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Gait deviations associated with post-stroke hemiparesis: improvement during treadmill walking using weight support, speed, support stiffness, and handrail hold

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Abstract

By comparing treadmill walking in hemiparetic and non-disabled individuals at matched speeds, Chen et al. [Chen G, Patten C, Kothari DH, Zajac FE. Gait differences between individuals with post-stroke hemiparesis and non-disabled controls at matched speeds. Gait Posture (2004)] identified gait deviations that were consistent with impaired swing initiation and single limb support in the paretic limb and related compensatory strategies. Treadmill training with harness support is a promising, task-oriented approach to restoring locomotor function in individuals with post-stroke hemiparesis. To provide a rationale for the proper selection of training parameters, we assessed the potential of body weight support, treadmill speed, support stiffness, and handrail hold to improve the identified gait deviations associated with hemiparesis during treadmill walking. In the six hemiparetic subjects studied, the adjustment of each training parameter was found to improve a specific set of the gait deviations. With increased body weight support or the addition of handrail hold, percentage single limb support time on the paretic limb increased and temporal symmetry improved. With increased treadmill speed, leg kinetic energy at toe-off in the paretic limb increased but remained low relative to values in the non-paretic limb. With increased support stiffness, the exaggerated energy cost associated with raising the trunk during pre-swing and swing of the paretic limb was improved. We conclude that the proper selection of training parameters can improve the gait pattern practiced by individuals with hemiparesis during treadmill training and may improve treatment outcome.

Keywords: Gait; Stroke; Rehabilitation; Treadmill; Biomechanics

1. Introduction

After suffering a stroke, many individuals are left with neurological and functional deficits, including hemiparesis, which impair their ability to walk. Approximately, two-thirds of acute hospitalized stroke patients cannot walk independently [2]. Of those individuals who recover their ability to walk, many are still disabled by slow walking speed and limited endurance.

By comparing treadmill walking in hemiparetic and non-disabled individuals at matched speeds, Chen et al. [1] identified gait deviations that were consistent with impaired swing initiation and single limb support in the paretic limb and related compensatory strategies. Leg kinetic energy at toe-off in the paretic limb was reduced, consistent with inadequate propulsion by the plantarflexors or hip flexors during swing initiation. As a result, percentage swing time was increased and peak knee flexion during swing was reduced. Energy cost associated with raising the trunk during pre-swing and swing of the paretic limb was exaggerated, consistent with compensatory pelvic hiking to clear the paretic limb with reduced knee flexion. Leg kinetic energy at toe-off in the non-paretic limb was increased and percentage swing time reduced, consistent with weakness or poor balance during single limb support on the paretic limb. The improvement of these

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gait deviations may improve locomotor function in hemiparetic individuals.

Treadmill training with harness support is an effective, task-oriented approach to restoring locomotor function in individuals with post-stroke hemiparesis [3–5]. With the assistance of the harness and treadmill, many hemiparetic individuals are able to walk with more normal gait kinematics and EMG timing [6] and improved symmetry [6–8]. Manual assistance is typically provided by one or more therapists to further guide the trunk and legs through a normal gait trajectory. As individuals improve their locomotor ability, training parameters, such as body weight support and treadmill speed, are usually adjusted; and manual assistance, if provided, is reduced. The appropriate selection of training parameters and use of manual assistance is believed to be key to obtaining optimal therapeutic results [9,10]. To provide a rationale for the proper selection of training parameters, we assessed the potential of body weight support, treadmill speed, support stiffness, and handrail hold to reduce the identified gait deviations associated with hemiparesis [1] during treadmill walking in a pilot group of subjects.

