Department of Veterans Affairs

JRRD Volume 48, Number 2, 2011
Pages 147–160
Journal of Rehabilitation Research & Development

Trainer variability during step training after spinal cord injury: Implications for robotic gait-training device design

Jose A. Galvez, PhD;[1] Amy Budovitch, PT;[2] Susan J. Harkema, PhD;[3] David J. Reinkensmeyer, PhD[1]*

[1]Departments of Mechanical and Aerospace Engineering and Biomedical Engineering, University of California at Irvine, Irvine, CA; [2]Rancho Los Amigos National Rehabilitation Center, Downey, CA; [3]Department of Neurology and Brain Research Institute, University of California at Los Angeles, Los Angeles, CA

Abstract—Robotic devices are being developed to automate repetitive aspects of walking retraining after neurological injuries, in part because they might improve the consistency and quality of training. However, it is unclear how inconsistent manual training actually is or whether stepping quality depends strongly on the trainers' manual skill. The objective of this study was to quantify trainer variability of manual skill during step training using body-weight support on a treadmill and assess factors of trainer skill. We attached a sensorized orthosis to one leg of each patient with spinal cord injury and measured the shank kinematics and forces exerted by different trainers during six training sessions. An expert trainer rated the trainers' skill level based on videotape recordings. Between-trainer force variability was substantial, about two times greater than within-trainer variability. Trainer skill rating correlated strongly with two gait features: better knee extension during stance and fewer episodes of toe dragging. Better knee extension correlated directly with larger knee horizontal assistance force, but better toe clearance did not correlate with larger ankle push-up force; rather, it correlated with better knee and hip extension. These results are useful to inform robotic gait-training design.

Key words: automation of therapy, locomotor training, neuromotor rehabilitation, physical therapy, quality of life, rehabilitation engineering, rehabilitation robotics, spinal cord injury, training consistency, walking impairment.

INTRODUCTION

Step training using manual assistance with body-weight support (BWS) on a treadmill (BWST) (**Figure 1(a)**) can help people with neurological injuries, such as stroke and spinal cord injury (SCI), improve their walking ability [1–6]. However, such training can be labor intensive, involving up to four experienced trainers. Partly in response to the labor-intensive nature of the training, commercial robotic devices, including the Gait Trainer GTI (Reha-Stim; Berlin, Germany) [7], the Lokomat (Hocoma Inc; Rockland, Massachusetts) [8], and the AutoAmbulator (HealthSouth; Birmingham, Alabama) have been developed and are being used for step training in many clinical centers. Training with these devices has been found to benefit patients [9–12], but in some cases to a lesser extent than training with human trainers [13–14]. Thus, the benefits of robotic-assisted gait training relative to human-assisted training appear to be incompletely unrealized at this time, suggesting that gait-training robotic design requires ongoing revision.

Abbreviations: BWS = body-weight support, BWST = BWS on a treadmill, CI = confidence interval, CV = coefficient of variation, RMS = root-mean-square, RMS-SD = RMS of the standard deviation, ROM = range of motion, SCI = spinal cord injury, SD = standard deviation, UCLA = University of California at Los Angeles.

***Address all correspondence to David J. Reinkensmeyer, PhD; Department of Mechanical and Aerospace Engineering, University of California at Irvine, 4200 Engineering Gateway, Irvine, CA 92697; 949-824-5218; fax: 949-824-8585. Email: dreinken@uci.edu**

DOI:10.1682/JRRD.2010.04.0067

Besides reducing labor, an often-used justification for automating gait training is that robots could make the training more consistent. However, it is currently unknown how inconsistent manual therapy actually is. It may be that trainers can learn to apply consistent patterns of force with only a small amount of training experience, in which case this is not a valid rationale for robotic devices. A related rationale for robotic gait training is that robots might be able to act like the most skilled of trainers, improving access to high-quality gait training. However, it is unclear how sensitive the elicitation of desirable gait patterns is to training skill. Again, it may be that trainers exhibit a comparable, efficacious level of skill given a reasonable amount of training experience. In this case, a robotic device would again not be a significant improvement over a reasonably experienced trainer.

Therefore, the purpose of this study was to quantify between-trainer variability in step training with BWST. In addition, we sought to determine whether perceived skill varies greatly between trainers, and if so, to identify the biomechanical correlates that account for this perception of skill. For this purpose, we developed an orthosis that measured the forces and motions applied by human trainers to the patient's lower-leg shank during BWST sessions with four patients with SCI. We used the orthosis to quantify between- and within-trainer repeatability of manual assistance forces. We also analyzed relationships between trainer skill rating, trainer forces, and key kinematic outcomes during different phases of the gait cycle. A reduced version of this work was published in two previous conference papers [15–16].

METHODS

Experimental Setup

Patients with SCI walked on a treadmill with a fraction of their weight unloaded from a pneumatic overhead support as three trainers assisted movement of the legs and hips (**Figure 1(a)**). We attached an orthosis with sensors to one of the legs (**Figure 1(b)**). This orthosis was a refined version of a previous two-hand sensor system that consisted of two 6-axis force-torque sensors attached to a custom orthopedic splint [17]. We extended it with a 6-degree-of-freedom linkage (Microscribe articulated-arm digitizer, Immersion Corp; San Jose, California) that measured the position and orientation of the patient's shank and developed braces that simulated the trainer's handholds on the knee and ankle (**Figure 1(b)**). We will refer to the forces applied to the upper handle as "forces on the knee" and to the lower handle as "forces on the ankle." Although these forces can be transformed into a single force and moment pair acting on the shank, we will work with and report the forces on the knee and ankle because they provide direct insight into the bimanual force pattern of the trainers.

