Development and Evaluation of the iWalker: An Instrumented Rolling Walker to Assess Balance and Mobility in Everyday Activities

by

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Abstract

The rollator is a mobility aid commonly used to facilitate balance and mobility for individuals with cardiorespiratory, musculoskeletal, or neurological deficits. Despite its popularity, there are also reports of adverse effects related to walker use linked to increased fall risks. Studies examining the effectiveness and consequences of rollator use have employed standard laboratory-based measurement methods that rely on performing specific tasks within a short time period and under controlled conditions, potentially limiting generalization to mobility in the everyday context. An instrumented rolling walker (iWalker) was developed as an ambulatory measurement tool applicable to the assessment of balance outside of the lab or clinic for assistive device users. The iWalker autonomously collects measurements of the upper and lower limb behaviour related to balance, walker kinematics, and video of the immediate spatial environment.

The design and development of the iWalker is first described, followed by two studies characterizing the involvement of the upper limbs for balance in standing and walking that served to address gaps in the literature and evaluate the utility of the upper limb measures. Overall, the upper limbs can become the primary effectors of balancing forces when lower limb capabilities are compromised. When lower limb involvement was experimentally constrained, the upper limbs became the primary effectors of balance control in healthy, young adults. In older adults, individuals demonstrating the highest upper limb usage during walking were associated with the largest reduction in frontal plane stepping parameters (i.e., step width). A third study evaluated the applicability of the iWalker to assess everyday mobility in a series of in-patients recovering from neurological injury (i.e., stroke, traumatic brain injury). Patients demonstrated significantly different upper limb balancing behaviour in everyday situations compared to in-laboratory assessments. Furthermore, the iWalker captured behaviours that may be precursors to falling, such as collisions, stumbling and lifting the assistive device. The implications of these studies on assessing the effectiveness of rollators and feasibility of using the iWalker in follow-up efforts are discussed.

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Chapter 1
Introduction

1 Problem

Occurring at high rates and often resulting in serious injury, falling is a widespread clinical problem faced by older adults [114]. Injuries, such as hip fracture or head trauma, can have long-lasting effects on an individual's motor capabilities and activity, which can spiral into further functional decline, deconditioning and increase the risk for subsequent falls [66]. As the most common reason for admission to an assisted living facility, falls also contribute to loss of independence for activities of daily living [131]. Furthermore, in light of a growing older adult demographic, the total economic burden associated with treating fall-related injuries, institutional care, and downward spiral in health status will likely increase [99]. Hence, examining the causes of falling and developing prevention strategies has been a major goal in research and health policy. However, despite substantial insight into the neurophysiological and biomechanical characteristics of the body that influence the control of posture and dynamic stability, the effectiveness of interventions to affect rates of falling have remained limited [44].

Falling, or failing to maintain upright posture and recover balance, is influenced by factors intrinsic to the individual and (extrinsic) characteristics of the surrounding environment. The intrinsic sensory, muscular, and integrative systems of the body essential to the control of balance and recovery has been the major focus of research of the causes of falling and forms the basis of the majority of intervention strategies. While clinical trials of multifactorial interventions incorporating tailored combinations of training and behaviour modification initially suggested promise [47], their effectiveness in reducing fall rates when implemented into clinical practice have been limited [44]. The inability of current interventions to deliver substantial reductions in falls suggests that our current models of balance control and approaches for fall prevention need to be re-examined.

While the factors intrinsic to the body have been the focus for study and intervention, extrinsic factors clearly place specific demands on balance control. Extrinsic factors that have been linked to falls are primarily characterized by the physical context that a person interacts with in everyday activities, such as their home, work, leisure and community environments [121, 46].

Moreover, the everyday environment includes complex and dynamic features that can present unpredictable challenges to balance that may be encountered in novel combinations. Maintaining balance in this context demands successful adaptation of intrinsic functional capabilities defined by the constraints of the task and environment [103]. For example, balance recovery reactions must resolve the availability of support surfaces and hazardous obstacles in the immediate spatial surroundings to prevent a fall [143]. We need to advance our established intrinsic-based models of postural control and recovery to include interactions with environmental context.

There exist at least two main options for study: 1) explore systematically the influence of specific extrinsic factors on postural and locomotor control in laboratory settings; and/or 2) explore the multi-dimensional influence of an array of extrinsic factors during everyday testing protocols. The latter reflects the opportunity to measure behaviour of individuals in their natural environments in which the extrinsic factors are not experimentally delivered or constrained. While this proposition poses an important challenge to the measurement of behaviour in a natural setting, it does provide the important advantage of informing about the possible interactions between extrinsic and intrinsic factors that may not have been previously hypothesized. Recent advances in ambulatory monitoring tools and techniques have created the opportunity to record and measure behaviour in the everyday environment. These advances in sensing and acquisition technology allow measurements to be taken outside the lab, over long durations and with greater sensitivity and reproducibility than self-report measures. Importantly, the potential of ambulatory measurement approaches to reconstruct the temporal sequence of events permits the association of behaviours and extrinsic factors towards establishing specific relationships.

