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A cable-driven locomotor training system for restoration of gait in human SCI

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ABSTRACT

A novel cable-driven robotic locomotor training system was developed to provide compliant assistance/resistance forces to the legs during treadmill training in patients with incomplete spinal cord injury (SCI). Eleven subjects with incomplete SCI were recruited to participate in two experiments to test the feasibility of the robotic gait training system. Specifically, 10 subjects participated in one experimental session to test the characteristics of the robotic gait training system and one subject participated in repeated testing sessions over 8 weeks with the robotic device to test improvements in locomotor function. Limb kinematics were recorded in one experiment to evaluate the system characteristics of the cable-driven locomotor trainer and the overground gait speed and 6 min walking distance were evaluated at pre, 4 and 8 weeks post treadmill training of a single subject as well. The results indicated that the cable driven robotic gait training system improved the kinematic performance of the leg during treadmill walking and had no significant impact on the variability of lower leg trajectory, suggesting a high backdrivability of the cable system. In addition, results from a patient with incomplete SCI indicated that prolonged robotic gait training using the cable robot improved overground gait speed. Results from this study suggested that a cable driven robotic gait training system is effective in improving leg kinematic performance, yet allows variability of gait kinematics. Thus, it seems feasible to improve the locomotor function in human SCI using this cable driven robotic system, warranting testing with a larger group of patients.

1. Introduction

Body weight supported treadmill training (BWSTT) is a promising rehabilitation method designed to improve motor function and ambulation in people with spinal cord injury (SCI) [1–5]. During BWSTT, the patient is given body weight support through a harness and their legs are moved by physical therapists in a "kinematically correct gait" [1]. This training paradigm is task-specific, utilizes both motor and sensory pathways of the relevant neuromuscular systems and has been shown to provide significant improvements in locomotor function [6]. However, a major limitation of BWSTT is that it requires considerable involvement of a physical therapist and is labor intensive. Thus, a robotic system that controls the stepping environment is an appealing approach to BWSTT.

Several robotic systems have been developed for automating locomotor training, such as the Lokomat [7] and the Gait Trainer (GT) [8]. The Lokomat is a motorized exoskeleton that drives hip and knee motion in a fixed trajectory using four DC motors [7], but it is difficult to back-drive the Lokomat because it uses high-advantage, ball screw actuators. The GT rigidly drives the patient's feet through a stepping motion using a crank-and-rocker mechanism attached to foot platforms [8]. These robotic systems are effective in reducing therapist labor during locomotor training and increasing the total duration of training, but can show relatively limited functional gains for some patients [4,5]. The reduced effectiveness may be due to a tendency to produce less effort during fixed trajectory motion [9]. In addition, fixed-trajectory training eliminates the variability in kinematics of the lower limbs, which is thought to be critical for successful motor adaptation, as demonstrated in both animal [10] and human studies [11]. The limited degrees of freedom of the Lokomat only allow movement of the limbs in the sagittal plane, which may affect gait dynamics [12,13]. As a consequence, there is a need for a robotic system that provides a more compliant assistance and encourages active involvement of the patient during gait training.

In this study, we tested a novel, compliant, cable-driven robotic system that provides controlled forces to the legs during the early swing phase of gait. The system was designed to allow more freedom for the subjects to voluntarily move their legs during BWSTT. We tested the feasibility of using the cable-driven locomotor trainer (CaLT) for gait training in 11 subjects with incomplete SCI.

