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Original Article

Treadmill walking in incomplete spinal-cord-injured subjects: 1. Adaptation to changes in speed

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Walking in spinal-cord-injured (SCI) subjects is usually achieved at a lower speed than in normal subjects.

Study design/methods: Time and distance parameters, angular displacements of lower limbs and electromyographic (EMG) activity were measured for seven normal and seven SCI subjects at several walking speeds. Analyses of variance were used for comparing groups and speeds.

Objectives: First, to measure the adaptability of SCI subjects' walking pattern to different speeds (0.1–1.0 m/s), and to compare it to that of normal subjects. Second, to characterize SCI subjects' walking pattern as compared to that of normal subjects at a matched treadmill speed (0.3 m/s).

Setting: University-Based Human Gait Laboratory, Montreal, Canada.

Results: SCI subjects' pattern adapted to a limited range of speeds. Longer cycle duration, flexed knee at foot contact, increased hip joint flexion at foot contact and during swing, and altered coordination of hip and knee joints were found for the SCI group. At all speeds, duration of muscle activity was longer in the SCI group and the increase in amplitude of soleus EMG activity at higher speeds was not specific to push-off. The importance of matching the walking speed of SCI and normal subjects in order to differentiate the features that are a consequence of SCI subjects' reduced walking speed rather than a direct consequence of the injury is demonstrated.

Conclusions: All SCI subjects could adapt to a narrow range of speeds and only three could reach the maximal tested speed. This limited maximal speed seems to be a consequence of SCI subjects having reached their limit in increasing stride length and not being able to increase stride frequency further. This limitation in increasing stride frequency is likely because of the altered neural drive.

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Keywords: spinal cord injury; gait; speed; treadmill walking

Introduction

The deficits resulting from a spinal cord injury can be very detrimental, even when the loss of motor and sensory functions is incomplete. Motor deficits include muscle weakness or paralysis and spasticity. Some spinal-cord-injured SCI individuals with an incomplete loss of motor function can have enough residual sensory and motor function to walk, but the speed is usually greatly reduced and the walking pattern differs from that of normal individuals. Gait patterns of SCI subjects are characterized by absent, weak or abnormal muscle activity, changes in hip, knee and ankle angular displacements,[1–3] and hyperactive reflexes[4–6] contributing to abnormal activation of triceps surae. The muscle weakness and spasticity result in deficient propulsion during stance, and foot drag during swing. Also, dyscoordination in the lower limb joints, as revealed by cyclographs of joint angular displacements, has been found in SCI subjects during level and incline walking.[7] The walking ability of SCI subjects can be greatly improved quantitatively and qualitatively with appropriate interventions.[8–10] However, it is not known if this is also reflected in the capacity to adapt to changes in the

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mechanical demand of the task, as often observed in a changing environment. Thus, an important question remains: can SCI subjects adapt their walking pattern over a wide range of speeds?

In normal subjects, the locomotor pattern is easily adapted to increase in the walking speed. This adaptation is generally obtained by increasing the cadence (stride frequency) and the stride length simultaneously,[11–17] although it is possible to adapt somewhat by modifying only one aspect.[15] The adaptation to higher walking speeds also includes greater joint angular excursions,[17–21] and changes in the timing or in the amplitude of the activity of muscles controlling lower limb joints.[17,22–23]

Gait impairments resulting from neurological injuries or disorders are usually determined by comparing the gait of neurologically impaired subjects to that of normal subjects walking at their preferred speed. However, in SCI subjects that were observed in the laboratory, the preferred speed was usually two to 10 times slower than the preferred speed of the average normal subject. Since there are important differences in the pattern that are related to speed, comparing SCI and normal subjects while walking at their respective preferred speeds gives an incomplete picture of the differences between the two groups. While this comparison allows a measure of how different the SCI subjects' walking is from normal walking, it fails to identify the differences that are a consequence of their low walking speed. Preliminary results showed that some of the characteristics in the gait of SCI subjects that have been reported could be attributed to the reduced walking speed rather than to the deficit caused by the neurological impairment.[24] Further, the gait of normal subjects at very low speeds (lower than 0.4 m/s) has not been documented. Since the severity of impairments limits the preferred treadmill walking speed to less than 0.3 m/s in many subjects among the SCI population, it is important to measure the gait of normal subjects at very low speeds.

Therefore, the main objective of this study was to measure the adaptation of the gait pattern to different treadmill walking speeds in normal and SCI subjects and to assess whether these adaptations are performed in a similar way in both groups. Further, preliminary data showed that there were large differences in temporal and angular values when the gait of SCI subjects at a reduced speed was compared to that of normal subjects at comfortable speed. For instance, the cycle duration for the SCI group at 0.3 m/s was more than twice the value of the normal group at 1.0 m/s. Thus, another objective was to compare gait patterns of SCI and normal subjects when walking at the same speeds. This study presents characteristics and limitations of gait in SCI subjects, and features that are directly associated with their deficits and are not a direct result of the low walking speed.

Methods

Subjects

Convenient samples of seven incomplete SCI subjects (six males, one female) and of seven normal male subjects (aged between 28 and 40 years) were selected for the study. SCI subjects' profiles are presented in Table 1. They were graded as 'D' according to the ASIA international standards,[25,26] able to walk overground independently with or without walking aids for at least 20 steps and free of any major medical problems. Their reported preferred walking speed on the treadmill ranged from 0.3 to 0.8 m/s, although some of the SCI subjects could walk beyond 1.0 m/s. The protocol of this study was approved by the University Departmental Ethics Committee.

Experimental procedure

Data were recorded in the Human Gait Laboratory at McGill University. Subjects walked on a custom-designed treadmill allowing precise control of the

Table 1 SCI subjects' profiles

Subjects	*Age (years)*	*Cause and level of injury*	*Time since injury*	*Walking aids*	*Overground walking speed*[a]	*Preferred treadmill speed*[a]	*Maximal treadmill speed*[a]
S1	20	Trauma from diving, C5	2.8 years	One forearm crutch	0.66	0.60	1.00
S2	38	Trauma from diving, C6	>20 years	Forearm crutches	0.45	0.30	0.60
S3	25	Trauma, T8–T9	>6 years	Forearm crutches	0.27	0.30	0.60
S4	46	Trauma from MVA,[b] T11	14 months	None or one cane	0.87	0.70	1.10
S5	16	Trauma from MVA,[b] C7–T1	10 months	Two canes	0.27	0.40	0.70
S6	20	Trauma from MVA,[b] C6	>5 years	One cane	0.95	0.80	1.30
S7	24	Spastic paresis, nonfamilial origin	>14 years	None	0.36	0.35	0.80

The walking aids are those used by the subjects for their daily activities. The comfortable overground walking speed was measured on a 4 m walkway. All subjects were graded as 'D' according to the ASIA international standards[25,26]
[a]Speed is in m/s; [b]MVA, motor vehicle accident

walking speed.[27] This treadmill is driven by a hydraulic motor allowing speeds ranging from 0.02 to 2.0 m/s.

Foot contacts were recorded bilaterally using footswitches. Kinematic and EMG data were collected from the left side for normal subjects and from the more affected side for SCI subjects. The more affected side was determined through an evaluation (manual muscle tests and spasticity tests of lower limb joints) done by a physiotherapist. Each subject had a familiarization period on the treadmill of approximately 5 min for normal subjects and 1–5 min for SCI subjects, depending on their endurance.

