THE EFFECT OF UNLOADING ON OVERGROUND AMBULATION IN STROKE CLIENTS

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ABSTRACT

This study aimed to determine whether the use of body weight (BWS) during overground ambulation leads to immediate improvements in the gait pattern of stroke clients, and whether an overground support system is feasible for use as a gait training tool in this population. Fourteen subjects 3 to 18 weeks post stroke and with ambulatory capacity but mobility problems (timed-up-and-go score > 25 s) participated in the study. Subjects were instructed to walk over a distance of 7 m with 0% (full weight bearing), 15% or 30% BWS provided by an overhead suspension system. Ground reaction forces, body kinematics and temporal distance factors were recorded. Subjects demonstrated significant improvements in force generation during both the loading and push off phase of gait ($p < 0.001$) with increasing support. There were also significant improvements in trunk postural control as well as a decrease in cycle duration and total double support time. The results indicate that providing partial BWS during overground walking allows for the expression of an improved walking pattern in stroke clients. The system was well tolerated by all subjects during testing and indicates therefore that this system is feasible for use in retraining walking in stroke clients.

RÉSUMÉ

Cette étude visait à déterminer (1) si le support partiel du poids du corps durant la marche au sol entraîne une amélioration immédiate du patron de marche chez des personnes ayant eu un accident vasculaire-cérébral (AVC) et (2) si un système de support de poids peut constituer un outil pour réentraîner ces personnes à la marche au sol. Quatorze sujets ayant eu un AVC depuis 3 à 18 semaines et possédant des capacités ambulatoires avec mobilité réduite (pointage supérieur à 25 s au test "timed-up-and-go") ont participé à cette étude. La tâche expérimentale consistait à marcher au sol sur une distance de 7 m en ayant 0%, 15% ou 30% du poids de leur corps supporté à l'aide d'un système de harnais suspendu. Au cours des essais de marche, les forces de réaction du sol, les mouvements du corps ainsi que les variables spatio-temporelles étaient enregistrées. Les résultats ont démontré que les forces générées par les sujets durant les phases d'amortissement et de poussée de la marche augmentaient lorsque le pourcentage de support de poids était augmenté ($p < 0.001$). Il y avait aussi une amélioration significative du contrôle postural du tronc et une diminution de la durée du cycle de marche et de la période de double support. Les résultats indiquent que le support d'une partie du poids du corps durant la marche au sol entraîne une amélioration significative du patron de marche chez des personnes ayant eu un AVC. De plus, le fait que les sujets aient bien toléré le système de support de poids durant les évaluations indique que ce système pourrait être utilisé pour réentraîner ces personnes à la marche au sol.

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An easy life does not teach us anything. In the end it is the learning that matters: what we've learned and how we've grown.

Richard Bach (One)

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PREFACE

Candidates have the option of including, as part of the thesis, the text of paper(s) submitted for publication, or the clearly duplicated text of published paper(s). These texts must be bound as an integral part of the thesis. If this option is chosen, **connecting texts that provide logical bridges between the different papers are mandatory**. The thesis must be written in such a way that it is more than a mere collection of manuscripts; in other words results of a series of papers must be integrated.

The thesis must still conform to all other requirements of the "Guidelines for Thesis Preparation". **The thesis must include**: A table of contents, an abstract in English and French, an introduction which clearly states the rationale and objectives of the study, a comprehensive review of the literature, a final conclusion and summary and a thorough bibliography or reference list. Additional material must be provided where appropriate (eg. in appendices) and in sufficient detail to allow clear and precise judgement to be made of the importance and originality of the research reported in the thesis.

This is a manuscript based thesis. ***The methodology, results and a detailed discussion*** is presented in the format of a paper to be submitted to the journal BRAIN. A more concise summary and conclusion is presented in the final chapter of the thesis. The research and manuscript were done by the candidate with corrections and revisions provided by the two co-authors on the paper.

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CHAPTER 1

INTRODUCTION AND OBJECTIVES

INTRODUCTION

The asymmetrical gait pattern and decreased weight bearing capacity in stroke clients has been a significant challenge in gait rehabilitation of these individuals. Studies have been conducted which looked at the effect of supporting a percentage of the body weight by a harness during treadmill training. Generally immediate improvements in posture, kinematics and temporal distance parameters have been noted and following training subjects demonstrated improvements in gait speed, weight bearing and posture. In addition subjects were shown to maintain improvements in gait speed and motor recovery at three months post training (Visintin et al. 1998). It is not known to what extent these results are applicable to overground locomotion since it is not clear whether treadmill walking is the same as overground walking in stroke clients. Normal control subjects demonstrate minor differences in kinematics, electromyography, and temporal distance parameters between the two modes of walking. Following a stroke the motor control system is impaired and these relatively minor differences may in fact be highly significant in this population. In addition, it is not certain whether the use of support promotes a more symmetrical pattern of weight bearing since none of the studies conducted to date looked at the ground reaction forces.

This study looked at the effect of unloading on the gait pattern of stroke clients during overground ambulation. The study was conducted at the Posture and Gait Lab at the Jewish

Rehabilitation Hospital, and consisted of 14 stroke clients admitted for rehabilitation following an acute stroke. Each subject was required to complete five to ten walking trials along a seven metre walkway with partial weight support being provided by an overhead suspension system. Trials were done at 0% (full weight bearing), 15% and 30% BWS. For all trials bilateral recordings were done for ground reaction forces, body kinematics and temporal distance factors. The ground reaction forces were recorded at 100Hz using two AMTI (Advanced Medical Technology Incorporated) force plates, Kinematic data were recorded at 50Hz using the ELITE (BTS) four camera system, and temporal distance factors were acquired using a pair of pressure sensitive footswitches worn inside the shoes. Data were analysed using software from ELITE (BTS), and Matlab (Mathworks) and Systat (SPSS) mathematical softwares. A one way analysis of variance was done and in cases where statistical significance was obtained ($p < 0.001$) further analysis for post hoc comparisons were conducted using the Tukey test.

OBJECTIVES OF THE STUDY

Main Objective:

To investigate the immediate effects of unloading on the locomotor pattern of stroke clients ambulating overground.

Specific Objectives:

1. To compare the ground reaction forces when ambulating overground with 0%, 15% and 30% body weight support (BWS).
2. To compare the sagittal plane angular displacement profiles of the trunk, hip, knee, and ankle when ambulating overground with 0%, 15% and 30% BWS.
3. To compare the frontal and horizontal plane angular displacement profiles of the pectoral and pelvis segment when ambulating overground with 0%, 15% and 30% BWS.
4. To compare temporal distance parameters when ambulating overground with 0%, 15% and 30% BWS.
5. To determine the feasibility of using an overground system for gait training in stroke clients.

CHAPTER 2

BACKGROUND REVIEW

STROKE EPIDEMIOLOGY

Incidence of Stroke

Stroke, cerebrovascular accident, apoplexy- these terms all refer to a sudden onset of focal neurological deficit caused by a major disturbance in the blood supply to the brain. Stroke has been identified as the third leading cause of death, and the single greatest neurological cause of disability (Clifford, 1986; Dombovy et al., 1986; Duncan, 1994). Mayo et al. (1996) reported a total of 335,283 discharges of stroke clients over a ten year period (1982-1991) within Canada. In the United States approximately 500,000 individuals suffer a new or recurrent stroke each year (Dombovy et al., 1986; Knutsson et al., 1979; Feigenson, 1981) and in Great Britain, stroke accounts for nearly 5% of the National Health Service expenditure (Wade et al., 1987). The incidence of stroke has been decreasing since 1950 due to improved neurodiagnostic techniques as well as more effective management of hypertension. However, better medical management has led to an increase in post stroke survivors and as a result the overall prevalence of stroke survivors remains at 794/100,000 (Clifford, 1986). Of the stroke survivors some 50-60% will be disabled and 70% will have a reduced capacity for work (Clifford, 1986; Duncan, 1994; Feigenson, 1981; Wade, 1987).

Ambulatory status following a stroke

Gait is a function that is basic to most individuals, and may be defined as the manner of moving the body from one place to another, by alternately and respectively changing the location of the feet, with the condition that at least one foot is in contact with the walking surface at all times (Smidt, 1990). The functional goals of this activity are to move from one place to another and to do so safely and efficiently. However, in individuals suffering neurological insult, these goals are frequently compromised. Indeed, gait dysfunction is one of the significant contributors to disability after stroke.

The degree of recovery of ambulatory status post stroke is uncertain, however studies have reported ranges from 60-85% returning to independent ambulation at six months post stroke (Bonita et al., 1988; Duncan, 1994; Feigenson, 1981; Gowland, 1982; Gresham et al., 1979; Jorgensen et al., 1995; Keenan et al., 1984; Mayo et al., 1991; Newman, 1972; Olney et al., 1996; Olsen, 1990; Smith et al., 1985; Wade et al., 1985, 1987). Independent ambulation however is not necessarily the equivalent of functional ambulation, as indicated by Lerner-Frankiel et al. (1986), where it was found that actual distances required for independence as a community ambulator were far greater than those typically used by physical therapists to deem a patient independent. In fact many 'independent' clients walk slowly and rarely venture out of doors (Wade, 1987). Independent community ambulation requires walking at or near normal velocity and stepping up and down curbs (Lerner-Frankiel et al, 1986). Keenan et al. (1984) reported a 59% return to community level of ambulation. Wade et al. (1987) found a 22% return to normal walking speed at three months post stroke and a review by Duncan (1994) cited that at six months post stroke only 25% of individuals had regained walking speeds that were within normal ranges.

HEMIPLEGIC GAIT

The performance of gait is directed to the accomplishment of four related tasks: maintaining balance of the trunk, arms and head; maintaining support of the limb segments during stance; clearing the floor with the swinging foot during swing; and supplying sufficient energy to the body system with each stride to cause it to move forward (Olney et al., 1996). Normal gait is characterized by a smooth forward progression of the centre of gravity, well-coordinated limb movement and should be accomplished using energy conservation measures.

The two immediate impairments to gait performance following a stroke are diminished strength and inappropriately timed or inappropriately graded muscle activity (Olney et al., 1996). Later on there is the added impairment of spasticity and changes in the mechanical properties of the muscle (Dietz et al., 1984). Stroke clients have great difficulty with weight bearing through the affected limb and as a result demonstrate poor single limb balance as well as difficulty controlling forward progression. Perry (1969) noted that these subjects lacked: adequate shock absorption at heel strike; control of momentum during stance; the ability to generate the force for push off to maintain forward propulsion; and quick adequate excursion of the paretic limb during swing. In general, their gait is qualitatively described as slow, precarious and asymmetrical, with a short quick step of the unaffected leg and a long slow step of the hemiplegic limb.

Kinematic Profile

The kinematic profiles of hemiplegics are quite variable among subjects, however, within subjects a distinct pattern emerges over steps. This pattern is frequently described as one of hip flexion, knee extension, ankle plantar flexion and lower limb circumduction during swing; hip flexion, decreased knee flexion, and ankle plantar flexion during loading; knee hyperextension during midstance; and lack of roll-off at toe off (Perry, 1969). Burdett et al. (1988) in their analysis of 19 subjects walking with and without orthoses reported similar findings: decreased hip flexion at initial contact, increased hip flexion at toe off, and decreased hip flexion during mid swing; more knee flexion at initial contact and less at toe off and mid swing; and more ankle plantarflexion at initial contact and mid swing and less at toe off.

There have been conflicting reports with regards to the decrease in hip flexion and hip extension. Unlike Burdett and colleagues, Olney et al. (1991) found that there was no decrease in hip flexion with initial contact. Leheman et al. (1987) found a decrease of 14 degrees in hip extension when compared to normal individuals walking at comparable speeds, whereas the decrease found by Olney et al. (1991) was significantly less. The variability in these findings may be due to differences in walking speeds in all three groups.

Wagenaar and Beek (1992) reported on trunk rotation in relation to gait speed. In the nine subjects included in the study it was found that the total range of thoracic rotation was greater in stroke clients walking at higher velocities (0.5 to $1.25ms^{-1}$) than that of healthy subjects. No significant differences were found however for total range of pelvic rotation, total range of trunk rotation and the phase difference of head rotation. Studies looking at the energy of walking in stroke clients have also found a dominance of head, arms and trunk movements in the total gait pattern, which in turn leads to an increase energy cost of walking (Olney et al., 1986, 1988, 1991).