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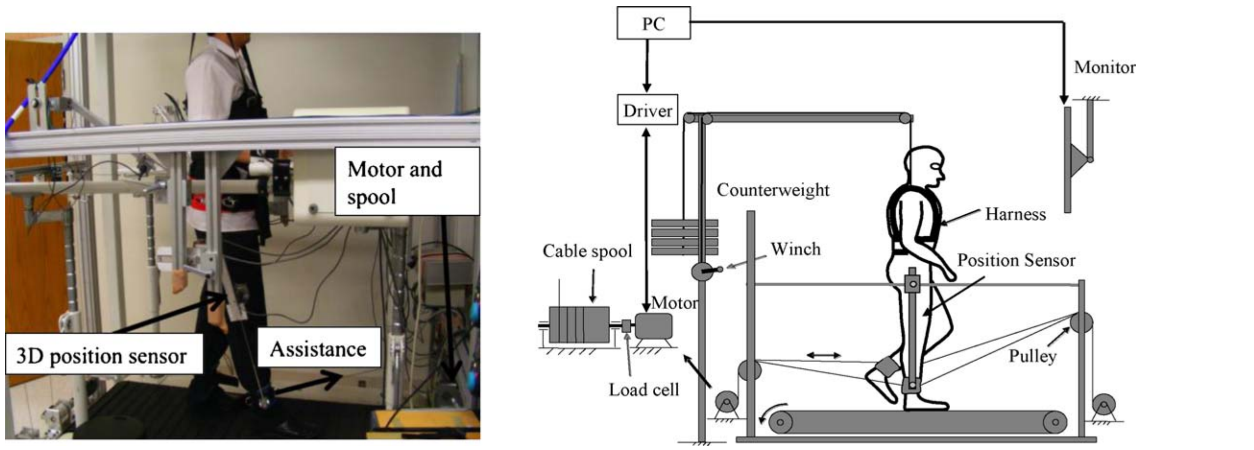

Fig. 1. This figure illustrates the cable-robot, a motor-driven cable apparatus that was used with a treadmill and body weight support system. Four cables driven by four motors, pulleys, and cable spools were used to apply resistance/assistance loads during the swing phase of walking. A personal computer was used to control the torques produced by the four motors, applying targeted resistance or assistance loads.

2. Methods

2.1. System design principle

One of the primary design goals of the CaLT was to encourage active involvement of subjects and allow freedom for subjects to control their stepping performance during treadmill training. Thus, the robotic system was designed to provide a controlled assistance or resistance force to assure a stable stepping and maximize patient effort. A secondary goal of the design was to create a system that allows step-to-step variation in leg kinematics. As a component of the design, we sought to provide a system that could be operated with minimal interference to stepping (i.e., an "off" mode). Our approach was to utilize a motorized system that is highly "backdriveable", meaning that it provides little resistance to motion when forces are not being actively applied.

2.2. Drive system

In the current design, see Fig. 1, forces were applied to the lower leg at the level of the ankles during treadmill stepping. Specifically, four nylon-coated stainless-steel cables (1.6 mm diameter), driven by four motors (AKM33H, Kollmorgen) through 4 cable spools and pulleys, were affixed to custom cuffs that were strapped to the legs around the ankles to produce up to 45 N assistance/resistance force. Four, one-degree of freedom (DOF) reaction torque load cells (TRT-200, Transducer Techniques, Temecula, CA) were integrated between the output shafts of the motors and the cable spools to record the applied torques (see Fig. 1). Bilateral ankle kinematics were measured using two custom, 3 dimensional position sensors. Each sensor consisted of a detection rod and two universal joints (U-joints) attached to the two ends of the rod. Two potentiometers (P2201, Novotechnik, Southborough, MA) and one linear position transducer (SP-2, Celesco, Chatsworth, CA) were used to measure the rotational angular and linear position of the rod. The ankle position signals were used by the operator to control the timing and magnitude of applied forces, at targeted phases of gait.

The CaLT fit well with subjects of different leg lengths because it used a single end attachment to deliver forces to the lower extremities, which also eliminated the requirement of alignment between the lower limb rotation center and the axis of rotation of a robot arm that is needed in other exoskeleton systems [7]. This approach reduced the time required for setup.