Data were recorded while subjects walked at different treadmill speeds. Normal subjects were evaluated at speeds ranging from 0.1 to 0.7 m/s, and at 1.0 m/s. SCI subjects were evaluated from 0.1 m/s to their maximal speed, with increments of 0.1 m/s. All subjects were required to hold the handlebars located on both sides on the treadmill. For normal subjects, the different speed conditions were presented by increments of 0.1 m/s either in an increasing or in a decreasing order. For each condition, normal subjects walked for at least 20 cycles to let them habituate to the new speed. At lower speeds, especially at 0.1 m/s, some subjects needed more time before establishing a stable gait pattern. For the SCI subjects, the preferred speed was determined while they were getting more familiar with treadmill walking. Thereafter, the treadmill speed was brought down to 0.1 m/s and speed was subsequently increased by increments of 0.1 m/s until they reached their maximal speed (as judged by the subject). This procedure allowed them to adapt gradually to the increased speed. SCI subjects were given time to adapt to the new speed condition providing that they had enough endurance to do so. Otherwise, only the maximum time (at least 10 consecutive cycles) that a subject could walk at a given speed was recorded. When needed, subjects took rest periods of about 2 min.

Data recording and instrumentation

Pressure-sensitive footswitches were attached to the soles of the subject's shoes, indicating the heel, fifth metatarsal and big toe contacts with the treadmill belt. Footswitch signals were used to determine the cycle, stance and swing durations, as well as the single and double limb support periods. Reflective markers were placed over the head of the fifth metatarsal, the lateral aspect of calcaneus, the lateral malleolus, the knee joint line and the greater trochanter (GT). Displacements of the markers were recorded unilaterally on VHS video recorders (Panasonic, AG-7350) with two CCD 60 Hz video cameras (Panasonic 5100) placed at an angle of about 90°. Halogen spotlights were positioned to optimize reflection of light from the markers towards the camera. A shutter speed of 1/1000 s was used to alleviate the blur occurring during faster movements. The activity of the following four muscles were recorded unilaterally: *tibialis anterior* (TA), *soleus* (SO), *semitendinosus* part of medial hamstrings (MH) and *vastus lateralis* (VL). Following appropriate skin preparation, surface preamplified electrodes were placed near the estimated motor point of the muscle.[28] EMG signals were differentially amplified and bandpassed (10–1000 Hz) and recorded along with footswitch signals on a 14-channel magnetic FM tape recorder (Honeywell #101) and processed off-line. Synchronization of video, EMG and footswitch signals were obtained using SMPTE time code generators (Skotel: TCG-80).

Data analysis

For each subject, five consecutive walking cycles were used for the averaging of the video and EMG signals in each of the different conditions. All signals were analyzed off-line. The video recordings were analyzed with the Peak Performance® 3-D motion analysis system (version 5.1). The 2-D coordinates for each marker were used to calculate angular displacements of hip, knee and ankle joints in the sagittal plane. For the hip joint, the angle is measured between the thigh segment and the vertical line passing through the GT. For the knee joint, the measured angle is that formed between the projection of the thigh segment and the leg segment. For the ankle joint, the angle formed between the leg segment and the foot segment is measured. Angular displacements were filtered with a fourth-order, double-pass Butterworth filter at 6 Hz. The footswitch and EMG signals were digitized at a sampling rate of 1200 Hz using a custom-designed program. EMG signals were digitally bandpassed (20–400 Hz), rectified, smoothed using a window of 20 samples, and averaged using the DataPac II® program (Run Technologies).

Stance and swing onset and offset were determined manually on the computer screen from the footswitch signals, using the DataPac II® program. This allowed the calculation of cycle, stance and swing durations. The displacement of GT was normalized to the height of GT during standing. In other words, the relative displacement was obtained by subtracting the height of GT during standing from the value obtained during walking. Before averaging, the kinematic and EMG data were normalized to the stance and to the swing durations separately, rather than to the cycle duration. The main reason for normalizing in this way was that the stance and swing durations were drastically different between 0.1 and 1.0 m/s, both in absolute and relative values. There was also a difference in the stance and swing durations between the two groups for any given walking speed. These differences make it difficult to compare directly the magnitude and timing of the changes occurring in the kinematic or EMG data among the different conditions and between the two groups. This method of normalization allows a better comparison of the magnitude and timing of any changes occurring in the kinematic or EMG data among the different conditions and between the two groups.

A mean angular displacement profile was calculated for the normal group for all speed conditions. This profile was compared to individual SCI subjects' profiles. In addition, hip angular excursions as well as knee and ankle angular values at specific points of the gait cycle were measured at 0.1, 0.3, 0.5 and 1.0 m/s. These specific points are presented in Table 2. In addition, in order to measure the intralimb coordination, hip/knee as well as knee/ankle cyclograms (the angular displacement of one joint plotted as a function of another joint) were analyzed. Cyclograms have been used to describe the adaptation of SCI subjects' gait to different treadmill inclines.[7] From subjects' EMG profile averages, mean patterns of the onset and offset of EMG activity were calculated for both groups, for all speed conditions. The onset and offset of muscle activation were determined manually. The onset of EMG activity was determined as the point when the amplitude of activity rose and remained above the resting noise level for at least 25 ms. Similarly, the offset of EMG activity was determined when the activity returned and remained at the baseline level. However, in some subjects, the activation patterns of TA and MH muscles were continuous with two clear peaks of activation, whereas in other subjects, two bursts separated by a brief moment of inactivity were observed. For clarity and simplicity, the onset and offset of the activity of these muscles were considered as the whole period of activation, even when the bursts were briefly interrupted.

In addition, the maximum amplitude of SOL during push-off was also measured. The epoch for push-off was determined as the portion between 50 and 90% of the whole stance duration. For each subject, the values obtained at each speed were expressed as a percent of the values obtained at 0.1 m/s. Then group averages for each speed were calculated. The SOL muscle was selected for this evaluation as it has been previously demonstrated that approximately most of the energy during walking in normal subjects is produced by the triceps surae during push-off.[29] Thus, it was anticipated that SOL will show important changes in amplitude as the walking speed changes.

Table 2 Angle measurement at specific points of the walking cycle for the kinematic comparison of normal and SCI groups

Hip	*Knee*	*Ankle*
Total excursion	Angle at foot contact	Angle at foot contact
Maximum	Minimum angle during stance	Angle at toe-off
Maximum flexion	Maximum angle during swing	Maximum dorsiflexion during swing

Some of the specific angular measurements were chosen because of their importance in the determination of some of the gait characteristics. Logically, and based on anecdotal observations, the total hip excursion and the knee angle at foot contact are determining factors of the stride length. Other specific measurements were chosen because they corresponded to points for which important differences in the angular displacement profiles were observed between the two groups. This was the case for the knee and ankle angles at foot contact and the minimal knee angle (maximal extension) during stance. Finally, the ankle angle at toe-off gives an indication of the amount of plantarflexion during push-off, whereas the maximal knee angle (maximal flexion) and the maximal ankle dorsiflexion during swing give an indication of the control of the swinging limb. These measurements allow the establishment of quantitative comparisons between the two groups for different speed conditions

Statistical analysis

Individual and group averages were calculated for cycle, stance and swing durations, stride length, joint angle values at specific points of the cycle, onset and offset of muscle activity, and maximum amplitude of soleus EMG activity. Comparisons between groups and among speed conditions were done for time and distance parameters and joint angle values using an analysis of variance (ANOVA) for repeated measures with a grouping factor (Systat® software). Comparisons between groups for the 0.3 m/s condition were done using a one-way ANOVA. Comparisons between 0.3 and 1.0 m/s for the normal group were done with a one-way ANOVA with repeated measures. A one-way ANOVA was also used to compare groups on the ratio of the maximal knee excursion over and the maximal hip excursion for stance and swing phases. An alpha value of 0.10 was used to determine significance of results.

