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Original Article

Treadmill walking in incomplete spinal-cord-injured subjects: 2. Factors limiting the maximal speed

A Pépin[1], M Ladouceur[2] and H Barbeau*,[3,4]

[1]Département de Kinanthropologie, Université du Québec à Montréal, Montréal, Québec, Canada; [2]Department of Physical Education and Kinesiology, Brock University, St Catharines, Ontario, Canada; [3]School of Physical and Occupational Therapy, McGill University, Montréal, Québec, Canada; [4]Jewish Rehabilitation Research Center, Laval, Quebec, Canada

Study design/methods: Five SCI subjects referred to the laboratory and a convenience sample of five normal volunteer individuals was selected. Stride length and frequency were measured at different walking speeds under three different conditions: preferred, highest possible and lowest possible stepping frequencies.
Objective: To determine which factors are limiting the maximal walking speed in spinal-cord-injured (SCI) individuals.
Setting: University-Based Human Gait Laboratory, Montreal, Canada.
Results: It is shown that maximal stride frequency was the predominant limiting factor of the maximal treadmill-walking speed in SCI subjects. These results were explained in the light of the forced hybrid mass–spring pendulum model. At all speeds, SCI subjects spent longer time in stance, swing and double support phases. The relative time spent in single support is greater at higher walking speed and the difficulty to reduce double support time is a limiting factor.
Conclusions: A better understanding of the factors limiting the maximal speed in SCI subjects should help developing rehabilitation interventions oriented towards increasing the control and the capacity of walking. Rehabilitation strategies should put the emphasis on improving the capacity to produce rapid alternate rhythmical stepping movements of the lower limbs.
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Keywords: spinal cord injury; gait; speed; treadmill walking

Introduction

Walking at different speeds is achieved by simultaneously varying the stride length and frequency.[1–3] For instance, in order to walk faster, stride length and stride frequency are increased simultaneously. Similar to able-bodied subjects, spinal-cord-injured (SCI) subjects with an incomplete motor function loss can also adapt to speeds other than their ‘comfortable’ or preferred one by lowering or increasing the length and frequency of their stride. In a previous study (see companion paper),[4] it was shown that changes in preferred stride frequency (1/cycle duration) as a function of treadmill speed occurred in a similar fashion in SCI and normal subjects, but over a narrower range of speeds. The major differences resided in a lower stride frequency (longer cycle duration) and a longer stride length in SCI subjects for a same walking speed. It was observed that SCI subjects tend to reach their maximal stride length at a lower speed than normal subjects do. This result shows that SCI subjects adapt to higher walking speeds by favoring an increase in stride length rather than an increase in stride frequency.

The level of impairment following incomplete SCI varies greatly among individuals and can be represented as a spectrum of deficits, as presented by the ASIA scale.[5] Muscle weakness is one of the major deficits observed in SCI subjects[6–8] and in most SCI subjects, the preferred as well as the maximal walking speed is usually greatly reduced. Similarly, it was reported that in stroke patients the strength of lower limb is related to gait velocity and cadence,[9] and that the maximal gait speed is correlated with hip flexor and plantarflexor strength in adult subjects with stroke.[10]

Walking speed being the product of the stride frequency and the stride length, one or both of these two variables can be a limiting factor of maximal speed. One important problem for those subjects in reaching

*Correspondence: H Barbeau, School of Physical and Occupational Therapy, McGill University, 3630 Rue Drummond, Montréal, Quebec, Canada H3G 1Y5

higher walking speeds seems to be their inability to increase their stride frequency.[4] In other words, the maximal stride frequency that SCI subjects can achieve during treadmill walking is greatly affected by their neural deficits. If this hypothesis is true, maximal stride frequency in SCI subjects will be inferior to that of able-bodied subjects. The main objective of this study was to assess whether one of the two parameters determining walking speed – stride frequency and stride length – has a greater responsibility in limiting the maximal speed in SCI subjects. In normal subjects, it has been shown that although both parameters are usually modified when changing the walking speed, it is possible to modify only one of the two parameters over a wide range of speeds.[11] Indeed, it was shown that while walking at speeds ranging from 1.0 to 3.0 m/s, the highest possible stride frequency remained more or less constant at about 2.5 strides/s, while the stride length increased linearly. On the other hand, imposing the lowest possible stride frequency at a certain speed implies a maximal utilization of stride length for that particular speed.[11] The measurements of the highest and lowest possible frequencies over a range of speeds will allow to determine if one of the two parameters (maximal stride frequency and maximal stride length) has a greater role to play in the limitation of SCI subjects' walking speed.

Methods

Subjects

Two groups (five SCI and five normal subjects) participated in the study. All SCI subjects were coming to the laboratory for periodic evaluation sessions and were accustomed to treadmill walking. Relevant characteristics of SCI and normal subjects are found in Table 1. Ethical approval was obtained and subjects gave informed consent prior to participation in the study.

