Foot Placement Patterns of Individuals with Multiple Sclerosis during Rollator-Assisted Community Mobility

by

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A thesis submitted in conformity with the requirements for the degree of Master of Science,
Graduate Department of Rehabilitation Science,
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Abstract

Individuals with Multiple Sclerosis commonly use assistive mobility devices, such as rollators, to compensate for their mobility impairments. However, the effect of these devices on their foot placement during gait has not been explored in the community. The objective of experiment one was to develop and validate a tool that quantifies medio-lateral foot placement characteristics during rollator use. In this study, a technique was developed for an instrumented rollator (i.e. iWalker) and validated against a Vicon motion capture system in able-bodied young adults. The two systems were in strong agreement. The objective of experiment two was to apply this iWalker-based technique to individuals with Multiple Sclerosis to identify environment-related foot placement changes. This study revealed that step width variability, but not step width, can be influenced by certain outdoor environments. Therefore, environmental context is important to consider when investigating user-device interactions and factors responsible for safe mobility in this population.
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List of Abbreviations

2D  Two-Dimensional
AMD  Assistive Mobility Device
CNS  Central Nervous System
FOV  Field of View
ICC  Intra-Class Correlation Coefficient
ICF  International Classification of Functioning, Disability, and Health
MS  Multiple Sclerosis
PPMS  Primary Progressive Multiple Sclerosis
PRMS  Progressive-Relapsing Multiple Sclerosis
PSFS  Penn Spasm Frequency Scale
RMS  Root-Mean-Square
RRMS  Relapsing-Remitting Multiple Sclerosis
SPMS  Secondary Progressive Multiple Sclerosis
SW  Step Width
SWV  Step Width Variability
1 Introduction

Multiple sclerosis (MS) is one of the leading causes of neurological disability in young adults around the World and is highly prevalent in Canada. (Hauser & Oksenberg, 2006; World Health Organization, 2008) This chronic disease causes a progressive deterioration of the central nervous system (CNS), and results in a long-term accumulation of symptoms giving rise to disability (Confavreux & Vukusic, 2006a). The mechanisms underlying the manifestations of its symptoms have been demonstrated to involve a number of processes occurring in the brain and spinal cord, including a widespread demyelination of axons (Rovaris et al., 1997), inflammation (Bitsch, Schuchardt, Bunkowski, Kuhlmann, & Brück, 2000), axonal damage (Ferguson, Matyszak, Esiri, & Perry, 1997), and lesion development (Peterson, Bö, Mörk, Chang, & Trapp, 2001). These neurological changes have been confirmed using a variety of technologies, such as magnetic resonance imaging (MRI) (e.g. T2-weighted lesions) (Ormerod et al., 1987), functional MRI (fMRI) (e.g. cortex activation patterns) (Lee et al., 2000), electrophysiological techniques (e.g. evoked potentials and excised tissues) (Cameron & Lord, 2010; Kapoor, Brown, Thompson, & Miller, 1992), and diffusion tensor imaging (DTI) (e.g. measures of mean diffusivity and fractional anisotropy) (Cercignani, Inglese, Pagani, Comi, & Filippi, 2001). In addition to the signs of CNS deterioration, a wide-range of sensory, motor, and cognitive deficits have been documented to emerge over the course of the disease. (DeLuca, Chelune, Tulsky, Lengenfelder, & Chiaravalloti, 2004; Nilsagård, Denison, Gunnarsson, & Boström, 2009) One of the most frequent sequelae causing disability in this population is gait impediments. (Morris, Cantwell, Vowels, & Dodd, 2002) Out of the wide-range of clinical symptoms induced by MS, eighty-five percent of individuals with this disease reported that disturbances to gait were their primary complaint. (Scheinberg et al., 1980) As the disease progresses over time, the ambulatory ability of individuals with MS tends to decline, resulting in an increased falls risk (Nilsagård, Lundholm, Denison, & Gunnarsson, 2009; Peterson, Cho, Vonkoch, & Finlayson, 2008; Soyuer, Mirza, & Erkorkmaz, 2006) and the eventual loss of mobility (Pittock et al., 2004; Weinshenker, 1995). Therefore, to improve the quality of life of individuals with MS and reduce their disability, there is a need for research into interventions that prolong their ambulatory ability and make walking safer and more effective for them in their communities.
2 Background

2.1 Epidemiology

2.1.1 Incidence and Prevalence of Multiple Sclerosis in Canada

In Canada, multiple sclerosis (MS) occurs in as many as 240 per 100,000 persons. (Beck, Metz, Svenson, & Patten, 2005) As a result, the Canadian prevalence rate is fifth highest in the world. (World Health Organization, 2008) There are regional variations, with significantly higher odds of MS occurring in the Prairies and Atlantic regions than other parts of the country. (Beck et al., 2005) In North America, the median estimated incidence of MS has been determined to be 1.5 new cases per 100,000 persons. (World Health Organization, 2008)

Although the incidence of MS in Canada has been on the rise in the past, it is currently unclear if this trend continues to persist. Longitudinal studies in some regions of the country have reported that there has been no change in the incidence of this disease in recent decades (Hader & Yee, 2007; Marrie, Yu, Blanchard, Leung, & Elliott, 2010; Warren, Svenson, & Warren, 2008). Over the past 50 years, however, there has been a disproportionate increase in the incidence of MS among Canadian women, compared to men. The ratio of female to male cases is believed to exceed 3.2:1 and continues to grow. (Orton et al., 2006) Nevertheless, in both sexes, the estimates of prevalence have increased significantly over time nationwide. (Hader & Yee, 2007; Marrie et al., 2010; Sloka, Pryse-Phillips, & Stefanelli, 2005; Warren, 2008) As a result, there are progressively more people living with MS and they are living longer. (Marrie et al., 2010)

With this increased longevity, individuals with MS must cope with living with disability for longer. Mobility problems play an important role in the reduced quality of life experienced by individuals with MS who have relapsing or progressive forms of the disease. (Hemmett, Holmes, Barnes, & Russell, 2004) Therefore, we must increase our knowledge of MS to optimize the ambulatory ability of those affected by this disease and improve their quality of life.

2.1.2 Natural History of Multiple Sclerosis

The pioneering work to differentiate MS as a separate, identifiable disease was first conducted in the mid-nineteenth century by Charcot and his predecessors. (Miller, 2003, p. 13; Murray, 2009) A first characterization of the disease began to emerge through the accumulation of clinical and pathological data for the initial forty years of work. The main focus of research
then shifted to understanding the cause and mechanisms underlying MS for the subsequent 120 years. Progress in this period was not marked by revolutionary breakthroughs, but rather, by an evolution of knowledge that stemmed from the work of thousands of researchers and fostered by advances in basic science, clinical disease, and medical technology. Only in the past 10 years, have therapies been developed that effectively change the course of the disease. Nevertheless, despite how treatable MS has become, a cure continues to remain elusive. (Compston & Coles, 2008)

Presently, there is a full spectrum of symptoms and characteristics known to be exhibited by individuals with MS. (Confavreux & Vukusic, 2006b). The onset of symptoms typically occurs between the ages of 20 and 40, although MS has also been documented to develop in children and adults over 60 years old. (Stüve & Oksenberg, 2010) In general, the early stages of the disease are characterized by episodes of neurological dysfunction, from which individuals typically recover. Episodes at this stage often involve sensory impairments, such as paresthesia, vertigo, and optic neuritis. Acute and insidious motor deficits are also common, including limb ataxia and difficulties with balance. (Weinshenker et al., 1989) Between episodic relapses are clinically stable periods, during which the presence of symptoms is not characteristic. (Matthews, Compston, Allen, & Martyn, 1991, p. 144) In the long-run, however, extensive neurodegeneration ensues, manifesting as a progressive accumulation of disability. (Compston & Coles, 2008) Four clinical phenotypes have, thus, been developed to help categorize the disease course of individuals with MS: relapsing-remitting, secondary progressive, primary progressive, and progressive-relapsing. Relapsing-remitting MS (RRMS) is typically present in approximately 80 percent of individuals, with over half converting to the secondary-progressive (SPMS) course within ten years of diagnosis. Another 20 percent of individuals with MS are initially diagnosed with the primary progressive (PPMS) course. The occurrence of the progressive-relapsing (PRMS) phenotype is quite rare, but is very potent in the cases in which it presents. Nevertheless, with the large inter-individual heterogeneity exhibited by individuals with MS and seemingly infinite variety of symptoms, the clinical course of MS is often considered unpredictable. (Compston & Coles, 2008; Stüve & Oksenberg, 2010; Vukusic & Confavreux, 2007)
2.2 Clinical Characteristics and Gait Irregularities

Regardless of which clinical course manifests, neurological deficits lead to an ongoing deterioration of body functioning. The indications of dysfunction can be grouped into those that are physical [i.e. affecting sensory (Alpini, Caputo, Pugnetti, Giuliani, & Cesaran, 2001; Fisher et al., 2006; Nilsagård et al., 2009; Soyuer et al., 2006; Weinshenker et al., 1989; Zeigelboim et al., 2008) and motor systems (Abou Zeid, Weinshenker, & Keegan, 2009; Alusi, Worthington, Glickman, & Bain, 2001; Cowan, Ormerod, & Rudge, 1990; Djaldetti, Ziv, Achiron, & Melamed, 1996; Kapoor et al., 1992; Kent-Braun et al., 1997; Koseoglu, Gokkaya, Ergun, Inan, & Yesiltepe, 2006; Nilsagård et al., 2009; Rice, Vollmer, & Bigland-Ritchie, 1992)] as well as those that are cognitive in nature (Beatty & Monson, 1996; Bergendal, Fredrikson, & Almkvist, 2007; DeLuca et al., 2004; Pelosi, Geesken, Holly, Hayward, & Blumhardt, 1997; Smestad, Sandvik, Landrø, & Celius, 2010). Since the performance of gait has sensory (Pearson, 1995), motor (Winter, 1991), and cognitive components (Lundin-Olsson, Nyberg, & Gustafson, 1997), a range of MS symptoms, either alone or in combination, can be considered to contribute to the development of gait irregularities. Indeed, a high incidence of gait impairments is observed in the MS population. (Morris et al., 2002) Therefore, to make sense of these irregularities, a wide range of symptoms must be examined.

Symptoms that arise out of the progressive deterioration of sensory, motor, and cognitive functions have been well-characterized and documented in clinical and laboratory reports. Compared to able-bodied individuals, people living with MS often experience sensory abnormalities, which affect the information entering the CNS. These deficits may include distortions in proprioception (e.g. diminished joint position sense and loss of postural sense in the distal lower limbs) (Nilsagård et al., 2009), vestibular disturbances (e.g. vertigo and nystagmus) (Alpini et al., 2001; Zeigelboim et al., 2008), reductions in vision (e.g. diplopia, optic neuritis, and loss of visual acuity) (Fisher et al., 2006; Weinshenker et al., 1989), as well as touch abnormalities (e.g. impaired vibration sense, reduced cutaneous sensation, and paresthesia) (Soyuer et al., 2006). The transmission of sensory information has also been proven to be affected by MS, as revealed by the delayed latencies of spinal somatosensory evoked potentials in individuals with MS compared to those of able-bodied individuals. (Cameron & Lord, 2010; Dorfman, Bosley, & Cummins, 1978)
In addition, manifestations of MS are evident in the motor system. For example, spasticity (e.g. in the lower limbs) (Nilsagård et al., 2009), increased levels of fatigue (Djaldetti et al., 1996; Kent-Braun et al., 1997) and decreased muscle strength (e.g. in the quadriceps muscles) (Rice et al., 1992) are common symptoms of individuals with MS. Weakness of the lower limbs (Cowan et al., 1990) and respiratory muscles (Koseoglu et al., 2006) are also prevalent and, in severe cases, may eventually lead to the development of hemiparesis or bilateral diaphragmatic paralysis respectively. MS has also been associated with tremors (Alusi et al., 2001), apraxia (Abou Zeid et al., 2009), and the occurrence of some involuntary movements (e.g. propriospinal myoclonus) (Kapoor et al., 1992). As well, many of these features may be, in part, responsible for the reduced postural stability as measured by increased elliptical area of centre-of-pressure deviation, postural sway velocity, and medio-lateral sway range in this patient population (Fjeldstad, Pardo, Bemben, & Bemben, 2010; Kalron, Dvir, & Achiron, 2010; Sosnoff, Shin, & Motl, 2010).

Furthermore, several impairments to cognition are characteristic of MS. For instance, decrements in memory (i.e. working and long-term memory) (Pelosi et al., 1997), executive functioning (e.g. concept formation) (Beatty & Monson, 1996), attention (e.g. selective attention) (Smestad et al., 2010), and the efficiency of information processing (Bergendal et al., 2007; DeLuca et al., 2004) have been reported. These have implications for the capacity of individuals with MS to adapt to their environment (Plow, Resnik, & Allen, 2009), demonstrate planning and goal-directed behaviour (Lima et al., 2007; Schulz, Kopp, Kunkel, & Faiss, 2006), succeed at learning (Demaree, Gaudino, DeLuca, & Ricker, 2000), and make appropriate decisions (Nagy et al., 2006) in a timely manner. Thus, although the general intellectual functioning of individuals with MS may be similar to able-bodied individuals (e.g. in performance scores on the Raven's Standard Progressive Matrices and the National Adult Reading Test) (Macniven et al., 2008), their cognitive deficits (e.g. impaired simple and focused attention) (Schulz et al., 2006) may interfere with their ability to engage in activities of daily living, such as taking part in a conversation while walking (Kalron et al., 2010) in the community.

In combination, the sensory, motor, and cognitive deficiencies observed in individuals with MS often result in alterations to gait, which are directly observable in clinics and laboratories. In fact, impairments affecting gait are the most apparent cause of disability in MS. (Morris et al., 2002; Souza et al., 2010; Whetten-Goldstein, Sloan, Goldstein, & Kulas, 1998)
Relative to the gait patterns of able-bodied individuals, individuals with MS differ in several characteristic ways. For example, individuals with MS usually walk with a reduced gait speed, wider base of support, decreased stride and step lengths, reduced cadence and prolonged double-support time. (Cameron & Lord, 2010; Givon, Zeilig, & Achiron, 2009; Kalron et al., 2010; Martin et al., 2006; Morris et al., 2002) In addition to spatio-temporal differences, they exhibit altered gait kinematics, such as significant reductions in knee extensions at heel strike, and sagittal plane joint excursions at the ankle and hip. (Kelleher, Spence, Solomonidis, & Apatsidis, 2010; Martin et al., 2006) The variability of many kinematic parameters is increased in this population, including increased hip, knee, and ankle joint variability. (Cameron & Lord, 2010; Crenshaw, Royer, Richards, & Hudson, 2006) Kinetic parameters of gait in individuals with MS can also differ from able-bodied individuals, as is evident through observed decreases in anterior-posterior propulsive and braking ground reaction forces, maximum ankle dorsiflexion moments, and hip flexion and extension moments. (Crenshaw et al., 2006; Kelleher et al., 2010) As well, muscle activation patterns during gait have been shown to differ in individuals with MS, compared to able-bodied individuals. For instance, they present reductions in the amplitude and altered timing of medial gastrocnemius and tibialis anterior activity during gait. (Martin et al., 2006) Furthermore, dual-tasking paradigms have been used to reveal decrements in both gait and cognitive performance compared to able-bodied controls, as participants walk while performing a concurrent cognitive task. (Hamilton, 2009; Kalron et al., 2010) Taken as a whole, these altered clinical and gait characteristics have implications for effective and safe mobility in this population.

