fig. 6). Increasing the treadmill grade from 0 to 15° led to a progressive increase in the duration of thigh muscles. This increase in duration was mainly attributed to a shift in the offset of EMG activity toward the end of the stance phase. This shift was also observed in the SOL and MG, although less pronounced. For the TA (not shown in Fig. 6), uphill walking did not alter either the timing or the duration of EMG activity. In SCI subjects, all muscles showed prolonged activation. Limited adaptation in terms of changes in timing and duration of EMG activities were observed in uphill walking. Significant main effects due to group (normal vs. SCI) were found in all muscles ($P<0.05$) except TA, whereas significant main effects due to grade were found only in the MH ($P<0.01$). Post-hoc tests revealed that significant differences were present between 0 and 10° grades. The net amplitude change from 0 to 10° grade was significantly

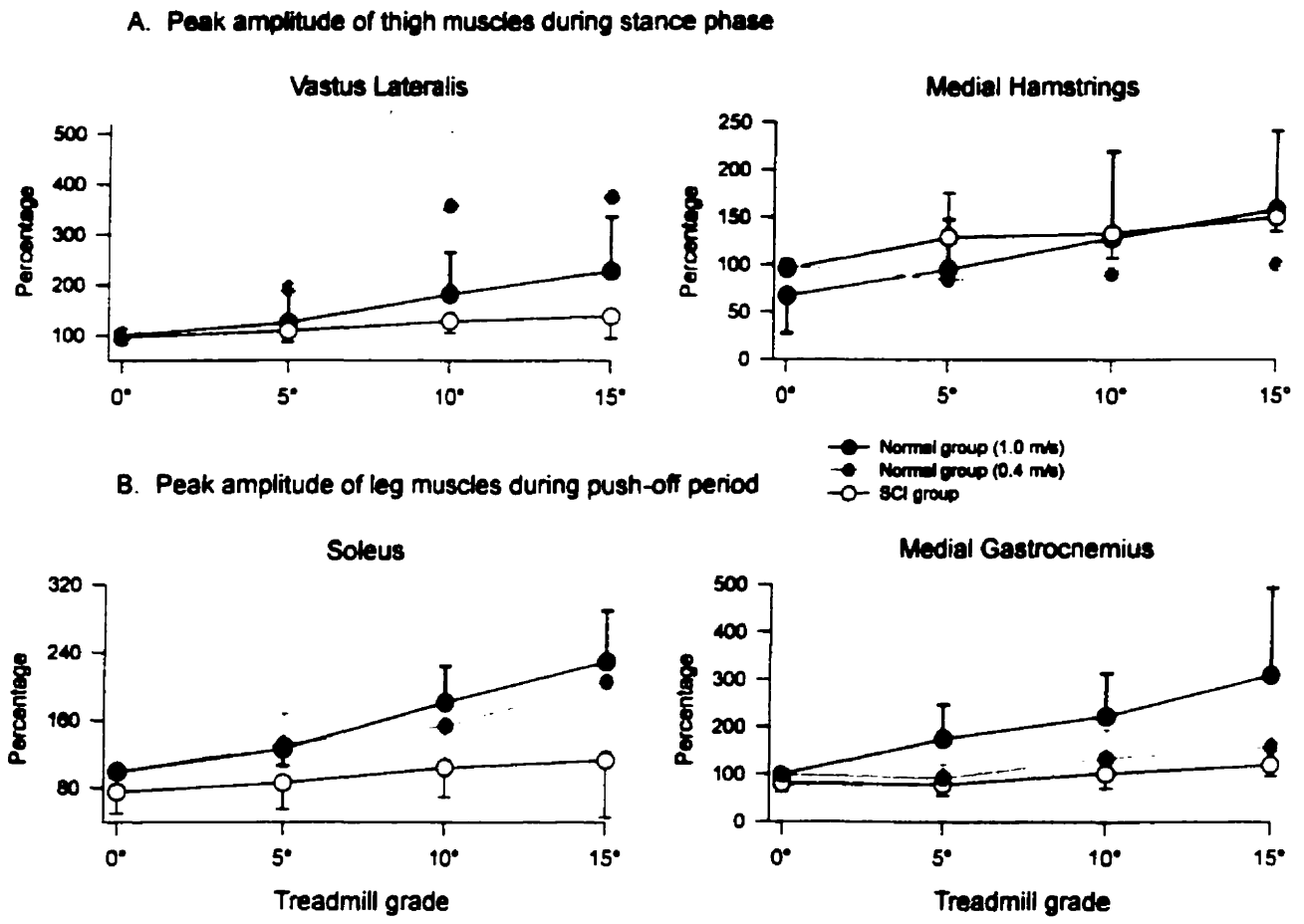


Fig. 5A, B Average peak amplitude (mean ± 1 SD) of thigh muscles during the stance phase (**A**) and leg muscles during push-off (**B**) for the normal group walking at 1.0 and 0.4 m/s and the SCI group walking at self-selected speeds during uphill conditions. The amplitude of electromyographic activity was normalized and expressed as a percentage of the level condition

larger ($P<0.01$) in the normal than in the SCI group (31.3% vs. 7.3%).

Discussion

This study showed that SCI subjects could adapt to uphill walking, albeit with certain limitations. Only three of the seven SCI subjects performed every walking grade, while the rest reached only the 10° grade. However, the strategies of adaptation were different from those of normal subjects.