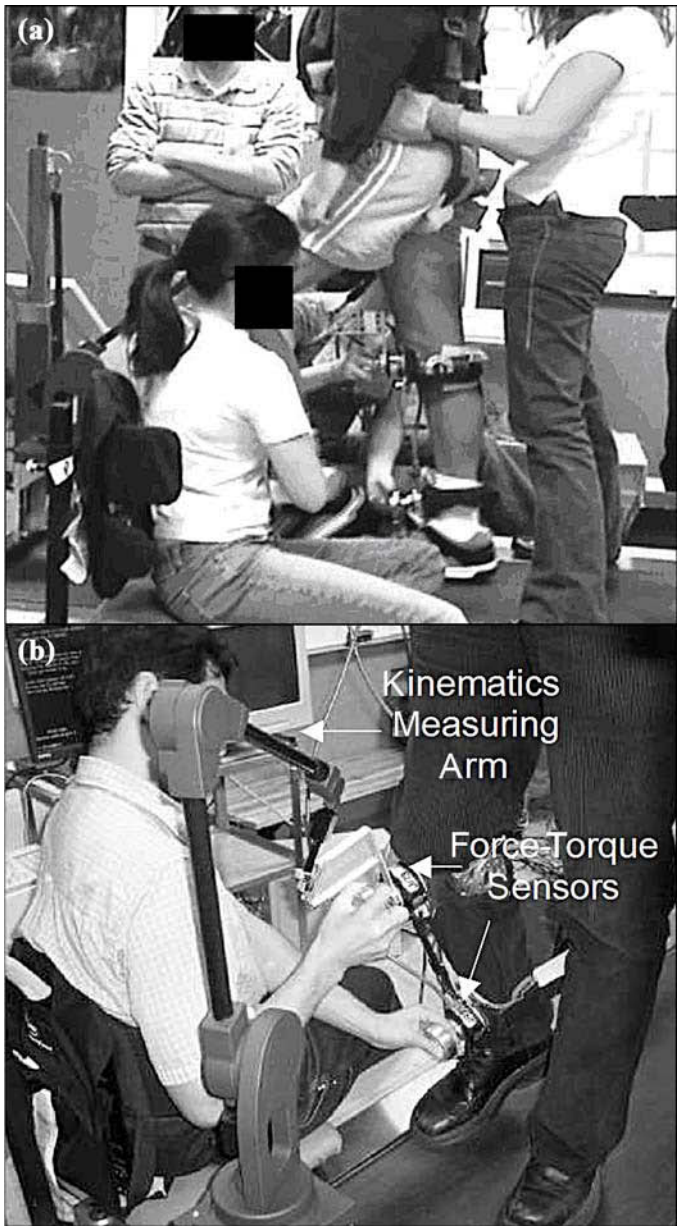

Figure 1.
Experimental setup. **(a)** Step training using body-weight support on treadmill with sensorized orthosis measuring shank motion and forces applied by trainer to patient's right leg. **(b)** Close-up view of sensor setup that measures motion and forces applied by trainers attached to right leg of nondisabled person.

Experimental Protocol for Patients with Spinal Cord Injury and Trainers

We performed six experiments with both experienced and novice trainers and with four patients with chronic, incomplete SCI. The lead trainer chose the treadmill speed and BWS and kept them constant for the experimental session. The trainer adjusted the speed to be comfortable for the patient, within the range of normal walking speeds, and adjusted the BWS to maximize limb loading without creating stance-phase knee buckling. **Figure 2(a)** illustrates the experimental sequence: the patient took about 22 steps with assistance from the trainers, the patient and trainers rested for at least half a minute, and then they started gait training again. After three such bouts, the trainers changed positions so that a different trainer assisted the sensorized leg. In the experiment shown in **Figure 2 (a)**, three different trainers rotated to assist the patient. In other experiments, four or five trainers rotated assistive positions. **Figure 2(b)** illustrates the pattern of forces and motions measured by the sensor system as the trainers assisted.

Table 1 summarizes the experiments that we performed for the study and the patients' characteristics. Six experienced and less-experienced trainers participated in six experimental sessions. The trainers ranged from 10 years of gait-training sessions several times per week to only four 30-minute sessions of experience; these less-experienced trainers had also participated in a 2-hour training program that involved practice with nondisabled subjects. A total of four patients with SCI participated in the experiments. Two of these patients participated in two sessions each, and we measured the assistance pattern of a different leg on each occasion. Therefore, we measured a total of six legs, and we will refer to these as the six experiments. For these six experiments, we had a total of 24 trainer/patient dyads to analyze, where a dyad is defined as a unique pairing of a specific trainer and a specific leg.

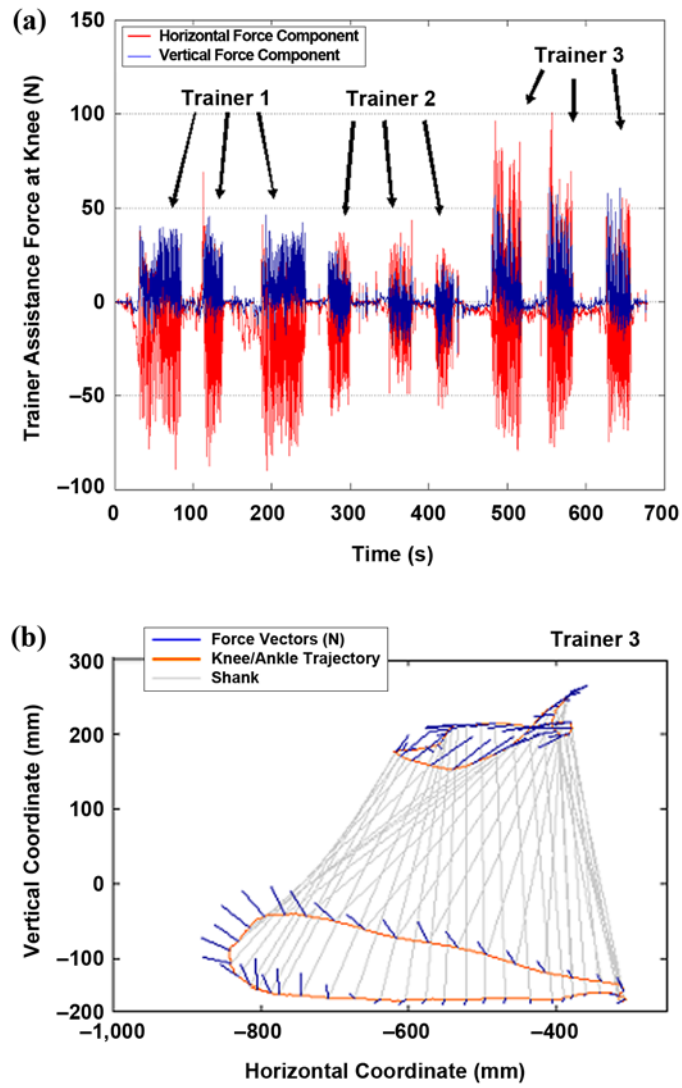

Figure 2.
Raw data illustrating experimental protocol. **(a)** Forces applied in horizontal (blue) and vertical (red) directions to knee by three different trainers during 11 minutes of locomotor training with one patient at 2 mph. Patient's impairment classification was American Spinal Injury Association level D. Patient stepped for three bouts of 20 to 30 steps for each of three trainers. **(b)** Example of knee and ankle trajectories and assistance force vectors of representative step of same patient. Treadmill was moving from right to left at 2 mph. Force vectors: 1 N = 1 mm.

Data Analysis

Trainer Skill Rating

For the six experiments, one of the more experienced trainers who had both participated in the experiments and trained the less-experienced trainers served as a rater. She viewed the videotapes of the training sessions, scoring each trainer's performance in eliciting stepping in each bout on a scale from 0.0 to 10.0, where 0.0 = poor and 10.0 = excellent. This trainer had 3 years of intensive step-training experience at the University of California at Los Angeles (UCLA) Locomotion Laboratory (Los Angeles, California). She did not rate several bouts in which the view was obstructed on the videotape, but did rate at

Table 1.
Experiment parameters and patient characteristics.

