ABSTRACT

Background: Though numerous studies have compared overground and treadmill walking there still exists a significant debate about whether the two modes of walking are equivalent. The present study provides a comprehensive evaluation of overground and treadmill walking at matched speeds and increasing treadmill speeds. Walking performance was compared in healthy adults, in people with stroke and between the groups. This is important to know because any differences may have implications for gait training in both groups. Methods: Ten healthy adults (50-73 years) and ten subjects with stroke (54-80 years) walked at their self-selected speed overground which was matched on a treadmill. Temporal parameters, angular kinematics and vertical ground reaction forces were recorded during walking once subjects were in steady state as determined from their heart rate and oxygen uptake, both of which were also recorded. Belt speed was then increased 10% and 20% above matched speed and steady state recordings obtained. Speed related adjustments were also evaluated and compared between the two groups of subjects. Results: For healthy adults, step, stride, and joint angular kinematics were similar for both modes of walking. Small reductions in double support time and decreased push-off force were evident on the treadmill. For subjects with stroke, step, stride, and stance times were longer when walking overground but the degree of symmetry was comparable for both surfaces. Kinematic data revealed interlimb asymmetry was more pronounced for all lower limb joint excursions during overground walking and vertical forces were higher. In comparison to healthy adults, stroke subjects walked with lower cadence, shorter strides, lower stance time, and smaller lower limb joint excursions than their healthy counterparts. When compared with overground
walking the metabolic requirements of treadmill walking for healthy adults and subjects with stroke however were about higher by 23% and 15% respectively. All temporal-distance parameters, hip joint excursion, F1 and F2 forces and metabolic costs showed main effects of speed. An interaction between speed and group indicated that oxygen consumption increased at a greater rate in stroke than healthy subjects. **Conclusions:** The findings suggest that, although overground and treadmill gait patterns are similar for each group of subjects, people with stroke adopt a more symmetrical kinematic walking pattern on the treadmill that is maintained at faster belt speeds. Although there are differences in gait patterns between healthy and stroke subjects, both groups respond to the challenge of increased walking speed in the same way. One important difference is the abnormal elevation of energy demands associated with treadmill walking at faster speeds in stroke. Clinically, this warrants consideration as it may lead to premature fatigue and undesirable cardiorespiratory challenge in this group of individuals.
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TABLE OF CONTENTS

ABSTRACT ................................................................................................................................. i
ACKNOWLEDGEMENTS ............................................................................................................ iii

CHAPTER 1
Introduction ................................................................................................................................. 1
  1.1 Objectives ......................................................................................................................... 6
REFERENCES .................................................................................................................................. 7

CHAPTER 2
Biomechanics and Metabolic Costs of Overground and Treadmill Walking in Healthy Adults ......................................................................................................................... 9
Abstract ....................................................................................................................................... 9
  2.1 Introduction ......................................................................................................................... 10
    2.1.1 Temporal and Distance Parameters ......................................................................... 10
    2.1.2 Kinematics ................................................................................................................. 11
    2.1.3 Kinetics ...................................................................................................................... 13
    2.1.4 Metabolic Costs of Walking in Healthy Adults ......................................................... 14
  2.2 Purpose .............................................................................................................................. 15
  2.3 Methodology ...................................................................................................................... 15
    2.3.1 Procedure ................................................................................................................... 16
    2.3.2 Instrumentation and Data Acquisition .................................................................... 16
    2.3.3 Overground Walking ............................................................................................... 18
    2.3.4 Treadmill Walking .................................................................................................... 19
    2.3.5 Data Processing ........................................................................................................ 20
      2.3.5.1 Temporal and Distance Parameters ................................................................. 22
      2.3.5.2 Kinematic and Kinetic Data ............................................................................. 22
      2.3.5.3 Metabolic Data ................................................................................................. 23
  2.4 Data Analysis .................................................................................................................... 23
  2.5 Results ............................................................................................................................... 23
LIST OF TABLES

CHAPTER 2
Table 2.1 Gait speed (m/s) and cadence (steps/min) for individual subjects............ 24
Table 2.2 Mean values (± 1SD) of temporal distance parameters for overground and treadmill walking ............................................................... 26
Table 2.3 Mean (± 1SD) of angular displacement at the hip, knee and ankle joints (in degrees) during stance................................................................. 27
Table 2.4 Mean (± 1SD) vertical ground reaction forces expressed as a proportion of body weight................................................................................ 30
Table 2.5 Metabolic data (mean ± 1SD) for overground and treadmill walking...... 32

CHAPTER 3
Table 3.1 Gait speed (m/s) and cadence (steps/min) for individual subjects............ 53
Table 3.2 Mean values (± 1SD) of temporal distance parameters for overground and treadmill walking ............................................................... 54
Table 3.3 Mean values (± 1SD) of temporal distance parameters for the unaffected and affected sides for overground and treadmill walking................. 55
Table 3.4 Mean (± 1SD) of angular displacements of the unaffected and affected hip, knee and ankle joints (in degrees) during stance ....................... 57
Table 3.5 Mean (± 1SD) vertical ground reaction forces for the unaffected and affected sides expressed as a proportion of body weight............... 61
Table 3.6 Metabolic data (mean ± 1SD) for overground and treadmill walking..... 64

CHAPTER 4
Table 4.1 Mean values (± 1SD) of temporal distance parameters for overground and treadmill walking (matched speeds) and the percentage change associated with increased treadmill speed........................................ 88
Table 4.2 Mean values (± 1SD) of joint excursions for overground and treadmill walking (matched speeds) and the percentage change associated with increased treadmill speed........................................ 90
Table 4.3  Mean (± 1SD) vertical ground reaction forces expressed as a percentage of body weight and the percentage change in the forces from treadmill walking at matched speed................................................................. 92

Table 4.4  Metabolic data (mean ± 1SD) at matched overground and treadmill walking speeds and the percentage change from treadmill walking at matched speed for healthy and stroke subjects................................................. 94
LIST OF FIGURES

CHAPTER 2

Figure 2.1 Mean angular profiles of the hip, knee and ankle for overground (thick solid line) and treadmill (thick dotted line) walking at matched speed. Positive is flexion, dorsiflexion. ............................................................... 28

Figure 2.2 Mean angular profiles of the hip, knee and ankle for treadmill walking at matched speed (thick solid line), 110% (grey solid line) and 120% (grey dotted line) of the matched speed. Positive is flexion, dorsiflexion. ....... 29

Figure 2.3 Mean vertical ground reaction force profiles during overground (thick solid line) and treadmill (thick dotted line) walking normalized to mean body weight. .......................................................... 31

Figure 2.4 Mean vertical ground reaction force profiles during treadmill (thick solid line) walking at matched speed, 110% (grey solid line) and 120% of the matched speed (grey dotted line) normalized to mean body weight. ...... 31

CHAPTER 3

Figure 3.1 Mean angular profiles of the hip, knee and ankle for overground (thick solid line) and treadmill (thick dotted line) walking at matched speed for the unaffected and the affected sides. ...................................................... 58

Figure 3.2 Mean angular profiles of the hip, knee and ankle for treadmill walking at matched speed (thick solid line), 110% (grey solid line) and 120% (grey dotted line) of the matched speed for the unaffected and the affected sides. ....................................................................................... 59

Figure 3.3 Mean vertical ground reaction forces during overground (thick solid line) and treadmill walking at matched speed (thick dotted line) for the unaffected and the affected sides. ......................................................... 62

Figure 3.4 Mean vertical ground reaction forces during treadmill walking at matched speed (thick solid line), 110% (thick grey line) and 120% (grey dotted line) of the matched speed for the unaffected and the affected sides. ...... 63
CHAPTER 1

Introduction

Treadmills are increasingly being used by clinicians and researchers to evaluate and retrain gait in people with gait abnormalities. In contrast to traditional gait laboratories, treadmills occupy relatively little space, allow control of walking speed, and facilitate the use of a safety harness and hand rails to provide security and stability as needed. Some models of treadmills are available that are instrumented with force sensors to provide kinetic information, thus increasing the capacity to evaluate gait performance.

Biomechanical gait analysis using a link segment model and an inverse dynamics approach has given important insight into the instantaneous kinematics and kinetics of gait. The word segment refers to rigid limb and body segments, where each segment represents a component of the body (for example, foot, shank, thigh, thorax, arms and head) and is joined to its adjacent segment(s) by hinges at which movement occurs. Inverse dynamics is the process by which the joint reaction forces and muscle moments are calculated using the kinematic and anthropometric data of the linked segments, and knowledge about the forces applied to the foot from the supporting surface (Winter, 2005). Link segment modeling requires estimates of segment mass, center of mass and moment of inertia which can be obtained from anthropometric tables and are based on the subject’s height, weight and sex. The conventional use of inverse dynamics involves iterative solution of the body segments equations of motion, that starts with measured ground reaction forces from force platforms and, beginning with those segments in contact with the ground, calculates joint forces and moments at each successive segment (Winter, 1991, 2005). In the present studies, which compared overground and treadmill
walking, kinetic analyses were limited to that which could be determined from vertical ground reaction forces that were available from the treadmill.

The link segment model and the analytical approach used for these studies carried out in the Motor Performance Laboratory were based on the following assumptions (see Winter, 1991, 2005):

- Each segment has a fixed mass represented as a point mass at its center of mass.
- The location of each segment’s center of mass remains fixed during movement.
- Joints are considered to be hinge or ball and socket joints.
- The mass moment of inertia of each segment about its center of mass is constant during the movement.
- Length of each segment remains constant during the movement.

Analysis of normal and pathological gait patterns has offered ample evidence of the body’s ability to vary its means of achieving the primary tasks involved in walking which have been summarized as (Winter et al., 1990, 1991, 2005):

1. Maintenance of support of the head, arms and trunk: that is, preventing collapse of the lower limb,
2. Maintenance of upright posture and balance of the body,
3. Control of the foot trajectory to achieve safe ground clearance and a gentle heel or toe landing,
4. Generation of mechanical energy to maintain the present forward velocity or to increase the forward velocity, and
5. Absorption of mechanical energy for shock absorption and stability or to decrease the forward velocity of the body.
The body is able to vary its means of accomplishing these tasks because of the redundancy of the motor system and its ability to use the redundancies in flexible ways. The normal person must adjust the walking pattern in response to, for example, unusual surfaces, pain, or a need to increase or decrease speed. Although some changes, such as those occurring with increased speed in normal subjects have been reported extensively, others have received little attention. It is important to compare higher treadmill walking speeds to self-selected walking speed as gait training in people with pathological gait often challenges them to walk faster. For people with disabilities, there are few reports even with simple changes, such as variations in speed.

It is also clear that subjects may use different strategies for accomplishing the tasks of walking, seemingly prioritizing factors that may be particularly important to their circumstances. For example, it appears that people concerned with stability, such as the elderly, will pay higher energy costs in favour of maintaining stability (Winter, 1991). It is clear that people can use different means of generating the work of walking, that is used to change their speed, and this may affect energy costs. For example, Kuo has persuasively shown that it is much more efficient to use ankle plantarflexors rather than hip flexors and/or extensors to perform the work of walking (Kuo, 2002). It is likely that strategies chosen may have an effect on the resulting metabolic costs, but these important measures are rarely included in mechanical analyses. Gait changes that could limit a person’s ability to adapt to specific situations are also of considerable interest, especially in clinical situations (Potter et al., 1995). In summary, despite knowing about redundancy and the adaptability of the human motor system, a great deal more information about
specific strategies adopted in response to different gait challenges such as walking surface and speed is needed, and these are the focuses of this thesis.

There is an underlying assumption that walking on a treadmill is similar to walking overground. It is important to ensure the validity of this assumption and demonstrate that the treadmill environment is as close to the overground environment as possible (Alton et al., 1998). In one of the early studies, Van Ingen Schenau (1980) used a theoretical mathematical model incorporating kinetic and potential energy calculations to compare treadmill and overground locomotion. Based on the results he hypothesized that the mechanics are the same provided that the motor of the treadmill is strong enough to produce a constant belt speed. He further suggested that the differences between treadmill and overground locomotion could exist if there were differences in mechanical compliance between the overground and treadmill walking surfaces or the surrounding visual environment.

Despite Van Ingen Schenau’s theoretical viewpoint, studies comparing overground and treadmill walking have shown a number of differences in temporal distance parameters of gait, lower limb joint excursions (Alton et al., 1998; Dingwell et al., 2001; Murray et al., 1985; Stolze et al., 1997; Strathy et al., 1982; Warabi et al., 2005), vertical ground reaction forces (White et al., 1998; Yack et al., 1995) and physiological measures of walking performance (Greig et al., 1993; Murray et al., 1985; Pearce et al., 1983).

The most common differences that have been reported are higher cadence, decreased stance time (and increased swing time), decreased double support time, shorter stride/step length and wider step width associated with treadmill walking compared to
overground walking at comparable speeds (Alton et al., 1998; Stolze et al., 1997; Warabi et al., 2005). In terms of kinematics, greater hip flexion in early stance and less hip extension in late stance was observed during treadmill walking (Alton et al., 1998). Of the few studies that have compared the kinetics of overground and treadmill walking White et al. (1998) and Yack et al. (1995) reported similar magnitudes of vertical ground reaction forces in early stance but higher forces in midstance and lower forces in late stance with treadmill walking. Overall the findings from healthy subjects suggest that the demands of walking may differ as a function of the surface. It is not known whether the same applies to people with hemiparesis secondary to stroke (Hesse, 1999).

One way to determine the demands of a physical task is to measure the metabolic cost. There are few studies that have compared the metabolic cost of walking on the two surfaces and the results are equivocal. At comparable self-selected speeds of walking Waters et al. (1988, 1999) reported similar energy costs, Murray et al. (1985) reported higher energy costs associated treadmill walking and Pearce et al. (1983) reported higher energy costs for overground walking. There were no clear explanations of the conflicting results of energy costs that have been reported in the aforementioned studies. Furthermore, none of these studies examined the gait kinematics and kinetics so it cannot be ascertained that the overground and treadmill walking patterns were comparable.

The literature suggests that there are differences between normal overground and treadmill walking. However, these studies are largely limited to very select outcomes in young, healthy individuals and it is important to explore in a more comprehensive manner the extent and nature of these differences. If motor and energy demands differ, this may have significant impact on skill transference from gait re-training on a treadmill
to overground walking. An integrated approach that simultaneously measures the
kinematics, kinetics and metabolic cost of walking is needed to evaluate and compare
walking on level ground and a treadmill. Clinically this knowledge is important to
understanding the relative demands in populations that may be exposed to gait re-
training.

1.1 Objectives

The overall purpose of this study was to characterize the biomechanical profiles
and metabolic costs of overground and treadmill walking in healthy older adults and in
adults who have had a stroke. The primary objectives are as follows:

1) To compare the biomechanical (kinematic and kinetic profiles) and metabolic
characteristics of overground and treadmill walking in healthy adults;
2) To compare the biomechanical (kinematic and kinetic profiles) and metabolic
characteristics of overground and treadmill walking in people who have had a
stroke;
3) To compare the biomechanical (kinematic and kinetic profiles) and metabolic
characteristics of overground and treadmill walking between healthy adults and
those who have experienced a stroke, and secondarily;
4) To explore, describe and compare the biomechanical (kinematic and kinetic) and
metabolic means by which healthy adults and those who have experienced a
stroke increase their speeds of walking on a treadmill.

The objectives are addressed in a series of three papers. The first paper focuses on
the gait of healthy older adults, the second on the gait of stroke survivors with an
emphasis on explaining the impact of walking surface and interlimb symmetry, and the
final paper compares the two groups of subjects and their means of increasing their speeds of walking on a treadmill. To facilitate interpretation of findings a general discussion is included.

REFERENCES


CHAPTER 2

Biomechanics and Metabolic Costs of Overground and Treadmill Walking in Healthy Adults¹

Abstract

**Background:** Although treadmill and overground walking appear to be biomechanically similar in young healthy subjects, it is not known whether this can be generalized to older subjects or if the energy demands are correspondingly comparable. **Methods:** Ten healthy adults between 50 and 73 years of age walked at their self-selected speed overground and each persons speed was matched on the treadmill. Temporal parameters, angular kinematics and vertical ground reaction forces were recorded during walking once subjects were in steady state, as determined from their heart rate and oxygen uptake. **Results:** Step, stride, and joint angular kinematics were similar for both modes of walking with the exception of the maximum hip flexion and knee extension which were both more pronounced with treadmill than overground walking, respectively, but in both instances differed by less than 3°. Vertical ground reaction force profiles were similar although the peak associated with push-off was 7.5% smaller during treadmill walking. The metabolic requirements of treadmill walking, however, were about 23% higher than that associated with overground walking. **Conclusions:** While treadmill and overground walking is biomechanically similar, the energy cost of treadmill walking is higher. This may be important to consider when a treadmill is used for gait retraining in patient populations.

2.1 Introduction

Similar to overground gait analysis, many systems have been developed to measure temporal-distance, kinematic, and kinetic parameters of normal and abnormal gait using instrumented treadmills (Alton et al., 1998; Strathy et al., 1983; Warabi et al., 2005). Treadmills offer many advantages over traditional overground walking for the analysis and training of gait as space requirements are minimal, environmental factors can be controlled, steady state locomotion speeds are selectable, and successive repetitive strides can be documented expeditiously (White et al., 1998). It is perhaps for these reasons that treadmills are increasingly being used for investigative and clinical gait evaluation as well as gait training. It is, however, important to fully understand if there are differences in the characteristics of walking on a treadmill and overground as this has implications for interpretation of the assessment and transference of training. Given the lack of consensus as to whether the differences between treadmill and overground walking are significant and the limited information on population other than young adults, this study characterizes the biomechanical profiles and metabolic costs of overground and treadmill walking in healthy older adults.

2.1.1 Temporal and Distance Parameters

Several authors have reported higher cadence and shorter stance times (and longer swing times) when walking on a treadmill compared to overground walking at comparable speeds (Alton et al., 1998; Stolze et al., 1997; Warabi et al., 2005). Alton et al. (1998) reported a significantly higher mean cadence of 122 ± 4 steps/min on a treadmill versus 117 ± 6 steps/min during overground walking for 17 young, healthy subjects. Similarly, cadences of 6.0% to 6.6% higher were associated with treadmill
walking in young subjects (Stolze et al., 1997; Warabi et al., 2005). Though Murray et al. (1985) found a trend of higher cadence, shorter swing phase, longer double support and shorter step length on the treadmill, no significant differences were reported in their sample of 7 young adults (range 20-36 years). Interestingly only Greig et al. (1993) reported a lower cadence and a 3% increase in stride length during treadmill walking, compared to overground walking in both young (21 - 37 years, n=12) and older (71 - 80 years, n=12) adults.