### 2.3 Multiple Sclerosis and Mobility

Within ten to fifteen years from disease onset, problems with gait are experienced by eighty percent of individuals with MS. Beyond those initial fifteen years, approximately half of all persons with MS require assistance to ambulate and, within twenty years, over one third lose their ability to walk altogether. (Givon et al., 2009; Noseworthy, Lucchinetti, Rodriguez, & Weinshenker, 2000; Pittock et al., 2004; Souza et al., 2010; Weinshenker, 1995) This progressive loss of ambulatory ability contributes to the increased physical disability observed in this population, which is associated with reduced levels of physical activity and a deterioration in the quality of life. (Miller & Dishon, 2006) It can also contribute to substantial patient burden (Sutliff, 2010) and impede participation in activities of daily living at work, among family
members, and during time at leisure. (Souza et al., 2010) As a result, a loss of mobility can have a profound influence on the well-being of an individual. (Metz, 2000) Consequently, maintaining mobility is associated with an increased health-related quality of life and, in older adults, is considered essential for active aging and independence. (Webber, Porter, & Menec, 2010) It is, thus, understandable that the maintenance of mobility in individuals with MS has been ranked as a high priority at all stages of the disease. (Sutliff, 2010)

As mobility deteriorates in individuals with MS, falls become a major concern as they can have a severe effect on levels of disability and quality of life. Given the nature of MS, it is not surprising that falls are prevalent in this population. (Cameron & Lord, 2010) However, not only do individuals with MS experience falls often, they also demonstrate a significant fear of falling. (Peterson, Cho, von Koch, & Finlayson, 2008) This fear is so intense that it has been associated with the curtailment of activity in middle-aged and older adults in this population. (Peterson, Cho, & Finlayson, 2007) Individuals with MS are also at an elevated risk of developing fall-related injuries as well as experiencing falls and accidents that culminate in death, compared to the population at large. (Brønnum-Hansen, Hansen, Koch-Henriksen, & Stenager, 2006; Cameron & Lord, 2010; Peterson et al., 2008) Therefore, a multi-factorial assessment of the impairments induced by MS should be conducted, in order to better identify the factors responsible for falls and develop more effective fall prevention programs for this population. (Souza et al., 2010)

2.3.1 Conceptualizing Mobility

Before discussing the factors known to contribute to mobility impairments in the MS population, it is necessary to clarify the meaning of mobility that is being presented throughout this work. On the whole, the concept of mobility describes a person's ability to change the position or location of his or her body. (World Health Organization, 2010) As such, it is inherently a complex construct (Metz, 2000) comprised of many parts. Walking is but one aspect of mobility. As set forth by the International Classification of Functioning, Disability, and Health (ICF), other aspects of this construct may include changing the position of your body (e.g. shifting the body's centre of gravity), maintaining the position of your body (e.g. continuing to stand), transferring yourself without changing body position (e.g. while sitting), interacting with objects (e.g. carrying, moving, and handling them), engaging in other forms of moving around in
addition to walking (e.g. jogging), moving around in different locations (e.g. the home, other buildings, or outside), and moving around with the use of transportation (e.g. driving a motor vehicle). (World Health Organization, 2010) While all of these are of vital importance to the overall mobility of a person, with deficits in any of these categories contributing to disability, the aspects of mobility that will be focused on in this work are those that involve walking and moving around in different locations.

In order to adequately assess the factors that affect the mobility of individuals with MS (which in this work, is their ability to ambulate in different environments in the community), one must take into consideration the following three factors and their interactions: (1) the individual characteristics of the person; (2) features of the walking environment; and (3) the effect of mobility interventions being used. The ICF model can be used to provide further clarity on the mobility-influencing factors of interest. The ICF is a classification of health and health-related domains that was adopted by the World Health Organization in 2001. (Delisa, 2004) This model ushered in a paradigm shift for the scientific community (Stucki, Ewert, & Cieza, 2002), by integrating the medical and social models of disability into a single bio-psychosocial model (Bernd, Van Der Pijl, & De Witte, 2009) and providing a coherent perspective on biological, individual and social aspects of health and functioning (Stucki et al., 2002). Currently, the ICF model acts as the guiding framework for research in the field of rehabilitation science and it has been used successfully to characterize aspects of functioning and disability that are affected by MS. (Holper et al., 2010; Svestkova et al., 2010) Future work on the interaction between person, environment, and intervention in the MS population will provide insight into new areas of the ICF model and contribute to the development of knowledge in the field of rehabilitation science.

2.3.2 Individual Determinants of Mobility in Multiple Sclerosis

In individuals with MS, the most common deficits that have been documented to result in disability are those that impair ambulation. (Noseworthy et al., 2000; Souza et al., 2010) Disordered movements in individuals with MS may arise out of several pathology-related sources, including demyelination, axonal destruction, and sclerotic plaque formation in the brain and spinal cord. (Morris, Cantwell, Vowels, & Dodd, 2002) The MS-provoked destruction of specific areas of the central nervous system can cause certain predictable symptoms to manifest. For example, lesions in the cerebral hemisphere, thalamus, or adjacent areas are often
responsible for sensory impairments from head to toe on the opposite side of the body. (Olek, 2005, p. 4) On the contrary, common motor symptoms (e.g. the development of paraparesis or quadriparesis) are often the result of spinal cord lesions. (Olek, 2005, p. 6) Slower spinal somatosensory conduction and impaired central integration can also play a role in symptom manifestation. (Cameron, Horak, Herndon, & Bourdette, 2008; Cameron & Lord, 2010) Overall, these neurological deficits give rise to the characteristic differences observed in clinical features of individuals with MS, compared to able-bodied individuals, as previously outlined. More importantly, the symptoms these deficits produce can severely diminish the capacity of individuals with MS to achieve effective mobility, making ambulation challenging and less safe.

Several symptoms of nervous system dysfunction, caused by MS, are known to affect ambulation, and thus mobility. These include loss of balance, muscle weakness, fatigue, cognitive impairment, fear of falling, spasticity, tremor, and visual impairment. (Klewer, Pöhlau, Nippert, Haas, & Kugler, 2001; Peterson et al., 2008; Souza et al., 2010) For example, when MS-related weakness is presented in the anterior tibial muscles, a common outcome is a limited ability to control dorsiflexion and so the foot drops. This feature can reduce toe clearance during the swing phase of gait, lower stance phase stability, and lead to poor placement of the affected foot at initial contact. Tripping may, thus, be commonly observed. (Bregman et al., 2010; Esnouf, Taylor, Mann, & Barrett, 2010; Olek, 2005, p. 6) Extensor spasms of the legs and ankle clonus, resulting from MS, can also interfere with ambulation. (Chua & Kong, 2001; Olek, 2005, p. 6) Furthermore, fatigue in individuals with MS can reduce endurance, limit physical activity, and alter fitness through deconditioning. (Judica et al., 2011) As well, even when a single sensory input is altered in individuals with MS (such as vestibular or visual input), during quiet standing, significant reductions in balance performance can be observed. (Cattaneo & Jonsdottir, 2009) Therefore, the mobility of individuals with MS can be influenced by manifestations of their neurological deficits, which can alter their ambulatory ability and may even play a role in the increased fall risk exhibited by this population.

Although the cause of falls in MS is uncertain, a number of factors have been directly associated with fall risk in this population. (Cameron & Lord, 2010) Common factors related to falls include diminished visual acuity, peripheral neuropathy, divided attention, reduced muscle endurance, impaired proprioception, spasticity, limited walking ability, higher Expanded Disability Status Score (EDSS) scores, fatigue, and heat sensitivity. The odds of falling were also
increased with higher scores on the Expanded Disability Status Score (EDSS). This scale, which is specific to the MS population, is used to assess the full-range of ambulatory impairments caused by this disease, ranging from 0.0 (normal neurological examination) to 10.0 (death due to MS). Surprisingly, use of a walking aid was also determined to be a fall risk factor. (Cameron & Lord, 2010; Nilsagård et al., 2009; Nilsagård, Denison, Gunnarsson, & Boström, 2009; Sandyk, 1999) When individuals with MS reached an EDSS score of 4, their disability progression proceeds at a predictable rate. This suggests that it might be possible to take advantage of early treatment interventions to prevent individuals with MS from reaching this landmark EDSS 4 score, and thus, slow the progression of the disease. (Hurwitz, 2011)

2.3.3 Environmental Factors that Influence Mobility

Before proceeding to discuss the interventions used to overcome mobility limitations caused by MS, it is first necessary to put into perspective the potential factors that can act on mobility, whether they are facilitators or barriers. Using a bio-medical approach (i.e. the perspective that a patient's functioning and health is an outcome of the disease process) (Stucki et al., 2002), the gait impairments observed in individuals with MS are a result of the neurological deficits of the disease and can be seen as impediments to successful mobility. While this is certainly true, the role of the walking environment on mobility is also an important consideration and has not been well-explored in gait studies on individuals with MS to-date. It, therefore, requires further investigation.

Previous work, in other populations, has demonstrated that gait characteristics can be influenced by environmental features. The positioning of the feet during gait is one such characteristic, as it is affected by the environment in both indoor and outdoor settings. (Carter, Campbell, Sanson-Fisher, & Gillespie, 2000; Gill, Williams, & Tinetti, 2000) Foot placement patterns play an important role in the prevention of falls and, thus, affect mobility. For example, inadequate toe clearance and other errors in foot placement can result in tripping. (Schulz, Lloyd, & Lee, 2010) At the present time, several factors are known to induce changes in an individual's gait pattern during walking in the community. Physical factors include the angle of inclination of the walking surface (e.g. ramps and curbs) (Crosbie & Ko, 2000; Lay, Hass, & Gregor, 2006; Prentice, Hasler, Groves, & Frank, 2004; Schmitt & Bonnono, 2009), features of the shoe-surface interface (e.g. the type of shoe worn by individuals with MS and coefficient of friction of
the surface) (Hanson, Redfern, & Mazumdar, 1999; Kelleher et al., 2010; Menant, Steele, Menz, Munro, & Lord, 2009; Redfern et al., 2001), lighting conditions (e.g. low lighting) (Marigold & Patla, 2005; Richardson, Thies, DeMott, & Ashton-Miller, 2005), and features of the surface terrain (e.g. its compliance, rigidity, and regularity) (Hanson et al., 1999; MacLellan & Patla, 2006; Marigold & Patla, 2005, 2008; Richardson, Thies, DeMott, & Ashton-Miller, 2004; Smothers, Ray, & Wildman, 2000; Thies, Richardson, & Ashton-Miller, 2005). There are also cognitive factors that can influence foot placement patterns, such as the attentional demands associated with walking through certain environments. The influence of cognitive factors has been demonstrated during intersection crossing (Shumway-Cook et al., 2002), which also has a temporal constraint (Hoxie & Rubenstein, 1994), as well as during conversational dual tasking during walking in individuals with MS (Hamilton et al., 2009), other forms of dual-tasking during gait (Dingwell, Robb, Troy, & Grabiner, 2008; Hollman, Kovash, Kubik, & Linbo, 2007), and investigations into visuo-spatial processing during navigation through a complex environment (Gérin-Lajoie et al., 2007). Although the influence of these factors has been recognized, how they affect the foot placement and the mobility of individuals with MS in everyday situations has yet to be ascertained.

Furthermore, using the Verbrugge scheme of the Disablement Process, disability can be perceived not as a personal characteristic, but rather as gap between personal capability and environmental demand. (Verbrugge & Jette, 1994) From this perspective, mobility interventions could be seen as environmental factors that bridge this gap, by affecting the direction, pace, and patterns of change in the trajectory of disease-related functional consequences over time. (Verbrugge & Jette, 1994) Since gait interventions are commonly used by individuals with MS to facilitate ambulation, these interventions must be taken into account when discussing the mobility of individuals with MS.

### 2.3.4 Interventions to Enhance Mobility

Factors that demonstrate great potential for improvement are important to consider when planning a mobility-enhancing intervention. Accordingly, there is a need to improve our understanding of mobility in order to enhance the effectiveness of our interventions. In the MS population, research has been traditionally focused on understanding the etiology of the disease and developing clinical treatment regimens. As a result, types of interventions that have been
studied to-date include training programs for individuals with MS (e.g. aquatic training, cycling resistance training, visuo-proprioceptive training, and balance & exercise training) (Cakt et al., 2010; Cattaneo, Jonsdottir, Zocchi, & Regola, 2007; Prosperini, Leonardi, De Carli, Mannocchi, & Pozzilli, 2010; Salem et al., 2010) and pharmacological agents (e.g. disease-modifying drugs, such as interferon beta-1a, interferon beta-1b, glatiramer acetate, and natalizumab) (Rice et al., 1999; Rotstein, Mamdani, & O'Connor, 2010). Relatively little emphasis in research studies to-date has been placed on investigating the effect of assistive technologies on individuals with MS. This is surprising given that this patient population frequently uses assistive mobility technologies (Finlayson, Guglielmello, & Liefer, 2001) and requires help with personal mobility both indoors and outdoors (Baum & Rothschild, 1983). Some assistive technologies believed to facilitate ambulation that have been studied in individuals with MS include ankle-foot orthoses (Bregman et al., 2010), functional electrical stimulation devices (Esnouf et al., 2010), hip flexion orthoses (Sutliff et al., 2008), and canes (Gianfrancesco et al., 2011). Many of these device types have been demonstrated to significantly improve the performance of gait or activities of daily living. (Bregman et al., 2010; Esnouf et al., 2010; Gianfrancesco et al., 2011; Sutliff et al., 2008) Walkers and wheeled-walkers, on the other hand, have not received as close of an examination in individuals with MS, despite their high prevalence of use among individuals with this disease (Finlayson, Guglielmello, & Liefer, 2001). Therefore, the only knowledge available on the effect of rollators on the gait patterns of their users has been derived from other populations.

