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Locomotor Adaptation to Resistance During Treadmill Training Transfers to Overground Walking in Human SCI

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Abstract

Treadmill training has been used as a promising technique to improve overground walking in patients with spinal cord injury (SCI). Previous findings showed that a gait pattern may adapt to a force perturbation during treadmill training and show aftereffects following removal of the force perturbation. We hypothesized that aftereffects would transfer to overground walking to a greater extent when the force perturbation was resisting rather than assisting leg swing during treadmill training. Ten subjects with incomplete SCI were recruited into this study for two treadmill training sessions: one using swing resistance and the other using swing assistance during treadmill stepping. A controlled resistance/assistance was provided to the subjects' right knee using a customized cable-driven robot. The subjects' spatial and temporal parameters were recorded during the training. The same parameters during overground walking were also recorded before and after the training session using an instrumented walkway. Results indicated that stride length during treadmill stepping increased following the release of resistance load and the aftereffect transferred to overground walking. In contrast, stride length during treadmill stepping decreased following the release of assistance load, but the aftereffect did not transfer to overground walking. Providing swing resistance during treadmill training could enhance the active involvement of the subjects in the gait motor task, thereby aiding in the transfer to overground walking. Such a paradigm may be useful as an adjunct approach to improve the locomotor function in patients with incomplete SCI.

Introduction

Treadmill training has been used as a promising technique to improve overground walking in patients with incomplete spinal cord injury (SCI) (Behrman and Harkema 2000; Field-Fote and Roach 2011; Field-Fote et al. 2005; Wernig and Muller 1992; Dietz et al. 1995). During treadmill training, the patient is given body weight support as necessary and provided with assistance to move their leg into a kinematically correct gait pattern. The assistance may be delivered by physical therapists or by a driven gait orthosis that precisely controls the kinematic trajectory (e.g., Lokomat) (Behrman and Harkema 2000; Colombo et al. 2000). This training paradigm largely meets the criteria for effective neuroplasticity: the training is task specific and allows the patients to use relevant sensorimotor pathways for walking (Harkema 2001). While treadmill training provides a convenient environment for the patients to experience a task-specific stepping practice, the effectiveness of this technique is dependent on whether the training effect can transfer to "real-world" overground walking, as the ultimate goal of treadmill training is to improve patients' ability to walk overground.

Previous studies indicate that robotic-assisted treadmill training may increase overground walking speed and endurance for some patients with incomplete SCI (Wirz et al. 2005; Field-Fote et al. 2005), suggesting that locomotor skill obtained during treadmill training may partially transfer to overground walking in patients with incomplete SCI. However, the functional gains obtained after robotic-assisted treadmill training are relatively small. For example, a recent randomized study showed that patients with incomplete SCI only obtained marginal improvements in overground walking speed (i.e., 0.01 ± 0.05 m/s) after 12 weeks of robotic-assisted treadmill training (Field-Fote and Roach 2011). As a consequence, there is a need to improve the transfer efficacy of robotic treadmill training.

Understanding the underlying mechanisms of motor adaptation during treadmill training is crucial for improving the transfer efficacy of robotic treadmill training in patients with SCI. Previous studies suggest that the central neural system (CNS) uses an internal model to control the leg kinematics during walking (Emken and Reinkensmeyer 2005; Noble and Prentice 2006). An internal model is defined as a neural representation of the body, task, and environment (Kawato 1999). When detecting a discrepancy between the predicted and actual movement due to an external environment perturbation (e.g., robotic-generated force field), the CNS will recalibrate the internal model to minimize the discrepancy (Bastian 2008).

Previous studies show that healthy people may adapt to a force perturbation and show aftereffects during treadmill stepping (Emken and Reinkensmeyer 2005; Noble and Prentice 2006; Blanchette and Bouyer 2009; Lam et al. 2006). Specifically, subjects' gait patterns deviate from the origin when they are first exposed to a force perturbation. Several steps later, subjects gradually adapt to the force perturbation and their gait pattern returns to the origin. Once the force perturbation is removed, an "aftereffect" is observed in the direction opposite to the force perturbation (Emken and Reinkensmeyer 2005; Noble and Prentice 2006). The occurrence of an aftereffect suggests that the internal model has been recalibrated to counterbalance the anticipated force perturbation in the upcoming steps.

Patients with incomplete SCI may preserve the ability to recalibrate the internal model when experiencing a force perturbation during treadmill walking. For instance, subjects with incomplete SCI show an aftereffect consisting of increased step length following removal of a resistance load during treadmill walking (Houldin et al. 2011). The increase in step length may lead to an improvement of walking speed, as walking speed is a function of step length and cadence. However, it remains unclear whether the aftereffect (e.g., the increased step length) induced by resistance load perturbations can transfer to overground walking in patients with incomplete SCI. This is crucial for the clinical application of such training paradigms.

In this study, we investigated the transfer of aftereffects obtained from treadmill training with force perturbations to overground walking in patients with incomplete SCI. We examined two types of force perturbation: leg swing resistance and assistance. We assumed that a resistance load would increase error in leg swing, while an assistance load would reduce error. Since error is a driving force for motor adaptation (Bastian 2008), resistive treadmill training may induce more robust aftereffects for transfer as compared to assistive treadmill training. It was also expected that the patients would increase active involvement during resistive treadmill training in order to counterbalance the resistance force. Previous studies have suggested that active involvement in the training is essential to increase neural descending drive to improve motor performance (Lotze et al. 2003; Lotze and Cohen 2006). We hypothesized that aftereffects would transfer to overground walking to a greater extent following resistive treadmill training than following assistive treadmill training.