2. Methods

Subject characteristics and the experimental setup and data analyses were presented in Chen et al. [1]. The subjects wore a Medical Harness (Robertson Mountaineering, Henderson, NV) attached to a custom-made support frame [11] as they walked on a Rehabilitation Treadmill (Biodex Medical Systems, Shirley, NY). Ten experimental conditions (Table 1) corresponding to different selections of body weight support, support stiffness, handrail hold, and treadmill speed were presented to the subjects in a randomized order. The level of body weight support (20%, 35%, or 50%) provided by the harness was set by changing the length of the support cable with a winch at the bottom of the frame and measured by an ATI force-torque sensor (ATI Industrial Automation, Apex, NC). Harness-support stiffness (11.7, 35.1 N/cm, or rigid) was adjusted by connecting music wire springs (Century Spring Corporation, Los Angeles, CA; $K = 35.1$ N/cm) in series with the cable. Handrail hold was provided using the subject's non-paretic arm. Treadmill speed was set to 70%, 100%, or 130% of the subject's comfortable treadmill speed (CTS) as determined during single pre-sessions where the subjects were familiarized to treadmill walking. The default settings of body weight support (35%), treadmill speed (100% CTS), support stiffness (35.1 N/cm), and handrail hold (none) were maintained, while variations in each training parameter were tested (Table 1). In addition, one condition of free treadmill walking, without harness support and handrail hold, was tested at the default speed (Table 1).

Table 1
Experimental treadmill conditions

Condition	BWS (% weight)	Speed (% CTS)	Stiffness (N/cm)	Handrail Hold
1 (free)	n/a	100	n/a	None
2	20[a]	100	35.1	None
3 (default)	35	100	35.1	None
4	50[a]	100	35.1	None
5	35	70[a]	35.1	None
6	35	130[a]	35.1	None
7	35	100	11.7[a]	None
8	35	100	Rigid[a]	None
9	n/a	100	n/a	Yes[a]
10	35	100	35.1	Yes[a]

Condition 1 (free), free walking on the treadmill without harness support and handrail hold. Condition 3 (default), default selection of BWS, speed, stiffness, and handrail hold. BWS, body weight support; CTS, comfortable treadmill speed; n/a, not applicable (harness support not provided).
[a] Variation in BWS, speed, stiffness, or handrail hold.

Differences between free and harness-supported treadmill walking at the default settings of body weight support (35%) and support stiffness (35.1 N/cm) were tested using the Wilcoxon signed-rank test. The influence of body weight support, treadmill speed, support stiffness, and handrail hold on each gait variable was tested on a multivariate basis using Friedman's method for randomized blocks. Significance was set at $P < 0.05$ for all tests.

3. Results

Table 2 presents the group means for gait variables during harness-supported and free treadmill walking. With harness support, leg kinetic energy at toe-off was reduced and percentage swing time increased in the non-paretic limb, which improved swing time symmetry (all $P = 0.03$) and brought these variables closer to values in non-disabled controls from Chen et al. [1] (Fig. 1). However, leg kinetic energy at toe-off, swing time, and peak knee flexion during swing in the paretic limb and energy cost associated with raising the trunk during pre-swing (NP-LR) and swing (NP-SS) of the paretic limb were not significantly different with harness support.

Table 3 presents the effect of each training parameter. With increased body weight support, percentage swing time of the non-paretic limb increased, resulting in improved swing time symmetry (both multivariate $P = 0.03$, Fig. 2). With increased treadmill speed, leg kinetic energy at toe-off in the paretic and non-paretic limbs increased (multivariate $P = 0.002$ and 0.006, respectively; Fig. 3) but did not become more similar. Also, energetic cost associated with raising the trunk increased with speed (multivariate $P = 0.03$; Fig. 3, Energetic cost, Trunk PEG in NP-LR and NP-SS). With increased support stiffness, energetic cost associated with raising the trunk was reduced (multivariate $P = 0.002$; Fig. 4). There was a statistically significant effect of support stiffness on leg kinetic energy at toe-off in the paretic limb (multivariate $P = 0.04$), but the magnitudes of these differences were small. Across the conditions of handrail hold,

Table 2
Comparison of free vs. harness-supported treadmill walking

	Free walking (condition 1)	Harness support (condition 3)	P	Non-disabled control
Swing time (% gait cycle)				
Paretic limb	39.8 (4.6)	41.0 (8.4)		32.2 (10.3)
Non-paretic limb	21.5 (4.5)	30.4 (6.1)	0.03	31.4 (8.4)
Asymmetry (%)	43.4 (16.5)	22.7 (25.5)	0.03	−0.1 (11.3)
Leg kinetic energy at toe-off (cJ/kg)				
Paretic limb	2.0 (1.2)	2.0 (1.8)		4.5 (2.0)
Non-paretic limb	7.5 (3.7)	4.9 (2.5)	0.03	4.9 (2.6)
Peak knee flexion during swing (°)				
Paretic limb	37.8 (9.8)	39.0 (13.9)		58.6 (7.4)
Component energetic cost (cJ/kg)				
Trunk PEG in NP-LR and NP-SS	14.6 (5.0)	11.6 (2.5)		6.1 (2.1)