Procedures

SCI subjects walked on a treadmill at speeds ranging from 0.1 m/s to maximal speed, whereas speeds for normal subjects ranged from 0.1 to 1.5 m/s. All subjects were asked to hold bars located on both sides of the treadmill. The number of walking speeds that were recorded for each SCI subject depended mostly on the capacity and endurance of the participant being evaluated. However, speeds of 0.1, 0.3, 0.5, 0.7 m/s as well as maximal speed were recorded for all SCI subjects. All SCI subjects could walk up to 0.7 m/s and three of them were able to walk at speeds beyond 1.0 m/s (see Table 1). Three different walking conditions were measured in the following order for each walking speed: preferred, highest possible and lowest possible stride frequencies. Measurements were recorded from the lowest to the highest speed. In order to minimize the effect of fatigue in SCI subjects, sufficient time was allowed in between speeds for the participants to recover. The highest possible frequency allowed the measurement of how often the subject is capable of repeating the task of stepping at a given speed. The lowest possible frequency allowed the measurement of the maximal stride length that a subject can achieve at a given speed. All conditions were filmed by a video camera. The signal from an SMPTE time code generator was recorded on the image. Foot contact and foot off were determined visually during playback with a

Table 1 Summary of SCI and normal subjects' profiles

(A)

SCI subjects	*Age*	*Level of the injury*	*Time since the injury (years)*	*Walking aids for daily activities*	*Maximal overground walking speed*[a,b]	*Preferred overground walking speed*[a,c]	*Maximal treadmill speed*[a]
S1	37	T8	8	Two elbow crutches	1.71	1.18	1.40
S2	41	T10	3	Two canes	1.47	1.08	1.10
S3	50	C5–C6	4.5	Two canes	0.70	0.59	0.70
S4	28	C6	11	Two elbow crutches	0.50	0.39	0.70
S5	35	T12–L1	13.5	Two canes	1.26	1.00	1.30

(B)

SCI group	*Age*	*Sex*	*Height (cm)*	*Weight (kg)*	*Normal group*	*Age*	*Sex*	*Height (cm)*	*Weight (kg)*
S1	37	M	183	77.3	S1	40	M	180	79.5
S2	41	M	173	81.8	S2	38	M	178	76.4
S3	50	F	161	68.2	S3	33	F	168	56.8
S4	28	M	173	72.7	S4	24	M	170	76.4
S5	35	M	168	61.4	S5	45	M	170	90.0

[a]Speed is in m/s; [b]The maximal overground walking speed was measured on a 15 m walkway; [c]The preferred overground walking speed was the average obtained during walking for at least 5 min on a 15 m walkway

minimal resolution of 16.7 ms (60 fields/s). In all, 10 consecutive walking cycles were used to calculate average stride frequency, as well as stance, swing and double support durations for all conditions. Each of the video records was monitored carefully to ensure that the subject was not moving forward or backward with respect to the treadmill. In conjunction with the constant speed of the treadmill, it made possible to calculate stride length from stride frequency. One SCI subject was evaluated twice, 5 weeks apart. This subject was part of a functional electrical stimulation (FES)-assisted walking training program (for more details on stimulation and training protocol, see Ladouceur and Barbeau[12]). Since the maximal walking speed of this subject had increased following the training program, it was of interest to measure whether the maximal stride frequency and/or maximal stride length had changed.

Analyses
Comparisons between groups and among speed conditions were made for all stride frequency conditions using an analysis of variance (ANOVA) for repeated measures with a grouping factor. The comparisons were made only for 0.1, 0.3, 0.5 and 0.7 m/s, representing the range of speed that all SCI subjects could attain. The parameters used for comparison were stride frequency, stride length, stance, swing and double support durations. An α value of 0.10 was used to determine significance of results. For the SCI group, the Pearson correlation coefficients were calculated between highest stride frequency and maximal stride length, between highest (maximal) stride frequency and maximal speed, and between maximal stride length and maximal speed. A multiple regression (stepwise) analysis was used to determine which of the two variables, maximal frequency or maximal stride length, has a greater contribution in limiting the maximal walking speed in SCI subjects. The software used for these analyses was Systat for Windows (v5.02) from Systat, Inc.

Results

The range of available stride frequency is reduced in SCI subjects
Results for frequencies and stride length for the normal group and SCI subjects are reported in Figures 1 and 2. It is important to mention that walking speeds of 1.0 and 1.5 m/s correspond approximately to the range of the comfortable speed observed in normal walking. In contrast, the maximal speed reached by SCI subjects in the present study ranged from 0.7 to 1.4 m/s. A first striking point in Figure 1 is that the range of possible frequencies (lowest to highest frequency) diminishes with increases in speed, both in normal and SCI subjects. This has been described elsewhere[11] in normal subjects at a higher speed range. It is noteworthy that this range is very small in three of the SCI subjects (S2, S3 and S4) and, as SCI subjects reach their maximal