Control was implemented through a LabVIEW program, which sent control signals to the motor drives through an analog output to set the applied forces. The controller automatically adjusted the load provided by the cables based on the kinematic performance of the subject. Two control algorithms were designed for either an assistance or resistance strategy. For both algorithms, the forces were applied starting at pre-swing (10% gait cycle prior to toe off) through mid-swing of gait on the legs (the time period of hip flexion as defined by Perry [14]). For the assistance paradigm, the assistance force provided was proportional to the kinematic error during the swing phase. Specifically, the force applied to the legs was determined in real time using the following equation:

$$F_a(t) = -k_P(x(t) - x_d(t)) - k_D(\dot{x}(t) - \dot{x}_d(t)) \quad (1)$$

where t is time; k_P and k_D are the position and velocity gains (e.g., k_P and k_D were adjustable depending the tolerance of the subject, 20.8–62.5 N/m for k_P and 0.13–0.3 N/m/s for k_D in the current study); $x(t), \dot{x}(t), x_d(t)$ and $\dot{x}_d(t)$ are the measured and desired ankle horizontal position and velocity during the swing phase; the desired positions were determined from the mean recorded ankle trajectory using the position sensor for two healthy subjects walking on the treadmill. These position signals were then normalized to the mean step duration via interpolation using a cubic spline. For the resistance paradigm, a similar equation was used, but only resistance load was applied.

2.3. System characteristics of the CaLT

Subjects: Eleven subjects with a chronic (>12 months) SCI were recruited to participate in this study with 10 out of 11 participating in experiment 2 and one subject participating in experiment 3, see Appendix Table A1. All subjects had an ASIA Impairment Scale Level of C or D. The average age of subjects was 49.5 $\pm$ 8.5 years old and time post injury was 6.0 $\pm$ 6.9 years. Inclusion criteria for participation in the study included: (a) age between 16 and 65 years; (b) medically stable with medical clearance to participate; (c) level of the SCI lesion between C1-T12; (d) passive range of motion of legs within functional limits for ambulation (i.e., ankle dorsiflexion to neutral position, knee flexion from 0° to 120°, and hip to 90° flexion and 10° extension); (e) ability to ambulate overground with assistive devices as needed, and with orthotics that do not cross the knee (i.e., people with complete SCI can ambulate with knee and ankle bracing-KAFOs). Exclusion criteria included the presence of unhealed decubiti, existing infection, severe cardiovascular and pulmonary disease, concomitant central or peripheral neurological injury (e.g., traumatic head injury or peripheral nerve damage in lower limbs), history of recurrent fractures, and known orthopedic problems in the lower extremities. The Institutional Review Board of Northwestern University approved the experimental protocol and each subject signed an informed consent form prior to participating in the study.

Experimental protocol: Experiment 1 was conducted in order to test the force bandwidth of the system. Sinusoidal force commands were sent to the motor with magnitudes ranging from 25 to 45 N and frequencies ranging from 0.5 to 15 Hz while the cable was affixed to a rigid point on the frame. Specifically, the cable was preloaded with a baseline load of 35 N, then a 20 N (peak to peak) sinusoidal force was applied at frequencies of 0.5–15 Hz. The amplitude of the force signal was then acquired for analysis.

In experiment 2, 10 subjects were fitted into an overhead body weight support system through a harness with body weight support provided as needed. The treadmill speed was set at a maximum comfortable speed for each subject. At the beginning of the test, the subject was instructed to step on the treadmill for 10 steps with no cable attached, which was defined as the baseline. Then, a cable was attached to each lower leg of the subject through ankle straps (a 4 N pre-tension load was applied to avoid cable slack) while the subjects walked on the treadmill for 10 steps. Then, a controlled assistance load was applied to the legs while the subjects walked on the treadmill for another 10 steps. The experimenter adjusted the position and velocity gains to gradually increase the assistance load until the subjects felt comfortable.

In order to test the feasibility of the CaLT to improve locomotor function, a subject with incomplete SCI was recruited to participate in an 8 weeks training protocol for experiment 3. An overhead harness was used for protection only with no body weight support provided. At the initiation of the locomotor training, the controlled load was applied to the legs at the ankle. For weeks 1–4, a controlled assistance load was applied during robot-assisted treadmill walking. For weeks 5–8, a controlled resistance load was applied. The amount of the load was determined based on the motor performance of the subject with peak values of 20–33 N for assistance training and 9–20 N for resistance training, respectively. The maximal comfortable treadmill speed was used for treadmill training, which increased from 0.78 m/s to 1.22 m/s. The total training time for each training session was 45 min, excluding set up.