Group means and standard deviations (in parentheses). Paretic and non-paretic limb variables are side-matched in non-disabled, speed-matched controls from Chen et al. (2004). *P*, significance between free and harness-supported walking conditions (Wilcoxon signed-rank test). PEG, potential energy of gravity; NP-LR, non-paretic limb loading response; NP-SS, non-paretic limb single support.

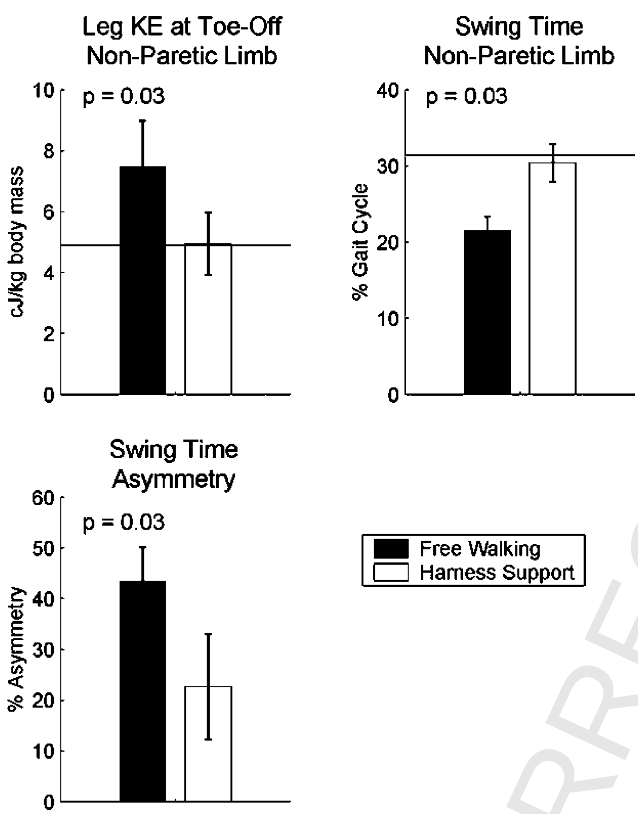

Fig. 1. Significant gait differences between free and harness-supported treadmill walking (solid and white bars, respectively) (leg kinetic energy at toe-off and percentage swing time in the non-paretic limb and swing time asymmetry. Values are means ± S.E.; *P*, significance between free and harness-supported walking conditions (Wilcoxon signed-rank test). Horizontal lines designate values during free treadmill walking in non-disabled, speed-matched controls from Chen et al. (2004). Paretic and non-paretic limb variables are side-matched in control subjects. *Note*: Swing time asymmetry in control subjects was close to zero. With the addition of harness support, leg kinetic energy at toe-off in the non-paretic limb was reduced, swing time increased, which improved swing time symmetry.

harness support, and combined harness support and handrail hold, swing time of the non-paretic limb increased, resulting in improved swing time symmetry (both multivariate P = 0.03, Fig. 5).

4. Discussion

The adjustment of each training parameter improved a specific set of gait deviations associated with post-stroke hemiparesis (Table 3). Our findings provide a rationale for the proper selection of training parameters during treadmill training in hemiparetic individuals.