In general, walkers have been recognized as a particularly effective gait aid, achieving high user satisfaction and promoting independence among their users. (Samuelsson & Wressle, 2008) To assist users, these devices may provide mechanical support, sensory feedback, physiological support, and balance stabilization. (Alkjaer, Larsen, Pedersen, Nielsen, & Simonsen, 2006; Bateni, Heung, Zettel, McClroy, & Maki, 2004; Eggermont, Heuvelen, Keeken, Hollander, & Scherder, 2006; Jeka, Schöner, Dijkstra, Ribeiro, & Lackner, 1997; Patla, Davies, & Niechwiej, 2004; Probst et al., 2004) However, there is a small but expanding body of evidence demonstrating that there is an increased incidence of falls in walker users, which may be related to characteristics of the device, the individual, or environment. For example, TBI patients who use these devices fall more than those who do not use them (Medley, Thompson, & French, 2006), and injuries and deaths were directly associated with walker use among the non-institutionalized U.S. population at large, according to the National Electronic Injury
Surveillance System (Charron, Kirby, & MacLeod, 1995). Similar findings regarding falls have been found in the MS population, with the odds of falling increasing fivefold if an individual with MS was using a walking aid indoors and outdoors. (Nilsagård, Denison, Gunnarsson, & Boström, 2009) Potential causes of the increased fall risk in rollator users may be that the devices excessively impose attentional demands, impede lateral compensatory balance reactions, and present a tripping hazard due to foot-device collisions. (Bateni et al., 2004; Tung, 2010; Wright & Kemp, 1992) However, these reasons are merely speculative, as there is no conclusive evidence as to which factors are specifically responsible for the increased risk of falling exhibited by walker users. More work needs to be performed to determine whether individuals with MS fall because: (1) they are intrinsically predisposed to falls, due to no fault of the walker; (2) the walkers are unable to provide adequate assistance in order to prevent the user from falling; or (3) environmental factors or hazards are acting on the walker and its user causing them to fall despite the biomechanical advantages provided by the walker. There is still much to learn about the dynamic control of balance during assisted gait, in order to improve our current interventions and develop more effective fall prevention strategies.

2.3.5 Current and Emerging Approaches to Assess Mobility

Mobility assessments are conducted in three settings: the clinic, the laboratory, and the community. The tools and procedures used in each of these settings differ, providing a glimpse into different aspects of mobility. Clinical and laboratory-based assessments have been the predominant means of collecting information; however, they are becoming increasingly complimented by assessments in the community, which offer more detail in to the everyday challenges to mobility that people with MS face.

2.3.5.1 Clinical Assessment

In a clinical setting, tests of balance control are often qualitative or ordinal. Accordingly, the data generated consists of a score or written assessment by a clinician who is observing his or her client perform a battery of performance tests. Some assessment tools that have been used to test motor function, disability level, and cognition in individuals with MS include: 'Timed-Up and Go' (TUG) test; Berg Balance Scale (BBS); 10-Metre Walk test; Modified Fatigue Impact Scale; Four Square Step Test (FSST); 12-item Multiple Sclerosis Walking Scale; Functional Reach Test; Expanded Disability Status Score (EDSS); Hauser Deambulation Index (DI);
Dynamic Gait Index (DGI); Dizziness Handicap Inventory (DHI); Multiple Sclerosis Functional Composite (MSFC) (i.e. 9-Hole Peg Test; Timed 25-Foot Walk; and Paced Auditory Serial Addition Test); and Activities-specific Balance Confidence (ABC) test. (Cattaneo, Jonsdottir, & Repetti, 2007; Cattaneo, Regola, & Meotti, 2006; Nilsagård, Lundholm, Denison, & Gunnarsson, 2009; Polman & Rudick, 2010; Powell & Myers, 1995; Salem et al., 2010) These tools are easily implemented by clinical staff. However, even for an expert clinician, the process of observing test performance remains subjective and, although the observations may be clinically-relevant (e.g. some tests can be used to help distinguish between fallers and non-fallers), they provide a limited insight into the underlying features of the gait impairment. Some of the tests have even been demonstrated to have poor sensitivity to distinguish between sub-groups of patients. (Martin et al., 2006)

2.3.5.2 Laboratory Assessment

In contrast with clinical assessments, laboratory-based measures of balance control may provide more impartial, objective, and detailed feedback on the stability issues that affect an individual's mobility. The costs associated with acquiring these measures, however, are sometimes prohibitive, thereby favouring a clinical alternative. Quantitative variables in the laboratory are often categorized according to how they describe and analyze movement. (Winter, 2004, p. 9) For example, kinematic variables are those that describe movement, irrespective of their underlying forces and are the net result of all internal and external forces acting on the individual. (Winter, 1991, p. 17) Conversely, kinetic variables describe the forces that generate his or her movement. (Winter, 2004, p. 9) Using a combination of laboratory devices, such as motion capture and force plate systems respectively, any changes that occur in these variables can be reliably tracked and compared between individuals or longitudinally within a particular person. Other commonly used laboratory apparatus include electromyography (i.e. for muscle activity) and accelerometry (e.g. for upper trunk accelerations).

2.3.5.3 Community-based Assessment and Ambulatory Monitoring

Several types of devices are available that enable the ambulatory monitoring of physical activity and walking mobility as patients conduct a typical day. Commonly used devices for this purpose are pedometers and accelerometers, both of which are reliable in individuals with MS. (Kayes et al., 2009; Motl, McAuley, Snook, & Scott, 2005; Motl et al., 2007; Rietberg, van
Wegen, Uitdehaag, de Vet, & Kwakkel, 2010; Snook & Motl, 2009) For example, pedometers can be used to measure levels of physical activity in order to understand how features of a built environment facilitate or impede the mobility of individuals belonging to a patient population. One study in individuals with MS determined that specific aspects of the community (e.g. the presence of stores within walking distance, proximity to public transit, and accessible buildings) are associated with enhanced mobility, which was inferred from step counts to quantify activity levels. (Doerksen, Motl, & McAuley, 2007) Another technology that is becoming increasingly used to assess community-based mobility is instrumented assistive mobility devices. These tools provide insight into the assistive needs of device users as they walk in different environments. In spite of the development of ambulatory monitoring technologies, however, our knowledge of gait in individuals with MS continues to stem mainly from clinic- and laboratory-based assessments. Little is known about the stability of individuals with MS, who use rollators, as they ambulate through an urban/suburban environment containing features that are common to those they encounter on a daily basis. (Keysor, Jette, & Haley, 2005; Keysor et al., 2010; Shumway-Cook et al., 2003, 2002) Developing an understanding of this element of gait in the MS population will be crucial to advancing the prevention of falls and enhancing assistive technologies to ameliorate the elevated fall risk associated with this disease.

2.4 Rationale and Objectives

Neurological deficits arising from multiple sclerosis culminate in the manifestation of physical and cognitive symptoms that affect mobility. The progressive neural degeneration leads to the eventual loss of mobility, resulting in a perpetual accumulation of disability and diminished quality of life. As a result, there is a need to prolong ambulation in individuals with MS while minimizing their already elevated fall risk.

Although emphasis in research and rehabilitation has been placed on understanding the etiology of the disease and developing clinical treatment regimens, assistive mobility devices are widely used and prescribed as interventions to enhance the mobility of individuals with MS. Rollators are particularly effective devices, providing such benefits as mechanical support, sensory feedback, physiological support, and balance stabilization. However, in spite of these supports, individuals with MS continue to experience falls during the use of assistive devices.
The generation of appropriately timed and context specific compensatory balance reactions are essential to avoid a fall (Maki & McIlroy, 1997). These reactions aim to maintain the centre of mass within the base of support through the generation of stabilizing torques at the ankle, knees, and hips with the feet in place (fixed-support) or through a change in support achieved through lower extremity and upper limb placement (change-in-support). Appropriate foot placement during walking and under circumstances of instability is recognized to be an important mechanism to avoid a fall. The positioning of the feet is susceptible to influence by elements of the walking environment, which has implications for fall risk and safety during ambulation. However, little work has been conducted to explore the effect of environmental features on the foot placement patterns of individuals with MS. Furthermore, no work has explored the impact of walking through different community-based environments, with a rollator, on the stability of individuals with MS, despite how commonly these devices are employed.

The overall objective of this work was to develop a more comprehensive understanding of the effect than an assistive mobility device has on the mobility of individuals living with multiple sclerosis. To explore this relationship, two specific objectives were formed and each was accomplished in distinct, but associated projects. The first experiment involved developing a novel technique to quantify foot placement parameters with an instrumented rollator and demonstrating the feasibility of the technique by implementing it a sample of able-bodied young adults. I hypothesized that step width could be accurately assessed as a quantitative measure of rollator-assisted gait using an on-board digital camera system. In experiment two, the objective was to investigate the relationship between foot placement characteristics and the environmental context when individuals with Multiple Sclerosis walk with a rollator. The hypothesis for this experiment was that the foot placement characteristics observed in an in-laboratory context will be different than those observed in an outdoor environment. This work has implications for the development of a better understanding of the challenges to mobility that individuals face as they ambulate through the community.
3 Experiment 1: The ambulatory monitoring of foot placement with an instrumented rollator: development, validation, and preliminary application.

3.1 Introduction

Assistive mobility devices (AMD), such as the rollator or four-wheeled walker, are widely used in community settings by many mobility-impaired populations. (Finlayson, Guglielmello, & Liefer, 2001; Russell, Hendershot, LeClere, Howie, & Adler, 1997) However, recent evidence has revealed the potential for AMDs to negatively impact the user's balance and stability. In fact, AMD use has been associated with an increased risk of functional decline resulting in hospitalization (Mahoney, Sager, & Jalaluddin, 1999) as well as an increased risk of experiencing falls in the community (Medley et al., 2006). Accordingly, there is only a limited understanding of how AMDs are used in these settings and what benefits and risks they impart on their users. To date, the understanding of AMD use has been founded predominantly on laboratory-based testing. For example, perturbation paradigms involving multi-directional platform translations have been used to provide insight into how walkers may impact the compensatory balance reactions that restore stability following a balance perturbation. (Bateni et al., 2004; Cheng, 2006) However, more recent work has become focused on the ambulatory monitoring of gait during AMD use outside of a laboratory, with the advent of instrumented devices. These studies provide a glimpse into the everyday challenges to gait that AMD users face in their communities and may help to better understand the factors responsible for falls in these individuals.

Foot placement is a particularly important element of gait to examine in community-based environments in the context of falls. Inadequate toe clearance during gait, for instance, can result in a trip. (Schulz et al., 2010) Furthermore, foot placement patterns and their variability can be influenced by such features of the walking environment as surface texture, angle of inclination, ambient lighting conditions, and the presence of attention-demanding distractions. (Al-Yahya et al., 2009; Richardson et al., 2004) Neurological patient populations, such as multiple sclerosis, often exhibit altered foot placement characteristics compared to able-bodied individuals (e.g. increased step-to-step variability) (Crenshaw et al., 2006), which may have
implications for their ability to ambulate safely through challenging environments (e.g. those with surface irregularities) (Thies, Richardson, & Ashton-Miller, 2005). Nevertheless, the influence of the environment on the foot placement patterns of these mobility-impaired populations has yet to be fully ascertained, particularly in individuals who use AMDs.

Therefore, the purpose of this experiment was to: 1) develop a foot placement detection technique that accurately quantifies step width and step width variability during the use of an instrumented rollator (i.e. iWalker); and 2) to validate this technique in a small cohort of able-bodied young adults. It was hypothesized that a technique can be developed to accurately quantify step width and step width variability during the use of a rollator that has been fitted with an on-board camera system and that this technique will be valid in a cohort of able-bodied young adults respectively. This experiment will set the foundation for future work that seeks to characterize environment-related differences in step width as individuals with multiple sclerosis walk through several community settings with the use of a rollator.

3.2 Methods

3.2.1 Component 1: Technique Development

3.2.1.1 Instrumentation

The iWalker is an instrumented rollator, developed by our team, to help to better understand community-based walker use characteristics (Figure 1). This device concurrently measures biophysical and environmental data, during use, and consists of a standard rolling walker (Invacare Dolomite AB, Sweeden) that has been fitted with four single-axis button load cells (SLB-250, Transducer Technologies, USA). Mounted to the base of each leg of the walker frame, these load cells capture walker loading characteristics. Their signals are amplified with in-line strain gage amplifiers (LCV-U5-CAB, Lorenz Messtechnik GmbH, Germany) discreetly placed underneath the walker seat. Optical encoders quantify wheel rotation and provide an estimate of rollator velocity. A portable digital video camera system (Archos Helmet Camcorder/404 Media Player, Archos, Inc., France), geared with a wide-angle lens (1.7 mm, Edmund Optics, USA), was installed on the lower front bar of the rollator. This camera, which was oriented towards the feet, provides visual information related to the rollator user's foot placement, at a sampling rate of 29.97 Hz.
3.2.1.2 Video Image Rectification

Each frame of video acquired during the initial camera calibration procedure, as well as throughout the entire duration of data collection, was rectified. This rectification process was used to correct for image distortion, which is caused by parallax error associated with the curvature of the camera lens. This parallax error is greatest at the external boundaries of the image. The correction process transforms each pixel on a video frame into pixels on a new image with less distortion. The procedure involved designing a checker board-patterned calibration instrument, composed of 3 x 3 cm squares, and placing this instrument within the frame of the walker in the full field of view of the foot camera. A brief video image of the instrument was collected in 20 different positions (e.g. varying the angle of inclination and distance from the lens), and served as an input into the Camera Calibration Toolbox for Matlab® (Bouguet, 2010), which calculated the calibration coefficient factors required to rectify each frame of video data.

3.2.1.3 Capture Volume Calibration

In order to calibrate the camera view for the analysis of foot placement information, a two-dimensional (2D) map of the three-dimensional (3D) space under the walker must be created. Consequently, I developed a unique calibration tool (Figure 2) that enabled us to compare the position of various points on the floor relative to the walker frame. The tool used passive spherical markers (diameter 2.5 cm) to create a virtual grid in the camera view (with the
X- and Y-axes representing the medio-lateral and vertical directions respectively), raised 5 cm above ground level (i.e. Y-offset). Using this grid, I created a Cartesian coordinate system with the origin located at the base of the right rear walker wheel. As such, the x- and y-axes ran along the medio-lateral and antero-posterior directions respectively. Furthermore, since I designed the calibration tool to fit securely around the rear walker wheels, I also used it as a means to aim the iWalker camera. Through aligning the tool relative to a marked position on the portable digital video system's viewer screen, I ensured that the camera view was consistent for all recordings. Accordingly, prior to each data collection with the iWalker, five seconds of video data was recorded with the calibration tool for use during post-processing. Following this brief recording, the calibration tool was removed from the camera field-of-view (FOV) (i.e. capture volume) and the iWalker was used to acquire gait data.