As an example application of an ambulatory measurement approach, the issue of mobility aids to assist in maintaining balance and preventing falls is examined. Mobility aids (i.e., canes, crutches and walkers) have received relatively little attention compared to exercise and training interventions as fall prevention strategies. In particular, the limited evidence examining the effect of using walkers to prevent falls is controversial. Hypothetically, walking frames permit the inclusion of the upper limbs in sensing instability and generating stabilizing forces to compensate for impaired lower limb abilities [13]. However, only a small number of studies have provided limited evidence to support this hypothesis [12, 13, 74]. Furthermore, epidemiological and prospective studies have found that walker users demonstrated greater rates of falling [82, 92, 78, 22, 127] and potential mechanisms underlying the increased risk have been proposed and assessed [12, 14, 41]. Despite the lack of evidence to support their benefit and indications of significant adverse effects, walkers continue to be prescribed to address balance impairments and help maintain independent mobility [66, 50].

The long term objective of this research is to assess the potential effectiveness and challenges associated with using rollators (or 4-wheeled walkers) to maintain balance in everyday conditions. In this thesis, we hypothesize that an ambulatory measurement approach will enable a deeper understanding of the potential balance benefits and issues related to rollator use. Specifically, three thesis objectives are identified: 1) Develop an autonomous measurement tool to measure balance control and characteristics of the environmental context in everyday situations; 2) Examine the characteristics of upper limb involvement in maintaining stability and factors influencing their control; and 3) Evaluate the utility of using an ambulatory measurement approach to directly observe the stabilizing benefits and potentially adverse consequences associated with rollator use in everyday life situations.

2 Overview of thesis

In addition to the current introductory chapter, this thesis comprises of 6 chapters:

- Chapter 2 is a background chapter that provides key definitions and a review of the existing literature. The objective of this chapter was to detail the relevant studies examining balance control and falling, fall prevention strategies, ambulatory monitoring techniques, and the effects of using walkers for balance and mobility.
- Chapter 3 details the design and development of a novel ambulatory monitoring tool, the iWalker. Entitled, ‘iWalker: Development of an Instrumented Rolling Walker to Assess Balance in Everyday Activities’, the main objective of this chapter was to describe the rationale for selecting the key sensors and components of the system. Validation data to support the implementation of the assembled system, and clinical examples to demonstrate the potential utility of the iWalker as a research tool are presented.
- Chapter 4 entitled, ’Bilateral Integration of Upper Limbs in Standing Balance Control with an Assistive Device’, describes a controlled laboratory study investigating the role of upper limbs in maintaining balance with rollator use during quiet and perturbed standing in healthy young adults. The objectives of this study were to: 1) characterize the upper limb

behaviour associated with maintaining balance while standing; and 2) develop a basis of upper limb measurements collected by the iWalker to help interpret subsequent studies.

- Chapter 5 extends the work conducted in the previous standing study to walking. Entitled, 'Frontal Plane Balance Control in Assisted Walking: Beam-Walking Controls and Chronic Older Adult Rollator Users', this chapter characterized upper limb involvement during straight-line walking in healthy adults. The secondary objective of this study was to examine the frontal plane upper limb behaviour in a population of older adults with balance impairments who are regular users of rollators.
- Chapter 6 entitled, 'Everyday Balance Related to Rollator-Assisted Mobility: Case Studies in Neurologic Rehabilitation In-Patients', presents a case series of comparing balance behaviour indicated by iWalker measurements under laboratory conditions to everyday life situations. The objectives of this study were to: 1) demonstrate the interactions between the everyday environment and balance behaviour related to rollator use; and, 2) evaluate the balance benefits and adverse consequences of using a rollator under conditions more typical of everyday life.
- Chapter 7 is the concluding chapter of the thesis which summarizes the key lessons and findings from the described studies and discusses their impact on clinical assessment, assistive device design, safety policies, and future research.

Chapter 2
Background

In 2006, nearly 3 million Canadian adults had difficulty with mobility tasks, representing more than 7 out of 10 persons with disabilities [101]. The capacity for mobility, defined as the ability to move from one place to another, influences the maintenance of physical, metabolic and cognitive functioning [123]. When mobility is compromised by musculoskeletal, neural and/or cardiorespiratory conditions, there is a 3-5 times increased likelihood for requiring some dependency when performing activities of daily living [55]. Mobility impairments can also restrict the capacity for social interactions [31], which an important factor associated with maintaining cognitive function [1]. Furthermore, mobility problems limit the amount of physical activity performed, such as leisure-time walking [8], impacting the ability to achieve health benefits associated with regular physical activity such as decreased risk for heart disease, stroke, diabetes, and osteoporosis [63]. Hence, maintaining or improving the ability to walk independently is an important goal in maintaining health-related quality of life.

