Improved Intralimb Coordination in People With Incomplete Spinal Cord Injury Following Training With Body Weight Support and Electrical Stimulation

Background and Purpose. Limb coordination is an element of motor control that is frequently disrupted following spinal cord injury (SCI). The authors assessed intralimb coordination in subjects with SCI following a 12-week program combining body weight support, electrical stimulation, and treadmill training. Subjects. Fourteen subjects with long-standing (mean time post-SCI = 70 months, range = 12–171 months), incomplete SCI participated. Three subjects without SCI provided data for comparison. Methods. A vector-based technique was used to assign values to the frame-by-frame changes in hip/knee angle, and vector analysis techniques were used to assess how closely the hip/knee angles of each step cycle resembled those of every other step cycle. Overground and treadmill walking speeds also were measured. Results. Following training, 9 of the 14 subjects with SCI demonstrated greater intercycle agreement. Mean overground and treadmill walking speeds improved (84% and 158%, respectively). Discussion and Conclusion. The intervention used in this study is based on our current understanding of the role of afferent input in the production of walking. Although the study sample was small and there was no control group, results suggest that training may improve intralimb coordination in people with SCI. [Field-Fote EC, Tepavac D. Improved intralimb coordination in people with incomplete spinal cord injury following training with body weight support and electrical stimulation. Phys Ther. 2002;82:707–715.]

Key Words: Body weight support, Coordination, Electrical stimulation, Spinal cord injury, Treadmill training, Walking.

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Since the late 1980s, treadmill training with body weight support (BWS) has been used in many research and rehabilitation settings to improve locomotor function in people with neurological deficits; this work has been widely discussed in the literature. In addition to deficits in locomotor function, people with spinal cord injury (SCI) often exhibit disordered spinal reflexes that have negative effects on locomotor function. The use of electrical stimulation (ES) to elicit reflex-based movements may result in restoration of some normal spinal reflex activity in these individuals, and it has been proposed that this restoration of reflex activity may be useful in decreasing some of the functional impairment in people with disordered motor activity. The combined use of BWS, ES, and treadmill training is a recent innovation, and it has been shown to increase walking speed in people with paresis due to incomplete SCI and cerebrovascular accident. In people with incomplete SCI, this increase in walking speed is accompanied by an increase in lower-extremity force production as well as increases in the distance they can walk and improvement in the quality of gait (eg, step rhythm, foot placement, step symmetry, weight shift). Although speed, force, and endurance are valuable measures of walking ability, they do not address the issue of whether training produces changes in motor control, as indicated by improved coordination of movement.

Motor behaviors can be distinguished from each other by the pattern of coordination between limbs (interlimb coordination) or between limb segments (intralimb coordination). Forward walking, for example, can be distinguished from backward walking by the coupling relationship, or relative movement, between the hip and the knee. The relative movement of joint angles produced during the performance of a motor task provides a means of assessing intralimb coordination. During the performance of a multilimb coordination. During the performance of a multicyclic behavior, the cycle-to-cycle agreement (consistency of the behavior or the extent to which cycles resemble each other) of this intralimb relationship furnishes a gauge of the ability to consistently reproduce the behavior and, therefore, can be considered a measure of the degree of coordination. We believe this consistency, in turn, may offer insight regarding the stability of the control mechanisms underlying coordination of the behavior.

Coordinated limb movement demands a complex interaction between the motor output of the central nervous system and the biomechanical constraints and advantages inherent in the anthropometry of the individual. We contend that the degree of coordination offers a measure of the integrated function of the systems involved in the control of movement. Researchers have demonstrated that in nonhuman animals, the spinal cord contains the neural circuitry (ie, central pattern generators) to produce well-coordinated movements that are highly reproducible and that this circuitry is amenable to training. Humans are thought to have a similar organization of spinal cord circuitry such that locomotor output is largely produced at the level of the spinal cord. However, although a highly consistent pattern of hip-knee coordination is observed during treadmill walking in people without SCI, this pattern is disrupted in people with SCI.