Spatial and Temporal Characteristics

Walking velocity is one of the most consistent differences reported between stroke clients and unimpaired individuals. The average range of walking speed reported for hemiplegic individuals is 0.2 to 0.7 ms^{-1} compared with 1 to 1.2 ms^{-1} for normal subjects (Olney and Richards, 1996; Giuliani, 1990; Finch and Barbeau, 1986; Peat et al., 1976; and Olney et al., 1974). In addition to the decrease in walking speed, stride length and cadence are also lower in these individuals. Brandstater et al. (1983) reported an average stride length of 0.6m compared to values of 1 to 1.4 for normal adults (Oberg et al., 1993).

Several researchers have reported abnormal stance/swing ratios in hemiplegic individuals (Olney and Richards, 1996; Knutsson, 1981; Peat et al., 1976; Olney et al., 1974; Brandstater et al., 1983; Mizrahi et al., 1982; Murray, 1967 and Wall, 1986). Three main differences were reported; first the stance phase of both the affected and unaffected sides were longer in duration and occupied a greater proportion of the gait cycle than normal, and secondly the stance phase of the unaffected limb was longer than that of the affected. The affected limb in 20 subjects was shown to have a stance phase of 67% of the cycle and swing of 33%, whilst the unaffected limb showed a stance phase of 80% and swing of 20% (Finch and Barbeau, 1986). Normal subjects spend 60% of the gait cycle in stance and 40% in swing (Murray, 1967; Perry, 1992; and Dubo et al., 1976). The third difference was an increase in the time spent in double support (Olney and Richards, 1996).

Ground Reaction Forces

Very few studies have looked at the ground reaction forces produced during hemiplegic gait. Wortis et al. (1951) reported greater variability and an initial low peak in the vertical force curve when compared to the double peaked curve obtained for normal subjects. Carlsoo et al. (1974) reported three patterns based on the vertical force curves of the paretic leg: one with two distinct crests and an intermediate trough, with the crests occurring at heel strike and push off; the second showing a constant vertical force for the entire support phase of stance; and a third showing a single peak at mid stance. The transversal horizontal force was found to be directed laterally during the major part of the support phase of stance. This was the same as for normal subjects, however, the magnitude of the force was found to be different for the two legs. In some cases the horizontal force was smaller for the unaffected leg whilst in others it was larger. Similar variability in the vertical force curve was reported by Crenna and Frigo (1985) using the Vector Diagram technique. These abnormalities indicate that individuals with hemiplegia have difficulty controlling force parameters, especially during weight acceptance and problems with transferring weight to the affected limb.

Rogers et al. (1993) found that major differences occurred in the proportion of the resultant lateral horizontal force contributed from underneath the flexing limb versus the stance limb during a rapid leg flexion task in stroke clients. Abnormalities were found in both the affected as well as the unaffected leg, and the changes indicated an overall reduction in the actively generated horizontal force under the paretic limb. Findings from the study were indicative of an attempt to cautiously regulate the dynamics of weight transfer to the affected side. Smaller changes were also observed in the timing of the ground reaction forces, with changes under the paretic limb starting later than the non-paretic limb. Even though the leg flexion task did not include a forward progression component as occurs with gait, both tasks require the lateral transfer of the body mass from bipedal to single limb stance. Ground reaction force profiles similar to those obtained by Rogers et al. were reported by Brunt et al. (1995), in a study looking at limb loading and control parameters in gait initiation of stroke clients. Stroke clients were found to have difficulty with loading the affected leg prior to unloading for swing. This study concluded that there was a correlation between symmetrical weight bearing and the ability to provide the forces necessary to generate the forward momentum in the initiation of gait.

Morita et al. (1995) attempted to correlate certain force parameters in 58 stroke clients with the stage of motor recovery using the Brunnstrom scale. The impulse (area between a component curve and baseline) was used to quantitatively assess the vertical and horizontal forces. For the vertical force curve the ratio of the impulse of the affected limb to that of the unaffected was found to be highly correlated with the stage of recovery. Values approached 1.0 when subjects were at stage 6 recovery (spasticity disappears and co-ordination approaches normal) Brunnstrom (1970). In addition the impulse of the unaffected limb was found to be higher than that of the affected, and the values for both limbs were higher at increasing stages of recovery.

For the longitudinal force curve, the ratio of the impulse of the acceleration force, divided by the absolute sum of the acceleration and deceleration forces, was also found to be highly correlated with the stage of recovery. The ratio for both limbs approached normal values of 0.5 with increasing stage of recovery. The acceleration force of the unaffected limb was found to be greater than the deceleration force whereas the reverse occurred for the affected limb. This may be related to synergistic muscle activity as well decreased hip extension and ankle dorsiflexion of the affected limb. The average value of the lateral force of the affected limb divided by that of the unaffected was also found to approach normal values with increasing stage of recovery however the correlation was much less than that observed for the vertical and longitudinal forces. Generally the ground reaction forces seems to provide a good reflection of the degree of motor recovery following a stroke.

Weight Bearing in Hemiplegics

Some of the gait asymmetries seen in hemiplegics is often attributed to a difficulty with weight bearing on the affected limb, as well as problems with weight shifting. Studies looking at weight bearing in stroke clients and foot- ground pressure patterns have indeed confirmed this (Bohannon et al., 1985; Dickstein et al., 1984; Seliktar et al., 1978). Twenty of the 23 subjects examined by Dickstein et al. (1984) demonstrated decreased weight bearing on their affected lower extremity. Bohannon and colleague (1985) also found that their subjects demonstrated decreased weight bearing on the affected lower extremity. The ratio of weight over the non paretic lower extremity to weight over the paretic extremity was found to be 1.46 which exceeds the upper limit documented by Dickstein et al. (1984) for a normal geriatric control group (Ratio 1.09, CI= 0.83-1.35).

All subjects were able to increase weight bearing on the affected extremity when shifting their weight as much as possible, however, they were unable to reach the values of the unaffected lower extremity. It was concluded that stroke clients cannot overcome fully their difficulty in bearing weight through the affected lower extremity by voluntarily shifting as much weight as possible onto the extremity. Traditional neurodevelopmental approaches to gait training, which focuses on weight shifting exercises to increase weight bearing on the affected limb, may therefore be limited in their effectiveness.

EFFECTS OF UNLOADING ON TREADMILL AMBULATION

Supported locomotion has become a topic of interest within the last ten years and has its foundations in animal studies. Barbeau and Rossignol (1987) demonstrated that adult spinalized cats were capable of regaining full weight bearing locomotion through an intensive interactive locomotor training program. The program consisted of appropriately graded weight support that was provided by supporting the cat's tail and allowing the animal to bear only the amount of weight which it was capable of. Training was done on a treadmill. Based on these findings as well as clinical studies on gait it was felt that this approach of supporting a percentage of the body weight may lead to more normal gait patterns in individuals with gait impairments. A treadmill apparatus with harness support for evaluation and rehabilitation of gait was proposed (Barbeau et al., 1987; Norman et al., 1995) and has been applied to two neurological patient populations, individuals with spinal cord injuries (Visintin et al., 1989;

Barbeau et al., 1992; Wernig et al., 1992) and stroke clients (Visintin et al., 1995; Hesse et al., 1995,1997; Hassid et al., 1997). The effect of unloading on the gait pattern of spinal cord injured subjects (SCI) and hemiplegics will be considered here.

Unloading in Spinal Cord Injuries

Visintin and Barbeau (1989) looked at the immediate effects of unloading on the locomotor pattern of seven spastic paretic clients. Despite the high degree of variability noted between subjects there was a decrease in mean burst amplitude with some muscles showing more appropriate EMG timing. Subjects demonstrated a straighter trunk alignment with 40% body weight support as well as improved knee extension during initial contact. Subjects who had excessive knee flexion during midstance demonstrate better extension with support. With regards to temporal distance parameters there was an increase in cycle duration from 5.6 to 20.6%. Single limb support time was increased within a range of 3.3 to 64.3% and the stance phase was found to approach normal in two of the subjects. Coinciding with the increase in single limb support was a decrease in the percentage of total double support time, which was more marked in clients with mild and moderate disability. Stride length was also seen to be increased from 13.7 to 42.9% and five clients demonstrated an increase in maximal walking speed ranging from 10.3 to 44.2%. In addition to these changes, greater cardio-pulmonary efficiency might result as reflected by the decreased heart rate during the supported trials (Ratliff et al., 1995).

Training effects over 6 to 28 weeks were reported by Wernig and colleague (1992). These included a decrease in support from 40% to 0%, increased walking distance from 0-104 metres at week one to 200-410 metres at the last week of training, and increased walking speed from 0-10m/min. to 14-23m/min.

Unloading in Hemiplegics

As mentioned previously, stroke clients have difficulty with weight bearing through the affected lower extremity, therefore the use of a body weight support system offers a solution to this problem during gait training. Hesse et al. (1997) looked at the immediate effects of varying degrees of body weight support on treadmill ambulation of 11 hemiparetic subjects. Individuals were shown to demonstrate a decrease in relative double support time from 29.1% of the cycle with full weight bearing to 12.0% with 60% body weight support. Relative single limb support time was found to increase from 45.5% with full weight bearing to 76% with 60% body weight support. Most consistent improvements in single limb stance time was reported by Hassid et al. (1997) to lie between 15 and 35% body weight support, the single limb loading ratio within this range being closest to one. By supporting a percentage of the body weight the individuals were able to spend more time on the impaired lower extremity and demonstrated phase changes in gait that were similar to those reported by Finch et al. (1991) for normal individuals.

With body weight support, individuals were found to walk with a more upright posture. Increased hip and knee extension were demonstrated during mid stance, however, there was also a reduction in hip and knee flexion during swing (Hesse et al., 1997). This pattern is a closer approximation to normal gait patterns. In some subjects the use of body weight support is shown to result in a change from forefoot or flat foot contact to heel contact, however, with 60% support, and in some cases 45% subjects were seen to walk on tip toe especially on the unaffected side (Hesse et al., 1997). These findings were similar to that of Visintin and Barbeau (1989) where it was seen that greater than 40% body weight support in spastic paretic clients resulted in loss of heel ground contact. From these studies it would appear that the optimal choice of support lies between 15 and 30% of the body weight.

The functional activity of vastus lateralis and soleus was found to decrease constantly with the most significant changes demonstrated for support of greater than 30% (Hesse et al., 1997). Similar findings were demonstrated in normal individuals (Finch et al., 1991). The activity of tibialis anterior, biceps femoris, and erector spinae was unaffected by body weight support. Vastus lateralis demonstrated decreased activity at all levels of support, whereas gluteus medius demonstrated increased activity with support of up to 45% of the body weight. Premature activity of the soleus did not change regardless of the degree of support that was provided. Qualitatively the pattern of muscle activation remained the same.

Long term training effects were investigated by Hesse et al. (1994,1995). It was found that within 15 days of training, support could be reduced from 30% to 0%, and speed was increased from 0.07-0.11m/s to 0.18-0.22m/s. Improvements were also noted on the Rivermead Motor Assessment Score. Significant improvements in gait parameters were reported by Visintin et al. (1995, 1998) for clients with severe and moderate disability. They conducted a randomized clinical trial involving 100 stroke clients. Following six weeks of training the BWS group scored significantly higher than the no BWS group on measures of functional balance, motor recovery, overground walking speed, and overground walking endurance. In addition the BWS group maintained higher scores on overground walking speed and motor recovery at 3 months post training.

Generally the literature suggests that body weight support results in improvements in the mechanics of gait in hemiplegic subjects however the degree of support provided has to be carefully chosen since values above 40% begin to produce changes that differ from normal physiological gait. Most recent studies seem to suggest that between 15 and 35% body weight support is optimal (Hassid et al., 1997; Hesse et al., 1997).