1 Balance control and falling

Our current knowledge on mobility stems from a history of research on walking and motor control, based on laboratory research equipped with sensitive measurement instruments, such as motion analysis (e.g., [138, 24]), forceplates (e.g., [26, 141]), and electromyography (e.g., [53, 39]). This body of work has defined the fundamental determinants of walking activity: progression, balance, efficiency and shock-absorption [123]. Progression is the production of rhythmic patterns of muscle activation of the legs and trunk to move the body in the desired direction, including the initiation and termination of walking. Balance control is needed to maintain an appropriate posture, counteracting against gravity and other destabilizing forces. These goals must be met through strategies that are energy-efficient and minimize stresses to maintain body structures and tissues over the lifespan of the person. While all of the aforementioned capabilities are required for independent mobility, this thesis is focused on maintaining balance.

The importance of balance control is evidenced by the serious consequences associated with failing to maintain upright stability. Falls and unstable balance rank high among the clinical problems faced by older adults [115]. In Canada, seniors are 9 times more likely to experience a fall-related injury than those less than 65 years of age and nearly 62% of injury-related hospitalizations for seniors are the result of falls [99]. Loss of independence is often a consequence of falls, accounting for 40% of all nursing home admissions [131]. A fall can also cause loss of self-confidence and self-imposed reduction in activity levels, which can lead to further decline in functioning and contribute to future falls [30].

The cause of falls is varied, generally resulting from an interaction of multiple and diverse risk factors and situations [115, 81, 66], including both extrinsic (e.g. prescription medications, poor lighting, loose carpets) and intrinsic (e.g., lower extremity weakness, visual deficits) factors. From a review of the literature examining the risk factors for falls, the major factors have been ranked according to the relative risk of falls [66] and are listed in Table 1. Importantly, the interaction between multiple factors has a dramatic effect on risk of falls. For example, Tinetti et al. found that the risk of falling in community-dwelling older adults increased from 27% for those with no or one risk factor to 78% for those with 4 or more risk factors [131].

2 Fall prevention strategies

Evaluating intervention programs designed to directly target risk factors has been the subject of a number of reviews and meta-analyses [66, 47, 21], indicating mixed success in affecting fall outcomes. The majority of interventions consisting of training approaches, such as gait, balance, strength, flexibility, or endurance exercise, have not demonstrated significant effectiveness in reducing fall rates [47, 21]. Behavioural modifications, such as education, medication modification, vision correction, and nutritional supplements have also demonstrated little to no impact on reducing the rates of falling [66, 47, 21], nor have environmental modification interventions, primarily removal of potential tripping or slipping hazards in the home, been proven to be effective [75].

Combinations of interventions designed to target the interaction between multiple risk factors that precipitate falls have demonstrated more significant benefits [66, 47, 21]. Multiple component training interventions, comprising of a combination of 2 or more of the aforementioned training and/or behavioural modification modalities have demonstrated significant reductions in fall risk [47, 21]. Multifactorial assessment and management programs have shown the greatest effectiveness [66, 47, 21]. In these programs, individual risk factors were identified on assessment and a tailored intervention program (e.g., balance training, drug review and home modification) was designed to specifically address the individual risk factors that were identified on assessment. However, randomized control trials examining the effectiveness of these approaches into clinical practice have failed to deliver significant effects on fall outcomes (for review see [44]).

The inability of current interventions to deliver substantial reductions in falls suggests that our current body-centric models of balance control and approaches for fall prevention need to be re-examined. The majority of intervention strategies have emphasized the identification and modification of intrinsic factors. For example, 10 of the 11 major risk factors associated with falls [66] (Table 1) describe intrinsic body capabilities. However, the translation of how these intrinsic factors are influenced by the extrinsic factors, to ultimately influence maintaining stability in everyday life, is poorly understood. Recently, researchers have suggested that more comprehensive models of postural control and mobility should incorporate the effect of extrinsic factors [103, 104].

3 Balance in everyday environments

Extrinsic factors are primarily characterized by the physical context that a person interacts with in everyday activities, such as their home, work, leisure and community environments [121, 46]. The environmental factors linked to falling have largely been drawn from self-report data (e.g., [99, 130, 18, 22]), placing a strong reliance on recall of events preceding falls. A major limitation of work to date has been that falls are often recorded and documented through self-report and lack essential environmental details, such as obstacle height, illumination or coefficient of friction. More specifically, there is a general consensus that falls likely result from the interactions between limitations in intrinsic capabilities and the demands arising from the constraints of the environment [121, 115, 81, 66, 75]. Not only is identification of the relevant environmental details essential, but the specific combination and the timing in which they are encountered are key factors determining behavioural strategies.

Laboratory-based studies have provided further insight the into balance strategies in responses to specific environmental conditions. For example, the proactive and reactive strategies in adaptive walking and dynamic stability control have been examined in tasks such obstacle avoidance [106, 138, 45], stair ascent/descent (for review see [126]), and changes to support surfaces [143, 20, 85]. While the modern gait and balance laboratory is adept at probing the biomechanics of walking and neuromotor control mechanisms with exceptional depth, it has been argued that the laboratory cannot duplicate the complex and unpredictable challenges that the everyday environment presents [103, 36, 28]. There continue to be improvements in the simulation of environments [42, 56, 97] and such simulations may be useful in probing specific extrinsic determinants of walking. However, the detailed combination and time-course of environmental factors encountered, along with the resulting balance behaviour in everyday scenarios has yet to be explored.