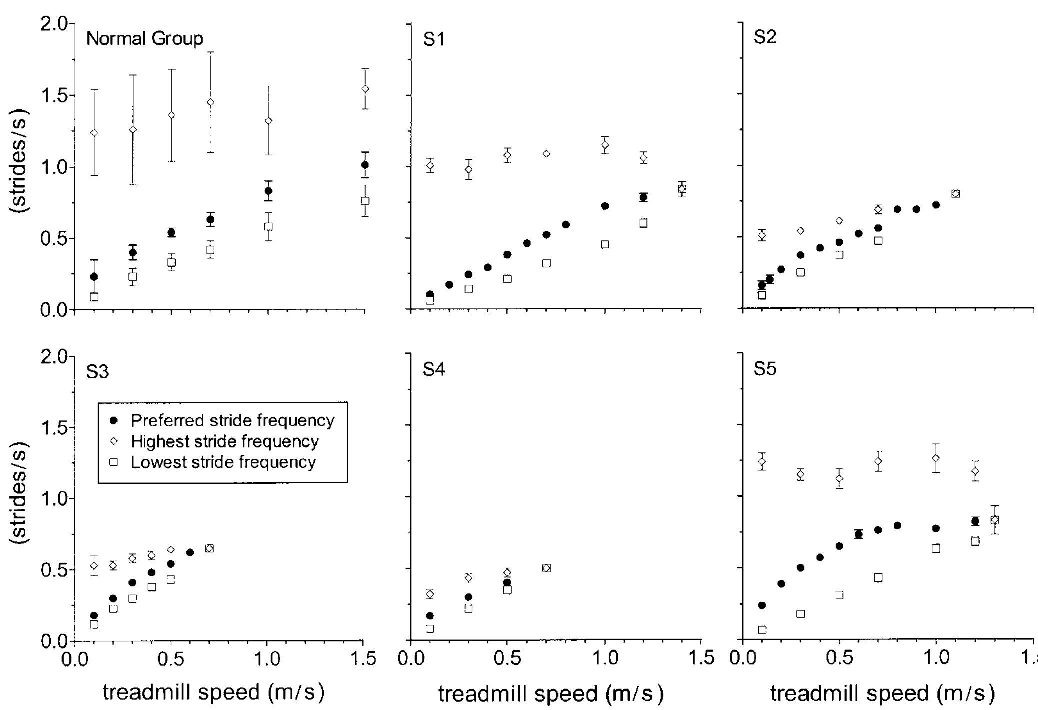

Figure 1 Preferred, highest and lowest stride frequencies as a function of walking speed. The preferred, the highest and the lowest stride frequencies for the normal group (upper left corner) and five SCI subjects as a function of walking speeds. The error bar represents the standard deviation. Note that the within subject variability was very small, which explains the absence of error bars in some of the conditions for SCI subjects

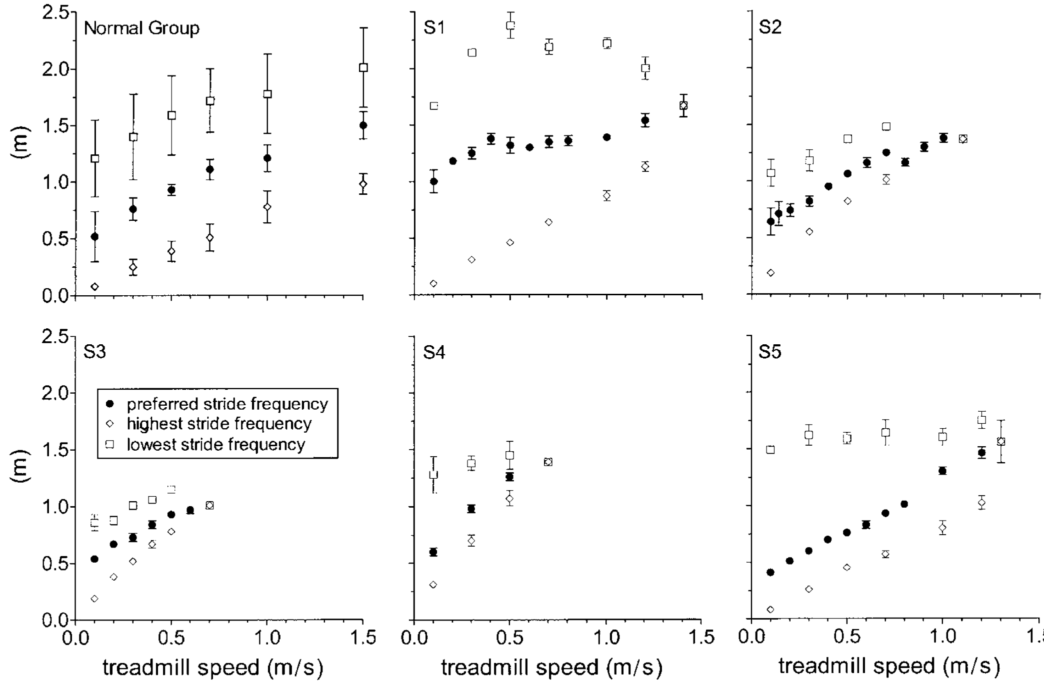

Figure 2 Stride length as a function of walking speed at preferred, highest and lowest stride frequencies. Stride length values for the preferred, the highest and the lowest stride frequencies conditions for the normal group (upper left corner) and five SCI subjects at different walking speeds. The error bar represents the standard deviation. Note that the within subject variability was very small, which explains the absence of error bars in some of the conditions for SCI subjects

speed, the modulation of frequency was no longer possible. In some cases, the highest frequency was smaller at maximal speed than it was at lower speeds (S1 and S5). In Figure 1, it is shown that normal subjects could reach a maximal stride frequency that varied from 1.24 Hz (±0.30 SD) at 0.1 m/s to 1.54 Hz (±0.37 SD) at 1.5 m/s. Except for S5, SCI subjects had lower values than those of the normal group. Further, three of the SCI subjects (S2, S3 and S4) showed only a small increase between the highest and the preferred frequency. For the lowest frequency conditions, SCI subjects had values that were similar to that of the normal group.

Figure 2 presents the stride length values for the three walking conditions for the normal group and all SCI subjects. The range of possible stride length as a function of speed remained the same in normal subjects. However, this range diminished with increases in speed in all SCI subjects. There was no frequency constraint for the maximal speed in SCI subjects. However, it is assumed that the frequency and stride length used by the subject at maximal speed were the only ones possible for that speed. Therefore, the values for stride frequency (Figure 1) and stride length (Figure 2) measured at maximal speed were considered as the end point of the three curves. In normal subjects, for all three conditions, the maximal stride length increased along with the walking speed. In SCI subjects, the stride length at preferred and highest frequency increased with speed. For the lowest frequency condition (maximal stride length), there were increases with speed in all subjects, but these increases were very minimal in two subjects (S4 and S5). One subject (S1) showed an increase in maximal stride length up to 0.5 m/s followed by a decrease at higher speeds. Except for S1, the range of maximal stride length in SCI subjects was similar or slightly lower than that of the normal group. A noteworthy observation is that, contrarily to normal subjects, SCI subjects had either a plateau or a decrease in their maximal stride length as they reached their maximal speed.