Reductions in stance phase on the treadmill of 4.7% (Alton et al., 1998) and 7% (Stolze et al., 1997; Warabi et al., 2005) combined with an increase in swing phase and a shortened double limb support time were reported for treadmill walking (81.1 ± 11.2 ms) in comparison to overground walking (111.7 ± 17.4 ms) (Stolze et al., 1997). Greater step width during treadmill walking (104.3 ± 17.6 mm) compared to overground (81.2 ± 19.6 mm) was also reported, which may have allowed subjects to maintain a more stable posture during walking on an unfamiliar surface (Stolze et al., 1997). Furthermore fear of the moving belt may translate into a sense of urgency to place the swing limb onto the treadmill as the stance limb is pulled backwards (Alton et al., 1998).

2.1.2 Kinematics

Few studies have compared the kinematics of walking on the treadmill in relation to overground walking. Dingwell et al. (2001) reported that the stride to stride variability of joint angular displacements was higher for overground than for treadmill walking in 10 young healthy subjects. This was the case for hip, knee and ankle joint excursions, noting that there was higher variability at more distal joints (Dingwell et al., 2001).
Alton et al. (1998) reported significantly greater hip range of motion (27°) due to hip flexion (32°) during treadmill walking when compared to overground walking (24° and 28°, respectively) at similar speeds. At self-selected walking speed, Alton et al. (1998) and Murray et al. (1985) reported hip extension angles in the range of 4° and -8° during overground walking and 2° and -5° during treadmill walking, respectively (positive is flexion, negative is extension and neutral is zero). Murray et al. (1985) examined 7 healthy subjects walking at slow, self-selected and fast speeds and noted a lower maximum hip extension angle during the stance phase during treadmill walking compared to overground walking. They hypothesized that subjects were flexing their hips more and using less extension as a protective response to avoid falling off the back of the treadmill and to keep up with the belt speed.

Alton et al. (1998) like Murray et al. (1985) reported similar maximum knee extension and knee flexion angles for overground walking (10° and 70°) and (positive = flexion, negative = hyperextension and zero = neutral) for treadmill walking (12° and 71°) as well as knee range of motion that was within one degree (57°-58°). Strathy et al. (1983) compared the knee kinematics of overground and treadmill walking at similar speeds in 10 healthy subjects (21-40 years) and reported significantly less knee extension just prior to foot contact during treadmill walking. The decrease in knee range of motion was attributed to subjects taking shorter steps during treadmill walking.

Murray et al. (1985) reported maximum ankle dorsiflexion angles of 11°, 9° and 7° for overground walking at slow, self-selected and fast speeds and 8°, 6° and 5° for treadmill walking at similar speeds. The ankle range of motion and maximum dorsiflexion were significantly smaller during treadmill walking at slow and self-selected
speeds when compared with overground walking but were similar at fast speeds of walking. Alton et al. (1998) reported similar maximum ankle dorsiflexion and plantarflexion angles of $24^\circ$ and $8^\circ$ for overground walking and $22^\circ$ and $9^\circ$ for treadmill walking.

2.1.3 Kinetics

Little is known about the kinetics of treadmill walking due to the challenges of measuring three dimensional (3D) forces beneath a moving belt. Most commercially available instrumented treadmills provide only the vertical ground reaction force. The characteristic ‘butterfly’ profile of the vertical ground reaction force showing an initial peak greater than body weight after foot contact (F1), a valley (less than body weight) during midstance (F2), and a second peak at push-off (F3) is prominent in treadmill walking although some differences in magnitude compared to overground have been reported (White et al., 1998; Yack et al., 1995).

Yack et al. (1995) tested 24 subjects (23-42 years) at comparable walking speeds overground and on the treadmill and reported that the F2 force was about 9.7% higher on the treadmill. Similar findings were described by White et al. (1998) who also controlled for cadence and stride length in addition to speed. The higher midstance (F2) force may reflect the constant velocity associated with the treadmill, such that the downward vertical acceleration of the centre of gravity is less than that of overground walking.

Lower F3 forces (~ 5%) were associated with self-selected speeds of treadmill walking compared to overground (White et al., 1998; Yack et al., 1995). White et al. (1998) theorized that during late stance the propulsive shear forces that increase the belt speed during push-off result in some transfer of energy from the subject to the belt thus
reducing the F3 force (due to increased belt speed at push-off). Alternatively, the lower F3 force may be secondary to the reduction in hip extension and ankle plantarflexion limiting push-off.

2.1.4 Metabolic Costs of Walking in Healthy Adults

Metabolic energy is required to generate muscle force to support the body’s weight and to propel and swing the legs during walking (Gottschall and Kram, 2002). In normal, healthy subjects the metabolic energy cost of walking is largely dependent on speed. The speed-energy relationship is parabolic, with the minimum cost at optimal walking speed, and increasing as walking speed increases or decreases from the optimal speed that was reported to be between 1-1.3m/s (Bernardi et al., 1999; Holt et al., 1991; Martin et al., 1992; Pearce et al., 1983; Waters et al., 1988).

Heart rate, an indicator of energy cost (Greig et al., 1993; Murray et al., 1985; Pearce et al., 1983), is reportedly 6 beats/min higher in young and older adults when walking on a treadmill compared to overground at speeds of 1.4 m/s. This has been attributed to anxiety associated with walking on the treadmill (Murray et al., 1985), as well as a true increase in the metabolic cost of walking as a result of the added cost of gripping the treadmill hand rails (Greig et al., 1993). In contrast, Pearce et al. (1983) reported significantly higher heart rates with overground walking (93 ± 13 beats/min) than treadmill walking (88 ± 14 beats/min) at the same speed (1.3 m/s) in 42 healthy subjects aged 19 – 66 years.

At comfortable steady state overground walking at speeds between 1.22-1.33 m/s, the rate of oxygen consumption in healthy adults (n = 73) was reported to be about 12 ml/min/kg (Waters et al., 1988). The average oxygen consumption was similar for
treadmill and overground walking at slow and self-selected walking speeds (Murray et al., 1985; Ralston, 1960). Pearce et al. (1983), however, reported higher rates of oxygen consumption for overground walking (11 ± 2 ml/min/kg) versus treadmill (10 ± 2 ml/min/kg) at comparable walking speeds in healthy adults. The explanation for these conflicting findings of oxygen consumption is not clear. In the absence of biomechanical information it cannot be determined whether or not the movement patterns differed between the two walking conditions.

2.2 Purpose

The purpose of this study was to compare the biomechanics and metabolic cost of overground and treadmill walking in healthy adults over 50 years of age. The specific objectives are as follows:

- To compare temporal distance parameters and lower limb kinematics between walking overground and on the treadmill;
- To compare the vertical ground reaction forces between walking overground and on the treadmill;
- To compare the energy cost of overground and treadmill walking;
- To explore alterations in the biomechanics and metabolic costs with increases in speed of treadmill walking.

2.3 Methodology

Prior to recruitment of subjects, ethics approval for the study was obtained from the Faculty of Health Sciences Research Ethics Board of Queen’s University and Affiliated Teaching Hospitals, Kingston, Ontario. A sample of convenience that included healthy adults over 50 years of age were recruited in the following ways:
1) newspaper advertisements,
2) advertisements posted at various locations in the city, and
3) letters sent to the members of the local Seniors Centre, describing the study purpose and protocol.

Subjects who were interested in participating in the study contacted the principal investigator and were screened by telephone to ensure they met all inclusion criteria. Interested subjects were included if they were over 50 years of age, were able to walk independently without any walking aid and reported themselves to be generally in good health. Subjects were excluded if they reported any cardiovascular or cardiopulmonary problem, any orthopaedic condition, uncontrolled metabolic diseases or other pre-existing condition limiting mobility. Those eligible to participate were invited to the Motor Performance Laboratory in the School of Rehabilitation Therapy at Queen’s University, Kingston, Ontario for final screening and testing.

2.3.1 Procedure

The protocol was conducted on a single day and required between three and four hours to complete. A minimum of three researchers were present during all testing to assist with data acquisition and monitoring of the subject.

Upon arrival in the Motor Performance Laboratory, subjects read the consent form, had their questions answered and provided signed consent (Appendix A). The subjects’ age, height, weight, past medical history and current medications were recorded.

2.3.2 Instrumentation and Data Acquisition

The kinematic data were collected using two Optotrak cameras (Northern Digital Inc., Waterloo, ON, Canada) placed on either side of the walkway or the treadmill.
Kinetic data were collected from two AMTI force platforms (Advanced Mechanical Technologies Inc., Newton, MA, USA) embedded in an 8 m walkway for the overground condition and two Kistler force plates (Kistler, Winterthur, Switzerland) positioned in front of each other beneath the belt of a Gaitway treadmill (H/P Cosmos, Germany). Metabolic data were collected using a CosMed K4b2 portable integrated system (Cosmed, Chicago, IL, USA) that performed breath by breath gas analysis.

Calibration of the Optotrak cameras and all force platforms was performed before data acquisition. A cube embedded with 16 infrared emitting diodes (IREDS) with a fixed spatial orientation (i.e., known relative co-ordinates) was placed at the corner of the first force platform in the direction of walking and 10 seconds of data were collected. The second force plate was in a fixed position relative to the first force platform and its coordinate position was automatically assigned. These data defined the co-ordinate system and force platform locations.

The metabolic unit was calibrated using a standard mixture of known oxygen (O2) and carbon dioxide (CO2) gas concentrations to ensure accurate sensor operation. Flow sensor calibration was also performed using a 3000 ml syringe.

Subjects were instrumented with eight IRED clusters made of moulded plastic with slots for four IREDS per cluster. Clusters were secured over the mid-foot, mid-shank, and mid-thigh bilaterally using stretchable Velcro straps with additional clusters placed on the sacrum and the seventh cervical vertebra. To define movement planes, axes, body and limb segment lengths specific landmarks were identified using a probe embedded with 6 IREDS, the relative co-ordinates of which were known as was their relationship to the tip. The tip of the probe identified the first and fifth metatarsal heads,
medial and lateral malleoli, medial and lateral epicondyles, greater trochanter, anterior superior iliac spine (ASIS), and the acromion process bilaterally for each subject. The subject remained in a comfortable standing position while the landmarking was completed. This process was repeated prior to treadmill walking. The landmarking was necessary to define the anatomically relevant locations in a segment coordinate system and to build a biomechanical model composed of linked segments (corresponding to a set of bones) that are based on the anatomical landmarks.

A heart rate monitor was strapped snugly to each subject’s chest and the heart frequency receiver was plugged into the control panel of the portable unit. The portable unit of the metabolic apparatus was supported on the torso with a harness provided by the manufacturer. A seal moulded of a polymer gel product was affixed to a face mask equipped with a turbine to provide an intimate seal over the mouth and nose and subjects were asked to breathe normally. The turbine allowed room air to enter and exhaled air to be transmitted to the gas exchange analyzer in the portable metabolic testing unit. The analyzers are temperature and humidity controlled. This system avoids the discomfort of a mouthpiece and is well tolerated. Finally, subjects were asked to wear a body weight support harness, which for treadmill walking only was secured to an overhead support system for safety.

2.3.3 Overground Walking

Once subjects were instrumented and familiar with the environment and the equipment, they were asked to sit quietly for 5 to 10 minutes while their resting metabolic rate was recorded. They were then asked to walk overground to become accustomed to the equipment they were wearing. The subjects walked back and forth on the overground
walkway for several minutes at their self-selected speed. When the heart rate stabilized to within a few beats per minute and minute oxygen consumption (VO₂) varied by less than 10%, indicating steady state, a series of walking trials were recorded every 10 seconds. The subjects did not have to wait at the end of the walkway but had to turn around to walk back and forth on the walkway. Three successful trials were required and were defined as those in which all IRED clusters were in view of the camera and each foot landed completely on a force platform. Subjects were not instructed to step on the plates or how they should walk other than to walk at a comfortable pace. In this way undesired adjustments of their gait pattern were avoided. Heart rate and breath by breath metabolic data were obtained throughout the test period. After completion of the overground trials, the subject was asked to sit quietly until the heart rate returned to resting values. None reported feeling fatigued.

The average walking speed over the three trials was calculated from the coordinate data associated with the pelvic marker. This speed was used as the criterion for matching the treadmill speed.

2.3.4 Treadmill Walking

Before mounting the treadmill, the subjects were again asked to sit quietly for 5 to 10 minutes while their resting metabolic rate was recorded. Then they were asked to stand on the treadmill and the safety harness was attached to an overhead support frame since the handrails had been removed. In the case of a slip or a fall the subjects’ weight would be supported by the frame to prevent injury, but no weight was supported during walking.
The treadmill was started and the subject walked while the speed was progressively increased to match that achieved during the overground trials. When subjects felt comfortable and steady state was achieved (about 2 to 4 minutes), subjects walked for an additional two minutes at that speed; Optotrak and vertical ground reaction force data were collected over the last 30 seconds of the two minute period. The speed was then increased by 10% (of the matched speed) and one minute was allowed for the subjects to accommodate. Subjects walked for an additional two minutes and data were again collected over the last 30 seconds. A final speed increase equivalent to 20% of the matched speed was introduced and data acquired as previously described. As with overground walking, three successful trials in which each foot landed on the front force plate only were required. Heart rate and metabolic data were collected continuously. The speed was then gradually reduced to zero and subjects were asked to sit quietly until their heart rate returned to baseline resting values.

2.3.5 Data Processing

The force platforms in the walkway operated in a voltage range of +/- 10Volts and measured the ground reaction forces in X, Y and Z axes. The overground and treadmill force platform vertical ground reaction force (VGRF) signals were digitized at a frequency of 200Hz and stored for off-line analysis on a laboratory computer. The Optotrak data were collected at 100Hz.

The co-ordinate data from the IRED clusters and the probed landmarks were merged into the motion analysis software environment (C-Motion Inc., Rockville, MD, USA) to define a 3-D, eight-segment model consisting of the feet, shanks, thighs, pelvis and trunk. The limb segments were defined from the tracking markers on the clusters and
the landmarks as follows: the foot = the mid-foot cluster, the first and fifth metatarsal heads and the medial and lateral malleoli; the shank = the mid-shank cluster, medial and lateral malleoli and the medial and lateral epicondyles; the thigh = the mid-thigh cluster, medial and lateral epicondyles, greater trochanter and the anterior superior iliac spine; the pelvis = the pelvic cluster, greater trochanter and the anterior superior iliac spine which is a landmark on the iliac crest; the trunk = thoracic cluster and the acromion processes.

The Optotrak and force plate data that were collected simultaneously were synchronized using C-Motion (C-Motion, Inc., Rockville, MD) motion analysis software that was used for processing the gait data. The events from the gait cycle, including the right and left foot contact and toe off, were identified based on the vertical ground reaction forces for each successful gait trial. The time corresponding to the first and last samples in which the heel and toe were visually observed for initial contact and terminal stance and vertical force measured at least 1N (after initial frames were zeroed) were defined as initial contact and terminal stance, respectively. A link segment model described above was used to process the kinematic data. Data were filtered using a dual pass Butterworth filter with a cut-off frequency of 6Hz. All data were normalized to 100% of the stance phase and processed gait trials were averaged. Temporal-distance parameters, joint angular displacements and the vertical ground reaction force profiles (normalized to the subject’s mass while fully instrumented) were determined for each subject and ensemble averaged to provide group data.

The minute oxygen consumption (VO₂) and heart rate (HR) were determined from the final 30 second rest period preceding the start of the walking trials and for a 30 second period during steady state walking (after 1.5 minutes from the start of data
collection for overground trials and last 30 seconds for treadmill trials). Oxygen consumption was normalized to the subject’s body mass when fully instrumented.

All the data were later transferred to Microsoft Excel and SPSS for plotting and further analysis, respectively.

2.3.5.1 Temporal and Distance Parameters

Self-selected overground walking speed (m/s) was determined from the distance traversed divided by time taken, whereas the treadmill walking speed was the speed of the belt.

Stance time (s) was defined as the period from initial foot contact to toe off of the same foot and swing time (s) was the period between toe off and initial foot contact of the same foot. Cadence was expressed as the number of steps taken per minute and was calculated from walking speed and step time (s). Stride length (m) was the distance traversed from foot contact to subsequent foot contact of the same foot. Step length (m) was the distance traversed from foot contact of one foot to foot contact of the opposite foot. Stride width (m) was determined as the medio-lateral linear perpendicular distance between the two feet during double support. Double support time (s) was the time during one gait cycle when both feet were in contact with the ground.

2.3.5.2 Kinematic and Kinetic Data

Angular displacements about the hip, knee and the ankle joints were calculated for both overground and treadmill walking. Only the vertical ground reaction force was considered since other forces were not available during treadmill walking.
2.3.5.3 Metabolic Data

Heart rate (HR, bpm) and oxygen consumption (VO₂, ml/min) that was normalized to body weight were recorded breath-by-breath. VO₂ was calculated as follows:

\[
VO₂ = \frac{[(FiO₂*VI)-(FeO₂*VE)]}{\text{body weight in kg (Cosmed, Chicago, IL, USA)}}
\]

where:

- \(FiO₂\) = fractional concentration of inspired \(O₂\)
- \(VI\) = the minute inhaled volume of air (ml/min)
- \(FeO₂\) = fractional concentration of expired \(O₂\)
- \(VE\) = the minute exhaled volume of air (ml/min)

2.4 Data Analysis

Statistical analyses were performed using SPSS (Version 14, SPSS Inc.) and a significance level of \(p<0.05\) was set. A paired t-test was performed for all dependent kinematic and kinetic measures to determine if there were any asymmetries between the right and left legs. Since there were no differences \((p>0.05)\) only the data from the right side were analyzed further.

Descriptive statistics (mean ± SD) were calculated for all the outcome measures. Repeated measures analysis of variance (ANOVA) was used to identify main effects of speed and walking surface as well as interactions for all dependent measures.

2.5 Results

Ten subjects (5 males and 5 females) with a mean age of 60.6 ± 7.4 years (range 50-73 years), mean mass of 78 ± 11.8 kg and mean height of 1.74 ± 0.07 m completed the protocol without incident.
2.5.1 Gait Speed and Cadence

The individual and descriptive data for gait speed and cadence during overground and treadmill walking are listed in Table 2.1.

The average self-selected overground walking speed and matched speed on the treadmill were identical at 1.15 m/s. At comparable speeds, the cadence during overground (109 ± 14 steps/min) and treadmill walking (110 ± 20 steps/min) was not different.

The speeds corresponding to 110% and 120% of the matched treadmill speed were 1.27 m/s (±0.16) and 1.38 m/s (±0.17), respectively. As expected they were significantly faster than the self-selected speed of walking (p < 0.001). Though there was a tendency for cadence to increase at faster walking speeds, it was only significantly different from self-selected walking when subjects walked 20% faster (p<0.001).