Methods

Subjects

Ten subjects with incomplete SCI, with the clinical features described in Table 1, were recruited in this study. Inclusion criteria included age between 18 and 65 years, level of injury between C1-T10, ASIA level at C or D, and ability to ambulate overground with assistive devices as needed. Exclusion criteria included multiple CNS lesion sites, urinary tract infection, other secondary infections, heterotopic ossification, respiratory insufficiency, significant osteoporosis, or inability to give informed consent. Informed consent was obtained, and all procedures were conducted in accordance with the Helsinki Declaration of 1975 and approved by the Institutional Review Board of Northwestern University, Chicago, IL.

Table 1 Subject information

Case	Age	Years post injury	Asia level	Level of injury	Use of aids
S1	43	2	D	C5–C6	No
S2	64	8	D	T9–T10	Bilateral forearm crutch
S3	50	7	D	C6–C7	No

S4	38	12	D	C5	Walker
S5	46	26	D	C5–C6	Bilateral cane
S6	47	3	D	T5–T7	No
S7	48	3	D	C6–C7	Walker
S8	63	2	D	C3–C7	Walker
S9	52	7	D	T7	No
S10	48	3	D	T5–T7	Walker

Test apparatus

A custom-designed cable-driven robot was used to provide controlled assistance/resistance load during treadmill training (Fig. 1a). A detailed description of the system has been reported previously (Wu et al. 2011). In brief, the system works in conjunction with a motorized treadmill and a body weight support system. It consists of four nylon-coated stainless-steel cables driven by four motors and cable spools. Each cable can be attached to a subject's leg through a strap to provide a controlled resistance or assistance load (Fig. 1b). Two cables were set in the front of the treadmill to provide an assistance load for leg swing. The additional two cables were set in the back of the treadmill to provide a resistance load against leg swing. The cable is light weight, compliant, and highly back-drivable and therefore has minimal constraints to patients' voluntary movement (Wu et al. 2011).

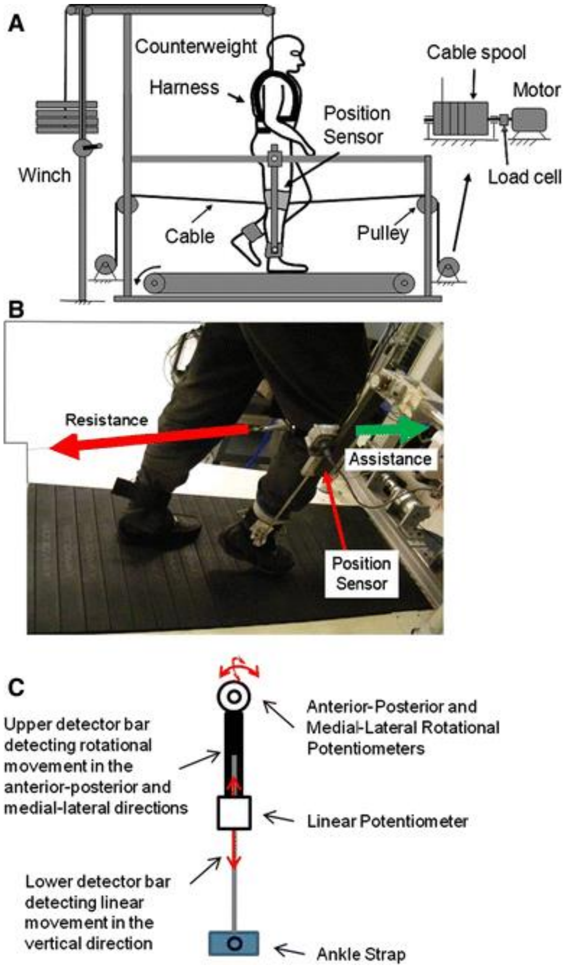

Fig. 1 a Illustration of the cable-driven robotic system from a lateral view. **b** Application of the resistance and assistance loads. **c** Illustration of the ankle position sensor from a lateral view

A customized three-dimensional (3D) position sensor was used to measure the ankle position during treadmill walking (Fig. 1c). The sensor consists of a detector rod and three potentiometers. Two potentiometers (P2201, Novotechnik, Southborough, MA) were used to measure rotational movements of the rod in the anterior-posterior and medial-lateral directions; one potentiometer (SP-2, Celesco, Chatsworth, CA) was used to measure the linear movement of the rod in the vertical direction. A customized program written in LabVIEW (National Instruments, Austin, TX) was used to acquire the ankle position data as well as to output the load command signals to the servomotor systems. In this study, the resistance and assistance loads were set to be applied from the late stance phase (when the ankle started to change its movement direction from backward to forward on the treadmill) to the first 40% of the swing phase. The ankle position data obtained from the 3D sensors were recorded throughout the training period using a data acquisition card (National Instruments, Austin, TX) on a personal computer with a sampling rate of 500 Hz.

Procedures

Each subject participated in two data collection sessions, one for resistive treadmill training and the other for assistive treadmill training. The two sessions were scheduled with 2 weeks in between to wash out any lingering carryover. Each subject completed resistive treadmill training before assistive treadmill training. In this study, resistance/assistance was applied to the subjects' right leg at the knee. Subjects were allowed to use the left leg to compensate for the kinematic changes of the right leg caused by the force perturbation. Additionally, subjects were allowed to hold onto a handrail during treadmill walking.

Each session consisted of three main components in sequence, including (a) a pre-training overground test, (b) resistive/assistive treadmill training, and (c) a post-training overground test. During (a) and (c), subjects were instructed to walk over an instrumented walkway (GaitMat II, E.Q. Inc, Chalfont, PA) twice at their comfortable speed. During resistive/assistive treadmill training, the training speed was set at the speed obtained in the pre-training overground test. Body weight support was provided as necessary to assure stable stepping. The amount of resistance/assistance provided was set at the maximum level that each subject felt comfortable to tolerate. The level was determined prior to treadmill training. Specifically, we gradually increased resistance load until the subject felt that he/she could walk with the load for 10 min without strenuous effort. The training speed, amount of body weight support provided, and level of resistance/assistance load are reported in Table 2.