Results

SCI subjects individual results for temporal and distance parameters

In Figure 1, results for the temporal and distance parameters for each SCI subject and the 95% confidence interval of the mean for the normal group are presented. The relation of cycle and stance durations as a function of the walking speed is curvilinear, both in normal and in SCI subjects. Only three of the SCI subjects could reach the speed of 1.0 m/s, but all subjects walked at speeds up to 0.5 m/s. Three of the SCI subjects show stride length and cycle durations values (Figure 1a and b) that are outside the 95% confidence interval of the normal subjects' average at 0.1 m/s, whereas at 0.3 and 0.5 m/s, five SCI subjects are outside the confidence interval. In contrast, at 0.7 and 1.0 m/s, only one subject remains below the confidence interval. For stance duration (*C*), two subjects are outside the confidence interval at 0.3 and 0.5 m/s. In contrast, for swing duration (*D*), all but one SCI subjects have values that are outside the confidence interval from 0.1 to 0.7 m/s, and at 1.0 m/s, the three subjects are below the interval.

Group means for temporal and distance parameters

In Table 3, group averages for cycle duration, stance and swing durations in absolute and relative values, and stride length are presented for the 0.1,. 0.3, 0.5 and

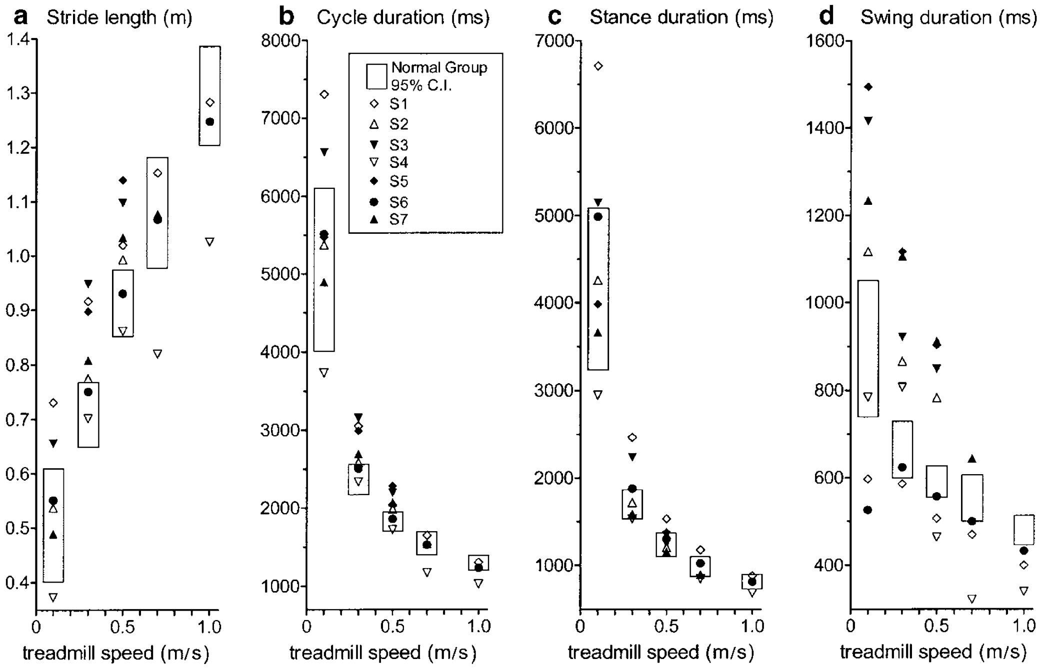

Figure 1 Temporal and distance parameters as a function of walking speed. Stride length (**a**), cycle duration (**b**), stance and swing (**c** and **d**) durations expressed as a function of the walking speed for all SCI subjects. The boxes represent the 95% confidence interval of the mean for the normal group

1.0 m/s conditions. Since only three SCI subjects could walk at 1.0 m/s, the ANOVA for the comparison between groups and between speed conditions was conducted only for the 0.1, 0.3 and 0.5 m/s conditions. Therefore, the results for the 1.0 m/s in the SCI group are presented but were not part of the statistical analysis.

General trends for speed-related changes There are significant decreases of cycle, stance and swing durations and an increase of stride length for both groups as speed increased. The relative stance duration is smaller and, consequently, the relative swing duration is greater with increases in the walking speed.. The cycle duration tends to be slightly longer for the SCI group, but this difference is not statistically significant. In contrast, longer swing duration values are observed in the SCI group, whereas no differences were found in average stance duration values between groups. On average, SCI subjects have a greater stride length than normal subjects for each of the three speeds up to 0.5 m/s. In both groups, when the speed is increased, the relative stance duration is decreased, whereas the relative swing duration is increased.

In the normal group, the comparison between 0.3 and 1.0 m/s conditions shows significant differences for cycle ($F_{(1,6)} = 138.434$; $P = 0.000$), stance ($F_{(1,6)} = 139.124$; $P = 0.000$) and swing ($F_{(1,6)} = 33.899$; $P = 0.001$) durations. Similarly, when stance and swing are expressed in percent of cycle duration, there is a significant difference ($F_{(1,6)} = 80.444$; $P = 0.000$) between 0.3 m/s (72 and 28%) and 1.0 m/s (63 and 37%).

Comparison between groups at the same speed At 0.3 m/s, there is a longer cycle duration for the SCI group than for the normal group ($F_{(1,12)} = 6.771$;. $P = 0.023$), primarily because of a longer swing duration ($F_{(1,12)} = 12.117$; $P = 0.005$). Therefore, it should be expected that the SCI group has a greater stride length than the normal group ($F_{(1,12)} = 6.770$; $P = 0.023$). In addition, the comparison of stance and swing expressed in percent of cycle duration shows small but significant differences ($F_{(1,12)} = 3.715$; $P = 0.078$) between the two groups (72 and 28% for the normal group versus 69 and 31% for the SCI group).

Angular displacement profiles
Figure 2a and b present the average angular displacement profiles as a function of the walking speed for the normal group (a) and for one SCI subject (b). Functionally, S1 could walk overground without aid, although he normally used one forearm crutch. The length of the stance and swing axes represent, respectively, 62 and 38% of the whole duration axis; this ratio was used because it was the ratio measured in the normal group for the 1.0 m/s condition.