Table 2.1 Gait speed (m/s) and cadence (steps/min) for individual subjects

<table>
<thead>
<tr>
<th>Subject Number</th>
<th>Overground (Self-selected Speed)</th>
<th>Treadmill (Matched Speed)</th>
<th>Treadmill (+10%)</th>
<th>Treadmill (+20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Speed</td>
<td>Cadence</td>
<td>Speed</td>
<td>Cadence</td>
</tr>
<tr>
<td>C1</td>
<td>1.29</td>
<td>142</td>
<td>1.26</td>
<td>151</td>
</tr>
<tr>
<td>C2</td>
<td>1.23</td>
<td>107</td>
<td>1.22</td>
<td>101</td>
</tr>
<tr>
<td>C3</td>
<td>1.04</td>
<td>99</td>
<td>1.03</td>
<td>100</td>
</tr>
<tr>
<td>C4</td>
<td>1.37</td>
<td>111</td>
<td>1.25</td>
<td>106</td>
</tr>
<tr>
<td>C5</td>
<td>0.92</td>
<td>103</td>
<td>0.98</td>
<td>125</td>
</tr>
<tr>
<td>C6</td>
<td>0.96</td>
<td>96</td>
<td>0.96</td>
<td>91</td>
</tr>
<tr>
<td>C7</td>
<td>1.35</td>
<td>122</td>
<td>1.39</td>
<td>122</td>
</tr>
<tr>
<td>C8</td>
<td>1.14</td>
<td>109</td>
<td>1.13</td>
<td>96</td>
</tr>
<tr>
<td>C9</td>
<td>1.27</td>
<td>113</td>
<td>1.26</td>
<td>124</td>
</tr>
<tr>
<td>C10</td>
<td>1.04</td>
<td>93</td>
<td>1.04</td>
<td>84</td>
</tr>
<tr>
<td>Average</td>
<td>1.15</td>
<td>109</td>
<td>1.15</td>
<td>110</td>
</tr>
</tbody>
</table>

* Indicates significantly different from overground condition (p<0.05)
2.5.2 Temporal and Distance Parameters

The individual and descriptive data for the temporal distance parameters during overground and treadmill walking are listed in Table 2.2.

At comparable speeds, the percentage of the gait cycle spent in double support was greater during overground walking than treadmill walking (p=0.002). This was accompanied by greater stride width (p<0.050). All other temporal distance measures were similar for both walking conditions (p>0.050).

At faster walking speeds there was a tendency for gait cycle time and % stance to be lower, and % swing, stride and step length to be higher but they were significantly different from self-selected walking only when subjects walked 20% faster (p<0.001). With increased speed, the % of time in double support was less (p<0.001 at 110% and at 120%) than that associated with overground walking. The stride width was greater at 110% of the matched speed (p=0.03) than that observed during comfortable walking but the difference was small and was not maintained at 120% of the matched speed.
Table 2.2 Mean values (± 1SD) of temporal distance parameters for overground and treadmill walking

<table>
<thead>
<tr>
<th>Variable</th>
<th>Overground</th>
<th>Treadmill</th>
<th>Treadmill (+10%)</th>
<th>Treadmill (+20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cycle Time (sec)</td>
<td>1.13 ± 0.14</td>
<td>1.12 ± 0.19</td>
<td>1.08 ± 0.15</td>
<td>1.06 ± 0.13</td>
</tr>
<tr>
<td>Stance Time (sec)</td>
<td>0.72 ± 0.10</td>
<td>0.70 ± 0.13</td>
<td>0.68 ± 0.11</td>
<td>0.65 ± 0.09</td>
</tr>
<tr>
<td>Swing Time (sec)</td>
<td>0.40 ± 0.05</td>
<td>0.42 ± 0.07</td>
<td>0.40 ± 0.05</td>
<td>0.41 ± 0.05</td>
</tr>
<tr>
<td>Double Support Time (sec)</td>
<td>0.33 ± 0.06</td>
<td>0.28 ± 0.08</td>
<td>0.26 ± 0.06</td>
<td>0.23 ± 0.05</td>
</tr>
<tr>
<td>Stance (% of gait cycle)</td>
<td>64.39 ± 1.81</td>
<td>62.89 ± 1.49</td>
<td>63.07 ± 2.42</td>
<td>61.63 ± 1.76</td>
</tr>
<tr>
<td>Swing (% of gait cycle)</td>
<td>35.61 ± 1.81</td>
<td>37.11 ± 1.49</td>
<td>36.93 ± 2.42</td>
<td>38.37 ± 1.76</td>
</tr>
<tr>
<td>Double Support (% of gait cycle)</td>
<td>29.29 ± 3.23</td>
<td>26.03 ± 2.60</td>
<td>24.64 ± 3.51</td>
<td>22.75 ± 2.59</td>
</tr>
<tr>
<td>Stance/Swing Ratio</td>
<td>1.82 ± 0.14</td>
<td>1.70 ± 0.10</td>
<td>1.72 ± 0.19</td>
<td>1.61 ± 0.12</td>
</tr>
<tr>
<td>Stride Length (m)</td>
<td>1.27 ± 0.11</td>
<td>1.22 ± 0.15</td>
<td>1.32 ± 0.17</td>
<td>1.42 ± 0.16</td>
</tr>
<tr>
<td>Step Length (m)</td>
<td>0.63 ± 0.06</td>
<td>0.61 ± 0.07</td>
<td>0.67 ± 0.08</td>
<td>0.71 ± 0.09</td>
</tr>
<tr>
<td>Stride Width (m)</td>
<td>0.123 ± 0.03</td>
<td>0.093 ± 0.04</td>
<td>0.093 ± 0.03</td>
<td>0.095 ± 0.04</td>
</tr>
</tbody>
</table>

* Indicates significantly different from overground condition (p<0.05)

2.5.3 Kinematic Data

The descriptive data associated with the angular displacement of the hip, knee and ankle joints during stance phase for overground and treadmill walking are listed in Table 2.3 and the average angular displacement profiles for all subjects are shown in Figures 2.1 and 2.2.

At comparable speeds, there was no difference in the total hip, knee or ankle range of motion. The maximum knee extension angle differed, measuring 2° during
treadmill walking and 0° during overground walking (p = 0.049). In isolation this finding is not considered important.

As walking speed increased by 20%, the total range of motion at the hip and the ankle increased (p=0.021); this was likely attributable to greater hip flexion and ankle plantarflexion (p<0.031).

<table>
<thead>
<tr>
<th>Joint Angle</th>
<th>Overground</th>
<th>Treadmill (+10%)</th>
<th>Treadmill (+20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip Flexion</td>
<td>22 (±5)</td>
<td>25 (±6)</td>
<td>27 (±7)</td>
</tr>
<tr>
<td></td>
<td>-13 (±5)</td>
<td>-10 (±7)</td>
<td>-10 (±7)</td>
</tr>
<tr>
<td></td>
<td>34 (±4)</td>
<td>35 (±7)</td>
<td>37 (±7)</td>
</tr>
<tr>
<td></td>
<td>46 (±3)</td>
<td>48 (±8)</td>
<td>47 (±10)</td>
</tr>
<tr>
<td>Extension</td>
<td>0 (±1)</td>
<td>2* (±2)</td>
<td>2 (±2)</td>
</tr>
<tr>
<td></td>
<td>46 (±3)</td>
<td>46 (±7)</td>
<td>46 (±9)</td>
</tr>
<tr>
<td>Excursion</td>
<td>13 (±4)</td>
<td>13 (±3)</td>
<td>13 (±4)</td>
</tr>
<tr>
<td>Dorsiflexion</td>
<td>-10 (±4)</td>
<td>-13 (±8)</td>
<td>-14 (±8)</td>
</tr>
<tr>
<td>Plantarflexion</td>
<td>23 (±3)</td>
<td>26 (±8)</td>
<td>27 (±8)</td>
</tr>
</tbody>
</table>

* Indicates significantly different from overground condition (p<0.05)

From the average angular displacement profiles (Figure 2.1) for the hip, knee and ankle, it appears that there is greater hip and knee flexion and greater ankle joint excursion during treadmill walking at matched speed, though statistical significance was not generally found.
Figure 2.1  Mean angular profiles of the hip, knee and ankle for overground (thick solid line) and treadmill (thick dotted line) walking at matched speed. Positive is flexion, dorsiflexion. Corresponding thin lines represent ± 1 standard deviation.
Figure 2.2  Mean angular profiles of the hip, knee and ankle for treadmill walking at matched speed (thick solid line), 110% (grey solid line) and 120% (grey dotted line) of the matched speed. Positive is flexion, dorsiflexion.
2.5.4 Kinetic Data

The two vertical ground reaction force maxima (F1 and F3) and the minimum force values (F2) are presented in Table 2.4 for all conditions. The peak vertical ground reaction force values occurring after foot contact (F1) and the characteristic unloading of the force plate during midstance (F2) were not different between overground and treadmill conditions at similar speeds. In contrast, the second peak associated with push-off (F3) was about 7.5% smaller when subjects walked on the treadmill compared to overground (p=0.003). Figures 2.3 and 2.4 illustrate the average vertical ground reaction force profiles.

As the treadmill speed increased, the magnitude of F2 decreased (p<0.05). The push-off (F3) force increased and at a walking speed 20% faster than self-selected speed, the magnitude of F3 was comparable to that measured at the slower overground walking speed (Table 2.4 and Figure 2.3). Note that there is attenuation of the maxima and minima when averaging across subjects due to temporal misalignment of F1, F2 and F3 points of interest.

Table 2.4 Mean (± 1SD) vertical ground reaction forces expressed as a proportion of body weight

<table>
<thead>
<tr>
<th>Ground Reaction Force</th>
<th>Overground</th>
<th>Treadmill</th>
<th>Treadmill (+10%)</th>
<th>Treadmill (+20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>F1</td>
<td>1.09</td>
<td>1.06</td>
<td>1.08</td>
<td>1.12</td>
</tr>
<tr>
<td></td>
<td>(±0.06)</td>
<td>(±0.08)</td>
<td>(±0.09)</td>
<td>(±0.11)</td>
</tr>
<tr>
<td>F2</td>
<td>0.75</td>
<td>0.74</td>
<td>0.70*</td>
<td>0.66*</td>
</tr>
<tr>
<td></td>
<td>(±0.09)</td>
<td>(±0.13)</td>
<td>(±0.14)</td>
<td>(±0.13)</td>
</tr>
<tr>
<td>F3</td>
<td>1.06</td>
<td>0.98*</td>
<td>1.01*</td>
<td>1.02</td>
</tr>
<tr>
<td></td>
<td>(±0.05)</td>
<td>(±0.07)</td>
<td>(±0.06)</td>
<td>(±0.06)</td>
</tr>
</tbody>
</table>

* Indicates significantly different from overground walking (p<0.05)
Figure 2.3  Mean vertical ground reaction force profiles during overground (thick solid line) and treadmill (thick dotted line) walking normalized to mean body weight. Corresponding thin lines represent ± 1 standard deviations.

Figure 2.4  Mean vertical ground reaction force profiles during treadmill (thick solid line) walking at matched speed, 110% (grey solid line) and 120% of the matched speed (grey dotted line) normalized to mean body weight.
2.5.5 Metabolic Data

At matched speed the average HR and oxygen consumption (VO₂) were about 6% and 23% higher during treadmill walking than during overground walking, respectively, though only VO₂ differed significantly (Table 2.5; p = 0.001). HR and VO₂ increased over the matched speed in a near linear fashion with increased walking speed (p = 0.002).

HR returned to comparable values after overground and treadmill walking; however, VO₂ remained higher after treadmill walking (p=0.039). The average time required for HR recovery was significantly higher for treadmill walking, but note that the final speed was 20% higher than that achieved with overground walking.

Table 2.5 Metabolic data (mean ± 1SD) for overground and treadmill walking

<table>
<thead>
<tr>
<th></th>
<th>Overground</th>
<th>Treadmill</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Resting</td>
<td>Self-selected</td>
</tr>
<tr>
<td>Heart Rate (beats/min)</td>
<td>75 (±10)</td>
<td>85 (±10)</td>
</tr>
<tr>
<td>Oxygen Consumption (ml/kg/min)</td>
<td>4.03 (±0.53)</td>
<td>11.39 (±1.10)</td>
</tr>
<tr>
<td>Time to Recovery (min)</td>
<td>2.60 (±0.97)</td>
<td>4.10 (±1.10)</td>
</tr>
</tbody>
</table>

* Indicates significantly different from self-selected (p<0.05)
† Indicates significantly different from overground resting

2.6 Discussion

While the kinematic and kinetic profiles associated with walking on a treadmill and overground at comparable speeds showed only minor differences, the significantly higher energy demands of treadmill walking was an important finding.
The average self-selected overground walking speed and cadence were comparable to norms reported elsewhere: walking speed of 1-1.3 m/s and cadence of 90-140 steps/min (Alton et al., 1998; Andriacchi et al., 1977; Warabi et al., 2005; Winter, 1991). Matching the speed on the treadmill resulted in similar cadence, cycle times and stance/swing ratios. Others have reported a higher cadence in young subjects walking on the treadmill compared to overground at similar walking speed (Alton et al., 1998; Stolze et al., 1997; Warabi et al., 2005), but this did not occur in the present study. It is possible that the sense of security offered by the harness, which was not used in other studies, resulted in strides that were similar to those of overground walking. It is also possible that the length of the treadmill used in this study, which is 10-20 cm longer than the two reported in other studies (Alton et al., 1998; Stolze et al., 1997), reduced subjects’ concern that they might be transported off the back of the treadmill, which may have resulted in shorter strides and faster cadence.

Double support time was significantly shorter in treadmill walking, whether expressed in seconds or as a percentage of the gait cycle, supporting the findings of Stolze et al., (1997). However unlike Alton et al. (1998), Stolze et al. (1997) and Warabi et al. (2005), we did not find any differences in stance or swing when expressed in units of time or as percentages of gait cycle. The small but insignificant reductions in stance time and increases in swing time with treadmill walking in addition to the significant reduction in double support time support their general results.

Joint excursions at the hip, knee, and ankle during treadmill and overground walking were within ranges reported by others (Alton et al., 1998; Murray et al., 1985; Winter, 1991). Examining the maximum flexion and extension angles achieved during
stance revealed that only the amount of knee extension differed between treadmill and overground walking; an average of 2° less knee extension was observed when subjects walked on a treadmill versus overground walking. This finding supports the one study that reported less knee extension just prior to initial contact when on the treadmill (Strathy et al., 1983). The present study did not find significant increases in hip flexion as was observed by Alton et al. (1998) or significant reduction in hip extension (Murray et al., 1985) during treadmill walking though the kinematic profiles showed these tendencies. It is noteworthy that the average differences between treadmill and overground joint angle displacements in this study and in previous reports were less than 5°, which may be of minor clinical significance.

The temporal and kinematic changes that occurred in treadmill walking as a result of increased speed were expected. Nonetheless this is important to quantify. Although there is a considerable amount of information pertaining to changes with increases in walking speed in overground walking, it is possible that different gait adaptations might be needed to respond to the surface motion imposed by the treadmill. With increasing treadmill speed there was an increase in cadence, a decrease in gait cycle time and stance time and an increase in swing time. Double support time decreased by 8.5% and 11.6% at speeds of 110% and 120% of the self-selected speed. These changes were consistent with expectations based on overground walking at increasing speeds (Murray et al., 1985). The progressive increase in joint excursions was also anticipated (Winter, 1991) though it would be interesting to know if the adjustments associated with treadmill walking would be proportional to the adjustments made while walking overground at similar faster walking speeds.
The characteristic butterfly pattern of the VGRF was clearly evident in both treadmill and overground walking and was similar to those reported by White et al. (1998), and Yack et al. (1995). No differences were detected in the magnitude of F1 or F2 forces suggesting that early accelerations and midstance decelerations of the centre of gravity were similar between treadmill and overground conditions. White et al. (1998), and Yack et al. (1995) found higher F2 forces on the treadmill but the differences were small in magnitude and occurred with normal and fast speeds of walking.

The push-off (F3) force was lower during treadmill walking in the present study. This is thought to be due to the moving belt of the treadmill. Whether on ground or on a treadmill, the body has been described as being in a state of instability during the latter half of double support stance, that is, in the final stage of push-off where the weight is over the toes (Winter, 1990; Winter et al., 1990). During overground walking the propulsive force contributes to vertical displacement, but on the treadmill the ‘propulsive’ force is not resisted thus contributing to the belt movement. As a result the vertical component is reduced and this is reflected in the lower magnitude of F3. Even at higher speeds of treadmill walking the magnitude of the F3 vertical ground reaction force never reached that achieved during overground walking in this study.

Considering the similarity of temporal distance, angular displacement and vertical ground reaction force data between overground and treadmill walking, the higher metabolic cost of treadmill walking was surprising. Heart rate and oxygen consumption, the two indicators of metabolic cost, were both significantly higher in treadmill walking. As noted in the review of literature, comparisons of metabolic costs between overground and
treadmill walking are extremely inconsistent. The fact that this study has two measures of metabolic cost that have given consistent results supports the validity of this finding.

There are a number of possible explanations for the increased metabolic costs observed in the presence of similar temporal and kinematic outcomes. Because very similar movement patterns have been accomplished at greater cost, this suggests that the kinetics must be different. The differences in F3 magnitude in the two walking conditions supports this view although not in the direction expected. There are three ways energy costs can be higher in the presence of similar kinematics. First, there may be increased muscle-generated forces against an immovable object, like gripping the hand rails tightly. Second there may be co-contraction of agonist and antagonist muscles. Third, there may be a reduction in energy savings resulting from differences in energy exchange and/or energy transfer during the movement. Of these the second or the third are the most likely.

With respect to the presence of increased force against an immovable object, Greig et al. (1993) attributed increased energy costs during treadmill walking to the subjects gripping the support rails, which resulted in unwanted arm muscle activation. On the other hand when compared to free arm swing, just resting the hands on the hand rails has been reported to result in a decrease in aerobic demands in healthy subjects at steady state exercise testing on a treadmill (Berling et al., 2006). In the present study no hand rails were used, so this explanation is not reasonable.

With respect to increased co-contraction, this might occur if the person felt either less stable or more anxious and therefore used muscle co-contraction to provide stability, which would result in increased cost. Studies comparing treadmill versus overground walking have suggested that subjects walking on a treadmill are less secure or stable than
they are when walking overground (Alton et al., 1998; Stolze et al., 1997). Indeed the finding of higher cadences (Alton et al., 1998) and wider stride width (Stolze et al., 1997) have been presented as evidence in support of this view. Additionally, stance period has been reported to be shorter on the treadmill which may have reflected exaggerated elevation of the swing limb toward the moving belt (Alton et al., 1998), although these authors did not have the kinematic data to determine if this was the case. In the current study, subjects wore a harness for safety and did not report any concern of falling. Furthermore, stride width was smaller than it was overground, the period of double support was shorter and cadence was not different. These are indicators that stability was not an issue.