Table 2 Training parameters

Case	% of BW support	AL (N)	RL (N)	TSA (m/s)	TSR (m/s)
S1	0	35	35	0.93	0.99
S2	5	25	25	0.26	0.27
S3	0	30	35	0.94	1.08
S4	20	40	30	0.52	0.78
S5	0	20	27	0.82	0.83
S6	0	30	25	0.83	0.87
S7	0	30	35	0.55	0.54
S8	0	45	25	0.54	0.43
S9	0	45	30	0.90	0.91
S10	0	30	25	0.71	0.87

BW body weight; *AL* assistance load; *RL* resistance load; *TSA* training speed in assistance session; *TSR* training speed in resistance session

The treadmill training component (b) was divided into five sequential periods. During the first period (pre-adaptation period), subjects stepped on a treadmill without receiving resistance/assistance for 2 min. During the second period (adaptation period), resistance/assistance load was applied to subjects' right knee for approximately 250 strides. During the third period (post-adaptation period), subjects continued stepping on the treadmill for another 2 min with the resistance/assistance being removed unexpectedly. The fourth period was a 10-min break. Following the break, subjects resumed the resistive/assistive treadmill training for another 5 min. The training was resumed to prepare subjects to re-adapt to the load before conducting a post-training overground test. A wheelchair was used for transportation of the subject from the laboratory with cable robot and treadmill to the laboratory with the GaitMat II.

Data analysis

Stride length, stance time, swing time, and speed during overground walking were calculated using the GaitMat II software. Two trials for each test condition were averaged for statistical analysis. Stride length, stance time, and swing time during treadmill walking were analyzed based on the ankle position data. The raw position data were smoothed using a fourth-order Butterworth low-pass filter with a cut-off frequency at 6 Hz. Based on the smoothed data, the events of heel contact and toe-off during the treadmill walking were identified to segment walking cycles. Heel contact was defined as the timing when the ankle trajectory changed its movement direction from forward to backward; toe-off was defined as the timing when the ankle trajectory changed its movement direction from backward to forward (Zeni et al. 2008).

Similar to overground walking, we defined stride length as the distance that the treadmill belt moved between two consecutive heel contacts of the right leg (Fig. 2). In Fig. 2, HC1 represents a point on the treadmill where the heel contacts the belt the first time. The point HC1 shifted along the belt in the direction of movement to a new location HC1′ when the heel contacts the belt again at location HC2. The distance from HC2 to HC1′ was defined as the stride length and can be expressed as:

$$\text{StrideLength} = S_2 + L,$$

(1)

where S_2 denotes the distance between HC2 and where the first toe-off occurs (TO1); L denotes the distance between TO1 and HC1′, which is equal to the distance that treadmill belt shifts during the time when the right leg moves from TO1 to HC2, i.e., swing time. The term L is written as:

$$L = VT_2,$$

(2)

where V denotes the treadmill belt speed; T_2 denotes swing time. The treadmill belt speed V *is* written as:

$$V = \frac{S_1}{T_1},$$

(3)

where S_1 denotes the distance between HC1 and TO1, i.e., the distance that the belt shifts during stance phase; T_1 denotes stance time. Integrating Eqs. 2–3 to 1, stride length is represented as following:

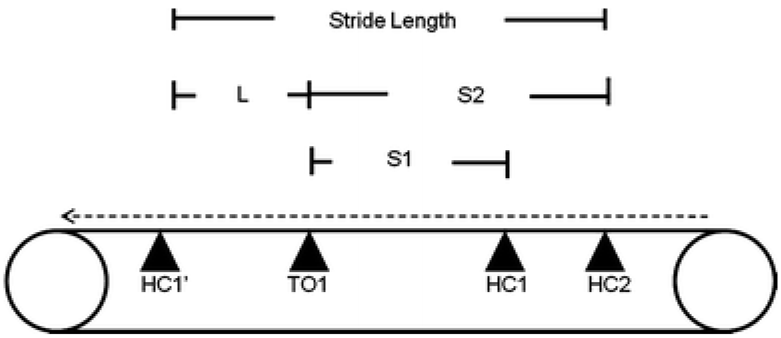

Fig. 2 Illustration for calculation of stride length during treadmill walking. HC1 = where the first heel contact occurs; TO1 = where the first toe-off occurs; HC2 = where the second heel contact occurs; HC1′ = the location of HC1 when the second heel contact occurs

$$\text{StrideLength} = S_2 + S_1 \times \frac{T_2}{T_1}.$$

(4)

Statistical analysis

Stride length, stance time, and swing time during treadmill walking were averaged across the last 5 strides of the pre-adaptation period (baseline), from the 101st to the 105th strides during the adaptation period (mid-adaptation), across the last 5 strides of the adaptation period (late adaptation), and across the first 5 strides of the post-adaptation period (early post-adaptation). The change score of each parameter was calculated from the baseline to mid-adaptation, from the baseline to late adaptation, and from the baseline to early post-adaptation. Two-way repeated-measures ANOVAs were conducted to determine whether the change scores varied as a function of training type (resistive vs. assistive), training period (mid-adaptation, late adaptation, and early post-adaptation), and the interaction between the type and period. Additionally, each parameter was compared between the baseline period and the first stride of the post-adaptation period using paired *t* tests or Wilcoxon signed-rank tests, depending on the normality of the data, determined by the Shapiro–Wilk test. Moreover, stride length, stance time, swing time, and speed between pre-training and post-training conditions were also compared using paired *t* tests or Wilcoxon signed-rank tests. All statistical analyses were conducted at the alpha level of 0.05 using Predictive Analytics Software (PASW) version 18 (SPSS Inc, Chicago, IL).