2.4. Data acquisition

For experiment 1, torque signals from the motor loadcells were recorded for the sinusoidal torque signals ranging from 0.5 to 15 Hz. For experiment 2, kinematics of the right leg were recorded for all test conditions using a custom designed 3 dimensional position sensor. All signals were sampled at 500 Hz using a data acquisition board (National Instruments, Austin, TX) on a PC with custom LabVIEW (National Instruments) software. For experiment 3, overground gait speed [15] and 6 min walking distance [16] were evaluated before, and after 4 and 8 weeks of training. The GaitMat II (Equitest, Chalfont, PA) was also used to measure spatiotemporal patterns of walking pre and post training.

2.5. Data analysis

All the load cell and kinematic data were analyzed using custom software written in Matlab (Mathworks, Inc., Natick, MA, USA). The position signals of the ankle were low-pass filtered using a 4th order Butterworth filter with a cutoff frequency at 10 Hz. These data were used to segment the entire dataset into step cycles, dependent on the measured ankle position using the position sensor. Seven steps of each condition were used for analysis (data from the first two steps and the last step were removed). Due to the variability in step duration from cycle to cycle, the data from each cycle were interpolated and then averaged across 7 steps to create a mean ankle trajectory for each condition. In order to quantify the variability of the ankle trajectory, we defined the path deviation as:

$$PD = \sqrt{\left(\frac{1}{N}\sum_{i=1}^{N} SD_{x_i}\right)^2 + \left(\frac{1}{N}\sum_{i=1}^{N} SD_{y_i}\right)^2} \quad (2)$$

where $N = 101$ points for one gait cycle; SD_{x_i} and SD_{y_i} are standard deviation of each point of the ankle trajectory in the horizontal and vertical direction, respectively, across 7 steps. In addition, the vertical and horizontal peak velocity of the ankle during the swing phase of gait were identified and averaged across 7 steps for each condition. The area encompassed by the mean ankle trajectory was calculated for each condition [17]. A repeated measures ANOVA was used to compare the ankle kinematics across different test conditions with significance noted at $p < 0.05$. If the ANOVA revealed significant differences, Tukey–Kramer post-hoc tests were used to identify differences between pairs, again with significance noted at $p < 0.05$.

3. Results

The results of experiment 1 indicated that CaLT's bandwidth (−3 dB) for tracking a 20 N sinusoidal force was 10.5 Hz (see Fig. 2), a bandwidth that is approximately 5 times that of human walking [18]. Above 10.5 Hz, the frequency response of the cable system declined at an average rate of −3.29 dB/decade.

The results of experiment 2 demonstrated the backdrivability of the CaLT. The CaLT did not substantially alter stepping trajectories when attached to the leg, compared with stepping without the cable attached. The ankle trajectories in the anterior–posterior direction, from one subject with chronic SCI, are shown in Fig. 3A. The group data indicated that CaLT did not significantly affect the variability in ankle trajectory when assistance load was applied during treadmill walking. Specifically, the average path deviation of the ankle trajectory was 10.7 ± 3.4 mm for the baseline, 11.4 ± 2.7 mm with cable attached and 10.5 ± 3.0 mm with assistance force applied, as shown in Fig. 3B. There were no significant changes in the variability of ankle trajectory for different loading conditions (i.e., at baseline, with cable attached and with assistance load applied) (ANOVA, $p = 0.6$).