For normal subjects, the mean angular displacement profiles are shown for speeds of 0.1, 0.3, 0.5 and 1.0 m/s along with the 95% confidence interval for the 1.0 m/s condition. Some striking changes as a function of the speed can be observed for all three joints. For the lower speed conditions, the amount of hip extension at the end of stance and the amount of hip flexion during swing and early stance are substantially reduced as compared

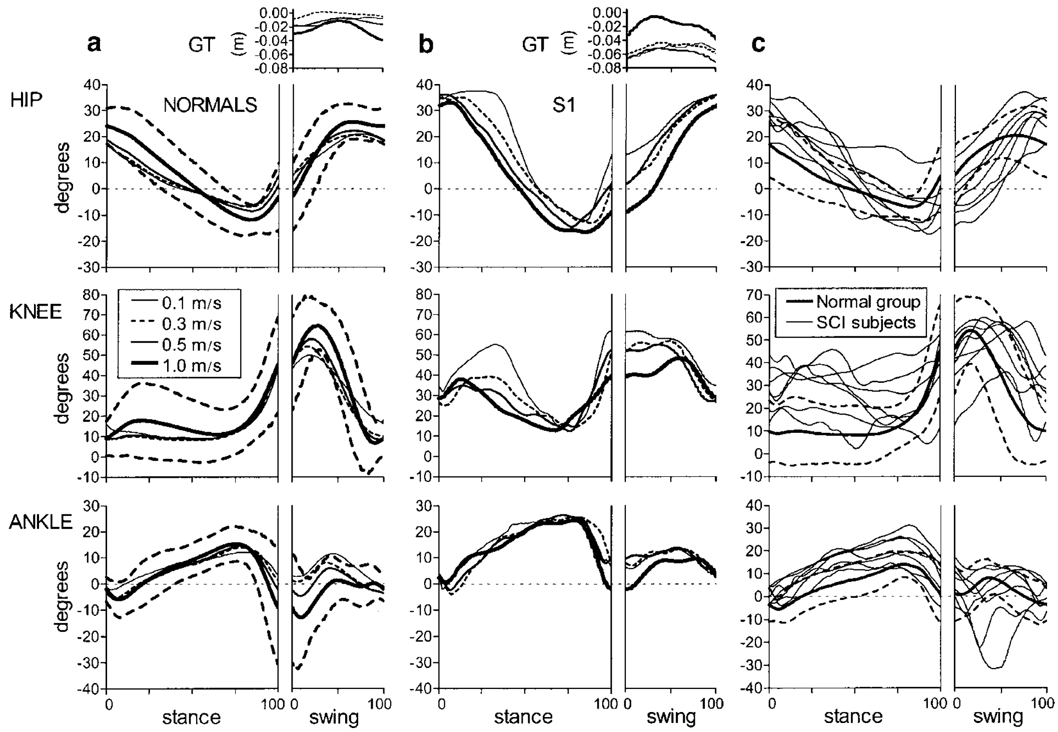

Figure 2 Angular displacements as a function of walking speed. Angular displacements of hip, knee and ankle joints at 0.1, 0.3, 0.5 and 1.0 m/s for the normal group (**a**) and one SCI subject (**b**). The thick dashed lines in the left panel represent the 95% confidence interval of the normal group average at 1.0 m/s. The upper part of panels (**a**) and (**b**) show the relative vertical displacements of GT relative to its position in quiet standing. The third panel (**c**) shows the normal group's average angular displacement profiles (bold lines) with the 95% confidence interval (dashed lines) at 0.3 m/s along with the individual profiles of all SCI subjects (thin lines). For hip and knee joints, positive values indicate flexion (upward displacement of the curve) and negative values indicate extension (downward displacement of the curve). For ankle joint, positive values indicate dorsiflexion (upward displacement of the curve) and negative values indicate plantarflexion (downward displacement of the curve)

to values obtained at 1.0 m/s. These results are directly linked to the reduced stride length described earlier (Figure 1 and Table 3). Similarly, at the knee joint, the brief flexion normally occurring in early stance, that is associated with weight acceptance, is greatly reduced or absent at speeds of 0.5 m/s or less, as well as the amount of knee flexion during swing. In contrast, the amount of knee flexion at foot contact in the 0.1 m/s condition is increased by about 8° as compared to the 1.0 m/s condition. At the ankle joint, there is a small decrease in the passive dorsiflexion during stance when speed decreases. In contrast, at push-off, the ankle joint remains in a dorsiflexed position during push-off at 0.1 and 0.3 m/s and the amount of dorsiflexion increased during swing.

In comparison, S1 shows a greater amount of flexion at the hip and knee joints at foot contact for all speeds. This finding is also observed in five of the other six SCI subjects. Furthermore, the total hip excursion is greater than that of the normal group at the same speed in five of the seven SCI subjects including S1.

The relative vertical displacement of the marker placed over GT during the swing phase is presented in the top part of Figure 2a and b. In the normal group, the height of GT is very slightly lowered as the speed increased. In contrast, S1 shows a net increase in the height of GT at 1.0 m/s. This strategy is also observed in other SCI subjects.

Figure 2c presents the average angular displacement patterns of the normal group at speeds of 0.3 m/s along with the individual averages of the SCI subjects (thin lines). A striking finding is the high variability in the angular displacement profiles among the SCI subjects and how they differ as compared to those of the normal group. For all SCI subjects, the profiles of the hip, knee and ankle angular displacements fall outside the 95% confidence interval of the normal group, either in part or for the whole profile. In some SCI subjects, the timing of certain events seems to be changed. The hip joint remains in extension longer than that of the normal group, and shows rapid flexion during the swing period. At the knee joint, foot contact is made with the knee in a flexed position and most subjects maintain the flexion throughout stance. During the swing phase at 0.3 m/s, the knee reaches its maximum flexion later than for the normal group. At the ankle joint, there is more dorsiflexion throughout stance than for the normal group, and during swing, the individual profiles vary from extreme plantarflexion in two subjects, to dorsiflexion beyond the confidence interval of the normal group in two other subjects.

Table 3 Mean temporal and distance data as a function of walking speed

Speed (m/s)	*Cycle (ms)*		*Stance (ms)*		*Swing (ms)*	
	Normal	*SCI*	*Normal*	*SCI*	*Normal*	*SCI*
0.1	5056 (1244)	5868 (1389)	4160 (1100)	4765 (1317)	895 (185)	1104 (426)
0.3	2363 (232)	2843 (370) *	1699 (199)	1967 (369) *	664 (82)	876 (199) *
0.5	1826 (146)	2113 (312)	1235 (158)	1387 (244)	591 (43)	726 (185)
1.0	1295 (109)	1186 (139)	816 (97)	795 (100)	480 (40)	391 (47)
	Not significant		Not significant		***	
	Stride (m)		*Stance (% of cycle)*		*Swing (% of cycle)*	
	Normal	*SCI*	*Normal*	*SCI*	*Normal*	*SCI*
0.1	0.506 (0.124)	0.587 (0.139)	81.9 (3.0)	80.7 (6.9)	18.1 (3.0)	19.3 (6.9)
0.3	0.709 (0.070)	0.853 (0.111) *	71.8 (3.0)	68.9 (7.0) *	28.2 (3.0)	31.1 (7.0) *
0.5	0.913 (0.073)	1.056 (0.156)	67.4 (3.8)	65.8 (7.0)	32.6 (3.8)	34.2 (7.0)
1.0	1.295 (0.109)	1.186 (0.139)	62.9 (3.0)	67.0 (1.8)	37.1 (3.0)	33.0 (1.8)
	****		Not significant		Not significant	

