Treadmill Training with Harness Support: A Biomechanical Basis for Selection of Training Parameters for Individuals with Post-Stroke Hemiparesis

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DOCTOR OF PHILOSOPHY

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ABSTRACT

Treadmill training with harness support is a promising, task-oriented approach to restoring locomotor function in individuals with post-stroke hemiparesis, but a scientific basis for the proper selection of training parameters is needed. The goal of this dissertation is to provide a biomechanical basis for the selection of training parameters (i.e., body weight support, treadmill speed, support stiffness, and handrail hold) to improve the gait pattern practiced by hemiparetic individuals during treadmill training.

By comparing gait characteristics of hemiparetic and non-disabled individuals at matched treadmill speeds, non-speed-related gait deviations associated with post-stroke hemiparesis were identified. In the hemiparetic subjects, leg kinetic energy at toe-off in the paretic limb was reduced, resulting in increased percentage swing time and reduced peak knee flexion during swing, consistent with inadequate leg propulsion by the plantarflexors or hip flexors during swing initiation. 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 exaggerated, resulting in reduced swing time, consistent with weakness or poor balance during single limb support on the paretic limb.

The adjustment of each training parameter during treadmill walking in the hemiparetic subjects was found to improve a specific set of the gait deviations identified. With increased body weight support or the addition of handrail hold, the exaggerated leg kinetic energy at toe-off and reduced swing time in the non-paretic limb were improved, resulting in increased single limb support time on the paretic limb. With increased treadmill speed, the reduced leg kinetic energy at toe-off in the paretic limb was improved but remained low relative to values in the non-paretic limb. With increased support stiffness, the exaggerated energy cost associated with raising the trunk was

improved. We conclude that the proper selection of training parameters can improve the gait pattern practiced by individuals with post-stroke hemiparesis during treadmill training. These actions, ostensibly, may improve treatment outcome.

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TABLE OF CONTENTS

List of Tables

Appendix B

LIST OF ILLUSTRATIONS

Discussion

Appendix A

Appendix B

(no figures)

INTRODUCTION

Stroke survivors are often left with neurological and functional deficits, including hemiparesis, which impair their ability to walk. Approximately a half a million Americans suffer a stroke each year with the estimated number of stroke survivors close to 3 million (Post-Stroke Rehabilitation, 1995). Approximately, two-thirds of acute hospitalized stroke patients cannot walk independently (Jorgensen *et al.*, 1995). Of those individuals who recover their ability to walk, many are still disabled by slow walking speed and limited endurance, which restrict their independent mobility at home and in the community.

Treadmill training with harness support is a promising, task-oriented approach to restoring locomotor function in individuals with post-stroke hemiparesis. With the assistance of the harness and treadmill, many hemiparetic individuals are able to walk with more normal gait kinematics and EMG timing (Hesse *et al.*, 1999) and improved symmetry (Hesse *et al.*, 1999; Hassid *et al.*, 1997). Manual assistance is typically provided by one or more therapists to further guide the trunk and legs through a normal gait trajectory. After treadmill training, both sub-acute patients and individuals with chronic hemiparesis have demonstrated improved locomotor ability overground, as reflected by improved ambulatory status and increased walking speed (Hesse *et al.*, 1995a; Visintin *et al.*, 1998; Sullivan *et al.*, 2002).

Treadmill training with harness support is believed to be effective in hemiparetic individuals, because it provides task-specific training (Hesse, 1999; Shepherd and Carr, 1999) that optimizes the sensory inputs facilitating spinal and supraspinal locomotor networks (Dobkin, 1999). Moreover, training parameters, such as level of body weight support and treadmill speed, can be adjusted to facilitate the gait pattern and accommodate for improvement in locomotor ability. The appropriate selection of training parameters and use of manual assistance is believed to be important to maximize the effectiveness of the technique (Hesse, 1999; Dobkin, 1999).

However, currently, there is not an objective basis for the selection of training parameters for hemiparetic individuals. Indeed, training approaches used by experimenters vary widely and are often in conflict (Gardner *et al.*, 1998; Dobkin, 1999; Hesse *et al.*, 1995a; Seif-Naraghi and Herman, 1999; Visintin *et al.*, 1998; Miller *et al.*, 2002). Considerable latitude exists in the application of treadmill training, because individuals are able to walk at different settings of training parameters and with variable degrees of manual assistance during training sessions. Experimenters attempt to adjust training parameters and manual assistance from session to session, and even within a session, based on the visual appearance of gait kinematics and the perceived force profile of manual assistance being provided. Further complicating the issue, different methods of body weight support have been used (i.e., support with a rigid cable, compliant springs, pneumatic lift, etc.), which allow different degrees of trunk motion. As the rehabilitation potential of treadmill training is more extensively studied in hemiparetic individuals, a scientific basis for the proper selection of training parameters is needed.

The goal of this dissertation is to provide a biomechanical basis for the selection of training parameters for individuals with post-stroke hemiparesis (see Figure 1). The facilitation of an improved gait pattern during treadmill walking, as defined by a reduction in gait deviations associated with hemiparesis, can be used as a rationale for the selection of training parameters. However, many characteristics of hemiparetic gait are likely a consequence of slow walking speed and may not be associated with factors that might limit gait performance. By comparing gait characteristics of hemiparetic and non-disabled subjects walking at matched speeds on a treadmill, Study 1 identifies non-speed-related gait deviations, which are more likely associated with impairment, resulting from hemiparesis, and related compensatory strategies. Study 2 assesses how significant gait deviations associated with hemiparesis can be improved during treadmill walking with the proper selection of training parameters (i.e., body weight support, treadmill speed, support stiffness, and handrail hold). The two studies provide a scientific rationale based on biomechanics for the proper selection of parameters during treadmill training in hemiparetic individuals.

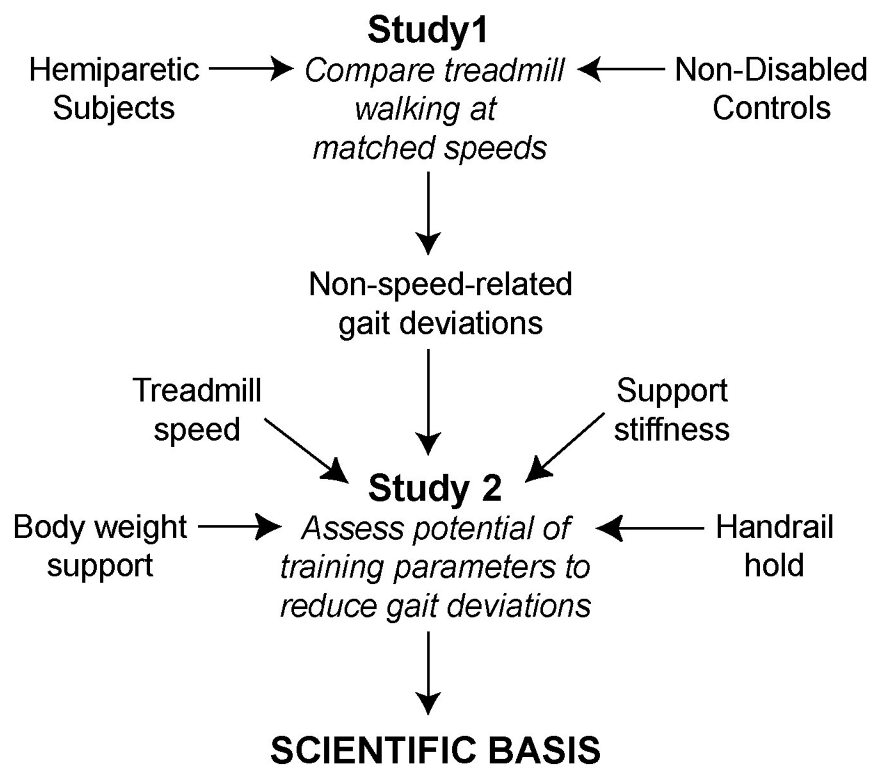

Figure 1. Development of a scientific basis for selection of training parameters for individuals with post-stroke hemiparesis. In Study 1, gait characteristics in hemiparetic subjects and non-disabled controls are compared at matched treadmill speeds to identify non-speed-related gait deviations associated with hemiparesis. In Study 2, the potential of training parameters (i.e., level of body weight support, treadmill speed, support stiffness, and handrail hold) to reduce the identified gait deviations during treadmill walking is assessed.

Study 1 Non-speed-related gait differences between individuals with post-stroke hemiparesis and non-disabled controls

Abstract

Treadmill walking was used to assess the non-speed-related gait differences between six individuals with post-stroke hemiparesis and six non-disabled, healthy controls. The treadmill allowed the collection of kinematic and insole pressure data from multiple, steady state gait cycles of subjects walking at matched speeds that were comfortable for the hemiparetic subjects (0.13-0.45 m/s). At matched speeds, a large set of gait differences between the hemiparetic and non-disabled subjects was consistent with impaired swing initiation in the paretic limb (i.e., inadequate acceleration of the leg during pre-swing, increased percentage swing time, and reduced knee flexion at toe-off and mid-swing in the paretic limb) and related compensatory strategies (i.e., pelvic hiking and swing-phase acceleration and circumduction of the paretic limb). Exaggerated positive work associated with raising the trunk during pre-swing and swing of the paretic limb, consistent with pelvic hiking, contributed to increased mechanical energetic cost during walking in the hemiparetic subjects. A second set of gait differences was consistent with impaired single limb support on the paretic limb (i.e., shortened support time on the paretic limb) and related compensatory strategies (i.e., exaggerated acceleration of the non-paretic limb during pre-swing to shorten its swing time). Other significant gait differences included asymmetry in step length and increased step width. Stride length and joint angular excursions at the hip and ankle were comparable in the speed-matched groups. The comparison of gait characteristics in hemiparetic and non-disabled subjects walking at matched treadmill speeds was useful in identifying non-speed-related gait differences that are likely associated with impairment, resulting from hemiparesis, and related compensatory strategies.

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1. Introduction

Stroke survivors are often 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 (Jorgensen *et al.*, 1995). Of those individuals who recover their ability to walk, many are still disabled by slow walking speed and limited endurance. In addition, gait in individuals with post-stroke hemiparesis is characterized by reduced cadence, stride length, and joint angular excursions (von Schroeder *et al.*, 1995; Wall and Ashburn, 1979; Dean *et al.*, 2000); asymmetry in temporal, spatial, kinematic, and kinetic gait variables (Brandstater *et al.*, 1983; Roth *et al.*, 1997; Kim and Eng, 2003; Wall and Turnbull, 1986); and increased mechanical energetic cost (Olney *et al.*, 1986; Iida and Yamamuro, 1987). The correction of these gait deviations has been a focus of gait rehabilitation, since it may improve locomotor function in hemiparetic individuals. However, many of these gait deviations are likely a consequence of slow walking speed and may not be associated with impairment or compensatory strategies.

In both non-disabled and hemiparetic individuals, gait parameters change with walking speed. When non-disabled individuals walk at slower than normal speeds, they exhibit reduced cadence, stride length, and joint angular excursions (Andriacchi *et al.*, 1977; Murray *et al.*, 1984; Lehmann *et al.*, 1987). In hemiparetic individuals, many temporal, kinematic, and kinetic gait variables are highly correlated with self-selected walking speed (Olney *et al.*, 1994; Roth *et al.*, 1997; Wagenaar and Beek, 1992; Kim and Eng, 2003). Even though many investigators have interpreted these correlations to indicate the variables' importance in determining gait speed in hemiparetic individuals (Olney *et al.*, 1994; Roth *et al.*, 1997; Wagenaar and Beek, 1992; Kim and Eng, 2003), it may be that these variables are speed-dependent, as in non-disabled individuals. Regardless, speed-dependency of gait variables in hemiparetic individuals may be partially attributed to differences in level of impairment between individuals (e.g., less impaired individuals tend to be more symmetrical and have faster self-selected walking speeds), while speed-dependency in non-disabled individuals are presumably attributed to altered strategies that are required to walk at slower speeds.

The comparison of gait characteristics in hemiparetic and non-disabled individuals walking at matched speeds may assist in identifying non-speed-related gait differences that are more likely associated with impairment, resulting from hemiparesis, and related compensatory strategies. Because gait speed in hemiparetic individuals is significantly slower than normal, some characteristics of hemiparetic gait may be a consequence of marked differences in speed. However, other characteristics may not be speed-related and, instead, result from a combination of the pathology and related compensatory strategies to accommodate for altered physiologic function. Using speed-matched data from non-disabled controls, gait deviations in both the paretic and non-paretic limbs can be identified. For example, asymmetry in stance/swing time in hemiparetic individuals has been attributed to prolonged swing time of the paretic limb (e.g., due to inadequate acceleration of the limb during pre-swing) (Lehmann *et al.*, 1987; Olney and Richards, 1996) and/or shortened swing time of the non-paretic limb (e.g., due to weakness or poor balance during single limb support on the paretic limb) (Olney *et al.*, 1994; Roth *et al.*, 1997; Wagenaar and Beek, 1992; Kim and Eng, 2003). By comparing swing times and leg kinetic energies at toe-off in the paretic and non-paretic limbs to values in non-disabled controls at matched speeds, unequivocal support for one or both viewpoints may be obtained.

Treadmill walking facilitates the collection of biomechanical data from hemiparetic and non-disabled subjects walking at matched speeds. The slow walking speeds of hemiparetic individuals are very unnatural for non-disabled individuals and difficult to achieve with steady cadences overground. Since walking at a prescribed speed on the treadmill only requires that subjects maintain their global position (i.e., relative to the room and handrails), the sensory cues provide strong feedback on proper speed, making it easier for hemiparetic and non-disabled subjects to walk at matched speeds with steady cadences. Since subjects do not advance in global space, a stationary motion camera setup can collect kinematic data over multiple gait cycles. Moreover, assuming the treadmill surface moves at a constant velocity (i.e., no slippage of the belt or deceleration of the motor), treadmill walking is mechanically equivalent to overground walking, given that air resistance is negligible (Ingen Schenau, 1980). These assumptions should be valid, especially at the slow walking speeds of hemiparetic individuals.