Maximal walking speed is limited by maximal frequency in SCI subjects

Figure 3 presents the means for the maximal stride frequency and stride length at all measured speeds for the normal group, and for speeds of 0.1–0.7 m/s for the SCI group (highest speed at which all five SCI subjects could walk). There was a significant increase of maximal stride frequency and maximal stride length for both groups as speed increased (Figure 3a and b). However, SCI subjects had a significantly lower maximal stride frequency than that of the normal group at any of the walking speeds. In fact, the maximal frequency in SCI subjects was closer to the range for the preferred frequency in normal subjects (see Figure 1). In addition, maximal frequency increased as a function of speed in both groups. As shown in Figure 3b, both groups had an increase in their maximal stride length as speed increased.

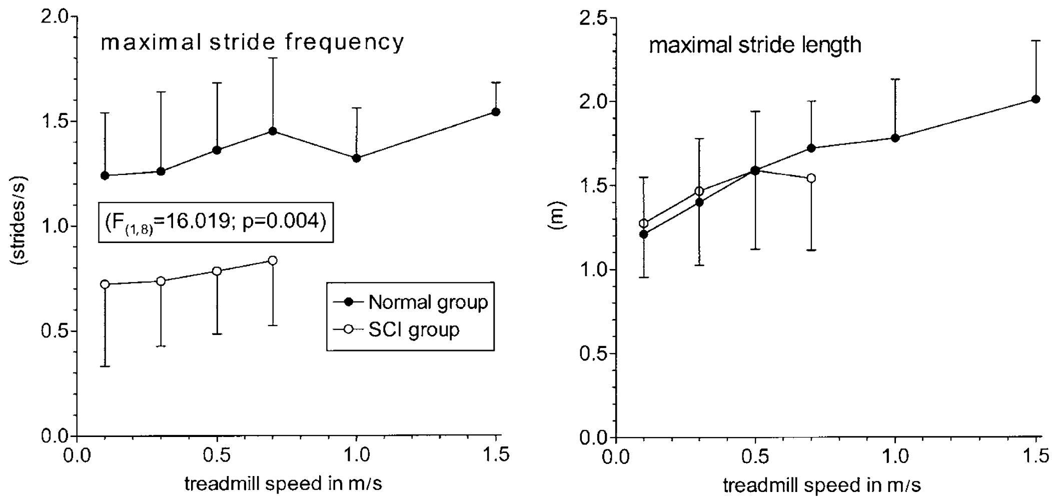

Figure 3 Maximal stride frequency and stride length as a function of walking speed. Mean values of maximal stride frequency (left panel) and maximal stride length (right panel) as a function of treadmill speed for normal and SCI groups. The error bar represents the standard deviation. The maximal stride frequency was greater for normal subjects ($F_{(1,8)}=16.019$; $P=0.004$). The maximal frequency ($F_{(3,24)}=19.184$; $P=0.000$) and the maximal stride length ($F_{(3,24)}=18.632$; $P=0.000$) increased speed in both groups as a function of speed

In SCI subjects, the two variables (maximal stride frequency and maximal stride length) were not significantly correlated. The correlation between maximal frequency and maximal speed, and between maximal stride length and maximal speed was high ($r=0.885$; $P=0.023$ and $r=0.860$; $P=0.031$, respectively). The determination of maximal speed appears to be bifactorial. A multiple regression (stepwise) analysis was conducted to determine the weight of each variable in the determination of maximal speed. As expected, the R^2 value for the regression was high ($R^2=0.99$). The b coefficients determined by the stepwise multiple regression for the maximal frequency and the maximal stride length were, respectively, 1.183 ($P=0.002$) and 0.633 ($P=0.001$). In other words, the maximal frequency is near twice more important than the maximal stride length in determining the maximal speed in SCI subjects.

All temporal components of the walking cycle are impaired in SCI subjects
The difference in the maximal stride frequency variable between SCI and normal subjects was just presented above. Since the stride frequency is the inverse of the cycle duration (F = 1/cycle duration), at least one of the temporal components of the walking cycle (stance, swing and double support) should also be affected. In Figure 4, values for the stance, swing, and double support are presented for the three frequency conditions from 0.1 to 0.7 m/s. Interestingly, for the highest frequency condition, stance across speeds was on average 2.0 ($\pm$0.22 SD) times longer for the SCI group than for the normal group, whereas double support duration was on average 2.2 ($\pm$0.46 SD) times longer. The swing duration was on average 1.7 ($\pm$0.21 SD) times longer for the SCI group.

Comparison between two evaluations in one SCI subject
Figure 5 illustrates the results for the stride frequency in S1 for two evaluations separated by approximately 5 weeks. This subject was part of a FES-assisted walking training program. On the second evaluation, his maximal walking speed had increased from 1.4 to 1.7 m/s, an increase of 21.4%. Interestingly, the subject's preferred and maximal stride frequency substantially increased across all speeds on the second evaluation, whereas the lowest stride frequency showed minimal change. The increases in maximal stride frequency for all of the measured speeds observed in S1's second evaluation were accompanied by average decreases of 14.7% ($\pm$5.3 SD) in stance duration, and 23.5% ($\pm$11.0 SD) in double support, whereas swing duration did not change (−0.3% ($\pm$10.5 SD)).