Adaptation of lower-limb movements

In all normal subjects, the strategy for adapting to uphill walking was to increase hip flexion and ankle dorsiflexion as well as to decrease knee extension of the forward limb from mid-swing to foot contact. Similar findings have been reported during uphill treadmill walking (Lange et al. 1996; Wall et al. 1981). However, in the SCI group, similar adaptation was found only at the hip joint. Major limitations were seen at the knee and at the ankle joints. Both neural and biomechanical factors could give rise to the limitations.

Pronounced knee flexion during foot contact and throughout stance even during level walking has been observed in subjects with spastic paresis of multiple origins (Conrad et al. 1985) and in cats with lesions of the dorsolateral funiculi and dorsal columns (Jiang and Drew 1996). This pronounced knee flexion may have been caused, in part, by hyperactivity in ankle extensors during early stance and also by weakness of knee extensors, as suggested by Winter (1991). When the knee is already in a flexed position, it becomes a biomechanical constraint such that it is more difficult to adapt to the new treadmill grade. Likewise, the limited adaptation observed at the ankle joint might be due to weakness or absence of dorsiflexor activity (as seen in subject PO, Fig. 4). Meanwhile, pronounced knee flexion during foot contact can give rise to increased ankle dorsiflexion during uphill stance. Consequently, increased triceps surae activation is needed to counteract the flexed posture and generate adequate push off, but this could not be achieved by SCI subjects.

Normal subjects use the same hip-knee coordination pattern in uphill walking as in level walking. The major