In this pilot study, we compared the gait of six individuals with post-stroke hemiparesis and six non-disabled controls while they walked on a treadmill at matched speeds that were comfortable for the hemiparetic subjects. We hypothesized that 1) a large set of gait differences will be consistent with impaired swing initiation and single limb support in the paretic limb and related compensatory strategies, and 2) swing time asymmetry is attributed to both increased percentage swing time in the paretic limb and reduced swing time in the non-paretic limb.

2. Methods

2.1 Subjects

Six individuals with a clinical presentation of a single cerebrovascular accident and resultant hemiparesis were selected from the VA Palo Alto Health Care System outpatient population. Inclusion criteria for this study were 1) clinical presentation of a single stroke at least six months prior to study, 2) ability to walk independently overground with use of an ankle foot orthosis (AFO) or assistive device, and 3) ability to advance the paretic limb independently while walking on a treadmill. Subjects were included independent of side of hemiparesis. Informed consent was obtained from all subjects. All procedures were approved by the Stanford University Human Subjects Committee. Each subject's lower extremity functional motor level was quantified using the Fugl-Meyer Assessment of motor function (Fugl-Meyer *et al.*, 1975). Six non-disabled individuals were recruited to serve as gender, age (within ± 10 years), height (within ± 6 cm), and weight (within ± 12 kg) matched controls for the hemiparetic subjects. The non-disabled controls exhibited normal joint range of motion and muscle strength and had no apparent gait abnormalities. Subject characteristics are presented in Table 1, ordered by the hemiparetic subjects' comfortable walking speeds on the treadmill.

Table 1: Hemiparetic (H1-6) and Non-Disabled (N1-6) Subject Characteristics

Subject No.	H1/N1	H2/N2	H3/N3	H4/N4	H5/N5	H6/N6	Mean (SD)
Gender	M/M	M/M	F/F	M/M	F/F	F/F	
Age (yrs)	52/60	66/60	64/67	68/72	56/58	56/49	60(7)/61(8)
Height (cm)	170/176	175/172	161/161	158/159	167/161	169/163	167(6)/165(7)
Weight (kg)	84/94	77/66	66/55	66/66	54/54	54/52	67(12)/64(12)
SSWS (cm/s)	22/111	33/138	59/149	72/113	56/146	77/147	53(22)/134(17)
CTS (cm/s)	13/13	31/31	36/36	36/36	45/45	45/45	34(12)/34(12)
Time post-stroke (months)	28	20	45	122	8	40	44 (41)
Affected side	R	R	L	L	R	L	
LE Fugl-Meyer (max=34)	16	20	16	24	27	22	21 (4)
Assistive device(s)	AFO cane	cane	AFO cane	AFO	cane		

Individual characteristics and group means and standard deviations (SD).
Abbreviations: LE, lower extremity; AFO, ankle-foot orthosis; SSWS, self-selected overground walking speed; CTS, comfortable treadmill speed.

2.2 Instrumentation

Subjects wore a Medical harness (Robertson Mountaineering, Henderson, NV) attached to an overhead support as they walked on a Rehabilitation treadmill (Biodex Medical Systems, Shirley, NY). During treadmill walking, the harness did not provide body weight support but served as a safety catch if subjects were to fall. Pedar insole pressure sensors (Novel, Munich, Germany) were placed inside the subjects' shoes to estimate the vertical ground reaction force and center of pressure. For subjects who wore an AFO, the sensors were placed inside the AFO.

Bilateral kinematics were captured at 50 Hz using a Qualisys Motion Analysis System (Qualisys Inc., East Windsor, CT), incorporating five digital ProReflex cameras. Eight clusters of three reflective markers were located on the upper trunk and pelvis and right and left thighs, shanks, and feet and calibrated to anatomical reference points to define each segment's position and orientation. A voltage signal coinciding with each camera exposure initialization was used to synchronize the insole pressure readings.

2.3 Protocol

Hemiparetic subjects and non-disabled controls walked on the treadmill at speeds comfortable for the hemiparetic subjects (range 13 to 45 cm/s, Table 1), as determined during single pre-sessions where hemiparetic subjects were familiarized to treadmill walking. Hemiparetic subjects who normally wore an AFO walked with the AFO on the treadmill. All subjects were asked to hold onto the handrails as the treadmill belt accelerated and release hand hold once the prescribed speed was reached. After subjects achieved steady state without handrail hold, data was collected for 20 seconds. Two subjects failed to walk for 20 seconds without handrail hold, but data for at least five complete gait cycles were collected from them.

2.4 Data reduction and analysis

The raw kinematic data, consisting of marker cluster coordinates, were post-processed in MARey (Center for Locomotion Studies, Penn State University, State College, Pennsylvania) to obtain knee flexion and ankle dorsiflexion angles; joint center trajectories of the hip, knee, and ankle; and anatomical trajectories of the acromion processes and tip of the second toe and heel of each foot. The joint center and anatomical trajectories were fitted to a 7-segment inertial model of each subject, consisting of a trunk (including the mass of the head and arms), two thighs, two shanks, and two feet (including the mass of the shoes), based on data collected by Dempster et al. (1955). Hip flexion/extension angle was defined to be the angle between the axes of the femur and trunk in the sagittal plane, which was defined by the mid-line between the hip joint centers and acromion processes. A measure of limb circumduction during swing was defined as the peak lateral displacement of the foot center of mass during swing relative to its lateral position during stance.

The post-processed kinematic data were interpolated from 50 to 100 Hz; filtered using a second-order Butterworth, low-pass filter (fc = 6 Hz); and differentiated to obtain segment linear and angular velocities. The kinetic (KE), potential (PEG), and mechanical (ME = KE + PEG) energies of the segments were then obtained from the segments' positions and velocities (Winter, 1979). Kinetic, potential, and mechanical energetic cost was defined as the summed positive increments in these energies during a gait interval or stride (Cavagna *et al.*, 1976) (see Figure 1). For visual inspection, the kinematic and energetic gait trajectories were normalized to percent gait cycle, beginning and ending with initial contact of each foot, and ensemble averaged over the gait cycles collected.

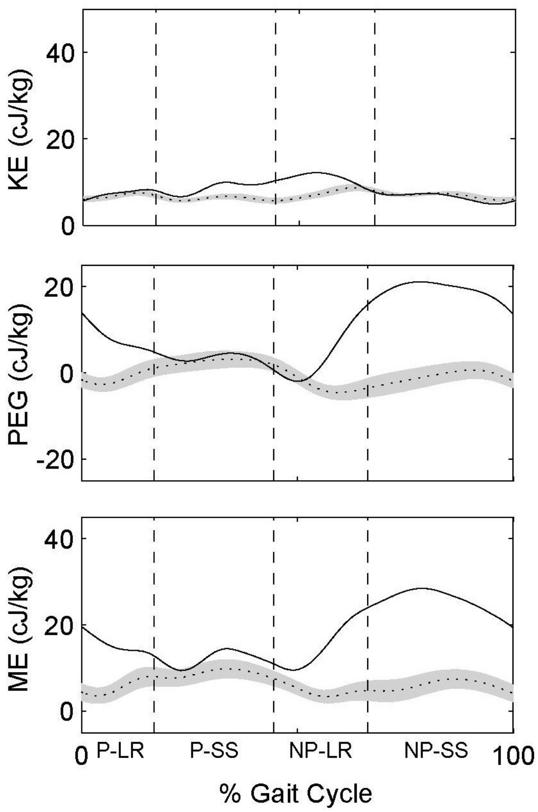

Figure 1. Kinetic energy (KE), potential energy of gravity (PEG), and mechanical energy (ME = KE + PEG) of the whole body during the gait cycle, beginning and ending with initial contact of the paretic limb, for hemiparetic subject H2 (solid line) and non-disabled control N2 (dotted line, shaded region ± SD across gait cycles collected). Initial contact of paretic limb is side-matched in the non-disabled control. Kinetic, potential, and mechanical energetic cost was defined to be the summed positive increments in these energies during a gait interval or stride. Note the large rise in ME and PEG during NP-LR and NP-SS, primarily attributed to the exaggerated increase in trunk height during pre-swing and swing of the paretic limb, consistent with pelvic hiking. Abbreviations: P-LR, paretic limb loading response; P-SS, paretic limb single support; NP-LR, non-paretic limb loading response; NP-SS, non-paretic limb single support. Vertical, dashed lines designate transitions between gait intervals for the hemiparetic subject.

Percent recovery in mechanical energy during a stride, through mirrored exchanges between kinetic and potential energies, was quantified using *Eq. 1*, modified from Cavagna et al. (1976):

$$\text{Percent Recovery} = 100\left(1 - \frac{ME_{\cos t}}{KE_{\cos t} + PEG_{\cos t}}\right) \tag{1}$$

In words, if the rises and falls in kinetic and potential energies during a gait cycle are anti-phased and cancel such that mechanical energetic cost is significantly lower than the sum of kinetic and potential energetic cost, percent recovery is high (e.g., 60-70% in non-disabled individuals walking at normal speeds) (Cavagna *et al.*, 1976).

The periods of stance and swing of each limb were determined using a threshold on the vertical ground reaction force (F > 35 N), as estimated by the insole pressure data (interpolated from 50 to 100 Hz). The ground reaction force center of pressure during loading response and pre-swing of each limb was defined as the weighted-average center of pressure during these phases of double limb support. Step length was defined as the forward distance between initial contact of one foot and the previous initial contact of the other foot.

Asymmetry in swing time was quantified using a modified version of an index (*Eq. 2*) proposed by Robinson et al. (1987):

$$Asymmetry\,(\%) = \frac{100\,(V_{paretic} - V_{non-paretic})}{\max(|V_{paretic}|, |V_{non-paretic}|)} \qquad (2)$$

where $V_{paretic}$ is the value of a gait parameter recorded for the paretic limb, and $V_{non\text{-}paretic}$ is the corresponding value for the non-paretic limb. The magnitude of the index represents the degree of asymmetry, and the sign indicates the direction of asymmetry. An index of zero indicates perfect symmetry. A positive index indicates a larger value of the gait parameter for the paretic limb, while a negative index indicates a larger value for the non-paretic limb. Since the direction of asymmetry was not always consistent across subjects in the hemiparetic and non-disabled groups, the value of the index (Asymmetry in Table 2) as well as its magnitude (Asymmetry Magnitude in Table 2) were considered.

2.5 Statistics

The joint-kinematic and segmental-energetic trajectories and temporal-spatial gait variables in the hemiparetic and non-disabled subjects were visually examined for differences between speed-matched pairs. Because of the small sample size in this pilot study, full statistical analyses on the data were not appropriate. However, consistencies were tested using a Wilcoxon signed-rank test (significance set at $p < 0.10$). With n = 6 speed-matched pairs, the signed-rank test was significant at $p < 0.10$ only when either all 6 differences were in the same direction ($p = 0.03$), or 5 out of 6 differences were in the same direction with the non-conforming difference being the smallest ($p = 0.06$) or second smallest in magnitude ($p = 0.09$). Thus, each of these three cases indicated strong and consistent differences between groups.

3. Results

3.1 Temporal and spatial gait variables

Table 2 presents the group means for temporal and spatial gait variables for the hemiparetic subjects and non-disabled controls. Walking speeds were nearly identical between groups and for each speed-matched pair. Cadence, stride time, and stride length were not appreciably different between groups. Step width was greater in the hemiparetic group ($p=0.06$). Percentage swing time of the paretic limb was increased relative to values in side-matched limbs in non-disabled controls ($p=0.06$, see also Figure 2), while swing time of the non-paretic limb was reduced ($p=0.06$, see also Figure 2). As a result, swing time asymmetry was much greater in the hemiparetic group ($p=0.03$, see also Figure 2). Step lengths and step length asymmetry were not statistically different between groups. However, this occurred because the direction of asymmetry was not consistent across the hemiparetic subjects. The magnitude of step length asymmetry was significantly greater in the hemiparetic group ($p=0.09$). The ground reaction force center of pressure during loading response of the paretic limb was more anterior than values in non-disabled controls ($p=0.06$).

Table 2: Temporal and Spatial Gait Variables

	Hemiparetic	Non-Disabled	p-value
Gait cycles collected	10.0 (3.8)	11.5 (3.3)	
Speed (cm/s)	34.6 (11.7)	34.4 (11.7)	
Cadence (steps/min)	83.4 (12.8)	78.9 (21.7)	
Stride time (s)	1.47 (0.21)	1.63 (0.50)	
Stride length (cm)	52.3 (22.0)	53.3 (18.8)	
Step width (cm)	17.3 (5.9)	11.5 (2.2)	0.06
Swing time (% gait cycle)			
Paretic limb	39.8 (4.6)	32.2 (10.3)	0.06
Non-paretic limb	21.5 (4.5)	31.4 (8.4)	0.06
Asymmetry (%)	43.4 (16.5)	-0.1 (11.3)	0.03
Asymmetry magnitude (%)	43.4 (16.5)	6.7 (8.5)	0.03
Step length (cm)			
Paretic limb	29.8 (12.2)	27.8 (10.6)	
Non-paretic limb	22.4 (14.4)	25.4 (8.4)	
Asymmetry (%)	27.4 (56.3)	6.2 (11.6)	
Asymmetry magnitude (%)	48.0 (36.1)	11.3 (5.2)	0.09
GRF A-P center of pressure			
Loading response (% insole length)			
Paretic limb	50.6 (14.7)	36.3 (14.1)	0.06
Non-paretic limb	37.9 (8.3)	34.9 (11.4)	
Pre-swing (% insole length)			
Paretic limb	69.6 (4.3)	71.4 (7.7)	
Non-paretic limb	70.3 (8.1)	71.5 (4.1)	

Group means and standard deviations (in parentheses).
Paretic and non-paretic limb variables are side-matched in non-disabled controls.
Abbreviations: GRF, ground reaction force, A-P, anterior-posterior.