Discussion

Higher walking speed is concomitant with increases in both stride frequency and stride length. In SCI subjects both parameters were well correlated with maximal walking speed, which was to be expected. It was also shown that maximal stride frequency was a greater limiting factor than maximal stride length upon maximal speed in SCI subjects. However, SCI subjects seemed to reach their limit in stride length as they reached maximal speed (see Figure 2). Therefore, the reduced maximal speed in SCI subjects cannot be attributed solely to their reduced stride frequency. Still, the major locomotor deficit resulting from SCI, at least for the subjects in the present study, is the incapability of producing high-frequency alternate rhythmical movements of the lower limbs.

Part of the limitation that SCI subjects have in increasing their stride frequency and, to a lesser extent

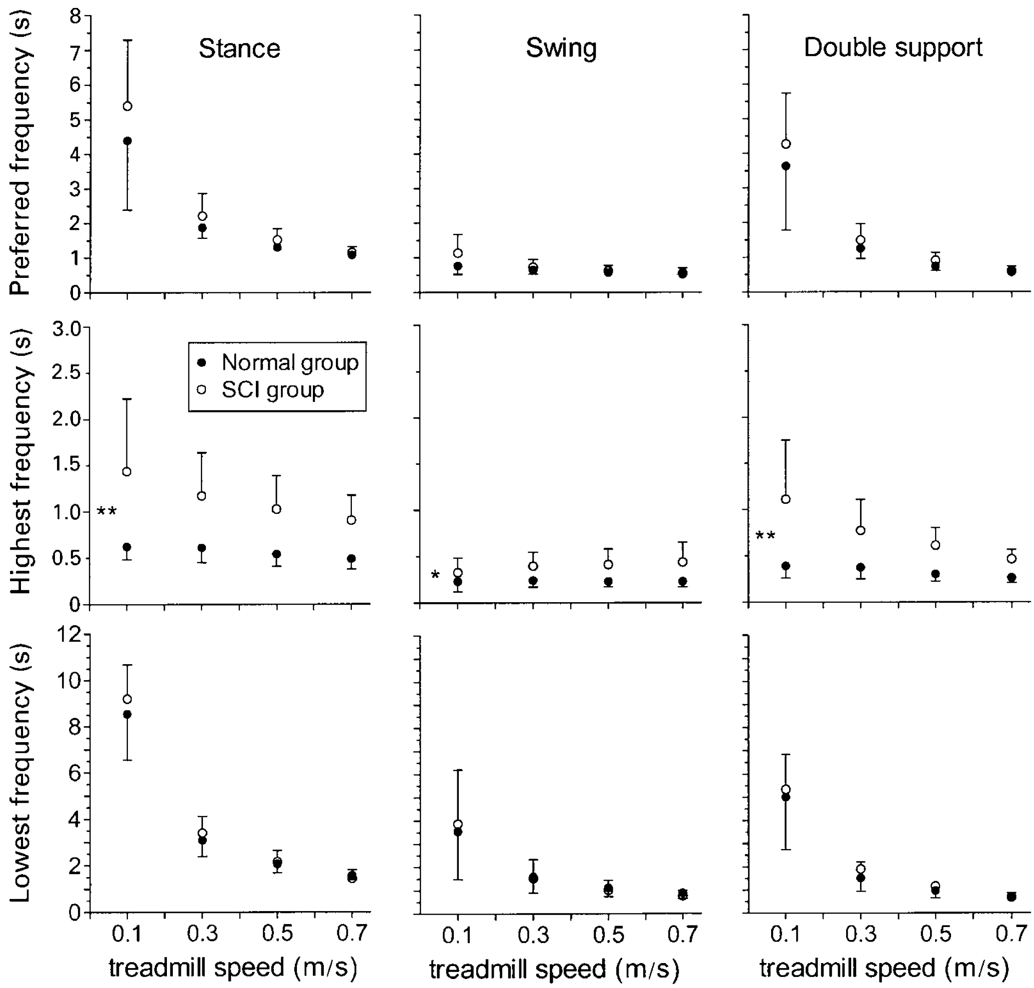

Figure 4 Temporal parameters as a function of walking speed at preferred, highest and lowest stride frequencies. Mean values for stance, swing and double support durations as a function of walking speed for the normal and SCI groups at preferred, highest and lowest stride frequencies. The error bar represents the standard deviation. Statistically significant differences between groups across speeds are represented by asterisks (*for $P<0.10$; **for $P<0.05$)

their stride length, could be explained in the light of a walking model based on limb oscillators theory. A hybrid model including inertial (pendular) and elastic (spring) components has been presented by Holt.[13] During walking, energy is conserved mostly by the pendular actions but also by the spring action of the body. In normal walking, the transfer of potential to kinetic energy of the body center of mass is about 68%.[14] This result shows the loss of energy in the transfer from one form to another (potential to kinetic and *vice versa*). As a result of the inefficient energy transfers from one limb oscillator to the next and the dampening effect of the soft tissues, the oscillations will damp out unless energy is injected into the system by the muscles. The forced hybrid mass–spring pendulum model includes one active and two conservative forces acting across the gait cycle to counteract these dissipative (damping) losses. The active force is a periodic forcing function from the muscles, whereas the two conservative forces are those coming from the body's inertia under the effect of gravity, and those resulting from the spring energy return of muscles and soft tissues.[13] In normal walking, about two-thirds of the injected energy comes from the triceps surae muscles during push-off.[15] Following SCI, there is often an increased stiffness of the lower limb joints because of changes in the mechanical properties of muscles and tendons[6,16–18] as well as coactivation of antagonist muscles. Further, the activity profile of the triceps surae muscles is often flattened, the amplitude reduced, and peak activity during push-off is commonly absent.[4,19] Similar muscle weakness has been shown in hemiparesis where, at maximal speed, the plantarflexor moment only reached 50% of that obtained in healthy subjects at self-selected speed, even though muscles were activated maximally.[20] A combination of the factors mentioned above could explain the reduced maximal stride frequency and maximal stride length in SCI subjects since forcing is linearly related to amplitude (stride length), and stiffness is linearly related to stride frequency.