4 Ambulatory measurement approach

An alternative approach to a systematic exploration of the influence of specific extrinsic factors on postural and locomotor control in laboratory settings is to measure and record the behaviour of individuals in everyday life activities. Advances in ambulatory sensing and acquisition technology are creating opportunities to measure and record outside the lab and in everyday environments, over extended periods, and with greater sensitivity and reproducibility than clinical and self-report measures [118, 64, 43]. For example, cardiac specialists have used non-invasive ambulatory systems to track electrocardiogram (ECG) signals (i.e., the Holter monitor) to assist in diagnosis and assessment of cardiovascular patients [144]. Ambulatory EEG recorders are used to record and identify epileptic spikes that are difficult to capture due to their infrequency [117]. A particular advantage of ambulatory devices is the continuous recording for days to weeks coupled with their small physical size, allowing the individual to wear the device with little restrictions to activity.

Inertial sensors, such as accelerometers and/or gyroscopes, have been used to measure kinematics of the limbs and/or trunk related to balance (e.g., postural sway [29, 86]) and walking stability (e.g., temporal stepping parameters [7]). Center-of-pressure (COP) shifts and temporal gait parameters can be estimated by embedding footswitches [52], pressure-sensitive insoles [110], or multi-axis force transducers [73] into the shoes. However, sampling both ambulatory behaviour and the features and conditions of the immediate environment is needed to directly associate balance behaviour in response to extrinsic factors. While the potential for ambulatory measurement methods to capture balance behaviour exist, methods to capture relevant details of the corresponding environmental context are lacking.

Environmental features of the immediate spatial surroundings, such as terrain conditions and obstacles, have been previously linked to falls using self-report or manually recorded data. Studies linking home hazards to falls in the elderly have used a ‘checklist’ approach by recording the presence or absence of tripping or slipping hazards in the home [75, 46]. These methods lack the depth of measurements needed to directly associate an interaction with the environmental feature, such as combinations and time in which features are encountered. Developing feasible techniques to record the spatial surroundings in an ambulatory fashion is a methodological challenge yet to be addressed.

Video is a promising sensing modality to capture useful details regarding the nature of objects in the spatial surroundings (e.g., size, shape, or movement) with the temporal precision required to reconstruct a sequence of events. From a practical perspective, video captured by hardware mounted to an individual is problematic due to movement of the camera from body motion. A stable reference is preferred to analyze and interpret video images. An alternative is to instrument the environment with cameras to record an individual’s interaction with their spatial surroundings. For example, computer vision has been used as a sensing agent to track hand motions in an instrumented bathroom [89]. Unfortunately, instrumenting the environment limits the availability of data to the areas or targets that are set up with sensors, which excludes novel environments.

The selection of the initial application for an ambulatory measurement approach considered both the methodological challenges involved in developing an ambulatory measurement system and the potential impact on preventing falls. As an intermediate approach to instrumenting individuals with cameras or mounting them in the environment, a stable frame of reference that moves with the individual can be used. A specific mobility aid, the rollator (or 4-wheeled walker) can provide such a reference. From a list of the most significant risk factors for falls [66] (Table 2,1), the lone extrinsic factor identified as a high risk factor for falls was the use of mobility aids. Despite being identified as a risk factor, the utility of mobility aids, such as canes, walkers and crutches, on preventing falls have received relatively little attention. Examining the effects of mobility aids on maintaining balance using an ambulatory measurement approach is expected to uncover novel behaviour to expand our knowledge of balance control with such devices and better expose potentially adverse effects.

5 Mobility aids

Mobility aids, such as canes, walkers and crutches, play an integral role in addressing mobility disability [13]. In the United States alone, and estimated 4.8 million people (1.55 % of the total population) used canes, 1.8 million (0.58%) used walkers, and 0.6 million (0.20%) used crutches [65]. These aids are used to address pain, endurance, weakness, and/or balance impairments in the lower limbs to facilitate independent movement and maintain stability [13]. Generally, mobility aids are considered to facilitate mobility by permitting the upper limbs to be involved in bearing weight, effecting stabilizing forces and providing additional haptic feedback of body orientation [13, 3, 25, 32, 62].