Anxiety could also cause subjects to become tense and increase co-contraction. Supporting increased anxiety as the cause of the increased cost is the fact that differences between overground and treadmill energy costs are more typical of older subject groups than of younger populations (Greig et al., 1993) as there is some evidence of anxiety in the elderly that relates to fear of falling (Lee et al., 2008). White et al. (1998) have stated that a period of familiarization with treadmill walking was required for accommodation or gait habituation. Since only two of our subjects had prior experience of walking on a treadmill and since instability was ruled out, co-activation associated with anxiety or task novelty may have accounted for the increased oxygen demand. Although if anxiety were an issue one would have expected a higher baseline HR as well as oxygen consumption prior to treadmill walking; this did not occur (see Table 2.5).

Increased co-contraction during treadmill walking might also occur because it was a new skill. This well-known phenomenon of increased joint stiffness to provide greater
stability around joints during skill acquisition diminishes as skill develops allowing the movement to be performed more efficiently (Payton and Kelley, 1972). The unfamiliarity with treadmill walking reported by most subjects makes this an attractive explanation, but without recording electromyograms from lower limb muscles, it is not possible to draw such a conclusion.

Failure to perform treadmill walking in ways that optimize energy exchanges and energy transfers also may explain the higher energy cost. Winter (1979, 1983) noted that about 60% of potential and kinetic energy are exchanged at self-selected speeds of walking, and any disruption in the normal pattern may result in lower levels of conservation. Even though walking speed and movement patterns were similar in the two situations, it is possible that energy transfer was less efficient with treadmill walking. Many years of walking overground results in optimal patterns of energy flow between the segments of the body. Given the differences in F3 magnitude it is conceivable that the kinetics of the movement are different in treadmill walking, it follows that the complex energy transfers between segments may also be disrupted. Using a theoretical model, Kuo has persuasively shown that if a subject uses hip work instead of ankle plantarflexion work in walking, the gait will be much less efficient (Kuo, 2002), which may have occurred on the treadmill. Full kinetic analysis for treadmill walking would be required to determine if this is the case, but it could not be done with only the VGRF data available.

2.7 Limitations

The present study provides a comprehensive evaluation of overground and treadmill walking in adults over 50 years of age that has shown the two modes to be mechanically similar in terms of temporal, kinematic and kinetic characteristics, but not in metabolic
requirements. This could be important clinically as the energy demands of treadmill walking may result in premature fatigue and undesirable cardiorespiratory challenge which may require consideration of its suitability as a training tool. Even though our subjects produced similar biomechanical gait patterns within several minutes of accommodation to walking on the treadmill and overground this may not have provided adequate time to adapt in terms of metabolic cost.

Speculatively, habituation can reduce anxiety and also increase the sense of security during treadmill walking which could normalize muscle activation patterns thereby reducing the oxygen consumption and energy costs to a level similar to overground walking. Further study is required to determine the association between muscle activity and metabolic cost add to establish and whether metabolic equivalency between the two modes of walking can be achieved with longer periods of familiarization.

2.8 Conclusion

Overground and treadmill walking were similar in terms of most temporal-distance, kinematic and kinetic indicators. Small reductions in double support time and decreased push-off force were evident on the treadmill; however, treadmill walking was more demanding in terms of metabolic cost than overground walking.

The findings suggest that even though the gait patterns on both treadmill and overground may be largely indistinguishable, the higher demands associated with treadmill walking may reflect a greater challenge for individuals. This may be particularly relevant in the presence of cardiovascular disorders.
REFERENCES


CHAPTER 3
Biomechanics and Metabolic Costs of Overground and Treadmill Walking in People with Stroke

Abstract

Background: Comparisons of treadmill and overground walking following stroke indicate that symmetry in temporal-distance measures is better on the treadmill suggestive of better gait economy. This study examined this issue by comparing the kinematic, kinetic and metabolic demands associated with overground and treadmill walking at matched speeds and increasing treadmill speeds. Methods: Ten people with hemiparesis due to stroke walked overground at their preferred speed that was matched on the treadmill. Temporal-distance outcomes, angular kinematics and vertical ground reaction forces were recorded once subjects were in steady state (stable heart rate and oxygen uptake). Belt speed was then increased 10% and 20% above preferred speed and steady state recordings obtained. Results: Step, stride, and stance times were longer when walking overground but the degree of symmetry was comparable for both surfaces. Kinematic data revealed interlimb asymmetry was more pronounced for all lower limb joint excursions during overground walking and vertical forces were higher. However, the metabolic demands were lower when walking overground than on the treadmill. Increasing the belt speed increased angular displacements and the vertical forces associated with both limbs such that symmetry remained unchanged. Metabolic demands increased significantly. Conclusions: People with stroke adopt a more symmetrical kinematic walking pattern on the treadmill that is maintained at faster belt speeds. The higher metabolic cost of walking on the treadmill compared to overground at matched
speeds challenges the assumption that gait symmetry reduces the energy costs of ambulation.

3.1 Introduction

The treadmill is gaining widespread use in clinical and research settings as a tool for evaluation and treatment of walking impairments post-stroke as it provides a controlled environment in which the walking speed and the amount of assistance provided can be adapted to the individual’s abilities and requirements. On a treadmill, patients can practice complete gait cycles even if they have balance deficits or cannot support their full body weight using body weight support systems. In this way users can engage in task oriented repetitive practice which has been shown to improve walking capabilities (Barbeau and Visintin, 2003; Hesse et al., 1994, 1999; Visintin et al., 1998; Waagfjord et al., 1990) and aerobic capacity (Eich et al., 2004; Macko et al., 1997; Pang and Eng, 2006) in stroke subjects. What is not known is the degree to which walking overground and on a treadmill is similar in this population, information that is important in relation to the transfer of learning across the two walking conditions.

Walking speed has been shown to discriminate between stroke survivors able to ambulate within the community and those who cannot (Lord et al., 2004). There is a wide variation in walking ability following a stroke and people with faster walking speeds have been observed to have more normal walking patterns (Olney et al., 1994). Indeed walking at faster than normal treadmill speeds has been associated with greater improvements in overground walking speed (Pohl et al., 2002) although the gait quality (i.e., biomechanics) and metabolic costs of increased speed have not been examined. It is important to investigate changes in both conditions because each poses different
challenges from the visual environment and from the walking surfaces (Harris-Love et al., 2004).

Symmetry is often assumed to be a characteristic feature of normal walking, with a lack of gait symmetry being related to possible functional differences between lower extremities (Sadeghi et al., 2000). The asymmetrical nature of hemiparetic gait in terms of temporal-distance parameters (Bayat et al., 2005, Harris-Love et al., 2001, 2004; Silver et al., 2000), joint angular displacements (Griffin et al., 1995; Nadeau et al., 1999) and vertical ground reaction forces (Hesse et al., 1994, 1999) has been well documented. Harris-Love et al., 2004 and Silver et al., 2000 found that people with stroke walked with increased symmetry and that muscle activity normalized when walking on the treadmill. They suggested that this was an indication of higher motor efficiency and better gait economy, which was attributed to the movement of the belt pulling the stance limb backward thus promoting a more timely initiation of swing (Harris-Love et al., 2001, 2004). The corresponding normalization in phasic muscle activity indicates that the improved symmetry is not solely mechanically induced, but is rather actively achieved (Harris-Love et al., 2004; Hesse et al., 1999). Biomechanical aspects, however, were not examined and would be important to adequately evaluate the impact of treadmill walking on gait patterns in stroke (Silver et al., 2000). To determine gait efficiency and energy cost a more detailed biomechanical evaluation including joint kinematic profiles and metabolic measurement is needed (Chen et al., 2005; Mian et al., 2006). This study compared the temporal distance parameters, biomechanical profiles and metabolic costs of overground and treadmill walking in people who had sustained a stroke.
3.1.1 Temporal and Distance Parameters

Overground walking speeds in stroke subjects have been reported in the range of 0.16-1.50 m/s (Bohannon, 1987, 1992; Olney et al., 2006; Roth et al., 1997; Teixeira-Salmela et al., 1999; Wall and Turnbull, 1986). Cadence and stride length are the main determinants of overground and treadmill walking speed (Bohannon, 1987, 1992; Nakamura, 1988; Roth et al., 1997; Wagenaar and Beek, 1992).

Hesse et al. (1999) had 18 subjects with hemiparesis due to stroke walk at their self-selected walking speed overground and on a treadmill and they found that overground speed (0.32 m/s) and cadence (62 steps/min) were higher than when walking on a treadmill (0.27 m/s and 53 steps/min), respectively. In contrast, Bayat et al. (2005) (10 stroke subjects, average age = 63 years) reported a higher cadence on the treadmill compared to overground although the walking speed was significantly lower on the treadmill. Even when they controlled for walking speed, cadence remained significantly higher when walking on the treadmill. This was associated with significantly shorter stride lengths (overground: 0.76 m; treadmill: 0.38 m).

Comparing overground and treadmill walking, Hesse et al. (1999) reported similar stride length (0.61 m), stance period (76-81%) and double support period (22%) on the unaffected and the affected sides (stance: 67%; double support: 22%). In contrast, Harris-Love et al. (2001) compared overground and treadmill walking at similar speeds (0.54 m/s) in 18 stroke subjects and reported similar cadence (78 steps/min), but significantly decreased stance period on the unaffected side (overground stance – 74%; treadmill – 71%) and significantly increased stance period on the affected side (overground stance – 60%; treadmill – 64%) during treadmill walking. The changes in
stance period resulted in an increase of 11% in stance period symmetry (more symmetrical ‘affected to unaffected’ stance period ratio) on the treadmill. Hesse et al. (1999) also reported 8% greater stance duration symmetry during treadmill walking.

Harris-Love et al. (2001) hypothesized that treadmill walking promotes interlimb symmetry as the belt pulls the stance limb backward which appears to hasten the swing phase and initiation of stance on the affected limb. It seems from these studies that at comparable walking speeds an increased cadence was attributed to the changes in proportions of the gait cycle occupied by the stance and swing phases that in turn promoted the stance period symmetry on the treadmill.

3.1.2 Kinematics

No studies have been identified that compare the kinematics of walking on the treadmill with overground walking in people with stroke. Comparisons of overground walking with healthy individuals have shown that the hip, knee and ankle excursions of individuals with stroke are lower than those of healthy individuals (Knutsson and Richards, 1979; Lehmann et al., 1987; Olney et al., 1989, 1991, 1994).

At initial contact subjects with stroke characteristically show less hip flexion, more knee flexion and more ankle plantarflexion than their healthy counterparts. During midstance they show less hip and knee flexion and at push-off they show less hip extension (more hip flexion) and less knee flexion and ankle plantarflexion (Kerrigan et al., 1999; Lehmann et al., 1987; Olney et al., 1991, 1994; Olney and Colborne, 1991; Olney and Richards, 1996). Some studies have also reported hyperextension or less knee flexion and/or more ankle plantarflexion at initial contact (Kerrigan et al., 1999; Olney and Richards, 1996; Perry, 1969). It is not known if stroke subjects exhibit similar
kinematic patterns when walking on a treadmill as described for their overground walking.

3.1.3 Kinetics

The characteristic profile of the vertical ground reaction force shows an initial peak greater than body weight after foot contact (F1), a valley (less than body weight) during midstance (F2), and a second peak at push-off (F3) in treadmill walking although some minor differences exist with overground walking.

Harris-Love et al. (2001) compared overground and treadmill walking at similar speeds (0.54 m/s) in 18 stroke subjects and reported that F1 forces were about 10% lower during treadmill walking on both the unaffected (overground: 9.21N/kg; treadmill: 8.45N/kg) and the affected sides (overground: 9.25N/kg; treadmill: 8.13N/kg) with values normalized to body weight. An earlier study also compared overground with treadmill walking in 18 stroke subjects (average age = 60 years) at self-selected walking speeds of 0.32 m/s and 0.27 m/s, respectively reporting significantly lower F1 (overground: 746 N; treadmill: 730 N) and F3 (overground: 746 N; treadmill: 727 N) forces only on the affected side during treadmill walking (Hesse et al., 1999). These authors (Hesse et al., 1994), also measured overground vertical ground reaction forces in 148 subjects with stroke (average age = 57 years) walking at an average speed of 0.88 m/s and average F1 forces were reported to be 1.03 and 1.05 (times body weight) on the unaffected and affected sides respectively; F3 forces were 1.03 and 1.01.

Harris-Love et al. (2001) attributed the reduction in the early force peak to the use of hand rails and to the compliant surface of the treadmill that attenuated peak forces through absorption. Hesse et al. (1999) attributed the lower F1 and F3 forces on the
affected side to the slower walking speed on the treadmill since they did not match the speed of overground walking.

### 3.1.4 Metabolic Costs of Walking in Subjects with Stroke

An early study on metabolic cost of walking reported that, unlike healthy subjects in whom the natural speed of walking corresponded to minimal energy expenditure, subjects with stroke walked less efficiently because they are unable to attain an optimal speed (Bard, 1963). The metabolic cost of walking in stroke subjects is variable depending on the degree of impairment, the stage of recovery and the protocol used to measure the costs (da Cunha-Filho et al., 2003). The cost of walking in stroke has been reported to be about 1.5-2 times greater than in healthy controls (Macko et al., 1997a,b). The increased energy demand secondary to stroke can lead to a decline in mobility, which in turn is linked to reductions in aerobic capacity, disuse atrophy and weakness further impairing function (Ada et al., 2003; Macko et al., 1997a,b; Olney et al., 1986).

Cardiovascular co-morbidities that have an effect on overall metabolic cost are quite common in stroke. A review article stated that cardiovascular disorders can be causal, consequential or coexistent in stroke and are found in 75% of patients who have suffered a stroke (Roth et al., 1993). Treadmill stress testing in 31 subjects with stroke reported that 29% of the subjects without known heart disease had signs of myocardial ischemia (Macko et al., 1997). Since cardiovascular co-morbidities are quite common in stroke, knowledge of the metabolic cost associated with treadmill walking relative to overground walking is important in order to accurately gauge the relative task demands. To date, there is no study that has directly compared the metabolic cost between overground and treadmill walking at steady state.
3.2 **Purpose**

The purpose of this study was to compare the biomechanics and metabolic cost of overground and treadmill walking in subjects with stroke. The objectives were as follows:

- To compare temporal-distance parameters and lower limb kinematics between walking overground and on the treadmill,
- To compare the vertical ground reaction forces between walking overground and on the treadmill,
- To compare the energy cost of overground and treadmill walking, and
- To explore alterations in biomechanics and metabolic costs with increased speed of treadmill walking.

3.3 **Methodology**

Prior to recruitment of subjects, ethical approval for the study was obtained from the Faculty of Health Sciences and Affiliated Teaching Hospitals Research Ethics Board of Queens University, Kingston, Ontario. Subjects with stroke were recruited in the following ways:

1) advertisements posted in the local newspaper and at various locations in the city,
2) letters sent to the members of the local Seniors Centre, describing the study purpose and protocol,
3) letters describing the study purpose and protocol distributed to the members of the Stroke Club at St. Mary’s of the Lake Hospital, Kingston, and
4) letters describing the study purpose and protocol distributed to local clinicians working in the area of stroke rehabilitation.
Subjects who were interested in participating in the study contacted the principal investigator and were screened by telephone to ensure they met all inclusion criteria (had sustained a hemispheric stroke, had residual weakness and/or spasticity in the affected lower limb, were able to walk independently without any walking aid for ten minutes continuously and reported themselves to be in good health). All subjects were given detailed information about the study and their questions were answered. Subjects were excluded if they self reported any cardiovascular or cardiopulmonary problem, any orthopaedic condition affecting mobility, uncontrolled metabolic diseases or other relevant pre-existing condition. Those eligible to participate were invited to the Motor Performance Laboratory at Queen’s University for final screening and testing.

The procedure, instrumentation, data acquisition and data processing for overground and treadmill walking were similar to that described in sections 2.3.2 – 2.3.5 of Chapter 2.

3.4 Data Analysis

Statistical analyses were performed using SPSS (Version 14, SPSS Inc.) and a significance level of p<0.05 was set. Descriptive statistics (mean ± SD) were calculated for all outcome measures. Two factor analyses of variance were performed to determine if there were differences in dependent measures between sides (unaffected and the affected) and condition (overground and treadmill) and if there were interactions between these factors. Interlimb asymmetry was the arithmetic difference between the unaffected and the affected sides. Similar analyses were performed to examine the effect of speed in association with overground walking at self-selected speed. Metabolic data were analyzed using a single factor analysis of variance.
3.5 Results

Ten subjects (6 males and 4 females) with a mean age of 64.9 ± 9.7 years (range 54-80 years), mean mass of 72.7 ± 17.1 kg and mean height of 1.7 ± 0.1 m completed the protocol without incident. All subjects walked independently without any personal assistance or walking aids. The average time since stroke was 5.9 ± 4.1 years ranging from 1.5 to 15.0 years; five subjects had right hemiparesis and 5 had left hemiparesis.

3.5.1 Gait Speed and Cadence

The individual and descriptive data for gait speed and cadence during overground and treadmill walking are listed in Table 3.1.

The average self-selected overground walking speed and matched speed on the treadmill were identical at 0.81m/s. The walking speeds corresponding to 110% and 120% of the matched treadmill speed were 0.89m/s (±0.17) and 0.99m/s (±0.18) respectively. As expected, speeds corresponding to 110% and 120% of the matched treadmill speed were significantly faster than the self-selected and matched speed of walking (p<0.001).

At comparable speeds, the cadence achieved during overground (87 ±11 steps/min) and treadmill walking (88 ±13 steps/min) were similar (p = 0.523). The cadence at faster walking speeds on the treadmill appeared higher than those observed during overground walking at self-selected speed, but the differences did not reach significance (p≥0.064).
Table 3.1  Gait speed (m/s) and cadence (steps/min) for individual subjects

<table>
<thead>
<tr>
<th>Subject Number</th>
<th>Overground (Self-selected Speed)</th>
<th>Treadmill (Matched Speed)</th>
<th>Treadmill (+10%)</th>
<th>Treadmill (+20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Speed</td>
<td>Cadence</td>
<td>Speed</td>
<td>Cadence</td>
</tr>
<tr>
<td>S1</td>
<td>0.98</td>
<td>98</td>
<td>0.98</td>
<td>81</td>
</tr>
<tr>
<td>S2</td>
<td>0.75</td>
<td>70</td>
<td>0.75</td>
<td>70</td>
</tr>
<tr>
<td>S3</td>
<td>0.96</td>
<td>91</td>
<td>0.95</td>
<td>86</td>
</tr>
<tr>
<td>S4</td>
<td>0.59</td>
<td>69</td>
<td>0.59</td>
<td>86</td>
</tr>
<tr>
<td>S5</td>
<td>0.56</td>
<td>82</td>
<td>0.57</td>
<td>97</td>
</tr>
<tr>
<td>S6</td>
<td>0.69</td>
<td>93</td>
<td>0.68</td>
<td>98</td>
</tr>
<tr>
<td>S7</td>
<td>0.96</td>
<td>103</td>
<td>0.95</td>
<td>92</td>
</tr>
<tr>
<td>S8</td>
<td>0.81</td>
<td>93</td>
<td>0.81</td>
<td>80</td>
</tr>
<tr>
<td>S9</td>
<td>0.96</td>
<td>92</td>
<td>0.95</td>
<td>108</td>
</tr>
<tr>
<td>S10</td>
<td>0.83</td>
<td>80</td>
<td>0.84</td>
<td>83</td>
</tr>
<tr>
<td>Average</td>
<td>0.81</td>
<td>87</td>
<td>0.81</td>
<td>88</td>
</tr>
</tbody>
</table>

* Indicates significantly different from overground condition (p<0.05)

3.5.2 Temporal and Distance Parameters

The descriptive data for the temporal distance parameters during overground and treadmill walking are listed in Tables 3.2 and for the unaffected and affected limbs in Table 3.3.