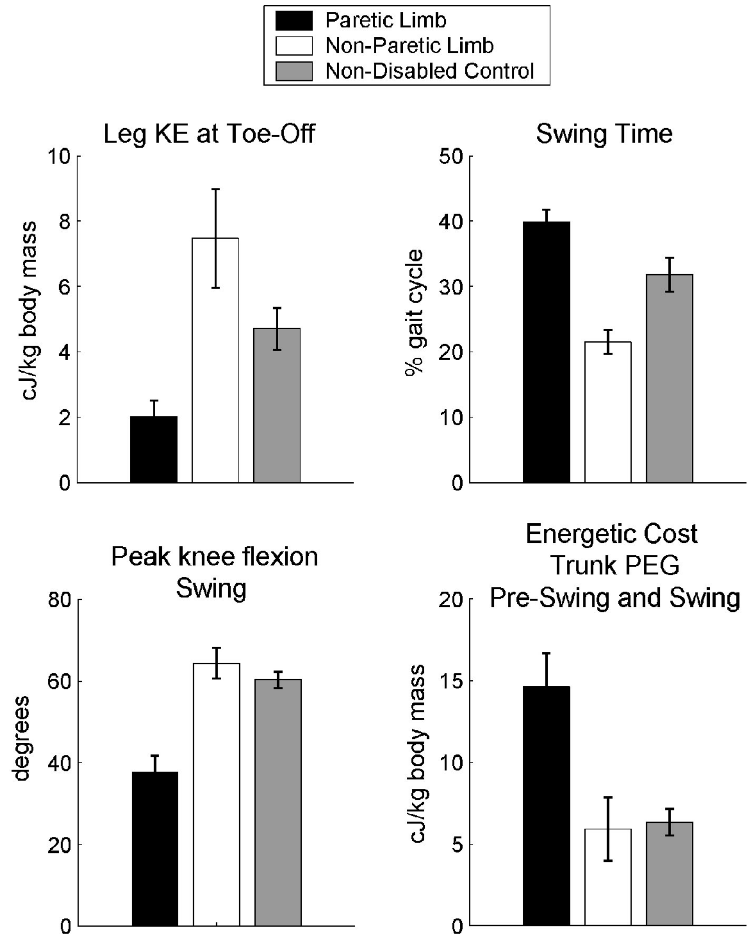

Figure 2. Significant gait differences between hemiparetic and non-disabled subjects. Leg kinetic energy at toe-off, percentage swing time, peak knee flexion during swing, and energetic cost associated with rises in trunk potential energy during pre-swing and swing of the paretic limb (solid bars), non-paretic limb (white bars), and in non-disabled controls (gray bars). Values are means ± SE. Means for non-disabled controls include values for both right and left limbs, since they did not differ appreciably. Leg kinetic energy at toe-off in the paretic limb was reduced compared to values in non-disabled controls, resulting in increased swing time and reduced peak knee flexion during swing. Leg kinetic energy at toe-off in the non-paretic limb was relatively high, resulting in shortened swing time. Energetic cost associated with raising the trunk during pre-swing and swing of the paretic limb was increased, consistent with pelvic hiking to clear the paretic limb with reduced knee flexion.

3.2 Mechanical energetic gait variables

Table 3 presents the group means for mechanical energetic gait variables for the hemiparetic subjects and non-disabled controls. Kinetic, potential, and mechanical energetic costs of transport were greater in the hemiparetic group (p=0.06, 0.03, and 0.03, respectively). Nonetheless, percent recovery between kinetic and potential energy was similar between groups. Mechanical energetic cost per stride was greater in the hemiparetic group (p=0.06), which was primarily attributed to increased energetic cost during pre-swing (NP-LR) and swing (NP-SS) of the paretic limb. Potential energetic cost associated with raising the trunk during pre-swing (NP-LR) and swing (NP-SS) of the paretic limb and kinetic energetic cost associated with accelerating the paretic limb during swing (NP-SS) were greater in the hemiparetic group (both p=0.03) (see also Figure 2). Leg kinetic energy at toe-off in the paretic limb was reduced compared to values in non-disabled controls (p=0.03), and leg kinetic energy at toe-off in the non-paretic limb was increased (p=0.09) (see also Figure 2).

Table 3: Mechanical Energetic Gait Variables

	Hemiparetic	Non-Disabled	p-value
Energetic cost of transport (cJ/kg m)			
Kinetic energy	35.4 (10.7)	20.9 (2.3)	0.06
Potential energy	58.8 (15.2)	36.4 (4.8)	0.03
Mechanical energy	67.9 (17.4)	42.5 (5.8)	0.03
Percent recovery (%)	28.2 (3.7)	26.3 (6.3)	
Energetic cost – Stride interval (cJ/kg)			
Non-paretic limb pre-swing (P-LR)	8.5 (6.9)	8.2 (4.2)	
Non-paretic limb swing (P-SS)	3.8 (2.1)	3.5 (1.3)	
Paretic limb pre-swing (NP-LR)	12.5 (6.7)	6.6 (3.3)	
Paretic limb swing (NP-SS)	7.4 (5.2)	4.0 (1.6)	0.09
Total: Stride	32.3 (11.4)	22.5 (8.1)	0.06
Component energetic cost (cJ/kg)			
Trunk PEG in P-LR and P-SS	5.9 (4.7)	6.7 (3.6)	
Non-paretic leg KE in P-SS	2.6 (1.6)	1.8 (0.8)	
Trunk PEG in NP-LR and NP-SS	14.6 (5.0)	6.1 (2.1)	0.03
Paretic leg KE in NP-SS	4.8 (2.8)	1.8 (1.0)	0.03
Leg kinetic energy at toe-off (cJ/kg)			
Paretic limb	2.0 (1.2)	4.5 (2.0)	0.03
Non-paretic limb	7.5 (3.7)	4.9 (2.6)	0.09

Group means and standard deviations (in parentheses).
Paretic and non-paretic limb variables are side-matched in non-disabled controls.
Abbreviations: P-LR, paretic limb loading response; P-SS, paretic limb single support; NP-LR, non-paretic limb loading response; NP-SS, non-paretic limb single support; PEG, potential energy of gravity; KE, kinetic energy.

3.3 Kinematic gait variables

Table 4: Kinematic Gait Variables

	Hemiparetic	Non-Disabled	p-value
Hip joint angle			
Angular excursion (deg)			
Paretic limb	31.4 (12.8)	31.4 (7.9)	
Non-paretic limb	36.4 (9.4)	31.0 (8.6)	
Peak extension (deg)			
Paretic limb	12.1 (10.0)	18.3 (5.4)	0.09
Non-paretic limb	16.6 (9.7)	18.4 (6.8)	
Knee joint angle			
Angular excursion (deg)			
Paretic knee limb	32.1 (14.2)	53.2 (14.0)	
Non-paretic limb	57.3 (12.6)	55.4 (14.5)	
Flexion at toe-off (deg)			
Paretic limb	26.0 (3.8)	40.0 (5.9)	0.03
Non-paretic limb	48.5 (7.4)	40.6 (6.9)	
Peak flexion – swing (deg)			
Paretic limb	37.8 (9.8)	58.6 (7.4)	0.03
Non-paretic limb	64.3 (9.3)	61.9 (6.4)	
Ankle joint angle			
Angular excursion (deg)			
Paretic limb	13.7 (6.3)	16.0 (5.0)	
Non-paretic limb	19.0 (5.8)	15.7 (6.0)	
Plantarflexion at toe-off (deg) †			
Paretic limb	-5.5 (9.4)	-5.3 (7.3)	
Non-paretic limb	-3.7 (9.3)	-5.1 (7.0)	
Foot lateral displacement – swing (cm)			
Paretic limb	4.6 (3.2)	1.5 (0.5)	0.03
Non-paretic limb	1.6 (1.5)	1.5 (0.4)	

Group means and standard deviations (in parentheses).
Paretic and non-paretic limb variables are side-matched in non-disabled controls.
† Negative values indicate ankle dorsiflexion.

Table 4 presents the group means for kinematic gait variables for the hemiparetic subjects and non-disabled controls. Peak hip extension in the paretic limb was reduced in the hemiparetic group (p=0.09), but the strongest differences were related to knee joint angles in the paretic limb. Knee flexion at toe-off and peak flexion during swing in the paretic limb were greatly reduced compared to values in non-disabled controls (both p=0.03) (see also Figure 2). Foot lateral displacement during swing of the paretic limb was greater than values in non-disabled controls (p=0.03).

4. Discussion

In support of our hypothesis, at matched walking speeds, a large set of gait differences between hemiparetic subjects and non-disabled controls was consistent with impaired swing initiation in the paretic limb and related compensatory strategies. Leg kinetic energy at toe-off in the paretic limb was reduced in the hemiparetic group, which resulted in increased energetic cost associated with accelerating the limb during swing and increased percentage swing time (see Figure 3 for illustration in subject H5).

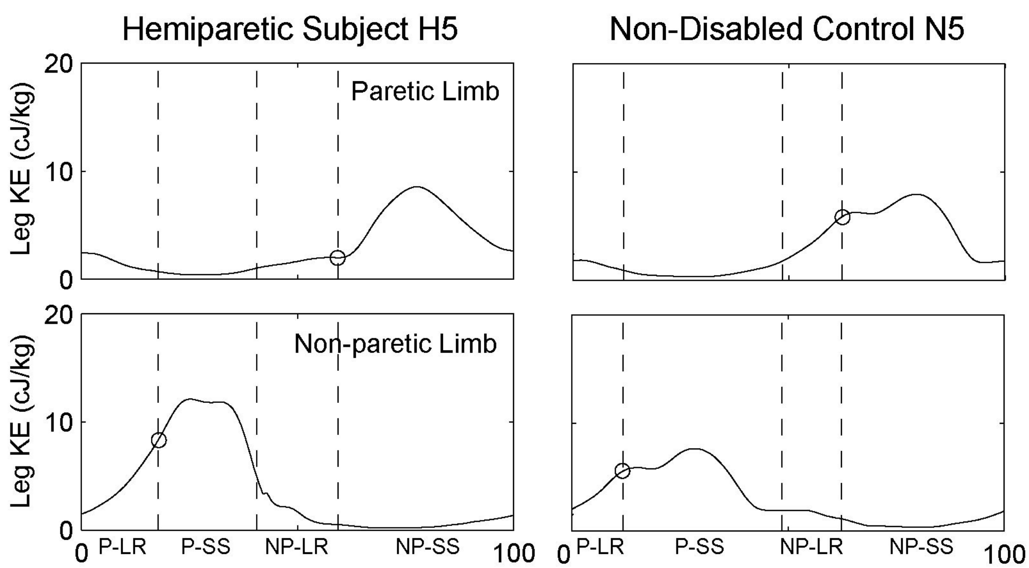

Figure 3. Leg kinetic energy (KE) of the paretic and non-paretic limbs for hemiparetic subject H5 and side-matched limbs in non-disabled control N5 during the gait cycle, beginning and ending with initial contact of the paretic limb. Note the lack of acceleration of the paretic limb during pre-swing (NP-LR), resulting in low leg kinetic energy at toe-off (marked by circle), and subsequent rise in leg kinetic energy during swing (NP-SS). In contrast, acceleration of the non-paretic limb was high during pre-swing (P-LR), resulting in increased leg kinetic energy at toe-off (marked by circle) and shortened swing time (P-SS). Abbreviations: P-LR, paretic limb loading response; P-SS, paretic limb single support; NP-LR, non-paretic limb loading response; NP-SS, non-paretic limb single support. Vertical, dashed lines designate transitions between gait intervals for each subject.

The low leg kinetic energy at toe-off in the paretic limb is presumably caused by inadequate leg propulsion by the plantarflexors (Neptune *et al.*, 2001; Nadeau *et al.*, 1999b; Hof *et al.*, 1993; Meinders *et al.*, 1998) or hip flexors (Nadeau *et al.*, 1999b; Nadeau *et al.*, 1999a) during pre-swing, which may limit how fast the paretic limb advances during swing and, consequently, gait speed. Moreover, compensatory swing-phase acceleration of the paretic limb, presumably by the hip flexors, could be energetically costly and compromise walking endurance.

Deviations in swing kinematics in the paretic limb were likely a consequence of inadequate leg kinetic energy at toe-off. Since leg acceleration during pre-swing is the result of flexion of the knee, which accelerates the thigh and shank forward relative to the trunk (illustrated in Figure 4), knee flexion at toe-off was not surprisingly reduced in the paretic limb. More importantly, reduced knee flexion at toe-off persisted through mid-swing, such that peak knee flexion was reduced. This impairment can interfere with limb clearance during swing. Indeed, using dynamical simulations of the swing phase of gait, Piazza and Delp (1996) found that reduced knee flexion velocity at toe-off can contribute to reduced peak knee flexion during swing. Since knee flexion velocity and leg kinetic energy are kinematically related at toe-off, our findings are consistent with the theoretical construct of these simulations.

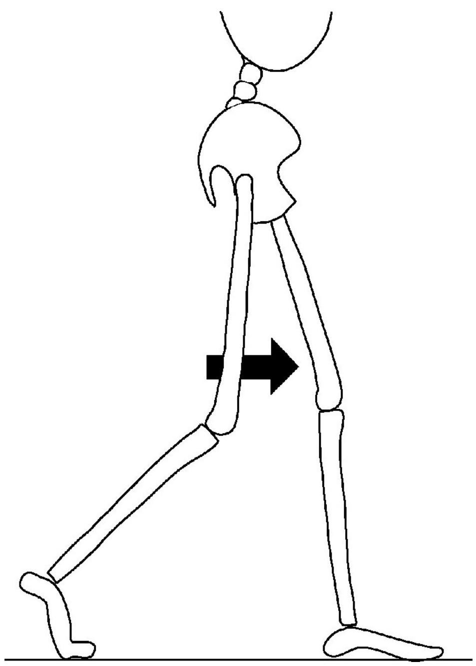

Figure 4. Limb acceleration during pre-swing is the result of flexion of the knee, which accelerates the thigh and shank forward relative to the trunk and pivoting foot. (Adapted from Perry, J. (1992) Gait Analysis Normal and Pathological Function. Thorofare, NJ: Slack).