All of the studies discussed have looked at the effect of unloading on treadmill ambulation. No study has been identified that looks at the effect on overground walking. In non disabled subjects differences were found between the two modes of walking, however they were considered to be relatively insignificant and it has been concluded that treadmill walking is a good approximation to overground walking in normal individuals (Arsenault et al., 1986). This assumption cannot be applied to a neurologically compromised population whose motor control system is impaired. In spinal cord injured subjects it has been observed that some individuals who have difficult walking overground demonstrate improvements in kinematics and temporal distance parameters on the treadmill (Barbeau, personal communication). It is important therefore to examine the effects of unloading on overground ambulation in stroke clients and not assume that it will be the same as on the treadmill.

Most studies on hemiplegic gait focused on the affected extremity, however, as cited in a review by Olney and colleague (1996), the unaffected lower extremity also demonstrate abnormal patterns and in some cases even more so than the affected lower extremity. Bilateral recording in gait studies involving hemiplegic subjects therefore is very important in order to obtain a better understanding of the compensations that may occur during hemiplegic gait.

None of the studies looking at supported locomotion in hemiplegics have reported on the ground reaction forces, and only one study to date has reported on this outcome in a normal population. Flynn et al. (1997) found that with 20% body weight support there was reduction in both the first and second vertical force peaks. Plantar pressure was found to be reduced from 6.8% to 27.8%, however, the reduction varied at different regions of the foot.

TREADMILL WALKING COMPARED TO OVERGROUND WALKING

The application of treadmill training to the field of rehabilitation has evolved from research with cats; where it was shown that the spinalized animal could be trained to walk on a treadmill (Lovely et al., 1985, 1986; Barbeau et al., 1987). Even though treadmill training has been shown to be a good approximation to overground walking there were still some differences that were identified. Wetzel et al. (1975) reported variations in the subcomponents of swing as well as interlimb timing in cats walking on a treadmill. The interval between touchdown for one hindlimb and the ipsilateral forelimb was reduced for treadmill stepping. Similarly comparisons in non disabled individuals have revealed differences between treadmill and overground walking for joint as well as temporal distance measures (Arsenault et al., 1986; Murray et al., 1985; Nelson et al., 1972; Pearce et al., 1983; Strathy et al., 1983).

Changes in knee function during treadmill ambulation was reported by Strathy et al. (1983). Sagittal plane motion showed a lack of maximal knee joint extension just prior to or at heel strike on the treadmill. Average heel contact time on the treadmill was reduced and toe contact time increased. This may have been due to the decreased knee extension observed at heel strike. Increased cadence together with shortened stride lengths were reported by Murray et al. (1985), Arsenault et al. (1986) and Stolze et al. (1996) for treadmill walking. Stolze found that stance was enhanced by 5% whilst stride length was reduced by 4%when compared to overground walking. In addition step width and foot angles were found to be increased during treadmill locomotion, which may indicate the adaptation of a more protective gait pattern. Murray et al. (1985) reported significantly more hip extension with overground walking than with treadmill walking at three different speeds.

Two studies were identified which reported on electromyographic (EMG) activity during treadmill walking, Murray et al. (1985) and Arsenault et al. (1986). Murray et al. reported significantly greater EMG activity in the quadriceps during treadmill walking as well as greater average EMG activity for all muscle groups during treadmill walking at three different speeds. In some cases the calf muscles were seen to start its activity earlier in the stance phase during treadmill walking. Similar increases in quadriceps activity was reported by Arsenault et al. (1986), more specifically in the biceps femoris and rectus femoris. Pooling of data across subjects however showed that these differences were not significant.

Van Ingen Schenau (1980) indicated that from a purely mechanical point of view, there is no difference between treadmill walking or running and overground when the mechanical variables are described with respect to the surface on which the subject locomotes. He suggested that variations may be attributed to differences in visual and auditory information. In overground walking the surroundings move with respect to the subject, this could cause a difference in the regulation of the movement pattern thereby resulting in differences in kinematic patterns and energy consumption.

From a neurophysiological point of view treadmill walking requires tracking of an externally produced velocity under different peripheral inputs (Wetzel et al., 1975). There are no external cues apart from the treadmill belt to inform the animal that it is moving, therefore it has to learn to utilize a variety of peripheral inputs but in a different manner from overground walking. Visual and vestibular information, which are important in maintaining equilibrium and stability during overground locomotion, has to be used quite differently when walking on a treadmill. On the treadmill the external environment is stationary, whereas in the natural environment the external environment constantly changes as both the individual as well as objects around are in motion. Generally with treadmill walking, the speed of walking is being externally driven by the treadmill rather than internally driven by the individual, the subject is lifting up to keep up with the treadmill belt rather than pushing off to initiate the swing, and the stance limb is being pulled backward under than trunk instead of the trunk gliding forward over the stance limb

(Bassille & Bock, 1995).

There is no documented study showing that treadmill walking is similar to overground walking in stroke clients and in fact these insignificant differences in non disabled subjects may be highly significant in this population. Due to this the results of treadmill studies in stroke clients cannot be applied to overground walking. Even though the effect of unloading on treadmill walking has been examined in this population the effects on overground walking is not known. This study therefore will look at the effects of constant load removal on overground walking in stroke clients.

CHAPTER 3

RATIONALE

The asymmetrical gait pattern and decreased weight bearing capacity has been a challenge in gait rehabilitation of stroke clients. Several studies have been conducted to analyse the mechanisms involved in hemiplegic gait, however, most of these studies have focused on the affected lower extremity with minimal or no attention to the unaffected side. In order to understand the compensations or over compensations that may occur, it is important to do bilateral comparisons in these individuals. Some of the compensations may actually predispose the unaffected side of the body to injury. Therefore, it is important to understand these mechanisms so this can be prevented.

None of the studies that looked at the effects of unloading on hemiplegic gait examined the ground reaction forces. There appears to be a relationship between symmetrical weight bearing and the ability to provide the forces that generate the forward momentum necessary for effective gait initiation. It would be important therefore to see whether unloading produces a more normal pattern in the force curves, which would indicate greater efficiency in initiating gait, or whether it creates even more distortions in the curve, thereby indicating a less efficient pattern.

All of the studies that have looked at the effects of unloading on hemiplegic gait have been conducted on the treadmill. Furthermore, all of the studies comparing treadmill and overground ambulation were conducted in normal populations, and even though differences were found to be minor in normal subjects, they may be quite significant in a population whose motor control mechanisms are impaired. The motor control of treadmill walking is also different from overground walking. In stroke clients some or all of these mechanisms may be compromised and their behavior on a treadmill may be very different to that overground. From the differences pointed out in the literature review the results of the treadmill studies cannot be applied to overground walking in a stroke population.

This study therefore proposes to look at the effects of unloading on overground ambulation in stroke clients. Kinematics, ground reaction forces and temporal distance factors will be investigated. The results of this study would add to the body of literature regarding ambulation in stroke. It would provide a more complete picture of the mechanisms involved in postural control during hemiplegic gait and the changes that occur with unloading. In addition it would form a basis for further investigations with regards to treadmill and overground ambulation in a neurologically compromised population.

HYPOTHESES

It was expected that BWS during overground walking would lead to immediate and significant improvements of the gait pattern of stroke clients. More specifically the following were expected to be seen:

1. Changes in kinematic profiles that would indicate a more upright posture and thus improved postural control during supported walking.
2. Changes in ground reaction forces that would indicate improvements in force generation and propulsion, with force patterns approaching normal during supported walking.
3. Changes in temporal distance parameters that would reflect a more symmetrical gait pattern with support.

ETHICAL ISSUES

The only ethical issue that arose in this research was that of consent to participate in the study. This was addressed by getting the subjects to sign an informed consent form (appendix 1). There was no risk of falls involved since the subjects were supported by the harness, and none of the measurement procedures involved the infliction of pain or any form of trauma.

SUMMARY OF RESEARCH

EXPOSURE Walking with body weight support: 0, 15, & 30% . (Ten trials at each level)			
PARAMETERS	**DATA COLLECTION**	**OUTCOME**	**STATISTICS**
Bilateral Ground Reaction Forces in sagittal and frontal planes	Two AMTI force plates. Data collected at 100Hz	Impulse Index, and force difference during loading and push off phase.	One way ANOVA, Tukey post hoc test and simple linear regressions.
Kinematics (sagittal, frontal and horizontal plane): trunk, bilateral hip, knee, ankle	ELITE 4 camera system. Data collected at 50Hz	Sagittal, frontal and horizontal plane excursions hip, knee, ankle, trunk, pelvis and pectoral segment. Knee angle at loading	One way ANOVA, Tukey post hoc test and simple linear regressions.
Temporal distance factors	Pair of pressure sensitive footswitches. Output converted to an analogue signal and collected in ELITE at 100Hz	Step and stride lengths, cycle duration, single limb support time, double support time, % swing, % stance, stance swing ratio.	One way ANOVA, Tukey post hoc test and simple linear regressions.

CHAPTER 4

RESULTS AND DISCUSSION

THE EFFECTS OF UNLOADING ON OVERGROUND LOCOMOTION FOLLOWING STROKE

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SUMMARY

Unloading of the lower extremities by providing partial body weight support (BWS) during treadmill training has been shown to be an effective gait training approach in stroke clients. However, it is not known how unloading impacts on the gait parameters. Moreover, several differences exist between treadmill and overground ambulation, and these may be highly significant in a neurologically compromised population. This study aimed to determine whether the use of BWS during overground ambulation leads to immediate improvements in the gait pattern of stroke clients, and whether an overground support system is feasible for use as a gait training tool in this population. Fourteen subjects 3 to 18 weeks post stroke and with ambulatory capacity but mobility problems (timed-up-and-go score > 25 s) participated in the study. Subjects were instructed to walk over a distance of 7 m with 0% (full weight bearing), 15% or 30% BWS provided by an overhead suspension system. Ground reaction forces, body kinematics and temporal distance factors were recorded. Subjects demonstrated significant improvements in force generation during both the loading and push off phases of gait ($p < 0.001$) with increasing levels of support. There were also significant improvements in trunk control. Subjects demonstrated a more upright trunk, more level scapular rotations, and increased pelvis rotation with increasing support. In addition, there was a decrease in cycle duration and total double support time with BWS. The results indicate that providing BWS during overground walking leads to immediate and significant improvements in kinetics, kinematics and temporal distance factors in stroke clients. The testing protocol was well tolerated by the subjects, which indicates that the system is feasible for use in retraining ambulation in stroke clients with mild to moderate disability.

INTRODUCTION

Stroke is the third leading cause of death and the single greatest cause of neurological disability (Clifford, 1986; Dombovy et al., 1986; Duncan, 1994). Gait dysfunction continues to be one of the significant contributors to disability following a stroke, and of the survivors, 59% return to community ambulation (Keenan et al., 1984), with only 22% regaining normal walking speeds at 3 months post stroke (Duncan, 1994). Both animal and human research have shown that the type of training strategy can significantly influence the degree of locomotor recovery (Barbeau et al., 1998; Hesse et al., 1995; Barbeau et al., 1994; Richards et al., 1993; Finch, 1986). However, the most efficient and effective gait training strategy remains to be determined. It is known that the earlier training is started, the more favorable the outcome (Novack et al., 1984). In stroke clients, the actual practicing of gait is often delayed due to an inability to bear weight on the paretic limb. One of the most promising approach to date, which addresses the issue of weight bearing capacity, is gait training with the use of body weight support (BWS) and treadmill stimulation (Visintin et al., 1989, 1995, 1998). This approach was developed on the basis of animal findings, which showed that adult spinalized cats could regain full weight bearing locomotion through intensive interactive locomotor training consisting of graded weight support during treadmill walking (Barbeau & Rossignol, 1987).

Subjects with spinal cord injuries demonstrated immediate improvements in trunk alignment as well as improved knee extension with 40% BWS (Visintin et al., 1989). In addition there was a more appropriate timing of electromyographic (EMG) activity of the lower extremities. Significant changes were also observed in temporal distance parameters, which included increased single limb support duration, decreased total double support duration, increased stride length and increased maximal walking speed. Training over 6 weeks led to a decrease in total double support duration, increased endurance and a straighter knee at loading (Wernig et al., 1992, Barbeau et al., 1992, Barbeau et al., 1991). Similar improvements were noted in stroke clients walking with BWS on the treadmill. Hesse et al., 1997, 1995, 1994; Hassid, 1997). Relative double support time decreased from 29.1% at full weight bearing to 12% at 60% BWS, and the single limb loading ratio approached one when support was provided. Furthermore long term training on the treadmill with BWS was seen to improve walking speed, functional balance, motor recovery and overground walking endurance in stroke clients (Hesse et al., 1994, 1995; Visintin et al., 1998). Visintin et al. (1998) also reported maintenance of a higher overground walking speed and motor recovery score in clients trained with BWS three months post training as opposed to those that were not trained with BWS.