![Figure 2. Camera calibration tool with superimposed coordinate system.](image)

### 3.2.1.4 Marker Preparation and Placement

In order to facilitate the quantification of foot placement parameters, passive spherical markers (2.5 cm diameter) were developed. The markers were uniformly covered with fluorescent orange reflective tape to clearly distinguish them from the surrounding environment.
Prior to recording the gait of an iWalker user, the markers were placed at the tips of the toes and at the same height as the y-offset on the calibration grid (i.e. 5 cm) (Figure 3). As a result, when the y-coordinates of the toe markers in the video output (acquired during gait) aligned with the y-coordinates of a recorded calibration grid marker (acquired during the capture volume calibration stage), their x-coordinate positions could be scaled by the same coefficients to obtain the equivalent, three-dimensional values.

![Figure 3. Position of toe markers in camera field-of-view during stance.](image)

### 3.2.1.5 Filtering and Feature Extraction

Prior to extracting the position of the toe markers, the video data was first rectified to remove parallax error. Secondly, each video frame was automatically colour-filtered using video editing software (VirtualDub 1.8.1) (Lee, 2009) to accentuate the colour of the toe markers appearing in the FOV. The filtered frames were then processed using a custom-designed marker localization algorithm (Matlab) with the purpose of extracting the toe markers from the background of the video image. This algorithm read the intensity of each RGB component for every pixel in every frame of video data and determined if the component intensities of that pixel fell within the distinctive range of orange found on the toe markers. If the pixel colour was
within the correct range, then the pixel intensity would have been brightened; otherwise, it would have been considered part of the background and its intensity reduced. The outcome of this marker localization technique was then imported into video marker tracking software (Peak Motus 7.0, Vicon, UK), which was used to extract the 2D coordinate position of both of the toe markers from each video frame through locating the centroid of the marker pixels.

3.2.1.6 Data Analysis

Step width was defined as the medio-lateral distance between two consecutive foot falls. In order to estimate real-life step width using the 2D coordinate data of the toe markers, a specific position in the camera FOV had to be selected in which the stance leg was in mid-stance and the stance toe marker consistently fell within the virtual boundaries of the calibration tool. As a result, a step-by-step analysis was focused on the times at which the stance toe marker intersected the virtual x-axis (i.e. the zero-line) on the calibration tool (i.e. y=0).

First, the position of the calibration tool markers that fell within this zone of interest had to be determined. Thus, the acquired five seconds of calibration tool video data was: 1) rectified to remove parallax error with Matlab, using code generated by Bouguet (Bouguet, 2010); 2) filtered to accentuate the markers with both VirtualDub (using built-in routines) and Matlab (using my custom-made program); and 3) processed to extract the position of the calibration tool markers, which were located on the zero-line, using PeakMotus (using built-in routines). The resulting coordinates were then averaged together for each marker. The average medio-lateral coordinate distance between each of these markers and the right-most marker, situated on the y-axis (i.e. the origin (0,0), in this case), was computed for the entire five second video. Since these virtual coordinate distances corresponded to known three-dimensional distances on the tool, a unique scaling coefficient was established by finding the quotient of the actual distance (in cm) and theoretical distances (in 2D pixels).

Afterwards, for each video frame of walking data in which the rollator user's stance foot crossed the zero-line of the coordinate system, the 2D coordinate distance between the toe markers and the origin of the coordinate system was calculated. This value was then scaled to reflect its three-dimensional value using the y-axis-specific scaling coefficient. To complete the calculation, the difference between scaled-2D distances for each consecutive foot fall was determined and considered to be representative of the rollator user's step width (Figure 4).
3.2.2 Component 2: Technique Validation in Able-bodied Young Adults

3.2.2.1 Participants

Five able-bodied young adults (mean age 27.5 ± 1.5 years; 2 males, 3 females) were recruited to participate in this study. These individuals had no history of neurological or musculoskeletal disorders nor did they possess any mobility impairments. Each participant was notified of the purpose of the study as well as the details of the experimental protocol, which had been approved by the Toronto Rehab Research Ethics Board. Written informed consent was obtained prior to their participation.

3.2.2.2 Experimental Procedure

The iWalker camera-based step width algorithm was validated using a laboratory-based motion capture system (Vicon). To accomplish this, a cross-sectional validation study was performed, in which each of the five able-bodied participants took part in one laboratory visit. The participants were instructed to walk across a 7-metre in-laboratory pathway three times, at their individual preferred walking speeds. Throughout these trials, their foot placement patterns were simultaneously recorded using the iWalker's foot camera as well as a seven-camera VICON® MX Motion Capture System (Vicon, CO, USA) (with a sampling rate of 100 Hz), as a reference. The Vicon marker set consisted of the same two custom-made markers (used with the iWalker-based technique) placed on the forefoot of both lower limbs, as displayed in Figure 3.
Since none of these individuals were rollator users, they were provided with an opportunity to practice walking with the rollator before commencing the trials, so that they could familiarize themselves with using the device and overcome the learning curve associated with first-time device use.

3.2.2.3 Data Analysis

For the iWalker camera data, step widths were calculated using the method described above. The motion capture data was processed using VICON® Nexus software and involved labeling the toe markers on each foot (i.e. the same markers used for the iWalker-based analysis), extracting the three-dimensional coordinate position of the marker data, determining the distance between the toes and walker frame as the user's stance foot crossed the line intersecting the back walker wheels, and calculating the medio-lateral distance between two consecutive foot falls relative to the walker frame. This calculation would be considered representative of the true value of the individual's step width and was performed using Microsoft Office Excel 2007. Since this calculation used only positional information at discrete time points, it was not necessary to filter the coordinate data output from the VICON® system nor to synchronize the iWalker and Vicon recordings. Subsequently, step width variability was calculated as the standard deviation of the step width values for each participant.

On average, each participant performed five, steady-state walking steps [i.e. steps not associated with gait initiation (the first three steps) (Miller & Verstraete, 1996) or termination (the last two steps) (Miff, Childress, Gard, Meier, & Hansen, 2005; Sparrow & Tirosh, 2005)] during each of the three trials. As a result, for each individual, approximately twelve step width calculations (i.e. four step widths calculated from the five steps of each walking trial) were performed for both the iWalker and Vicon camera systems, and the system-specific results were averaged together. This provided a total of sixty data points overall, with which to compare the two methods. First, an intra-class correlation coefficient [ICC(3,k)] was computed, using a two-way mixed effects model with an absolute agreement approach. As with many clinical measurements, reasonable validity would be assumed if the ICC exceeded 0.90. (Portney & Watkins, 2009) Secondly, the root-mean-square (RMS) difference was calculated to assess the spread of the variability of differences in pairs of measurements from the two techniques, thus providing not only a measure of agreement, but also a value for the measurement error between
the techniques. In order for the iWalker-based technique to be capable of detecting meaningful changes, the RMS difference must be smaller than the environment-related changes one would expect to observe. Lastly, a Bland-Altman analysis was performed in order to further assess the level of agreement between the Vicon and iWalker techniques as well as to detect bias or any relationships between the variance in measures and the size of the mean. It was presented as a plot of the difference against the mean for the two measurement techniques, along with the mean difference between pairs of measurements (i.e. the bias) as well as the upper and lower 95% confidence intervals (i.e. 1.96 times the standard deviation of the measurement results). Agreement would be confirmed visually if no considerable bias, trends, or heteroscedasticity (i.e. when random data points have different variances) are evident in the data points on the plot.

3.3 Results

3.3.1 Validation in Able-Bodied Young Adults

A strong level of agreement between the Vicon motion capture system and iWalker foot camera-based technique was established and corroborated with a Bland-Altman analysis. Figure 5 depicts the resulting Bland-Altman plot. The range of agreement between the step width estimations of the two systems (i.e. the mean bias of the systems ± 1.96 standard deviations) had a maximum of 1.43 cm and minimum of -1.38 cm. Each data point on the plot corresponds to the average step width of each participant (x-axis) relative to the average difference between pairs of measurements from the two techniques for all observations on a participant (y-axis).

Furthermore, the level of concordance between the Vicon and iWalker calculations was very high. The intra-class correlation coefficient was determined to be 0.999 (95% CI: 0.987 - 1.000, P <0.001) using a two-way mixed effects model. Table 1 exhibits the mean step width and step width variability results generated by both systems. The step widths estimated using the iWalker method were highly comparable to those generated by the Vicon system, with similar variability, as the two methods differed in these parameters by only 0.03 cm and 0.27 cm respectively.

Overall, the average root-mean-square (RMS) difference between the two methods was 0.70 cm. A scatter plot of step width values computed by both the Vicon and iWalker systems for one representative participant is presented in Figure 6.
Figure 5. Difference against the mean (Bland-Altman Plot) for the iWalker and Vicon calculations of step width in able-bodied young adults.

Table 1. Means and standard deviations of foot placement parameters for the devices during in-laboratory testing on able-bodied young adults.

<table>
<thead>
<tr>
<th>Devices</th>
<th>Mean Step Width (range) [cm]</th>
<th>Mean Step Width Variability (range) [cm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>iWalker</td>
<td>$14.40 \pm 4.64 \text{ (9.82-21.85)}$</td>
<td>$2.64 \pm 0.36 \text{ (2.34-3.22)}$</td>
</tr>
<tr>
<td>Vicon</td>
<td>$14.37 \pm 4.34 \text{ (10.13-21.22)}$</td>
<td>$2.37 \pm 0.43 \text{ (1.83-2.94)}$</td>
</tr>
</tbody>
</table>
3.4 Discussion

In this study, my objective was to design a foot placement detection technique that would enable us to quantitatively track foot placement during rollator use and validate it against a high-precision Vicon motion capture system in a cohort of able-bodied young adults. I determined that my technique was valid, according to a Bland-Altman analysis, the result of the intra-class correlation, and the RMS difference between systems. The mean error of the iWalker-based technique was 0.70 cm, which was the average RMS difference between the iWalker- and Vicon-based step width calculations. This error was in the same range as that of instrumented walkway technologies commonly used to quantify the spatiotemporal characteristics of gait (e.g. GAITRite, with 0.51-0.66 cm error) (Selby-Silverstein & Besser, 1999), and in some cases, it was even smaller (e.g. GaitMatII, with 1.17 cm error) (Barker, Craik, Freedman, Herrmann, & Hillstrom, 2006). Compared to stationary video-based motion tracking systems, my mean error was expectedly larger (e.g. the Peak Motus system and Rivermead video-based clinical gait analysis method (RIVCAM) have errors of 0.091 cm (Richards, 1999) and 0.30 cm (Churchill, Halligan, & Wade, 2002) respectively). Also, as expected, my mean error exceeded that of three-
dimensional optoelectronic motion capture systems. For example, the Vicon-460 and Optotrack 3020 systems exhibit differences of only 0.063 mm (Windolf, Götzen, & Morlock, 2008) and 0.03 mm respectively (Maletsky, Sun, & Morton, 2007) compared to rigid bodies that are precisely controlled by a robot or using a machinist's rotary table. Although the error associated with my technique was considerably larger than these gold-standard systems, the calculated difference constituted a small proportion of the expected variations in support-base gait parameters of healthy (Barker et al., 2006) and patient (Gutierrez et al., 2005) populations (e.g. 2.9-5.2 cm, for normal adult gait, and 4.0-9.0 cm, for MS gait). It was also evident that my approach differed from that of other rollator-based computer vision systems, such as the method put forward by Ng et al. (Ng, Fakih, Fournier, Poupart, & Zelek, 2009). My technique was focused on acquiring a more focal measure of foot placement from the on-board video recordings, whereas the other method sought to achieve a global characterization of lower limb position.

The observed difference between the results of my technique and the Vicon system (i.e. 0.70 cm) may be attributed to a number of factors. First, it is important to differentiate between the marker localization strategies performed by the Vicon and iWalker methods. The Vicon cameras contained filters that permitted the detection of only light that was the colour of the camera's stroboscopic diodes (i.e. red and infra-red) and had been reflected off of the spherical markers (Tebbutt, Wood, & King, 2002). The iWalker camera, on the other hand, captured the full spectrum of visible light within the capture volume, but utilized a custom filtering program during post-processing to emphasize the colour of the markers (i.e. orange) and effectively extract them from the background image. As well, since the iWalker's camera was fitted with a wide-angle lens, the video data recorded by the iWalker was inherently more distorted by parallax error, associated with the curvature of the lens. The effect of this factor, however, was minimized by rectifying every acquired image of iWalker video data with the camera calibration toolbox for Matlab (Bouguet, 2010). It is also probable that the difference between the two systems is, in part, a consequence of the iWalker camera using a mobile axis system as opposed to the Vicon system's stationary coordinate system, fixed to the laboratory. The method with which the actual visual environment was calibrated using the iWalker camera may have an effect. Once the calibration process with my unique calibration tool was complete, the accuracy of the results produced by my technique was sensitive to any unanticipated movements of the
walker camera relative to the walker frame. This was controlled for by ensuring that camera was firmly mounted to the frame, thus minimizing its effect. Nevertheless, the iWalker-based foot placement detection technique still enabled the accurate tracking of quantitative foot placement parameters during rollator use outside of laboratory environment.

The observed step width positions obtained using my technique were similar to those observed in previous studies, i.e. for individuals in their mid-twenties who were able-bodied (Grabiner & Troy, 2005; Rosenblatt & Grabiner, 2010). The values varied, however, depending on the anatomical placement of the foot markers (Owings & Grabiner, 2004; Rosenblatt & Grabiner, 2010), which ultimately influenced the outcome of the step width (SW) calculation, as well as whether the participants' gait patterns were being observed on a treadmill or over-ground (Grabiner & Troy, 2005). The average SW that I obtained in this study, which is 14.40 cm, corresponded most closely to the results presented in the study by Grabiner et al. (Grabiner & Troy, 2005), which reported a SW of 15.2 cm. That study was conducted using a treadmill, which is known to induce wider step widths (Rosenblatt & Grabiner, 2010), and may account for some of the perceived difference. Furthermore, my average SW calculation appeared to be slightly larger than that reported by the over-ground walking study performed by Rosenblatt et al. (Rosenblatt & Grabiner, 2010), which obtained a SW of 11.48 cm. In that study, the toe marker used for the SW calculation was placed more medially than in this study (i.e. on the second metatarsal as opposed to the third), which may have contributed to the discrepancy. There were also reports of much smaller SW calculations, such as those of Owings et al. (Owings & Grabiner, 2004) with a value of 8.1 cm. However, since that study calculated SW using the position of the participants' heel markers, it is not surprising that their SWs were smaller than those in this study. In terms of step width variability (SWV), the values acquired in this study (i.e. average of 2.64 ± 0.36 cm) were similar to those reported by Rosenblatt & Grabiner (2010) (i.e. 2.19 ± 0.39 cm and 2.33 ± 0.58 cm). Therefore, my results for step width and step width variability were deemed to be externally valid.