The purpose of our investigation was to test the hypothesis that intralimb coordination improves in people with incomplete SCI following participation in a locomotor training program. We defined coordination as the ability to produce a consistent pattern of hip-knee coupling over multiple cycles. We addressed the following questions: (1) Can training affect the consistency of the

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hip-knee coupling relationship even when walking at faster speeds? and (2) Following training, does the coupling relationship in subjects with SCI more closely resemble that observed in subjects without SCI? We also were interested in the effects of training on overground and treadmill walking speeds.

Methods

Subjects and Data Collection

A total of 17 subjects participated in this study. Fourteen participants had incomplete SCI (5 women, 9 men; mean age=31 years, SD=10.3 range=18–50). Nine of these subjects had tetraplegia, and 5 subjects had paraplegia. Using the American Spinal Injury Association (ASIA) Impairment Scale, all subjects with SCI were classified as ASIA C (sensory and motor function are preserved below the level of the lesion, but at least half of the muscles below the level of the lesion had a manual muscle test grade of less than 3). All of these subjects had long-standing SCI as all had sustained an SCI above the T10 neurologic level at least 1 year prior to participation in the study (mean time post-SCI=70 months, SD=47.6, range=12–171).

In our opinion, subjects with SCIs below the level of T10 often have well-preserved lower-limb muscles, paralysis accompanied by profound atrophy in the distal lower-limb musculature, and a lack of spasticity, defined as a velocity-dependent increase in resistance to passive stretch. Our observations are consistent with the extensive loss of motoneurons known to occur within the caudal regions of the lumbosacral enlargement. These individuals, we believe, are not likely to respond to ES directed at eliciting a spinal-level reflex. In addition to the subjects with SCI, 3 participants with no known orthopedic or neurological deficits provided data for purposes of comparison (1 woman, 2 men; mean age=33.7 years, SD=8.4, range=24–39). These subjects were a sample of convenience drawn from our laboratory staff. A larger sample size was deemed not to be necessary because cycle-to-cycle variations in the gait of people without SCI have been shown to be minimal. Each subject provided written informed consent consistent with regulations for protection of human subjects. Subjects with SCI were tested prior to and following participation in the training program. The subjects without SCI were tested on a single occasion.

Testing

Reflective markers were placed over the fifth metatarsal head, lateral malleolus, lateral tibial plateau, greater trochanter, and mid-trunk. For safety, the subjects with SCI were strapped into a harness* that was suspended from an overhead winch, but no BWS was provided during the testing sessions. Subjects with SCI walked on the treadmill at their maximum comfortable walking speed and were allowed to use whatever footwear and orthotic device they typically used when walking. Treadmill speed was gradually increased in 0.1 m/s intervals. Subjects were allowed 10 steps to adjust to each new speed. They were instructed to say "too fast" at the point where they felt walking speed was no longer comfortable. At that point, the treadmill speed was decreased by 0.1 m/s, and this speed was defined as that subject’s maximum comfortable walking speed. The subjects without SCI walked at a treadmill speed of 1.0 m/s, which is slower than normal walking speed and which we believed would allow more appropriate comparisons to be made with data from subjects with SCI. Subjects with SCI were videotaped (Panasonic VHS VTR†; 60 Hz) from the sagittal view on the side of the weaker limb while they walked unassisted (in the absence of both BWS and ES). The subjects without SCI were videotaped from the left side.

Training

Subjects with SCI participated in a 36-session (3 days a week for 12 weeks) training program of BWS- and ES-assisted treadmill walking. Subjects were allotted a 1½-hour block of time during which they were permitted to determine their own walk/rest bouts. Body weight support was provided by the harness/overhead winch complex. The level of BWS provided to each subject could be adjusted via this motorized winch and could be monitored via a light-emitting diode display. Electrical stimulation applied to the common peroneal nerve via a Grass S88 stimulator‡ coupled to a Grass SIU5 stimulus isolation unit§ was triggered using a hand switch at the time of terminal stance to elicit a flexion withdrawal response to assist with stepping in the weaker limb. Stimulator settings were: 500- to 750-millisecond train, 50 to 80 pulses per second, 1.0- to 1.5-millisecond pulse duration, and 60 to 150 V. Voltage amplitude was dependent on subject tolerance and the level of ES necessary to elicit a brisk flexion withdrawal reflex. Within each training session, the treadmill speed and amount of BWS provided was adjusted to allow the subject to walk optimally, as determined by the professional judgment of the physical therapist-trainer, on the treadmill. Stimulator settings were adjusted within and between sessions to elicit an optimal flexion withdrawal response in which a brisk dorsiflexion, knee flexion, and hip flexion response resulted in the lower limb withdraw-

