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RESEARCH ARTICLE

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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

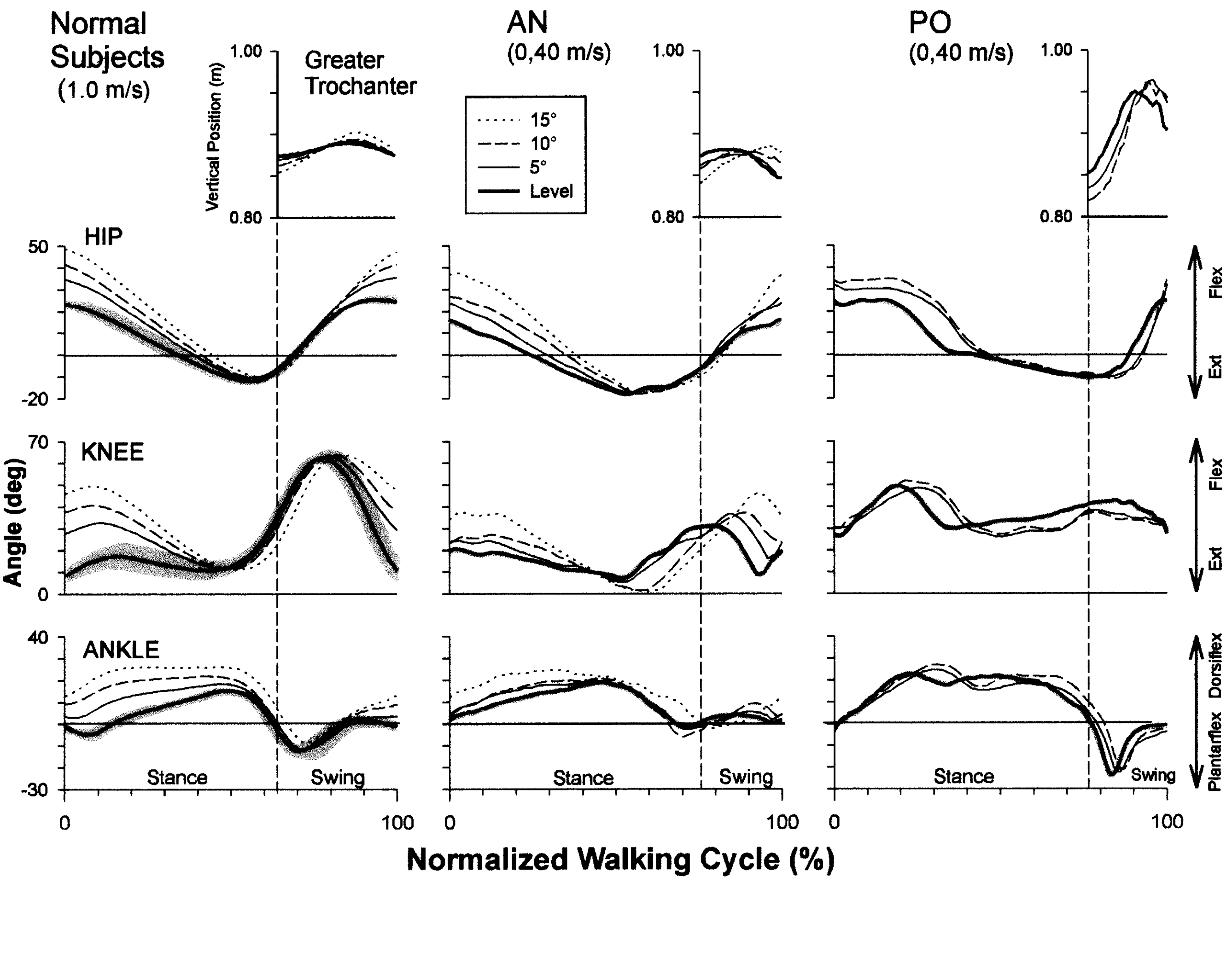

Fig. 1 Angular excursions of the hip, knee and ankle during uphill walking, normalized to the gait cycle (from one foot contact, *0%*, to the next, *100%*). The trajectory of the greater trochanter is shown at the top. This figure shows the average of the seven normal subjects and two SCI subjects (each *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