Future directions for this work will include further refining the algorithm (e.g. exploring marker-less tracking techniques to improve the ecological validity of the results), upgrading the camera system or adding additional cameras, expanding to a sample population of individuals with MS, and increasing the number of environmental settings under investigation. The development of novel measures of foot placement will also be explored, specifically measures
that are walker-specific may be more relevant to report when examining community-based rollator use. An example of such a measure involves using the medio-lateral distance between stance foot position and the most proximal walker wheel as an indicator of collision risk.

3.5 Conclusions

In this study, I developed an accurate technique with which to quantify foot placement parameters during the use of an instrumented rollator. I also validated the technique in a small cohort of able-bodied young adults. As a result, this technique can be used outside of a laboratory to help discover differences in medio-lateral foot placement behaviour based on environmental context.
Experiment 2: Assessing the Medio-Lateral Foot Placement Patterns of Individuals with Multiple Sclerosis during Rollator-Assisted Community Mobility

4.1 Introduction

Multiple sclerosis (MS) is a chronic degenerative, autoimmunogenic disorder that affects the central nervous system (CNS) (Stüve & Oksenberg, 2010). The compromised function of the CNS results in sensory, motor, and cognitive (Boissy & Cohen, 2007; Cattaneo et al., 2002; Franklin, Nelson, Filley, & Heaton, 1989; Gutierrez et al., 2005; Hamilton, 2009; Kalron, Dvir, & Achiron, 2010; Souza et al., 2010) defects. Individuals with MS, therefore, often experience ataxia as well as difficulties with balance, coordination, postural control, and gait mechanics (Ashburn, 1988; Frzovic, Morris, & Vowels, 2000; Gutierrez et al., 2005). The gait deviations induced by MS have been linked to a higher risk of falling (Cameron & Lord, 2010; Cattaneo et al., 2002; Finlayson, Peterson, & Cho, 2006; Nilsagård, Denison, Gunnarsson, & Boström, 2009). Thus, gait deficits are associated with a poor quality of life in individuals with MS (Nogueira, Nóbrega, Lopes, Thuler, & Alvarenga, 2009) and play a role in the progression of disability that characterizes this disease (Compston & Coles, 2008).

Assistive mobility devices (AMDs) are a common form of intervention employed by individuals with MS. They help to overcome functional limitations and improve mobility. Walkers have long been recognized as a particularly effective AMD, achieving high user satisfaction, improving walking endurance, and promoting independence (Gupta, Goldstein, & Brooks, 2006; Samuelsson & Wressle, 2008). Paradoxically, the use of AMDs has been identified as a significant risk factor for falling in community-dwelling adults (Medley et al., 2006). This unexpected contradiction calls for further research to better understand the relationship between AMD users and their devices.

Regardless of AMD use, appropriate foot placement with every step is critical to avoid a fall. For example, tripping is one outcome of inadequate toe clearance and other errors in foot placement. Step width, in particular, is an important gait characteristic to investigate. Defined as the medio-lateral distance between the feet during two consecutive foot falls, this parameter plays a role in maintaining frontal plane stability, which is crucial to balance (MacKinnon &
Winter, 1993). In fact, both increases and decreases in step width and step width variability have been associated with a higher risk of experiencing falls in older adults (Brach, Berlin, VanSwearingen, Newman & Studenski, 2005; Maki, 1997). As well, given the limited medio-lateral space in which to place the feet between the back wheels of a walker, this parameter may have implications for tripping in rollator users. Foot placement, in general, can be influenced by several factors occurring in an individual's walking environment, in both indoor (Carter et al., 2000; Gill et al., 2000) and outdoor (Carter et al., 2000) settings. The current understanding of foot placement and gait patterns in individuals with MS stems mainly from laboratory- and clinic-based assessments. These environments contain few of the factors that have been demonstrated to alter gait patterns that would be typically encountered in a community where many of these individuals reside, such as having to ambulate across walking surfaces that are inclined and contain irregularities (Marigold & Patla, 2008; McIntosh, Beatty, Dwan, & Vickers, 2006). As a result, little is known about the everyday challenges to mobility for individuals with MS, particularly for those who use rollators (i.e. rolling walkers), as they ambulate through common environments (Keysor et al., 2005; Keysor et al., 2010; Shumway-Cook et al., 2003, 2002). Therefore, since a large proportion of falls occur outside of an institutional setting, it would be important to identify environment-related changes in stepping behaviour during community ambulation in order to begin to understand how features of the walking environment influence gait. This knowledge would be beneficial to the development of more effective fall prevention strategies (e.g. enhanced AMDs, gait training programs, etc.).

The objective of this study was to investigate the relationship between medio-lateral foot placement characteristics and the environmental context when individuals with MS walk with an assistive mobility device. It was hypothesized that foot placement characteristics observed in a laboratory context will be different than those observed outside of a laboratory. In particular, step width and step width variability will be increased outside of a laboratory. This examination will provide insight into the everyday gait patterns of individuals with MS who use assistive mobility devices, in order to better understand the lower limb strategies they use in response to challenges in the environment. As a result, this work will be useful for further research into factors leading to falls in the community for individuals living with MS.
4.2 Methods

4.2.1 Study Population

In this study, ten community-dwelling adults with MS, residing in the Greater Toronto Area, were either recruited through referrals of a clinician at the Elkie Adler MS Clinic or had responded to a printed advertisement the electronic newsletter of the local chapter of the MS Society of Canada. The sample consisted exclusively of women between the ages of 38 and 67. For inclusion, the participants were required to have: 1) received a medical diagnosis of MS; 2) been adult (i.e. over 18 years of aged); and 3) reported daily use of a rollator. Participants were excluded from participating in the study if they were unable to communicate in English or if they possessed cognitive impairments that precluded them from understanding instructions or providing written informed consent. Ethical approval was obtained from the University of Toronto (UofT) and the Toronto Rehabilitation Institute (TRI) and all participants signed an approved informed consent form prior to commencing with the study. Participant characteristics are displayed in Table 2.

<table>
<thead>
<tr>
<th>Participant</th>
<th>Sex (M/F)</th>
<th>Age (Years)</th>
<th>Height (cm)</th>
<th>Mass (kg)</th>
<th>Clinical Phenotype of MS</th>
<th>Time Since Diagnosis with MS (Years)</th>
<th>Penn Spasm Frequency Scale Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>F</td>
<td>38</td>
<td>158</td>
<td>54</td>
<td>SPMS</td>
<td>5</td>
<td>2</td>
</tr>
<tr>
<td>2</td>
<td>F</td>
<td>39</td>
<td>180</td>
<td>112</td>
<td>RRMS</td>
<td>4</td>
<td>2</td>
</tr>
<tr>
<td>3</td>
<td>F</td>
<td>58</td>
<td>159</td>
<td>49</td>
<td>SPMS</td>
<td>16</td>
<td>0</td>
</tr>
<tr>
<td>4</td>
<td>F</td>
<td>67</td>
<td>157</td>
<td>82</td>
<td>PPMS</td>
<td>11</td>
<td>1</td>
</tr>
<tr>
<td>5</td>
<td>F</td>
<td>59</td>
<td>160</td>
<td>48</td>
<td>SPMS</td>
<td>33</td>
<td>2</td>
</tr>
<tr>
<td>6</td>
<td>F</td>
<td>53</td>
<td>165</td>
<td>54</td>
<td>SPMS</td>
<td>21</td>
<td>2</td>
</tr>
<tr>
<td>7</td>
<td>F</td>
<td>60</td>
<td>157</td>
<td>50</td>
<td>SPMS</td>
<td>39</td>
<td>2</td>
</tr>
<tr>
<td>8</td>
<td>F</td>
<td>60</td>
<td>165</td>
<td>46</td>
<td>SPMS</td>
<td>41</td>
<td>1</td>
</tr>
<tr>
<td>9</td>
<td>F</td>
<td>45</td>
<td>155</td>
<td>102</td>
<td>RRMS</td>
<td>12</td>
<td>1</td>
</tr>
<tr>
<td>10</td>
<td>F</td>
<td>60</td>
<td>155</td>
<td>66</td>
<td>SPMS</td>
<td>14</td>
<td>1</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Mean ± SD</th>
<th>Sex (M/F)</th>
<th>Age (Years)</th>
<th>Height (cm)</th>
<th>Mass (kg)</th>
<th>Clinical Phenotype of MS</th>
<th>Time Since Diagnosis with MS (Years)</th>
<th>Penn Spasm Frequency Scale Score</th>
</tr>
</thead>
<tbody>
<tr>
<td>53.9 ± 9.9</td>
<td>M:F</td>
<td>161.1 ± 7.5</td>
<td>66.3 ± 24.1</td>
<td>7 SPMS</td>
<td>19.6 ± 1.4</td>
<td>1.4 ± 0.7</td>
<td></td>
</tr>
</tbody>
</table>

N.B.: SPMS = secondary-progressive MS; RRMS = relapsing-remitting MS; PPMS = primary-progressive MS.
4.2.2 Study Design

A cross-sectional study was conducted to: 1) validate a rollator-based foot placement detection technique, developed by Chee et al. (2010), for use in the MS population; and 2) assess the medio-lateral foot placement characteristics of the participants as they ambulated through several community-based environments with a rollator. The study protocol was implemented in the gait laboratory located at the University of Toronto Rehabilitation Sciences Building and its immediate vicinity in downtown Toronto.

Participants were initially assessed in the gait laboratory to collect demographic information (e.g. age, height, and mass) and clinical data (i.e. clinical phenotype, years since diagnosis, and level of spasticity). Spasticity levels were documented using the Penn Spasm Frequency Scale (PSFS), a self-report measure in which participants rate their perceptions of spasticity based on the frequency of muscle spasms, using a 5-point ordinal scale (Penn, 1988; Priebe, Sherwood, Thornby, Kharas, & Markowski, 1996). This measurement tool is commonly used by clinicians and researchers to characterize individuals with MS. (Meca-Lallana, de Mingo-Casado, Amorín-Díaz, Martínez-Navarro, & Barreiro, 2010)

4.2.2.1 Component 1: Validation of foot placement detection technique in MS population

Participants performed one pass of a 7-metre laboratory-based walkway while using an instrumented rollator (i.e. iWalker). Their step widths were recorded simultaneously using both a video camera mounted on-board the iWalker and the laboratory motion capture system. The iWalker-based foot detection technique that was developed has undergone preliminary validation testing in able-bodied young adults. This testing has demonstrated a high level of agreement between the iWalker and Vicon measures (Chee, Tung, Gage, McIlroy, & Zabjek, 2010).

4.2.2.2 Component 2: Assessment of foot placement during a community-based course walk.

In the course walk component, step width, step width variability, and walking velocity were measured with the iWalker as participants used the device in four outdoor community settings: 1) urban sidewalk; 2) up-ramp; 3) down-ramp; and 4) busy intersection crossing (Figure 7). Each of these settings were encountered in the same sequence by all participants (i.e.
randomization of order was not possible), as they were guided along a pre-determined course in the vicinity of the UofT and TRI precincts. The participants were instructed to walk at a comfortable speed and the assessors offered no encouragement, in order to avoid influencing gait speed. Frequent, seated breaks were enforced in order to avoid fatiguing the participants.

4.2.3 Instrumentation

A seven-camera VICON® MX Motion Capture System (Vicon, CO, USA), sampling at 100 Hz, was used to capture three-dimensional coordinate data for the in-laboratory validation of the iWalker method of calculating step width. The Vicon marker set consisted of the same two custom-made markers (used with the iWalker-based technique) placed on the forefoot of both lower limbs. The iWalker, which was used both in and out of the laboratory, concurrently measured biophysical and environmental data during its use and consisted of a standard rolling
walker (Invacare Dolomite AB, Sweeden) that had been fitted with a portable digital video camera system (Archos Helmet Camcorder/404 Media Player, Archos, Inc., France), geared with a wide-angle lens (1.7 mm, Edmund Optics, USA), and installed on the lower front bar of the rollator. This camera, which was oriented facing the participant’s feet to collect foot placement information (Figure 8), operated at a sampling rate of 29.97 Hz. An optical encoder, installed on one of the wheels, was used to quantify the average velocity at which the walker was being pushed (i.e. the distance travelled as measured using the number of wheel rotations measured by the optical encoder, over the time time taken to perform the steps observed in each environment).

4.2.4 Data Analysis

The mean and standard deviation of the demographic and clinical characteristics were calculated. During the validation component, five consecutive step widths were calculated for each participant, with Microsoft Office Excel 2007, using the technique developed by Chee et al., which uses markers on the middle toe of each foot (Chee et al., 2010). These step widths were compared to the output of the Vicon system. Since the calculations used only positional information at discrete time points, it was determined that the principles of standard filtering techniques that are applied for time series data were not relevant to apply. It is recognized that if
additional analysis focused on the measures of velocity and acceleration derived from this positional data, that the trajectory of the markers over the duration of the sampling period would be reconstructed and filtered. An intra-class correlation coefficient [ICC(3,k)] was computed for the two systems (i.e. iWalker and Vicon), using a two-way mixed effects model with an absolute agreement approach. As well, a Bland-Altman analysis was performed in order to assess the level of agreement between the systems. In the course walk component, six consecutive step widths of steady-state walking [i.e. steps not associated with gait initiation (the first three steps) (Miller & Verstraete, 1996) or gait termination (the last two steps) (Miff et al., 2005; Sparrow & Tirosh, 2005)] were calculated in each of the four community settings using the iWalker. Step width variability was also calculated in each setting as the standard error of the mean (i.e. the standard deviation divided by the square root of the number of observations). A one-factor repeated-measures analysis of variance (ANOVA) was conducted to compare the effect of the five environments (i.e. in-lab and the four out-of-lab settings) on step width and step width variability (α < 0.05). If the environmental effect was significant, a post-hoc analysis comprising of pairwise comparisons using Bonferroni-adjusted t tests would be performed.