Results

Adaptation of stride length during treadmill walking

The stride length from a typical subject (S9) over the course of resistive treadmill training is shown in Fig. 3a. The baseline of stride length was 0.92 ± 0.02 m before resistive treadmill training. When the resistance was initially applied, the subject demonstrated a decrease in stride length from the baseline. Several strides later, the stride length increased at or even above the baseline. When the resistance was removed unexpectedly at the first stride of the post-adaptation period, the subject showed an aftereffect consisting of a greater stride length. The stride length still remained above the baseline in the following several strides although the magnitude sharply declined.

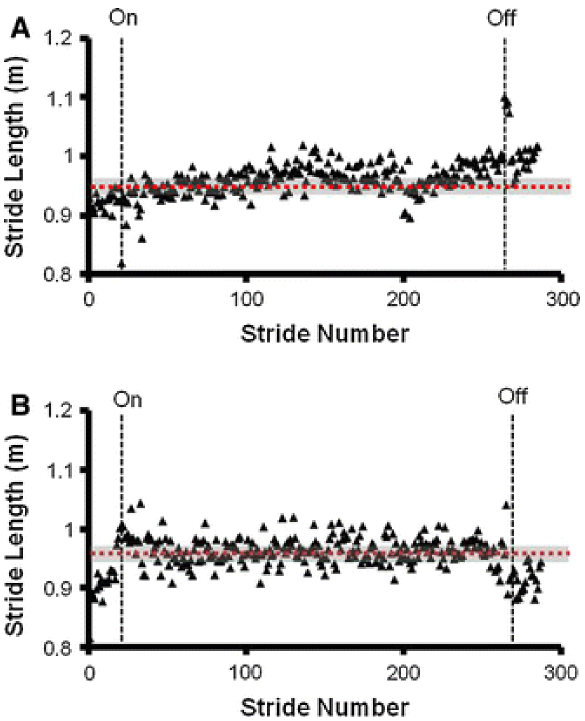

Fig. 3 a A representative subject's (S9) stride length over the last 20 strides of the pre-adaptation period (before load is on), the 250 strides of the adaptation period, and the first 20 strides of the post-adaptation period (after load is off) in resistive treadmill training. **b** The same subject's stride length in assistive treadmill training. The *red dashed horizontal line* in each plot represents the baseline value, which was the averaged stride length across the last 5 strides of the pre-adaptation period. The *gray* area represents one SD above and below the baseline

The same subject demonstrated a different adaptation pattern to the assistance load in stride length (Fig. 3b). The baseline of stride length was 0.95 ± 0.03 m before assistive treadmill training for this subject. The subject demonstrated an increase in stride length from the baseline right after the assistance was initiated. Several strides later, the stride length returned to the baseline. When the assistance was removed unexpectedly at the first stride of the post-adaptation period, the subject showed an aftereffect consisting of a reduced stride length. The stride length remained below the baseline in the following strides.

The group data also showed that the resistive treadmill training induced an aftereffect consisting of an increase in stride length, and the assistive treadmill training induced an aftereffect consisting of a decrease in stride length (Fig. 4a). Specifically, a paired *t* test showed that the first stride length of the post-adaptation period following resistive treadmill training was significantly greater than that of the baseline ($P = 0.009$). In contrast, the first stride length of the post-adaptation period following assistive treadmill training was significantly smaller than that of the baseline ($P = 0.016$).

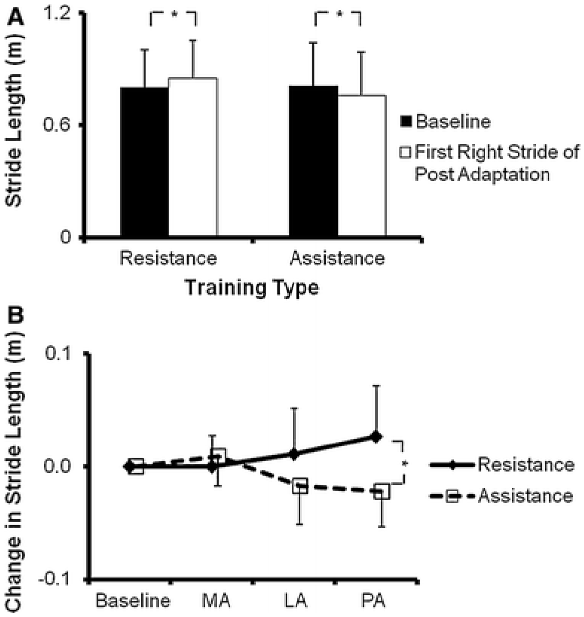

Fig. 4 a Stride length during the baseline and the first stride of the post-adaptation period in resistance and assistive treadmill training. **b** Changes in stride length from the baseline to the mid-, late-, and post-adaptation periods during resistance and assistive treadmill training. **P* < 0.05. *MA* mid-adaptation; *LA* late adaptation; *PA* post-adaptation

On average, the resistance load did not result in a substantial change in stride length during the mid-adaptation period, but resulted in an increase in stride length in the late adaptation period, and a further increase during the post-adaptation period (Fig. 4b). In contrast, the assistance load resulted in an increase in stride length during the mid-adaptation period, but resulted in a decrease during late adaptation and a further decrease during the post-adaptation period (Fig. 4b). The interaction between training type and training period on the change in stride length was significant (*P* = 0.004), as indicated by a two-way repeated measures ANOVA. The ANOVA also indicated a significant main effect of training type (*P* = 0.035), suggesting that subjects tended to produce a longer stride length during resistance than assistive treadmill training. The main effect of training period was not significant (*P* = 0.528).