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† Matsushita Electric Corporation of America, One Panasonic Way, Secaucus, NJ 07094.
‡ Astro-Med Industrial Park, 600 E Greenwich Ave, West Warwick, RI 02893.
ing from the support surface. Subjects were encouraged to walk at their fastest comfortable walking speed and were told that time was not important. Subjects were allowed to use the treadmill handrails for balance, if they deemed it necessary, but they were discouraged from using their upper extremities for weight bearing. No attempts were made to wean the subjects off the weight support during training. Additional details regarding the intervention are published elsewhere.\(^{13}\)

**Data Analysis**

Data from 30 seconds of walking were analyzed for each subject. The number of complete step cycles that the subjects with SCI could perform varied from 4 to 8 steps during the pretraining test session. Videotaped data were digitized at 30 frames a second, and hip and knee joint angles were calculated using a 2D Peak Motion Measurement System\(^{29}\) according to conventional definitions\(^{29}\) and filtered at 4 Hz using a Butterworth filter.\(^{29}\) The hip angle was defined as the angle formed by the segments represented by the trunk and by the thigh. The knee angle was defined as the angle formed by the segments represented by the thigh and by the shank. The movement direction of the knee relative to the hip in each frame-to-frame interval of videotaped data was quantified based on a vector coding technique developed by Tepavac and Field-Fote.\(^{30}\) This technique represents an alternative to relative phase analysis and has been shown to yield valid and reliable measurements.\(^{30}\)

It was designed to assist clinicians in interpreting the data because we believe they are more likely to think of movement in terms of joint angles as opposed to phase values. In addition, the vector coding technique may have an advantage over other methods of quantifying angle-angle because it allows the simultaneous comparison of multiple cycles, whereas other techniques are limited to pair-wise comparisons.\(^{17,31}\)

The step cycle period was normalized to the mean cycle period for each subject using a spline interpolation technique. We selected the video frame wherein heel-strike occurred as the starting point (frame 1) for encoding. The difference between frame 1 and frame 2 for the hip angle values \((x_{1,2})\) and the knee angle values \((y_{1,2})\) was determined. These values represent the change in the \(x\) and \(y\) directions, respectively, in the frame-to-frame interval between frame 1 and frame 2. The vector formed by the line segment joining frame 1 and frame 2 has both direction and magnitude. If the line segment joining frames 1 and 2 of the first step cycle has the same direction as the vector joining frames 1 and 2 of the second cycle, and if this is true of all frame-to-frame intervals in the 2 cycles, then the relative motion plot for cycle 1 and cycle 2 will have the same shape (although not necessarily the same area). The angular direction of the line segment, \(l_{1,2}\), between 2 consecutive points or frames (Fig. 1) was calculated using the formula:

\[
l_{1,2} = \sqrt{(x_{1,2})^2 + (y_{1,2})^2}
\]

The cosine and sine of \(l_{1,2}\) were found using the formulas:

\[
\cos \theta_{1,2} = x_{1,2}/l_{1,2}
\]

and

\[
\sin \theta_{1,2} = y_{1,2}/l_{1,2}
\]

This process was repeated for each frame-to-frame interval within each cycle. The mean cosine \((\cos \bar{\theta})\) and sine \((\sin \bar{\theta})\) for a given frame-to-frame interval over multiple cycles (eg, frame 1–2 of cycles 1–6), was calculated, and the mean vector length for that frame-to-frame interval was then determined using the formula:

\[
\sqrt{(\cos \bar{\theta}_{1,2})^2 + (\sin \bar{\theta}_{1,2})^2} = a_{1,2}
\]

The length of the mean vector, \(a\), denotes the degree of dispersion (or conversely of concentration) of the hip-knee values about the mean\(^{32}\) over multiple cycles for that particular frame. The larger the value of \(a\) (between 0 and 1), the less variable (ie, less randomly distributed, more consistent) is the hip-knee relationship.