The exaggerated energetic cost associated with raising the trunk during pre-swing and swing of the paretic limb was consistent with pelvic hiking (see Figure 1 for illustration in subject H2) to compensate for reduced peak knee flexion during swing in the paretic limb. Moreover, increased lateral displacement of the foot during swing in the paretic limb was consistent with limb circumduction to further assist limb clearance. Importantly, the increased positive work to raise the trunk contributed to increased mechanical energetic cost per stride. Similarly, Olney et al. (1986) found that excessive rises and falls in trunk potential energy tended to dominate total mechanical energetic cost during walking in hemiparetic individuals. They suggested that the restoration of normal kinetic and potential energy exchanges at the trunk was key to energy conservation. However, in our study, the increased energy cost associated with raising the trunk was primarily evident during pre-swing and swing of the paretic limb. This strategy presumably compensates for reduced knee flexion during swing in the paretic limb. Therefore, our data suggest that improvement of swing initiation in the paretic limb is an important objective of rehabilitation. Improved swing initiation may increase knee flexion during swing and allow the paretic limb to clear while reducing the energy cost associated with raising the trunk. These actions can improve energy efficiency during walking.

A second set of gait differences between hemiparetic subjects and non-disabled controls was consistent with impaired single limb support on the paretic limb and related compensatory strategies. The exaggerated leg kinetic energy at toe-off in the non-paretic limb resulted in reduced percentage swing time, which could be a consequence of weakness or poor balance during single limb support on the paretic limb (see Figure 3 for illustration in subject H5). Due to the high acceleration of the non-paretic limb during pre-swing, knee flexion at toe-off in the non-paretic limb was increased in the hemiparetic group. The restoration of single support time on the paretic limb, equivalent to swing time of the non-paretic limb, has been stressed in gait rehabilitation, because it is regarded as a positive training stimulus in improving locomotor function in hemiparetic individuals (Harris-Love *et al.*, 2001; Hesse *et al.*, 1999; Hassid *et al.*, 1997). The gait pattern requires that support and equilibrium be maintained over the paretic limb for a longer period of time and may provide a higher training stimulus for impaired equilibrium reflexes (Hesse, 1999).

In support of our second hypothesis, swing time asymmetry in the hemiparetic group was attributed to both increased percentage swing time in the paretic limb and reduced swing time in the non-paretic limb. Swing time asymmetry would appear to be an important measure of gait function in hemiparetic individuals, since it reflects the large disparity in leg kinetic energy at toe-off between the paretic and non-paretic limbs. The disparity is likely a consequence of both impaired swing initiation and single limb support in the paretic limb, which appeared to be related to many gait deviations in the hemiparetic group. Swing time symmetry has been found to be correlated with self-selected walking speed in hemiparetic individuals (Roth *et al.*, 1997; Kim and Eng, 2003; Titianova and Tarkka, 1995; Tyson, 1994), though the strength of correlation has varied in these studies.

Other gait differences were observed between hemiparetic subjects and non-disabled controls that did not appear to be directly related to impaired swing initiation or single limb support in the paretic limb. The ground reaction force center of pressure during loading response indicated initial contact of the paretic limb with the mid- to fore-foot, consistent with equinus (Perry *et al.*, 1978). Step width was greater in the hemiparetic subjects, which could be a consequence of poor balance. Step length asymmetry was prominent in the hemiparetic group, but the direction of asymmetry was not consistent across subjects. Four hemiparetic subjects exhibited a shorter step length in the non-paretic limb, and two exhibited a shorter step length in the paretic limb. Peak hip extension in the paretic limb was reduced in the hemiparetic group. The reduction was more evident in the subjects who exhibited shorter step length in the non-paretic limb, since their center of mass did not advance as far past the base of support during single limb support on the paretic limb. Kim and Eng (2003) also reported inconsistency in the direction of step length asymmetry and suggested that differences in compensatory strategies may increase or decrease step length of either limb and influence symmetry. Considering that the limbs advance forward during swing, it’s intriguing that the direction of step length asymmetry is quite inconsistent across hemiparetic individuals when the direction of swing time asymmetry is very consistent (i.e., swing time predominantly longer in the paretic limb) (Kim and Eng, 2003; von Schroeder *et al.*, 1995; Wall and Turnbull, 1986). Further research is needed to clarify the factors governing step length asymmetry.

Although percent recovery between kinetic and potential energies was equally low in the hemiparetic and non-disabled subjects, mechanical energetic cost of transport was greater in the hemiparetic group due to larger rises and falls in kinetic and potential energies during the gait cycle. In non-disabled individuals walking at normal speeds, energy recovery through pendulum-like exchanges between kinetic and potential energies has been credited to improve energy efficiency (Cavagna *et al.*, 2000; Cavagna *et al.*, 1977). However, since a jerky walking motion with large fluctuations in kinetic and potential energies is more energetically costly than a smooth motion, independent of percent recovery, the quantification of gait efficiency using energy recovery is probably inappropriate in pathological gait. Our results suggest that reducing costly accelerations and lifting of body segments in hemiparetic individuals can contribute more to energy efficiency than restoring energy exchanges typical of non-disabled individuals walking at faster speeds.

At matched walking speeds, many gait variables in the hemiparetic group were not appreciably different from values in non-disabled controls. For instance, cadence, stride time, and stride length were not significantly different. Joint angular excursions in either limb, except at the paretic knee joint, were not appreciably different; neither were ankle plantarflexion angles at toe-off in either limb. Moreover, energetic cost associated with the gait intervals corresponding to pre-swing and swing of the non-paretic limb were not different from values in non-disabled controls. Since many of these variables in the hemiparetic group would have undoubtedly differed from values in controls walking at normal speeds, we conclude that speed-matching using the treadmill was important to our study.

Even though treadmill and overground walking are believed to be mechanically equivalent (Ingen Schenau, 1980), the gait patterns exhibited by subjects during treadmill walking may differ from their natural gait patterns (Murray *et al.*, 1985; Alton *et al.*, 1998; Strathy *et al.*, 1983), particularly in hemiparetic individuals. Due to the limited width of the treadmill belt, the hemiparetic subjects who normally walk with a cane overground could not use a cane on the treadmill. Moreover, the treadmill speeds comfortable for the hemiparetic subjects were generally slower than their self-selected overground walking speeds. A number of factors could have contributed to the slower walking speeds on the treadmill, including 1) lack of an assistive device (i.e., a cane in four of the subjects), 2) differences in sensory experience associated with treadmill walking (e.g., visual flow of environment and traction of rubber treadmill belt), and 3) insecurity during a novel locomotor task. However, from visual inspection, the overall gait pattern of the hemiparetic subjects, particularly the asymmetrical nature, appeared to be preserved on the treadmill. Moreover, our study focused on non-speed-related gait differences between hemiparetic subjects and non-disabled controls, which minimized the significance of differences between comfortable speeds on the treadmill and overground.

While this study identified strong and consistent gait differences between hemiparetic individuals and non-disabled controls walking at matched treadmill speeds, future studies could attempt to identify gait deviations particular to a hemiparetic individual. However, conclusions about gait differences between a single hemiparetic subject and non-disabled control should be made cautiously, since gait patterns observed in our non-disabled subjects were variable, even at similar speeds. The variability in non-disabled gait patterns was mostly attributed to differences in cadence, which did not prevent us from identifying strong and consistent gait differences between the hemiparetic and non-disabled subjects. However, if gait deviations particular to a hemiparetic individual were of interest, the comparison should ideally be made between the hemiparetic subject and a group of non-disabled controls walking at the same speed or a single non-disabled control walking at the same speed and cadence. Further work is needed to determine the most appropriate comparisons between hemiparetic and non-disabled subjects, which can provide the most insight into the gait deficiencies and compensatory patterns of hemiparetic individuals.

In conclusion, the comparison of gait characteristics in hemiparetic and non-disabled subjects walking at matched treadmill speeds was useful in identifying non-speed-related gait differences that are likely associated with impairment, resulting from hemiparesis, and related compensatory strategies. More importantly, the comparisons allowed the identification of interrelationships between biomechanical variables and deviations in both the paretic and non-paretic limbs. At matched walking speeds, a large set of gait differences between hemiparetic and non-disabled subjects was consistent with impaired swing initiation and single limb support in the paretic limb and related compensatory strategies. These findings should be extended in a larger group of hemiparetic subjects and non-disabled controls.

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Study 2 **Gait deviations associated with post-stroke hemiparesis: Improvement during treadmill walking using weight support, speed, support stiffness, and handrail hold**

Abstract

By comparing gait characteristics in hemiparetic and non-disabled individuals at matched treadmill speeds, Chen et al. (2003) identified non-speed-related gait deviations associated with post-stroke hemiparesis 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 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 identified. With increased body weight support or the addition of handrail hold, exaggerated leg kinetic energy at toe-off and reduced percentage swing time in the non-paretic limb were improved, resulting in improved temporal symmetry and increased single limb support time on the paretic limb. With increased treadmill speed, leg kinetic energy at toe-off in the paretic limb increased but still remained low relative to values in the non-paretic limb. With increased support stiffness, exaggerated energy cost associated with raising the trunk during pre-swing and swing of the paretic limb was improved. The proper selection of training parameters can improve the gait pattern practiced by individuals with hemiparesis during treadmill training and may improve treatment outcome.

G. Chen, C. Patten, D.H. Kothari, F.E. Zajac. Submitted to *Gait & Posture*, 2003

1. Introduction

After suffering a stroke, many individuals are left with neurological and functional deficits, including hemiparesis, which impair their ability to walk. Jorgensen et al. (1995) studied 804 acute, hospitalized stroke patients and found that 63% of them could not walk independently prior to rehabilitation. Moreover, many individuals who achieved independent gait were still hindered by reduced walking speed and limited endurance. Hemiparetic gait has been characterized by reduced speed, cadence, stride length, and joint angular excursions (von Schroeder *et al.*, 1995; Wall and Ashburn, 1979; Dean *et al.*, 2000); asymmetry in temporal, spatial, kinematic, and kinetic gait variables (Brandstater *et al.*, 1983; Roth *et al.*, 1997; Kim and Eng, 2003; Wall and Turnbull, 1986); and increased mechanical energetic cost (Olney *et al.*, 1986; Iida and Yamamuro, 1987). The improvement of these gait deviations has been stressed in gait rehabilitation, since it may enhance locomotor function in hemiparetic individuals. However, many of these gait deviations are likely the consequence of marked reductions in gait speed and may not be associated with factors that limit gait performance in hemiparetic individuals.

By comparing gait characteristics of hemiparetic and non-disabled individuals walking at matched treadmill speeds, Chen et al. (2003) identified non-speed-related gait deviations associated with hemiparesis. Leg kinetic energy at toe-off in the paretic limb was found to be reduced, resulting in increased percentage swing time and reduced peak knee flexion during swing, consistent with inadequate leg propulsion by the plantarflexors or hip flexors during swing initiation. 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, resulting in reduced percentage swing time, consistent with weakness or poor balance during single limb support on the paretic limb. The improvement of these non-speed-related gait deviations associated with post-stroke hemiparesis 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 (Hesse, 2001;

Hesse *et al.*, 1995b; Hesse *et al.*, 1995a; Hesse *et al.*, 1994; Sullivan *et al.*, 2002; Visintin *et al.*, 1998). With the assistance of the harness and treadmill, many hemiparetic individuals are able to walk with more normal gait kinematics and EMG timing (Hesse *et al.*, 1999) and improved symmetry (Hesse *et al.*, 1997; Hesse *et al.*, 1999; Hassid *et al.*, 1997). 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 (Hesse, 1999; Dobkin, 1999).

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 (Chen *et al.*, 2003) during treadmill walking in a pilot group of subjects. Body weight support and treadmill speed are currently used as training parameters, and their immediate impact on temporal gait variables has been studied (Harris-Love *et al.*, 2001; Hassid *et al.*, 1997; Hesse *et al.*, 1999). However, their potential effect on kinematic and energetic gait variables has not been quantified. Handrail hold has generally been provided to individuals for security, but its importance to training has not been well documented. Nevertheless, handrail hold, by itself, or as an adjunct to harness support, can provide additional mechanical-sensory input that may improve the gait pattern exhibited by individuals. Support stiffness is generally predetermined by the harness support system. Some systems are relatively compliant, allowing significant vertical motion of the trunk (Hesse *et al.*, 1999; Sullivan *et al.*, 2002), and others are very rigid (Barbeau *et al.*, 1987; Miller *et al.*, 2002). If the proper selection of training parameters (i.e., level of body weight support, treadmill speed, support stiffness, and handrail hold) can improve specific gait deviations associated with post-stroke hemiparesis, the parameters can be prescribed individually, or as a set, to improve the gait pattern practiced by hemiparetic individuals on the treadmill. These actions, ostensibly, may improve treatment outcome.

2. Methods

2.1 Subjects

We recruited six individuals with a clinical presentation of a single stroke and resultant hemiparesis from the VA Palo Alto Health Care System outpatient population. Inclusion criteria for this study were 1) clinical presentation of a stroke at least six months prior to study, 2) ability to walk independently overground with use of an ankle-foot orthosis (AFO) or assistive device, and 3) ability to advance the paretic limb independently while walking on a treadmill. The protocol was approved by the Stanford University Human Subjects Committee, and informed consent was obtained from the subjects. The Fugl-Meyer Assessment (Fugl-Meyer *et al.*, 1975) was used to quantify each subject's lower extremity functional motor level. Subject characteristics are presented in Table 1, ordered by the subject's comfortable walking speed on the treadmill.

Table 1: Subject Characteristics

Subject No.	1	2	3	4	5	6	Mean (SD)
Gender	M	M	F	M	F	F	
Age (yrs)	52	66	64	68	56	56	60 (7)
Height (cm)	170	175	161	158	167	169	167 (6)
Weight (kg)	84	77	66	66	54	54	67 (12)
Self-selected overground walking speed (cm/s)	22	33	59	72	56	77	53 (22)
Comfortable treadmill speed (cm/s)	13	31	36	36	45	45	34 (12)
Time post-stroke (months)	28	20	45	122	8	40	44 (41)
Affected side	R	R	L	L	R	L	
LE Fugl-Meyer (max = 34)	16	20	16	24	27	22	21 (4)
Assistive device(s)	AFO cane	cane	AFO cane	AFO		cane		

Individual characteristics and group means and standard deviations (SD).
Abbreviations: LE, lower extremity; AFO, ankle-foot orthosis.