In addition, the results from experiment 2 indicated that the CaLT improved the leg kinematic performance of people with incomplete SCI during treadmill stepping. For instance, the peak horizontal and vertical velocity of the ankle during swing phase of gait significantly increased with controlled assistance force applied during treadmill walking in patients with SCI as shown in Fig. 4A and B (ANOVA, $p < 0.05$). Specifically, the peak horizontal velocities were 0.92 ± 0.20 m/s at the baseline, 0.92 ± 0.22 m/s with the cable attached and 1.01 ± 0.24 m/s with assistance force applied. The peak vertical velocities were 0.24 ± 0.11 m/s at baseline, 0.25 ± 0.12 m/s with the cable attached and 0.27 ± 0.11 m/s with assistance force applied. The post-hoc Tukey test indicated that the peak horizontal velocity of the ankle was significantly larger with

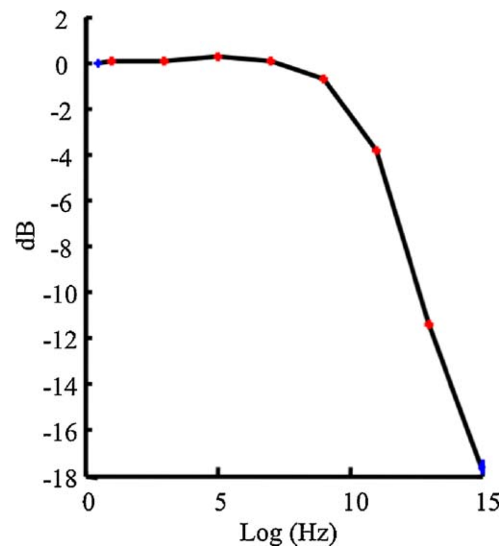

Fig. 2. Magnitude of the frequency response of CaLT for a commanded horizontal sinusoid force at 20 N. A pretension force at 35 N was applied to the cable.

assistance force compared to the baseline and the condition with the cable attached, $p < 0.05$). The peak vertical velocity of the ankle was also significantly larger with assistance force compared to the baseline (Tukey test, $p < 0.05$). In addition, the area encompassed by the ankle trajectory in the sagittal plane significantly increased with assistance load compared to the baseline and with the cable attached (Fig. 4C) (Tukey test, $p < 0.05$). The area was 123.02 ± 64.33 mm^2 at baseline, 122.39 ± 57.18 mm^2 with the cable attached and 160.20 ± 57.24 mm^2 with assistance force applied.

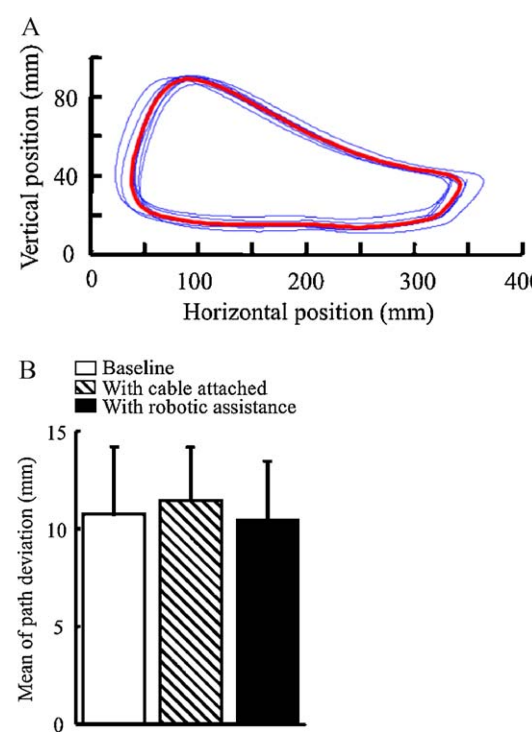

Fig. 3. (A) Ankle trajectories in the sagittal plane are shown for one subject with incomplete SCI during treadmill stepping without the attachment of the robot. The ensemble-average trajectory across 7 step cycles is shown with solid thick lines. (B) Variability of ankle trajectory for 3 different loading conditions: no attachment of cable robot, cable robot attached, and controlled assistance load applied. Path deviation of ankle trajectory in the sagittal plane for each condition was used to quantify the variability for each subject. The bar and error bar indicate the mean and standard deviation of the RMS error of ankle trajectory across subjects.