The arithmetic average, \(\bar{a}\), of all the mean vector lengths is found by:

\[\text{Figure 1. Direction of movement from origin, "x," and the associated sine, cosine for a sample of 8 possible positions for the subsequent point [frame]. The actual possible positions are infinite.}\]

\^6 Peak Performance Technologies Inc, 7388 S Revere Pkwy, Englewood, CO 80112.
Table 1.
Subject Characteristics and Performance Values$^a$

<table>
<thead>
<tr>
<th>Subject No.</th>
<th>Age (y)</th>
<th>Sex</th>
<th>Neurologic Level$^b$</th>
<th>Time Postinjury (mo)</th>
<th>Orthotic Device</th>
<th>Overground Walking Speed (m/s) (no ES/no BWS)</th>
<th>Treadmill Walking Speed (m/s) (BWS and ES)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Pretraining</td>
<td>Posttraining</td>
</tr>
<tr>
<td>1</td>
<td>27</td>
<td>M</td>
<td>C6</td>
<td>64</td>
<td>R KAFo</td>
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<td>0.08</td>
</tr>
<tr>
<td>2</td>
<td>50</td>
<td>M</td>
<td>C5</td>
<td>111</td>
<td>B AFOs</td>
<td>0.04</td>
<td>0.06</td>
</tr>
<tr>
<td>3</td>
<td>24</td>
<td>M</td>
<td>C5</td>
<td>43</td>
<td>R AFO</td>
<td>0.03</td>
<td>0.04</td>
</tr>
<tr>
<td>4</td>
<td>27</td>
<td>M</td>
<td>C6</td>
<td>84</td>
<td>R AFO</td>
<td>0.16</td>
<td>0.29</td>
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<tr>
<td>5</td>
<td>31</td>
<td>F</td>
<td>T6</td>
<td>12</td>
<td>R AFO</td>
<td>0.13</td>
<td>0.20</td>
</tr>
<tr>
<td>6</td>
<td>37</td>
<td>M</td>
<td>C4</td>
<td>60</td>
<td>NA</td>
<td>0.05</td>
<td>0.08</td>
</tr>
<tr>
<td>7</td>
<td>33</td>
<td>M</td>
<td>T3</td>
<td>141</td>
<td>B AFOs</td>
<td>0.11</td>
<td>0.29</td>
</tr>
<tr>
<td>8</td>
<td>19</td>
<td>M</td>
<td>C5</td>
<td>16</td>
<td>R AFO</td>
<td>0.22</td>
<td>0.26</td>
</tr>
<tr>
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<td>M</td>
<td>C5</td>
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<td>0.25</td>
<td>0.29</td>
</tr>
<tr>
<td>10</td>
<td>44</td>
<td>F</td>
<td>T3</td>
<td>105</td>
<td>B AFOs</td>
<td>0.08</td>
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</tr>
<tr>
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<td>36</td>
<td>F</td>
<td>C5</td>
<td>171</td>
<td>B AFOs</td>
<td>0.11</td>
<td>0.14</td>
</tr>
<tr>
<td>12</td>
<td>18</td>
<td>M</td>
<td>C5</td>
<td>18</td>
<td>B AFOs</td>
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<td>0.08</td>
</tr>
<tr>
<td>13</td>
<td>45</td>
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<td>T7</td>
<td>56</td>
<td>B AFOs</td>
<td>0.08</td>
<td>0.20</td>
</tr>
<tr>
<td>14</td>
<td>20</td>
<td>F</td>
<td>T4</td>
<td>41</td>
<td>B AFOs</td>
<td>0.19</td>
<td>0.38</td>
</tr>
</tbody>
</table>

$^a$ES = electrical stimulation, BWS = body weight support, R = right, B = bilateral, NA = not applicable, AFO = ankle-foot orthosis, KAFo = knee-ankle-foot orthosis.

$^b$Neurologic level" is an American Spinal Injury Association term defined as “the most caudal segment of the spinal cord with normal sensory and motor function on both sides of the body.$^{26}$

\[
(5) \quad a_{1,2} + a_{2,3} + \ldots + a_{n-1,n}/n = \bar{a}
\]

where \(n\) is the number of frames per cycle and \(\bar{a}\) is the angular component of the coefficient of correspondence (ACC), which indicates the overall variability of the hip-knee relationship for all included cycles. If the relative motion between the hip and the knee is in perfect agreement over multiple cycles, then \(\bar{a} = 1\), indicating maximal consistency between cycles.