Adaptation of stance time during treadmill walking

The stance time of subject S9 over the course of resistive treadmill training is shown in Fig. 5a. The baseline of stance time for this subject was 0.93 ± 0.01 s before resistive treadmill training. The stance time had no significant deviation from the baseline when the resistance was initially applied. However, the stance time started increasing after the 100th stride of the adaptation period. When the resistance was removed unexpectedly during the post-adaptation period, the subject showed an aftereffect consisting of an increased stance time. The stance time remained above the baseline in the following strides.

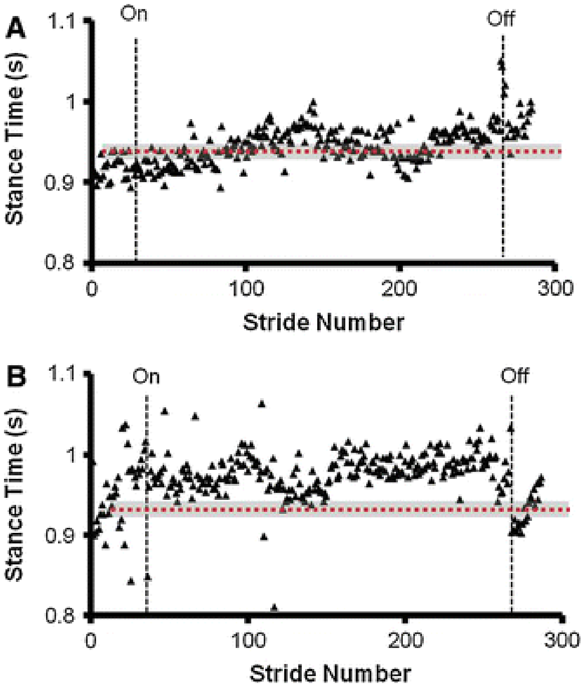

Fig. 5 a A representative subject's (S9) stance time over the last 20 strides of the pre-adaptation period (before load is on), the 250 strides of the adaptation period, and the first 20 strides of the post-adaptation period (after load is off) in resistive treadmill training. **b** The same subject's stance time in assistive treadmill training. The *red dashed horizontal line* in each plot represents the baseline value, which was the averaged stance time across the last 5 strides of the pre-adaptation period. The *gray* area represents one standard deviation above and below the baseline

The stance time of the same subject over the course of assistive treadmill training is shown in Fig. 5b. The baseline of stance time was 0.94 ± 0.03 m before assistive treadmill training. After the assistance load was applied, the stance time mostly stayed above the baseline during the adaptation period. After the assistance load was removed during the post-adaptation period, the stance time mostly stayed below the baseline.

The group data showed that the resistive treadmill training induced an aftereffect consisting of a significant increase in stance time (Fig. 6a). A paired *t* test indicated that the stance time of the first stride of the post-adaptation period following resistive treadmill training was significantly greater than that of the baseline ($P < 0.001$). In contrast, no significant changes in stance time were observed after assistive treadmill training (Fig. 6a). A paired *t* test indicated that the stance time of the first stride of the post-adaptation period following assistive treadmill training was not significantly different from that of the baseline ($P = 0.9$).

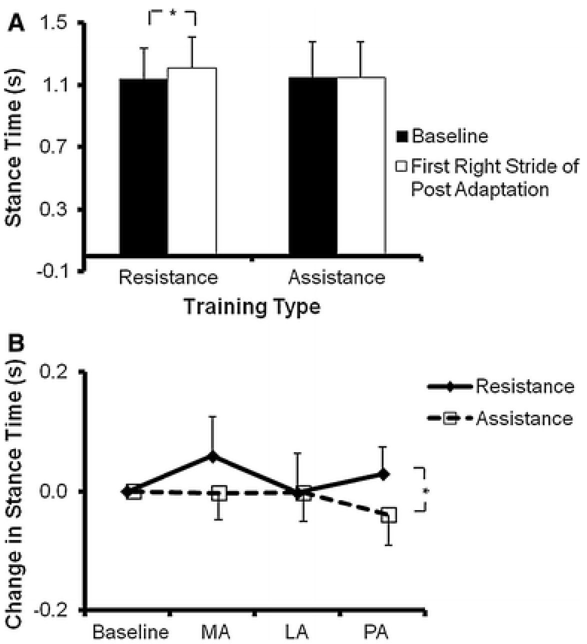

Fig. 6 a Stance time during the baseline and the first stride of the post-adaptation period in resistance and assistive treadmill training. **b** Changes in stance time from the baseline to the mid-, late-, and post-adaptation periods during resistance and assistive treadmill training. *$P < 0.05$. *MA* mid-adaptation; *LA* late adaptation; *PA* post-adaptation

On average, the stance time increased from the baseline during the mid-adaptation period, returned to the baseline during the late adaptation period, and again increased from the baseline during the post-adaptation period in resistive treadmill training (Fig. 6b). In contrast, the stance time stayed close to the baseline during the mid- and late-adaptation period and was decreased from the baseline during the post-adaptation period in assistive treadmill training (Fig. 6b). A two-way repeated-measures ANOVA indicated that the interaction between training type and training period on the change in stance time was significant ($P = 0.006$). However, the main effects of training type ($P = 0.083$) and training period ($P = 0.76$) were not significant.