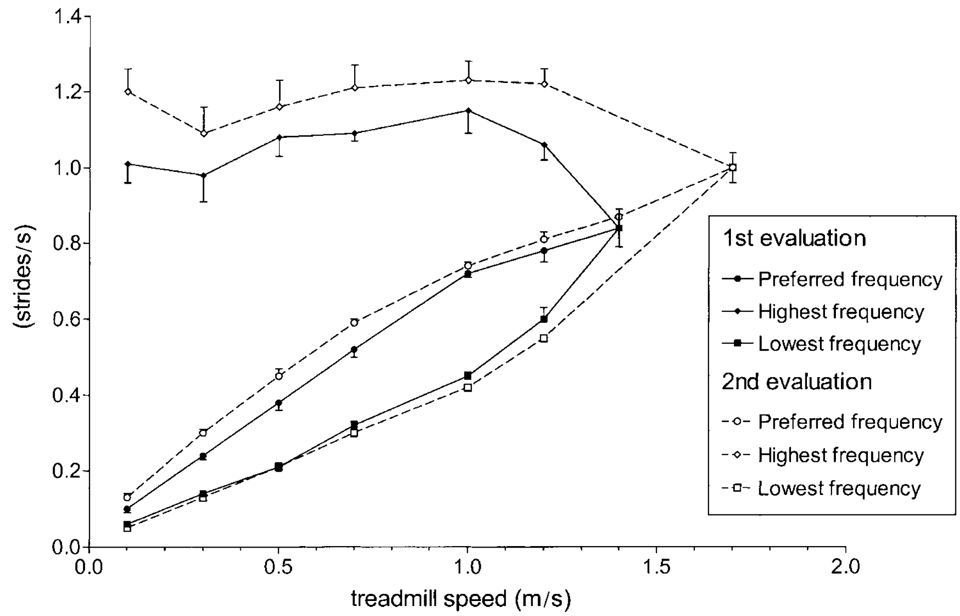

Figure 5 Preferred, highest and lowest stride frequencies as a function of walking speed – training effect of FES-assisted walking as an active orthotic device in one SCI subject. Comparison of preferred, highest and lowest stride frequency at different treadmill speeds, before and after FES-assisted walking intervention in one SCI subject. Note that the maximal speed for the second evaluation increased from 1.4 to 1.7 m/s. The error bar represents the standard deviation

Overall, the reduced maximal stride frequency in SCI subjects was reflected by longer stance, swing and double support durations than those measured in the normal group for the same walking conditions. Interestingly, it has been shown that in hemiplegic subjects, double support duration on the affected side was one of the good predictors of overground gait speed, whereas the percent of stance phase was one of the good predictors on the unaffected side.[21] The longer time that SCI subjects have to stay in double support in order to perform the task could also be linked, in part, to an altered control of balance. However, the longer double support can also be attributed to an incapacity to reduce swing duration. It has been shown that electrical stimulation of the weak plantarflexor muscles during walking could improve forward and upward propulsion of the swinging leg and reduce swing duration.[22–24]

In both groups, the maximal stride frequency and the maximal stride length both increased as the speed increased. This was reported before in normal subjects at higher range of speeds.[11,25] The increase in lowest frequency values as speed increased was accompanied by a decrease in stance, swing and double support durations (see Figure 4). Nilsson *et al*[2] and Nilsson and Thorstenssion[11] argued that in normal subjects, the increase in stride length implied that the lower limb had to be brought forward at a higher velocity, meaning that a higher torque had to be generated by the hip flexor muscles, the moment of inertia being assumed to remain approximately constant. It was concluded that there was a need to increase the neuromotor output during the swing phase for increasing the stride length at higher speeds. Therefore, the limitation of stride length in some of the SCI subjects could be accounted, at least in part, by a deficiency in producing enough flexor moment at the hip. Although, this view is not really in agreement with the model presented above, it is not necessarily incompatible. However, the diminished activity of plantarflexor muscles observed in SCI subjects has more likely a more important role to play in the limitation of maximal stride length as they reached their maximal speed. In a study on ground reaction forces,[26] it was reported that the first and second vertical peak forces did not change with an increase or a decrease of the stride length by 10% of the leg length. However, both braking and propulsive forces were affected by the 10% change, the increase in stride length being accompanied by an increase in the propulsive force (push-off). Hence, the lack or the diminution of the propulsive forces in SCI subjects can be associated with the smaller maximal stride length. Another non-negligible limitation could be mechanical, such as an altered range of motion that is often observed in SCI subjects.