Group averages ($n = 7$) at speeds of 0.1, 0.3, 0.5 and 1.0 m/s for cycle duration, stance and swing durations, stride length, and stance and swing in percent of cycle duration. The numbers in parentheses represent the standard deviation of the mean. Note that there were only three subjects in the SCI group at 1.0 m/s and results from this speed were excluded from the analysis of variance
$^{*}F_{(2,24)}22.876$; $P = 0.000$ in all cases; $^{**}F_{(2,24)} = 86.595$; $P = 0.000$; $^{***}F_{(1,12)} = 5.959$; $P = 0.031$; $^{****}F_{(1,12)} = 3.704$; $P = 0.078$

Group means for specific angular measurements

Table 4 summarizes the results obtained for the specific angle measurements for all three joints. Only values for 0.1, 0.3 and 0.5 m/s were used in the statistical analysis. A first general observation is that the motion of the lower limbs during walking is speed sensitive. An increase in speed is associated with an increase in joint excursions in both groups. At the hip joint, significant increases in the total excursion, maximal extension and maximal flexion are observed in both groups when speed increases. Total excursion and maximal flexion are significantly greater for the SCI group at all speeds.

Flexion at the knee during stance is greater for the SCI group across all speeds. The difference is significant at foot contact, but not for the minimum angle because of the wide variability in this parameter among SCI subjects. There is no difference between groups for maximum knee flexion during swing. The knee flexion increases with speed, but the increase is mostly because of the normal group. The behavior of the two groups is significantly different when treadmill speed increases ($F_{(2.24)} = 5.995$; $P = 0.008$).

The ankle angle at foot contact is significantly different between groups across all speeds. Both groups behave similarly as speed increases, by changing the ankle angle towards plantarflexion at foot contact and at toe-off. However, ankle position remains in dorsiflexion at foot contact and at toe-off for the SCI group. The amount of dorsiflexion during swing is significantly diminished as walking speed increases.

Table 5 compares groups at 0.3 m/s and compares values from normal subjects across 0.3 and 1.0 m/s. One of the most striking differences occurs at the hip joint. The SCI group shows greater hip flexion and total hip excursion than the normal group at 0.3 m/s. In fact, the hip excursion in the SCI group at 0.3 m/s is in the same range as that seen in the normal group at 1.0 m/s (see Table 4).

Joint angular relationships

In Figure 3, the hip–knee angular displacement relations of all SCI subjects are compared to that of the normal group average at 0.1, 0.3, 0.5 and 1.0 m/s. From these cyclographs, it is clear that each SCI subject has a different coordination pattern between these two joints. In some cases (S1, S3, and S4), the patterns show some similarities to that of the normal group, whereas in others (S5 and S7) they are completely different. The orientation and the shapes of the cyclograms for SCI subjects is changed because of the angle of the knee joint at foot contact (see point 1, Figure 3). The cyclographs show couplings between the two joints. Coupling is defined by a quasilinear portion of the cyclograph curve with a positive slope (see point 2, Figure 3) and means that both joints are moving at the same time in the same direction. In most SCI subjects, there is a coupling in

Table 4 Specific angular measurements as a function of walking speed

Speed (m/s)	*Total excursion* Normal	*Total excursion* SCI		*Maximal extension* Normal	*Maximal extension* SCI		*Maximal flexion* Normal	*Maximal flexion* SCI	
Hip angles (in degrees)									
0.1	28.0 (5.2)	33.0 (10.5)	* $F_{(2,24)} = 8.775$;	−6.7 (3)	−5.9 (12)	* $F_{(2,24)} = 18.350$;	21.3 (5.8)	27.1 (6.9)	* $F_{(2,24)} = 10.747$;
0.3	28.7 (4.4)	37.7 (7)	$P = 0.001$	−7.5 (2.4)	−8.6 (9.5)	$P = 0.000$	21.3 (5)	29.2 (6.4)	$P = 0.000$
0.5	32.1 (3.7)	42.9 (6.6)		−9.3 (2.5)	−11.9 (9.6)		22.9 (4.5)	31.1 (5.8)	
1.0	38.4 (1.4)	44.4 (3.6)		−11.7 (3.5)	−11.6 (8.4)		26.7 (3.9)	32.8 (7.6)	
	$F_{(1,12)} = 16.762$; $P = 0.001$			Not significant			$F_{(1,12)} = 17.865$; $P = 0.034$		
Speed (m/s)	*Angle at foot contact* Normal	*Angle at foot contact* SCI		*Minimal angle (stance)* Normal	*Minimal angle (stance)* SCI		*Maximal angle (swing)* Normal	*Maximal angle (swing)* SCI	
Knee angles (in degrees)									
0.1	16.2 (10.4)	32.5 (10.5)	* Not significant	7.4 (6.7)	14.1 (15.4)	* Not significant	51.9 (9.6)	55.9 (10)	* $F_{(2.24)} = 2.610$;
0.3	9.8 (7.1)	28.6 (10.2)		7.4 (5)	16.5 (11.3)		55.9 (7.1)	53.6 (9)	$P = 0.094$
0.5	8.9 (5.7)	33.4 (9.8)		7.1 (6)	17.5 (12.2)		60.0 (5.9)	54.1 (9.5)	
1.0	9.0 (4.5)	34.5 (5.5)		8.7 (5.5)	21.0 (11.4)		66.6 (5.3)	57.1 (9.1)	
	$F_{(1,12)} = 17.865$; $P = 0.001$			Not significant			Not significant		
Speed (m/s)	*Angle at foot contact* Normal	*Angle at foot contact* SCI		*Maximal plantarflexion* Normal	*Maximal plantarflexion* SCI		*Dorsiflexion during swing* Normal	*Dorsiflexion during swing* SCI	
Ankle angles (in degrees)									
0.1	−0.9 (4.7)	4.3 (5.6)	* $F_{(2,24)} = 8.759$;	3.2 (11.2)	7.8 (8.8)	* $F_{(2,24)} = 4.906$;	12.8 (3.5)	11.2 (5.8)	* $F_{(2,24)} = 20.928$;
0.3	−3.6 (3.9)	0.1 (4)	$P = 0.001$	1.3 (6.5)	7.0 (7.9)	$P = 0.016$	9.5 (3.4)	9.9 (6.4)	$P = 0.000$
0.5	−3.7 (3.8)	0.9 (6.1)		−2.2 (9.3)	5.0 (7.7)		7.7 (2.4)	9.0 (6.8)	
1.0	−2.2 (2.6)	4.1 (5.2)		−12.2 (11.1)	4.6 (6.7)		3.1 (4.2)	10.4 (0.8)	
	$F_{(1,12)} = 3.802$; $P = 0.075$			Not significant			Not significant		

Normal and SCI group averages of joint angular values at different points of the walking cycle (see text for more details). The numbers in parentheses represent the standard deviation of the mean. Note that there were only three subjects in the SCI group at 1.0 m/s and results from this speed were excluded from the analysis of variance

Table 5 Comparison of angular values between groups at 0.3 m/s, and between 0.3 and 1.0 m/s conditions for the normal group