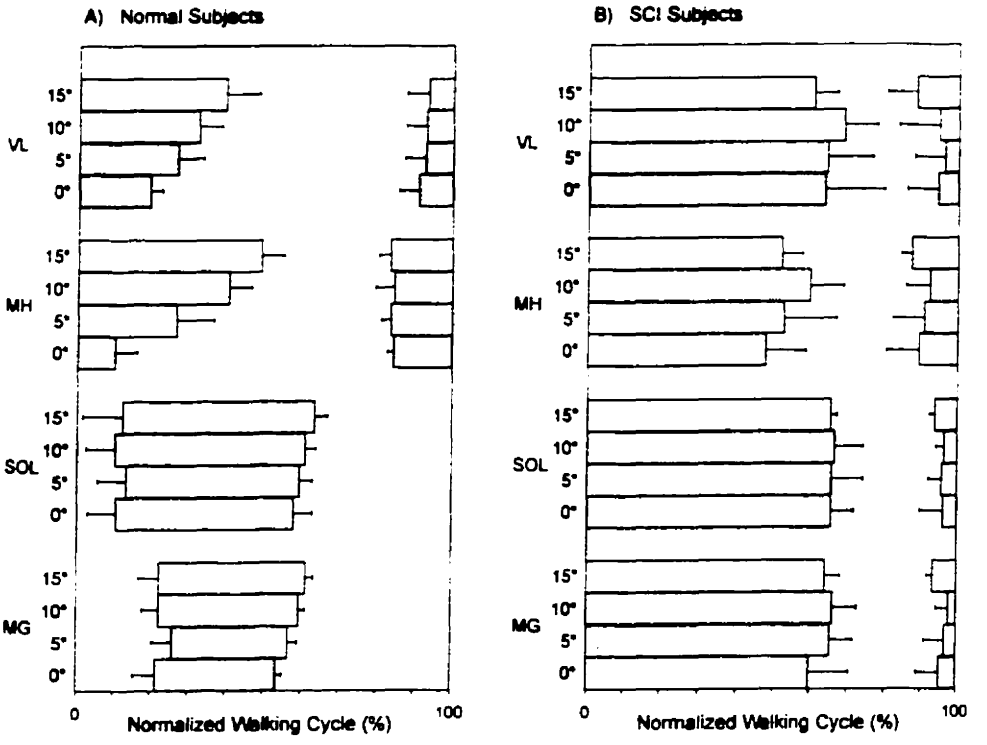


Fig. 6A, B Average timing and duration of electromyographic activity of thigh and leg muscles for the normal (**A**) and the SCI (**B**) groups during walking at 0, 5, 10 and 15°. *VL* Vastus lateralis, *MH* medial hamstrings, *SOL* soleus, *MG* medial gastrocnemius

adaptation feature is a progressive shift of the starting position, which involves a progressive increase in hip flexion and a simultaneous decrease in knee extension towards the end of swing (see Fig. 3). Contrary to normal subjects, many different hip-knee coordination patterns were found in SCI subjects, even during level walking. However, as found in the normal subjects, these patterns were preserved in adapting to uphill walking, at least during the swing phase. Alterations in the coupling between the hip and knee joints during locomotion have also been reported in cats with partial lesions of the spinal cord (Jiang and Drew 1996). In the present study, changes in the timing and magnitude of hip and knee angular movements at specific phases lead to modifications in the coupling between these two joints and resulted in a variety of coordination patterns in SCI subjects.

Adaptation of motor recruitment pattern

Uphill walking induced a progressive increase in the duration and amplitude of the EMG activity of thigh muscles. The increase in duration was caused by a shift in the offset of MH and VL activities toward the end of the stance phase, as seen in both cats (Smith and Carlson-Kuhta 1995) and humans (Tokuhiro et al. 1985) during uphill walking. Such motor adaptation may be required to counteract the increased hip and knee flexion in uphill stance and to bring the whole limb back to an extended position before initiating swing. While a large increase in the amplitude of the SOL and MG is required to generate more propulsion during push off in uphill walking, the offset of these plantarflexors is minimally affected in normal subjects. This is not an unexpected finding because the offset of plantarflexor activity is directly related to toe-off timing, and that event is invariant across grades.

In SCI subjects, all muscles showed prolonged activation throughout stance, without much adaptation to changes in walking grades. This prolonged activation indicates an overall exertion, possibly due to impaired motor recruitment even during level walking, and becomes a limiting factor in adapting to uphill walking. Premature activities were observed in plantarflexors of SCI subjects during all walking grades. Abnormal activity in plantarflexors at the end of the swing phase and in early stance was previously described in subjects with spastic paresis and SCI subjects (Conrad et al. 1985; Fung and Barbeau 1989; Knutsson 1985). This abnormal activity could arise from stretch activation of the triceps surae, which may be a result of defective gating mechanisms of Ia afferents (Fung and Barbeau 1994; Yang et al. 1991). The presence of co-contraction patterns of all lower-limb muscles observed in the present study has previously been reported in SCI subjects (Fung and Barbeau 1989). Co-contraction may be used to counteract muscle weakness and stabilize lower-limb joints throughout the stance phase in SCI subjects.

The lack of voluntary control over the leg muscles in SCI subjects is undoubtedly due to the loss of descending pathways following spinal-cord injury. The absence of an increase in the level of activation of plantarflexors likely results in a weak push off, and compensatory mechanisms from proximal segments might be used to

adapt to uphill walking. For instance, the hip flexors can be recruited to pull the swinging limb upward and forward during uphill walking. Compensatory mechanisms from hip muscles have been previously identified in some gait-related pathologies (Olney et al. 1988; Winter and Sienko 1988). Pelvic elevation on the unsupported side during the swing phase can partially compensate for weakness or paralysis of ankle dorsiflexors and helps to provide adequate toe floor clearance (as shown in subject PO, Fig. 1). Other emerging adaptation strategies may include accentuation of pelvic and trunk rotations during walking, as reported in patients with hemiplegia (Wagenaar and Beek 1992) or unilateral hip pain (Thurston 1985), to compensate for push-off weakness and assist foot swing during uphill walking.