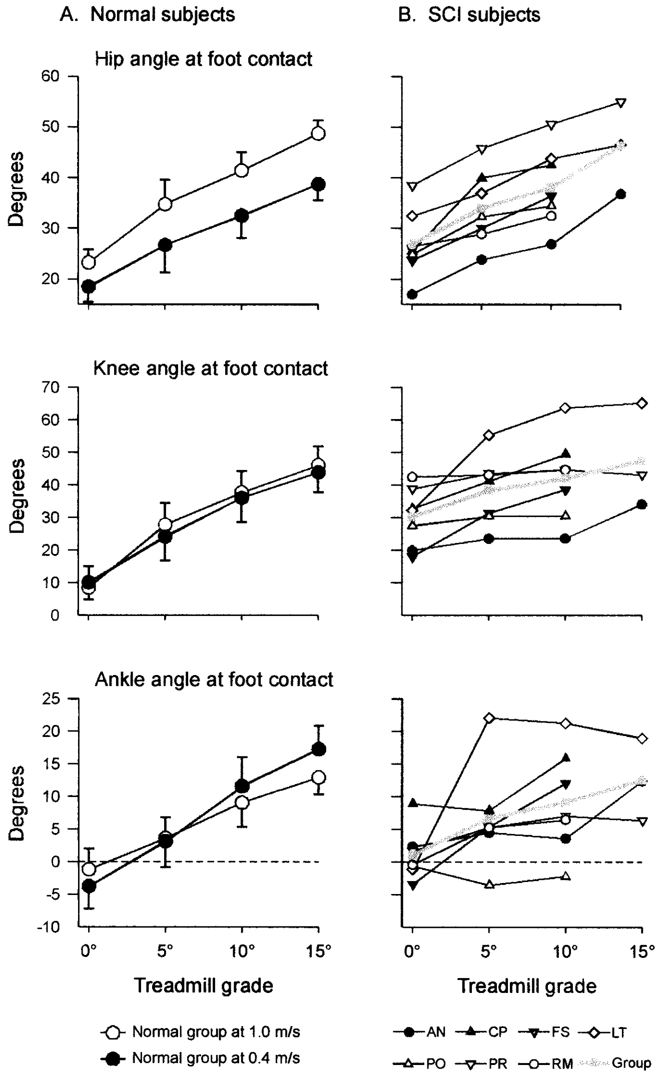

Fig. 2A, B Average hip, knee and ankle angles at foot contact during level and uphill walking for the group of normal subjects (n=7) (**A**) and the individual SCI subjects (**B**). *Gray line* represents the grand average of the seven SCI subjects. (*Initials* Individual subjects)