Specific angle values	*Comparison between groups at 0.3 m/s*	*Normal group, 0.3 versus 1.0 m/s*
Total hip excursion	↑ in SCI ($F_{(1,12)}=8.249$; $P=0.014$)	↑ at 1.0 m/s ($F_{(1,6)}=29.359$; $P=0.002$)
Maximum hip extension	No statistical difference	↑ at 1.0 m/s ($F_{(1,6)}=30.054$; $P=0.002$)
Maximum hip flexion	↑ in SCI ($F_{(1,12)}=6.537$; $P=0.025$)	↑ at 1.0 m/s ($F_{(1,6)}=21.726$; $P=0.003$)
Knee angle at foot contact	↑ in SCI ($F_{(1,12)}$ 15.808; $P=0.002$)	No statistical difference
Knee minimum angle (stance)	↑ in SCI ($F_{(1,12)}=3.413$; $P=0.089$)	No statistical difference
Maximal knee flexion (swing)	No statistical difference	↑ at 1.0 m/s $F_{(1,6)}=54.319$; $P=0.000$)
Ankle angle at foot contact	↑ dorsiflexion in SCI ($F_{(1,12)}=3.171$; $P=0.010$)	↓ dorsiflexion at 1.0 m/s ($F_{(1,6)}=4.778$; $P=0.071$)
Ankle angle at toe-off	No statistical difference	↑ plantarflexion at 1.0 m/s ($F_{(1,6)}=9.458$; $P=0.022$)
Ankle maximal dorsiflexion (swing)	No statistical difference	↓ dorsiflexion at 1.0 m/s ($F_{(1,6)}=16.429$; $P=0.007$)

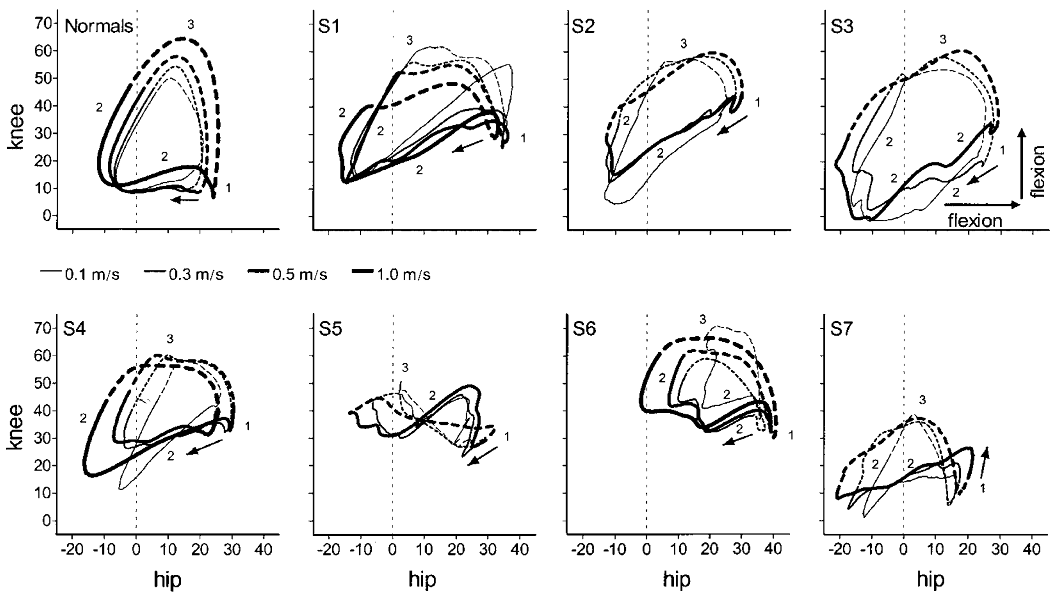

Figure 3 Relation between hip and knee joint angular displacements as a function of walking speed. Cyclographs representing the average hip–knee angular displacement relation for the normal group and for all SCI subjects at 0.1, 0.3, 0.5 and 1.0 m/s. The black solid part of the cyclographs represent the stance and the dashed part represents the swing. The arrows indicate the direction of the movement. See text for explanation of numbered points on the graphs

extension in the early part of stance which is also present in normal subjects, but only at 1.0 m/s. The coupling in flexion in the beginning of swing seen in normal subjects is also present in most SCI subjects, but its timing is often changed, and is either not well defined or happens for a shorter portion of the curve. During swing, the point of inflection of the curve (see point 3, Figure 3) observed in the normal group is not as well defined in most of the SCI subjects and its timing varies among subjects.

For the normal group, the shape of the curves remains almost identical at all speeds, except for the bottom part because of the increased knee flexion at higher speeds. Except for S3, cyclograph curves for SCI subjects are flattened as compared to the normal group, which reflects the greater hip and smaller knee excursions in these subjects. In most SCI subjects, a change in the shape or in the orientation of the cyclographs is observed especially between the 0.1 m/s condition and higher speeds.

The relation between knee and ankle joints is not presented because of idiosyncrasies in the patterns of knee–ankle cyclograms in SCI subjects, reflecting the variability of knee and ankle kinematics described previously.

EMG activity

In general, for both groups, muscle EMG activity increases in amplitude when speed is increased, although

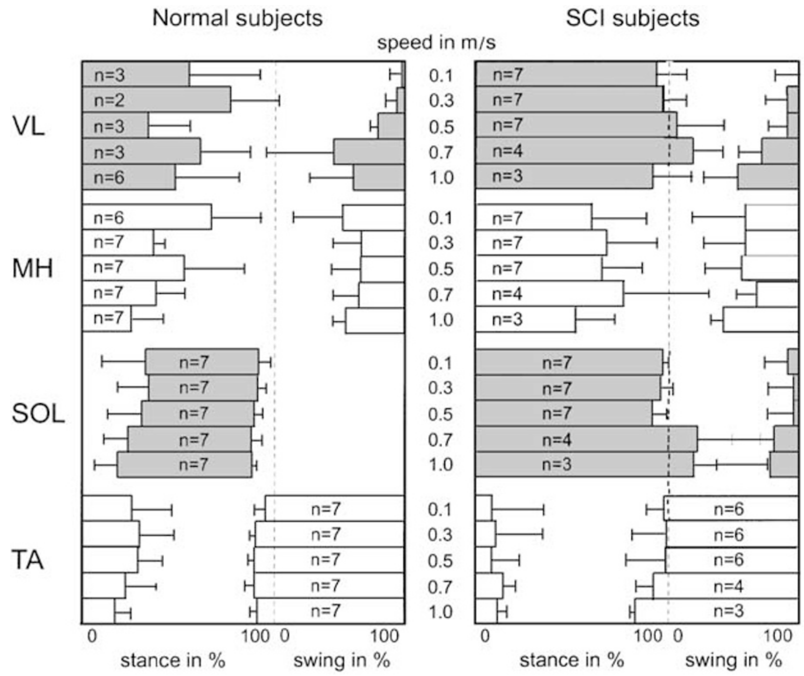

Figure 4 Onset and offset of EMG activity as a function of walking speed. Average onset and offset of EMG activity in normal and SCI groups ($n=7$) at all measured speeds. The error bars represent the standard deviation of the mean. Inside the bars are the number of subjects that recruited the muscle at a given speed

for the SCI group the increase in EMG amplitude is variable. Further, the activation of VL and SOL for the SCI group begins sooner within the walking cycle as speed increases. There are notable differences between the SCI and the normal groups. While the normal group shows low variability of the activation profile among subjects, there is a high variability among SCI subjects. Interestingly, at very low speeds, normal subjects also exhibit a similar variability in the more proximal muscles (VL and MH). For instance, only three of the seven normal subjects activate their VL at speeds lower than 1.0 m/s.