At comparable speeds, there was a significant decrease in the percentage of time spent in stance and a corresponding increase in the percentage of time spent in swing on both sides during treadmill walking compared to overground (p<0.001). It follows that the stance/swing ratio on the unaffected and the affected sides was also less during treadmill walking (p<0.029). The average stride length was similar in both walking conditions although the step length on the affected side was significantly shorter on the treadmill at matched speed (p=0.020). At matched speeds, all other temporal distance parameters including gait cycle time, double support time and stride width were similar for overground and treadmill walking (p>0.050).
At both increments of fast walking speeds the stance and swing phases as percentages of gait cycle were significantly different from those of overground walking (p<0.030). When subjects walked 20% faster than self-selected overground walking speed, the step length and stride length increased significantly (p<0.011).

Table 3.2 Mean values (± 1SD) of temporal distance parameters for overground and treadmill walking

<table>
<thead>
<tr>
<th>Variable</th>
<th>Overground</th>
<th>Treadmill</th>
<th>Treadmill (+10%)</th>
<th>Treadmill (+20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cycle Time (sec)</td>
<td>1.29 ±0.20</td>
<td>1.30 ±0.22</td>
<td>1.25 ±0.16</td>
<td>1.21 ±0.14</td>
</tr>
<tr>
<td>Stride Length (m)</td>
<td>1.16 ±0.17</td>
<td>1.08 ±0.22</td>
<td>1.17 ±0.19</td>
<td>1.27* ±0.19</td>
</tr>
<tr>
<td>Stride width (m)</td>
<td>0.12 ±0.03</td>
<td>0.11 ±0.02</td>
<td>0.13 ±0.04</td>
<td>0.13 ±0.04</td>
</tr>
<tr>
<td>Double Support Time (sec)</td>
<td>0.37 ±0.08</td>
<td>0.34 ±0.11</td>
<td>0.37 ±0.18</td>
<td>0.33 ±0.12</td>
</tr>
<tr>
<td>Stance Period Symmetry (Affected/Unaffected)</td>
<td>0.96 ±0.03</td>
<td>0.93 ±0.07</td>
<td>0.98 ±0.04</td>
<td>0.91 ±0.07</td>
</tr>
</tbody>
</table>

* Indicates significantly different from overground condition (p<0.05)
### Table 3.3
Mean values (±1SD) of temporal distance parameters for the unaffected and affected sides for overground and treadmill walking

<table>
<thead>
<tr>
<th>Variable</th>
<th>Overground</th>
<th>Treadmill</th>
<th>Treadmill (+10%)</th>
<th>Treadmill (+20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Unaffec</td>
<td>Affec</td>
<td>Unaffec</td>
<td>Affec</td>
</tr>
<tr>
<td><strong>Stance Time (sec)</strong></td>
<td>0.87±0.16</td>
<td>0.82±0.14</td>
<td>0.71±0.14</td>
<td>0.65±0.09</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>-0.04±0.04</td>
<td>0.59±0.13</td>
</tr>
<tr>
<td><strong>Swing Time (sec)</strong></td>
<td>0.44±0.05</td>
<td>0.48±0.09</td>
<td>-2.98±2.01</td>
<td>54.71±7.05</td>
</tr>
<tr>
<td><strong>Stance (%) of gait cycle</strong></td>
<td>66.32±3.14</td>
<td>63.33±3.19</td>
<td>50.76±8.03</td>
<td>1.28±0.17</td>
</tr>
<tr>
<td><strong>Swing (%) of gait cycle</strong></td>
<td>33.68±3.14</td>
<td>36.67±3.19</td>
<td>49.24±8.03</td>
<td>1.11±0.54</td>
</tr>
<tr>
<td><strong>Stance/Swing Ratio</strong></td>
<td>1.99±0.30</td>
<td>1.75±0.25</td>
<td>0.53±0.10</td>
<td>0.53±0.10</td>
</tr>
<tr>
<td><strong>Step Length (m)</strong></td>
<td>0.58±0.10</td>
<td>0.56±0.09</td>
<td>0.54±0.12</td>
<td>0.54±0.12</td>
</tr>
</tbody>
</table>

* Indicates significantly different from corresponding overground condition (p<0.05)

Significantly different values for unaffected and affected sides are bolded (p<0.05)

Unaffec = Unaffected side; Affec = Affected side; Asym = Asymmetry (unaffected minus affected)
3.5.3 Kinematic Data

The descriptive data associated with the angular displacements of the unaffected and affected hip, knee and ankle joints during the stance phase and the interlimb asymmetry for overground and treadmill walking are listed in Table 3.4. The average angular displacement profiles for all subjects are shown in Figures 3.1-3.2.

At comparable speeds, there were no differences between overground and treadmill walking in terms of the total hip, knee or ankle range of motion on the unaffected or the affected sides. The hip and knee flexion and extension angles as well as ankle dorsiflexion and plantarflexion angles were also similar on the unaffected and the affected sides at comparable speeds. The maximum hip flexion appeared greater on the unaffected side than the affected side for both surfaces though the values did not reach significance. The dynamic range of motion was in all cases significantly higher on the unaffected side than the affected side for all lower limb joints during overground walking only. Asymmetries in joint excursions were much reduced with treadmill walking and while increased belt speed was generally accompanied by increases in joint excursion, this occurred in both limbs such that symmetry remained unaffected (p>0.186).

At fast walking speeds of 110% and 120% of the matched speed on the treadmill, the total range of motion at the hip (p<0.007) and the knee (p<0.050) were significantly greater than overground walking on the affected side.

From the angular displacement profiles (Figures 3.1) there appeared to be less hip extension on the unaffected side, greater hip flexion during early stance on the affected side and greater knee flexion during late stance on both sides during treadmill walking at matched speed. These observed differences were not supported statistically.
### Table 3.4  
Mean (± 1SD) of angular displacements of the unaffected and affected hip, knee and ankle joints (in degrees) during stance

<table>
<thead>
<tr>
<th>Joint Angle</th>
<th>Overground</th>
<th>Treadmill</th>
<th>Treadmill (+10%)</th>
<th>Treadmill (+20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Unaffec</td>
<td>Affec</td>
<td>Asym</td>
<td>Unaffec</td>
</tr>
<tr>
<td><strong>Hip</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexion</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>24 (±7)</td>
<td>21 (±8)</td>
<td>2.71 (±2.58)</td>
<td>24 (±7)</td>
</tr>
<tr>
<td></td>
<td>-8 (±6)</td>
<td>-7 (±6)</td>
<td>-1.45 (±6.32)</td>
<td>-5 (±8)</td>
</tr>
<tr>
<td><strong>Excursion</strong></td>
<td>32 (±5)</td>
<td>28 (±7)</td>
<td>4.16 (±6.76)</td>
<td>29 (±5)</td>
</tr>
<tr>
<td>Flexion</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>45 (±6)</td>
<td>39 (±7)</td>
<td>5.91 (±10.41)</td>
<td>48 (±9)</td>
</tr>
<tr>
<td></td>
<td>3 (±3)</td>
<td>1 (±6)</td>
<td>2.96 (±5.57)</td>
<td>4 (±3)</td>
</tr>
<tr>
<td><strong>Excursion</strong></td>
<td>42 (±6)</td>
<td>39 (±8)</td>
<td>2.95 (±11.93)</td>
<td>44 (±9)</td>
</tr>
<tr>
<td><strong>Knee</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexion</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>13 (±2)</td>
<td>12 (±2)</td>
<td>1.08 (±2.13)</td>
<td>13 (±2)</td>
</tr>
<tr>
<td></td>
<td>-6 (±5)</td>
<td>-4 (±3)</td>
<td>-2.62 (±6.65)</td>
<td>-7 (±4)</td>
</tr>
<tr>
<td><strong>Excursion</strong></td>
<td>19 (±5)</td>
<td>16 (±4)</td>
<td>3.69 (±7.18)</td>
<td>20 (±4)</td>
</tr>
</tbody>
</table>

* Indicates significantly different from corresponding overground condition (p<0.05)  
Significantly different values for unaffected and affected sides are bolded (p<0.05)  
Unaffec = Unaffected side; Affec = Affected side; Asym = Asymmetry (unaffected minus affected)
Figure 3.1  Mean angular profiles of the hip, knee and ankle for overground (thick solid line) and treadmill (thick dotted line) walking at matched speed for the unaffected and the affected sides. Corresponding thin lines represent ± 1 standard deviation.
Figure 3.2  Mean angular profiles of the hip, knee and ankle for treadmill walking at matched speed (thick solid line), 110% (grey solid line) and 120% (grey dotted line) of the matched speed for the unaffected and the affected sides. Corresponding thin line represents ± 1 standard deviation for treadmill walking at matched speed.
3.5.4 **Kinetic Data**

The average maximal vertical ground reaction force peak values of F1 and F3, and minimum force values of F2 from the unaffected and affected sides are presented in Table 3.7 for all conditions.

A significant interaction between side and condition reflected the fact that the peak vertical ground reaction force values occurring after foot contact (F1) were significantly lower (p=0.032) on the unaffected side but similar on the affected side during treadmill compared to overground walking. F2 forces were similar on the unaffected and the affected sides for both surface conditions at matched speed. The second peak (F3) associated with push-off was smaller on the treadmill, a pattern demonstrated on the unaffected (p=0.007) and the affected sides (p=0.022) when compared with overground walking.

At fast walking speeds on the treadmill, F1 magnitudes were similar on the unaffected and the affected sides, F2 was significantly lower at 120% of the matched speed (p<0.05), and F3 was lower at 110% (p<0.05) and 120% (p=0.04) of overground and matched treadmill conditions. The average vertical ground reaction force profiles for all subjects are illustrated in Figures 3.3 and 3.4. Note that the magnitudes are attenuated maxima and minima due to temporal misalignment of F1, F2 and F3.
Table 3.5  Mean (± 1SD) vertical ground reaction forces for the unaffected and affected sides expressed as a proportion of body weight

<table>
<thead>
<tr>
<th>Force</th>
<th>Overground</th>
<th>Treadmill</th>
<th>Treadmill (+10%)</th>
<th>Treadmill (+20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Unaffec</td>
<td>Affec</td>
<td>Asym</td>
<td>Unaffec</td>
</tr>
<tr>
<td>F1</td>
<td>1.07</td>
<td>1.03</td>
<td>0.04</td>
<td>1.01*</td>
</tr>
<tr>
<td></td>
<td>(±0.06)</td>
<td>(±0.08)</td>
<td>(±0.09)</td>
<td>(±0.08)</td>
</tr>
<tr>
<td>F2</td>
<td>0.91</td>
<td>0.89</td>
<td>0.02</td>
<td>0.88</td>
</tr>
<tr>
<td></td>
<td>(±0.13)</td>
<td>(±0.12)</td>
<td>(±0.06)</td>
<td>(±0.13)</td>
</tr>
<tr>
<td>F3</td>
<td>1.08</td>
<td>1.03</td>
<td>0.05</td>
<td>0.99*</td>
</tr>
<tr>
<td></td>
<td>(±0.07)</td>
<td>(±0.09)</td>
<td>(±0.10)</td>
<td>(±0.08)</td>
</tr>
</tbody>
</table>

* Indicates significantly different from overground condition (p<0.05)
Significantly different values for unaffected and affected sides are bolded (p<0.05)
Unaffec = Unaffected side; Affec = Affected side; Asym = Asymmetry (unaffected minus affected)
Figure 3.3  Mean vertical ground reaction forces during overground (thick solid line) and treadmill walking at matched speed (thick dotted line) for the unaffected and the affected sides. Corresponding thin lines represent ± 1 standard deviation.
Figure 3.4  Mean vertical ground reaction forces during treadmill walking at matched speed (thick solid line), 110% (thick grey line) and 120% (grey dotted line) of the matched speed for the unaffected and the affected sides.
3.5.5 Metabolic Data

At matched walking speed the average HR and oxygen consumption (VO₂) were significantly higher by 5% and 15% respectively during treadmill walking than overground walking (p<0.003). Marked increases in HR and VO₂ accompanied increases in walking speed (p<0.035). These data are summarized in Table 3.6.

HR and VO₂ returned to resting values almost 4 minutes after overground walking. The time required to fully recover after treadmill walking was significantly longer (p<0.001). The longer duration related in large part to the longer duration and greater walking speeds achieved during treadmill walking.

<table>
<thead>
<tr>
<th>Table 3.6</th>
<th>Metabolic data (mean ± 1SD) for overground and treadmill walking</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Overground</td>
</tr>
<tr>
<td></td>
<td>Resting</td>
</tr>
<tr>
<td>Heart Rate (beats/min)</td>
<td>73 (±8)</td>
</tr>
<tr>
<td>Oxygen Consumption (ml/kg/min)</td>
<td>3.87 (±0.16)</td>
</tr>
<tr>
<td>Time to Recovery (min)</td>
<td>3.90 (±0.74)</td>
</tr>
</tbody>
</table>

* Indicates significantly different from overground walking (p<0.05)
† Indicates significantly different from treadmill walking (p<0.05)

3.6 Discussion

The study demonstrates that, while the kinematic and kinetic differences between overground and treadmill walking were small, the energy demands of treadmill walking were significantly higher.
The average self-selected walking speed and cadence in stroke subjects is in accordance with other studies regardless of the stage of motor recovery (Bohannon, 1987, 1992; Olney et al., 2006; Roth et al., 1997; Teixeira-Salmela et al., 1999; Wall and Turnbull, 1986). At matched walking speeds gait cycle time and cadence were similar for overground and treadmill walking as observed in other studies (Harris-Love et al., 2001; Hesse et al., 1999). Bayat et al. (2005), suggested that subjects were less secure or stable when walking on a treadmill as evidenced by higher cadence and shorter stride length than walking overground. In the current study, stride length and cadence were similar for overground and treadmill walking suggesting that stability was not likely experienced differently by our subjects. Perhaps the harness worn for safety alleviated any concerns about stability.

Asymmetry in stroke has been defined in several ways; ratios (unaffected/affected) (Griffin et al., 1995; Harris-Love et al., 2001; Hesse et al., 1999), difference and ratios (i.e. unaffected-affected/unaffected) (Hsu et al., 2003; Patterson et al. 2009) and as differences (unaffected-affected) (Brouwer et al., 2009). Differences were opted for in this descriptive study as this is most useful in identifying the magnitude of the inter-limb discrepancies in the unit of measurement. In self-paced overground walking in stroke, asymmetry in kinematic and kinetic measures is well documented (Griffin et al., 1995; Harris-Love et al., 2001, 2004; Parvataneni et al., 2007) and is reaffirmed in the current study in terms of joint excursions and push-off forces. Harris-Love et al. (2001) and Hesse et al. (1999) reported greater stance symmetry on the treadmill. Harris-Love et al. (2001) hypothesized that walking on a treadmill would induce greater symmetry because of the constraints imposed by a belt moving at a
constant speed. Decreased interlimb differences in temporal parameters and vertical ground reaction force profiles supported their argument although the joint biomechanics associated with the changes were unknown. In the Harris-Love et al. (2001) study subjects were provided with intermittent body weight support in early stance, and Hesse et al. (1999) permitted the use of hand rails. In contrast, subjects in the current study had no external support which may have accounted for the similarity in stance time symmetry ratio observed in treadmill and overground walking. Our findings also reveal an overall reduction in interlimb asymmetry of all lower limb joint excursions associated with walking on a treadmill which was particularly marked at the hip and the knee. In rehabilitation, the achievement of symmetry is often considered a goal of gait re-education (Griffin et al., 1995; Hesse et al., 1994, 1999) and an indicator of walking efficiency (Harris-Love et al., 2001; Silver et al., 2000). Others suggest that the neuromuscular deficits that predominate on the side contralateral to the lesioned hemisphere result in motor strategies that exploit the asymmetry for optimal performance (Griffin et al., 1995; Nadeau et al., 1999; Parvataneni et al., 2007). Indeed, asymmetrical kinematic and kinetic gait patterns in healthy subjects who demonstrate dominance of one leg over the other are not uncommon (Sadeghi et al., 2000). In the present study the immediacy and consistency of the biomechanical adaptations observed in subjects as they switched from overground to treadmill walking is suggestive of a response to restricting the degrees of freedom permitting the adoption of a walking pattern that may not be possible in the overground condition but optimal on the treadmill.

The kinematic profiles were very similar across conditions. The small differences in angular displacement of hip flexion and knee flexion observed from the profiles (Fig
3.1) at isolated joints were insignificant on their own, though the cumulative effect across joints may have resulted in subjects curtailing the stance phase and increasing swing phase thereby bringing the stance/swing ratios closer to normal (approximately 1.25). It is also possible that subjects flex their knee during late push-off as a strategy to prevent the limb from being pulled backwards by the belt. In overground walking subjects can adjust their walking speed within a stride to control, for example, the rate at which the limbs are brought into extension (Bayat et al., 2005) whereas during treadmill walking the adjustment within a stride needs to be optimally timed to accommodate the constant speed of the belt.

The typical butterfly pattern of vertical ground reaction forces was preserved during both overground and treadmill walking although the magnitudes of the F1 peak on the unaffected side and the F3 peak on both sides were lower when walking on the treadmill. In comparison, Hesse et al. (1999) reported lower F1 and F3 forces during treadmill walking on the affected side only. Harris-Love et al. (2001) reported 10% lower F1 forces during treadmill walking on both unaffected and affected sides, which they attributed to the use of hand rails, absorption by the compliant surface of the treadmill, and a more symmetrical walking pattern that enabled improved deceleration of the lower limb before initial contact of the foot with the support surface. The first two reasons are not applicable in the current study, as the force plates beneath the belt were non compliant and hand rails had been removed. However, there was more kinematic symmetry in treadmill walking, which supports Harris-Love’s explanation. It seems likely that a shorter stance phase associated with earlier knee flexion and limited push-off contributed to the lower F1 and F3 forces.
To date, no studies have directly compared the metabolic costs of overground and treadmill walking. Heart rates of 90-100 beats/min and VO2 values of 7-11 ml/kg/min have been reported for overground (da Cunha-Filho et al., 2003; Platts et al., 2006) and treadmill walking (Danielsson and Sunnerhagen, 2000; Macko et al., 1997a) in people with stroke. The findings in the present study are consistent with these reports but further demonstrate that HR and VO2 were both higher in treadmill walking than overground walking, a finding not previously reported. Considering the similarity of temporal distance parameters between overground and treadmill walking and the increased symmetry in joint excursion with treadmill walking, the higher energy cost of treadmill walking was unexpected.