The parallel bars may also be used to provide weight support and facilitate walking however, it encourages uneven weight bearing, with most weight being placed on the non-paretic limb (Visintin et al., 1994). In addition support by the parallel bars leads to an inappropriate swing phase with respect to EMG activity of the tibialis anterior muscle, as well as decreased excursion of the hip, knee and ankle (Barbeau et al., 1991). An overhead suspension harness allows for symmetrical unloading of the limbs (Barbeau et al., 1987; Norman et al., 1995). A comprehensive review of 20 studies showed that a gain in walking speed can be achieved with task specific locomotor training, but not with conventional methods of rehabilitation focusing on tone and strength impairments (Barbeau et al., 1998).

Treadmill training is commonly used as a tool for gait rehabilitation, and is based on the assumption that treadmill walking is similar to overground walking (Arsenault et al., 1986; Murray et al., 1985; Nelson et al., 1972; Pearce et al., 1983; Strathy et al., 1983; Stolze et al., 1996). However, several differences were identified between the two modes of walking, which may be highly significant in neurologically impaired individuals. These include differences in joint and temporal distance measures (Strathy et al., 1993; Murray et al., 1985), EMG activity (Murray et al., 1985) and energy demands (Pearce et al., 1983). Subjects adopted a more protective pattern of walking on the treadmill. From a neurophysiological point of view, treadmill walking requires tracking of an externally produced velocity with different peripheral proprioceptive inputs (Wetzel et al., 1975). The external environment is stationary and visual and vestibular information, which are important in maintaining equilibrium has to be used quite differently when walking on a treadmill. In general, the speed of walking is being externally driven by the treadmill, rather than internally driven by the individual.

In view of the differences between the two modes of walking, it is debatable whether there is sufficient carry over into functional ambulation following treadmill training. Even though unloading during treadmill training has been shown to be efficacious in gait training, perhaps even greater benefit may be derived from overground training with support. Overground walking is the most task specific approach to gait rehabilitation and a system that would facilitate an early start to overground ambulation would be ideal.

The objectives of this study therefore were to determine the immediate effects of unloading on the gait pattern of stroke clients during overground ambulation, and to determine whether an overhead suspension system, providing support during overground walking, is feasible for use as a gait training tool in stroke clients. The target population was stroke clients with mild to moderate gait dysfunction.

SUBJECTS AND METHODS

Subjects

The study was conducted at the Posture and Gait laboratory at the Jewish Rehabilitation Hospital and was approved by the ethics committee at the said hospital. A total of fourteen subjects (5 females, 9 males) participated in the study. Subjects were between 3 to 18 weeks post stroke and had a Timed-up-and-go score (TUG) of > 25 seconds. The ages ranged from 51 to 85 years with the mean age being 69. Seven subjects used an ankle foot orthosis (AFO) and seven did not. Three of the subjects used a walker for daily ambulation and eleven used a cane (table 1). Subjects were excluded on the following basis: onset of CVA greater than six months; inability to understand simple verbal commands due to language impairments; inability to complete a TUG test; TUG score < 25 seconds; any coexisting orthopaedic conditions that may affect gait performance; and not consenting to participating in the study.

<table 1 near here>

Experimental protocol

Prior to data acquisition, subjects walked at each level of support in order to familiarize themselves with walking in the harness at all levels of support. All individuals were required to complete five to ten walking trials on a seven metre walkway at each BWS level (0% - full weight bearing, 15%, 30%). Individuals were asked to walk at their comfortable speed. The use of an AFO was allowed but not the walking aid. This decision was based on the fact that most subjects, during the adaptation period, preferred to walk without an assistive device at increasing levels of support. Nonetheless, a physical therapist walked along with the subject and provided any manual assistance that was needed. The order of the trials was randomized to control for carry over and learning effects. Sufficient rests were given in between trials to minimize the influence of fatigue and decreased endurance. Age, initial general functional level, depression, cognitive impairment, and the number of existing comorbid conditions have been shown to influence walking ability (Barron et al., 1967; Gowland, 1982; Gabell et al., 1982; Parikh et al., 1987). However, these confounders were controlled in the present study, which utilized a within-subject design. Furthermore, all the trials were done on the same day thereby eliminating the possibility of day to day changes.

Overground system

The overground system used for the study was custom-built by Les Ateliers Fabrica Inc. at the Posture and Gait laboratory at The Jewish Rehabilitation Hospital. The system consists of standard Unistrut metal channels bolted to the ceiling over a 7 m walkway as shown in fig. 1A. A metal bar with a trolley assembly slides at low friction along the entire length of the Unistrut channel (fig. 1B). This metal bar contains a small electric torque motor, which is connected to an overhead suspension bar by cables operating over a pulley system (fig 1A). The suspension bar has two short straps which attaches to the harness by strap hooks. Once the subject is strapped into the harness and secured to the suspension system by the strap hooks, lifting is provided by the electric motor. A strain gauge load transducer, which also runs along the track in a trolley assembly, is connected at the other end of the suspension cable (fig. 1B). When the lift is provided, the amount of weight supported is detected by the transducer, and the output (in pounds) is transmitted to a digital meter (fig. 1C). The amount of weight supported is displayed in pounds by the meter. Details of the BWS harness have been described in previous studies (Norman et al., 1995;Visintin et al., 1998).

<figure 1 near here>

Data Acquisition

Subjects were only instructed to walk at their comfortable pace, and not to look at their foot placements on the force plates. Thus in some walking trials, the force signals from each plate were resultants of the left and right feet combined. In such cases the loading forces had to be mathematically decomposed into those of the individual foot using the alogarithm of Davis & Cavanagh (1993). Details are provided in the appendix. Kinetic data were collected at 100 Hz using two AMTI force plates mounted within the floor in the middle of the walkway. The triaxial forces (loading, longitudinal and transverse shears) and moments were acquired.

Reflective markers were affixed to anatomical landmarks as shown in Fig. 2A. Three-dimensional and bilateral motions of the full body were acquired using the BTS-ELITE system at 50Hz with four high-resolution cameras.

Stance and swing durations under each foot were determined with footswitch signals, using a pair of in-sole pressure devices. Pressure signals from the heel, ball and big toe of the foot were output as analog signals and acquired at 100 Hz by the BTS-ELITE system. This data was used to determine selected temporal distance variables.

<figure 2 near here>

ANALYSIS

Five walking trials at each BWS level were analyzed. The data were normalized to the gait cycle (from one foot floor contact, 0%, to the next, 100%). The magnitude of the loading force was also normalized to the subject's weight [(Vertical force/weight)x100] which changed as weight was removed by the BWS system. The integrated area under the normalized force curve was calculated and an Impulse Index (Fig. 3A) was derived as follows:-

Impulse Index = A/t where

A = area under the normalized force curve and

t = stance duration (% gait cycle)

This Impulse Index would indicate the overall efficiency in force generation under the loaded limb. We also analyzed the actual (non-normalized) forces in terms of the maximal load acceptance (first force peak, P_1, Fig. 3B) during initial stance and maximal push-off force (second force peak, P_2, Fig. 3C) generated through mid to terminal stance. The two parameters of interest were:

(1) ΔP_1 (Fig. 3B), (the difference in load acceptance from weight) which indicate the

amount of impact made by foot floor contact and how well the limb can support weight. Normal subjects accept load above and beyond one's weight.

(2) ΔP_2 (Fig. 3C), (the difference in the push-off load from weight) which indicates how well the limb is propelled into swing. Normal subjects push-off at a loading force above and beyond one's weight.

<figure 3 near here>

ELITE software was used to track the 3-D spatial positions of each marker and calculate hip, knee, ankle and trunk angular displacements in the sagittal plane (Fig. 2B). Orientation of the pectoral and pelvic segments in the frontal and horizontal planes were also calculated (Fig. 2C & D).

Critical gait events such as foot-floor contact and toe-off were determined using an interactive computer program in ELITE. Temporal distance parameters analyzed were cycle duration, single limb support duration, double support duration, percentage swing, percentage stance, and the stance-swing ratio. Step and stride lengths were calculated from the sagittal trajectories of the calcaneus and fifth metatarsal markers. All mathematical analyses were performed with Excel (Microsoft) and Matlab (Mathworks).

Statistical analyses for all the variables were done in Systat (SPSF) using a one-way analysis of variance with BWS as the independent variable. In cases where a significant main effect due to BWS was obtained ($p < 0.001$), further post hoc analyses were done using the Tukey test of pairwise comparisons.

RESULTS

Kinetics

<figure 4 near here>

An example of the normalized loading force profile changes due to BWS is shown in Fig. 4A. At full weight bearing (0% BWS), there were small peaks under the paretic limb, whereas a constant vertical force was observed for the non-paretic limb, neither of which exceeded body weight. With BWS there was two distinct force peaks which were seen to exceed body weight for both limbs. In other subjects not illustrated there was the appearance of two force peaks with increasing levels of support.

The impulse index provided an indication of the total force generation during stance (fig. 4B). There was a progressive increase in the impulse index with increasing levels of support for both extremities. For the paretic limb this increase was significant at 30% support ($P < 0.001$), whereas for the non-paretic limb the changes were significant at both 15 and 30% support ($p < 0.001$). These changes indicated that there was an increase in the total force generation of both extremities when weight support was provided.

Figure 4C shows the force change during the loading phase of the gait cycle. For both extremities there was a negative force difference when no support was provided, which indicated that the maximal vertical force during the loading phase did not exceed body weight. The force difference was seen to increase with increasing levels of support. For the paretic limb the increase was significant at both levels of support ($P < 0.001$), however the difference was only positive at 30% support, which indicated that at this level of support subjects were able to generate forces that exceeded body weight. For the non-paretic limb the changes were also significant at both levels of support, ($P < 0.01$ at 15% support and $P < 0.001$ at 30% support), however, the force difference was positive even at 15% support. This indicated that 15% support was already sufficient to allow the non-paretic limb to generate forces that exceeded body weight.

The changes observed for the push off force change (fig. 4D) were similar to those seen during the loading phase. There was a progressive increase in the force difference for both extremities which was significant at both levels of support. For the paretic limb the vertical force was seen to exceed body weight at 30% support (+ve force difference), whereas, for the non-paretic limb the vertical force exceeded body weight at both 15 and 30% support. The magnitude of the force generated was improved for both limbs but figure 4 shows that there remained a degree of asymmetry.

Kinematics

<figure 5 near here>

Most of the significant changes in kinematics were observed in the trunk. Figure (5A) showed a decrease in trunk excursion with increasing levels of support, which was significant at 30% BWS ($P < 0.05$). This decrease in trunk excursion indicates that the subjects were able to maintain a more upright posture with body weight support. In fact the kinematic profile of the subject in figure (5A) shows that the trunk was approximately neutral for the entire gait cycle at 15 and 30% support. At full weight bearing the subject was seen to be maintaining a more forward bending posture throughout the gait cycle.

Postural assessments of stroke clients usually reveal the paretic shoulder to be in a depressed position with respect to the non-paretic side. In figure (5B) it can be seen that the subject was able to maintain approximately equal shoulder levels when support was provided. This was reflected in a decrease in pectoral excursion in the frontal plane, which was significant at both 15 and 30% support ($P < 0.001$). With no support the paretic shoulder was in a depressed position for most of the gait cycle.

Subjects demonstrated a significant increase in pelvis excursion in the horizontal plane with 30% BWS ($P < 0.01$). Stroke clients commonly experience difficult with bringing the pelvis on the affected side forward during the gait cycle. In fact the paretic hip is usually seen to be lagging behind throughout the gait cycle. In figure (5C) it can be seen that the paretic hip was lagging behind for the entire gait cycle when no support was provided. With 15 and 30% BWS there was increased rotation of the pelvis with the paretic hip crossing the midline, however, the degree of rotation is less than that of unimpaired individuals (approximately 10 degrees, Perry, 1992).