Increased body weight support and the addition of handrail hold increased percentage single limb support time on the paretic limb, since swing time of the non-paretic limb

Table 3
Training parameters: effect summary

	BWS	Speed	Stiffness	Handrail
Swing time				
Paretic limb	X	X	X	X
Non-paretic limb	√ (+)	* (+)	X	√ (+)
Asymmetry	√ (−)	* (−)	* (−)	√ (−)
Leg kinetic energy at toe-off				
Paretic limb	X	√ (+)	√ (small)	X
Non-paretic limb	* (−)	√ (+)	X	* (−)
Peak knee flexion during swing				
Paretic limb	X	X	X	X
Component energetic cost				
Trunk PEG in NP-LR and NP-SS	X	√ (+)	√ (−)	* (−)

(√) Statistically significant effect (mult-$P < 0.05$; Friedman's method for randomized blocks); (*) some tendencies (not statistically significant); (*X*) little effect; (+) variable increased with level of parameter; (−) variable decreased with level of parameter. BWS, body weight support; PEG, potential energy of gravity; NP-LR, non-paretic limb loading response; NP-SS, non-paretic limb single support.

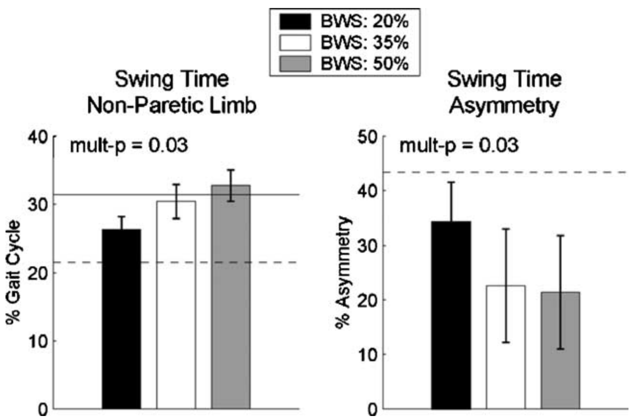

Fig. 2. Significant effects of body weight support during treadmill walking (percentage swing time of the non-paretic limb and swing time asymmetry. Each condition was tested at the default level of treadmill speed (100% CTS), support stiffness (35.1 N/cm), and handrail hold (none); mult-*P*, multivariate significance between body weight support conditions (Friedman's method for randomized blocks). *Note*: Swing time asymmetry in control subjects was close to zero. With increased body weight support, swing time of the non-paretic limb increased, resulting in improved swing time symmetry. In this figure, and in Figs. 4–6, values are means ± S.E. Horizontal lines designate values during free treadmill walking in the subjects (dashed lines) and non-disabled, speed-matched controls from Chen et al. (2004) (solid lines). Paretic and non-paretic limb variables are side-matched in control subjects.

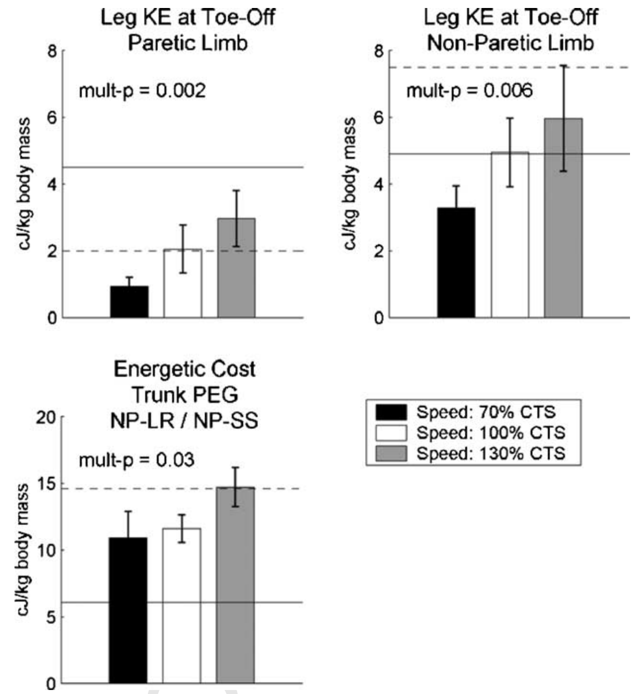

Fig. 3. Significant effects of treadmill speed during supported treadmill walking (leg kinetic energy at toe-off in the paretic and non-paretic limbs and energetic cost associated with rises in trunk potential energy during pre-swing (NP-LR) and swing (NP-SS) of the paretic limb. Each condition was tested at the default level of body weight support (35%), support stiffness (35.1 N/cm), and handrail hold (none); mult-*P*, multivariate significance between treadmill speed conditions (Friedman's method for randomized blocks). With increased treadmill speed, leg kinetic energy at toe-off in the paretic and non-paretic limbs increased but did not become more similar. Energetic cost associated with rises in trunk potential energy increased with speed, probably due to larger vertical displacements of the trunk associated with increased stride length.