There are a number of possible explanations for the increased metabolic cost in the presence of similar biomechanical outcomes. In stroke subjects stability is often compromised which causes increases in joint stiffness. For example, if the person felt less stable while walking on the treadmill, more muscle co-contraction would serve to increase joint stiffness which would result in increased metabolic cost. Chen et al. (2005) suggested that subjects with stroke were less stable when treadmill walking based on findings of asymmetry in step length and wider stride width compared to overground walking. These characteristics were not evident in the present study, although the step length on the affected side (as compared to overground) and the stance period were shorter on the treadmill. These differences may have been due to exaggerated elevation of the swing limb away from the moving belt but without kinematic data associated with the swing phase this cannot be determined. The similarity in step width and the willingness to
increase speed by 20% on the treadmill suggest that the elevated metabolic cost of treadmill walking is unlikely secondary to instability.

It is, however, possible that joint stiffness was increased in association with the novelty of that task. Increased co-contraction frequently accompanies performance of novel tasks (Payton and Kelley, 1972). Only one subject had reported previous experience with treadmill walking. Increased joint stiffness diminishes energy efficiency and is associated with greater total muscle work. The higher HR (see Table 3.8) and oxygen consumption during treadmill along with the unfamiliarity with treadmill walking reported by most subjects supports this explanation; however, electromyographic data are required to make such a determination.

It is unclear what role deconditioning may have had in causing the energy differences between overground and treadmill walking. The stroke subjects may have been deconditioned due to a relatively physically inactive lifestyle after stroke that limits their performance during physical activities and reduces the overall aerobic capacity. Both overground and treadmill walking are repetitive tasks, but treadmill walking does not provide the opportunity to adjust speed within a stride and as such might be more demanding. Whether the increased energy cost of treadmill walking could induce premature fatigue and thus challenge the suitability of treadmill walking remains unclear.

Disturbed mechanical energy patterns resulting in higher overall energy costs have been reported in stroke subjects as they fail to achieve the walking speeds that permit the optimal degree of energy conservation through kinetic and potential energy exchange of the body segments (Bard, 1963; Olney et al., 1986). It is unclear whether the subjects in this study exhibited better conservation of energy with overground walking
than treadmill walking although the differences in push-off force between the two conditions suggest this could be the case. The reduced propulsive thrust by the ankle plantarflexors at F3 that was evident in treadmill walking on both affected and unaffected sides would mean that additional work by the hip flexors and/or extensors would be needed to accomplish the work of walking. Using a simple theoretical model, Kuo (Kuo, 2002) has persuasively shown that use of the hip extensors and hip flexors to generate the positive work of walking is much less efficient than use of the ankle plantarflexors. Kuo’s analysis lends considerable weight to the possibility that differences in muscle work were responsible, at least in part, for increased costs of treadmill walking.

3.7 Limitations

It was important that the experimental conditions for the two situations be as similar as possible so that comparisons could be attributed only to the different mode of walking. For this reason the hand rails on the treadmill were removed necessitating that the harness be attached to the overhead support system for treadmill testing. This may have interfered in a subtle way with subjects’ gait pattern by providing a greater sense of stability than the overground condition. If true, the effect on HR and VO2 would be to reduce them, not increase them, as seen in the current study.

Our protocol required overground walking at self-selected speed to be tested first in order to enable matching of the speed on the treadmill, thereby allowing comparison of gait profiles. This may have influenced the metabolic measures during treadmill walking even though there was a rest period provided between overground and treadmill walking. Considering that the HR and VO2 had returned to baseline levels prior to treadmill walking, it is unlikely that the fixed order of trials had any significant impact.
3.8 Conclusion

Overground and treadmill walking were similar in terms of temporal-distance parameters when performed at similar walking speed. Small differences in stance and swing time and consequently the stance/swing ratios and decreased push-off force were evident on the treadmill; however, treadmill walking was more demanding in terms of metabolic cost. The higher demands associated with treadmill walking may increase the challenge for stroke subjects, which could be particularly relevant since these individuals generally have low exercise activity tolerance.

In combination, the findings of this study suggest that for independent ambulators following hemiparetic stroke, overground walking is less energy demanding than unsupported treadmill walking, despite greater kinematic asymmetry. It could be that if given the option, subjects may have self-selected a slower walking speed on the treadmill than their preferred overground speed as reported by Bayat et al., 2005. Metabolic cost is lowest at the preferred speed of walking (Gordon et al., 2009; Mian et al., 2006), so forcing our subjects to match their overground walking speed may have been suboptimal resulting in an elevated energy cost. The similarity in stride and cadence the two conditions however, makes the higher cost of treadmill walking unexpected.

This study has clearly illustrated that people with stroke can be challenged to walk faster on a treadmill and can meet the metabolic demands without compromise to their walking pattern. There is evidence to suggest that training at higher than natural speed translates into improvements in overground walking speed (Pohl et al., 2002) although it is not known whether there is any transference in walking pattern. It may be that once the constraints of the constant belt speed are removed, that the body recalibrates to perform
optimally in the overground condition suggesting very specific training induced adaptations. Nonetheless, treadmill training in stroke has proven very effective at promoting cardiovascular fitness and gains in strength (Macko et al., 2001), which are determinants of walking ability. For clinicians, it would be valuable to explore the effects of extended treadmill gait retraining on the biomechanics of overground walking in order to fully appreciate the extent any carry-over effects.

REFERENCES


72


CHAPTER 4

Biomechanics and Metabolic Costs of Overground and Treadmill Walking:
Comparisons between Healthy Adults and People with Stroke and Changes with Increasing Speed on the Treadmill

Abstract

Background: In promoting faster than self-selected walking speed in people with stroke, it is not clear whether the speed related adjustments in kinematic, kinetic and metabolic measures are similar to those of healthy individuals. This is important because differences could reflect an inability to adapt to the challenge and may adversely affect movement patterns and endurance. Methods: Secondary analyses of biomechanical and metabolic data from ten healthy subjects and ten similarly aged stroke subjects were conducted to compare overground and treadmill walking across groups and to explore the means by which each group increased their speed of walking on the treadmill. Results: Stroke subjects walked with lower cadence, shorter strides, lower stance time and smaller lower limb joint excursions than their healthy counterparts. All temporal-distance parameters, hip joint excursion, F1 and F2 forces and metabolic cost showed main effects of speed. An interaction between speed and group indicated that oxygen consumption increased at a greater rate in stroke than healthy subjects. Conclusions: Although there are differences in gait patterns between healthy and stroke subjects, in general both groups responded to the challenge of increased walking speed in the same way. One important difference is the abnormal elevation of energy demands associated with treadmill walking at faster speeds in stroke. Clinically, this warrants consideration as it may lead to premature fatigue and undesirable cardiorespiratory challenge in this group of individuals.
4.1 Introduction

Compared to healthy subjects, stroke gait has been characterized by reduced speed and cadence, asymmetry in temporal-distance parameters, asymmetry in kinematic and kinetic variables, and increased metabolic costs. Clinicians involved in retraining gait patterns in stroke have repeatedly stressed the importance of modifying these variables, specifically increasing gait speed and symmetry not only to improve the overall locomotor performance, but also to promote gait efficiency and functional independence (Barbeau and Visintin, 2003; Eich et al., 2004; Hesse et al., 1994, 1995, 1999; Macko et al., 1997a; Pang and Eng, 2006; Visintin et al., 1998; Waagfjord et al., 1990). To achieve this, treadmills are being used to retrain walking in stroke because variables such as walking speed can be controlled and most importantly, the amount of assistance provided and the environmental factors can be adapted to the individual’s abilities. Because of these advantages, treadmills are currently being used in a wide variety of settings including clinical and research settings, at public fitness centres and in homes not only for recreation but also to help improve functional capacity and to retrain walking in people with abnormal gait patterns.

Walking speed in subjects with stroke is a global indicator of gait performance and self-perceived physical function (Luukinen et al., 1995). It is also positively related to functional independence (Potter et al., 1995) and social activity (Cwikkel et al., 1995), hence efforts to improve walking speed. Providing task oriented repetitive practice, treadmill gait training has been shown to improve walking capability (Barbeau and Visintin, 2003; Hesse et al., 1994, 1995, 1999; Visintin et al., 1998; Waagfjord et al., 1990) and aerobic capacity (Eich et al., 2004; Macko et al., 1997a; Pang and Eng, 2006)
in people with stroke. Furthermore, increasing the belt speed provides added benefit in terms of walking capacity (Pohl et al., 2002; Silver et al., 2000).

There is limited information quantifying the effect of promoting faster than self-selected walking speed on gait parameters in subjects with stroke (Hesse et al., 1999). In Chapter 3 the findings indicate that stroke subjects were able to walk 20% faster than self-selected overground speed, but it is not clear whether the kinematic, kinetic and metabolic adjustments made are similar to how healthy individuals adapt. This is important to know because any differences could reflect as inability to adapt to the challenge and may adversely affect movement patterns and energy cost, which could be undesirable in this group.

4.1.1 Temporal and Distance Parameters

In overground walking, people with stroke have slower speeds, lower cadence, and smaller step lengths than healthy adults. Further, the step length and stance time are lower on the affected side compared to the unaffected side resulting in marked asymmetry (Nakamura et al., 1988; Roth et al., 1997; Wall and Turnbull, 1986; see also Chapter 3). Generally, in healthy individuals temporal-distance parameters are comparable for both limbs (see also Chapter 2). This is also the case when walking on a treadmill suggesting that they adapt easily to the walking surface (Alton et al., 1998). In contrast, the asymmetries evident during overground walking in stroke subjects become much less on the treadmill (Harris-Love et al., 2001; Hesse et al., 1999; see also Chapter 3). In combination these findings suggest that people with stroke adapt differently than healthy subjects. The extent to which this may be true is explored in this study.
Stride length and cadence have been reported to be co-dependent on each other with regards to increases in walking speed in healthy adults (Winter, 1991) and in stroke (Bohannon, 1992; Olney et al., 1994; Roth et al., 1997). Specifically there is a trade-off between the two parameters such that speed can be increased by higher cadence and shorter strides, or lower cadence and longer strides. Healthy individuals can select either option to increase walking speed. Such redundancy in control might be limited in people with stroke as a result of hemiparesis, incoordination, and other physical impairments that limit the repertoire of possible solutions to the challenge of walking faster. For example, walking speed was found to have a stronger correlation with the stance and swing phases on the unaffected side in stroke victims compared to the affected side (Olney et al., 1994; Roth et al., 1997). This could be because the ability to alter parameters on the affected side is limited. With the improvement in gait symmetry during treadmill walking in subjects with stroke (see Chapter 3), these individuals may have few remaining physical adaptations or compensations at their disposal to increase walking speed, particularly on the affected side. Understanding these adaptations relative to the norm healthy adults would provide valuable information to clinicians and rehabilitation specialists.

4.1.2 Kinematics

In healthy adults, at self-selected overground walking speeds, the hip normally extends from about 18° (SD 7°) of flexion at heel strike to 11° (SD 8°) of extension occurring near the end of stance phase; the knee flexes from about 4° (SD 5°) at the beginning of stance to about 35° (SD 6°) at the end of stance; the ankle plantarflexes from about 9° (SD 4°) of dorsiflexion at the beginning of stance to about 18° (SD 5°) of plantarflexion at the end of stance (Winter, 1991). In contrast, stroke subjects have
smaller amplitude total hip, knee and ankle joint excursions and the magnitude of the difference between healthy adults and stroke has been related to the slower walking speed adopted by subjects with stroke (Kerrigan et al., 1999; Moseley et al., 1993; Olney et al., 1994; Olney and Richards, 1996a,b). At initial contact people with stroke characteristically show less hip flexion, more knee flexion and less ankle dorsiflexion on the affected side than healthy adults. In addition less hip and knee flexion and more ankle plantarflexion is evident at midstance followed by less hip extension, knee flexion and ankle plantarflexion at push-off (Olney et al., 1994; Olney and Richards, 1996a,b).

With cadence increases of 20% in overground walking speed healthy adults show average increases in hip, knee and ankle joint excursions of 2.3°, 3.3° and 2.8° respectively (Winter, 1991). On the treadmill increases in hip and ankle joint excursions of 5° were observed when healthy adults walked 20% faster than their self-selected overground walking speed (Chapter 2), whereas stroke subjects increased their hip and knee joint excursions by 5° respectively on their affected side (Chapter 3) with little adjustment of ankle excursion. Considering that the ankle plantarflexors are generally weak following stroke (Nadeau et al., 1999; Olney et al., 1994), the capacity to increase the ankle’s dynamic range of motion may be compromised requiring compensations elsewhere.

4.1.3 Kinetics

In healthy adults the characteristic ‘butterfly’ profile of the vertical ground reaction force shows an initial peak greater than body weight after foot contact (F1), a valley (below body weight) during midstance (F2), and a second peak at push-off (F3). At comparable overground and treadmill walking speeds Yack et al. (1995) reported a
similar mean F1 force but the F2 force was about 9.7% higher on the treadmill attributed to a lower vertical acceleration of the centre of gravity secondary to the constant velocity of the belt. Push-off (F3) forces were lower during treadmill walking compared to overground (Riley et al., 2007; White et al., 1998; Yack et al., 1995; see Chapter 2) because of decreased push-off by the ankle plantarflexors. In stroke subjects, the F1 forces are about 10% lower during treadmill walking on both the unaffected and the affected sides when compared with overground walking at similar speeds (Harris-Love et al., 2001). In Chapter 3 lower F1 forces were reported on the unaffected side and F3 forces were lower bilaterally when walking on the treadmill.

To walk faster healthy adults use a combination of increased power generation at the ankle and hip (Winter, 1991). The relative contribution to the work of walking faster is variable as a result of multiple normal dynamic strategies that can be employed (Vardaxis et al. 1998). In stroke, recruitment of the hip flexor muscles was required to achieve faster walking speeds in the presence of plantarflexor weakness (Chen et al., 1997; Nadeau et al., 1999; Olney et al., 1994; Parvataneni et al., 2007), whereas others without marked paresis primarily used their plantarflexors in combination with the hip flexors and extensors (Winter, 1991). The strategies available depend on the severity and distribution of impairments. Regardless, the impact of the kinetic strategies used to increase walking speed could be evaluated by measuring temporal-distance parameters, kinematics, and most importantly the metabolic costs of walking.

4.1.4 Metabolic Costs of Walking

In healthy subjects the metabolic costs of walking are largely dependent on speed. Walking is most energy efficient at an optimal walking speed (about 1.3m/s) and the
metabolic demands increase as walking speed increases or decreases from the optimal speed (Bernardi et al., 1999; Holt et al., 1991; Martin et al., 1992; Pearce et al., 1983; Waters et al., 1988). In stroke, it is still quite unclear how self-selected walking speed relates to the metabolic costs as the metabolic demands are variable and are dependent on the degree of impairment, the stage of recovery and the protocol used (da Cunha-Filho et al., 2003). At similar speeds of walking, stroke subjects have been reported to have metabolic costs of about 1.5-2 times greater than healthy adults (Macko et al., 1997a).

Heart rate provides an estimate of the metabolic cost of walking and is reportedly higher for treadmill than overground walking in young and older adults (Greig et al., 1993; Murray et al., 1985; see Chapter 2) as well as in stroke (see Chapter 3). Oxygen consumption showed the same trend (Chapters 2 and 3). Whether healthy adults and those with stroke show similar, proportional increases in metabolic demand with increased speeds of walking on the treadmill is not known. This is especially important in stroke subjects since cardiovascular co-morbidities that have an effect on O2 demands of the body and the overall metabolic cost are quite common (Macko et al., 1997a,b; Roth et al., 1993). Furthermore, as treadmill gait training for people with stroke becomes more widespread, a clear understanding of the impact of speed on metabolic cost is warranted.

4.2 Purpose

Because walking speed is of primary importance to the person’s function and independence, the means by which gait speed is increased is of considerable interest. Although there is a substantial body of literature describing changes in overground gait with speed in normal adults, there is very little for treadmill walking. Limited research was found that describes modifications in treadmill gait with increasing speeds for
persons with stroke. The present study is a secondary analysis undertaken to compare the gait of subjects with stroke with normal adults at self-selected speeds both overground and on a treadmill, and to explore and describe the means by which each group increases their speed of walking on a treadmill. Simultaneous collection of kinematic, kinetic and metabolic data permits this issue to be appropriately addressed. Specifically analyses were conducted to:

- Compare the kinematic, kinetic and metabolic profiles of healthy subjects and people with stroke during overground walking at self-selected speed and treadmill walking at matched speed; and
- To explore the kinematic, kinetic and metabolic changes associated with increases in belt speed while walking on a treadmill and whether they differ in healthy individuals and those with stroke.

4.3 Methodology

This study used the data collected from two groups of subjects (healthy and stroke) under identical laboratory test conditions. The details of the subject recruitment and the protocol followed are provided in the methods sections of Chapters 2 and 3.

4.4 Data Analysis

Descriptive statistics (mean ± SD) were calculated for all outcome measures. Statistical analyses were performed using SPSS (Version 14.0) and a significance level of p < 0.05 was adopted. Analyses of variance were performed to determine if there were differences in dependent measures as a function of group (healthy and stroke), condition (overground and treadmill walking at matched speeds), and interactions of both. Similar analyses were performed to determine if there were main effects of speed and group on
the percentage change observed in dependent measures as belt speed was increased 10% and 20% over the matched speed. Interactions between group and speed were also examined. Percent change was calculated by subtracting the matched speed value from the faster speed value (10% or 20%), dividing by the matched speed value and multiplying by 100. The biomechanical data obtained from the unaffected and affected sides of the body of stroke subjects were compared to the data from healthy subjects in separate ANOVAs.

4.5 Results

Ten healthy subjects (5 males and 5 females) with a mean age of 60.6 ± 7.4 years (range 50-73 years) and ten similarly aged stroke subjects (6 males and 4 females) with a mean age of 64.9 ± 9.7 years (range 54-80 years; p = 0.764) completed the protocol without incident. The average time since stroke was 5.9 ± 4.1 years ranging from 1.5 to 15 years; five subjects had right hemiparesis and 5 had left hemiparesis. All subjects could walk independently without any personal assistance or a walking aid.

4.5.1 Gait Speed and Cadence

The descriptive data for gait speed and cadence during overground and treadmill walking and the percentage change with increasing treadmill belt speeds for all measures are listed in Table 4.1.