The only significant change in lower extremity kinematics was observed in the knee at loading (fig. 6). With increasing levels of support, there was a significant decrease in knee flexion observed in both the paretic and non-paretic extremities during load acceptance (15% of the gait cycle). Coinciding with the changes at the knee was a decrease in passive dorsiflexion at the ankle, which usually results from a flexed knee (fig. 6).

<figure 6 near here>

Temporal Distance Factors

<figure 7 near here>

With BWS there was a progressive decrease in the cycle duration of both lower extremities (fig. 7A). This decrease was significant at 30% support and indicates an increase in cadence. Following a stroke, there is a decrease in single limb stance time on the paretic limb, which results in an increase in the total double support time. With increasing levels of BWS there was a decrease in the total double support time, which was significant at both 15 and 30% support (fig. 7B). This decrease in double support indicates that subjects were spending more time in stance on the paretic limb, and this is reflected in a trend for an increase in single limb support time (fig. 7B). Even though there were improvements in temporal distance factors figure 7A and C shows that there was still a bit of asymmetry. Other temporal distance parameters were not significantly affected by BWS.

DISCUSSION

The results of the study showed that the use of BWS during overground walking leads to immediate improvements in kinetics, kinematics and temporal distance factors in stroke clients.

During full weight bearing, neither of the extremities exhibited the typical loading force pattern which consists of two peaks exceeding body weight by approximately10% (Perry, 1992). The non-paretic limb was even more abnormal than the paretic limb. Crenna and Frigo (1985) have reported similar findings of a constant vertical force development in the non-paretic side of hemiplegic gait. The patterns observed indicated that there were problems with loading and propulsion for *both* lower extremities post stroke. Some of the variation may be related to the walking speed, however it has been shown that the two peaks are also seen in normals walking at slow speeds, but the amplitude of the peaks is less than that for higher speeds (Crenna & Frigo, 1985). With BWS, there was improved weight acceptance for both limbs, however, there remained a degree of asymmetry in the amount of weight borne on each limb. Perhaps with training, a symmetrical pattern of weight bearing may be attained. The peak at push off indicates a downward acceleration and lowering of the centre of gravity as the body weight falls forward over the forefoot rocker in terminal stance (Perry 1992). These changes provide the acceleration for forward propulsion of the body. The improved force change at push-off therefore indicated improved propulsion with support, which may lead to a more energy efficient gait pattern.

The kinematic changes observed in the trunk indicate improved postural control with BWS, which may result in improvements in balance and stability during walking. The changes observed in the sagittal plane were similar to that observed in both stroke clients and individuals with SCI during supported walking on the treadmill (Visintin et al., 1989; Hesse et al., 1997). However, none of the prior studies reported on kinematics in the frontal or horizontal planes. The finding of a straighter knee at loading was also reported for stroke clients and individuals with SCI when walking on a treadmill with support (Barbeau et al., 1998; Hesse, 1997; Visintin et al., 1989). These changes indicate improved loading and may be related to the improvements in the vertical ground reaction forces. From a clinical perspective, a straighter knee at loading decreases the risk of buckling of the knee, which can lead to falls in stroke clients. The changes in temporal distance parameters indicate that subjects were spending more time in single limb stance when support was provided, and there was also an increase in cadence as reflected by a decrease in cycle duration for both limbs.

Task specific gait training in stroke

Traditional models of rehabilitation have focused on training isolated components of gait in order to improve walking in stroke clients. Training incorporates a variety of facilitation techniques and is based on a reflex-hierarchical model of motor control. Current developments in the motor control literature have given rise to a new model of rehabilitation based on dynamical systems and focuses on a more task-oriented approach to training (Kamm et al., 1990; Carr et al., 1987; Shumway-Cook et al., 1995; Horak, 1987). This model is still in the process of being refined and validated, however, studies that have looked at its application to gait training have been favorable (Richards et al., 1993; Malouin et al., 1992; Finch et al., 1991). The key component of training was the actual practicing of gait in order to improve gait. Waajfjord et al (1990) using a single subject design found that treadmill training led to improvements in base of support and increased symmetry of step length. Richards et al. (1993) in a study of 27 stroke patients found that task specific gait training was well tolerated and led to significant improvements in gait, as opposed to conventional training.

The task specific approach to gait training has led to an increase in the use of the treadmill for gait rehabilitation in stroke clients. However, with more severe deficits, it is exceptionally difficult to start training early since these individuals have major difficulty with weight bearing. This is an important factor to consider since it is known that the earlier training is started, the more favorable the outcome (Novack et al., 1984). The use of a harness, which provides partial support during walking, facilitates an early start to training walking. Treadmill training is based on the assumption that it is similar to overground walking. In normal individuals, the two modes of walking were seen to be quite similar (Arsenault et al., 1986; Murray et al., 1985; Nelson et al., 1972; Pearce et al., 1983; Strathy et al., 1983; Stolze et al., 1996). However, several differences were identified which may be highly significant in a neurologically compromised population. On the treadmill the excursion of the knee joint is decreased throughout both phases of the gait cycle (Strathy et al., 1983). Stride length is decreased at the individual's comfortable walking speed and cadence is increased (Murray et al., 1985, Arsenault et al., 1986). The EMG activity of the quadriceps increases with treadmill walking (Murray et al., 1985) and Pearce et al., (1983) also demonstrated that a higher predicted energy cost value was associated with treadmill walking.

Van Ingen Schenau (1980) stated that, from a purely mechanical point of view, there is no difference between the two modes of walking when the mechanical variables are described with respect to the surface on which the subject locomotes. From a motor control perspective, walking on a treadmill is considered to be a more regulatory task. Speed is dictated by the treadmill belt, which also acts as a stimulus to facilitate stepping. With overground walking speed is internally controlled and involves more feedforward, anticipatory motor processing, as opposed to treadmill walking where motor control is more reactionary involving feedback processing. One of the major assumptions underlying the dynamical systems theory is that control of a specific motor task can best be regained by practice of that specific motor task, and that such task needs to be practiced in their various environmental contexts. From this perspective therefore, treadmill training is not sufficient. Training overground walking would be the most task-specific approach to gait rehabilitation since it allows for training in several different contexts, and functional walking requires the ability to walk overground. Applying the concept of BWS to an overground system would allow for an early start to training overground ambulation. This study has shown that an overground suspension system is feasible for use as a gait training tool in stroke clients with mild to moderate disability. Such a system would allow for greater variability in gait training. The main advantage is that it would facilitate an early start to training overground walking which is required for functional community ambulation.

The constraints of load in stroke

Following a stroke, there is decreased weight bearing capacity of the paretic limb, as well as problems with weight shifting (Bohannon et al., 1985; Dickstein et al., 1984; Seliktar et al., 1978). This is thought to contribute significantly to some of the asymmetries observed in the gait pattern of stroke clients. The ratio of weight over the non-paretic lower extremity to weight over the paretic limb was found to be 1.46 which exceeds the upper limit of 1.09 for a normal geriatric control population (Dickstein et al., 1984). The majority of these individuals use an assistive device for ambulation, which further encourages less weight bearing on the paretic limb. Traditional rehabilitation has focused on weight shifting activities in order to increase weight bearing on the paretic limb. This approach assumes that increased weight bearing would facilitate normal activity of the lower extremity muscles.

Peat et al., (1976) found that previously unloaded or inactive muscles are activated if subjects bear more weight on the affected limb. The increased load on a muscle stretches the spindle receptors, thereby increasing Ia discharge and facilitating the homogynous and synergistic motorneurons that can facilitate a muscle contraction. Even though there was increased activation with weight bearing, the timing was inappropriate. It is possible that the weight transferred to the paretic limb was too much for appropriate recruitment of muscles, whereas grading the weight to the muscles' capabilities may lead to a more efficient pattern of recruitment. Apart from the effect on the Ia afferents, appropriate loading during gait may cause excitation of the Ib afferents of the extensor muscles during stance (Pearson, 1992; Pearson et al 1992, 1993; Van de Crommert, 1998). This would lead to increased extensor activity, which may explain the improved knee extension during stance observed in both stroke and SCI subjects walking with BWS. The BWS system allows for graded loading of the extremity, and in subjects with spinal cord injuries, a more appropriate timing of muscle activity was noted during supported walking (Visintin et al., 1989). Even though the initial stages of training involves load removal, the ultimate goal is to promote weight bearing.

The Ib afferents from the extensor muscles are known to exert a widespread inhibitory influence on the motor neurons of the lower extremity extensors. However, during gait this effect is reversed and instead of being inhibitory the pathways become facilitatory during stance (Pearson et al., 1992; Pearson et al., 1993; Van de Crommert et al., 1998; Duysens et al., 1998). As the limb moves into swing the Ib afferents become inhibitory once more allowing the limb to flex. This reversal allows for a smooth gait pattern. The Ib afferents have a higher threshold of activation than the Ia afferents, therefore sufficient load must be transferred onto the limb in order to attain this reversal. Too much or too little load may cause the limb to yield into flexion. Hemiplegic subjects often complain of the sensation of "heaviness" in the paretic limb. Graded loading through the use of BWS systems allows for the amount of stretch on the muscle to be controlled at a level that meets the subject's capabilities. Once the demand of coping with weight is satisfied, subjects may then be able to focus on other important components of gait such as postural control and balance.

The concept of graded weight bearing may be supported by infant literature. A conventional assumption in human motor development was that the stepping reflex needed to be suppressed between 4 to 5 months after birth, a phenomenon known as astasia, in oredr for cortical control to take over (Forssberg, 1985). However, Thelen (1984) showed that stepping could reappear in infants with astasia when they were submerged up to their trunk in water. The weight support provided by the water allowed for expression of the gait pattern. In addition Forssberg, (1985) noted a continuous maturation of the locomotor pattern in infants during the entire phase of supported walking. By the end of the support phase infant's pattern were similar to that of independent walking. In the same manner progressive weight bearing in stroke clients may lead to a smooth progression of the development of load compensating mechanisms.

Functional implications

The use of an overground BWS system allows for an early start to retraining actual overground walking. However, the degree of support provided needs to be carefully choosen. On the treadmill, support levels exceeding 40% produced changes that differed from normal physiological gait. Recent studies on the treadmill seem to suggest that between 15 and 30% support is optimal (Hesse et al., 1997; Hassid, 1997). The grouped data from this study showed that 30% support led to most improvement, however some subjects performed better at 15% support. The vertical ground reaction force appeared to be a good parameter for determining the level of support at which to start training. The level of support which produces a force pattern with two peaks exceeding body weight may be the most appropriate starting point. In cases where the pattern is very similar, then the lowest level of support should be used in order to challenge the system.

Functional ambulation requires walking at different speeds as well as stopping and starting again. Not only is the individual moving but also the environment as seen by the individual. An overground system allows for interactive gait training in which the environment can be altered to mimic certain aspects of a community environment. The surface may be changed so that the individual learns to walk on different surfaces, and obstacles may be placed along the walkway. It would also be possible to have people walking around the client as well as across his path so that he learns to cope with a moving environment in which he is also moving. The potentials for this system in gait training is numerous. An important next step is to investigate the effects of long term training to verify whether improvement continues and if this is transferable to community ambulation. The full spectrum of the stroke population in which the system is applicable also needs to be determined.

In conclusion, the use of BWS during overground ambualtion leads to immediate and significant improvements in ground reaction forces, body kinematics, and temporal distance factors in stroke clients with moderate disability. This overhead suspension system which provides support during overground walking is feasible for use as a gait training tool in the rehabilitation of stroke clients with moderate disability.

Table 1

Profile of the 14 subjects who participated in the study

Subject	Sex	Age	Affected side	Time post stroke (weeks)	Assistive Device	TUG score (sec)
1	F	85	L	6	walker	29
2	F	82	L	7	walker	26
3	F	66	L	9	cane	38
4	M	73	L	4	cane + AFO	30
5	M	51	L	16	cane	25
6	M	69	R	18	Cane + AFO	42
7	M	75	L	4	Cane + AFO	48
8	F	75	R	8	Walker + AFO	38
9	M	57	L	9	cane	30
10	M	57	L	11	cane	28
11	M	78	L	4	cane	33
12	M	63	L	12	Cane + AFO	31
13	M	73	R	3	Cane + AFO	31
14	F	62	L	7	Cane + AFO	32

LEGENDS

FIGURES

Figure 1

(A) Overground support system and posterior view of a subject walking with weight support. (B) Expanded view of the strain gauge transducer and motor lift. (C) Digital meter providing a read out of the amount of weight supported in pounds.