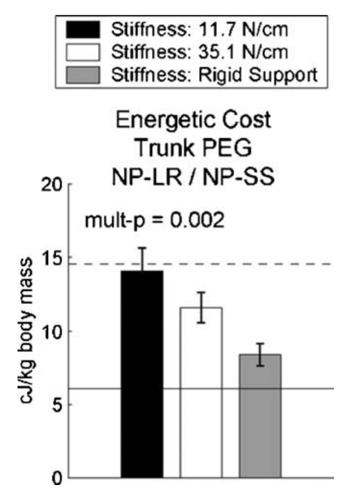

Fig. 4. Significant effect of support stiffness during treadmill walking (energetic cost associated with the rises in trunk potential energy during pre-swing (NP-LR) and swing (NP-SS) of the paretic limb. Each condition was tested at the default level of body weight support (35%), treadmill speed (100% CTS), and handrail hold (none); mult-*P*, multivariate significance between support stiffness conditions (Friedman's method for randomized blocks). With increased support stiffness, energetic cost associated with rises in trunk potential energy decreased, probably due to restriction of vertical displacements of the trunk.

increased. Increased single limb support time may provide a higher training stimulus for impaired equilibrium reflexes [6,8,9,12]. Previous studies have also reported increased single limb support time on the paretic limb during treadmill walking with harness support [6–8]. Additionally, our study found that single limb support time was increased when handrail hold was provided by itself, though to a smaller extent than with harness support, and further increased when handrail hold was combined with harness support. Reduced single limb support time on the paretic limb is a prominent characteristic of hemiparetic gait [1,13,14] and is consistent with weakness or poor balance during support on the paretic limb. Both harness support and handrail hold assist in weight support and balance, which may allow hemiparetic individuals to achieve longer support time on the paretic limb.

Faster treadmill speeds increased leg kinetic energy at toe-off in the paretic limb, which could be important to achieving faster walking speeds overground and improving swing initiation at slower speeds. Leg kinetic energy at toe-off in the paretic limb only improved with increased treadmill speed and was not appreciably affected by body weight support, support stiffness, or handrail hold. Low leg kinetic energy at toe-off in the paretic limb, which can result from inadequate propulsion by the plantarflexors or hip flexors [15,16], may limit how fast the paretic limb advances during swing and, consequently, gait speed. When the gait of hemiparetic and non-disabled subjects were compared at the same speeds, deviations in swing time and peak knee flexion during swing in the paretic limb were thought to result from inadequate leg kinetic energy at toe-off [1]. In

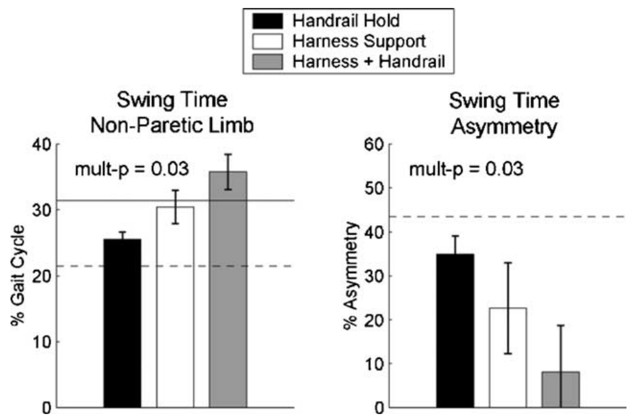

Fig. 5. Significant effects across the conditions of treadmill walking with handrail hold, harness support, and combined harness support and handrail hold (percentage swing time of the non-paretic limb and swing time asymmetry. Each condition was tested at the default level of treadmill speed (100% CTS). Harness support was provided at the default level of body weight support (35%) and support stiffness (35.1 N/cm); mult-*P*, multivariate significance between the three conditions (Friedman's method for randomized blocks). *Note*: Swing time asymmetry in control subjects was close to zero. Across the conditions of handrail hold, harness support, and combined harness support and handrail hold, swing time of the non-paretic limb increased, resulting in improved swing time symmetry.