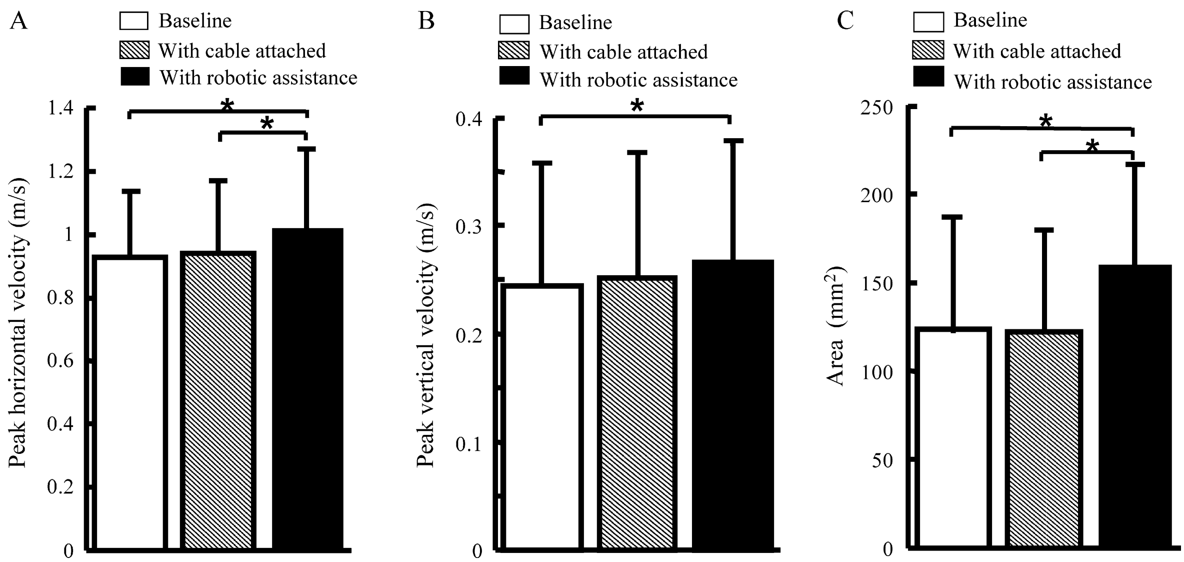

Fig. 4. Average peak forward (A) and upward (B) velocity of the ankle and area enclosed by the ankle trajectory in the sagittal plane (C) during robotic-assisted treadmill walking. Three different loading conditions, i.e., no cable robot attached (baseline), with cable robot attached, and with controlled assistance load applied, were compared. Averaged values across 7 step cycles and 10 subjects with incomplete SCI are shown. * Indicates significance, $p < 0.05$.

The results from experiment 3 demonstrated feasibility of using the CaLT to improve locomotor function in people with incomplete SCI. The overground gait speed improved following 8 weeks robot-assisted treadmill training. The self-selected and fast overground gait speed improved from 0.78 m/s to 1.06 m/s and 0.95 m/s to 1.49 m/s, respectively for this subject, Fig. 5. In addition, the 6 min walking distance improved from 312 to 397 m following 8 weeks of treadmill training.

4. Discussion

A novel cable-driven robotic gait training system was developed and tested in 11 subjects with incomplete SCI. A compliant force control strategy was used in CaLT, which allowed freedom of the subject to voluntarily move their legs outside a fixed trajectory during treadmill stepping. The cable-driven system provided limited restrictions on the variability in foot trajectory while a controlled force was applied to improve ankle kinematics, including increased horizontal and vertical ankle velocity during the swing phase of gait. In addition, results from a pilot experiment indicated that it is feasible to improve the walking function in human SCI using the CaLT.