Adaptation of swing time during treadmill walking

The swing time from subject S9 over the course of resistive treadmill training is shown in Fig. 7a. The baseline swing time was 0.52 ± 0.01 s before resistive treadmill training. When the resistance was initially applied, the swing time did not deviate much from the baseline. However, the swing time slightly increased after the 100th stride of the adaptation period. During the early post-adaptation period, the swing time stayed close to the baseline.

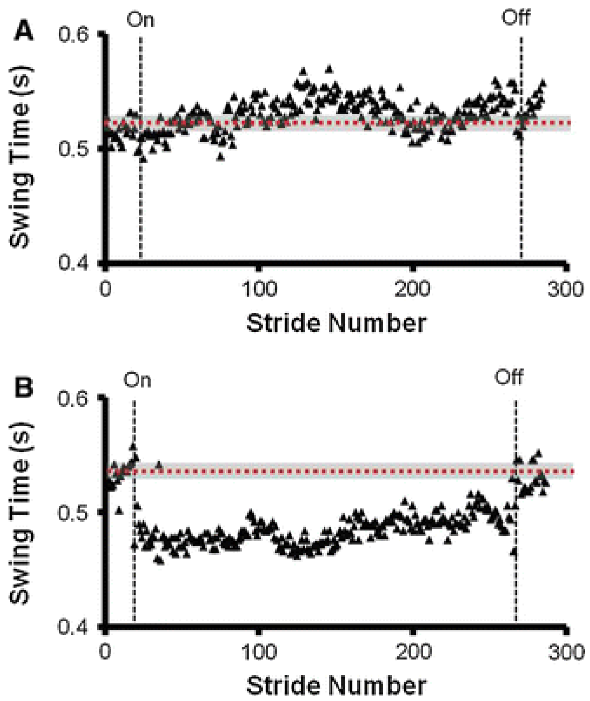

Fig. 7 a A representative subject's (S9) swing time over the last 20 strides of the pre-adaptation period (before load is on), the 250 strides of the adaptation period, and the first 20 strides of the post-adaptation period (after load is off) in resistive treadmill training. **b** The same subject's swing time in assistive treadmill training. The *red dashed horizontal line* in each plot represents the baseline value, which was the averaged swing time across the last 5 strides of the pre-adaptation period. The *gray* area represents one SD above and below the mean baseline

The swing time from the same subject over the course of assistive treadmill training is shown in Fig. 7b. The baseline stance time was 0.53 ± 0.03 m before assistive treadmill training. The swing time substantially dropped below the baseline during the adaptation period. When the assistance was removed unexpectedly, the swing time returned to the baseline in the following strides, i.e., no aftereffect.

The group data showed that neither resistive nor assistive treadmill training induced a significant aftereffect in swing time (Fig. 8a). A Wilcoxon signed-rank test indicated that the swing time of the first stride of the post-adaptation period following resistive treadmill training was not significantly different from that of the baseline (*P* = 0.12). A paired *t* test showed that the swing time of the first stride of the post-adaptation period following assistive treadmill training was not significantly different from that of the baseline (*P* = 0.11).

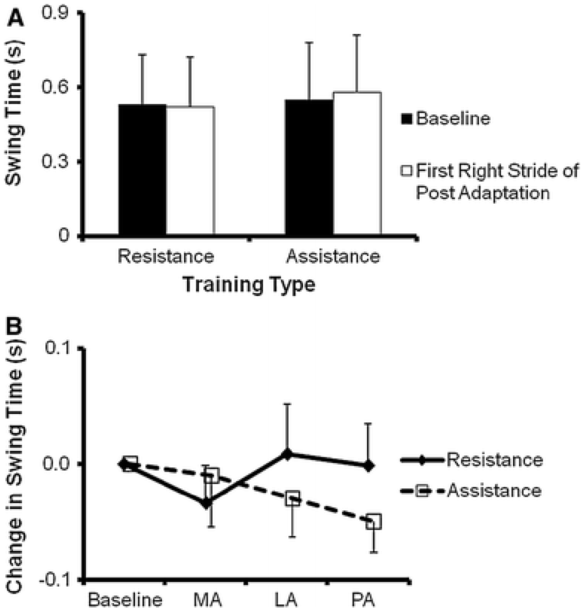

Fig. 8 a Swing time during the baseline and the first stride of the post-adaptation period in resistance and assistive treadmill training. **b** Changes in swing time from the baseline to the mid-, late-, and post-adaptation periods during resistance and assistive treadmill training. *MA* mid-adaptation; *LA* late adaptation; *PA* post-adaptation

On average, the swing time decreased from the baseline during the mid-adaptation period and increased from the baseline during the late- and post-adaptation periods in the resistance condition (Fig. 8b). On the other hand, the swing time was trending downwards from the mid-, late-, to the post-adaptation period in the assistance condition. The two-way repeated-measures ANOVA detected a significant interaction effect ($P = 0.049$). However, it did not detect significant main effects of training type ($P = 0.37$) and training period ($P = 0.53$).

Spatial and temporal parameters during overground walking

On average, stride length during overground walking increased significantly from 1.05 ± 0.20 to 1.12 ± 0.22 m after resistive treadmill training ($P = 0.005$, paired *t* test, Fig. 9a). In contrast, there were no significant changes in stride length after assistive treadmill training (1.11 ± 0.23 vs. 1.10 ± 0.26 m, $P = 0.816$, paired *t* test, Fig. 9a). There was a significant increase in overground walking speed after resistive treadmill training from 0.70 ± 0.23 to 0.78 ± 0.25 m/s ($P = 0.01$, Wilcoxon signed-rank test, Fig. 9b). In contrast, there was no significant change in walking speed after assistive treadmill training ($P = 0.197$, paired *t* test, Fig. 9b), although the averaged values decreased from 0.76 ± 0.25 to 0 0.72 ± 0.24 m/s.