our study, these deviations in swing kinematics did not improve at faster treadmill speeds, probably because leg kinetic energy at toe-off in the paretic limb, though increased, was still inadequate for the faster speed of walking. However, if the increased leg kinetic energy elicited by faster training can be maintained at slower walking speeds overground, we believe that it may lead to improved swing kinematics in the paretic limb and, perhaps, the reduction of other costly compensatory strategies (e.g., pelvic hiking and circumduction of the paretic limb) that might limit walking endurance.

The reduction of energy cost associated with raising the trunk during pre-swing and swing of the paretic limb provides a rationale for the use of a stiffer harness support during treadmill training in hemiparetic individuals. The restoration of normal displacements of the trunk has been stressed in treadmill training because it strongly affects the sensory experience that is believed to be important to achieving optimal training results [17,18]. Moreover, exaggerated displacements of the trunk during walking in hemiparetic individuals contribute to increased mechanical energetic cost [1,19]. Ironically, advocates for the use of a compliant harness support have generally emphasized that compliance allows for more natural displacements of the trunk during the gait cycle, which a rigid support was thought to eliminate. However, since vertical displacements of the trunk were abnormally large in the subjects, a reduction in these displacements actually improved the overall motion profile of the trunk. On the other hand, much of the increased vertical displacement of the trunk in the subjects was attributed to the large rise in trunk height during pre-swing and swing of the paretic limb, which compensates for reduced knee flexion in the limb. Thus, the clinical importance of reducing these displacements during training could be challenged. For instance, if reduced knee flexion during swing in the paretic limb cannot be improved in the individual, reduction in compensatory pelvic hiking would not be expected to improve locomotor performance overground. In addition, some of the subjects preferred a compliant support because a rigid support was uncomfortable.

Deviations in swing time and peak knee flexion during swing in the paretic limb were not improved with the adjustment of training parameters (Table 3). Indeed, manual assistance to advance the paretic limb during swing is commonly needed in hemiparetic individuals who cannot walk independently on the treadmill even when harness support or handrail hold is provided [4,5,9]. Manual assistance greatly increases the physical demand on therapists and has driven the push for fully mechanized gait trainers [18] and powered orthoses [20] that can assist even severely impaired individuals to produce a gait-like movement pattern. Nevertheless, our study found that gait deviations associated with swing of the paretic limb (i.e., increased swing time and reduced peak knee flexion) are also resistant to improvement in hemiparetic individuals who are ambulatory on the treadmill. In this case, manual assistance or other facilitatory techniques (e.g., functional electrical stimulation) should, perhaps, be provided to insure proper kinematics of the paretic limb during swing, even though the subjects can advance the limb independently.

Our study provides a rationale for the proper selection of training parameters for treadmill training in hemiparetic individuals, but some important limitations should be noted. First, because of the small sample size in our study, the findings should be substantiated in a larger group of subjects. Second, even if certain gait deviations associated with hemiparesis are reduced during treadmill walking, the practice of an improved gait pattern, as defined, may not improve the individual's locomotor ability overground. Ultimately, the importance of these factors to treatment outcome in hemiparetic individuals needs to be verified in clinical trials. Lastly, our findings, based on data from ambulatory subjects, may not be relevant to the training of more severely impaired individuals who require a great amount of manual assistance to walk on the treadmill.

In conclusion, the proper selection of training parameters can improve specific gait deviations associated with post-stroke hemiparesis during treadmill walking. With harness support or handrail hold, increased treadmill speed, and increased support stiffness, deviations in single limb support time in the paretic limb, leg kinetic energy at toe-off in the paretic limb, and energy cost associated with raising the trunk were improved, respectively. However, deviations in swing time and knee flexion during swing in the paretic limb resisted improvement and probably need to be addressed using manual assistance or other facilitatory techniques. The practice of an improved gait pattern during treadmill training, as defined by a reduction in these gait deviations, may improve treatment outcome in hemiparetic individuals.

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