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-

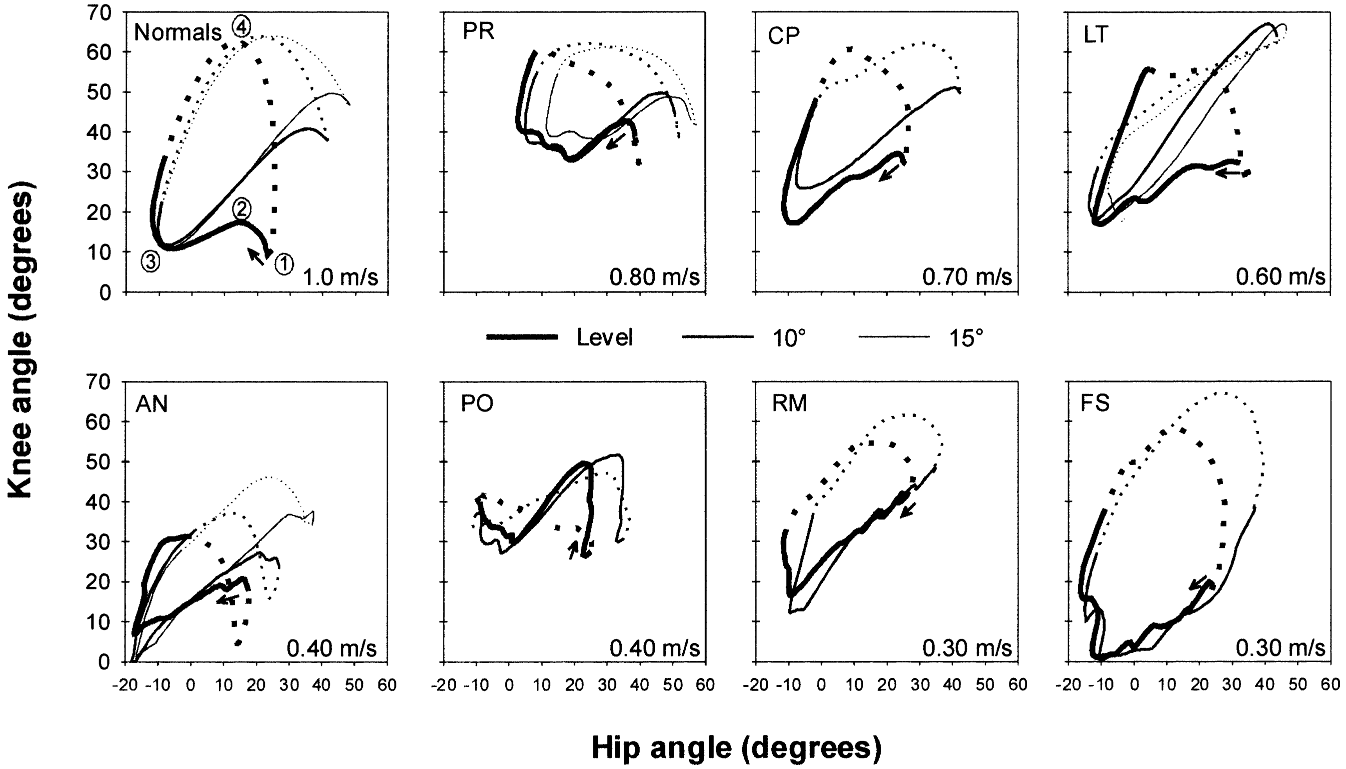

Fig. 3 Cyclographs showing the coupling between the hip and knee joints during uphill walking. This figure shows the average of the seven normal subjects (*normals*) and the seven individual SCI subjects (*initials*) rank-ordered by their comfortable walking speed. The *solid lines* represent the stance 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-