Changes in the timing of EMG activity Figure 4 presents the changes in the timing of the EMG activity of the four muscles as a function of the walking speed. In normal subjects that show EMG activity in VL at lower speeds, the muscle is active from late swing to midstance. VL is activated for all SCI subjects at all of the measured speeds and the duration of VL activity is prolonged for the whole stance duration at all speeds in comparison to normal subjects. As speed increases, the activation of VL starts earlier in swing in both groups. In normal subjects, the MH muscle is activated from midswing to early stance at 1.0 m/s. At lower speeds, MH burst is prolonged to about midstance. As is the case for VL, the duration of MH activity is longer in the SCI group for all speeds, which leads to coactivation of both muscles for most of stance.

The timing of SOL muscle activity in normal subjects changes as a function of the walking speed. There is an earlier recruitment of SOL muscle during stance in normal subjects as the speed increases; however, the timing of the EMG activity offset relative to end of stance does not change. Further, the alternating pattern of activation between TA and SOL is preserved at all speeds. In contrast, the recruitment of SOL in SCI subjects happens well before stance, starting in late swing for all speed conditions and the activation lasts for the whole stance duration. Further, there are periods of coactivation between TA and SOL, especially from late swing to early stance. At speeds from 0.1 to 0.5 m/s, the activation of TA in the SCI group starts later than in the normal group in late stance and ends earlier than in the normal group.

Modulation of the amplitude of EMG activity with increases in speed For the three normal subjects that show EMG activity in VL at lower speeds and for all SCI subjects, the amplitude of VL activity is increased at higher speeds. For MH, the EMG amplitude is also substantially higher in all normal and most SCI subjects at higher speeds. In the normal group, the amplitude of the TA EMG burst associated with the foot placement (from late swing to early stance) increases as well. The amplitude of the second burst associated with ankle dorsiflexion during swing (from late stance to early swing) showed little change throughout the different speeds. Changes in TA varied greatly among SCI

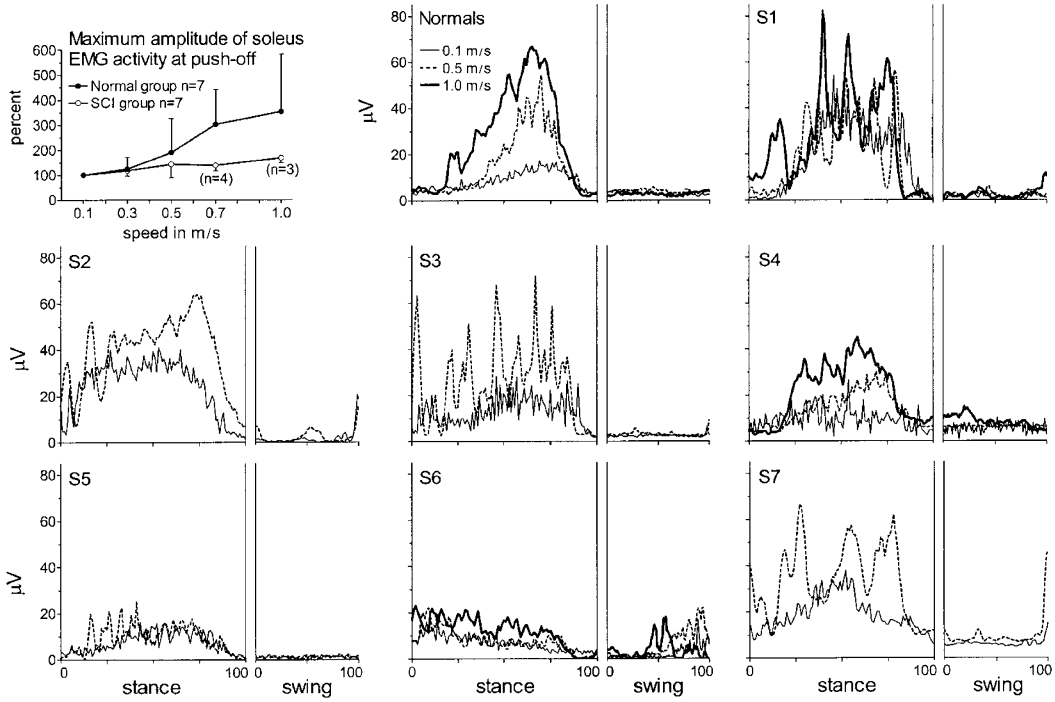

Figure 5 EMG profiles of soleus muscle as a function of walking speed. Average soleus EMG profiles for the normal group along with individual profiles for SCI subjects at 0.1, 0.5 and 1.0 m/s. The top left corner shows the maximal amplitude of soleus activity during push-off as a percent of the maximal activity recorded at 0.1 m/s

subjects and some showed very little change in EMG amplitude.

As a result of its major role for propulsion during walking,[29] the changes in SOL EMG profiles and amplitude are presented in Figure 5. In the normal group, there is a greater amplitude of SOL EMG activity as speed increases, and this increase is associated with push-off. There is also an increase in the amplitude of the SOL activity in five of the SCI subjects (S1, S2, S3, S4 and S7), but the increase was not specific to the push-off (except for S4). Clonus is present in most SCI subjects as they walk faster. In the top left corner of Figure 5, the changes in peak amplitude of SOL during push-off as a function of speed are presented for both groups. The maximum amplitude measured during push-off demonstrate that the SCI group showed a smaller increase than the normal group as a function of an increase in speed. Although the trend seems to be clear, the difference between groups was not statistically significant.

Discussion

It has been shown in other studies that the walking pattern of normal subjects is very adaptable to changes in the mechanical demands of the task. This pattern can adapt easily to different speeds,[15,17] inclines,[30–31] surfaces such as overground versus treadmill[32–35] and uneven surfaces.[36] The main objective of the present study was to measure how SCI subjects adapt to changes in the walking speed. Studies of hindlimb locomotion in spinalized cats[37–39] have shown that adaptation to increasing treadmill speed was possible, but that this adaptation was limited to a maximal speed varying between 0.8 and 1.0 m/s. In the spinal cat, the limited adaptation to increases in speed is presumably because of the absence of descending tracts and achieved with input solely from peripheral influence. Since some descending tracts are partially preserved in incomplete SCI subjects, it was likely that they would have some degree of capacity for adaptation.

In this study, it was demonstrated that the SCI have a limited capacity to adapt to changes in speed. In the normal population, the adaptation to increases in the walking speed is generally obtained by increasing the cadence (stride frequency) and the stride length simultaneously[11–16] and there is a curvilinear relation between the cycle duration and the walking speed in normal subjects.[15–17] This relation was also observed in the present study. In fact, temporal and distance parameters changed in a similar fashion as a function of speed for both groups. However, the average values of the two groups differed significantly. Only two SCI subjects could reduce their cycle duration, by increasing their stride frequency, to the same level as that of normal subjects suggesting that SCI subjects have a limited capacity to modulate their stride frequency because of an altered neural drive, and therefore, favor an increase in their stride length to adapt to higher treadmill speeds. This point will be further discussed later.