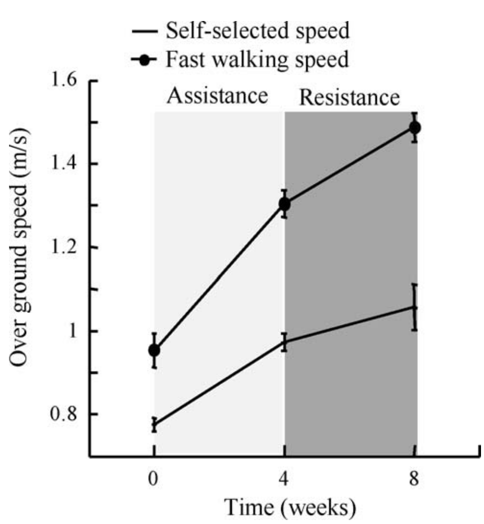

Fig. 5. Overground gait speed of one subject with incomplete SCI pre, post 4 and 8 weeks robotic-assisted treadmill training. An instrumented walkway (GaitMat II, E.Q., Inc.) was used to measure the overground gait speed. Three trials were tested and averaged for each test condition.

The key feature of the CaLT system is that the trajectory of the gait pattern is not fixed as it is for other systems. The advantage of this design approach is that leg motion is not rigidly restricted and it allows some level of variation in the kinematics of the leg, a fundamental feature of the neural control of repetitive movement such as stepping [19]. In addition, it is highly backdrivable, which has advantages over the mechanisms used for other systems, allowing patients to make and correct errors across steps. The new system uses a light weight cable drive with controlled forces applied to the legs (rather than a controlled trajectory).

The fixed trajectory strategy used in currently available robotic gait training systems may be suboptimal for improving locomotor function in human SCI. During robotic BWSTT, some robotic training devices passively move the legs in a kinematically correct pattern. This approach provides appropriate afferent input related to stepping, but the robot may essentially take over the task for the subjects, reducing the participation level [20]. Muscle activity is significantly lower during passively guided, robotic locomotor training than with physical therapist assisted treadmill training in patients with SCI [9], resulting in a modest gait speed improvement following robot-assisted treadmill training [4]. In the CaLT, assistance as needed or resistance as tolerated control strategies can be used to control the applied load. The control strategies may encourage an active involvement by the patient during robotic training, which is critical for motor learning [21].

Results from this pilot study indicate the CaLT as a promising training strategy to improve locomotor function in people with incomplete SCI. In a single subject, the self-selected overground gait speed improved 0.28 m/s (36% improvement), which is promising compared with the gait speed gain obtained using other robotic gait training systems (0.05–0.11 m/s, [4,5]). Outcome studies with larger numbers of subjects with SCI are needed to demonstrate the clinical benefit of this promising system for gait rehabilitation of patients with SCI.

5. Conclusion

The cable driven locomotor training system proposed in this study provides a promising alternative for treatment of patients with incomplete SCI through robot-assisted treadmill training. This new system is highly backdrivable, complaint, and allows freedom for patients to voluntarily move their legs during BWSTT. A pilot long term training study from one subject with incomplete SCI also indicated that it is feasible to improve the locomotor function in people with SCI using the CaLT.

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Conflict of interest: There is no conflict of interest regarding the publication of this manuscript.

Appendix A

Table A1

Table 1
Subject information indicating age, injury level, ASIA classification, years since injury, and medications the subjects were prescribed at the time of the study.

Subject	Age (year)	SCI Level	ASIA Level	Time post injury (year)	Medications
A	52	T7	D	5.5	None
B	46	C5/C6	D	25	None
C	63	T11/T12	D	1	None
D	48	C6/C7	D	2	None
E	40	C6/C7	C	2.3	Baclofen
F	51	C3/C7	D	8.8	Baclofen 10 mg/day
G	48	T4/T6	D	2.5	None
H	53	C5	D	4	Baclofen 80 mg/day
I	64	T9	D	8.9	Baclofen 20 mg/day
J	37	C4	D	4.2	None
K	43	C5/C6	D	1.3	None

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