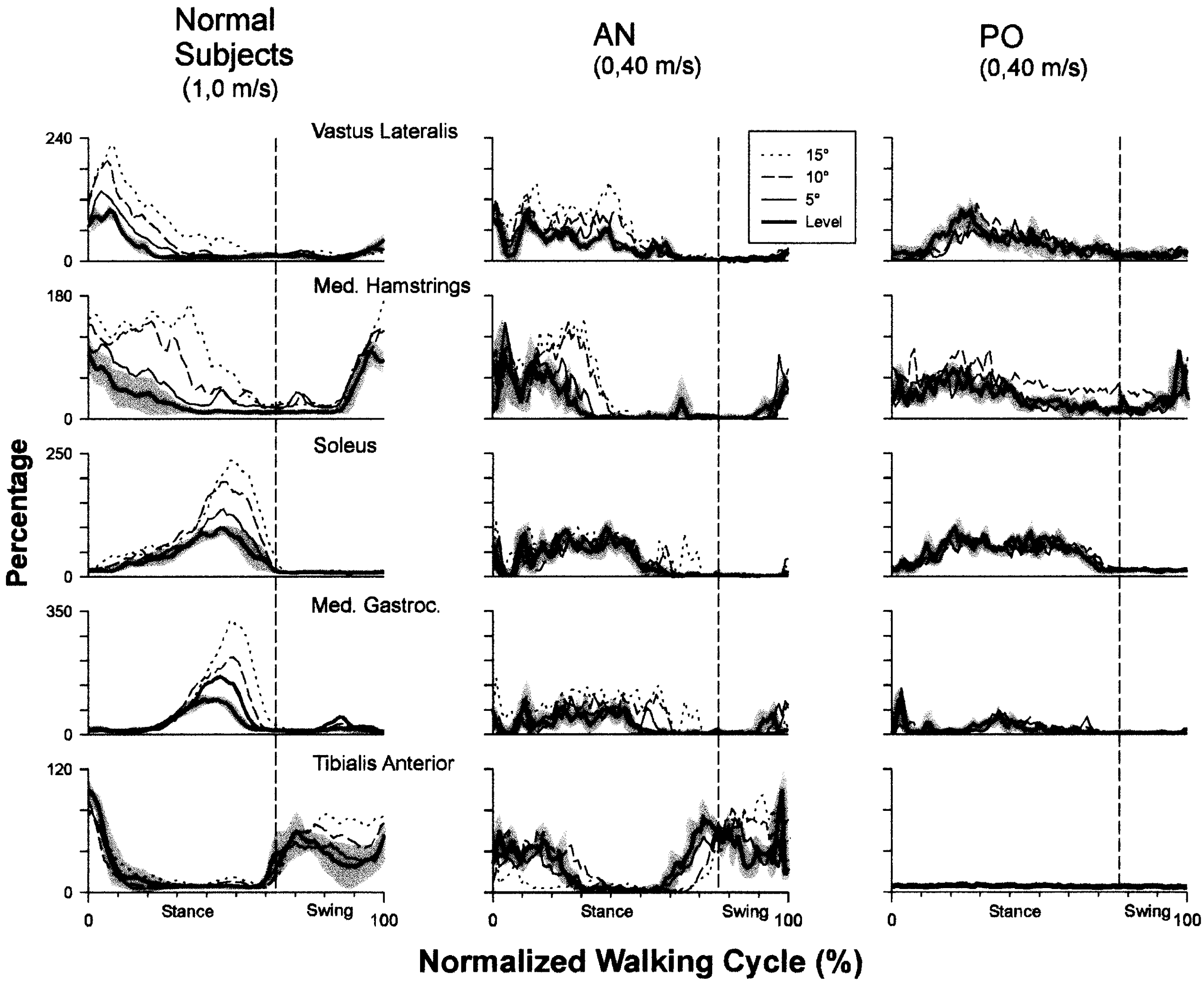

Fig. 4 Electromyographic (EMG) linear envelope of lower-limb muscles during uphill walking, normalized to the gait cycle (from one foot contact, *0%*, to the next, *100%*). Amplitude of EMG activity was normalized and expressed as a percentage of the level condition. The average of the seven normal subjects 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