2.2 Instrumentation

Subjects wore a Medical harness (Robertson Mountaineering, Henderson, NV) attached to a custom-made support frame (Chen *et al.*, 2001) as they walked on a Rehabilitation treadmill (Biodex Medical Systems, Shirley, NY). The level of body weight support provided by the harness was measured by an ATI force-torque sensor (ATI Industrial Automation, Apex, NC). Harness-support stiffness was adjusted by connecting music wire springs (Century Spring Corporation, Los Angeles, CA; K = 35.1 N/cm) in series with the cable. Body weight support was adjusted by stretching out the springs using a mechanical winch at the bottom of the frame. Pedar insole pressure sensors (Novel, Munich, Germany) were placed inside the subjects' shoes to estimate the vertical ground reaction force. For subjects who use an AFO, the sensors were placed inside the AFO.

A Qualisys Motion Analysis System (Qualisys Inc., East Windsor, CT), incorporating five digital ProReflex cameras, was used to capture bilateral kinematics at 50 Hz during treadmill walking. Reflective marker clusters were affixed to the upper trunk and pelvis and right and left thighs, shanks, and feet. The clusters were calibrated to anatomical reference points to define each segment's position and orientation. A voltage signal from the camera was used to synchronize the insole pressure readings.

2.3 Protocol

The subjects walked on the treadmill with different selections of body weight support, treadmill speed, support stiffness, and handrail hold as prescribed by ten experimental conditions (see Table 2), which were presented in a randomized order. Body weight support was set to 20%, 35%, or 50% by changing the length of the support cable with a mechanical winch and locking the length when the support force was at the prescribed level while subjects stood on the treadmill. 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. Support stiffness was set to 11.7 or 35.1 N/cm or made rigid by connecting 3, 1, or 0 springs in series with the support cable. Handrail hold was allowed using the subject's non-paretic arm. 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 (see Table 2). In addition, one condition of free treadmill walking, without harness support and handrail hold, was tested at the default speed (see Table 2).

Table 2: Experimental Treadmill Conditions

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

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.
* variation in BWS, speed, stiffness, or handrail hold.
Abbreviations: BWS, body weight support; CTS, comfortable treadmill speed; n/a, not applicable (harness support not provided).

Subjects who normally wore an AFO walked with the AFO on the treadmill. In conditions without handrail hold, the subjects were asked to hold onto the handrail as the treadmill belt accelerated and release the hold after the prescribed speed was reached. After subjects achieved steady state, data were collected for 20 seconds. A few subjects failed to walk for 20 seconds in a small number of trials, but data for at least five complete gait cycles were collected in each of these trials.

2.4 Data reduction and analysis

The procedures for data reduction and analysis were described in detail in Chen et al. (2003). In summary, the raw kinematic data were post-processed in MARey (Center for Locomotion Studies, Penn State University, State College, Pennsylvania) to obtain knee flexion angles and joint center and anatomical trajectories, which were fitted to a 7-segment inertial model of each subject, consisting of a trunk (including the mass of the head and arms), two thighs, two shanks, and two feet.

The post-processed kinematic data were differentiated to obtain segment linear and angular velocities. The kinetic (KE), potential (PEG), and mechanical (ME = KE + PEG) energies of the segments were then obtained from the segment's positions and velocities (Winter, 1979). The periods of stance and swing of each limb were determined using a threshold on the vertical ground reaction force (F > 35 N), as estimated by the insole pressure data. Energetic cost associated with raising the trunk during pre-swing and swing of the paretic limb was defined to be the summed positive increments in trunk potential energy during these intervals (Cavagna *et al.*, 1976) (see Figure 1).

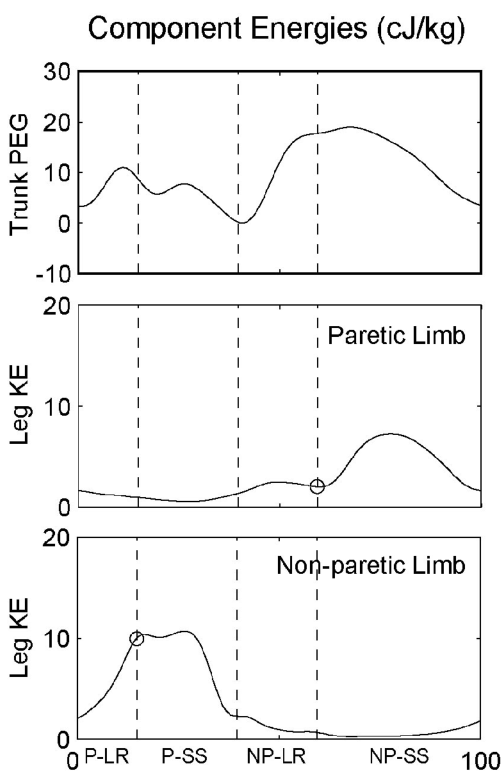

Figure 1. Trunk potential energy (PEG) and leg kinetic energies (KE) of the paretic and non-paretic limbs for subject 6 during free treadmill walking normalized to percent gait cycle, beginning and ending with initial contact of the paretic limb. Energetic cost associated with raising the trunk during pre-swing and swing of the paretic limb (NP-LR and NP-SS, respectively) was defined to be the summed positive increments in trunk potential energy during these intervals. Note the lack of acceleration of the paretic limb during pre-swing (NP-LR), resulting in low leg kinetic energy at toe-off (marked by circle) and prolonged swing time (NP-SS). In contrast, acceleration of the non-paretic limb was relatively high during pre-swing (P-LR), resulting in increased leg kinetic energy at toe-off (marked by circle) and shortened swing time (P-SS). Abbreviations: P-LR, paretic limb loading response; P-SS, paretic limb single support; NP-LR, non-paretic limb loading response; NP-SS, non-paretic limb single support. Vertical, dashed lines designate transitions between gait intervals.

Asymmetry in swing time was quantified using a modified version of an index (*Eq. 1*) proposed by Robinson et al. (1987):

$$Asymmetry\,(\%) = \frac{100\,(V_{paretic} - V_{non-paretic})}{\max(|V_{paretic}|,|V_{non-paretic}|)} \quad (1)$$

where $V_{paretic}$ is the value of a gait parameter recorded for the paretic limb, and $V_{non\text{-}paretic}$ is the corresponding value for the non-paretic limb. The magnitude of the index represents the degree of asymmetry, and the sign indicates the direction of asymmetry. An index of zero indicates perfect symmetry.

2.5 Statistics

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

3.1 Harness-supported vs. free treadmill walking

Table 3 presents the group means for gait variables during harness-supported and free treadmill walking. With harness support, leg kinetic energy at toe-off in the non-paretic limb was reduced ($p = 0.03$, see also Figure 2), resulting in increased percentage swing time ($p = 0.03$, see also Figure 2) and improved swing time symmetry ($p = 0.03$, see also Figure 2), which brought these variables closer to values in non-disabled controls from Chen et al. (2003) (see also Figure 2). However, leg kinetic energy at toe-off, swing time, and peak knee flexion during swing in the paretic limb were unchanged with harness support. A tendency towards reduced energetic cost associated with raising the trunk during pre-swing and swing of the paretic limb was observed with harness support, but this difference did not reach statistical significance (see Table 3, Component energetic cost – Trunk PEG in NP-LR and NP-SS).

Table 3: Comparison of Harness-Supported vs. Free Treadmill Walking

	Free Walking	Harness Support	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 (deg)				
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).
Each condition was tested at the default treadmill speed (100% CTS) without handrail hold. Harness support was provided at the default level of body weight support (35%) and support stiffness (35.1 N/cm).
Paretic and non-paretic limb variables are side-matched in non-disabled, speed-matched controls from Chen et al. (2003).
p = significance between free and harness-supported walking conditions (Wilcoxon signed-rank test).
Abbreviations: PEG, potential energy of gravity; NP-LR, non-paretic limb loading response; NP-SS, non-paretic limb single support.

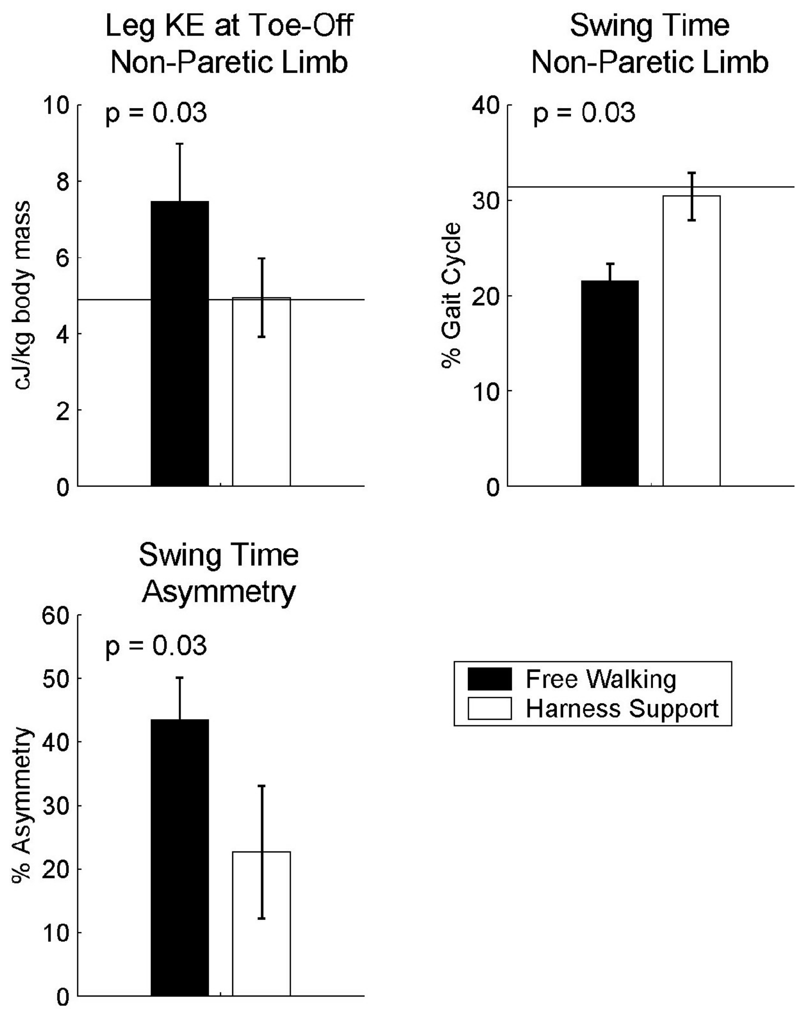

Figure 2. Significant gait differences between free and harness-supported treadmill walking (solid and white bars, respectively) – leg kinetic energy at toe-off and percentage swing time in the non-paretic limb and swing time asymmetry. Values are means ± SE. p = significance between free and harness-supported walking conditions (Wilcoxon signed-rank test). Horizontal lines designate values during free treadmill walking in non-disabled, speed-matched controls from Chen et al. (2003). 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, resulting in increased swing time and improved swing time symmetry.

3.2 Influence of body weight support

With increased body weight support, percentage swing time of the non-paretic limb increased (multivariate $p = 0.03$, see Figure 3), resulting in improved swing time symmetry (multivariate $p = 0.03$, see Figure 3). The differences between 20% and 35%

body weight support were more evident than between 35% and 50% body weight support (see Figure 3). However, leg kinetic energy at toe-off, swing time, and peak knee flexion during swing in the paretic limb and energetic cost associated with raising the trunk were not strongly affected by body weight support. A tendency towards reduced leg kinetic energy at toe-off in the non-paretic limb was observed with increased body weight support (multivariate p = 0.07). A summary of the effect of body weight support is presented in Table 4.

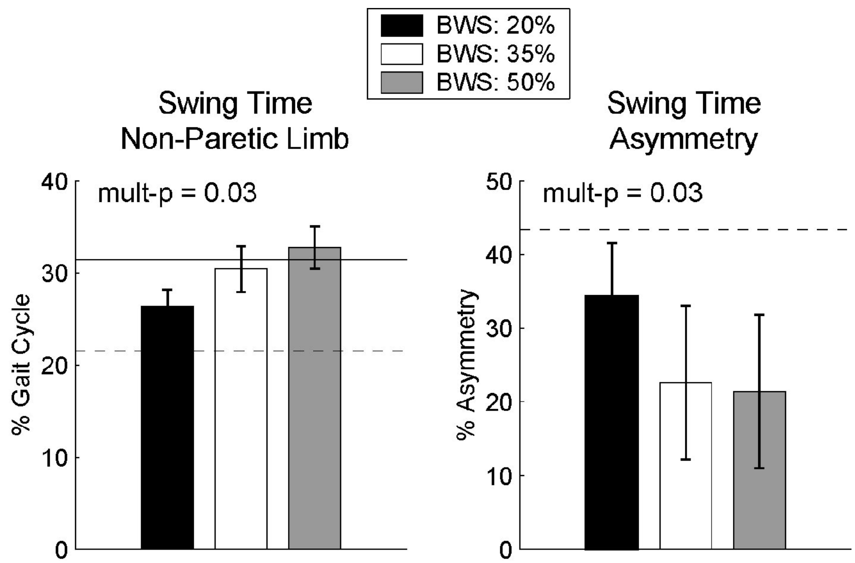

Figure 3. Significant effects of body weight support during treadmill walking – percentage swing time of the non-paretic limb and swing time asymmetry. Each condition was tested at the default level of treadmill speed (100% CTS), support stiffness (35.1 N/cm), and handrail hold (none). Values are means ± SE. mult-p = multivariate significance between body weight support conditions (Friedman's method for randomized blocks). Horizontal lines designate values during free treadmill walking in the subjects (dashed lines) and non-disabled, speed-matched controls from Chen et al. (2003) (solid lines). Paretic and non-paretic limb variables are side-matched in control subjects. 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.

Table 4: 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
Abbreviations: BWS, body weight support; PEG, potential energy of gravity; NP-LR, non-paretic limb loading response; NP-SS, non-paretic limb single support

3.3 Influence of treadmill speed

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; see Figure 4) but remained asymmetrical. Energetic cost associated with raising the trunk increased with speed (multivariate p = 0.03; see Figure 4, Energetic cost – Trunk PEG in NP-LR and NP-SS), probably due to larger vertical displacements of the trunk associated with increased stride length. Percentage swing time and peak knee flexion during swing in the paretic limb were not appreciably affected by treadmill speed. A tendency towards increased swing time of the non-paretic limb and improved swing time symmetry was observed with increased treadmill speed. A summary of the effect of treadmill speed is presented in Table 4.