Despite the recommended use of mobility aids to address gait and balance disorders for fall prevention [66, 115], the evidence examining their effectiveness in reducing falls is unclear. Mobility aid use has been linked to an increased fall risk [82, 92, 78], ranking ahead of vision impairment and arthritis as risk factors [66]. Mobility aid use can be interpreted as an indicator of balance impairment or functional decline [78, 79, 127]. Alternatively, mobility aids may introduce an increased risk of adverse consequences that may lead to falls. For example, unexpected perturbations to balance may result from inadvertent contact with an object in the environment (e.g., doorframes) [13] or by causing tripping [12]. Slipping or tipping may result from excessive horizontal forces coupled with insufficient downward force applied to a walking frame [41, 102]. Furthermore, mobility aid use has been linked to increased attentional and neuromotor demand which may interfere with upper limb reactions to recover balance, such as grasping a handrail or other support surfaces [13, 139]. In light of the controversial literature regarding the potential impact of mobility aids, the need to better understand the balance benefits and potential risk of adverse consequences associated with using these devices has been recommended [13, 127] .

6 Rollators

Rollators are a class of walkers used to address a broad range of mobility limitations, particularly to address respiratory, weakness, pain, or balance impairments [13]. This class of aids is typified by a walking frame with either 2-, 3-, or 4-wheels. The 2-wheeled version is distinguished with standard posts on the rear legs of the frame. On the 3- and 4-wheeled versions, the rear legs are wheeled and the front leg(s) feature swivel-mounted wheel(s) for increased manoeuvrability.

Since the 2-wheeled version can only be moved forward when the frame is unloaded, this version is more appropriate for individuals requiring higher levels of transfer support and stability [84]. In comparison, elderly subjects using the 4-wheeled version demonstrated tighter turning and improved mobility on varied terrain compared with using a 2-wheeled rollator [129]. A preference for the fully wheeled rollator over the 2-wheeled rollator to facilitate mobility has been reported in frail elderly [77] and patients with Parkinson's disease [25].

Although prevalence estimates have not been well-reported, the available numbers indicate a high number of 4-wheeled rollator users worldwide. In Sweden, a reported 250 000 people use 4-wheeled rollators (4% of total population) [17] and in the province of Ontario (Canada), an estimated 50 000 (0.4% total population) new 4-wheeled rollators were publicly subsidized between the years 2001-2006 [100]. Since prescriptions are not required to obtain these devices, private purchases are common and are not accounted for in these user estimates and considering the projected growing elderly demographic, actual user numbers are likely to rise. Due to the prevalence of the 4-wheeled version of rollators, for the remainder of this thesis the term 'rollator' will reference the 4-wheeled version.

Rollators are a popular type of aid prescribed to address impairments that affect mobility activities. In particular, the effect of using rollators on mobility have been examined for their potential benefits to walking efficiency, lower limb loading, and balance control [13]. The potential benefit of rollator use on walking endurance has been examined in the chronic obstructive pulmonary disorder (COPD) population. Solway et al. found that rollator use was associated with decreased dyspnea (shortness of breath) and rest times over a six-minute walk test (6MWT) in COPD patients enrolled in a respiratory rehabilitation program [125]. In a subgroup with greater disability, indicated by an unaided 6MWT distance <300m, there was also an improvement in the walking distance associated with rollator use [125]. Probst et al. observed improved respiration capacity and efficiency, indicated by increased ventilation volumes and reduced oxygen uptake, in walking with a rollator [109]. Gupta et al. examined longer-term effects of the use of a rollator on walking distance [48] and quality of life [49] in COPD patients following completion of a pulmonary rehabilitation program. The reduced dyspnea and increased walking distance benefits associated with using a rollator were confirmed to be consistent 4- and 8-weeks following rehabilitation [48]. Compared to a control group that did not receive a rollator following rehabilitation, the rollator intervention group demonstrated improved mastery, indicating a higher extent to which an individual feels they can cope with the limitations of their condition [49].

Use of rollators has been suggested to alleviate pain associated with lower limb joint loading in the arthritis population by permitting a portion of body weight to be shifted to the upper limbs [13, 32] . However, there is little empirical evidence to support this view [13]. Ishikura found a substantial reduction in peak lower-limb loads walking with a pickup walker, indicating the potential effect of rollators [58], but may be confounded by the distinct walking pattern adopted in pickup walker use [13]. A biomechanical analysis of gait with a rollator conducted by Alkjaer et al. found reduced peak knee and ankle moments and a corresponding increase in hip extensor contributions for forward progression [3]. Both the cited studies have been limited to data collected from healthy adult controls, and generalization to patients has been assessed.

Over 50% of the older adult population who use assistive devices in Canada use them to prevent falls or to ‘make me feel steady/balanced’ [111] indicating the potential importance of the rollator in addressing balance control impairments. Despite widespread clinical acceptance of balance benefits, little empirical evidence is available to indicate the advantages of rollator use in the control of balance [13]. The existing body of work consists primarily of laboratory-based studies investigating the effect of rollators on lower limb behaviour. Liu et al. examined the effects of rollators on spatiotemporal gait parameters in older adults with history of falls (but without experience using assistive devices) [74]. Compared to unaided gait, rollator use was associated with adopting a slower and more cautious gait pattern, characterized by decreased cadence and speed, coupled with a decrease in time spent in single support [74]. Bateni et al. investigated the consequences of walking frames on performing balance recovery reactions in response to support surface translations and found a significant reduction in stepping behaviour with the walker [12]. Although upper limb behaviour was not reported, this finding suggests that the upper limbs contributed to sensing and/or generating torques to maintain stability.