Fig. 9 a Stride length during overground walking before and after resistance and assistive treadmill training. **b** Overground walking speed before and after resistance and assistive treadmill training. $^{*}P < 0.05$

Stance time and swing time during overground walking had no substantial change after either type of training. On average, stance time was slightly decreased from 1.18 ± 0.57 to 1.15 ± 0.54 s after resistive treadmill training, although a Wilcoxon signed-rank test suggested that the change was not significant ($P = 0.2$). Similarly, stance time was slightly decreased from 1.17 ± 0.57 to 1.15 ± 0.52 s after assistive treadmill training. A Wilcoxon signed-rank test also suggested that the change was not significant ($P = 0.68$). On the other hand, swing time during overground walking was slightly decreased from 0.47 ± 0.07 to 0.46 ± 0.07 s after resistive treadmill training. A paired *t* test indicated that the change was not significant ($P = 0.36$). Also, the swing time during overground walking was slightly decreased from 0.48 ± 0.07 to 0.47 ± 0.08 s after assistive treadmill training. A Wilcoxon signed-rank test suggested that the change was not significant ($P = 0.39$).

Discussion

The results of this study showed that subjects with incomplete SCI adapted to resistance/assistance loads applied to the leg at the knee. Following load release, subjects demonstrated aftereffects consisting of a longer stride length following resistive treadmill training and demonstrated an aftereffect consisting of a shorter stride length following assistive treadmill training. These findings were consistent with previous findings—force perturbations can induce aftereffects in the direction opposite to the perturbation (Emken and Reinkensmeyer 2005; Houldin et al. 2011; Noble and Prentice 2006). The overground stride length and walking speed increased after resistive treadmill training, but had no significant changes after assistive treadmill training. These findings suggested that a transfer of locomotor adaptation to overground walking occurred after resistive treadmill training but not after assistive treadmill training. On the other hand, the stance time during overground walking did not change significantly after resistive treadmill training although the training induced an aftereffect of increased stance time. The finding suggests that spatial (stride length) and temporal (stance time) gait parameters may be controlled differently during transfer.

Why transfer occurred following resistive but not assistive treadmill training

The resistance load applied to the leg may have increased subjects' active involvement during treadmill training. Subjects needed to make a greater effort to take a stride in order to counterbalance the resistance load applied to the leg. A previous study has suggested that both neural descending drive and alpha motoneuron excitability is increased while people are moving against resistance (Aagaard et al. 2002). It has also been suggested that enhanced neural descending drive is associated with development in motor memory, which is essential to motor skill acquisition (Kaelin-Lang et al. 2005). The findings from current study suggest that an increase in active involvement in resistive treadmill training may facilitate transfer of motor adaptation in patients with incomplete SCI.

In contrast, the assistance load applied to the leg during treadmill training may have reduced the subjects' effort during training, which may decrease the voluntary drive and have a negative impact on the transfer. For instance, we observed a decrease in stride length following assistive treadmill training (Fig. 4a). The finding suggested that subjects anticipated the assistance load to help their leg swing and made less effort during training. The motor control system may reduce the effort to optimize energy cost when there is a force assisting leg swing (Emken et al. 2007). Reduced effort may be a reason why previous studies observed modest improvement in overground walking ability following long-term robotic-assisted training (Field-Fote and Roach 2011).

Different responses in spatial and temporal gait parameters

While the aftereffect in stride length transferred to overground walking following resistive treadmill training, the aftereffect in stance time did not. The timing control is somewhat different between treadmill walking and overground walking. Specifically, the timing of treadmill walking is constrained by the treadmill belt speed, but such a constraint does not exist in overground walking. It has been suggested that transfer is more likely to occur if the tasks are more similar (Schmidt and Lee 2011). The dissimilarity between the timing control of treadmill walking and overground walking may partially explain why the aftereffect of increased stance time following resistive treadmill training did not transfer.

The results also suggest that spatial and temporal gait parameters may be controlled differently in locomotor adaptation, which is also consistent with previous studies. For example, the ability to adapt spatial gait parameters and the ability to adapt temporal gait parameters mature at different ages in healthy people (Vasudevan et al. 2011). In addition, healthy people adapt temporal gait parameters two times faster than that they adapt spatial gait parameters when walking on a split-belt treadmill (Malone and Bastian 2010). These findings collectively imply that different adaptive approaches may be needed to address spatial and temporal gait parameters. For example, resistive treadmill training may be more effective in addressing spatial than temporal gait parameters in patients with incomplete SCI.

Clinical implication

Patients with incomplete SCI usually demonstrate an abnormal gait pattern including decreases in stride length and speed (Amatachaya et al. 2009; Krawetz and Nance 1996). The results of this study suggest that resistive treadmill training may have potential clinical application for gait restoration. For example, the subjects in this study had an average stride length of 1.05 ± 0.2 m during overground walking before resistive treadmill training, which was much shorter than that of the age-matched healthy people (approximately 1.6 ± 0.1 m) (Murray et al. 1964). After the resistive treadmill training, the subjects' stride length was increased by 0.07 m, and the increase in stride length was accompanied with an increase in walking speed (by 0.08 m/s). While only modest improvements in gait speed and stride length were obtained after one training session in this current study, the improvement may accumulate following prolonged training. A long-term treadmill training study with a controlled resistance load applied to the leg in patients with incomplete SCI is ongoing.