Figure 2

(A) marker placements and body model used for data acquisition and tracking in the ELITE system. (B) Angles calculated in the sagittal plane. (C) Angles calculated in the frontal plane. (D) Angles calculated in the horizontal plane.

Figure 3

(A) A method for calculating the impulse index. (B & C) P_1 and P_2 corresponds to the two peaks involved with the different phases of loading and push off respectively. The difference between P_1, P_2 and weight provides the two (B) $< P_1$ and (C) $< P_2$.

Figure 4

(A) Vertical ground reaction force normalized to weight, at different levels of BWS plotted against a normalized gait cycle (0% - one foot floor contact to 100% - next foot floor contact) for subject 2. The vertical force exceeds body weight with increasing levels of support. Figure 4 B, C & D Mean ! S.E. for selected force parameters. (B) Impulse Index increases with increasing levels of support. (C) Load acceptance force change: a positive value indicates the presence a vertical force exceeding body weight. (D) Push-off force change: values increase with increasing levels of support.

Figure 5

Kinematics of the trunk and girdles. (A) Trunk excursion in the sagittal plane. There is a significant decrease in trunk excursion at 30% support. (B) Pectoral excursion in the frontal plane. With support the shoulder levels become more equal. (C) Pelvis rotation in the horizontal plane. There is increased rotation with the pelvis crossing the midline at 15 and 30% support.

Figure 6

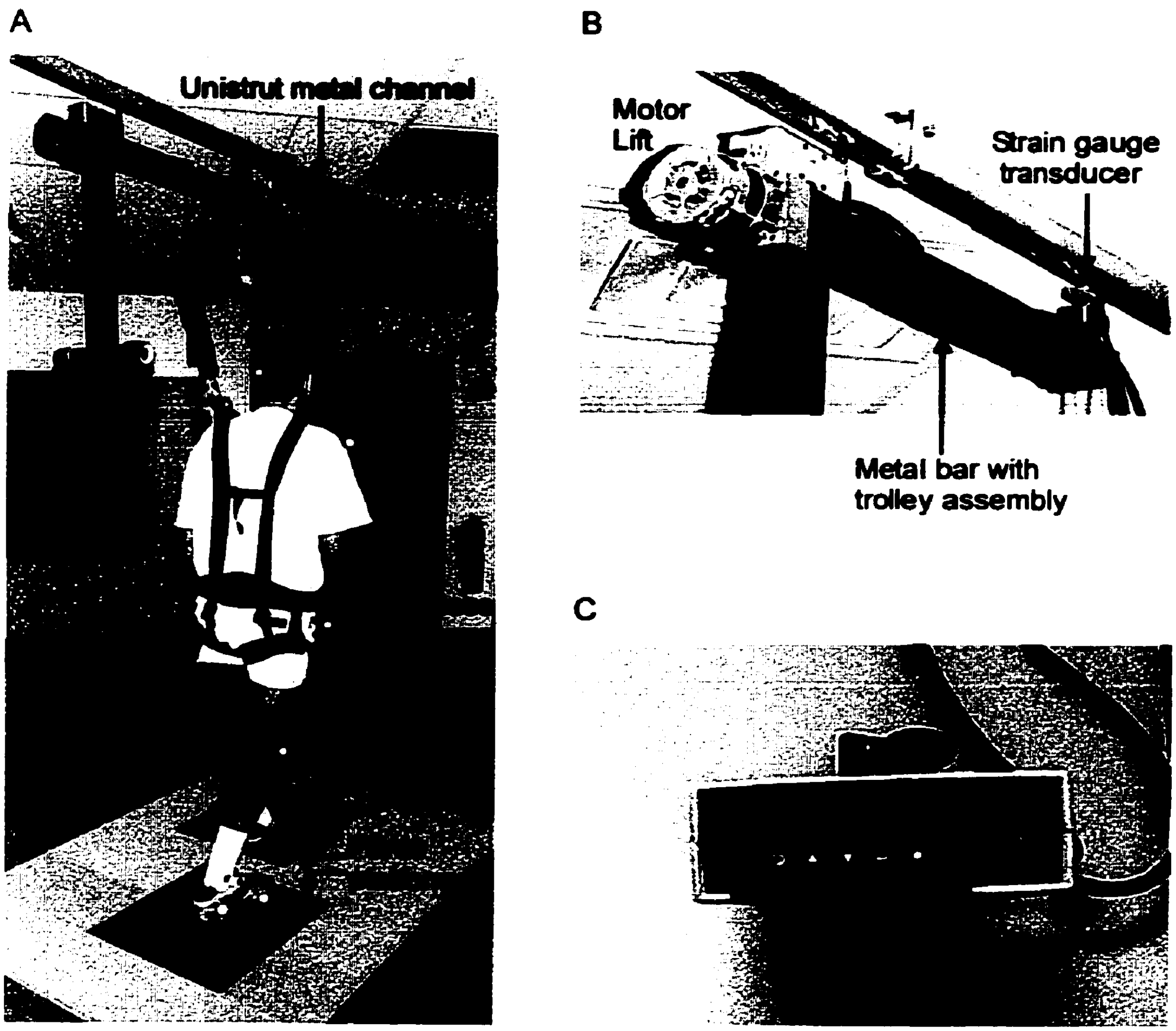
(A) Kinematic profile for subject 6 at different levels of support plotted against a normalized gait cycle. A significant decrease in knee flexion is noted at loading together with a passive decrease in dorsiflexion. (B) Mean ! S.E for knee angle at loading.

Figure 7

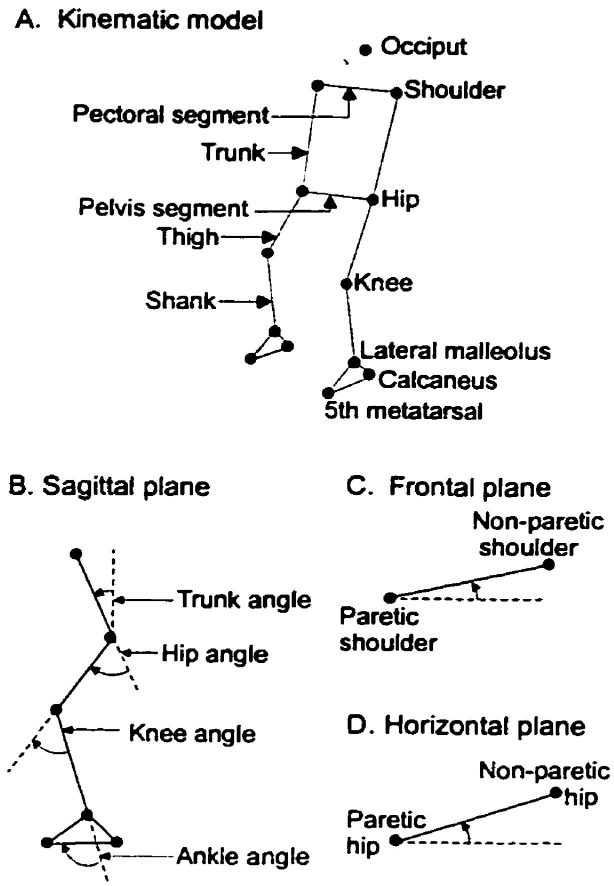
Mean ! S.E for certain temporal distance factors. (A) Cycle duration (B) Total double support duration and (C) Single limb support duration.

APPENDIX

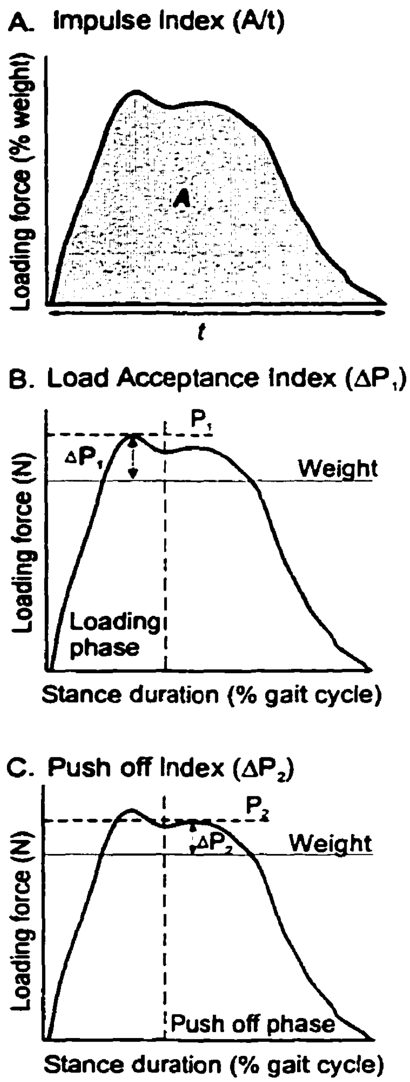
Method used for decomposing the ground reaction forces into left and right force profiles.