Conversely, there have also been reports that walking frames may actually increase the risk for falling [12, 14, 22, 127, 83]. Examination of hospital admissions indicate the majority of walker-related injuries occurred as a result of falling [22], and walkers were associated with seven times as many injuries as canes [127]. However, the fact that walker users typically have higher levels of balance impairments than non-users (e.g., cane users) confounds the potential hazard associated with using walkers. The potential mechanisms of increased risks for falling have been assessed. The potential for walker tipping has been described during use of standard pickup walkers [34] and in rollators during the sit-to-stand transition [41]. In their study investigating compensatory stepping behaviour with walkers, Bateni et al. found that the walking frame interfered with lateral stepping reactions, demonstrated by significantly shortened steps and frequent collisions between the foot and frame [12]. Furthermore, holding onto a walker interferes with upper limb reactions to recover balance, such as grasping a handrail or other support surface [14].

7 Examining rollator use to maintain balance in everyday activities

In light of the prevalence of use and the lack of studies on examining effectiveness in preventing falls [13], the current work will focus on the long term research objective of assessing the effectiveness and challenges associated with using rollators to maintain balance in everyday conditions. In the current work, we hypothesize that adopting an ambulatory measurement approach, will expose specific circumstances influencing balance benefits and additional fall risks.

To address these objectives, three objectives specific to this thesis are identified: 1) Develop an autonomous measurement tool to measure balance control and information about the environmental context in everyday situations; 2) Examine the characteristics of upper limb involvement in maintaining stability and factors influencing their control; and 3) Evaluate the utility of using an ambulatory measurement approach to directly observe the stabilizing benefits and adverse consequences associated with rollator use in everyday life situations.

The objective of developing an autonomous measurement tool to measure balance control and information about the environmental context in everyday situations is first addressed in Chapter 3 of the thesis. The two initial studies (Chapters 4 and 5) address the lack of understanding of the characteristics of upper limb involvement in maintaining stability and factors influencing their control in standing and walking, respectively. Following the laboratory studies establishing the link between upper limb involvement and maintaining stability, the utility of employing the iWalker tool in conditions reflective of everyday life is assessed in Chapter 6. The thesis concludes with a chapter summarizing the key findings and their implications on clinical use of rollators, future assistive device designs, and guide future research.

Risk Factor	Significant/Total[+]	Mean RR-OR[*]	Range
Muscle weakness	10/11	4.4	1.5-10.3
History of falls	12/13	3.0	1.7-7.0
Gait deficit	10/12	2.9	1.3-5.6
Balance deficit	8/11	2.9	1.6-5.4
Use mobility aid	8/8	2.6	1.2-4.6
Visual deficit	6/12	2.5	1.6-3.5
Arthritis	3/7	2.4	1.9-2.9
Impaired ADL	8/9	2.3	1.5-3.1
Depression	3/6	2.2	1.7-2.5
Cognitive Impairment	4/11	1.8	1.0-2.3
Age >80 y	5/8	1.7	1.1-2.5

[+]Number of studies with significant odds ratio or relative risk ratio in univariate analysis

[*]Relative risk ratios (RR) calculated for prospective studies. Odds ratios (OR) calculated for retrospective studies

ADL = activities of daily living

Table 1: Most Common Risk Factors for Falls in 16 Studies, adapted from [31]

Chapter 3
iWalker: Development of an Instrumented Rolling Walker to Assess Balance and Mobility in Everyday Activities

1 Introduction

The capacity for independent walking, or the ability to move with an upright posture from one place to another, strongly influences health-related quality of life [122]. Independent walking promotes maintenance of healthy physical, metabolic and cognitive functioning [63], facilitates social interactions [31], which is an important factor associated with maintaining cognitive function [1]. Conversely, when walking ability is significantly compromised there is a 3 to 5 times increase in likelihood for dependency with activities of daily living [55]. With mobility disability currently affecting 1 out of every 3 Canadian senior citizens [101] and likely to rise, the impact of addressing mobility issues will be immense.

Mobility aids, such as canes, crutches and walkers, play an integral role in facilitating independent walking [13]. In particular, rollators (4-wheeled walkers) are a class of mobility assistive devices used to address a wide range of gait and balance issues. For example, these devices facilitate independent walking in the chronic obstructive pulmonary disorder (COPD) population by improving walking efficiency [125] and providing a convenient seat for resting. In the arthritis population, rollators alleviate pain associated with lower limb joint loading by permitting body weight to be shifted to the upper limbs [32]. However, over 50% of the older adult (65 years or older) population who use assistive devices in Canada use them to prevent falls or to ‘make me feel steady/balanced’ [111] indicating the importance of the rollator in addressing balance control impairments.