Data were analyzed using Microsoft Excel 97 SR-2 Statistical Tool Pac$^1$ and customized statistical programs. The required level of significance for all tests was set at \(P < .01\). Nonparametric statistics were used to compare changes in the ACC and to assess the relationship between change in this value and change in overground and treadmill walking speeds; nonparametric statistics were used because we chose not to make assumptions about the distribution of the measurements from the population we sampled. To compare differences between pretraining and postraining ACC values, we used the Wilcoxon test, a matched-pairs, signed-rank test for nonparametric data.$^{33}$ Treadmill speed during testing was recorded for all subjects to assess change in walking speed over the course of training. Spearman rank correlation coefficients \((r_s)\) were used to assess the relationship between change in the ACC and change in overground walking speed. In addition, we acknowledged that a movement pattern might be highly reproducible over time, but that this should not be construed to mean that the movement pattern resembles that of individuals without SCI. Therefore, to assess whether the timing of the hip-knee coordination pattern observed in subjects with SCI resembled that of subjects without SCI, we evaluated the timing of knee extension onset within the hip flexion-extension cycle. This variable has been used previously for the purpose of assessing similarities among different forms of behavior.$^{18}$ Cycle period was defined as the time from the onset of hip flexion (0) to the onset of the subsequent hip flexion (1). The phase value of the first knee extension onset during each cycle was calculated. A one-tailed, matched-pairs \(t\) test was used to test for differences between pretraining and postraining walking speed and for differences between pretraining and postraining knee extension onset phase.

**Results**

Descriptive information for the 14 subjects with SCI who participated in the training program is given in the Table. Representative relative motion plots for one subject without SCI and one subject with SCI before and after training are shown in Figure 2. Nine of the 14 subjects with SCI showed increases in the ACC over multiple cycles during treadmill walking, as measured by a larger ACC for the postraining test compared with pretraining values (Fig. 3). The mean values of the ACC in the group with SCI prior to and following training were 0.56 and 0.65, respectively. The mean value of the ACC for the 3 subjects without SCI was 0.94. Postraining overground and treadmill walking speeds were also greater than pretraining values. On average, subjects walked 84% (range = 16%–220%) faster overground and

$^1$ Microsoft Corp, One Microsoft Way, Redmond, WA 98052.
158% (range=41%-763%) faster on the treadmill during the posttraining test compared with the pretreatment training. There was a good correlation between change in overground walking speed and change in the ACC (Spearman $r_s=.75$), and there was a moderate correlation between change in treadmill walking speed and change in ACC (Spearman $r_s=.54$). These levels of correlation were within the ranges defined as “good” and “moderate” by Portney and Watkins.34 Of the 5 subjects with SCI who did not show a change in the ACC following training, 4 ranked in the lowest third of the subject pool with regard to change in overground walking speed.

In the subjects without SCI, the onset of knee extension occurred at a mean phase of 0.2 (Fig. 4A). In the subjects with SCI, the onset of knee extension occurred at a mean phase of 0.32 (SD=0.15) prior to training and of 0.28 (SD=0.09) following training (Figs. 4B, 4C). There was no difference between pretraining and posttraining knee extension onset phase in subjects with SCI. There was no change in the within-subject variance between pretraining and posttraining values.

**Discussion**

Our goal was to assess changes in intralimb coordination in individuals with incomplete SCI who participated in a 12-week walking program consisting of BWS, ES, and treadmill training. The results showed that there was an increase in the consistency of the walking pattern, as indicated by the higher posttraining ACC values compared with the pretreatment values. This was true despite the fact that the subjects were also walking faster after training.