Experiment	Patient	Age (yr)	Injury Grade	Time Since Injury (yr)	Level of Injury	Body Weight (kg)	BWS (%)	Treadmill Speed (m/s)	No. of Trainers
SCI-D1	1	62	ASIA C	18	C7	76	29	0.80	3
SCI-CD1	2 (left leg)	36	ASIA C and D*	5	C4 (sensory) C6 (motor)	92	66	0.85	5
SCI-CD2	2 (right leg)	36	ASIA C and D*	5	C4 (sensory) C6 (motor)	92	53	0.85	3
SCI-B1	3	44	ASIA B	3	C5–C6	72	75	0.89	5
SCI-B2	4 (left leg)	24	ASIA B	3	C7	94	71	0.80	4
SCI-B3	4 (right leg)	24	ASIA B	3	C7	94	67	0.76	4

Note: Grade according to ASIA impairment scale.
*Subject classified as unsure between ASIA C and ASIA D.
ASIA = American Spinal Injury Association, BWS = body-weight support, C = cervical.

least one bout for all trainer/patient dyads. For each trainer/patient dyad, we averaged the ratings across the total number of stepping bouts. For each bout, the rater also commented on what she perceived the trainers were doing well or poorly. Beyond this subjective analysis, we blinded the rater to the data analysis. In the rest of the article, we refer to this rating as "trainer skill rating."

Force and Kinematic Data

We recorded the sagittal plane force at 1,000 Hz and the kinematic data at 500 Hz and extracted key variables of interest from this data. For each bout, the treadmill belt took about three patient steps to accelerate to the target speed; these steps were not analyzed, leaving about 54 steps for each trainer/patient dyad, generated in three bouts of an average of 18 steps. For one of the 24 trainer/patient dyads, only one bout was performed; for three dyads, just two bouts were performed.

We obtained velocity and acceleration of the orthosis by differencing and smoothing with a zero-phase low-pass filter consisting of a single-pole recursive filter with a cut-off frequency of 10 Hz cascaded four times forward and backward.

We time-normalized data by first determining initial foot contact for each step using the time of maximum ankle vertical acceleration (caused by impact of the foot with the ground) at the end of swing. We then expressed time as a percentage of the stride duration. A step began at initial foot contact (0% of the gait cycle) and ended at foot contact of the subsequent step (100% of the gait cycle). We averaged force and kinematic data and calculated their standard deviation (SD) across steps every 1 percent of the stride.

We estimated shank angle and knee and ankle positions from the orthosis kinematic measurements. We calculated toe position in early swing (used for estimating toe dragging) using the ankle angle at early swing measured from the videotape of each subject.

Measures

For the manual force data, we focused on the force at two key time periods during the gait cycle. We defined the knee extension force as the force applied in the backward, horizontal direction against the knee during midstance, defined as 10 to 30 percent of the stride. We defined the ankle push-up force as the force applied in the vertical direction to the lower shank at the ankle during the initial 15 percent of swing. We defined the beginning of swing as the time when the ankle horizontal position reached its minimum. We selected these two measures, knee extension force and ankle push-up force, because they capture manual assistance that is related to two critical aspects of the gait cycle difficult for people with gait impairment after SCI to achieve on their own: adequate knee-extension during stance and adequate toe clearance during swing. Note that we chose the coordinate frame for describing these forces such that backward forces are negative and upward forces are positive.

For the kinematics data, we quantified knee extension using the angle of the shank with respect to the horizontal during midstance (10%–30% of the stride). To quantify a variable related to toe dragging, we measured mean toe height in the initial 15 percent of swing, when the risk of

toe dragging was highest. We obtained the horizontal knee position range of motion (ROM) by subtracting the minimum from the maximum mean horizontal knee position.

We calculated the between-step (i.e., within-trainer) variability of key measures over the portions of interest of the stride, using the root-mean-square (RMS) of the SD (RMS-SD) [18]. We used this measure to estimate the between-step variability of knee extension force, ankle push-up force, shank angle during midstance, and toe height during early swing. Specifically, we considered a walking bout with n steps, each step cycle with 100 samples (1%–100% of the gait cycle as explained earlier). For each of the 100 points, we first calculated the SD across n steps. Then, to obtain an average of the variability over 10 to 30 percent of the gait cycle (in the case of the knee extension force measure), we calculated the RMS-SDs over samples 11 to 30 percent. This process yields the square root of the arithmetic mean of the variance, averaged across samples of the relevant period of the gait cycle.

We determined a coefficient of variation (CV) to assay between-step variability of different measures. We defined the CV as the ratio of the RMS-SD and the mean of the absolute values of the means over the selected stride portion [19].

We calculated the between-trainer variability of key measures by taking the SD of the measure of interest across the trainer-leg dyads of each experiment. We then averaged the SDs across all six experiments, once again using the RMS operation.

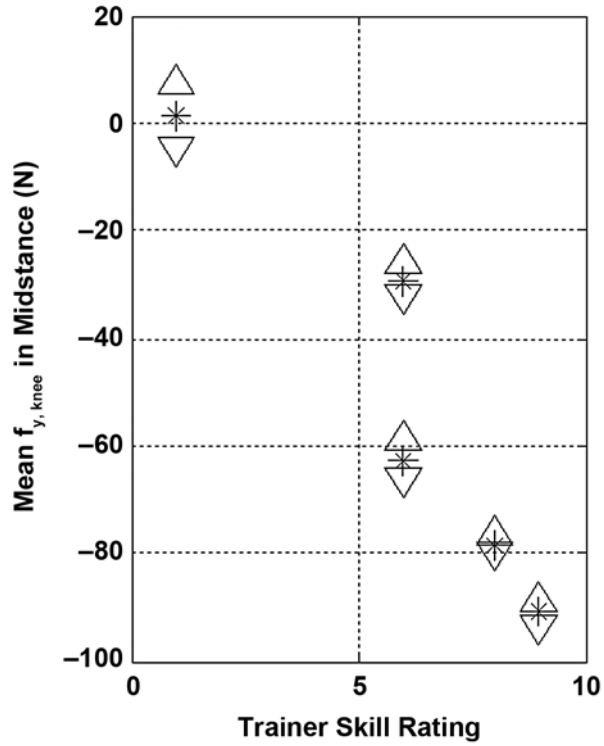

Figure 3.
Example of statistical analyses. Plot shows knee extension force during midstance for each trainer versus trainer skill rating in training session for leg of patient with spinal cord injury. From data such as these, we calculated correlation coefficient and significance of correlation. Triangles denote confidence intervals at $p < 0.05$ for means of knee extension force ($f_{y, knee}$) for each trainer (for three bouts of 20–30 steps each). We noted significant differences between pairs of trainers when such confidence intervals did not overlap.

Statistical Analysis

As an assessment of whether different trainers assisted differently for the same patient, we calculated 95 percent confidence intervals (CIs) for each measure, then used the intervals to identify significant differences between two different trainers for the same leg in the same experimental session (**Figure 3**). We obtained the CIs for the interstride SDs at each 1 percent stride interval by using the one-sample chi-square statistic and the CIs for the RMS-SDs by averaging the upper and lower bounds of the SDs' CIs. We obtained the CIs for the CVs by dividing the bounds of the RMS-SDs' CIs by the means. For each experiment, we then checked whether the parameters for each possible pair of two different trainers assisting on the same leg were significantly different by finding out, pairwise, whether the 95 percent CIs overlapped. The number of possible pairwise comparisons across all six experiments was 38, since an experiment with n trainers gave rise to $(n - 1)!$ possible pairwise comparisons. Specifically, we had 2 experiments with 5 trainers, each one giving rise to $4 + 3 + 2 + 1 = 10$ possible pairwise comparisons; 2 experiments with 4 trainers, each one giving rise to $3 + 2 + 1 = 6$ possible pairwise comparisons; and 2 experiments with 3 trainers, each one giving rise to $2 + 1 = 3$ possible pairwise comparisons. If the number of comparisons that showed significant differences was high, we took this to indicate that the differences between trainers were large enough and that the experiment had a large enough sample size (number of steps) to allow us to infer that different parameters were consistently different from each other for different trainers.