4.3 Results

4.3.1 Clinical characteristics

Clinical characteristics of the participants in this study are provided in Table 2. The majority of participants were diagnosed with secondary-progressive MS [SPMS] (n=7); the remainder were diagnosed with relapsing-remitting MS [RRMS] (n=2) or primary-progressive MS [PPMS] (n=1). Levels of spasticity ranged from none (PSFS 0) (n=1) to spasms occurring less than once per day (PSFS 2) (n=5). Many participants reported that spasms were induced only by stimulation (e.g. extreme temperatures or prolonged exercise) (PSFS 1) (n=4). Nine of the ten participants (n=9) used walking aids daily in the course of their lives, identifying the rollator as their primary device. The remaining participant (n=1) used a rollator primarily for long distance outdoor walking or during aerobic exercise in order to increase endurance. Half of participants (n=5) had been prescribed a rollator through an assistive technologies clinic for individuals with MS. The other participants (n=5) were either recommended to use a rollator by a physician and/or physiotherapist or had independently decided to use this device and the appropriateness of their decision confirmed by a health care professional. Regardless, all
participants (n=10) were well-adjusted to using rollators while walking and had user experience spanning several months to fourteen years. In addition to rollator use, three of the participants reported that they would use a cane on occasion. Another two participants sometimes used wheelchairs. Some of the participants also had to use secondary gait aids to ambulate safely, including a concealed ankle-foot orthosis or air cast (n=3) and orthotic shoes (n=1), which they wore throughout testing.

4.3.2 Validation of the foot placement detection technique.

A strong level of agreement between the Vicon motion capture system and the iWalker foot camera-based technique was observed. The intra-class correlation coefficient was determined to be 0.997 (95% CI: 0.982 - 0.999, p < 0.001). Overall, the average root-mean-square (RMS) difference between the two methods was 0.72 cm. Figure 9 depicts the resulting Bland-Altman plot. The range of agreement between the step width estimations of the two systems (i.e. the mean bias of the systems ± 2 standard deviations) extended from approximately -1.13 cm to 1.69 cm.

![Figure 9. Difference against the mean (Bland-Altman Plot) for the iWalker and Vicon calculations of step width in individuals with MS.](image-url)
4.3.3 Spatio-temporal gait characteristics during course walk.

Figures 10 and 11 display the distribution of step width and step width variability respectively in each of the five walking environments. There was no significant effect of the environment on step width \(F[4, 28] = 1.317, p = 0.288\). However, the walking environment did have a significant effect on the step width variability of the participants \(F[4, 28] = 7.224, p < 0.001\). Table 3 reports step width variability in each of the environments when it is computed as the standard deviation, standard error of the mean, and coefficient of variation. Post-hoc analysis (pairwise comparisons using Bonferroni-adjusted t tests) revealed differences between the in-lab and up-ramp conditions \(p = 0.012\), and the up-ramp and down-ramp conditions \(p = 0.027\). This analysis involved the results of only eight participants. Two participants were excluded because they were unable to complete all sections of the walking course. In addition, the velocity of the participants as they traversed each of the environments was determined and is depicted in Figure 12. The walking environment had a significantly effect on velocity \(F[4, 28] = 12.184, p < 0.001\). Post-hoc comparisons revealed that participants walked faster during intersection crossing than in either the in-lab \(p = 0.001\) or up-ramp \(p = 0.012\) environments.

![Figure 10. Distribution of step width in each environment, as depicted by the sample minimum, lower quartile (Q1), median (Q2), upper quartile (Q3), and sample maximum.](image-url)
Figure 11. Distribution of step width variability in each environment.
Note: * denotes statistical significance ($p < 0.05$)

Table 3. Values of step width variability (mean ± standard deviation) in each of the environments, calculated as the standard deviation, standard error of the mean, and coefficient of variation.

<table>
<thead>
<tr>
<th>Environment</th>
<th>Average Number of Samples per Participant</th>
<th>Standard Deviation (mm)</th>
<th>Standard Error of the Mean (mm)</th>
<th>Coefficient of Variation (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>In-Lab</td>
<td>5</td>
<td>9.8 ± 4.3</td>
<td>4.5 ± 1.8</td>
<td>5.8 ± 2.2</td>
</tr>
<tr>
<td>Sidewalk</td>
<td>6</td>
<td>17.7 ± 7.6</td>
<td>7.1 ± 3.0</td>
<td>11.8 ± 7.1</td>
</tr>
<tr>
<td>Up-Ramp</td>
<td>6</td>
<td>24.4 ± 8.4</td>
<td>10.1 ± 4.1</td>
<td>14.5 ± 5.0</td>
</tr>
<tr>
<td>Down-Ramp</td>
<td>7</td>
<td>13.8 ± 4.6</td>
<td>5.4 ± 2.1</td>
<td>8.7 ± 4.4</td>
</tr>
<tr>
<td>Intersection Crossing</td>
<td>6</td>
<td>12.7 ± 7.3</td>
<td>5.0 ± 2.8</td>
<td>8.9 ± 7.3</td>
</tr>
</tbody>
</table>
4.4 Discussion

The objective of this study was to determine if medio-lateral foot placement characteristics (i.e. step width and step width variability) are influenced by the walking environment as individuals with MS use a rollator. The iWalker-based method of calculating step width in the MS population was valid, with a strong level of agreement with the Vicon system (according to a Bland-Altman analysis), an intra-class correlation coefficient of 0.997, and a RMS difference of 0.72 cm. Step width variability, but not step width, was influenced by the walking environment. Significant differences were detected between the in-lab and up-ramp conditions as well as between the up-ramp and down-ramp conditions. Participants also exhibited a significant increase in walking velocity when crossing an intersection compared to when they were walking either in-lab or up-ramp environments.
The sample of individuals with MS used in this study had a higher proportion of assistive device users than others typically characterized in the literature, such as in the study by Nilsagård et al. (2009). As well, the participants in the current study demonstrated a reduced walking velocity than both the able-bodied participants and individuals with MS reported by Martin et al. (2006). This was to be expected because rollator users have been demonstrated to exhibit a reduced walking speed (Liu, McGee, Wang & Persson, 2009) and diminished ambulatory capacity (Kelleher, Spence, Solomonidis, & Apatsidis, 2010). Reasons for prescribing a rollator to individuals with MS may include: 1) decreasing weight bearing; 2) decreasing pain; and/or 3) to account for balance deficits. Nevertheless, spasticity levels were similar to those reported in other studies (Meca-Lallana et al., 2010). In terms of clinical categorization, the proportion of participants with a PPMS course of the disease (10%) was quite representative of the population at large (Stüve & Oksenberg, 2010). However, the proportion of participants with a SPMS course, relative to a RRMS course (i.e. 7 SPMS:2 RRMS), was well over the 1:1 proportion that would be expected for a sample of individuals initially diagnosed over ten years prior. (Stüve & Oksenberg, 2010) A mean of 19.6 ± 13.5 for the number of years since initial diagnosis may be responsible for this imbalance.

4.4.1 The effect of environment on medio-lateral foot placement during rollator use

4.4.1.1 Effect on step width

This is the first study to observe the foot placement patterns of individuals with MS as they use an assistive mobility device in community environments. During rollator use, step width is a vital parameter to observe due to the potential of the feet to come into contact with the back wheels of the walker, leading to a possible trip and/or loss of balance. It has also been suggested that step width is a measure of lateral balance control and had been positively associated with falls in older adults. (Maki, 1997) Human walking requires active stabilization in the frontal plane and medio-lateral foot placement is an effective means to provide this stability. As a result, humans tend to avoid taking narrow steps because narrow steps are less stable. This reduced stability may be derived from such factors as the greater metabolic cost of walking at very narrow step widths and the smaller margin of safety between the centre-of-mass and centre-of-pressure during narrow-base walking. (Donelan, Shipman, Kram, & Kuo, 2004)
The step widths exhibited by my cohort was in the same range as previously reported values for individuals with MS walking on level-ground without a rollator, with similar step width variability (e.g. 17.0 ± 0.98 cm vs. 11.6 ± 0.5 cm in Givon et al. (2009) and 19 ± 3 cm in Gutierrez et al. (2005)). Compared to level-ground walking, previous studies with passive mechanical walking systems (such as the models used by Kuo et al. (1999) and Bauby et al. (2000)) have shown that step width adjustments (i.e. increases in step width) can be an efficient and effective means to maintain stability without the assistive support, in situations that are less stable, such as when descending a slope (Bauby & Kuo, 2000; Kuo, 1999). However, the step widths of my participants were unchanged between the laboratory setting and the four outdoor environments studied, despite the challenges to gait in these settings (e.g. ramps, which can lead to slips or trips). It is, therefore, possible that step width is dependent on environment, but is not affected because of the effect of the walker on gait. Any increases in step width that were caused by environmentally-induced instability could have been offset by the tendency to decrease step width in response to: a) mechanical stabilization (e.g. increased base of support), which was provided by the rollator; or b) spatial constraint, which may have been imposed by the rollator frame. It is also possible that the task challenges in each of my environments were not as difficult as the challenges performed in studies that reported significant differences due to environment. If this is the case, then a different set of conditions (i.e. community environments containing more difficult challenges) may have resulted in significant changes in step width. A third possibility is that step width is not dependent on the environment, similar to the findings of Kawamura et al. (1991) in which neither up- nor down-slope walking modified the medio-lateral placement of two consecutive foot falls (i.e. stride width, as it is referred to in that study). Further work on discerning walker loading patterns during gait in these environments may help to determine if the walker is used to compensate for instability as opposed to the feet (e.g. through the use of the upper body to control frontal plane trunk accelerations, such that variation in step width would not be necessary to maintain stability).

4.4.1.2 Effect on step width variability

Both features of the walking environment (e.g. surface irregularity) (Thies, Richardson, & Ashton-Miller, 2005) and characteristics of an individual (e.g. muscle strength, cognitive function, postural control, proprioception, and walking speed) have been demonstrated to have an effect on step width variability. (Callisaya, Blizzard, McGinley, Schmidt, & Srikanth, 2010;
Kang & Dingwell, 2008) In the present study, the changes in step width variability observed may be related to aspects of the pathology of the participants. For example, individuals with MS often experience muscle weakness, which can affect the hip (Sutliff et al., 2008). The role of the hip abductors during gait is of particular interest, as they are used to raise the pelvis while the contralateral leg is in swing and also contribute to achieving an appropriate step width, which helps to control the medio-lateral stability of the centre-of-mass. During ramp ascent, additional work must be performed in order to raise the centre-of-mass to a higher level and the hip musculature is particularly important for accomplishing this task. Although increased hip extensor activity helps with forward propulsion, hip abduction is also important as it affects pelvic position and consequently toe clearance during swing (i.e. by raising the pelvis). Weak hip abductors would, therefore, make it more difficult for an individual to attain an adequate toe clearance to successfully accommodate the raised walking surface. It may also result in reduced control of the medio-lateral positioning of the feet, thus affecting step width variability. During level-ground walking with a rollator, optimal position of the upper extremities that are in contact with the walker handles will assist in the generation of stabilizing torques to support the head, arms and trunk and decrease the torque required by the hip abductors to control pelvic position, frontal plane motion of the centre of mass, and lower limb trajectory. However, it is postulated that during ramp ascent, the optimal contribution of the upper extremities to supporting the head, arms and trunk may be compromised due to changes in the postural demands of the task (e.g. increased forward inclination of the trunk) (Leroux, Fung, & Barbeau, 2002). As a result, the walker may be held farther in front of the user and his or her step width variability would be increased, as was observed in the individuals with MS. On the contrary, during ramp descent, the rollator may be used strategically to help absorb some of the kinetic energy (e.g. through use of the brakes and by holding it closer to the body in combination with a backward tilt of the trunk (Leroux et al., 2002)). The device would, thus, be able to compensate for hip abductor weakness by providing improved frontal plane stability through upper extremity control and decreasing the need to control pelvic position, thus reducing the variability of step width relative to up-ramp walking. The possible contribution of the upper extremities to stabilization of the head, arms and trunk is a reflection of other factors that may contribute to the observed changes in step width variability that could not be ascertained given the current experimental set-up. Therefore, it would be beneficial, in future work, to determine if the changes in step width variability correlate with changes in other characteristics of the participants (e.g. muscle strength or performance on
clinical tests that assess such qualities as their balance or ability to engage in activities of daily living).

4.4.1.3 Effect of velocity on foot placement

Walking velocity is another factor that can influence step width variability. Faster walking speeds often result in increased step width variability, as demonstrated by Brach et al. (Brach, Berthold, Craik, VanSwearingen, & Newman, 2001) in community-dwelling older adults. However, in the current study, there was no significant difference in velocity between the in-lab, up-ramp, and down-ramp conditions, and therefore, velocity may not have contributed to the changes observed in step width variability. Nevertheless, changes in velocity were still observed, i.e. the participants walked significantly faster in the intersection-crossing environment, compared to both the in-lab and up-ramp conditions. This was expected since crossing the street had a temporal constraint (i.e. crossing must be completed before the traffic light changes to red). It is interesting that the participants achieved this faster velocity without making significant changes to their medio-lateral foot placement (e.g. by increasing step width variability). As a result, either the individuals with MS adopt a different control strategy to increase velocity that is independent of medio-lateral balance control or they are using the walker in a compensatory manner such that medio-lateral foot placement changes are not necessary.

4.4.2 Limitations

Step-to-step differences in the angle of the feet relative to the plane of progression may have contributed to some of the variability in step width observed in the current study. This is the case because the most feasible location to place the foot markers was over the toes. Alternatively, markers could have been placed over the navicular; however, it would have been more difficult to regulate the vertical height of the marker, which is necessary to use my foot detection technique. Nevertheless, the variability of my results was not greater than that observed in other studies of individuals with MS. (Givon et al., 2009; Gutierrez et al., 2005) In addition, movements of the walker that were not in the direction of forward progression (e.g. side-to-side swerving) may have also contributed to the variability of my results. Within the laboratory, participants were instructed to walk a straight walking path, and no side to side swerving or turning movements were evident. The out of lab testing included environmental contexts with a straight walking trajectory. However, the impact of swerving on the amplitude and variability of
step width could not be quantified in the walking course, given the current experimental apparatus, and thus, remained a potential confounding factor. As well, another potential confound may have been the use of secondary gait aids (e.g. ankle-foot orthoses) by some of the participants (n=4), since this factor may alter the mechanics of walking. It is also important to note that the number of samples used to estimate the variability of step width was different between conditions (in lab: n = 5; sidewalk: n=6; up-ramp: n=6; down-ramp: n=7; intersection: n=6). In recognition that this difference in samples is reflective of the natural differences in the length of walking paths under each condition, a standard error of the mean was chosen for the purpose of statistical analysis. This approach provides a conservative estimate of variability between environments. However, since the standard error of the mean is less commonly used for expressing the variability of foot placement parameters, it is difficult to compare my results to those of other studies. I have, therefore, included the standard deviation and coefficient of variations to facilitate the interpretation and comparison of this work to other research. Lastly, the iWalker-based technique itself may have presented some limitations that could be potentially overcome in future work. For instance, it may be possible to improve the colour filtering algorithm, so that the technique is not as sensitive to differences in lighting condition between environments. As well, exploring less conspicuous testing apparatus (e.g. marker-less tracking techniques) may help to improve the ecological validity of the findings. In spite of these potential limitations, the present study had sufficient statistical power to detect changes in the main outcome measures (i.e. 83% for step width and 100% for step width variability, with α set at 0.05).