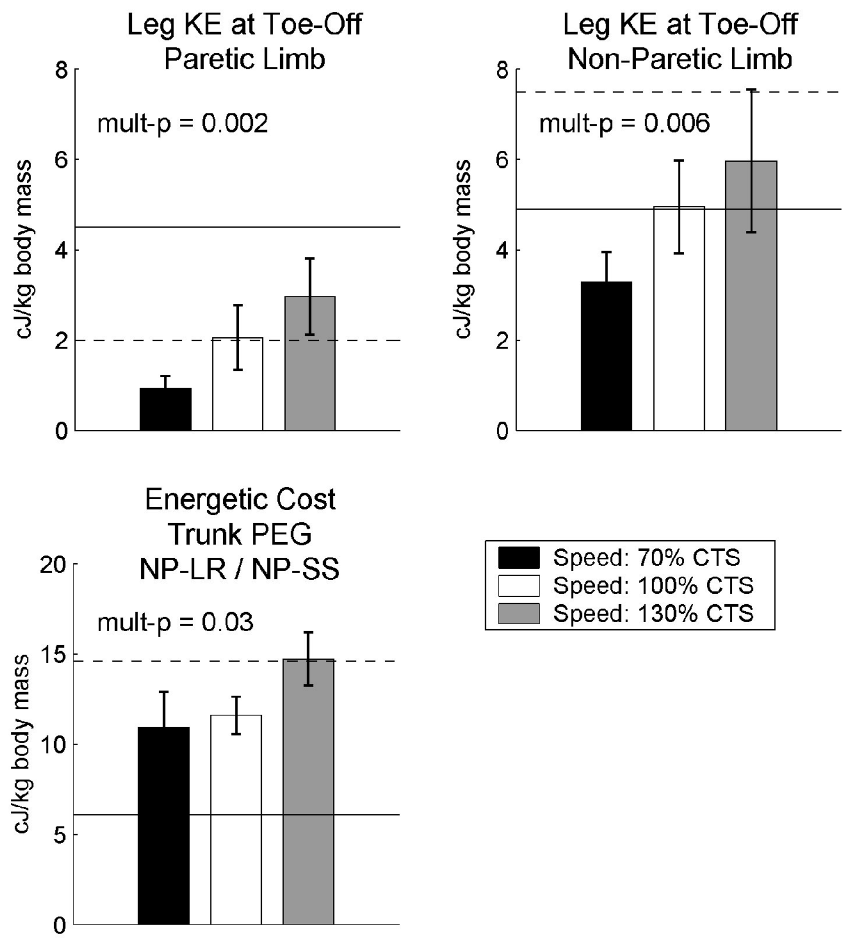

Figure 4. Significant effects of treadmill speed during supported treadmill walking – leg kinetic energy at toe-off in the paretic and non-paretic limbs and energetic cost associated with rises in trunk potential energy during pre-swing (NP-LR) and swing (NP-SS) of the paretic limb. Each condition was tested at the default level of body weight support (35%), support stiffness (35.1 N/cm), and handrail hold (none). Values are means ± SE. mult-p = multivariate significance between treadmill speed conditions (Friedman's method for randomized blocks). Horizontal lines designate values during free treadmill walking in the subjects (dashed lines) and non-disabled, speed-matched controls from Chen et al. (2003) (solid lines). Paretic and non-paretic limb variables are side-matched in control subjects. With increased treadmill speed, leg kinetic energy at toe-off in the paretic and non-paretic limbs increased but remained asymmetrical. 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.

3.4 Influence of support stiffness

With increased support stiffness, energetic cost associated with raising the trunk was reduced (multivariate p = 0.002; see Figure 5, Energetic cost – Trunk PEG in NP-LR and NP-SS), probably due to restriction of vertical displacements of the trunk. 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 the differences were small and probably not clinically meaningful. Leg kinetic energy at toe-off and percentage swing time in the non-paretic limb and swing time and peak knee flexion during swing in the paretic limb were not consistently affected by support stiffness. A tendency towards greater swing time symmetry at the two higher levels of support stiffness (i.e., 35.1 N/cm and rigid support) was observed. A summary of the effect of support stiffness is presented in Table 4.

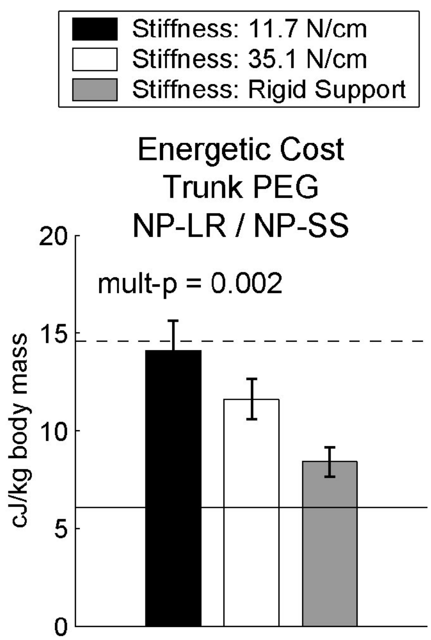

Figure 5. Significant effect of support stiffness during treadmill walking – energetic cost associated with the rises in trunk potential energy during pre-swing (NP-LR) and swing (NP-SS) of the paretic limb. Each condition was tested at the default level of body weight support (35%), treadmill speed (100% CTS), and handrail hold (none). Values are means ± SE. mult-p = multivariate significance between support stiffness conditions (Friedman's method for randomized blocks). Horizontal lines designate values during free treadmill walking in the subjects (dashed lines) and non-disabled, speed-matched controls from Chen et al. (2003) (solid lines). Paretic and non-paretic limb variables are side-matched in control subjects. With increased support stiffness, energetic cost associated with rises in trunk potential energy decreased, probably due to restriction of vertical displacements of the trunk.

3.5 Influence of handrail hold

Across the conditions of handrail hold, harness support, and combined harness support and handrail hold, swing time of the non-paretic limb increased (multivariate p = 0.03, see Figure 6), resulting in improved swing time symmetry (multivariate p = 0.03, see Figure 6). However, leg kinetic energy at toe-off, swing time, and peak knee flexion during swing in the paretic limb were unchanged across these conditions. A tendency towards reduced leg kinetic energy at toe-off in the non-paretic limb (multivariate p =

0.07) and reduced energetic cost associated with raising the trunk (multivariate p = 0.07) were observed across conditions. A summary of the effect of handrail hold is presented in Table 4.

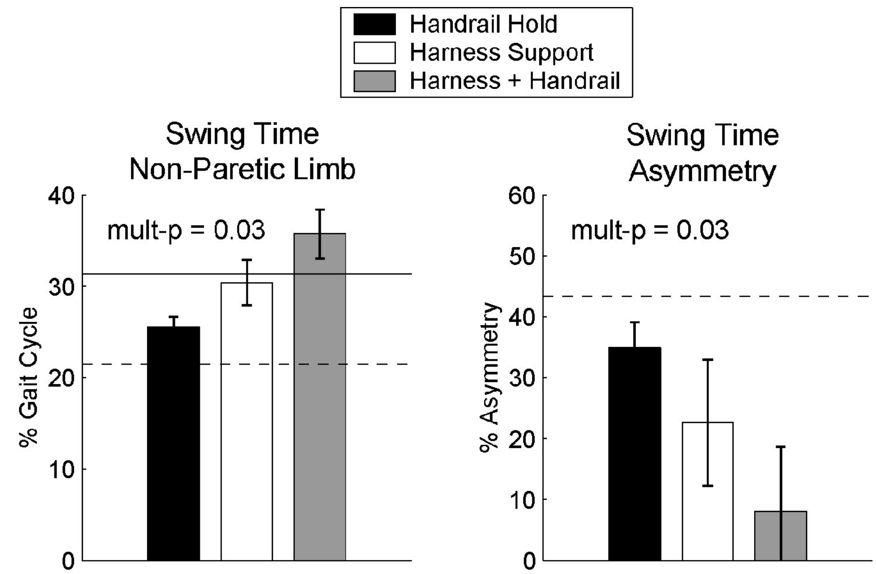

Figure 6. Significant effects across the conditions of treadmill walking with handrail hold, harness support, and combined harness support and handrail hold – percentage swing time of the non-paretic limb and swing time asymmetry. Each condition was tested at the default level of treadmill speed (100% CTS). Harness support was provided at the default level of body weight support (35%) and support stiffness (35.1 N/cm). Values are means ± SE. mult-p = multivariate significance between the three conditions (Friedman's method for randomized blocks). Horizontal lines designate values during free treadmill walking in the subjects (dashed lines) and non-disabled, speed-matched controls from Chen et al. (2003) (solid lines). Paretic and non-paretic limb variables are side-matched in control subjects. 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.

4. Discussion

The adjustment of each training parameter was found to improve a specific set of gait deviations associated with post-stroke hemiparesis (see Table 4). Our findings provide a rationale for the proper selection of training parameters during treadmill training in hemiparetic individuals. With increased body weight support by the harness or the addition of handrail hold, the reduced percentage swing time in the non-paretic limb improved, resulting in improved swing time symmetry. These changes appeared to result from a reduction in the exaggerated leg kinetic energy at toe-off in the non-paretic limb ($p = 0.03$ for harness-supported vs. free treadmill walking, multivariate $p = 0.07$ for both the effect of handrail hold and level of body weight support). With increased treadmill speed, the reduced leg kinetic energy at toe-off in the paretic limb improved 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 improved.

Increased percentage swing time of the non-paretic limb during treadmill training corresponds to increased single limb support time on the paretic limb and may be important to restoring locomotor function in hemiparetic individuals. The gait pattern requires that support and equilibrium be maintained over the paretic limb for a longer period of time and may provide a higher training stimulus for impaired equilibrium reflexes (Hesse, 1999; Harris-Love *et al.*, 2001; Hassid *et al.*, 1997; Hesse *et al.*, 1999). Previous studies have also reported increased single limb support time on the paretic limb during treadmill walking with harness support (Hassid *et al.*, 1997; Hesse *et al.*, 1999; Hesse *et al.*, 1997). 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 (Brandstater *et al.*, 1983; Roth *et al.*, 1997; Kim and Eng, 2003; Wall and Turnbull, 1986; Chen *et al.*, 2003) and is likely due to weakness or poor balance during support on the paretic limb. Both harness support and handrail hold assist in weight support and lateral balance, which may allow hemiparetic individuals to achieve longer support time on the paretic limb.

Increased treadmill speed elicited 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. Inadequate acceleration of the paretic limb during pre-swing, presumably caused by inadequate propulsion by the plantarflexors (Nadeau *et al.*, 1999a; Nadeau *et al.*, 1999b; Neptune *et al.*, 2001; Hof *et al.*, 1993) or hip flexors (Nadeau *et al.*, 1999a; Nadeau *et al.*, 1999b), 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 attributed to inadequate leg kinetic energy at toe-off (Chen *et al.*, 2003). However, in our study, the increased leg kinetic energy at toe-off in the paretic limb at higher treadmill speeds did not lead to improved swing time or peak knee flexion during swing. We suggest that these deviations in swing kinematics did not improve at faster treadmill speeds 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, 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 gait rehabilitation because of its strong contribution to energy efficiency (Olney *et al.*, 1986) and the sensory experience associated with walking (Gordon *et al.*, 2000; Hesse and Uhlenbrock, 2000). 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, since 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 presumably compensated for reduced knee flexion in the limb, 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, most of the subjects preferred a compliant support, because a rigid support was uncomfortable.

The increased swing time and reduced peak knee flexion during swing in the paretic limb were not improved with the adjustment of training parameters (see Table 4) and probably need to be addressed using manual assistance. 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 (Hesse, 1999; Hesse *et al.*, 2001; Hesse *et al.*, 1994; Visintin *et al.*, 1998; Sullivan *et al.*, 2002), which suggests that treadmill training may not adequately facilitate the swing phase of the paretic limb. Manual assistance greatly increases the physical demand on therapists and has driven the push for fully mechanized gait trainers (Hesse *et al.*, 2000; Uhlenbrock *et al.*, 1997) and powered orthoses (Ferris *et al.*, 2001; Jezernik *et al.*, 2001) 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 should, perhaps, be provided to insure proper kinematics of the paretic limb during swing, even though the subjects can advance the limb independently.

In treadmill training of less-impaired individuals, handrail hold could be a reasonable alternative to harness support, since handrail hold, by itself, significantly improved the same gait deviations that were also improved by harness support. Indeed, treadmill training without harness support has been found to enhance locomotor ability overground in hemiparetic individuals (Silver *et al.*, 2000; Laufer *et al.*, 2001) and improve their cardiovascular fitness (Macko *et al.*, 2001; Macko *et al.*, 1997). Treadmill walking has also been found to induce a more consistent and symmetric gait pattern as compared to overground walking at the same speed (Harris-Love *et al.*, 2001). Handrail hold was provided in all of these studies (Silver *et al.*, 2000; Laufer *et al.*, 2001; Macko *et al.*, 2001; Macko *et al.*, 1997; Harris-Love *et al.*, 2001), although its importance was not emphasized. Handrail hold assists with weight support and stability and provides the subject with a constant-velocity, tactile reference frame relative to the treadmill surface. Based on our finding that handrail hold significantly improved the same gait deviations that harness support did, we believe that handrail hold has much potential to elicit an improved gait pattern during treadmill walking in hemiparetic individuals, especially when harness support is not provided.

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 extended 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, as reflected by an improvement in these gait deviations, or an increase in walking speed or endurance. Ultimately, the importance of these factors to treatment outcome in hemiparetic individuals needs to be verified in clinical trials. Third, our study only examined the independent effect of adjusting each training parameter because of the number of parameters considered and the large number of trials that would have been required to examine the interaction of these effects. Nevertheless, interactive effects from simultaneous adjustment of training parameters (e.g., the effect of treadmill speed at each level of body weight support) may be important and should be examined in future studies. 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 swing time of the non-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 were not improved with the adjustment of training parameters and probably need to be addressed using manual assistance. The practice of an improved gait pattern during treadmill training, as defined by a reduction in these non-speed-related gait deviations, may improve treatment outcome in hemiparetic individuals.