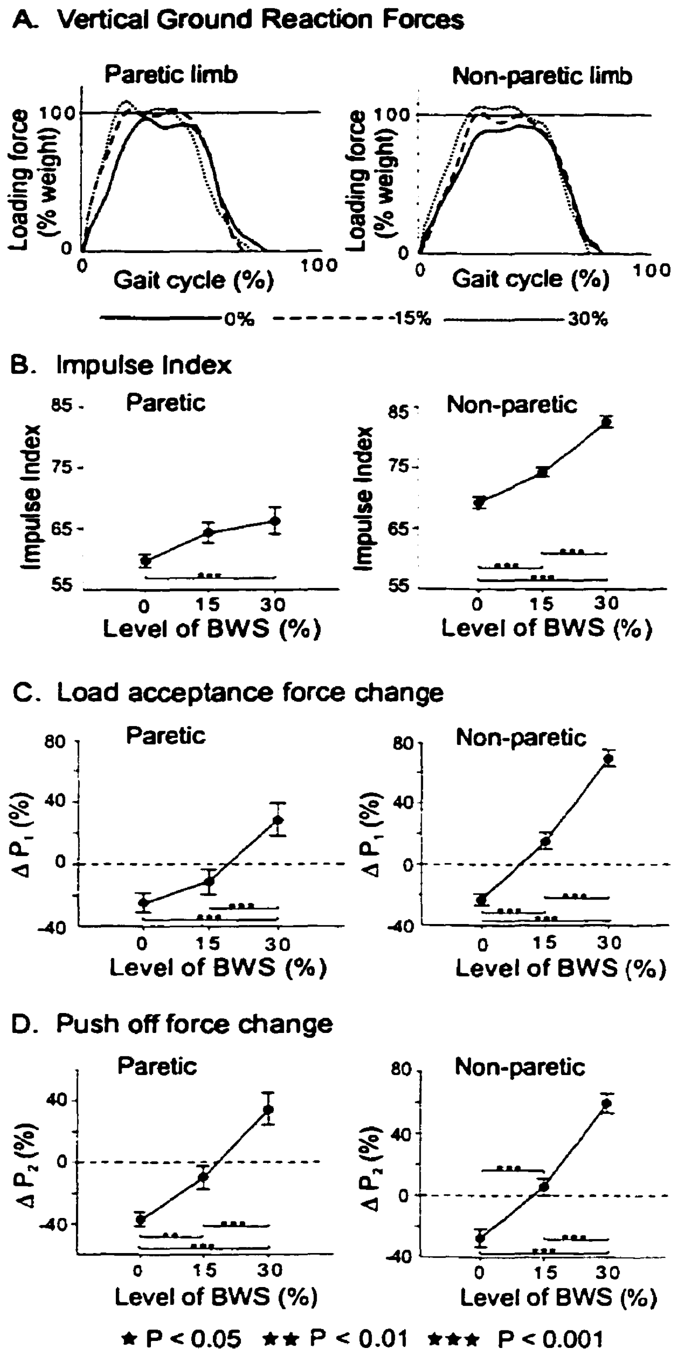
Roopchand et al., Fig. 1



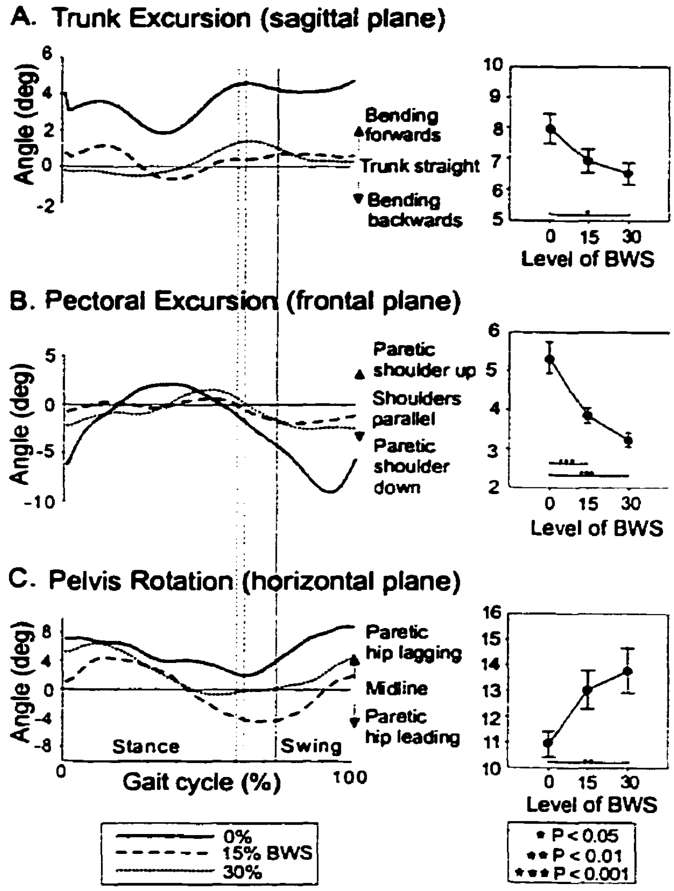
Roopchand et al., Fig. 2



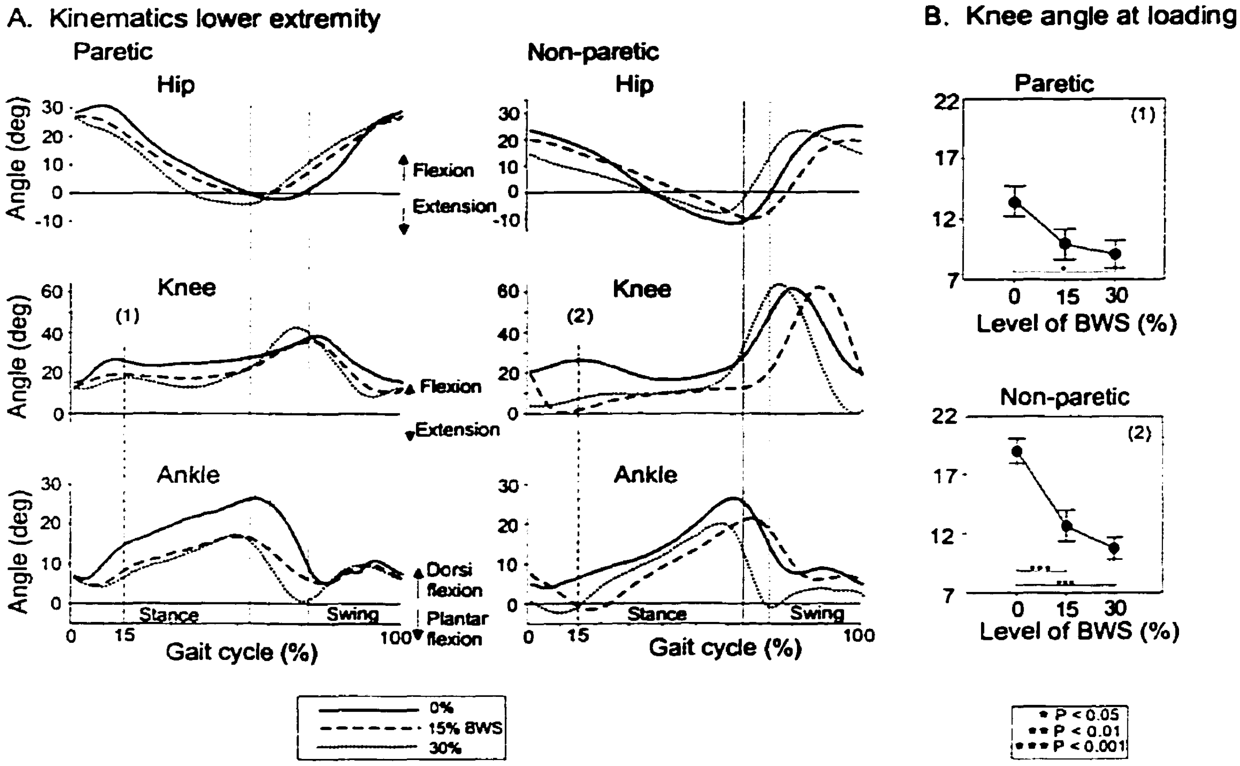
Roopchand et al., Fig. 3



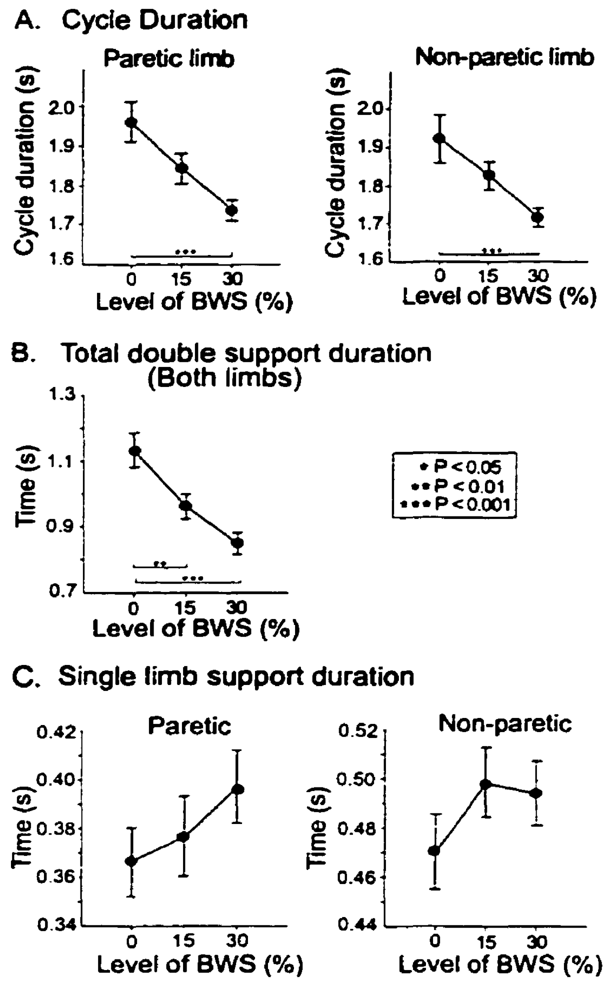
Roopchand et al., Fig. 4



Roopchand et al., Fig. 5



Roopchand et al., Fig. 6

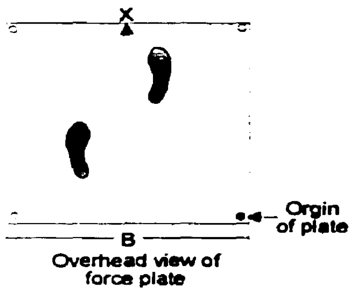


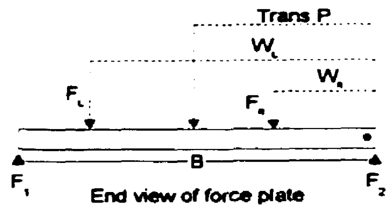
Roopchand et al., Fig. 7

Appendix

METHOD FOR DECOMPOSING LEFT AND RIGHT LOADING FORCES

The ELITE system outputs the total force exerted on the force plates (V_{Fc}) and the side to side movement of the centre of pressure (Trans P) measured from the origin of the plate.





Taking moments about the origin of the plate:

$$F_L + F_R = F_1 + F_2 \quad (1)$$

$$F_L W_L + F_R W_R = F_1 B \quad (2)$$

Where F_L is the vertical force of the left foot on the plate and Fr is the vertical force of the right foot. Solving the simultaneous equation with F_r as the subject gives:

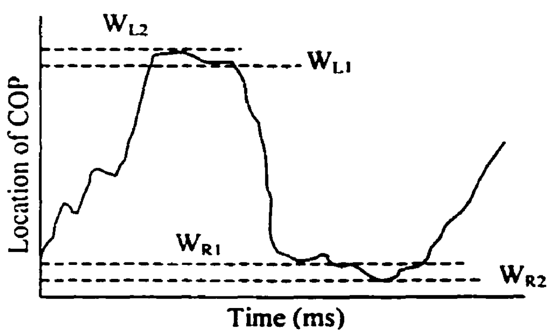
$$F_R = \frac{(F_1 + F_2)W_L - F_1 B}{W_L - W_R} \quad (3)$$

From ELITE output $F_1 + F_2 = V_{Fc}$

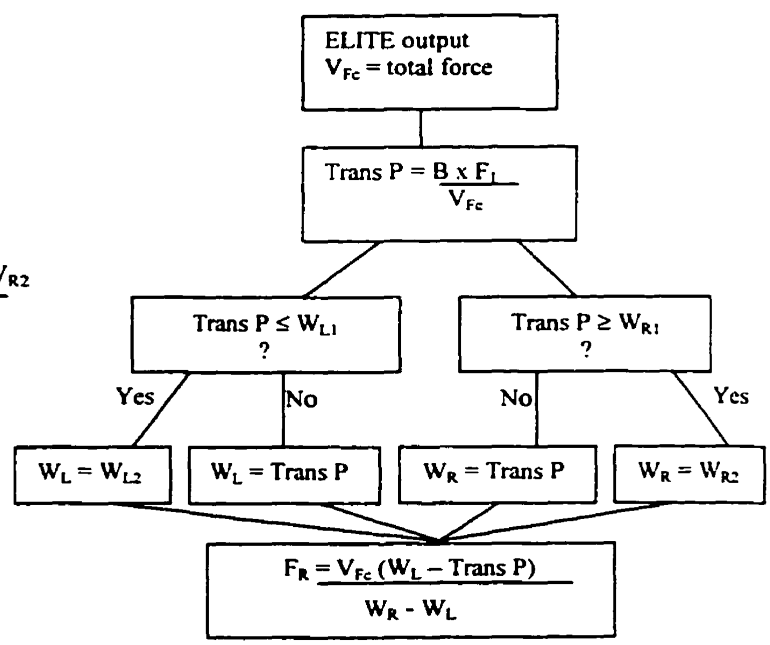
$F_1 B$ = total moment about the plate.
So $F_1 B = V_{Fc}$ x Trans P
Substituting this in equation (3)

$$Fr = \frac{V_{Fc}(W_L - \text{Trans P})}{W_L - W_R}$$

Applying the same logic to the left foot gives $F_L = \dfrac{V_{Fc}(W_R - \text{Trans P})}{W_R - W_L}$



Side to side location of the centre of pressure (COP) as subject transfers weight from left foot to right foot. Obtained from patient data collected by the ELITE system. W_{L2} and W_{R2} corresponds to the maximum and minimum excursion. W_{L1} and W_{R1} are inflection points in the trace. W_L and W_R are obtained using the alogarithm derived by Davis and cavanagh (1993) as shown in the flow diagram.



CHAPTER 5

SUMMARY AND CONCLUSIONS

Ambulation is an activity that is an integral part of the daily functioning of most individuals, and decreased ambulatory function is one of the significant contributors to disability following a stroke. The Dynamical Systems or Task Oriented Model of Rehabilitation have added a new dimension to gait training: that is emphasizing an early start to the actual training of gait in order to improve gait (Kamm et al., 1990; Shumway-Cook & Woollacott, 1995). The use of treadmill training with BWS has its foundations in animal studies, which demonstrated that adult spinalized cats were capable of regaining full weight bearing through an intensive interactive training program consisting of treadmill walking and graded weight support provided at the tail (Barbeau & Rossignol, 1987).