The increase in the maximal speed of S1 following the experimental FES-assisted walking training program was accompanied by a substantial increase in maximal stride frequency over the whole range of speeds, whereas only a small decrease for the lowest frequency condition was observed. This suggests that maximal stride length was not the most important limiting factor for the maximal treadmill speed in this SCI subject. Even though the observation of the changes in the measured parameters was made in only one participant, it is reasonable to hypothesize that improvement in maximal walking speed in SCI subjects following a therapeutic intervention (in this case, FES-assisted walking) could be principally because of an improvement in maximal stride frequency.[27] The stride frequency observed across the whole range of speeds for the preferred frequency condition was also increased for the second evaluation. In other words, when there was no frequency restriction imposed for any given speed, the stride frequency chosen freely by S1 for the second evaluation was always higher than the stride frequency used for the first evaluation. Why is the preferred frequency reset to higher values when the maximal stride frequency and the maximal speed were increased? In normal subjects, it has been shown that the preferred stride frequency, within a certain range of walking speeds, is strongly linked to the lowest metabolic cost for executing the task,[28–31] and that the natural stride frequency during walking could be predicted by the resonant frequency of a 'force-driven harmonic oscillator'.[30,32] The natural frequency was found to be the frequency at which the metabolic cost was the lowest, as determined by oxygen consumption ($\dot{V}O_2$). It can be hypothesized that the increase in the preferred stride frequency observed in S1 for the second evaluation is a sign that, following an improvement in its capacity, the system has reset itself to new values in terms of stride frequency and stride length. The forced hybrid mass–spring pendulum model states that the natural stride frequencies and stride lengths measured at any speed are emergent gait parameters arising as a consequence of a linear dynamical system operating in the resonant frequency mode.[13,32–34] The system always tends to readjust the stiffness in function of the forcing that is used so that it remains in its resonance mode. It is possible to infer that the changes in muscle activation following training with FES were likely because of changes in the forcing and in the stiffness of the system. It has been shown that passive stiffness and reflex stiffness of the ankle were decreased following a FES-assisted walking training protocol.[35] In other words, tendons, joint capsule and ligaments became more compliant and the stretch reflex reduced. However, more importantly, the intrinsic stiffness of triceps surae (active muscle stiffness) was increased.

Although it was not measured specifically in this study, it can be established that SCI subjects generally have muscle weakness. The forced hybrid mass–spring pendulum model predicts that the weaker muscles will generate less propulsive force reducing the magnitude of the stride length. The weaker muscles will also alter the total stiffness reducing also the frequency of the lower limb acting as a pendulum. According to the model, in order to produce a higher frequency of the 'lower limb pendulum', the stiffness must be increased to store and release energy in a spring-like fashion. The increased stiffness must be active or in other words, must be produced muscularly, because the passive elastic forces at the joint level are not sufficient. In SCI subjects, the exaggerated flexion of hip and knee joints at foot contact and of the knee joint during most of stance[4] lead to a mechanical disadvantage for increasing the stiffness of lower limbs. Therefore, the activity of extensor muscles during early stance would be mostly to counteract the effects of gravity bringing the lower limbs further in flexion rather than increasing the stiffness of lower limb joints. This interpretation suggests a deficit in the mechanisms by which potential energy is stored and transferred quickly into kinetic energy to keep the motion of the pendulum going, which will limit the intersegmental transfer of energy.[36] On the other hand, it is important to keep in mind that the limitation of stride length in SCI subjects is not only because of muscle weakness and can be linked to a reduced range of motion at the joints. A reduced hip range of motion and a flexed position of the knee will limit the stride length because the foot will not reach as far forward at foot contact.

The results obtained in this study can be generalized only for treadmill walking and for SCI subjects who are good functional walkers. It is understood that the maximal stride length that SCI subjects could attain was likely influenced by the passive nature of the movement. The motion of the treadmill can bring the lower limb in extension just by following the movement of the treadmill belt, without a real need for propulsion. However, stride length is also determined by the forward distance that can be reached during swing, which is obviously not directly affected by the movement of the treadmill belt. Nevertheless, the enhanced hip extension because of treadmill motion may affect the pendular motion of the swinging limb.

Maximal stride frequency and length are both limiting the maximal speed of SCI subjects. However, the maximal stride frequency is a more important limitation. The active muscle stiffness acting to conserve energy during walking is deficient in SCI subjects. It is expected that during overground walking, the weakness of propulsive muscles would limit the maximal stride length to a greater extent than during treadmill walking. This needs to be further investigated. Future overground studies should include measurements of kinetics and trunk movements as well as assessing the role of balance in the adaptation to higher speeds.

It is suggested that walking rehabilitation interventions should focus on strategies aimed at improving the capacity to produce rapid alternate rhythmical stepping movements of the lower limbs. The finding that therapeutic interventions, such as FES-assisted walking, not only improve the maximal walking speed but also change the preferred frequency[27] is noteworthy and deserves more investigation.