Within each experiment, we tested for correlations of the different parameters with trainer ratings and among the parameters (e.g., **Figure 3**), obtaining the p-value of each correlation. The sample size of each experiment was small (between 3 and 5 trainers in each experiment), but we

gained statistical power by using the Stouffer method to calculate a combined *p*-value that aggregated the evidence from all experiments [20]. Among the many methods that have been developed to combine the *p*-values from different experiments, the Stouffer method was the most appropriate for our data because it equally emphasizes all sizes of *p*-values [20] and combines a series of outcomes that can go in positive and negative directions so that relative evidence in one direction cancels relative evidence in the other direction [21]. The *p*-values are one-tailed, as recommended in the literature when *p*-values are to be combined (e.g., Whitlock [22]), and because it makes sense since both negative and positive correlations are possible so that outcomes are directional.

RESULTS

Forces Applied by Same Trainer Were Repeatable Across Steps

Figure 4 shows the horizontal forces exerted on the knee by two different trainers during 54 steps with the same patient with SCI and illustrate the low within-trainer variability of the forces across steps. Within-trainer variability was similar to those shown in **Figure 4** with other patient/trainer dyads. As a baseline for comparing between-trainer variability, we analyzed the within-trainer variability for two key features of the step cycle: knee extension force during midstance and push-up force at the ankle during initial swing. The SDs for these two measures were 16.7 N and 11.4 N, averaged across trainers and experiments (i.e., legs) using the RMS-SD technique described in the "Methods" section. For comparison, the mean values of these measures were –61.5 N and 34.2 N. Thus, variability (defined as 1 SD) was about 30 percent of the mean value of these measures on average for individual trainers.

Forces Applied by Different Trainers Differed Substantially

We next quantified the variability in the two key force measures between trainers. The average between-trainer SD was 40.3 N for the horizontal knee force through midstance and 19.0 N for the push-up ankle force during initial swing. Thus, for the horizontal knee force through midstance, the typical variability in force magnitudes between trainers was 2.4 times larger than the typical variability for a single trainer for step to step, a significant difference ($p = 0.02$, *t*-test). For the ankle push-up force, the average between-trainer variability was 1.7 times larger

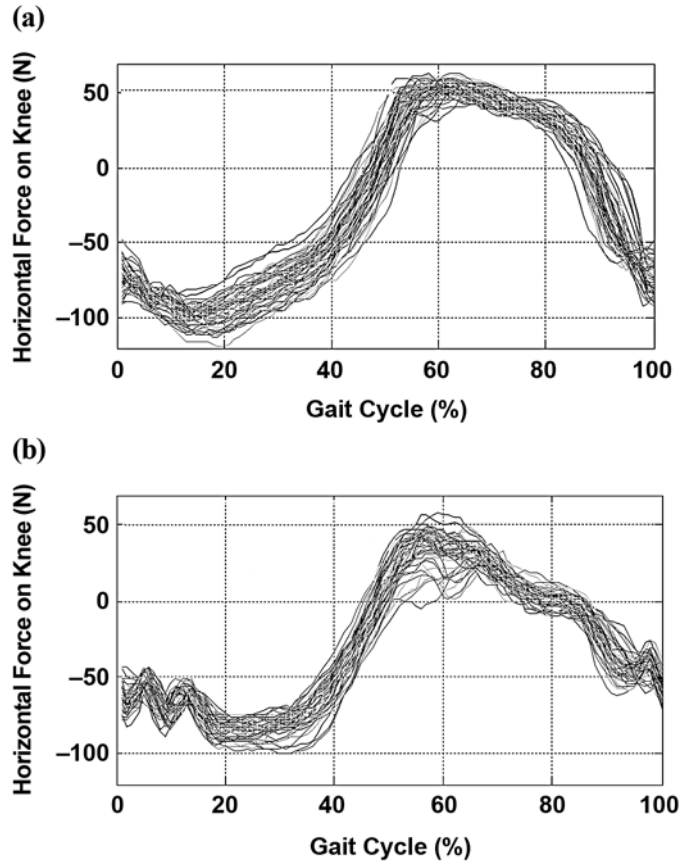

Figure 4.
Horizontal trainer force exerted on knee by two different trainers on leg of patient with spinal cord injury (American Spinal Injury Association level B) during the same session for 54 steps. Patient weight was 73 kg, body-weight support was 75%, and speed was 0.9 m/s. Initial foot contact occurs at 0% of gait cycle. **(a)** Trainer rated 9.0 out of 10.0. **(b)** Trainer rated 8.0 out of 10.0.

than the within-trainer variability across steps, but this difference was not significant ($p = 0.55$).

We also statistically assessed the presence of differences between trainers by performing pairwise comparisons of several measures between all possible pairs of trainers within each experiment ($n = 38$) using a CI approach (**Table 2**). The number of pairwise differences was high relative to the number of possible significant differences, indicating that different trainers commonly assisted in statistically detectable different ways on the same leg.

Even Trainers with Higher Skill Ratings Assisted Differently

We questioned whether the large between-trainer variability (relative to within-trainer variability) and frequent

Table 2.
Summary of statistical comparisons between two trainers assisting on same leg.

Measure	Statistical Comparison
Knee Extension Force During Stance	35
Shank Angle During Stance	33
Push-Up Force at Ankle in Early Swing	27
Toe Clearance in Early Swing	24
Horizontal Range of Motion of Knee	31

Note: Number of pairwise comparisons that were different between trainers out of 38, $p < 0.05$.

significant differences between trainers could be due to the large variation in experience between the trainers who participated in the experiment. We therefore separately analyzed the trainers who were rated as most skilled, which were a group of three of the six total who received an average rating across all six experiments >7.0 (average ratings were 7.1, 7.4, and 8.5 out of 10.0, compared with the other three trainers rated 5.2, 5.0, and 3.0 out of 10.0). The between-trainer SDs for the three trainers with higher skill ratings was 31.4 N and 20.7 N for knee extension force and ankle push-up force, which were not significantly different from the SDs calculated across all trainers (i.e., 40.3 N and 19.0 N).

Trainers with Higher Skill Ratings Elicited Better Leg Extension and Toe Clearance

As stated earlier, the average subjective rating of trainer skill varied from 3.0 to 8.5 on a scale of 1.0 to 10.0, and from 5.2 to 8.5 when considering only the experienced trainers (trained in the same program). We therefore sought to determine which biomechanical features of the training pattern accounted for this variability in subjective rating.

In all six experiments, the expert rater commented that leg extension in stance was less than desired from some trainers. Indeed, there was a significant positive correlation between trainer skill rating and shank angle during stance (**Table 3**). Also, more-skilled trainers exerted significantly larger (more negative) forces on the knee (**Table 3**). Larger knee-extension force was strongly correlated to better knee extension (**Table 4**). **Figure 5** shows an example of incomplete knee extension during training.