1.1 How can rollators assist balance control?

Considering the inverted pendulum model, upright balance is maintained by controlling the position and motion of the body’s center of mass (COM) within the base of support (BOS), typically defined by the placement of the feet [140]. Under this model, stabilizing torques comprise of two components: 1) the moment arm, defined as the distance between the point of application of ground reaction forces (center of pressure (COP)) and COM; and 2) vertical load

(F_z) [76, 142]. The net righting torque acting on the COM is the product of the vertical load and moment arm. During quiet standing (without upper limb involvement), stability is maintained through moments predominantly generated by the hip and ankle muscles [142, 105]. During locomotion, stability is challenged because the COM and BOS are dynamically changing according to the gait cycle. In fact, the COM is only within the BOS during double stance, typically only 20% of the time [105]. However, the body is able to estimate upcoming changes in COM and BOS from ongoing movements to adjust posture and stepping patterns to produce stable gait [105].

Despite its importance, the literature examining the effects of rollators on balance control is relatively sparse, primarily consisting of laboratory-based studies investigating the effect of rollators on lower limb behaviour. Liu et al. examined the effects of rollators on spatiotemporal gait parameters in older adults with history of falls (but without experience using assistive devices) [74]. Compared to unaided gait, rollator use was associated with adopting a slower and more cautious gait pattern, characterized by decreased cadence and speed, coupled with a decrease in time spent in single support [74]. Bateni et al., investigated the consequences of walking frames on performing balance recovery reactions in response to support surface translations. The available lateral space for compensatory stepping was limited by the legs of the walking frame, resulting in shortened steps and frequent collisions between the foot and frame [12]. While the focus of the study was on the potential hazards associated with using the walking frame, a significant reduction in stepping behaviour with the walker was observed [12]. The rationale for alterations in stepping patterns suggested in these studies is that walking frames facilitate balance control by providing an increased BOS and additional sensory feedback to the upper limbs [13, 74].

Using a rollator permits the upper limbs to be combined with the lower limbs to sense body position and generate stabilizing torques. Our research group has investigated the upper limb contribution to maintain balance by measuring the upper limb forces generated through the rollator in healthy young controls. In quiet and perturbed standing, upper limb involvement (indicated by upper limb COP excursion) was selected preferentially over the lower limbs to effect stabilizing moments for frontal plane control [134]. During gait, upper limb involvement while walking under balance challenged conditions (i.e., along a narrow beam) produced increased upper limb COP excursion and mean vertical loading (F_z) compared to baseline (i.e., level ground) [137]. This evidence, combined with the studies examining the consequences of walker use on lower limb behaviour, highlight the importance of measuring the upper limb contribution to stability while using a rollator.

1.2 Are rollators effective in assisting balance control in everyday life situations?

Evidence to evaluate the effectiveness of these devices typically employed laboratory tools, such as forceplates to record forces and movements, or pressure-sensitive mats to measure foot placement. While the modern gait and balance laboratory is adept at probing the biomechanics and neuromotor control of walking with exceptional depth, there have been concerns that laboratory-based assessments may not be transferable to walking in everyday life [36, 121, 28]. For the purposes of this thesis, everyday life is defined primarily by the physical environment encountered in day-to-day activities, including the unpredictable and dynamic conditions that challenge the control of walking (e.g., distractions, moving obstacles). Intuitively, walking along a straight, level and well-lit path is less challenging than moving in the complex settings that are typically encountered in everyday situations. However, data to support (or reject) this claim are lacking and the connection between assessments conducted in the laboratory and capability to walk independently in everyday life situations remains to be addressed.

Two factors that could limit the transferability of existing laboratory and clinical assessments of walking to the everyday life are considered: 1) the brief duration of laboratory assessments; and 2) the unpredictable and complex nature of the everyday environment. One of the challenges for falls research is that the occurrences of fall events are relatively rare. For example, the annual rate of falls in adults over 65 years is 47.7 per 1000 individuals in Canada [99], roughly translating to 1 fall every 2 years. Due to their rarity, methods to capture falls data have relied on self-reporting instruments, such as calendars [33] or falls diaries [90], and are limited in detail. For example, Stevens et al. used self-report data to examine United States emergency room admissions and found that older adults demonstrated a 7 times higher incidence of walker-related injuries compared to cane-related incidents [127]. Considering that walker users typically have higher levels of neuromuscular impairments, the potential hazard associated with using walkers is confounded by the intrinsic balance deficits of the population. Importantly, Stevens et al. recommended that more information detailing the immediate circumstances preceding the fall are needed to better understand the contribution of fall risk factors [127]. One of the main barriers to study these factors is a lack of available methods to record the detailed behavioural and environmental information prior to fall events, which occur rarely.