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References

1 Grillner S, Halbertsma J, Nilsson J, Thorstensson A. The adaptation to speed in human locomotion. *Brain Res* 1979; **165:** 177–182.
2 Nilsson J, Thorstensson A, Halbertsma J. Changes in leg movements and muscle activity with speed of locomotion and mode of progression in humans. *Acta Physiol Scand* 1985; **123:** 457–475.
3 Zatsiorsky VM, Werner SL, Kaimin MA. Basic kinematics of walking. *J Sports Med Phys Fitness* 1994; **34:** 109–134.
4 Pépin A, Norman KE, Barbeau H. Treadmill walking in spinal-cord-injured subjects with an incomplete motor function loss: 1) adaptation to changes in speed. *(Companion paper submitted to *Spinal Cord*).
5 Maynard Jr FM *et al*. International standards for neurological and functional classification of spinal cord injury. *Spinal Cord* 1997; **35:** 266–274.
6 Dietz V, Quintern J, Berger W. Electrophysiological studies of gait in spasticity and rigidity. Evidence that altered mechanical properties of muscle contribute to hypertonia. *Brain* 1981; **104:** 431–449.
7 Conrad B, Benecke R, Meinck HM. Gait disturbances in paraspastic patients. In: Delwaide PJ, Young RR (eds). *Clinical Neurophysiology in Spasticity*. Elsevier Science Publishers: Amsterdam 1985, pp 155–174.
8 Fung J, Barbeau H. A dynamic EMG profile index to quantify muscular activation disorder in spastic paretic gait. *Electroenceph Clin Neurophysiol* 1989; **73:** 495–491.
9 Bohannon RW. Strength of lower limb related to gait velocity and cadence in stroke patients. *Physiother Can* 1986; **38:** 204–206.
10 Nadeau S, Arsenault AB, Gravel D, Bourbonnais D. Analysis of the clinical factors determining natural and maximal gait speeds in adults with a stroke. *Am J Phys Med Rehabil* 1999; **78:** 123–130.
11 Nilsson J, Thorstensson A. Adaptability in frequency and amplitude of leg movements during human locomotion at different speeds. *Acta Physiol Scand* 1987; **129:** 107–114.
12 Ladouceur M, Barbeau H. Functional electrical stimulation assisted walking for persons with incomplete spinal injuries: 1. Longitudinal changes of the maximal overground walking speed. *Scan J Rehabil Med* 2000; **32:** 28–36.
13 Holt KG. Constraints in the emergence of preferred locomotory patterns. In: Rosenbaum DA, Collyer CE (eds). *Timing of Behavior: Neural, Computational and Psychological Perspectives*. MIT Press: Cambridge, MA 1998.
14 McMahon TA. *Muscles, Reflexes and Locomotion*. Princeton University Press: Princeton, NJ 1984.
15 Winter DA. Biomechanical motor patterns in normal walking. *J Mot Behav* 1983; **15:** 302–330.
16 Katz RT, Rymer WZ. Spastic hypertonia: mechanisms and measurement. *Arch Phys Med Rehabil* 1989; **70:** 144–155.
17 Dietz V, Berger W. Normal and impaired regulation of muscle stiffness in gait: a new hypothesis about muscle hypertonia. *Exp Neural* 1983; **79:** 680–687.
18 Sinkjaer T, Magnussen I. Passive, intrinsic, and reflex-mediated stiffness in the ankle extensors of hemiparetic patients. *Brain* 1994; **117:** 355–363.
19 Pépin A, Barbeau H. Effects of treadmill speed and incline on the walking pattern of normal and spastic paretic subjects. *Soc Neurosci Abstr* 1992; **18:** 860.
20 Nadeau S, Gravel D, Arsenault AB, Bourbonnais D. A mechanical model to study the relationship between gait speed and muscular strength. *IEEE Trans Rehabil Eng* 1996; **4:** 386–394.
21 Olney S, Griffin MP, McBride ID. Temporal, kinematic, and kinetic variables related to gait speed in subjects with hemiplegia: a regression approach. *Phys Ther* 1994; **74:** 872–885.
22 Bajd T *et al*. Influence of electrically stimulated ankle plantar flexors on the swinging leg. *Artif Organs* 1997; **21:** 176–179.
23 Bajd T *et al*. Improvement in step clearance via calf muscle stimulation. *Med Biol Eng Comput* 1997; **35:** 113–116.
24 Bajd T, Kralj A, Stefancic M, Lavrac N. Use of electrical stimulation in the lower extremities of incomplete spinal cord injured patients. *Artif Organs* 1999; **23:** 403–409.
25 Nilsson J. On the adaptation to speed and mode of progression in human locomotion. PhD dissertation Department of Physiology III, Karolinska Institute and University College of Physical Education, Stockholm, Sweden; 1990.
26 Martin PE, Marsh AP. Step length and frequency effects on ground reaction forces during walking. *J Mot Behav* 1983; **15:** 302–330.
27 Ladouceur M, Barbeau H. Functional electrical stimulation assisted walking for persons with incomplete spinal injuries: 2. Changes in the kinematics and physiological cost of overground walking. *Scand J Rehabil Med* 2000; **32:** 72–79.
28 Cotes JE, Meade F. The energy expenditure and mechanical energy demand in walking. *Ergonomics* 1990; **3:** 97–119.
29 Zarrugh MY, Todd FN, Ralston HJ. Optimization of energy expenditure during level walking. *Eur J Appl Physiol* 1974; **33:** 293–306.
30 Holt KG, Hamill J, Andres RO. Predicting the minimal energy costs of human walking. *Med Sci Sports Exercise* 1991; **23:** 491–498.
31 Minetti AE *et al*. Effects of stride frequency on mechanical power and energy expenditure of walking. *Med Sci Sports Exercise* 1995; **27:** 1194–1202.
32 Holt KG, Hamill J, Andres RO. The force-driven harmonic oscillator as a model for human locomotion. *Hum Mov Sci* 1990; **9:** 55–68.
33 Holt KG, Obusek JP, Fonseca ST. Constraints on disordered locomotion – a dynamical systems perspective on spastic cerebral palsy. *Hum Movement Sci* 1996; **15:** 177–202.
34 Obusek JP, Holt HG, Rosenstein RM. The hybrid mass–spring pendulum model of human leg swinging: stiffness in the control of cycle period. *Biol Cybernet* 1995; **73:** 139–142.
35 Mirbagheri MM, Ladouceur M, Barbeau H, Kearney RE. The effects of long term FES-assisted walking on intrinsic and reflex dynamic stiffness in spinal cord-injured subjects. *IEEE Trans Neural Syst Rehabil Eng* 2002; **10:** In press.
36 Zernicke RF, Schneider K, Buford JA. Intersegmental dynamics during gait: implications for control. In: Patla AE (ed). *Adaptability of Human Gait: Implications for the Control of Locomotion, Advances in Psychology*, vol. **78**. Elsevier Science Publishers BV: Amsterdam 1991.