Limitation of the study

There are several limitations in this study. First, we only tested the right leg for all subjects and did not examine whether the right leg was more affected than the left leg in terms of sensorimotor function (e.g., muscle strength and kinesthesia). Thus, we were not able to determine whether the adaptive strategies were different for the stronger and weaker legs. Second, we determined the amount of assistance/resistance based on subjects' self-report, which was subjective. Third, we only measured the spatial and temporal gait parameters from the right leg during treadmill training. How the subjects with incomplete SCI adapted their gait symmetry to the unilateral load was unknown. A previous study has shown that unilateral load changed healthy people's gait symmetry during and after adaptation (Noble and Prentice 2006). Thus, it is likely that there was a change in gait symmetry in our subjects. Finally, the level of injury was varied across subjects. However, we were not able to determine the relationship between motor adaptation strategy and the level of injury due to a small sample size.

Conclusion

The results of this study showed that patients with incomplete SCI transferred locomotor adaptation to overground walking following resistive treadmill training but not following assistive treadmill training. The resistance load applied to the leg may enhance patients' active involvement during treadmill training and facilitate transfer. This suggests that applying a controlled resistance load during treadmill training may be used as an adjunct approach to improve the locomotor function in patients with incomplete SCI.

References

Aagaard P, Simonsen EB, Andersen JL, Magnusson P, Dyhre-Poulsen P (2002) Neural adaptation to resistance training: changes in evoked V-wave and H-reflex responses. J Appl Physiol 92:2309–2318

Amatachaya S, Keawsutthi M, Amatachaya P, Manimmanakorn N (2009) Effects of external cues on gait performance in independent ambulatory incomplete spinal cord injury patients. Spinal Cord 47:668–673

Bastian AJ (2008) Understanding sensorimotor adaptation and learning for rehabilitation. Curr Opin Neurol 21:628–633

Behrman AL, Harkema SJ (2000) Locomotor training after human spinal cord injury: a series of case studies. Phys Ther 80:688–700

Blanchette A, Bouyer LJ (2009) Timing-specific transfer of adapted muscle activity after walking in an elastic force field. J Neurophysiol 102:568–577

Colombo G, Joerg M, Schreier R, Dietz V (2000) Treadmill training of paraplegic patients using a robotic orthosis. J Rehabil Res Dev 37:693–700

Dietz V, Colombo G, Jensen L, Baumgartner L (1995) Locomotor capacity of spinal cord in paraplegic patients. Ann Neurol 37:574–582

Emken JL, Reinkensmeyer DJ (2005) Robot-enhanced motor learning: accelerating internal model formation during locomotion by transient dynamic amplification. IEEE Trans Neural Syst Rehabil Eng 13:33–39

Emken JL, Benitez R, Sideris A, Bobrow JE, Reinkensmeyer DJ (2007) Motor adaptation as a greedy optimization of error and effort. J Neurophysio 97:3997–4006

Field-Fote EC, Roach KE (2011) Influence of a locomotor training approach on walking speed and distance in people with chronic spinal cord injury: a randomized clinical trial. Phys Ther 91:48–60

Field-Fote EC, Lindley SD, Sherman AL (2005) Locomotor training approaches for individuals with spinal cord injury: a preliminary report of walking-related outcomes. J Neurol Phys Ther 29:127–137

Harkema SJ (2001) Neural plasticity after human spinal cord injury: application of locomotor training to the rehabilitation of walking. Neuroscientist 7:455–468

Houldin A, Luttin K, Lam T (2011) Locomotor adaptations and after effects to resistance during walking in individuals with spinal cord injury. J Neurophysiol 106:247–258

Kaelin-Lang A, Sawaki L, Cohen LG (2005) Role of voluntary drive in encoding an elementary motor memory. J Neurophysiol 93:1099–1103

Kawato M (1999) Internal models for motor control and trajectory planning. Curr Opin Neurobiol 9:718–727

Krawetz P, Nance P (1996) Gait analysis of spinal cord injured subjects: effects of injury level and spasticity. Arch Phys Med Rehabil 77:635–638

Lam T, Anderschitz M, Dietz V (2006) Contribution of feedback and feedforward strategies to locomotor adaptations. J Neurophysiol 95:766–773

Lotze M, Cohen LG (2006) Volition and Imagery in Neurorehabilitation. Cog Behav Neurol 19:135–140

Lotze M, Braun C, Birbaumer N, Anders S, Cohen LG (2003) Motor learning elicited by voluntary drive. Brain 126:866–872

Malone LA, Bastian AJ (2010) Thinking about walking: effects of conscious correction versus distraction on locomotor adaptation. J Neurophysiol 103:1954–1962

Murray MP, Drought AB, Kory RC (1964) Walking patterns of normal men. J Bone Joint Surg 46:335–360

Noble JW, Prentice SD (2006) Adaptation to unilateral change in lower limb mechanical properties during human walking. Exp Brain Res 169:482–495

Schmidt RA, Lee TD (2011) Motor control and learning: a behavioral emphasis. Human Kinetics, Champaign

Vasudevan EV, Torres-Oviedo G, Morton SM, Yang JF, Bastian AJ (2011) Younger is not always better: development of locomotor adaptation from childhood to adulthood. J Neurosci 31:3055–3065

Wernig A, Muller S (1992) Laufband locomotion with body weight support improved walking in persons with severe spinal cord injuries. Paraplegia 30:229–238

Wirz M, Zemon DH, Rupp R, Scheel A, Colombo G, Dietz V, Hornby TG (2005) Effectiveness of automated locomotor training in patients with chronic incomplete spinal cord injury: a multicenter trial. Arch Phys Med Rehabil 86:672–680

Wu M, Hornby G, Landry J, Roth H, Schmit B (2011) A cable-driven locomotor training system for restoration of gait in human SCI. Gait Posture 33:256–260

Zeni JA, Richards JG, Higginson JS (2008) Two simple methods for determining gait events during treadmill and overground walking using kinematic data. Gait Posture 27:710–714

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