Table 3.
Correlations between trainer skill rating (TSR) and extension-related variables (measured during mid stance). Extension force variability is between-step root-mean-square of the standard deviation of extension force during midstance. Extension force coefficient of variation (CV) normalizes extension force variability by force magnitude. Symbols $p+$ and $p-$ refer to p-values for positive and negative correlations, respectively, and are given only when different experiments exhibited positive and negative correlations for given relationship.

Relationship	Experiment						Combined p-Value
	SCI-D1	SCI-CD1	SCI-B1	SCI-B2	SCI-CD2	SCI-B3	
TSR and Shank Angle							
r	0.79	0.86	0.93	0.78	0.96	0.96	—
p	0.21	0.03	0.01	0.11	0.08	0.02	<0.001
TSR and Extension Force							
r	–0.92	–0.90	–0.94	–0.87	–0.91	–0.64	—
p	0.13	0.02	0.01	0.07	0.13	0.18	<0.001
TSR and Extension Force Variability							
r	0.57	0.44	–0.66	0.27	–0.55	0.13	—
$p+$	0.31	0.23	0.89	0.37	0.63	0.44	0.47
$p-$	0.69	0.77	0.11	0.63	0.37	0.56	0.53
TSR and Extension Force CV*							
r	–0.99	–0.67	–0.93	–0.90	–0.98	–0.75	—
p	0.04	0.11	0.01	0.05	0.06	0.13	<0.001
TSR and Shank Angle Variability*							
r	–0.31	–0.73	–0.17	0.89	–0.99	–0.96	—
$p+$	0.60	0.92	0.61	0.06	0.97	0.98	0.96
$p-$	0.40	0.08	0.39	0.94	0.03	0.02	0.04

*Significant relationship according to combined p-value calculated using Stouffer method.

Table 4.
Correlations between various forces and kinematic measures. Symbols *p*+ and *p*– refer to *p*-values for positive and negative correlations, respectively, and are given only when different experiments exhibited positive and negative correlations for given relationship. Shank angle during terminal stance was measured during 30% to 50% of stride.

Relationship	Experiment						Combined
	SCI-D1	SCI-CD1	SCI-B1	SCI-B2	SCI-CD2	SCI-B3	*p*-Value
Extension Force and Shank Angle During Stance*							
r	–0.48	–0.98	–0.93	–0.98	–0.78	–0.44	—
p	0.34	0.001	0.01	0.012	0.22	0.23	<0.001
Push-Up Force and Toe Clearance							
r	0.70	0.46	–0.86	0.69	0.80	–0.09	—
p+	0.25	0.22	0.97	0.16	0.20	0.55	0.30
p–	0.75	0.78	0.03	0.84	0.80	0.45	0.70
Horizontal ROM of Knee and Toe Clearance							
r	1.00	0.72	0.73	0.96	–0.77	0.83	—
p+	0.02	0.09	0.08	0.02	0.56	0.09	<0.001
p–	0.98	0.91	0.92	0.98	0.44	0.91	>0.99
Shank Angle During Terminal Stance and Toe Clearance During Swing*							
r	0.25	0.29	0.93	0.74	0.82	0.94	—
p	0.42	0.32	0.01	0.13	0.25	0.03	0.003

*Significant relationship according to combined *p*-value calculated using Stouffer method.
ROM = range of motion.

The less-skilled trainers also had more problems with a lower toe height during initial swing and thus toe dragging (**Table 5**). One way to prevent toe drag would be to push up with a larger force on the ankle during initial swing. For the bouts in **Figure 6**, the trainer who did not have problems with toe dragging (rated 9.5/10.0, dotted line) exerted a larger upward force on the ankle in initial swing. However, this relation was not generally observed in the other experiments. Correlations were disparate and not significant for trainer skill rating and ankle push-up forces (**Table 5**) or for mean push-up force and mean toe clearance (**Table 4**).

If the obvious mechanical action of pushing upward on the ankle at the beginning of swing did not usually account for improved toe clearance, then what did? Ipsilateral-stance hip-joint kinematics are important for swing initiation [23–25]. The sensors did not measure hip-joint angle directly but did measure horizontal ROM of the knee, which is dictated by the range of hip flexion and extension. Horizontal ROM of the knee correlated with toe clearance (**Table 4**, $p < 0.001$). Toe clearance in initial swing was also correlated with a good terminal extension in stance of the same leg (measured during 30%–50% of the stride, **Table 4**, $p = 0.003$).

Within-Trainer Variabilities Decreased with Trainer Skill Rating

Research using animal models of step training have suggested that variability in the assistance pattern is a desirable feature for enhancing spinal plasticity [26]. We examined whether trainers who were rated as more skilled had lower (or greater) between-step variability in the assistance pattern. The correlations between trainer ratings and extension-force variability and trainer rating and push-up force variability were disparate and not significant, as seen in **Table 3**. However, when the magnitude of the forces was taken into account, we found that more skilled trainers had smaller CVs (i.e., variability divided by magnitude) in the extension force, with negative correlations for all six experiments, as seen in **Table 3** (pooled $p < 0.001$). The within-trainer CV of the extension force averaged (RMS) for all six trainers was 27 percent, significantly different from the within-trainer CV averaged for the three higher-rated trainers, which was 18 percent. The within-trainer CV averaged for the three lower-rated trainers was 34 percent. As for the CV of ankle push-up forces at initial swing, we found no significant differences between higher-rated and lower-rated trainers. The push-up force CV averaged for the six trainers was 33 percent.