The average self-selected overground walking speed and the matched speed on the treadmill were significantly faster for healthy subjects than stroke subjects (p=0.006). The cadence was also higher for healthy subjects than stroke for both walking surfaces (p=0.025). With increased walking speeds cadence increased in a linear manner for all subjects reflecting a main effect of speed (p=0.001).
4.5.2 Temporal and Distance Parameters

The descriptive data for the temporal distance parameters during overground and treadmill walking and the percentage change with increasing walking speeds on the treadmill are listed in Table 4.1.

The gait cycle time was similar for healthy and stroke subjects during overground (p=0.069) and treadmill (p=0.101) walking. Compared to healthy subjects at self-selected overground walking speed, the stance time and the percentage of time spent in stance on the unaffected side (p=0.045), the swing time and the percentage of time spent in swing on the affected side (p=0.020) were significantly higher. During treadmill walking, stance time and the percentage of time spent in stance were lower for stroke subjects on both the unaffected (p=0.007) and the affected sides (p=0.001). Swing time and the percentage of time spent in swing was correspondingly higher (p<0.007). Stroke subjects had a significantly shorter stride length (p=0.044) and step length on the affected side (p=0.011) during overground walking than healthy adults.

Stance time on the affected side of stroke subjects and healthy adults decreased proportionately with speed by 5% and 6%, respectively (p=0.006). The percentage of time spent during stance showed an interaction effect reflecting a much greater reduction on the unaffected side in stroke than was observed in healthy adults (-3.35% versus -1.97%, respectively (p=0.005). Main effects of speed were found for stride length (p<0.001) and step length (p<0.001). No group effects were evident on any measure (p>0.060) although an interaction was found for step length which increased by 20% on the affected side in stroke but only 16% in healthy subjects.
Table 4.1  Mean values (± 1SD) of temporal distance parameters for overground and treadmill walking (matched speeds) and the percentage change associated with increased treadmill speed

<table>
<thead>
<tr>
<th>Variable</th>
<th>Overground (Self-selected Walking Speed)</th>
<th>Treadmill (Matched Speed)</th>
<th>% Change (matched + 10%)</th>
<th>% Change (matched + 20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Healthy</td>
<td>Stroke</td>
<td>Healthy</td>
<td>Stroke</td>
</tr>
<tr>
<td>Speed (m/s)</td>
<td>1.15 (±0.15)</td>
<td>0.81* (±0.16)</td>
<td>1.15 (±0.14)</td>
<td>0.81* (±0.15)</td>
</tr>
<tr>
<td>Cadence (steps/min)</td>
<td>109 (±14)</td>
<td>87* (±11)</td>
<td>110 (±20)</td>
<td>88* (±11)</td>
</tr>
<tr>
<td>Cycle Time (s)</td>
<td>1.13 (±0.14)</td>
<td>1.29 (±0.20)</td>
<td>1.12 (±0.19)</td>
<td>1.30 (±0.22)</td>
</tr>
<tr>
<td>Stride Length (m)</td>
<td>1.27 (±0.11)</td>
<td>1.16* (±0.17)</td>
<td>1.22 (±0.15)</td>
<td>1.08 (±0.22)</td>
</tr>
</tbody>
</table>

| Stance Time (s)   | 0.72 (±0.10)                             | 0.87 (±0.16)              | 0.82 (±0.14)            | 0.70 (±0.13)            | 0.71 (±0.14)             | 0.65 (±0.09)             | -2.69† (±4.86)           | -3.70† (±6.77)           | -0.18 (±6.96)           | -6.45† (±6.01)           | -4.00† (±8.21)           | -5.36 (±6.15)           |
| Stance %          | 64.39 (±1.81)                            | 66.32* (±3.14)            | 63.33 (±3.19)           | 62.89 (±1.49)           | 54.71* (±7.05)           | 50.76* (±8.03)           | -0.28† (±2.73)           | -1.97† (±3.55)           | -3.62† (±6.06)           | -1.97† (±2.90)           | -3.35† (±7.69)           | -1.68† (±10.73)         |
| Step Length (m)   | 0.63 (±0.06)                             | 0.58 (±0.10)              | 0.56* (±0.09)           | 0.61 (±0.07)            | 0.54 (±0.12)             | 0.53 (±0.10)             | 9.48† (±3.21)            | 7.33† (±6.53)            | 10.42† (±8.18)           | 16.19† (±4.42)           | 18.38† (±9.48)           | 20.01† (±8.90)          |

*Indicates significantly different from healthy adults for the same walking surface (p<0.05)
† Indicates a main effect of treadmill speed (p<0.05)
Interaction effect of group (healthy and stroke) and treadmill speed are bolded to show different response among stroke subjects compared with healthy subjects
4.5.3 Kinematic Data

Table 4.2 summarizes the angular excursions of the hip, knee and ankle joints during stance for healthy and stroke subjects and the percentage change with increased belt speeds.

During overground walking at self-selected speed, there were no differences in the hip or knee range of motion on the unaffected side when compared to healthy subjects, but the total ankle range of motion was significantly less (p=0.019). On the affected side in stroke, the hip (p=0.022), knee (p=0.023) and ankle (p=0.001) excursions were less than in healthy subjects.

Main effects of treadmill speed were found for hip range of motion which increased by 11% in healthy adults and 12% on the unaffected and 14% on the affected sides for stroke subjects (p<0.023) with 20% speed increases. No main effects of group or interaction effects were found for other joint excursions.
Table 4.2  Mean values (± 1SD) of joint excursions for overground and treadmill walking (matched speeds) and the percentage change associated with increased treadmill speed

<table>
<thead>
<tr>
<th>Variable</th>
<th>Overground (Self-selected walking Speed)</th>
<th>Treadmill (Matched Speed)</th>
<th>% Change (matched + 10%)</th>
<th>% Change (matched + 20%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Healthy</td>
<td>Unaffec</td>
<td>Affec</td>
<td>Healthy</td>
</tr>
<tr>
<td>Hip Excursion</td>
<td>34 (±4)</td>
<td>32 (±5)</td>
<td>28* (±7)</td>
<td>35 (±7)</td>
</tr>
<tr>
<td>Knee Excursion</td>
<td>46 (±3)</td>
<td>42 (±6)</td>
<td>39* (±8)</td>
<td>46 (±7)</td>
</tr>
<tr>
<td>Ankle Excursion</td>
<td>23 (±3)</td>
<td>19* (±5)</td>
<td>16* (±4)</td>
<td>25 (±8)</td>
</tr>
</tbody>
</table>

*Indicates significantly different from healthy adults for the same walking surface (p<0.05)
† Indicates a main effect of treadmill speed (p<0.05)
Interaction effect of group (healthy and stroke) and treadmill speed are bolded to show different response among stroke subjects compared with healthy subjects
4.5.4 Kinetic Data

The average peak maximal vertical ground reaction forces at weight acceptance (F1) and push-off (F3), and minimum force values in midstance (F2) are shown for both subject groups (unaffected and affected limbs for stroke) in Table 4.3. The percentage changes with increased treadmill belt speeds are also presented.

The peak F1 and F3 vertical ground reaction force values were similar for healthy subjects and those with stroke (unaffected and affected sides), $p>0.111$. The midstance (F2) forces were higher on the unaffected ($p=0.027$) and the affected sides ($p=0.021$) than those recorded for healthy adults during overground walking only.

Main effects of treadmill speed were evident for the changes in magnitude of F1 and F2 forces ($p \leq 0.008$) when considering the unaffected side in stroke. Changes in F2 magnitudes showed an interaction effect ($p=0.001$) indicating that the degree to which the F2 forces decreased at higher speeds was greater in healthy subjects than on the affected side in stroke. No main effects or interaction effects were found in association with F3.
Table 4.3  Mean (± 1SD) vertical ground reaction forces expressed as a percentage of body weight and the percentage change in the forces from treadmill walking at matched speed

<table>
<thead>
<tr>
<th>Variable</th>
<th>Healthy</th>
<th>Overground</th>
<th>Unaffec</th>
<th>Affec</th>
<th>Healthy</th>
<th>Treadmill</th>
<th>Unaffec</th>
<th>Affec</th>
<th>% Change (matched + 10%)</th>
<th>Healthy</th>
<th>Unaffec</th>
<th>Affec</th>
<th>% Change (matched + 20%)</th>
<th>Healthy</th>
<th>Unaffec</th>
<th>Affec</th>
</tr>
</thead>
<tbody>
<tr>
<td>F1</td>
<td>1.09</td>
<td>1.07</td>
<td>1.03</td>
<td>1.06</td>
<td>1.01</td>
<td>1.02</td>
<td></td>
<td></td>
<td>2.29†</td>
<td>3.38†</td>
<td>5.26</td>
<td></td>
<td>5.53†</td>
<td>8.25†</td>
<td></td>
<td>3.45</td>
</tr>
<tr>
<td></td>
<td>(±0.06)</td>
<td>(±0.06)</td>
<td>(±0.08)</td>
<td>(±0.08)</td>
<td>(±0.08)</td>
<td>(±0.07)</td>
<td></td>
<td></td>
<td>(±3.95)</td>
<td>(±6.37)</td>
<td>(±11.87)</td>
<td></td>
<td>(±5.67)</td>
<td>(±12.59)</td>
<td></td>
<td>(±12.97)</td>
</tr>
<tr>
<td>F2</td>
<td>0.75</td>
<td>0.91*</td>
<td>0.89†</td>
<td>0.74</td>
<td>0.88</td>
<td>0.87</td>
<td></td>
<td></td>
<td>-5.32†</td>
<td>-3.60†</td>
<td>-3.98</td>
<td></td>
<td>-10.54†</td>
<td>-8.07†</td>
<td></td>
<td>-6.65</td>
</tr>
<tr>
<td></td>
<td>(±0.09)</td>
<td>(±0.13)</td>
<td>(±0.12)</td>
<td>(±0.13)</td>
<td>(±0.08)</td>
<td>(±0.12)</td>
<td></td>
<td></td>
<td>(±3.62)</td>
<td>(±2.80)</td>
<td>(±5.31)</td>
<td></td>
<td>(±6.45)</td>
<td>(±6.18)</td>
<td></td>
<td>(±6.66)</td>
</tr>
<tr>
<td>F3</td>
<td>1.06</td>
<td>1.08</td>
<td>1.03</td>
<td>0.98</td>
<td>0.99</td>
<td>0.95</td>
<td></td>
<td></td>
<td>2.82</td>
<td>2.74</td>
<td>1.22</td>
<td></td>
<td>4.22</td>
<td>3.41</td>
<td></td>
<td>0.43</td>
</tr>
<tr>
<td></td>
<td>(±0.05)</td>
<td>(±0.07)</td>
<td>(±0.09)</td>
<td>(±0.07)</td>
<td>(±0.07)</td>
<td>(±0.07)</td>
<td></td>
<td></td>
<td>(±2.66)</td>
<td>(±3.75)</td>
<td>(±4.00)</td>
<td></td>
<td>(±2.52)</td>
<td>(±5.14)</td>
<td></td>
<td>(±5.64)</td>
</tr>
</tbody>
</table>

*Indicates significantly different from healthy adults for the same walking surface (p<0.05)
† Indicates a main effect of treadmill speed (p<0.05)
Interaction effect of group (healthy and stroke) and treadmill speed are bolded to show different response among stroke subjects compared with healthy subjects
4.5.5 Metabolic Data

The metabolic data for overground and treadmill walking are summarized in Table 4.4 and the percent changes associated with increased belt speeds are also presented.

Resting metabolic data (HR and VO$_2$) were comparable prior to beginning the walking trials on both surfaces (p=0.732). Compared to healthy adults, VO$_2$ consumed by stroke subjects was lower during overground walking (p=0.035) as well as during treadmill walking (p=0.005). The recovery period (i.e. time for HR and VO$_2$ to return to resting values), however, was significantly longer for stroke subjects for both walking surfaces (p<0.004).

Increasing walking speeds on the treadmill had a significant impact on HR and VO$_2$ (main effects of speed, p<0.001) for both groups. In the case of oxygen uptake, an interaction effect indicated that stroke subjects increased their consumption to a much greater degree than their healthy counterparts (31% versus 17%, respectively), p=0.011.
Table 4.4  Metabolic data (mean ± 1SD) at matched overground and treadmill walking speeds and the percentage change from treadmill walking at matched speed for healthy and stroke subjects

<table>
<thead>
<tr>
<th></th>
<th>Overground</th>
<th>Treadmill</th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>Rest</td>
<td>Rest</td>
<td>% Change</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Steady state</td>
<td>Steady state</td>
<td>(matched</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>+10%)</td>
</tr>
<tr>
<td>Healthy Adults</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Heart Rate (beats/min)</td>
<td>75 (±10)</td>
<td>85 (±10)</td>
<td>73 (±10)</td>
<td>90 (±16)</td>
<td>6.43†</td>
</tr>
<tr>
<td>Oxygen Consumption (ml/kg/min)</td>
<td>4.03 (±0.53)</td>
<td>11.39 (±1.10)</td>
<td>4.78 (±1.03)</td>
<td>13.98*</td>
<td>11.15†</td>
</tr>
<tr>
<td>Recovery Time (min)</td>
<td>2.60 (±0.97)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stroke Subjects</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Heart Rate (beats/min)</td>
<td>73 (±8)</td>
<td>87 (±11)</td>
<td>74 (±9)</td>
<td>92* (±14)</td>
<td>7.61†</td>
</tr>
<tr>
<td>Oxygen Consumption (ml/kg/min)</td>
<td>3.87 (±0.16)</td>
<td>9.58* (±2.69)</td>
<td>3.93 (±0.16)</td>
<td>10.91*</td>
<td>14.79†</td>
</tr>
<tr>
<td>Recovery Time (min)</td>
<td>3.90 (±0.74)</td>
<td></td>
<td></td>
<td></td>
<td>6.50* (±1.51)</td>
</tr>
</tbody>
</table>

*Indicates significantly different from healthy adults (p<0.05)
† Indicates main effect of treadmill speed (p<0.05)
Interaction effect of group (healthy and stroke) and treadmill speed are bolded to show different response among stroke subjects compared with healthy subjects.
4.6 Discussion

Overground and treadmill walking were compared between healthy adults and those with stroke revealing differences in temporal-distance parameters and kinematics while kinetic parameters were similar for both surfaces of walking. Main effects of increasing the treadmill belt speed were found for cadence, gait cycle, step and stride length, and hip joint excursions. With increasing walking speed, vertical ground reaction forces (F1) increased in the healthy adults and only on the unaffected side in stroke and F2 forces decreased. The magnitude of the decrease in F2 forces was greater in healthy subjects than on the unaffected and affected sides in stroke. Compared to overground walking the metabolic cost was higher for both groups of subjects when walking on the treadmill, and higher in healthy adults compared to stroke for both conditions. With increased belt speed, stroke subjects showed a much greater increase in oxygen consumption than healthy subjects.

In relation to their healthy counterparts, the lower average cadence of stroke subjects, increased percentage of time spent in unaffected stance and affected swing, decreased percentage of time spent in unaffected swing and affected stance during both modes of walking is in accordance with other studies (Bohannon, 1987, 1992; Olney et al., 1994; Roth et al., 1997; Teixeira-Salmela et al., 1999; Wall and Turnbull, 1986). The main explanation for these differences is the fact that people with stroke walk considerably more slowly than healthy adults of similar age as was the case in the current study.

With increases in speed there was a predictable difference in the way this was accomplished in subjects with stroke. Both groups increased their cadence and stride
length and similar results have been reported by Bayat et al., 2005. In addition, stroke subjects were able to increase their step length on the affected side more than on the unaffected side, likely as a result of the higher work output of the unaffected standing leg. Even though increasing speed is most readily accomplished by increased cadence leaving stride length comparatively stable, the subjects chose to increase both cadence and stride length. Increasing cadence alone has resulted in perceived instability in older adults (Turnbull et al., 1995), but as both groups of subjects chose to increase cadence and stride length, instability on the treadmill may not have been an issue. The fact that subjects were able to adapt to the increased belt speed while maintaining their overall gait patterns suggests that any alteration in perceived stability was probably minimal.

At self-selected speed the kinematic profiles of stroke subjects showed smaller total joint excursions at the hip, knee and ankle on the affected side when compared with healthy subjects for overground walking. These findings are consistent with previous studies (Knutsson and Richards, 1979; Lehman et al., 1987; Olney et al., 1989, 1991, 1994) and reflect the dominance of the unaffected side. On the treadmill, however, people with stroke showed greater symmetry in joint excursions which led to the normalization of kinematic patterns. Only the hip excursion from the unaffected side and the ankle excursion from the affected sides differed significantly from the healthy adults when walking on the treadmill.

Increasing walking speed on the treadmill was associated with increased hip excursions of about 40°-50° in both groups of subjects. Unlike reports from overground walking in which faster speeds are accomplished by smaller (~2°-3°) increases in all lower limb joint excursions (Winter, 1991), the present study revealed adaptations that
were restricted to the hip. It could be that during overground walking subjects would attempt to increase their speed by increasing ankle plantarflexion push-off power. The benefit of using this strategy on the treadmill would likely be limited because part of the push-off force would contribute to the work performed on the belt rather than on limb propulsion (Parvataneni et al., 2009; White et al., 1998; see also Chapter 2).

In terms of VGRF, the magnitudes of F1 and F3 forces were comparable for healthy and stroke subjects for overground and treadmill conditions. The F2 forces, however, were higher on the unaffected and affected sides for stroke subjects compared to healthy subjects during overground walking, but were similar on the treadmill. In overground walking the stroke subjects were able to adjust their speed throughout a gait cycle which could minimize fluctuations in acceleration of the centre of mass thus flattening the typical midstance trough in the VGRF profile. During treadmill walking the speed is constant and dictated by the belt, it follows that with this constraint the midstance deceleration of the center of gravity was faster resulting in higher F2 forces during treadmill walking which more closely approximated the normal pattern.

The magnitudes of F1 and F3 peaks are known to increase as a function of walking speed (Keller et al., 1996; Winter, 1991). These trends were observed in healthy and stroke subjects as belt speed was increased, however significance was limited to the F1 peak only. The absence of speed related increases in F3 has been reported elsewhere for stroke (Nadeau et al., 1999; Olney et al., 1994) although the fact that the magnitudes failed to reach the levels attained during natural speed overground walking for either healthy or stroke subjects is unexpected. The most reasonable explanation is that the finding is an artefact of treadmill walking. The moving belt likely attenuates the vertical
force by directing it rearward thus contributing to the belt movement. Evaluation of the instantaneous belt speed indicated fluctuations of up to 4% from the mean belt speed which supports this view (Parvataneni et al., 2009; White et al., 1998; see also Chapter 2). The speed related kinematic changes that were observed occurred only at the hip and not the ankle, which supports this argument. The similarity in the VGRF across speeds offers little insight into the mechanisms underlying the temporal-distal differences observed between groups as they increase their walking speeds. More detailed kinetic information may be required to better understand alterations in gait with increased speed.