The results of EMG activity for the normal subjects are in concordance with those obtained in other studies[17,18,22,40] and have extended the knowledge about the timing of activation for the four studied muscles to speeds as low as 0.1 m/s. For most of the SCI subjects in this study, neural deficits resulted in changes in the muscle activation profiles, particularly in extensor muscles (SOL and VL). These changes in EMG activity along with the changes in knee kinematics indicate an inability to recruit normally the muscles acting on that joint, and could also be part of a strategy to enhance their support and balance during stance. Indeed, it has been shown in SCI[41] and hemiplegic subjects[42] that providing weight support and balance with the use of a harness reduces significantly the amount of knee flexion at foot contact and for most of stance.

In both groups, the hip total excursion increased as speed increased. Similar results have previously been reported for normal subjects.[18–21] The SCI group had larger hip excursion values than the normal group at each of the speeds, this difference was larger at a treadmill speed of 0.5 m/s compared with 0.1 m/s. Further, the values reached for hip excursion at 0.3 m/s in SCI subjects were almost the same as those reached by normal subjects at 1.0 m/s. Since the total hip excursion and the knee angle at foot contact directly influence the stride length, the greater than normal hip excursion combined with a flexed knee at foot contact suggest that SCI subjects reached their maximal stride length at lower speeds than normal subjects.

In SCI subjects, the ankle angular values at push-off changed towards plantarflexion as a function of the increasing speed, and remained in dorsiflexion even at 0.5 m/s, which was due, in large part, to the pronounced knee flexion. This ankle position also suggests a lack of propulsion at the end of stance. Plantarflexor muscles normally play a major role for propulsion during walking.[26] Studies in rats[43] and in cats[44] have shown that increases in speed and incline are associated with increased output of selective extensor muscles. Similarly, in normal subjects, an augmentation of quadriceps and triceps surae EMG activity was associated with an increase of speed or upward tilt during walking.[23] In the present study, SCI subjects showed differences in the timing and amplitude of SOL activity. Earlier onset (early stance or late swing) and prolongation of SOL EMG activity was present at all speeds. The amplitude of SOL activity increased with speed, but this increase was not specific to the part of the burst normally associated with push-off. Instead, in most cases, the whole burst was increased in amplitude, even at the beginning of stance. The increase of the early part of the burst is likely because of an increase in stretch-induced activation of the muscle as revealed by the presence of clonus, and was described previously in subjects with spastic paresis.[2–4,45–47] This abnormal activation has been associated, in part, with defective gating of Ia afferents[5,48,49] and with changes in the mechanical properties of muscles and tendons.[6,46,50] These changes in SOL, paralleled by similar changes in the VL activity (increased amplitude and prolonged activity), are also likely related to the flexed knee position adopted by these subjects at foot contact and throughout most of stance. The lack of an increase in EMG amplitude of SOL muscle that is specific to push-off and the lack of plantarflexion at the end of stance strongly suggest an inability to generate the propelling force that would be necessary to increase the stride frequency and stride length in order to reach higher walking speeds.

In the swing phase, the normal subjects augmented the maximal knee flexion as the speed increased, whereas the SCI subjects showed very little difference or sometimes even a decrease. This could indicate an abnormal control of the knee flexor muscles, but is very likely due, in part, to a reduced flexor moment at the hip joint. In normals, the knee flexion at the beginning of swing is linked to the hip flexor moment.[29,51] In this study, the greater knee flexion observed in normal subjects during swing at 1.0 m/s brings the center of mass of the whole lower limb closer to the hip joint, which is mechanically more efficient for the increase in hip joint velocity that is needed during swing at higher walking speeds. SCI subjects managed to increase their walking speed, but obviously used different strategies to swing the limb through. Results from the vertical displacement of the GT suggest that some of the subjects used a hip hiking strategy to adapt to increases in speed and it is likely that SCI subjects used the handlebars. Hip hiking strategy was also reported for adapting to incline walking.[7] However, since the handlebars were not instrumented, their contribution in SCI subjects' adaptation to higher speeds cannot be assessed. Hip circumduction is another possible strategy, but it was not measured. Adaptations using pelvis and trunk movements are also very likely and have been reported in hemiplegic subjects[52] as well as in normal subjects during grade walking.[53] Nevertheless, the inability to increase the knee flexion during swing in SCI subjects also influences their capacity to increase the stride frequency.

The coordination of the hip and knee joints during the walking cycle is a major factor determining the efficiency of walking. For instance, the amount of hip and knee extension at the end of stance together with the amount of hip flexion and knee extension at the end of swing greatly influences the stride length. Similarly, the combined movements of hip and knee flexion in late stance and early swing play an important role in the toe clearance during swing. Cyclograph curves revealed that there was an alteration in the relation between hip and knee joints in SCI subjects. These changes included modification of the timing of hip and knee coupling during the walking cycle, and were present even at lower speeds. They were even more substantial when the walking speed increased and clearly reflected a dyscoordination of lower limb movements. This altered intralimb coordination affects stride frequency and length and therefore contributes to limitations in maximal walking speed in SCI subjects. Overall, SCI subjects seem to rely more on hip excursion rather than on the knee excursion at any of the speeds.

The differences obtained for the normal group with respect to changes in the walking speed are in concordance with the findings in the literature, showing important differences in the temporal and distance parameters,[15,17] in the angular excursions[17,20–21] and in the activity of the muscles.[17,22,40] As a result of such differences, and because the SCI population generally has a significantly reduced walking speed as compared to the normal population, the comparison with the two populations while walking at a matched speed has many advantages.

Comparing the gait of SCI subjects to that of normal subjects walking at 'comfortable' or preferred speed does not allow the differentiation of features that are a consequence of their reduced walking speed rather than a direct consequence of their injury. Results from this study show that the characteristics observed in the SCI subjects walking pattern are related both to their reduced walking speed and to their neural deficits.

Conclusions and functional implications

The walking deficits in SCI subjects are shown by comparing their walking pattern to that of normal subjects. To measure the deficits that are linked to the injury and not just a consequence of the reduced walking speed, the speed of normal subjects was matched to that of SCI subjects in order to compare tasks that are mechanically similar. For the rehabilitation specialist, the characterization of the deficits observed in SCI subjects is important, particularly for developing new rehabilitation approaches. It is also important to have a reference regarding normal walking patterns at reduced speed.

The results presented herein set the basis for a better understanding of the ability of SCI subjects to adapt to different walking speeds on a treadmill. It was shown that SCI subjects have a limited capacity to adapt to speeds higher than their preferred or 'comfortable' one. The limited maximal walking speed observed in SCI subjects seems to be a consequence of having reached their limit in increasing stride length and not having the ability to increase stride frequency further. A question remains as to which of the two factors plays a greater role in setting the upper limit on walking speed. The companion paper will address this question.

In future studies, it would also be important to determine if the adaptation mechanisms observed in this study for treadmill walking are similar for overground walking.

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