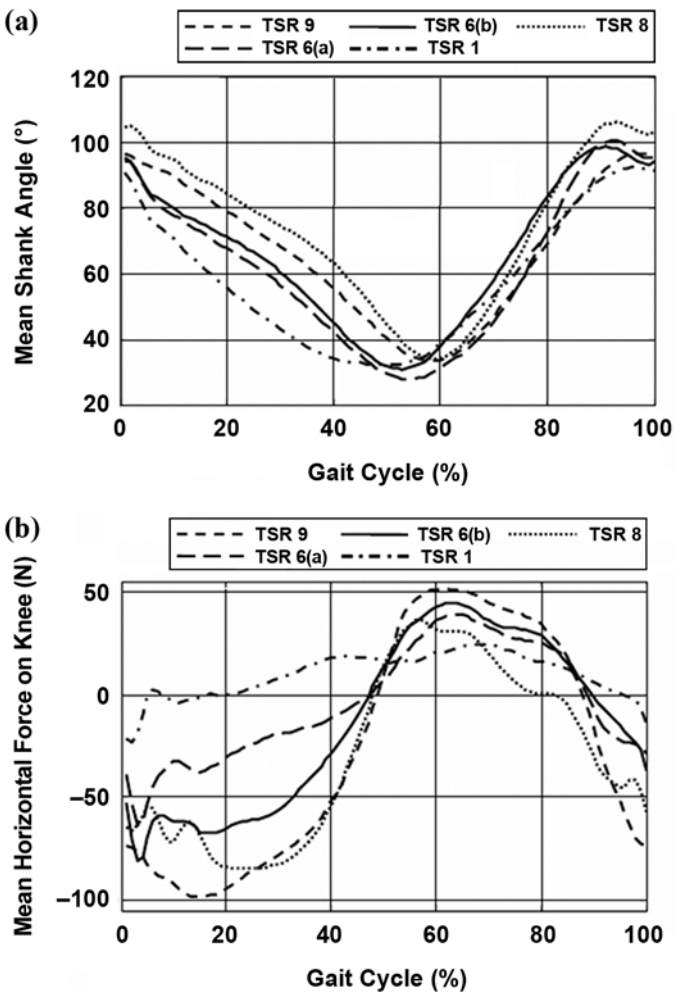

Figure 5.
(a) Mean shank angle and **(b)** mean horizontal trainer force on knee for one patient with spinal cord injury for five different trainers in same session (see **Figure 2**). Vertical shank angle is 90°. Rater noted some trainers did not hold leg extension long enough with this patient, which is evident in **(a)**: shank angle is >90° at 0% of gait cycle. After initial foot contact (0%–10% interval), shank angle decreased more rapidly when three lower-rated trainers assisted and continued to be more inclined for remainder of stance phase. Lower-rated trainers also helped create stance phase that terminated more prematurely, as seen when shank reached its minimum inclination at about 50% to 60% of gait cycle. Less-skilled trainers also exerted lower forces (less negative values) in stance, which likely contributed to poor leg extension. TSR = trainer skill rating.

For the kinematic measures of stepping performance, more-skilled trainers had less overall variability in shank angle (**Table 3**), with a combined *p*-value of 0.04 for negative correlation. Correlations of toe clearance variabilities with trainer ratings were disparate and not significant (**Table 4**).

DISCUSSION

This study examined the pattern of manual forces applied by trainers to the leg of patients with SCI during step-training using BWST. Our goals were to evaluate two hypotheses that are being used to justify the development of robotic gait-training devices: that there is substantial between-trainer variability in assistance patterns and that this variability makes a difference in the quality of the stepping pattern. We found that the variability between trainers in the magnitude of forces applied at key moments during stance and swing was about two times greater than the variability of a single trainer from step-to-step. This substantial between-trainer variability corresponded with frequent statistically significant differences when we compared pairs of trainers assisting on the same leg. We also found that these quantitative differences between trainers were not simply random but apparently generated the observed, strong correlations between an expert trainer's assessment of each trainer's skill and desirable kinematic gait features (i.e., adequate knee extension during stance and toe clearance during swing). Thus, we confirmed the two hypotheses—there is substantial between-trainer variability and it matters in the elicited stepping quality.

Implications for Clinical and Robotic Implementations of Step Training Using BWST

One implication for clinical trials of locomotor training is that between-trainer variability may be a confounding variable for evaluating efficacy of locomotor training with BWST. We note that the trainers who participated in this study were all trained in the same method of step training using BWST (developed by the UCLA Locomotion Laboratory) and worked together in one facility; thus, these differences were not attributable to differences in the taught technique or locale. The quantitative differences between trainers trained with different techniques, or in different locales, are probably even greater. We also note that trainers who were rated as more skilled in this study did in general have more experience with locomotor training (3–4 years of regular exposure), although one trainer had 10 years of experience but was among the lower-rated trainers. This confirms what is already clear to anyone who has attempted to learn how to provide manual assistance to the leg during step training: manual skill in gait training is learned and requires intensive repetitive practice. Further, as with any motor

Table 5.
Correlations between trainer skill rating (TSR) and swing-related measures. Symbols *p*+ and *p*– refer to *p*-values for positive and negative correlations, respectively, and are given only when different experiments exhibited positive and negative correlations for given relationship.

Relationship	Experiment						Combined
	SCI-D1	SCI-CD1	SCI-B1	SCI-B2	SCI-CD2	SCI-B3	*p*-Value
TSR and Toe Clearance*							
r	0.78	0.53	0.75	0.92	0.81	0.97	—
p	0.22	0.13	0.07	0.04	0.20	0.03	<0.001
TSR and Push-Up Force							
r	0.09	0.52	–0.91	0.48	1.00	–0.25	—
p+	0.45	0.18	0.99	0.26	0.004	0.62	0.23
p–	0.55	0.82	0.01	0.74	0.99	0.38	0.77
TSR and Toe-Clearance Variability							
r	0.36	0.79	–0.79	–0.78	–0.07	0.09	—
p+	0.38	0.06	–0.94	–0.78	–0.07	0.46	0.64
p–	0.62	0.94	0.06	0.89	0.52	0.54	0.36
TSR and Push-Up Force Variability							
r	0.63	–0.20	–0.63	0.31	0.63	0.79	—
p+	0.28	0.63	0.87	0.34	0.28	0.11	0.29
p–	0.72	0.37	0.13	0.66	0.72	0.89	0.71
TSR and Push-Up Force CV							
r	0.57	–0.53	–0.79	0.41	–0.77	–0.83	—
p+	0.31	0.82	0.94	0.20	0.22	0.92	0.76
p–	0.69	0.18	0.06	0.80	0.78	0.08	0.24

*Significant relationship according to combined *p*-value calculated using Stouffer method.
CV = coefficient of variation.

skill, the amount of practice required to become skilled depends on the particular person who is learning.

With respect to robotic gait training, this study validates at least part of the rationale typically prof fered for these devices. The manual provision of step training is indeed substantially variable between trainers and skill plays a role in this variability. The present study also provides some insight into which biomechanical features of the gait pattern correspond to the subjective impression of skill, which may be useful for robotic-therapy device design. Less-skilled trainers had problems with leg extension in stance and toe dragging in swing. More-skilled trainers ensured knee extension in stance by exerting a larger horizontal force against the knee. Exerting a large, controlled force at the right time during stance against the knee is something straightforward to implement with a robotic device and something robotic devices are well suited for.

We did not find, however, that trainers with a better skill rating exerted a larger ankle push-up force in initial swing to reduce the risk of toe dragging, which would be the "brute force" approach one would first be tempted to apply with a robot. Rather, toe clearance was correlated with other aspects of the stepping pattern, including ipsilateral leg extension in stance and hip flexion and extension measured by the horizontal ROM of the knee. The importance of ipsilateral hip-joint kinematics in stance for the proper initiation of swing is well known [23–25]. The implication is that robotic step-training devices should be designed, if they are to mimic expert trainers, to monitor and control the complete pattern of stepping and to provide little direct assistance in some phases of the gait cycle (e.g., the push-up force in early swing) if the overall pattern is good. Current gait-training robots only partially achieve this goal of adaptively controlling the complete pattern. For example, while an exoskeletal device like the Lokomat may provide a means to provide a more repeatable hip and knee pattern, it may exert unwanted contact forces as it constrains the legs to the parasagittal planes.