4.4.3 Future Directions

In order to develop a more comprehensive understanding of the challenges to stability that individuals with MS face in the community, an enhanced understanding of walker-user interactions in relation to the environment must be established. As a result, future work will involve exploring the walker loading patterns collected on-board the iWalker to discern the involvement of the upper limb during rollator-assisted gait and the implications it has for foot placement variability. I also plan to develop a comprehensive theoretical framework for categorizing community-based gait challenges and stability-related phenomena in order to better understand the factors that influence safe mobility in different environments. This knowledge will be fundamental to the development of more effective fall prevention strategies and the
design of future assistive mobility devices so that individuals with MS will be able to ambulate safely and efficiently for as long as possible as they cope with their progressively debilitating disease.

4.5 Conclusion

This study provides a first examination of the medio-lateral foot placement patterns exhibited by individuals with MS as they ambulate through several environments in the community with a rollator. Through validating my foot placement detection technique in a cohort of individuals with MS, the technique can be used in this population to further enhance the understanding of their rollator-assisted gait patterns in the community, which is where an increased risk of falling has been demonstrated. This study reports that step width variability, but not step width, is different between certain environmental settings, such as between a laboratory and ramp ascent or between ramp ascent and ramp descent. Walking velocity was also found to be altered in certain settings, such as during intersection crossing compared to in-lab and up-ramp walking. The velocity changes observed were not accompanied by changes in medio-lateral foot placement.
5 Discussion

The purpose of the present work was to develop a better understanding of how an assistive mobility device affects the mobility of individuals living with multiple sclerosis. This goal was approached by addressing two questions. First of all, can a technique be developed that enables the objective measurement of foot placement parameters, using an instrumented rollator, and is feasible to implement in able-bodied young adults? Secondly, what is the relationship between foot placement characteristics and the environmental context when a rollator is used by individuals with MS? It was first hypothesized that the medio-lateral positioning of the feet could be accurately assessed during rollator use by means of a camera system mounted on-board the device. A novel foot placement detection technique was developed to extract step width and step width variability parameters from the video output of the camera system. Through applying this technique to a cohort of able-bodied young adults, my hypothesis was supported. This result laid the foundation for Experiment 2. In order to examine the medio-lateral foot placement characteristics of individuals with MS as they ambulated through different outdoor urban environments with a rollator, I applied the foot placement detection technique to the MS population. In this experiment, I hypothesized that step width and step width variability would be increased in the community-based settings relative to a laboratory context. I validated the foot placement detection technique in individuals with MS, at the onset of the experiment, to confirm the concurrent validity of the technique when it is implemented in this population. Subsequently, I observed the foot placement characteristics of the individuals with MS in four community-based settings and compared the results to those obtained in the laboratory. In doing so, I determined that the second hypothesis was only partially supported. The walking environment did not have a significant effect on the step widths exhibited by the participants in any of the conditions; however, significant differences in step width variability, as well as walking velocity, were detected. It was concluded that circumstances exist in which the medio-lateral foot placement patterns of individuals with MS are influenced by an individual's outdoor walking environment during the use of a rollator. This warrants further study in order to attain a more comprehensive understanding of the factors that impact safe mobility through everyday environments and how stability is concurrently facilitated by lower limb placement and assistive mobility device use. This knowledge will be essential for designing more effective assistive
mobility devices as well as for improving fall prevention programs for individuals with MS, with the hope of reducing the impact of the disease on their lives and enhancing their quality of life.

5.1 Analysis of the Foot Placement Detection Technique

I found that the iWalker-based foot placement detection technique and the Vicon MX motion capture system were in agreement, in both the able-bodied and MS participant groups. Agreement was assessed using three approaches. First, intra-class correlation coefficients (ICC) (two-way mixed effects model, absolute agreement) were used to examine the extent to which the two measuring systems were associated, as an indicator of concurrent validity. Second, the RMS difference between pairs of measurements was used to provide insight into the variability (or spread) of those differences. Third, a Bland-Altman analysis was performed to further examine agreement, by helping to determine whether bias occurred and if any relationships exist between the variance in measures with the size of the mean.

According to the results of the ICCs, the iWalker-based foot placement detection technique exhibited a stronger agreement with three-dimensional motion capture systems than other commonly used clinic- and laboratory-based gait assessment tools, when measuring medio-lateral foot placement parameters (e.g. a GAITRite instrumented walkway system). (Menz, Latt, Tiedemann, Kwan, & Lord, 2004) In addition, the RMS differences (i.e. 0.70 cm) obtained for both of the participant groups were below what would be considered meaningful changes in clinical settings. There are no suggested clinically-meaningful, within-subject changes in step width that have been reported in previous work in the MS population. However, it would be meaningful to detect significant changes in step width between individuals with MS and other populations. For example, a clinically-meaningful difference in the base of support of individuals with MS, relative to able-bodied individuals, would be 2.5 cm. (Givon et al., 2009) This difference is over three times the magnitude of the RMS error of the iWalker-based foot placement detection technique, thus, it would be able to distinguish between these populations. Furthermore, to determine the ability of the technique to identify differences within a population and within subjects, clinically-meaningful changes in other populations can be used as a basis of comparison. For example, older adults who experience falls to the side and those who fall in other directions exhibit a difference in step width of 2.4 cm. (Ko et al., 2007) As well, the step widths of older adults who experience a fear of falling are, on average, 1.9 cm wider than those
who are fearless. (Chamberlin, Fulwider, Sanders, & Medeiros, 2005) Such changes would be detectable using the technique, as they exceed the magnitude of the RMS error by over two to three times. The technique would also be capable of detecting step width differences within individuals, as observed in other populations under different task or environmental conditions. For instance, step width has been demonstrated to change significantly with differences in surface regularity and lighting (e.g. 1.9 cm in individuals with peripheral neuropathy and 1.0 cm in young and older adults) (Menant et al., 2009; Richardson, Thies, DeMott, & Ashton-Miller, 2004), surface compliance (e.g. 1.9 cm in young adults walking on foam) (MacLellan & Patla, 2006), the use of gait aids (e.g. 1.6 cm and 2.0 cm respectively for ankle-foot orthoses and canes in stroke patients) (Abe, Michimata, Sugawara, Sugaya, & Izumi, 2009; Kuan, Tsou & Su, 1999), the presence of haptic feedback (e.g. 2.84 cm with vertical touch in individuals with peripheral neuropathy), the type of walking environment (e.g. in adults, 2.31 cm on a treadmill and 1.4 cm in virtual reality simulations compared to overground) (Hollman, Brey, Robb, Bang, & Kaufman, 2006; Stolze et al., 1997), the extent of lateral external stabilization (e.g. 6.6 cm in older adults) (Dean, Alexander, & Kuo, 2007), and the likelihood of experiencing a tripping perturbation (e.g. 1.23 cm for young adults) (Pijnappels, Bobbert, & van Dieën, 2001). These changes all surpass the values of the RMS error, thus indicating that they would be detected by the technique. Nevertheless, the RMS error of this technique is still larger than that exhibited by three-dimensional motion analysis systems. It is therefore limited in ability to detect smaller meaningful differences than 0.71 cm, which would include changes in footwear (e.g. 0.03 cm for a soft shoe) and wetness of the walking surface (e.g. 0.6 cm in a wet compared to dry surface). (Menant et al., 2009)

Furthermore, to detect meaningful changes in step width variability in the MS population, the iWalker-based foot placement detection technique must be able to perceive changes that exceed the intrinsic variability of step width exhibited by this population. The baseline step width variability of the individuals with MS, observed in the lab condition of this study, was 0.4 cm. This value is slightly less than the variability obtained by Givon et al. (2009) of 0.6 cm for individuals with MS. Consequently, changes in step width variability that exceed this 0.4-0.6 cm range could be considered clinically-meaningful. Previous work in other populations has demonstrated that changes in step width variability of this magnitude can be observed under conditions, such as increased surface compliance (MacLellan & Patla, 2006) and altered surface
regularity and/or lighting (Richardson et al., 2004; Thies et al., 2005). Changes of this magnitude would be identifiable using the iWalker-based technique. On the contrary, this technique would not be suitable to detect certain other meaningful changes in variability, which are smaller. Such changes would include those that take place when external lateral stabilization is provided (Donelan et al., 2004) and when the eyes are open or closed during gait (Bauby & Kuo, 2000).

When comparing the difference between the means of the two measurement techniques, there appears to be a slight bias of 0.03 cm and 0.28 cm for the able-bodied and MS groups respectively. This indicates that the iWalker-based technique, on average, tends to overestimate step width to a small degree. As well, the range of agreement between measurements obtained for both groups (i.e. both at 2.81 cm, using the Bland and Altman limits of agreement) is considerably larger than the corresponding RMS error calculations (i.e. 0.70 cm and 0.71 cm). These wide ranges may be an outcome of the small sample sizes used in this study relative to the recommended minimum sample size of 50 participants for calculating the limits of agreement (Rankin & Stokes, 1998). Increasing the sample size would reduce the effect of outliers and expectedly improve the range of agreement by decreasing its upper and lower limits.

In addition, factors related to experimental or methodological errors may also be playing a role in the variability observed between the techniques. For instance, the distortion of the camera lens on the iWalker necessitated the use of an image rectification process to remove the resulting parallax error from the video data. Since this correction factor increased in magnitude towards the extremities of the screen, the effectiveness of this correction would impact the accuracy of marker localization, particularly as markers approached the periphery of the field of view of the iWalker camera. In addition, characteristics of the markers themselves may contribute to the differences observed between the two systems. Factors such as the sphericity, size, and reflective properties of the markers can affect how the two systems reconstruct the positions of the markers within the capture volume of the lab (for the Vicon system) and the area under the walker (for the iWalker-based technique). In spite of these shortcomings, the iWalker-based foot placement detection technique appears to be a useful tool in the assessment of medio-lateral foot placement as it is reliable and sufficiently sensitive to detect some clinically-relevant changes in parameters of interest.
5.2 Environmental Effect on Medio-lateral Foot Placement

In the present work, the foot placement detection technique was utilized to assess the medio-lateral foot placement characteristics of the individuals with MS during their use of the iWalker in four outdoor environments and the laboratory. The prediction was that under more challenging environmental conditions, individuals would adopt a wider step width during walking, as a reflection of the real or perceived challenges to stability. However, contrary to my hypothesis, there were no significant differences in step width observed between any of the environments. There are several reasons why this may have occurred. First, it is possible that step width is not influenced by the walking environment. Alternatively, it could be that step width is in fact affected by the environment, but that the use of a rollator served to help individuals compensate for real or perceived challenges to stability. In effect, this could nullify any changes in step width that may have been anticipated. The measurement of other aspects of walker-user interactions, such as the loading forces applied to the rollator frame, may provide further insight into how these assistive devices are being used to adapt to environmental challenges. On the other hand, if the environment does influence step width, it is also possible that the lack of change observed in this work could be attributed to the observed environments not being sufficiently challenging to elicit a change in this parameter. In this scenario, increasing the number of environments or the challenges within a specific situation may help to reveal an environmental effect on step width.

In contrast to step width, significant environment-related changes were observed in the step width variability of the individuals with MS. The first difference emerged between the up-ramp and laboratory environments, with ramp ascent eliciting an increase in step width variability. This difference may be related to the pathology of the individuals with MS. Muscle weakness and spasticity are prevalent sequelae of MS, often affecting the hip musculature. (Snow et al., 1990) Ramp ascent requires additional work to be performed in order to successfully raise one's centre-of-mass to the higher level. Throughout this biomechanical challenge, a large amount of the extra work being performed is generated by concentric contractions of the hip and knee extensors. (Lay, Hass, Nichols, & Gregor, 2007) This increased activity benefits the sagittal plane component of the task, increasing forward propulsion. At the same time, in order to achieve the increased toe clearance that is necessary to accommodate for the raised support surface, the hip abductors are also involved, as they are used to raise the pelvis
while the contra-lateral leg is in swing. Therefore, if hip abduction is weak in individuals with MS, it may challenge their ability to achieve the required toe clearance and step length. Since modifying the activity of the hip abductors and adductors is the primary mechanism by which to control step width (Shumway-Cook & Woollacott, 2007, p. 370), weakness in these muscles may also contribute to decreased control of the lateral positioning of the feet, thus resulting in the increased step width variability. In addition, it is also necessary to take into account how the rollator may affect task performance. During level-ground walking, rollators can be used to help compensate for reductions in hip abductor and adductor strength, by helping the user to maintain his or her centre-of-mass within the lateral base-of-support through the use of the upper limbs. During ramp ascent, however, the forward inclination of the trunk tends to increase. (Leroux et al., 2002) This postural adjustment shifts the centre-of-mass anteriorly, thus assisting the lower limbs by generating additional forward momentum to compensate for the resistance due to gravity during ascent. Through subjective observations of the participants as they walked up the ramp, it appeared that this postural change may have persisted during rollator use, resulting in a modification of their interaction with the device by pushing it farther in front of them than during level-ground walking. In doing so, their capacity to exert vertical loads on the walker would have been reduced, thus diminishing the extent to which the device could have been used to compensate for muscular weakness. As a result, hip abductor strength could be more important to successful task performance and the frontal plane consequence of this could be an increase in step width variability.