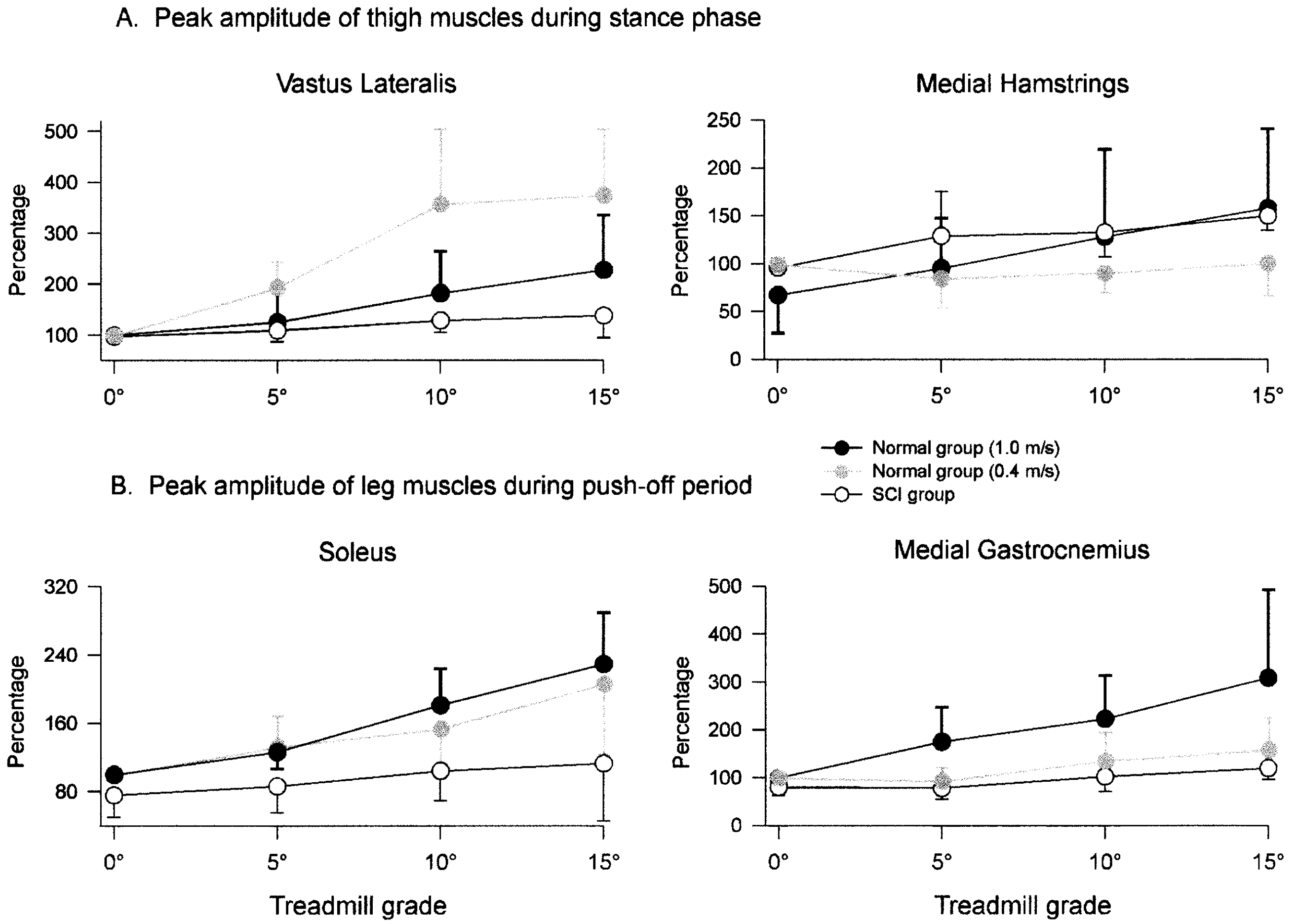

Fig. 5A, B Average peak amplitude (mean ± 1 SD) of thigh muscles during the stance phase (**A**) and leg muscles during push-off (**B**) for the normal group walking at 1.0 and 0.4 m/s and the SCI group walking at self-selected speeds during uphill conditions. The amplitude of electromyographic activity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

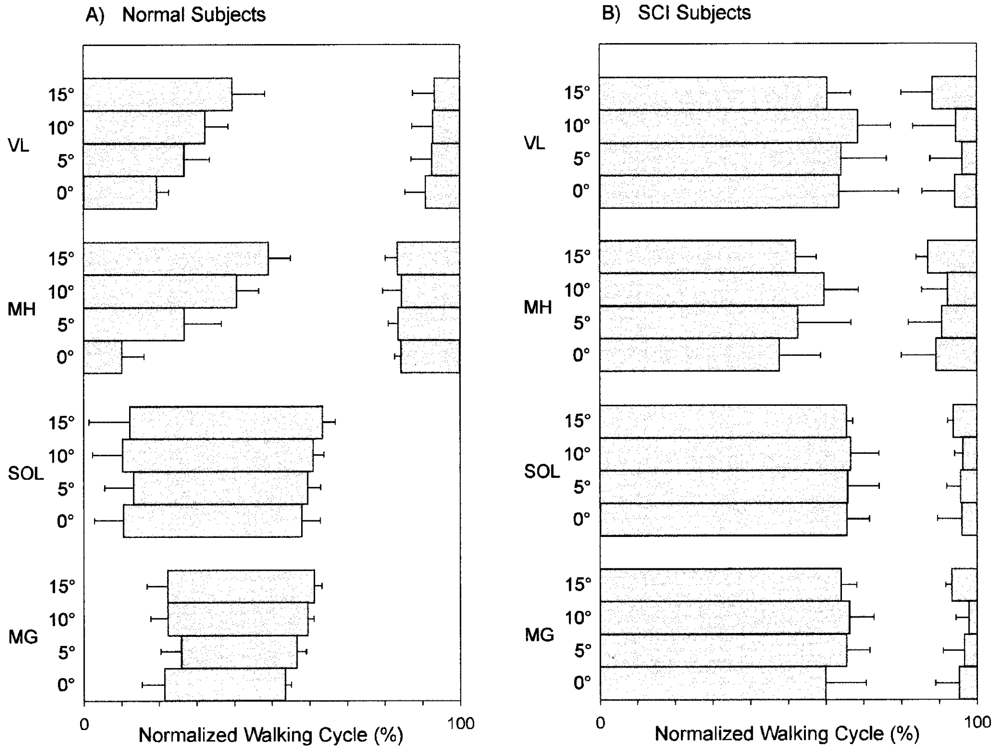

Fig. 6A, B Average timing and duration of electromyographic activity of thigh and leg muscles for the normal (**A**) and the SCI (**B**) groups during walking at 0, 5, 10 and 15°. *VL* Vastus lateralis, *MH* medial hamstrings, *SOL* soleus, *MG* medial gastrocnemius

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 subgroups, 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 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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