Finally, we found that as the perceived rating of skill increased, the step-to-step variability in the gait-training pattern decreased when variability was normalized by force magnitude. Step-to-step variability has been hypothesized to promote use-dependent plasticity [26–27], but the optimal amount of variability is unknown. The levels of

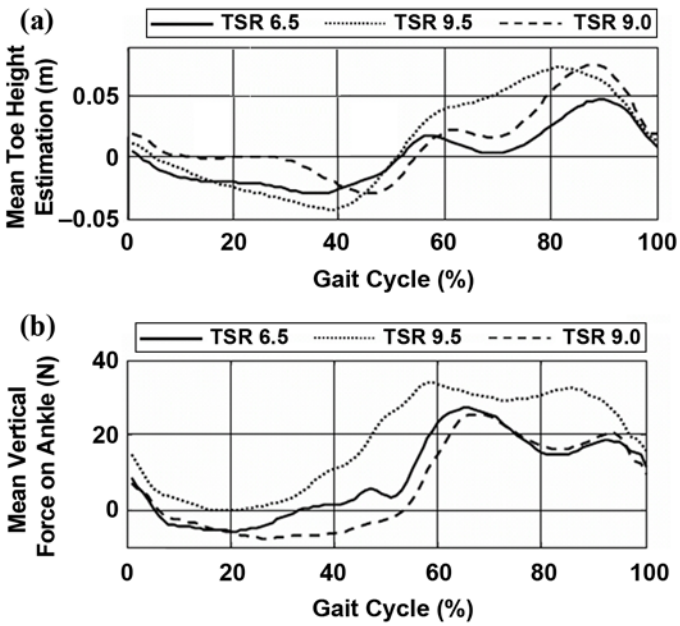

Figure 6.
(a) Mean toe height estimated for three different trainers assisting one patient with American Spinal Injury Association level D spinal cord injury in same session. Note toe height estimate is <0 m (treadmill level) during stance because ankle angle is assumed to be same as during swing. **(b)** Mean vertical ankle force applied by trainer in this experiment. Patient weight was 77 kg, body-weight support was 29%, and speed was 0.8 m/s. Mean toe clearance was too small at around 70% of gait cycle for two trainers, and in particular for least experienced trainer (TSR 6.5/10.0, solid line). Toe dragging was seen in videotapes with this trainer, consistent with finding of standard deviation across steps of 0.017 m in swing. Toe dragging was also observed with TSR 9.0/10.0 (dashed line). TSR = trainer skill rating.

variability of the trainers perceived as more skilled measured here (~18% CV in force) may be useful as starting targets for design variability in gait-training robots.

Possible Limitations

To measure the manual forces applied by trainers, it was necessary to insert a sensor system between the trainer's hands and the patient's leg. We worked with a skilled orthotist to design a sensorized orthosis that allowed the trainer to use similar hand positions as during normal training, and the orthosis pushed on the upper and lower shank at anatomical locations similar to those the trainers pushed on with their hands during normal training. The participating trainers thought that they could adequately assist through the orthosis. Nevertheless, the sensorized orthosis represented an intrusion in the normally tight coupling between trainer's hands and patient's leg, and this limitation should be considered.

Another possible limitation is that we rotated trainers from assisting on the sensorized leg, to assisting at the pelvis, to assisting at the nonsensorized leg instead of keeping the trainers on the pelvis and nonsensorized legs the same and rotating only through the sensorized leg. A possible effect is that more-skilled trainers may have, on average, worked at a disadvantage when "demonstrating" manual skill at the sensorized leg, since maintaining good stepping requires adequate assistance from the other assisting trainers. Similarly, less-skilled trainers may have worked at a relative advantage. On average, then, this confounding effect might have tended to decrease the dependence of stepping quality on trainer skill rating, but we found this dependence to be highly significant anyway.

A major limitation of this study is that we did not examine the long-term therapeutic effect of trainer skill or intertrainer variability on patient stepping ability; we simply examined the subjective quality of elicited stepping in a single session. Thus, any suggestion that what we measured matters for long-term gait outcomes should be conditioned by the degree of confidence in the hypothesis that within-session stepping patterns deemed "good" by expert trainers affect long-term therapeutic outcomes. The fact that trainer-applied step training was recently found to be significantly more effective than robot-applied training after stroke [13–14] supports this hypothesis.

CONCLUSIONS

The manual provision of step training is substantially more variable between trainers than within a single trainer and perceived skill correlates with specific biomechanical features of this variability, including knee extension during stance and toe drag during swing. Because trainers vary substantially and some trainers can establish gait patterns that are perceived as better, "manual skill" could potentially be systematically embedded in a repeatable control algorithm using robotics, providing wider access to a high-quality, step-retraining component of locomotor training. As with human trainers, one way to achieve this goal may be to have robotic gait-training devices learn how to apply therapeutic-efficacious assistance based on experience and feedback.

ACKNOWLEDGMENTS

Author Contributions:
Study design: D. J. Reinkensmeyer, S. J. Harkema, J. A. Galvez.
Experiment conception and instrumentation development: J. A. Galvez, D. J. Reinkensmeyer, S. J. Harkema.
Experiment process and data acquisition: J. A. Galvez.
Experiment supervision: A. Budovitch.
Stepping quality ratings: A. Budovitch.
Data analysis and interpretation: J. A. Galvez, D. J. Reinkensmeyer.
Drafting of manuscript: J. A. Galvez, D. J. Reinkensmeyer, S. J. Harkema.
Critical revision and approval of manuscript: D. J. Reinkensmeyer, S. J. Harkema, J. A. Galvez, A. Budovitch.
Financial Disclosures: The authors have declared that no competing interests exist.
Funding/Support: This material was based on work supported by the Christopher and Dana Reeve Foundation (grant GA1-0506-2), the National Institute on Disability and Rehabilitation Research (grant H133E020732), and the Spanish Government's Ramón y Cajal program (grant RYC-2005-98).
Additional Contributions: Dr. Galvez is now with the Instituto de Biomecánica de Valencia, Valencia, Spain, and Dr. Harkema is now with the Department of Neurological Surgery, University of Louisville, and the Frazier Rehab Institute, Louisville, Kentucky.
Institutional Review: All experiments were approved by the institutional review boards of the University of California at Irvine and Rancho Los Amigos National Rehabilitation Center, and participants provided informed consent.
Participant Follow-Up: The authors do not plan to inform participants of the publication of this study because of a lack of contact information.