The use of a suspension system providing support during treadmill walking has been one of the most successful approaches to gait training. However, there are differences in both biomechanical and motor control mechanisms involved with treadmill and overground ambulation. While treadmill gait training may be useful in rehabilitation, treadmill walking alone may not prepare an individual optimally for the dynamics of community ambulation. In fact, the most task specific approach to gait training would be to start overground ambulation at an early stage. In traditional approaches this was not possible due to the inability to weight bear on the paretic limb following a stroke. The use of a BWS mechanism has provided a solution to this problem.

This study examined the effects of systematic unloading during overground walking. Three levels of support were provided, 0%, 15% and 30% BWS. The objectives of the study were to determine if the use of BWS during overground walking would lead to immediate improvements in the gait pattern of stroke clients, and whether an overground suspension system is feasible for use in gait rehabilitation of stroke clients. Very few gait studies have looked at bilateral comparisons in stroke clients and it is often wrongly assumed that the non-paretic side is normal. This study looked at bilateral comparisons of ground reaction forces, body kinematics and temporal distance variables at all three levels of support. Ground reaction forces were acquired at 100Hz using two AMTI force platforms, kinematic data were acquired at 50Hz using the ELITE four camera system and reflective markers placed on selected anatomical landmarks, and temporal distance data were obtained using pressure sensitive devices worn inside the shoes.

BWS is based on the idea that providing support would lead to a more symmetrical pattern of weight bearing in stroke clients, however none of the support studies in stroke clients have looked at ground reaction forces to determine if this is true. This study therefore provides novel information on the behavior of the ground reaction forces with weight support. A total of 14 subjects with mild to moderate gait impairments participated in the study. Subjects were allowed to wear an AFO but no assistive devices were allowed once they were placed in the harness.

When no support was provided it was noted that the vertical force curve of both the paretic and non-paretic limbs were abnormal relative to the typical force pattern of an unimpaired individual. The vertical force was constantly below body weight and in some cases the two peaks that are present in a typical force curve were completely absent. With increasing levels of support there was the appearance of two distinct peaks for both limbs which begin to exceed body weight (body weight being the amount of weight on the limb for that particular trial). More detailed analyses revealed a significant increase in force generation during loading and push-off. With support the vertical force during these two phases were seen to exceed body weight, which is as expected in a normal force curve. The impulse Index showed that there was an increase in the total force generation when support was provided. Weight acceptance was seen to improve for both limbs, however, the pattern of weight bearing remained asymmetrical. BWS therefore leads to increased force generation and weight acceptance for both limbs but does not necessarily produce a more symmetrical pattern of weight bearing. Perhaps this may only be achieved with long term training.

The changes in kinematics revealed improved postural control of the trunk with support. There was a more upright trunk posture, shoulder levels approached neutral and pelvis rotation was increased when support was provided. Lower extremity kinematics were relatively unaffected by BWS, however, there was a significant increase in extension of both knees at load acceptance. This indicated improved weight acceptance at loading and may be related to the improvements noted in the vertical forces. Analysis of the temporal distance variables revealed a significant decrease in cycle duration for both limbs with increasing levels of support. This may indicate improved cadence. The total double support duration was also decreased with increasing levels of support, and this was accompanied by a trend for increased single limb support duration.

The results of this study confirmed the initial hypotheses and it can be concluded therefore that the use of BWS during overground walking leads to immediate and significant improvements in ground reaction forces, body kinematics and temporal distance factors in stroke clients with moderate gait impairments. The experimental protocol was well tolerated by all subjects and no one complained of discomfort, therefore this study have also shown that a suspension system which provides support during overground walking is feasible for use as a gait training tool in rehabilitation of stroke clients with mild to moderate disability.

Prior studies looking at the effects of unloading in a neurological population assumes that weight support leads to a more symmetrical pattern of weight bearing, but none of these actually measured the ground reaction forces. This study therefore has provided novel information on the effects of unloading on lower extremity force generation in stroke clients. The bilateral comparisons of ground reaction forces and kinematic data adds support to the concept that the "unimpaired" limb is also affected following a stroke and needs to be addressed in stroke rehabilitation programs. The use of BWS during gait appears to have a positive impact on both parameters. Prior studies on BWS have only looked at the trunk in the sagittal plane. This study has shown that frontal and horizontal plane kinematics of the trunk may also be improved with BWS, which provides a more comprehensive picture of the effects of unloading during gait.

CHAPTER 6

LIMITATIONS

1. The subjects who participated in this study were already ambulatory and represents mild to moderate disability only. Thus the results cannot be generalized to the entire stroke population.
2. The system used did not provide constant support throughout the gait cycle. The amount of support varied as the centre of gravity moved up and down. A system that provides constant support may be better.
3. Subjects were allowed to use an AFO and this may influence the kinematics of the paretic lower limb.

CHAPTER 7

FUTURE DIRECTIONS FOR RELATED RESEARCH

1. Studies need to be conducted with more impaired individuals to determine the maximal level of disability with which this system can be used. It is possible that severely impaired clients may require initial training on the treadmill before overground training can be done.
2. Training studies need to be conducted to see if long term training would lead to a better gait pattern and whether improvements would be maintained. It would also be important to assess community ambulation following training with this system. The long term goal of gait training is to improve community ambulation and not just ambulation within a laboratory setting.
3. More detailed work needs to be done to determine if the ground reaction force is indeed the best indicator of the level of support at which training should be started. If it is, then the elements of the ground reaction force which should be used to determine progression needs to be isolated.
4. Training studies should also look at relating laboratory measures with clinical measures. In doing so a clinical tool may be identified which can be used to determine which clients would benefit from training. Most hospitals are not equipped with force platforms and a clinical measure for determining progression would be important.
5. From a motor control point of view, it would be important to look at treadmill training as opposed to overground training. This would provide information on the

amount of carry over into functional ambulation from each method of training and may provide some insights into motor control mechanisms associated with the two modes of walking.

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Appendix 1

School of Physical and Occupational Therapy
McGill University
3650 Drummond
Montreal

INFORMED CONSENT FORM

Coordinators: Sharmella Roopchand, (MSc candidate McGill University)
Tel. (514) 931-0816.

Joyce Fung, PhD, Director of the Posture and Gait Laboratory.
Jewish Rehabilitation Hospital.
Tel. (514) 688-9550 ext. 529.

Objective: The main purpose of this study is to see whether the walking pattern of stroke clients will improve when they are supported in a harness which reduces the amount of weight placed on their legs as they walk along the ground.

The evaluation: The study will take place at the Posture and Gait Laboratory of the Jewish Rehabilitation Hospital. The session will take approximately three hours of your time. You will be asked to walk along a seven metre walkway fifteen times. For some of the walking trials a small lift will be provided by the harness so that you will be placing less weight on your legs as you walk. You will be supported in the harness at all times to prevent you from falling and a therapist will also walk behind you for additional safety. During the session you will also be allowed several rest periods. Electrodes will be taped over different muscles on your legs in order to record their activity as you walk. In order to put on these electrodes, it will be necessary to shave off a small patch of hair over the area where they will be placed, and to clean the skin with alcohol. In order to record your movement as you walk small markers; which look buttons; will be placed at the back of your head, over your shoulders, hips, knees, ankles, heels, the side of your little toe and also at the middle of each thigh.

Risks related to the experiment: There are no expected risks to you during participation in this study. A therapist will always be there to provide any assistance that may be needed and the harness will prevent you from falling.

Advantages: There may be no direct benefit to you from this study, however, the results will provide information that will help in developing better techniques for teaching stroke clients to walk which can be used with clients in the future.

Confidentiality: All personal information will be held confidential and the research data will appear only in the form of a scientific presentation or publication.

Contact persons: Should you have any questions or require further information regarding the study please do not hesitate to contact Sharmella Roopchand (tel: 931-0816 or 688-9550 ext.533) or Dr. Joyce Fung (tel: 688-9550 ext529).

Please note that consent to participate in this study must be a personal choice. Also you are free to discontinue the study at any point if you choose to do so with no effect to your treatment at the hospital.

CONSENT: I have read and completely understand the above information. I am aware that I am free to withdraw from the study at any point if I wish to do so and my treatment will in no manner be affected. I hereby freely consent to participate in the study.

Subject: ______________________ Date: __________
(Signature)

______________________ Tel: __________
(Name)

Witness: ______________________ Date: __________
(Signature)

______________________ Tel: __________

(Name)

Appendix 2

Ecole de physiothérapie et ergothérapie
Université McGill
3654 rue Drummond
Montréal

Effets du support de poids sur la marche au sol, chez les patients ayant subi un accident vasculaire cérébral

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FORMULAIRE DE CONSENTEMENT

Coordonatrices: Sharmella Roopchand, PT (candidate Msc, Université McGill)
Tel. (514) 931-0816.

Joyce Fung, PhD, PT (Directrice, Laboratoire de posture et marche) Hôpital Juif de Réadaptation.
Te. (514) 688-9550 ext. 529

Objectif: Le but de cette étude est d'examiner la possibilité d'améliorer la marche après un accident vasculaire cérébral en supportant une portion du poids d'une personne avec un harnais.

L'évaluation: L'étude se déroulera au Laboratoire de posture et marche de l'Hôpital Juif de Réadaptation. La session durera environ 3 heures. On vous demandera de marcher le long d'une allée de 7 mètres, 15 fois en tout. Pour certain des essais de marche, un peu de votre poids sera supporté par le harnais. De cette façon, vous allez porter moins de poids sur vos jambes pendant la marche. À tout moment, vous allez être supporté(e)s par le harnais pour éviter que vous fassiez une chute. De plus, une thérapeute marchera derrière vous pour plus de sécurité. Pendant la session, vous pourrez prendre autant de repos que vous aurez besoin entre les essais de marche. Des électrodes jetables vont être placées sur la peau recouvrant différent muscles de vos jambes, pour enregistrer leur activité pendant que vous marchez. Pour placer ces électrodes, il sera nécessaire de raser une petite zone sur la peau et de nettoyer la peau avec de l'alcool. Pour enregistrer vos mouvements pendant que vous marchez, de petits marqueurs ronds et réfléchissants vont être placés derrière votre tête, sur vos épaules, hanches, genoux, chevilles, pieds, le côté de vos petites orteils et au milieu de chaque cuisse.

Risques reliés à l'expérience: Il n'y a aucun risque prévu à votre participation à cette étude. Une thérapeute sera toujours présente pour vous aider et le harnais vous empêchera de chuter.

Avantages: Cette étude peut ou non vous apporter un bénéfice direct. En effet, vous pourrez ou non avoir une amélioration immédiate de votre habileté à marcher suite à la session expériementale. Cependant, les résultats de cette étude vont apporter des informations qui vont aider à développer de meilleure techniques pour enseigner aux patients hémiplégiques à marcher.

Confidentialité: Toute information personnelle sera gardée confidentielle, et les données de recherche vont apparaitre seulement sous forme de présentation scientifique ou de publication, sans que votre nom y apparaisse.

CONSENTEMENT: Vous pouvez être assuré(e) que l'information que vous avez reçu concernant ce projet est exacte et complète. Votre participation à ce projet est entièrement volontaire. Votre refus à participer n'affecterait en rien le traitement que vous recevez dans cet hôpital. De plus, vous pouvez vous retirer de cette étude à tout moment.

Pour obtenir réponse à toute question ou information supplémentaire en rapport à cette étude, n'hésitez pas à contacter Sharmella Roopchand (tel. 931-0816 ou 688-9550 ext. 533) ou Dr. Joyce Fung (tel. 688-9550 ext. 529).

Votre signature indique que vous avez lu ce formulaire, que vous comprenez le but de la recherche, que ce projet peut ou non vous apporter un bénéfice direct, et que vous acceptez de participer. Une copie de ce formulaire vous sera remise pour vos dossiers.

Sujet: ______________________ Date: ___________
(Signature)

______________________ Tel: ___________
(Nom)

Témoin: ______________________ Date: ___________
(Signature)

______________________ Tel: ___________
(Nom)