No studies to date have reported detailed biomechanics of gait combined with an evaluation of the metabolic cost. At natural speed walking HR was similar for both groups although the average VO$_2$ consumption in stroke was lower than in healthy subjects. Essentially then, the heart is working as hard to provide less oxygen to the working muscles. The requirement of less oxygen makes sense considering that stroke subjects walked slower and performed less work; however the finding that heart rates of stroke subjects were comparable to healthy controls may be attributable to physical deconditioning (Mackay-Lyons and Howlett, 2002; Mackay-Lyons and Makrides, 2005) or is symptomatic of cardiovascular disease which affects more than 75% of stroke survivors (Roth et al, 1993). In either case, the poor cardiovascular fitness may leave little reserve capacity to meet the activity demands that require them to walk faster thereby resulting in greater than expected energy costs. With faster walking speeds on the treadmill, HR increased linearly in both groups. In contrast, VO$_2$ increased by about 11% when speed was increased by 10% then the rate of increase slowed to 17% at 20% above self-selected walking speed in healthy adults. In stroke, the VO$_2$ response was more marked, increasing
15% and 31% at speed increments of 10% and 20%, respectively. This implies that with similar increases in demands of a walking task for both groups of subjects, stroke subjects accomplish the task at a much greater increase in costs. The much longer recovery period required by stroke subjects compared to healthy subjects is consistent with this viewpoint. Considering that the metabolic cost reflects the oxygen required by the working muscles it is reasonable to expect that such a dramatic increase as that observed in stroke would be accompanied by biomechanical adaptations.

The sagittal plane biomechanical profiles reported in this study do not appear to show differences that are sufficiently large to explain the increased metabolic cost observed in stroke with increasing speed. It is possible that kinematic differences in the frontal plane between healthy adults and stroke subjects may have contributed to the different metabolic demands. Although there have been no studies reporting three dimensional angular kinematics for overground and treadmill walking in stroke subjects, two studies involving healthy adults concluded that there were no differences between treadmill and overground walking in either the sagittal or frontal planes (Lee and Hidler, 2008; Riley et al., 2007). Indeed the only speed related kinematic change observed in the current study was an increase in hip joint excursion suggesting that adaptations are made more easily at the hip. Graf et al. (2005) reported average increases in peak powers between comfortable and fast walking to be 23% for plantarflexors, 54% for hip flexors and 67% for hip extensors. Although effective for increasing gait speed, this strategy would be associated with a high energy cost (Collen et al., 1990; Kuo 2002). Using a powered bipedal walking model, Kuo (2002) calculated that with no ankle push-off and the hip fully compensating, such a pattern yielded a four-fold increase in the mechanical cost. The increases in hip excursion for both
groups of subjects in this study were between 11% and 14% for healthy adults and the unaffected and affected sides of the stroke subjects. The higher metabolic cost of treadmill walking may be a reflection of the reliance on the use of this strategy, and more so in stroke subjects who have larger ankle push-off deficits on the affected side (Nadeau et al., 1999; Parvataneni et al., 2007).

Ogliati et al. (1988) suggested that the metabolic cost of walking increases in subjects with locomotor impairment as a result of spastic co-contraction of agonist-antagonist muscles and an inefficiency of body kinematics resulting in wasted mechanical energy. If this were the case for the stroke subjects in this study, the energy costs would have been higher than that of healthy adults. The slower speed of walking; however, may have masked such an effect. Walking 20% faster than their self selected speed stroke subjects averaged 0.99m/s and consumed 13.95 ml/kg/min of O₂ with a HR of 104 beats/min. The speed corresponds to only 86% of the self selected speed of healthy subjects, whereas O₂ consumption and HR correspond to 100% and 115% of that associated with healthy subjects who walked at 1.15 m/s. Although abnormal muscle activation patterns cannot be discarded as an explanation for the metabolic findings, more detailed studies involving electromyography are required.

4.7 Limitations

There were some limitations that may have reduced the power to detect differences between groups. The sample size for the subject groups was quite small and the stroke subjects were quite heterogenous having had their strokes between 1.5 to 15 years prior to the study. However, examination of the standard deviations shows that although a few variables show substantially greater variability for stroke subjects and
values for the affected side tend to be marginally higher than the unaffected side, overall they compared well with healthy subjects. Because of these two factors, the differences that were detected are almost certainly important ones, but similar studies involving more subjects with greater homogeneity may be warranted.

There were factors that were not controlled resulting in different test conditions for the two groups. The gait speed was different between groups which may have explained the observed differences in temporal-distance parameters and metabolic variables. The speed differential seems to have affected joint kinematics much less; in many cases the unaffected side kinematics were very similar to those of healthy subjects and differences were much more frequent on the affected side.

Concerning generalizability, the shortest time since stroke occurrence was 1.5 years and the average was nearly 6 years. Also, these subjects had walking speeds comparable to fast walking stroke subjects (Olney et al., 1991). As a result, the findings may apply only to faster walking individuals showing chronic effects of stroke, and may be different for slower walking chronic stroke subjects, or those with more acute lesions.

Finally the gait data that were analyzed were limited to the sagittal plane. Although most movement occurs in the plane of progression it is possible that additional data from other movement planes may have presented a different picture and interpretation. The addition of electromyographic measurements would also be of value.

4.8 Conclusion

The present study provides a comprehensive evaluation of overground and treadmill walking in healthy adults and subjects with stroke. Differences in temporal-distance parameters, kinematic variables and metabolic costs were noted although kinetic
data were similar. Although the heart rates increased proportionately for both groups, the percentage increase in metabolic cost at faster walking speeds on the treadmill was greater in subjects with stroke than for healthy adults. This could be a clinically important finding as the elevated energy demands of treadmill walking at faster speeds combined with oxygen debt may result in premature fatigue and undesirable cardio respiratory challenge which may require consideration of its suitability as a training tool especially in subjects with cardiovascular disease such as stroke. Further study is warranted to determine whether the metabolic differences between the two modes of walking in healthy adults and stroke subjects diminishes over time as subjects became more familiar with treadmill walking.

REFERENCES


CHAPTER 5

General Discussion and Conclusions

Though numerous studies have compared overground and treadmill walking there still exists a significant debate with regards to the differences between the two modes of walking. The present study adds to what is known by providing a comprehensive evaluation of overground and treadmill walking at matched speeds in healthy adults, in subjects with stroke and the comparisons between the two groups. From the three studies completed, the following main conclusions are drawn:

- For healthy adults, overground and treadmill walking were similar in terms of temporal-distance, kinematic and kinetic indicators. Small reductions in double support time and decreased push-off force were evident on the treadmill. Treadmill walking was more demanding in terms of metabolic cost than overground walking.
- For stroke subjects, overground and treadmill walking were similar in terms of temporal-distance measures. Significant interlimb asymmetries in joint excursions at the ankle, knee and hip were evident in overground walking but not on the treadmill. Lower push-off forces were evident on the treadmill; however, treadmill walking was more demanding in terms of metabolic cost than overground walking.
- Comparisons of overground and treadmill walking in healthy adults and stroke subjects revealed differences in temporal-distance parameters, kinematic variables and metabolic costs although kinetics were similar between groups. All temporal distance measures, hip excursion and ground reaction forces were affected by increasing the treadmill belt speed in a similar fashion for healthy subjects and
those with stroke. The metabolic cost of increasing walking speed was much greater in stroke than in healthy subjects.

The three most important and relevant findings from these studies are the improvement in symmetry of walking on the treadmill in subjects with stroke, the higher energy costs of treadmill walking and the abnormally large increase in metabolic demands associated with increasing walking speed in people with stroke. The first has implications for clinical practice, and the second and third have implications both for clinical practice and for the directions that are needed for further research in this area. These findings will be addressed within these topics.

- Clinical Implications of the Findings

Let us take the position that a clinician is considering when she/he should use a treadmill as part of a treatment program, and let us ask what findings from this study are relevant to discussion. First, the excursions of the joints and movements of the limbs would be very similar to overground walking although kinematic symmetry is improved for stroke subjects with treadmill walking. These findings support the reports of improvement in symmetry in two other studies (Harris-Love et al., 2001; Hesse et al., 1999) which gives assurance that the differences are real. Asymmetrical walking has been reported by most who have studied stroke gait (Griffin et al., 1994; Nakamura et al., 1988; Roth et al., 1997; Wall and Turnbull 1986). One can ask whether this is a desirable result, and what implications this might have. Although at first glance there seems to be little doubt that symmetry of gait is desirable, the literature has failed to show that symmetry relates to other measures of gait competence, notably gait speed (Griffin et al., 1995; Olney et al., 1994) or more efficient walking. The finding that subjects work at a higher metabolic level
when using the treadmill rather than walking at the same speed overground, at least initially, suggests that this might not be the case. If gait symmetry is desirable, which undoubtedly is the clinical preference for early gait retraining, the therapist may choose to have the person begin treadmill walking at a slower pace than overground, and monitor the heart rate and subjective perceived exertion as the subject progresses.

It would also be important to know if the increased symmetry observed with treadmill walking would carry over to overground walking, and if so, how long this would take. Though clinicians advocate for gait symmetry as an objective of treatment, it is also not clear if optimal walking performance can be achieved through encouragement of biomechanical symmetry especially in persons whose limbs have unequal capabilities (Griffin et al., 1995; Winter et al., 1990; Winter and Eng, 1995). In the studies discussed here, people with stroke were able to walk at faster speeds on the treadmill by modifying cadence and step/stride length and increasing hip excursion without any loss of symmetry, but this came at a substantial metabolic cost. Clinicians will want to carefully monitor cardiovascular responses to increasing speed since this strategy is being advocated in order to maximize the benefit of treadmill gait training (Pohl et al., 2002).

In summary, the therapist can make immediate use of these finding in clinical practice. Further research may lead to more precise recommendations for applications.

- Implications of findings for further research

For both groups of subjects, factors such as unfamiliarity with treadmill walking, co-contraction of agonist and antagonist muscles, anxiety, inefficient energy exchange and mostly the differences in kinetics have been identified as possible contributors to the higher energy costs of treadmill walking.
One of the first explorations of interest is to see if the differences between the two modes of walking in healthy adults and stroke subjects diminish over time as subjects became more familiar with treadmill walking. This research also has implications for clinical practice.

Considering the many similarities in temporal distance and kinematic parameters in both groups of subjects, the higher energy cost of treadmill walking was unexpected but consistent across all subjects. This finding presents a fascinating subject for further biomechanical and motor control research; understanding the work of gait is important in itself, but also it is needed in developing recommendations for rehabilitation. The problem must start with recognition that although we cannot be certain that there are kinematic differences between overground and treadmill walking, there must be kinetic variations. Furthermore, the different metabolic responses to increased belt speed in the two subject groups suggest that the kinetic variations diverge as a function of the challenge of walking faster. Further analysis of existing data, or use of existing methods, could be carried out to shed light on this finding.

Segment-by-segment energy analysis using the methods popularized by Winter (1991) could be performed. In this, no force plates are used, and the segmental energies are determined using kinematic information and calculated segment masses calculated from total body mass and anthropometric constants. Using this method several assumptions must be made including complete energy transfer within and between segments (Winter, 1991). Due to the coarseness of this method and its dependence upon the kinematics (which were not different for the two walking surfaces in the current study) it seems unlikely that it would offer any further insight.
Link segment analysis in three dimensions can be used to assess power flows or transfer of energy through segment ends (McGibbon et al., 2001; Winter, 1991). Because the inter-segment transfer could be different between the two walking conditions, indicating different motor strategies, it could be very useful if it had been available for treadmill walking. Until it is possible to obtain a full three dimensional link-segment analysis for both conditions, this method will be limited in interpretive value.

Even full link segment analysis when used alone cannot detect co-contraction of agonist and antagonist muscles, and for both groups of subjects energy costs would be increased if co-contraction was increased by anxiety or lack of skill. Co-contraction is, however, difficult to quantify. Electromyography itself may yield some useful information, especially when electrodes can be left in place through the two conditions to minimize application differences. Profiles from the two conditions could be compared visually and mathematically though some means of modelling the input of co-contracting muscles is needed to quantify differences. Electromyographic-force models during dynamic activities that incorporate multiple muscles are becoming feasible (Doorenbosch et al., 2005; White and Winter, 1993), and represent promising means of clarifying the fundamental neuromuscular patterns that underlie the movement profiles. In addition to enabling quantification of co-contraction, these methods would offer additional insight by being able to determine individual temporal muscle histories and direct measurement of activation patterns. The provision of meaningful muscle activation histories and quantification of co-contraction would be extremely valuable in exploring neuromuscular control and in attempting to explain energy cost differences found in the present study. Because of its relative explanatory power this might be the first approach that would be chosen.
In summary, methodologically more complex studies are warranted to determine the specific factors causing the metabolic differences observed.

REFERENCES


Appendix A: Consent Form

TITLE OF PROJECT: Biomechanics and energy costs of overground and treadmill walking in healthy adults and in stroke subjects

INVESTIGATORS: Dr. Brenda Brouwer, School of Rehabilitation Therapy, Dr. Sandra Olney, School of Rehabilitation Therapy, Dr. Anne Brown, School of Nursing, Krishnaji Parvataneni, Ph.D. candidate, School of Rehabilitation Therapy

BACKGROUND INFORMATION

You are being invited to participate in a research study to evaluate the differences between overground and treadmill walking in healthy adults and in subjects with stroke. One of the investigators will read through this consent form with you and describe the procedures in detail and answer any questions you may have. This study is being sponsored by the Botterell Foundation of Queen’s University.

DETAILS OF THE STUDY

The purpose of this study is to provide a comprehensive evaluation of overground and treadmill walking conditions in healthy adults and in subjects with stroke and then compare the walking performance between the two groups. You will be considered for the study if you have had a stroke more than 6 months ago that affected one side of your body OR if you are healthy and have no known medical conditions. You must be able to walk without an aid at a self-selected walking speed of 30 cm/sec or more.

Description of visits and tests to be performed as a part of the study

An initial visit to the Motor Performance Laboratory in the School of Rehabilitation Therapy will be required at which time we will measure your normal walking speed and ask you some questions about your general health and your activity level. We will also familiarize you with the equipment that will be used during the testing
and have you try on the mask that we use to measure how much oxygen you use. This visit serves as a final screen for eligibility and provides an opportunity for you to ask questions about the research. This visit will take no more than 45 minutes. One additional visit will be scheduled within a one-week period.

During your visit you will be asked to walk on a walkway at your self-selected speed for about 10 minutes. Later you will be made to walk on a treadmill for about 6-10 minutes starting at the self-selected speed on the walkway or that you are comfortable with. After two minutes the speed will be increased by 10% and then after another 2 minutes increased again by the same amount. While you are walking you will be wearing a safety harness to prevent you from having a slip or a fall.

You will also have many instruments attached to you. Recording electrodes will be placed on your skin overlying the muscles of your legs so we can record the activity of these muscles while you walk and electrodes will also be placed on your chest so we can monitor the activity of your heart. Light emitting diodes will be placed on your ankle, knee and hip joints and also at several points on your leg to allow us to measure how your legs move. You will also be fit with a mask that will cover your mouth and nose so we can monitor your breathing and measure how much oxygen you use while you walk. Your blood pressure will be taken every couple of minutes. This visit will require about 2 to 2.5 hours of your time.

**Risks/Side-Effects**

All the tests we will do have been conducted in healthy people and in people who have had strokes. Treadmill walking can be awkward or destabilizing however the harness worn will prevent you from falling. In the presence of cardiovascular disease, physical exertion can pose a risk. The test we use does not require maximal exertion and can be terminated at your request or if indicated from blood pressure or heart monitoring. An investigator will carefully supervise you at all times. The electrodes used are sticky and may leave red marks on the skin after removal; these generally disappear within hours. Your muscles may be a bit sore within 48 hours after the testing though this is usually avoided by having you warm-up before testing and stretch and cool down afterwards. You may feel somewhat fatigued after testing depending on how active you
normally are. This should not be excessive but just the result of having done some physical activity.

If you have any concerns, please ask the investigators immediately or contact Dr. Brouwer at 533-6087 at a later time.

**Benefits**

While you may not benefit directly from this study, the results from this study will contribute to our understanding of the factors that limit physical mobility following stroke and may benefit stroke patients in the future through the development of targeted interventions.

**Exclusions**

The testing involved requires some physical exertion. You will not be considered for this study if you have any of the following: a pre-existing mobility restriction, unstable angina, peripheral vascular disease or congestive heart failure. If you are not sure if you have any of these things please consult your doctor first.

**Confidentiality**

All information obtained during the course of this study is strictly confidential and your anonymity will be protected at all times. You will be identified by a code based on subject group (stroke or control), the date of entry into the study and the project identifier. Data will be stored in locked files and will be available only to the investigators associated with this study. You will not be identified in any publication, presentation or reports.

**Voluntary nature of study/Freedom to withdraw or participate**

Your participation in this study is voluntary. You may withdraw from this study at any time for any reason and your withdrawal will not affect future care.
Withdrawal of subject by principal investigator or co-investigators

A study investigator may decide to withdraw you from this study if you exhibit signs of cardiovascular stress (based on blood pressure and heart monitoring) or an inability to follow the protocol.

Liability

In the event that you are injured as a result of the study procedures, health care will be provided to you until resolution of the problem. By signing this consent form, you do not waive your legal rights nor release the investigators and sponsors from their legal and professional responsibilities.

Payment

We appreciate your involvement in this study. After completion of all testing you will be given a stipend of $60 to compensate you for time and travel. We can provide you with parking at the rear of the School of Rehabilitation Therapy building if needed.

SUBJECT STATEMENT AND SIGNATURE SECTION:

I have read and understand the consent form for this study. I have had the purposes, procedures and technical language of this study explained to me. I have been given sufficient time to consider the above information and to seek advice if I chose to do so. I have had the opportunity to ask questions which have been answered to my satisfaction. I am voluntarily signing this form. I will receive a copy of this consent form for my information.

If at any time I have further questions, problems or adverse events, I can contact:
Dr. Brenda Brouwer, Principal Investigator at 533-6087
OR
Dr. Sandra Olney, Director of the School of Rehabilitation Therapy at 533-6102

If I have questions regarding my rights as a research subject I can contact
Dr. Albert Clark, Chair, Research Ethics Board at 533-6081
By signing this consent form, I am indicating that I agree to participate in this study.

_______________________  _________________
Signature of Subject       Date

_______________________  _________________
Signature of Witness       Date

STATEMENT OF INVESTIGATOR:
I, or one of my colleagues, have carefully explained to the subject the nature of the above research study. I certify that, to the best of my knowledge, the subject understands clearly the nature of the study and demands, benefits, and risks involved to participants in this study.

____________________________  _________________
Signature of Principal Investigator  Date
(Krishnaji Parvataneni)