Examination of the relative motion plots generated by subjects with SCI revealed that no 2 individuals with SCI generated similar patterns of coordination. This finding is consistent with reports from other investigators.25 Furthermore, it was not possible to discern subject category (eg, cervical versus thoracic level of injury) based on these plots, an observation that was made by other researchers.25 This was in contrast to the subjects without SCI whose plots were similar, as has been noted by other researchers.25 Based on these findings, we recognized that in individuals with SCI, it is possible to

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**Figure 2.** Exemplary relative motion plots from one subject without spinal cord injury (SCI) and one subject with SCI allows a qualitative assessment of these data. The plot for the subject without SCI (Fig. 2A) suggests a high degree of consistency among cycles and is in accordance with the low variability of the hip-knee coupling relationship given by the angular component of the coefficient of correspondence (0.96). These values were calculated using the vector coding and analysis technique.30 The angle-angle diagram for the subject with SCI, prior to participation in the assisted walking program (Fig. 2B), suggests less consistency than that of the subject without SCI. The value for these data (0.40) was in accordance with that assessment. Following participation in the assisted walking program, the value for the hip-knee coupling relationship increased (to 0.70; Fig. 2C), reflecting the greater consistency observed in the angle-angle diagram.
develop a movement strategy that becomes more consistent with training and yet remains very different from that typical of people without SCI. For this reason, we selected knee extension onset phase as a kinematic marker to permit a comparison of temporal measurements between subjects with and without SCI. In the pretraining test, the onset of knee extension within the hip cycle occurred much later in the subjects with SCI compared with the subjects without SCI.

The combined use of BWS, ES, and treadmill training may provide an optimal sensory environment to promote improved walking in people with SCI. This view has considerable theoretical support. First, the application of ES to the common peroneal nerve to produce a flexion withdrawal response makes use of spinal-level neural circuitry, and this type of stimulation has been shown to attenuate abnormal reflex activity in subjects with spasticity. Second, BWS decreases lower-extremity load and can be varied to meet the needs of the individual. Furthermore, some authors have shown that lower-extremity load increases extensor muscle activity, the presence of which, hypothetically, could interfere with the ability to initiate the limb flexion necessary for swing phase. By decreasing lower-extremity load through BWS, step initiation may be facilitated. The harness provided a secure environment in which the subjects did not need to fear a loss of balance, and subjects were free to experiment with movement strategies that might otherwise not be attempted. Third, the motorized treadmill provides temporal cues associated with stepping and also assists with hip extension in the stance limb. This hip extension may be critical to the initiation of the swing phase. Finally, this program makes use of a task-oriented approach with the goal of improving the performance of that task. The need for task specificity in training is well-established.

Outcome measures in locomotor rehabilitation often, we believe, focus on variables such as speed and muscle force. Such measures are important, but they offer little evidence as to whether there has been a meaningful change in a person’s ability to control movement. We argue that the consistency of the pattern of coordination in a multicyclic behavior such as walking is a practical way to measure improvements in control of movement. Such a measure, in our view, provides the means to assess change in the integrated functions of the neuromuscular and the musculoskeletal systems. We contend that the results of our study are important because they suggest that, with training, it is possible to improve limb coordination during walking in individuals with long-standing, incomplete SCI. We believe the technique used in our study to encode relative hip-knee motion is mathematically equivalent to a previously reported technique, yet it offers the advantage of being able to simultaneously compare multiple cycles of behavior.

**Limitations**

Our study sample was limited to 14 subjects for whom a single type of intervention was investigated. It would be useful to compare changes in both intralimb and interlimb coordination that occur with other forms of locomotor training to discern which methods are associated with the greatest improvements. The speed of the treadmill was faster during the posttraining test than it was during the pretraining test, as the subjects’ maximum comfortable walking speed had increased. This difference in treadmill speed may have had some effect on our results. Future studies should consider comparing pretraining and posttraining test measurements obtained at the same walking speed.

At the time the data were collected, limitations in the layout of our laboratory precluded our ability to videotape subjects walking overground; thus, our kinematic assessments of walking were limited to the treadmill. Future investigations will be needed to determine whether the improvements in coordination observed during treadmill walking are transferable to overground walking.
Conclusions
The training program used in our study was chosen in an attempt to make use of our current understanding of the affect of afferent input on spinal-level neural circuitry and the importance of task-oriented training. Our results indicate that, in people with SCI who have some ambulatory capacity, intralimb coordination can be improved with training. This improvement persists even when walking at faster speeds.

References