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CONCLUSION

Summary

We conducted two studies that provide a scientific basis for the selection of parameters for treadmill training in individuals with post-stroke hemiparesis. In Study 1, gait characteristics in hemiparetic and non-disabled individuals were compared at matched treadmill speeds to identify non-speed-related gait deviations associated with post-stroke hemiparesis. Many of the gait deviations were consistent with impaired swing initiation and single limb support in the paretic limb and related compensatory strategies (see Figure 1).

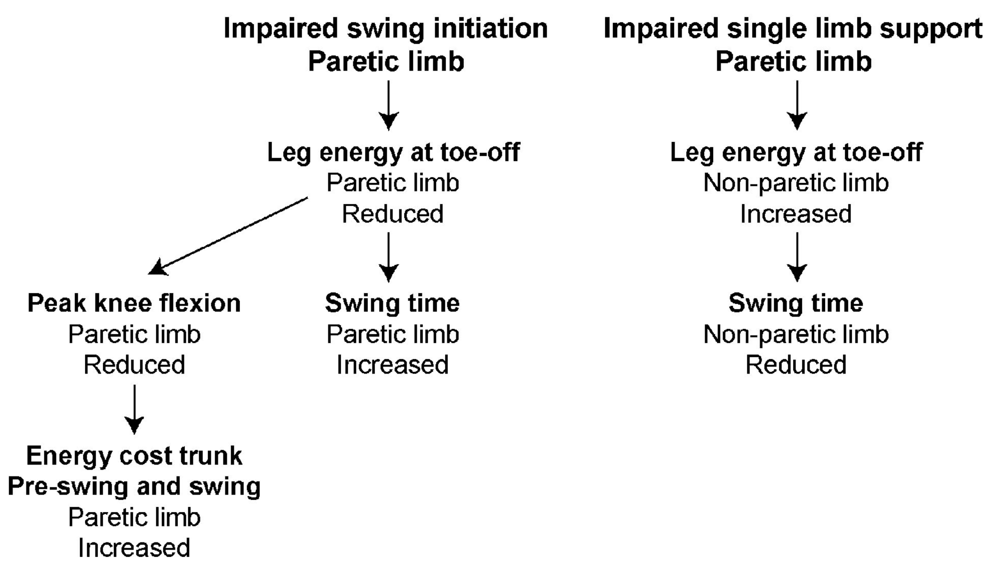

Figure 1. Summary of non-speed-related gait deviations associated with post-stroke hemiparesis. Many deviations were consistent with impaired swing initiation and single limb support in the paretic limb.

Reduced leg kinetic energy at toe-off in the paretic limb was consistent with inadequate leg propulsion by the plantarflexors or hip flexors during swing initiation. Reduced peak knee flexion during swing and increased percentage swing time in the paretic limb could be attributed to reduced leg kinetic energy at toe-off. Increased energy cost associated with raising the trunk during pre-swing and swing of the paretic limb was consistent with pelvic hiking to clear the paretic limb with reduced knee flexion. Increased leg kinetic energy at toe-off and reduced swing time in the non-paretic limb was consistent with compensation for weakness or poor balance during single limb support on the paretic limb.

In Study 2, the adjustment of training parameters during treadmill walking was found to improve specific gait deviations associated with post-stroke hemiparesis (see Figure 2).

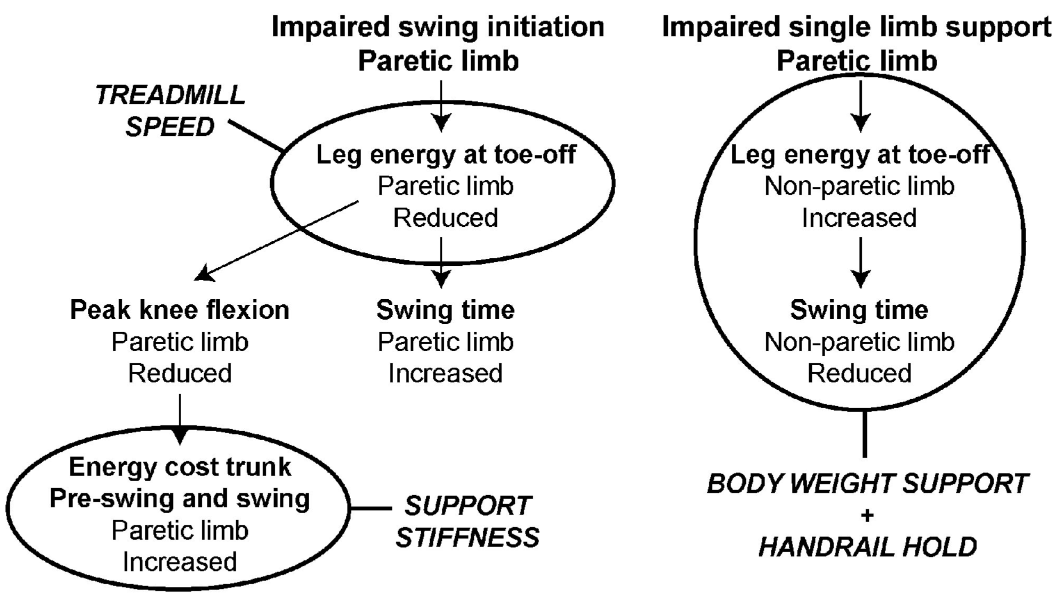

Figure 2. Summary of gait deviations improved with adjustment of training parameters.

With increased body weight support or the addition of handrail hold, the increased leg kinetic energy at toe-off and reduced swing time in the non-paretic limb were improved, resulting in increased single limb support time on the paretic limb. With increased treadmill speed, the reduced leg kinetic energy at toe-off in the paretic limb was improved. With increased support stiffness, the increased energy cost associated with raising the trunk was improved. However, with the adjustment of each training parameter, deviations in peak knee flexion during swing and swing time in the paretic limb were not improved. Our results provide a rationale for the use of manual assistance to improve swing kinematics in the paretic limb. However, if the increased leg energy elicited by faster training in the paretic limb can be maintained at slower walking speeds, we believe that it may also lead to improved swing kinematics in the paretic limb. The practice of an improved gait pattern during treadmill training, as defined by a reduction in gait deviations associated with hemiparesis, may improve treatment outcome in hemiparetic individuals.

Our findings provide a biomechanical basis for the selection of parameters for treadmill training in individuals with post-stroke hemiparesis. The selection criterion, based on the improvement of temporal, kinematic, and energetic gait deviations during training, is consistent with the use of task facilitation during skill acquisition (Holding, 1965; Newell, 1981) and the use of perceived improvement of gait movement pattern to guide the adjustment of training parameters. The practice and reinforcement of an improved gait pattern may optimize the sensory inputs facilitating spinal and supraspinal locomotor networks, which is believed to be important to maximizing the effectiveness of treadmill training (Dobkin, 1999). Moreover, walking with a more normal gait pattern, such as with increased single limb support time, can increase the functional demands on the paretic limb, which may also contribute to positive treatment outcome (Harris-Love *et al.*, 2001; Hesse *et al.*, 1999).

The Future

Further work is needed to extend the research and examine issues that were not addressed in this dissertation. First, the interactive effects from simultaneous adjustment of training parameters (e.g., the effect of treadmill speed at each level of body weight support) may be significant and should be examined in future studies. Second, other training parameters not examined in our study could be useful during treadmill training. For instance, manual assistance is typically provided during training and can have a strong impact on the gait pattern exhibited by hemiparetic individuals. Elastic cords are currently used in the UCLA locomotion laboratory to provide additional lateral stability to the trunk during treadmill training (Daniel P. Ferris, personal correspondence). Moreover, the relative distribution of harness support to the upper trunk and pelvis may be important. Future studies may examine how these additional training parameters affect the gait pattern exhibited by hemiparetic individuals during treadmill walking. Lastly, the effect of training parameters may be dependent on ambulatory status or other characteristics of the subject, and these factors should be considered in future studies.

Our scientific basis for selection of training parameters provides hypotheses regarding training application that can be tested in future clinical trials. For instance, our findings suggest that treadmill training with body weight support and handrail hold at a treadmill speed faster than the subject's comfortable speed with a stiff harness support may lead to better treatment outcome than training with other selections of training parameters (see Figure 3).

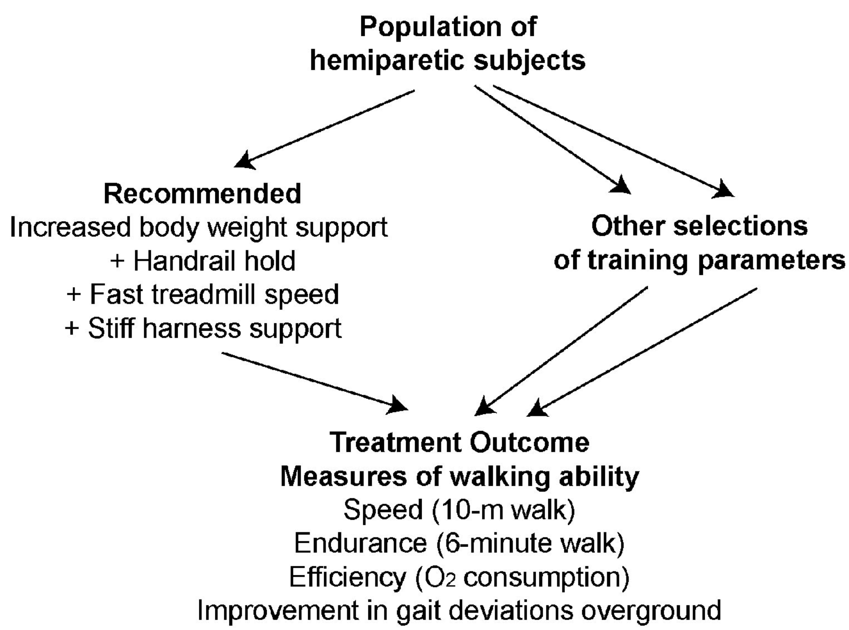

Figure 3. A model of a clinical trial that compares the effectiveness of treadmill training with different selections of training parameters.

A population of hemiparetic subjects can be randomized to different treadmill training groups, one of which receives training with the recommended selection of parameters.

Treatment outcome in the different groups can be assessed using measures of walking ability, such as speed, endurance, efficiency, and the improvement in gait deviations during overground walking. Speed, for example, can be measured by a timed 10-m walk, endurance by distance walked in 6 minutes (Guyatt *et al.*, 1985), and efficiency by oxygen consumption during submaximal treadmill walking (Macko *et al.*, 2001).

Following treadmill training in the hemiparetic subjects, the relative improvement between functional (i.e., walking faster with better endurance and efficiency) and quality (i.e., walking with a more normal gait pattern) measures of walking ability could suggest hypotheses regarding factors governing locomotor recovery. For instance, if faster overground walking speeds are achieved with a concurrent improvement in gait deviations associated with hemiparesis, it could be hypothesized that the improvement of gait deviations enabled the achievement of faster walking speeds. However, if faster walking speeds are achieved without improvement in gait deviations, it would suggest that the practice of a more normal gait pattern on the treadmill improved motor output (e.g., by promoting greater motor recovery, cardiovascular fitness, or strength), but that improvement in gait deviations during overground walking is not important to increasing walking speed in hemiparetic individuals. Lastly, if gait deviations during overground walking are improved following training, but walking speed, endurance, and efficiency are not improved, our rationale for the selection of training parameters could be challenged. However, the improvement in gait deviations during overground walking could lead to greater long-term improvement in functional measures of walking ability. This possibility would need to be confirmed in follow-up examinations. A better understanding of the factors governing locomotor recovery following treadmill training can lead to further refinement of training technique and the advancement of new approaches to gait rehabilitation for individuals with post-stroke hemiparesis.

APPENDIX A

Motion capture setup for treadmill walking with harness support

Reflective marker setup

Our reflective marker setup was based on standards described in MARey: Movement Analysis Routines for joint kinematics and kinetics (Center for Locomotion Studies, Penn State University, State College, Pennsylvania). Eight clusters of three reflective markers were located on the subject's upper trunk and pelvis and right and left thighs, shanks, and feet (see Figure 1). Markers of the upper trunk were affixed to skin over the sternum and clavicle region using double-sided tape. Markers of the pelvic, thigh, and shank clusters were glued to molded thermoplastic pieces, which were held in place by double-sided tape and elastic Velcro straps. The pelvic cluster was positioned to bridge across the bony prominences of the right and left anterior superior iliac spines. The thigh and shank clusters of the right and left limbs were staggered high and low (see Figure 1) to minimize marker blending in the camera views as the legs pass each other during the gait cycle. Markers of the foot clusters were glued to the shoes over the metatarsal region of the foot. The position of each marker cluster was calibrated to anatomical reference points during standing calibration trials.

Camera and treadmill configuration

The three-dimensional coordinates of the reflective markers during treadmill walking were reconstructed from data captured by four ProReflex cameras (Qualisys Inc., East Windsor, CT), positioned approximately as shown in Figure 2. A fifth camera (not shown in Figure 2), positioned opposite to the other four cameras, was used during calibration but not during walking trials. The treadmill belt was run in reverse as subjects walked towards the back of the treadmill (see Figures 1 and 2) to provide the cameras with a less-obstructed view. Moreover, the vertical supports of the custom-made harness support frame (Chen *et al.*, 2001) were positioned to the sides 2.45 m apart, so that they did not obstruct the camera views (see Figure 1).