Impact due to level and extent of spinal cord lesion

Our SCI group was comprised of three subjects (LT, PR, RM) with cervical injuries, two subjects (CP, FS) with thoracic injuries and one subject (PO) with both cervical and thoracic injuries (Table 1). The etiology of the injury was traumatic in these six subjects. Results obtained from angular excursions of lower limbs and intralimb coordination during graded walking revealed that important differences were present within each sub-group of injury level, but not between the sub-groups. For instance, RM and FS, who belonged to different injury level sub-groups, show very similar hip-knee coordination patterns when adapting to uphill walking (Fig. 3). In contrast, RM and LT, who belonged to the same sub-group, showed very distinct coordination patterns during both level and uphill walking. Krawetz and Nance (1996) also found no significant difference in lower-limb joint excursions during overground walking in a much larger group (n=27) of SCI subjects with lesions at cervical, thoracic and lumbar levels. Moreover, the injury level did not affect either the amplitude or the timing and duration of lower limb EMG activity during level and inclined walking.

The injury level did not seem to influence the overall performance of the SCI subjects. For the three subjects with cervical injuries, one was able to perform the most elevated treadmill grade, whereas the two others were unable to go beyond the 10° grade. None of the two subjects with thoracic injuries achieved the steeper grade. Furthermore, the two SCI subjects (PR, CP) who achieved the fastest walking speed on the treadmill belonged to two different sub-groups. This is in contrast with Krawetz and Nance (1996), who, in a larger scale study, reported a reduced walking speed for subjects with thoracic lesions compared with subjects with cervical lesions. Other factors, such as spasticity, hypertonia and muscle weakness, could also alter the overall performance of walking and, thus, influence its adaptability.

The extent of the spinal cord lesion can also have an impact on locomotor adaptability. Following partial spinal cord lesions involving the dorsal columns and the dorsolateral funiculi, cats can recover the ability to perform treadmill walking at different inclinations (both roll and tilt) without any support or assistance (Jiang and Drew 1992). However, permanent deficits were seen in the performance of anticipatory adjustments during locomotion, such as stepping over obstacles or walking through a horizontal ladder (Jiang and Drew 1992, 1993). More recently, it has been shown that after ventral and ventrolateral spinal lesions leading to extensive damage to reticulospinal and vestibulospinal pathways, cats had major difficulties in adapting to inclined walking (Brustein and Rossignol 1998). The deficits observed during level walking, which included problems in maintaining lateral stability, inconsistent stepping and occasional stumbling, were accentuated during inclined walking. The lack of hindlimb muscle adaptation during uphill and downhill locomotion was attributed to the loss of reticulospinal axons (Brustein and Rossignol 1998). In the present study, the extent of the spinal cord lesions could not be determined. However, none of our SCI subjects had any major asymmetrical impairment. Clinically, all SCI subjects were classified as "mildly impaired" (D) using the ASIA impairment scale (Ditunno et al. 1994), with motor and sensory functions preserved below the neurological level. Active movement against gravity can be elicited from the majority of key muscles below the lesion. However, the ability to adapt to uphill walking was not correlated with the actual ASIA motor and sensory scores.

In general, the treadmill walking speed seems to be a good indicator of the capability of locomotor adaptation in SCI subjects. Subjects who had faster maximal treadmill speed were the ones who achieved the steeper grades, except for CP, who felt uneasy walking upslope (for fear of falling). On the other hand, no single muscle or kinematic pattern could predict the capability to adapt to the steepest grade. The study of other factors, such as the kinetics of incline walking and the contribution of pelvic and trunk segments, would likely help to discriminate between the capabilities of adaptation in SCI subjects.

Conclusion

The results of this study showed that all normal subjects used the same kinematic and motor strategies to adapt to uphill walking. However, in SCI subjects, a variety of strategies exists. Some of the lower-limb deficits seen after an incomplete spinal cord injury led to alterations in the basic walking pattern and changed the kinematic and motor strategies in adapting to uphill walking. For instance, the lack of changes in knee-flexion angle, the large variations in ankle angle and the limited changes in the peak amplitude of plantarflexor muscles during uphill walking were three major abnormal features observed in the SCI subjects. In general, the capability to adapt to the steepest grade of locomotion cannot be de-

termined on the basis of the level of the injury, the clinical motor and sensory scores, the kinematic pattern or the EMG profile. In contrast, the maximum speed achieved during level treadmill walking was always a good indicator of the patient's ability to walk on an inclined plane.

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