REFERENCES

1. Wernig A, Nanassy A, Müller S. Maintenance of locomotor abilities following Laufband (treadmill) therapy in para- and tetraplegic persons: Follow-up studies. Spinal Cord. 1998;36(11):744–49. [PMID: 9848480] DOI:10.1038/sj.sc.3100670
2. Dietz V, Colombo G, Jensen L, Baumgartner L. Locomotor capacity of spinal cord in paraplegic patients. Ann Neurol. 1995;37(5):574–82. [PMID: 7755351] DOI:10.1002/ana.410370506
3. Behrman AL, Harkema SJ. Locomotor training after human spinal cord injury: A series of case studies. Phys Ther. 2000;80(7):688–700. [PMID: 10869131]
4. Edgerton VR, Kim SJ, Ichiyama RM, Gerasimenko YP, Roy RR. Rehabilitative therapies after spinal cord injury. J Neurotrauma. 2006;23(3–4):560–70. [PMID: 16629637] DOI:10.1089/neu.2006.23.560
5. Barbeau H, Nadeau S, Garneau C. Physical determinants, emerging concepts, and training approaches in gait of individuals with spinal cord injury. J Neurotrauma. 2006;23(3–4):571–85. [PMID: 16629638] DOI:10.1089/neu.2006.23.571
6. Dobkin B, Apple D, Barbeau H, Basso M, Behrman A, Deforge D, Ditunno J, Dudley G, Elashoff R, Fugate L, Harkema S, Saulino M, Scott M; Spinal Cord Injury Locomotor Trial Group. Weight-supported treadmill vs over-ground training for walking after acute incomplete SCI. Neurology. 2006;66(4):484–93. [PMID: 16505299]
7. Hesse S, Uhlenbrock D. A mechanized gait trainer for restoration of gait. J Rehabil Res Dev. 2000;37(6):701–8. [PMID: 11321006]
8. Colombo G, Joerg M, Schreier R, Dietz V. Treadmill training of paraplegic patients with a robotic orthosis. J Rehabil Res Dev. 2000;37(6):693–700. [PMID: 11321005]
9. Hornby TG, Campbell DD, Zemon DH, Kahn JH. Clinical and quantitative evaluation of robotic-assisted treadmill walking to retrain ambulation after spinal cord injury. Top Spinal Cord Inj Rehabil. 2005;11(2):1–17. DOI:10.1310/14Q9-AD7M-FXX9-1G2J
10. Werner C, Von Frankenberg S, Treig T, Konrad M, Hesse S. Treadmill training with partial body weight support and an electromechanical gait trainer for restoration of gait in subacute stroke patients: A randomized crossover study. Stroke. 2002;33(12):2895–2901. [PMID: 12468788] DOI:10.1161/01.STR.0000035734.61539.F6
11. Mayr A, Kofler M, Quirbach E, Matzak H, Fröhlich K, Saltuari L. Prospective, blinded, randomized crossover study of gait rehabilitation in stroke patients using the Lokomat gait orthosis. Neurorehabil Neural Repair. 2007;21(4):307–14. [PMID: 17476001] DOI:10.1177/1545968307300697
12. Husemann B, Müller F, Krewer C, Heller S, Koenig E. Effects of locomotion training with assistance of a robot-driven gait orthosis in hemiparetic patients after stroke: A randomized controlled pilot study. Stroke. 2007;38(2):349–54. [PMID: 17204680] DOI:10.1161/01.STR.0000254607.48765.cb
13. Hornby TG, Campbell DD, Kahn JH, Demott T, Moore JL, Roth HR. Enhanced gait-related improvements after therapist- versus robotic-assisted locomotor training in subjects with chronic stroke: A randomized controlled study. Stroke. 2008;39(6):1786–92. [PMID: 18467648] DOI:10.1161/STROKEAHA.107.504779
14. Hidler J, Nichols D, Pelliccio M, Brady K, Campbell DD, Kahn JH, Hornby TG. Multicenter randomized clinical trial evaluating the effectiveness of the Lokomat in subacute stroke. Neurorehabil Neural Repair. 2009;23(1):5–13. [PMID: 19109447] DOI:10.1177/1545968308326632
15. Galvez JA, Kerdanyan G, Maneekobkunwong S, Weber R, Scott M, Harkema SJ, Reinkensmeyer DJ. Measuring human trainers' skill for the design of better robot control

algorithms for gait training after spinal cord injury. Proceedings of the 2005 IEEE International Conference on Rehabilitation Robotics; 2005 Jun 28–Jul 1; Chicago, IL. Los Alamitos (CA): IEEE; 2005. p. 231–34. DOI:10.1109/ICORR.2005.1501092

16. Galvez JA, Budovitch A, Harkema SJ, Reinkensmeyer DJ. Quantification of therapists' manual assistance on the leg during treadmill gait training with partial body-weight support after spinal cord injury. Conf Proc IEEE Eng Med Biol Soc. 2007;2007:4028–32. [PMID: 18002883]
17. Bejczy A. Towards development of robotic aid for rehabilitation of locomotion-impaired subjects. Proceedings of the First Workshop on Robot Motion and Control; 1999 Jun 28–29; Kiekrz, Poland. Los Alamitos (CA): IEEE; 1999. p. 9–16. DOI:10.1109/ROMOCO.1999.791044
18. Rao SS, Boyd LA, Mulroy SJ, Bontrager EL, Gronley JK, Perry J. Segment velocities in normal and transtibial amputees: Prosthetic design implications. IEEE T Rehabil Eng. 1998; 6(2):219–26. [PMID: 9631330] DOI:10.1109/86.681188
19. Winter DA. Kinematic and kinetic patterns in human gait: Variability and compensating effects. Hum Mov Sci. 1984; 3(1–2):51–76. DOI:10.1016/0167-9457(84)90005-8
20. Loughin TM. A systematic comparison of methods for combining *p*-values from independent tests. Comput Stat Data Anal. 2004;47(3):467–85. DOI:10.1016/j.csda.2003.11.020
21. Rice WR. A consensus combined *p*-value test and the family-wide significance of component tests. Biometrics. 1990; 46(2):303–8. DOI:10.2307/2531435
22. Whitlock MC. Combining probability from independent tests: The weighted *Z*-method is superior to Fisher's approach. J Evol Biol. 2005;18(5):1368–73. [PMID: 16135132] DOI:10.1111/j.1420-9101.2005.00917.x
23. Pang MY, Yang JF. The initiation of the swing phase in human infant stepping: Importance of hip position and leg loading. J Physiology. 2000;528(Pt 2):389–404. [PMID: 11034628] DOI:10.1111/j.1469-7793.2000.00389.x
24. Pearson KG. Proprioceptive regulation of locomotion. Curr Opin Neurobiol. 1995;5(6):786–91. [PMID: 8805415] DOI:10.1016/0959-4388(95)80107-3
25. Dietz V, Müller R, Colombo G. Locomotor activity in spinal man: Significance of afferent input from joint and load receptors. Brain. 2002;125(Pt 12):2626–34. [PMID: 12429590] DOI:10.1093/brain/awf273
26. Cai LL, Fong AJ, Otoshi CK, Liang Y, Burdick JW, Roy RR, Edgerton VR. Implications of assist-as-needed robotic step training after a complete spinal cord injury on intrinsic strategies of motor learning. J Neurosci. 2006;26(41): 10564–68. [PMID: 17035542] DOI:10.1523/JNEUROSCI.2266-06.2006
27. Lewek MD, Cruz TH, Moore JL, Roth HR, Dhaher YY, Hornby TG. Allowing intralimb kinematic variability during locomotor training poststroke improves kinematic consistency: A subgroup analysis from a randomized clinical trial. Phys Ther. 2009;89(8):829–39. [PMID: 19520734] DOI:10.2522/ptj.20080180

Submitted for publication April 15, 2010. Accepted in revised form September 22, 2010.

This article and any supplementary material should be cited as follows:
Galvez JA, Budovitch A, Harkema SJ, Reinkensmeyer DJ. Trainer variability during step training after spinal cord injury: Implications for robotic gait-training device design. J Rehabil Res Dev. 2011;48(2):147–60. DOI:10.1682/JRRD.2010.04.0067