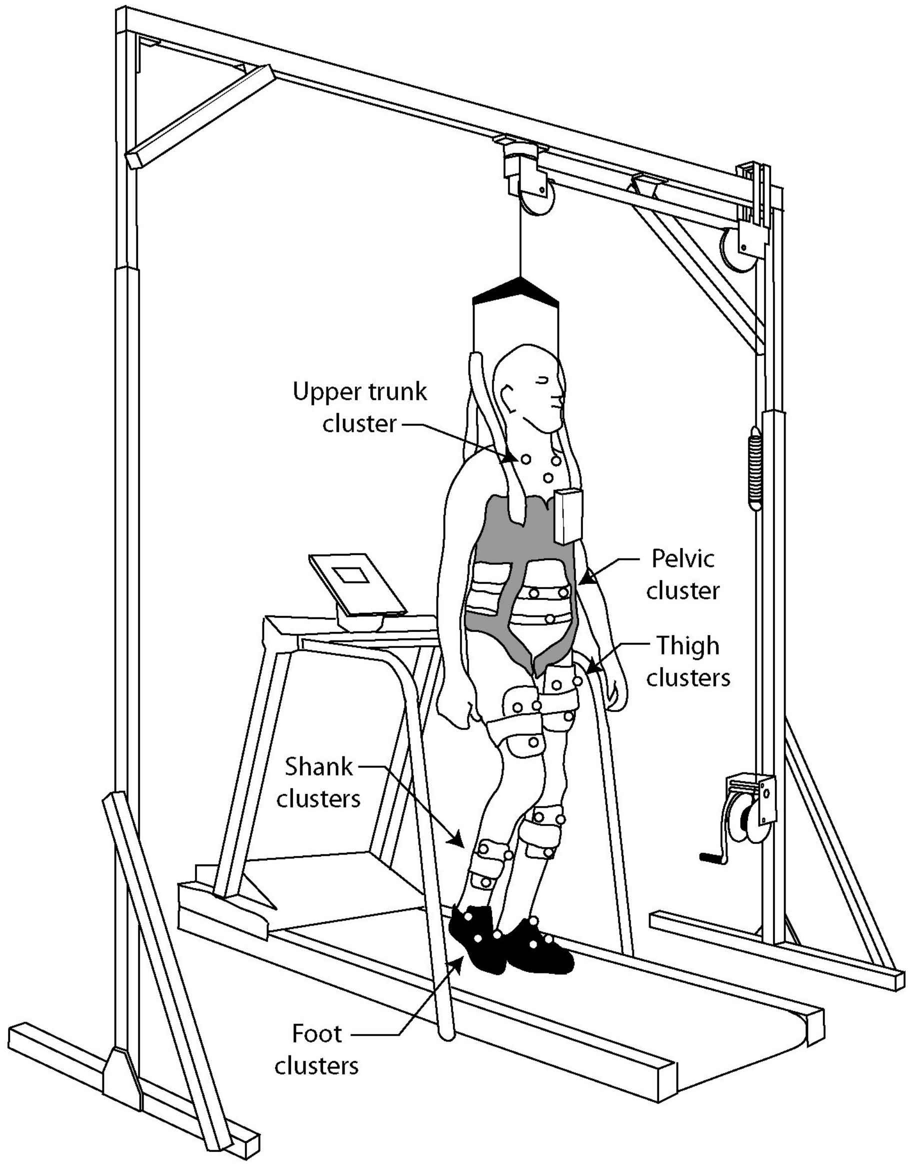

Figure 1. Configuration of treadmill and harness-support frame relative to the subject. Eight clusters of three reflective markers were located on the subject's upper trunk and pelvis and right and left thighs, shanks, and feet.

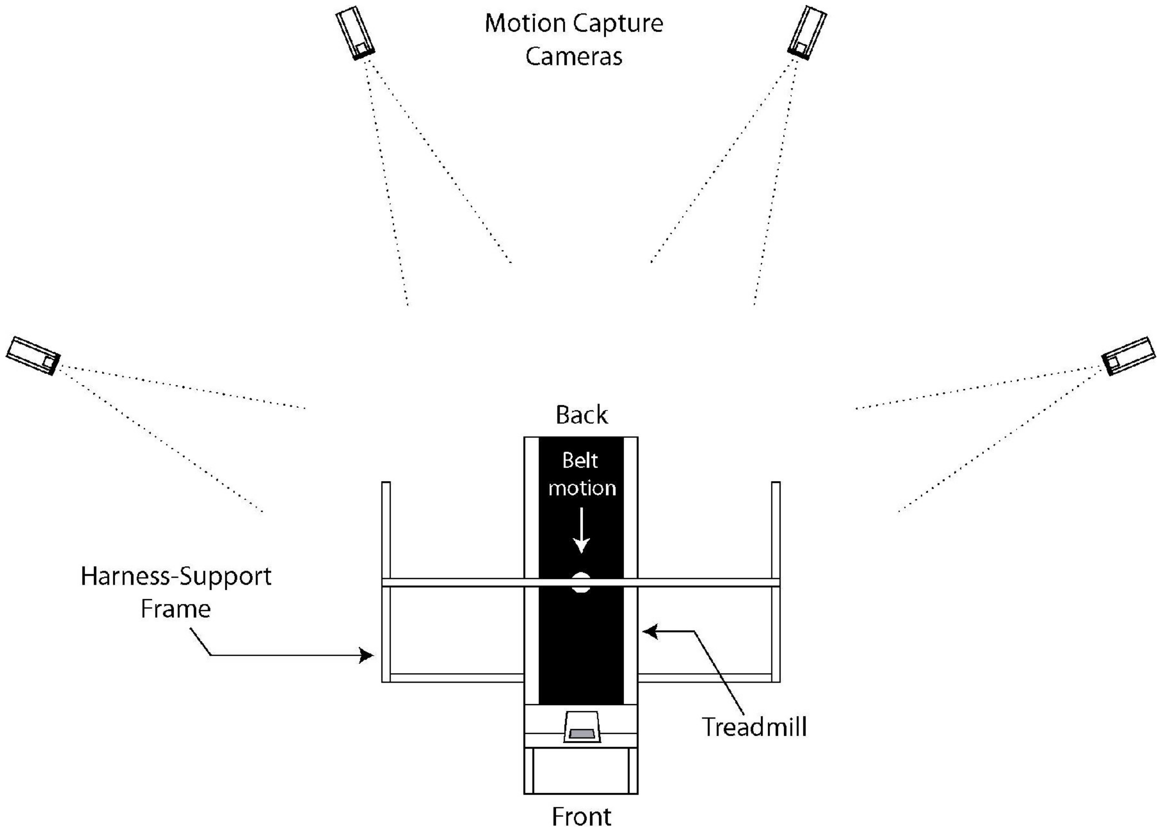

Figure 2. Configuration of the motion capture cameras relative to the treadmill and harness-support frame. Each camera was positioned approximately 2.0 m above ground level and 3.5 m from the center of the treadmill belt where the subjects walked.

APPENDIX B

Effect of harness support and training parameter settings on gait characteristics in non-disabled subjects

Study 2 examined the effect of harness support and training parameter settings on gait characteristics in hemiparetic subjects during treadmill walking. Data was also collected on non-disabled subjects as they walked on the treadmill during each of the 10 experimental conditions described in Study 2 (for description, see Table 2 in Study 2). For completeness and as theoretical comparison, the effect of harness support and training parameter settings on gait characteristics in non-disabled subjects are presented in Tables 1 through 5 in this appendix, focusing on the gait variables discussed in Study 2. Since data from the right and left limbs did not differ appreciably in the non-disabled subjects, they were averaged and analyzed together. Therefore, the group means in the tables represent values for the right and left limbs.

Harness-supported vs. free treadmill walking

With harness support, leg kinetic energy at toe-off was reduced (p=0.03), resulting in increased percentage swing time (p=0.03). A tendency towards reduced peak knee flexion during swing and energy cost associated with raising the trunk was observed, but the differences were small and did not reach statistical significance.

Table 1: Harness-supported vs. free treadmill walking

	Free Walking	Harness Support	p
Swing time (% gait cycle)	31.8 (9.2)	37.8 (8.7)	0.03
Leg kinetic energy at toe-off (cJ/kg)	4.7 (2.3)	3.5 (1.9)	0.03
Peak knee flexion during swing (deg)	60.3 (5.8)	57.6 (9.3)	
Trunk PEG cost – pre-swing/swing (cJ/kg)	6.4 (2.3)	5.6 (1.9)	

Group means and standard deviations (in parentheses)
Each condition was tested at the default treadmill speed (100% CTS) without handrail hold. Harness support was provided at the default level of body weight support (35%) and support stiffness (35.1 N/cm).
p = significance between free and harness-supported walking conditions (Wilcoxon signed-rank test).
Abbreviations: PEG, potential energy of gravity; CTS, comfortable treadmill speed of matched hemiparetic subject.

Influence of body weight support

The effect of body weight support on each of the gait variables did not reach statistical significance. However, a tendency towards increased swing time (multivariate p = 0.07) and reduced leg kinetic energy at toe-off was observed with increased body weight support. Peak knee flexion during swing and energy cost associated with raising the trunk were not affected by body weight support.

Table 2: Influence of body weight support

	20% BW	35% BW	50% BW	mult-p
Swing time (% gait cycle)	34.6 (6.2)	37.8 (8.7)	39.4 (11.2)	
Leg kinetic energy at toe-off (cJ/kg)	3.6 (1.9)	3.5 (1.9)	3.0 (1.3)	
Peak knee flexion during swing (deg)	57.2 (8.8)	57.6 (9.3)	57.5 (6.5)	
Trunk PEG cost - pre-swing/swing (cJ/kg)	5.7 (1.8)	5.6 (1.9)	5.8 (1.9)	

Group means and standard deviations (in parentheses)
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).
Abbreviations: BW, body weight; CTS, comfortable treadmill speed of matched hemiparetic subject; PEG, potential energy of gravity.

Influence of treadmill speed

With increased treadmill speed, leg kinetic energy at toe-off increased (multivariate p = 0.006). A tendency towards increased swing time, peak knee flexion during swing, and energy cost associated with raising the trunk was observed with increased treadmill speed, but these effects did not reach statistical significance.

Table 3: Influence of treadmill speed

	70% CTS	100% CTS	130% CTS	mult-p
Swing time (% gait cycle)	37.0 (12.4)	37.8 (8.7)	38.9 (5.7)	
Leg kinetic energy at toe-off (cJ/kg)	1.8 (0.8)	3.5 (1.9)	4.5 (2.4)	0.006
Peak knee flexion during swing (deg)	54.2 (8.0)	57.6 (9.3)	58.7 (8.9)	
Trunk PEG cost - pre-swing/swing (cJ/kg)	4.9 (1.1)	5.6 (1.9)	6.5 (2.8)	

Group means and standard deviations (in parentheses)
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).
Abbreviations: CTS, comfortable treadmill speed of matched hemiparetic subject; PEG, potential energy of gravity.

Influence of support stiffness

With increased support stiffness, energy cost associated with raising the trunk was reduced (multivariate p = 0.006). Swing time, leg kinetic energy at toe-off, and peak knee flexion during swing were not appreciably affected by support stiffness.

Table 4: Influence of support stiffness

	11.7 N/cm	35.1 N/cm	Rigid support	mult-p
Swing time (% gait cycle)	38.2 (7.7)	37.8 (8.7)	37.3 (9.6)	
Leg kinetic energy at toe-off (cJ/kg)	3.2 (1.7)	3.5 (1.9)	3.3 (1.7)	
Peak knee flexion during swing (deg)	56.5 (8.4)	57.6 (9.3)	58.8 (8.5)	
Trunk PEG cost - pre-swing/swing (cJ/kg)	6.7 (2.1)	5.6 (1.9)	3.5 (0.9)	0.006

Group means and standard deviations (in parentheses)
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).
Abbreviations: PEG, potential energy of gravity; CTS, comfortable treadmill speed of matched hemiparetic subject.

Influence of handrail hold

The effect of handrail hold on each of the gait variables did not reach statistical significance. However, a tendency towards increased swing time, and reduced leg kinetic energy at toe-off (multivariate p = 0.07) and energy cost associated with raising the trunk (multivariate p = 0.07) was observed across the conditions of handrail hold, harness support, and combined harness support and handrail hold. There was a tendency towards a small reduction in knee flexion in the two conditions that included harness support.

Table 5: Influence of handrail hold

	Handrail hold	Harness support	Harness and Handrail	mult-p
Swing time (% gait cycle)	35.2 (6.6)	37.8 (8.7)	39.4 (6.1)	
Leg kinetic energy at toe-off (cJ/kg)	3.8 (2.0)	3.5 (1.9)	2.9 (1.5)	
Peak knee flexion during swing (deg)	60.6 (4.2)	57.6 (9.3)	57.6 (6.6)	
Trunk PEG cost - pre-swing/swing (cJ/kg)	6.8 (2.2)	5.6 (1.9)	4.6 (1.9)	

Group means and standard deviations (in parentheses)
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 conditions (Friedman's method for randomized blocks).
Abbreviations: PEG, potential energy of gravity; CTS, comfortable treadmill speed of matched hemiparetic subject.

Discussion

The strongest effects of harness support and training parameter settings on gait characteristics in the non-disabled subjects were consistent with effects observed in the hemiparetic subjects, in particular for the gait variables corresponding to the non-paretic limb. For instance, with the addition of harness support, leg kinetic energy at toe-off was reduced and swing time was increased in both limbs in the non-disabled subjects, but these effects were observed only in the non-paretic limb in the hemiparetic subjects. However, with increased treadmill speed, leg kinetic energy at toe-off was similarly increased for both limbs in the non-disabled and hemiparetic subjects, though these energies remained asymmetrical in the hemiparetic subjects.

We believe that alterations in the sensory-mechanical environment with different selections of training parameters elicited changes in gait characteristics during treadmill walking, some of which were similar in the non-disabled and hemiparetic subjects. In the hemiparetic subjects, the elicited changes brought certain gait variables closer to values observed in the non-disabled subjects during free treadmill walking at matched speeds (i.e., without harness support and handrail hold), which reduced certain gait deviations associated with post-stroke hemiparesis. In the non-disabled subjects, the elicited changes caused gait variables to deviate from values observed during free treadmill walking. The reduction in leg kinetic energy at toe-off and increase in swing time with harness support are consistent with alterations in locomotor pattern during simulated reduced gravity (Donelan and Kram, 1997). Similar changes occurred with the addition of handrail hold. Since the forces applied to the handrail by the non-disabled subjects were probably much less than body weight, differences in sensory experience, like increased sense of stability during single limb support, probably contributed to these changes.

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