The second environment-related change in step width variability, exhibited by the individuals with MS, was a significant decrease in variability during ramp descent compared to ramp ascent. Contrary to uphill walking, walking down a ramp is characterized by power absorption. (Lay et al., 2007) In order to remain in control of the lowering of the centre-of-mass, eccentric muscle control is used to counteract the conversion of gravitational potential energy into kinetic energy, thus resisting the forward momentum of the body that would otherwise increase walking speed. Muscles of the lower limb joints, particularly the knee extensors and ankle plantarflexors, are involved with this absorption of energy. (Lay et al., 2007) Throughout downhill walking, the absolute angular orientations of the trunk and pelvis are usually tilted backwards (Leroux et al., 2002), thus moving the centre-of-mass posteriorly to further reduce forward momentum. This postural adjustment seemed to have been adopted by the individuals
with MS and incorporated into a strategy involving the rollator. Subjective observations of the participants during ramp descent revealed that they were holding the device closer to their bodies. In doing so, the rollator could have been loaded more, thus providing additional stability by increasing the user’s base of support, while ensuring that the centre-of-mass did not move as far ahead of the base of support as during level-ground walking. In conjunction, the participants also appeared to be engaging the brakes on the rollator’s wheels through the walker handles during ramp descent, as observed subjectively. The brakes could be used to generate friction with the support surface, which would have helped to further absorb kinetic energy. The device could, therefore, be used to compensate for hip abductor or adductor weakness, reducing the importance of the strength of these muscles and resulting in a lower step width variability compared to uphill walking.

Gait speed, as reflected by walker movement speed, was quantified to determine if it may have been a factor that related to the medio-lateral placement of the feet. Environment-related differences in speed were revealed between two pairs of environmental settings. The first significant difference was observed between the laboratory and intersection conditions, and the second, between the ramp ascent and intersection conditions. In both of these cases, speed was elevated during the intersection crossing. The relationship between walking speed and variability has not been explored in the MS population. However, work in other populations, such as older adults, has revealed that, in individuals who were capable of modifying their walking speeds by more than 0.1 m/s, faster speeds were associated with significant increases in step width variability. (Brach et al., 2001) In the present work, the individuals with multiple sclerosis were able to achieve faster walking speeds during intersection crossing without having to adapt by increasing their medio-lateral step-to-step variability in foot placement. Therefore, it is possible that they were interacting with the rollator in a way that compensated for the factors typically responsible for elevating step width variability at faster speeds in the absence of assistive support. Exploring additional behavioural measures of the participants, such as their device loading patterns, could be used to complement the observations of their foot placement patterns and may provide additional insight into the results. Nevertheless, walking speed on its own is a good indicator of walking ability because individuals who are able to fluctuate their speeds have a better ability to adapt to their environment and to do so safely (e.g. enabling them to cross an intersection in an appropriate amount of time). In fact, gait speed has been demonstrated to be a
good predictor of future hospitalization risk as well as deterioration in health and function, with slower walking speeds associated with worse outcomes. (Studenski et al., 2003) Therefore, since the individuals with MS demonstrated that they were able to significantly change their walking speeds, this parameter might be a good outcome measure with which to assess community-based mobility in the MS population in the future.

Lastly, there are additional factors to acknowledge that may have influenced the results obtained during the course walk. In particular, three potential threats to internal validity may have had an effect: order effects, instrument decay, and fatigue. Order effects may have been present, since all of the environmental conditions were encountered in the same order. It was not possible to randomize the sequence of these environments, because the participants had to use ramps to enter and exit the building in which testing was conducted. As well, a possible instrument decay factor could have been the potential shifting of the camera, as a result of walker movement and vibrations occurring throughout the walking course. This aspect was controlled for by ensuring that the camera was: 1) firmly mounted to the walker frame; and 2) aligned to the calibration jig before and after each data collection. In addition, fatigue may have been a potential confounding factor. Its effects were minimized by enforcing regular breaks along the course and confirming that the participants’ heart rates had returned to their individual baseline levels before proceeding further. However, fatigue may have still played a role in two of the participants not being able to complete the walking course. In spite of these limitations, the environment-related changes in medio-lateral foot placement obtained in the present work can be generalized to individuals with MS who are also regular rollator users and the ecological validity of these results can be further improved upon be developing less conspicuous testing apparatus (e.g. marker-less tracking).

5.3 Contributions of the Present Work

Relative to other neurological patient populations, such as stroke and Parkinson's disease, the understanding of gait in individuals with MS is limited, with the emphasis of research on this disease focused instead on other areas, such as understanding its etiology or the impact of clinical treatments (e.g. strength training and disease-modifying drugs). Nevertheless, some studies were conducted that have investigated the gait patterns of individuals with MS in both laboratory settings (Benedetti et al., 1999; Givon, Zeilig, & Achiron, 2009; Kelleher et al., 2010) as well as clinical environments (Cattaneo, Regola, & Meotti, 2006; Goldman, Marrie, & Cohen,
These studies were instrumental in characterizing the gait characteristics of individuals with MS and understanding the differences that exist between these patterns and those of able-bodied individuals. They also demonstrated that reliable techniques can be used to assess differences in ambulatory ability between individuals, who experience different levels of disease severity, as well as within individuals, as the disease progresses over time. (Kelleher et al., 2010; Khaleeli et al., 2008) However, these studies only measured the gait patterns of individuals with MS under ideal conditions, for example, in well-lit environments with smooth surfaces and no overt distractions. As a result, they did not explore the effect of common outdoor environments on the gait patterns of individuals with MS, in spite of the fact that environmental factors have been determined to significantly impact stability during gait. Furthermore, many of these studies excluded assistive mobility device users who could not ambulate without their devices for a specified distance. (Givon et al., 2009; Souza et al., 2010) In some studies that permitted assistive device use, the results of device-users were not distinguishable from those of independent ambulators nor were the devices used by participants always specified. (Goldman et al., 2008) The use of an assistive device was, counter-intuitively, revealed to be a factor associated with the high incidence of falls exhibited by individuals with MS. (Cattaneo et al., 2002) Consequently, there was a need to explore how assistive devices were used in this population and to examine their effect on gait in different environments.

In light of this, the present work provides new insight into the mobility of individuals with MS. It is novel since it is the first study to examine the environmental effect of several commonly-encountered urban community settings on the gait characteristics of individuals with MS, all of whom were using a rollator. Whereas previous work on community-based mobility in MS has been mostly based on self-reported data, the current results offer a first objective glimpse into the everyday mobility of individuals with MS requiring assistive support. Given that the individuals with MS assessed in this study were primarily rollator users, who used these devices regularly in the course of their everyday lives, it is possible that this sample of individuals with MS may have comprised a sub-group of the MS population at large. Compared to previous studies, the sample used had a higher proportion of assistive device users (e.g. 100% compared to the 75% observed in Nilsagård et al. (2009)) and more impaired gait performance (e.g. a
slower walking velocity). As a result, the findings could only be generalized to this sub-group of individuals with MS, i.e. individuals dependent on rollators during ambulation.

The present work also contributes to the advancement of knowledge that may benefit future rehabilitation strategies, such as the development of more effective fall prevention programs. The findings can be understood in terms of several components of the ICF model and, perhaps more importantly, enables us to explore new areas of this model. While previous studies on the etiology of MS have expanded the understanding of the body structures and functions component of the ICF, the present work explored how rollators intervene with functional mobility deficits to influence factors in the activities and participation components. The Badley approach to the ICF improves the conceptual clarity of these components (i.e. activities and participation), by further sub-dividing them into the categories of acts, tasks, and participation. (Badley, 2008) This approach is suitable and effective in the context of the present work. Acts, in general, describe the things a person can do irrespective of context or purpose. Thus, similar to previous gait studies, the laboratory-based component of this work provides an indication of how gait is performed using a rollator without the distractions of a community-based situation. The tasks category, on the other hand, is concerned with how individuals perform acts within a specific context of daily living. Given the high incidence of falls experienced by rollator users in the community, context is a vital factor to consider. Currently, there is only a limited knowledge of rollator-assisted gait in community-based environments. Therefore, new areas of the ICF model were examined using the course walk component of this work. The monitoring of factors related to mobility during the course walk has also laid the foundation for future work that can investigate the barriers to participation in daily tasks. These societal barriers may arise out of features of the walker user's surroundings that contribute to the handicap creation process for individuals with MS and other assistive device users. (Fougeyrollas, 1995) In fact, the rollator, itself, can be perceived of as an environmental factor that influences each component of the ICF model (i.e. body functions and structures, acts, tasks, and participation). Therefore, through developing a better understanding of the impact that the environment has on foot placement patterns during rollator use, this work fosters the growth of knowledge in the field of rehabilitation.

In addition, the present work further advances approaches to ambulatory monitoring. Ambulatory monitoring devices detect the physiological function or behaviour of patients in a
non-laboratory environment, particularly as they live freely in their homes. (Mathie, Coster, Lovell, & Celler, 2004; Uiterwaal, Glerum, Busser, & van Lummel, 1998) These devices have been used to monitor a wide-range of human bio-signals, including cardio-pulmonary and metabolic functions. (Alwan, 2009; Battelino & Phillip, 2010) With recent technological advances, the assessment of mobility has also become very common. While a variety of wearable devices have been developed (Bamberg, Lastayo, Dibble, Musselman, & Raghavendra, 2006; Brehm, Knol, & Harlaar, 2008; Favre, Aissaoui, Jolles, de Guise, & Aminian, 2009; Grant, Dall, Mitchell, & Granat, 2008; Greene et al., 2010; Kishimoto et al., 2009), accelerometry is one of the most common techniques used to provide feedback on an individual's ambulatory characteristics. (Dijkstra, Kamsma, & Zijlstra, 2010). However, the acquisition of spatial data on gait characteristics has, in general, been a challenge for ambulatory monitoring devices. Although previous studies have explored quantifying spatiotemporal parameters of gait (Ramstrand & Nilsson, 2007), including step length (Alvarez, Gonzalez, Lopez, & Alvarez, 2006), techniques capable of detecting medio-lateral foot placement characteristics have not been developed before the present work with the iWalker. In fact, this is the first study to directly analyze the positioning of the feet during rollator use outside of a laboratory environment. To accomplish this, the technique developed in this study tracks the position of the forefoot during walking with a rollator fitted with a digital video camera. This technique has, therefore, expanded the possibilities available for the community-based monitoring of behaviour in users of this type of mobility device. Although this contribution is novel, the iWalker on the whole is one of several rollators that have been fitted with ambulatory monitoring technology to-date. The monitoring instruments incorporated into these rollators include: force/torque sensors (for loading forces applied to the walker as well as stride length and frequency); localization cameras, infrared sensors, and sonar arrays (for guidance and obstacle detection); odometers and optical encoders (for speed); accelerometers (for tri-axial accelerations of the device); and ECG-based pulse monitors (for short- and long-term monitoring of health trends). (Spenko, Yu, & Dubowsky, 2006; Tung, 2010; Wasson, Gunderson, Graves, & Felder, 2001) A major focus of earlier rollator-based studies has been on extracting information from hand-device interactions in order to obtain an understanding of the user's stability during gait. (Spenko et al., 2006; Tung, 2010; Wasson et al., 2001) In fact, some of these devices used feedback from the sensors in the walker's handles to provide information on spatio-temporal parameters of gait involving the feet, including stride time, step and stride length, single- and double-support time, and the timing of
heel strike and toe off events. (Alwan, Ledoux, Wasson, Sheth, & Huang, 2007; Spenko et al., 2006). These studies were conducted in a number of settings, ranging from in the laboratory for validation (Alwan et al., 2007) to eldercare facilities (Spenko et al., 2006), hospitals, and outdoor urban environments (Tung, 2010). Some of these previous studies have also explored the implementation of behaviour modifying interventions through a rollator, which would, for example, help users avoid obstacles in their travel path. (Rentschler, Cooper, Blasch, & Boninger, 2003; Rentschler, Simpson, Cooper, & Boninger, 2008; Spenko et al., 2006; Wasson et al., 2001) However, the success of these attempts has been limited to-date. The future of ambulatory monitoring with the use of these assistive mobility devices will likely involve a combination of behaviour-sensing and -modifying technologies, provided that some of the current limitations of this approach to movement analysis can be effectively overcome.

5.4 Future Directions

In the present work, it was determined that medio-lateral foot placement characteristics can be objectively assessed using a rollator that has been instrumented with a digital video camera system and that this ambulatory monitoring device could be used to perceive changes in the step width variability of individuals with MS during device use in certain walking environments. Currently, my focus has been on understanding how lower limb behaviour is modified by the walking environment during rollator use. In order to achieve a more comprehensive understanding of the walker-user interactions taking place, other behavioural characteristics of device use must also be concurrently taken into account. The role of the upper limbs during assisted gait would be particularly informative. However, there may be other gait parameters that have not yet been explored that would also prove useful. For example, changes in step timing between these environments may provide important insight into the challenges to stability that these individuals face in their communities, thus supplementing the foot placement information. It may also be possible to develop more effective, walker-specific measures of foot placement, which could be used to better understand walker-user interactions (e.g. quantifying the foot-walker collision risk). Once all of these parameters have been characterized in multiple environments, a beneficial next step would be to determine what constitutes a clinically-meaningful change in individuals with MS and to understand the relationship of these parameters to falls risk in this population. This could be accomplished by conducting a prospective cohort study that quantifies these parameters as predictor variables at baseline and collects data on
subsequent outcomes, such as the incidence of experiencing falls over the course of a year. Once it has been determined which changes in gait parameters are linked to falls in this population, this knowledge could be used in the long-run to develop and implement an intelligent assistive mobility device that, not only monitors a user's behaviour, but also promotes safe community mobility by providing real-time biofeedback about how it is being used. An advanced body-device interface could provide tactile, visual, and/or auditory cues to the device users when their device utilization patterns indicate that they are approaching the limits of their stability, thus proactively avoiding falls and reinforcing safe device use practices. Going forward, there are also several stakeholders who should be engaged at each stage of the research process, including patients, rehabilitation-focused clinicians, physicians, and inter-disciplinary researchers. This collaboration will be beneficial to ensure that further advances in this line of research are clinically-relevant and that potential commercialization efforts result in the development of an end product that is useful for members of many of these stakeholder groups. In doing so, future work may help to determine the causes underlying the high incidence of falls experienced by individuals with MS, particularly in everyday circumstances in the community, thus, making their ambulation safer and more effective.

5.5 Conclusions

This work demonstrated that the iWalker-based foot placement detection technique, which was developed, is capable of accurately quantifying the medio-lateral foot placement patterns of a rollator user, since its measurements are in agreement with a Vicon motion capture system. This technique was first validated in a cohort of able-bodied young adults. It was subsequently validated in individuals with MS and used to determine that medio-lateral foot placement patterns outside of a laboratory can be different than inside a laboratory in certain situations. In particular, step width variability, but not step width, has been shown to be affected by the walking environment. This has implications for the stability of individuals living with this disease who use assistive mobility devices in the course of their daily lives in the community. With further development in this area of research, future assistive mobility devices may better reflect the assistive needs of individuals with MS, in circumstances of their everyday lives, thereby reducing their risk of experiencing falls, while augmenting their mobility, and ultimately, enhancing their quality of life.
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