Walking in everyday environments includes encounters with novel combinations and sequences of physical factors, such as other pedestrians, crosswalks, and terrain changes, which demand successful adaptation of functional capabilities. Studies investigating the adaptive control of locomotion has focused primarily on the influence of the immediate spatial surroundings, including obstacles (e.g., pedestrians) [45], terrain (e.g., icy surfaces) [98], or objects to be manipulated (e.g., purse) [14]. The motivation for studying these factors stems from epidemiological studies linking the occurrence of falls to the presence of environmental hazards, such as throw rugs or grab bars [116]. By identifying the environmental factors that pose a fall risk, interventions to reduce the frequency of encountering them can be designed. However, reliance on an assistive device, such as a rollator, is likely to introduce different sets of environmental factors that impact safe walking. While some potential factors may be generalized from unaided gait, specific factors that influence balance behaviour in walking with a rollator remains to be identified [13].

1.3 An alternative testing approach: Ambulatory measurement in everyday environments

An alternative approach to the systematic laboratory approach is to measure and record the behaviour of individuals in everyday life activities, where environmental factors are not experimentally delivered or constrained. Advances in ambulatory sensing and acquisition technology are creating opportunities to measure and record in everyday environments, over extended periods and with sensitivity and reproducibility rivalling laboratory measures. For example, ambulatory measurement methods have been applied to other health related concerns such as measuring cardiovascular output in daily activities in stroke survivors [43] and providing temporal gait parameters to quantify movement patterns [64]. Enabling measurement outside of the laboratory will permit researchers to observe balance behaviour within the complexities of the everyday setting and identify influential environmental factors, potentially informing about new associations that may not have been previously hypothesized. Extending data collection durations is a key point needed to increase the probability of capturing a fall or near-fall event.

While one needs to extend collection sessions to accommodate the complexities of the changing intrinsic and environmental factors, this must not be done at the expense of temporal precision. Balance recovery reactions are rapid events, taking place in hundreds of milliseconds [39], highlighting the need to record with a high temporal resolution. Records prior to such events are also needed to reconstruct the sequence and context in which the reaction took place. The challenge is to develop tools to provide the sensitivity and collection length needed to execute the proposed approach to examining everyday rollator use.

The scope of this chapter is to present the development of an ambulatory measurement tool that could serve to examine the effectiveness of rollators in facilitating balance in everyday life, and to determine any potential unintended consequences associated with rollator use. Specifically, this chapter presents the development of a novel instrumented 4-wheeled walker (iWalker) to examine the influence of environmental factors on balance control under everyday circumstances. The iWalker is designed to simultaneously capture details of walking behaviour, particularly indicators pertaining to maintaining balance, and the physical features of the immediate environment to provide contextual information. This chapter describes the iWalker development process by detailing: 1) the rationale for selection of the sensors; 2) implementation details of the iWalker, including calibration and experimental validation; and 3) clinical examples to demonstrate the utility of the iWalker as a data acquisition tool to capture behaviour in everyday life contexts.

2 Measurement rationale, hardware and validation

2.1 Overall criteria for design

The rationale for selection of sensors and support systems (e.g., power supply), implementation into the rollator frame, and validation testing data are described for each of the iWalker measurement capabilities. Subsections are divided into measurement of balance control, walking kinematics, spatial surroundings, and the data acquisition and power systems. To achieve the objective of an ambulatory measurement tool for everyday activity study protocols, the following design criteria are considered:

1. **Match between measurement and factor**: The primary requirement of the system is to collect as accurate a representation of the desired factors (i.e., balance control output from the limbs, movement kinematics, and spatial surroundings).

2. **Minimal supervisory input**: The requirements for researcher (and participant) input are to be minimized towards a system capable of autonomous collection over extended collection durations (4 hrs or greater). This includes minimizing setup time and eliminating needs for on-line control.

3. **Maintain accessibility to the everyday environments and activities**: To permit participants' access to everyday environments and activities, additional size and mass of the components added to the rollator are to be minimized.

4. **Maintain appearance and form**: The perception of a typical mobility assistive device is desired to encourage natural behaviour from the participants (and others). Similarly, maintaining the participant's typical appearance and form is desired (e.g., typical clothing and footwear, no sensors affixed to participant).

2.2 Upper limb kinetics: vertical load and COP estimates

Design/measurement rationale

A key distinction between unaided walking and walking with a rollator is the availability of upper limb support. The upper limbs are unlikely to generate movement substantial enough to measure using kinematics, such as hand accelerations. Hence, quantifying their contribution requires measuring the forces generated through the rollator. In studies conducted previously by our group [134, 136], the upper limb vertical loading (F_z) and COP measured through the forces applied to the rollator indicated the importance of considering the upper limb contributions. A shift in control from the lower to the upper limbs in the frontal plane was demonstrated in standing and level ground walking, likely due to the wider base of support afforded by the position of the hands [134, 135, 136]. Hence, the contribution from the upper limbs needs to be measure to assess a participant's balance behaviour in rollator-assisted walking.

Mechanically, the moments that contribute to stabilizing the body arise from vertical and transverse (shear) force components (see equation 10 in Appendix A). While the shear forces can play a substantial role in maintaining stability, estimates of their contributions in quiet standing indicates that they play a smaller role compared to the vertical loading shifts (Figure 33 in AppendixA). Combined with the overall vertical load (F_z), the COP estimated from shifts in vertical loading can be employed as a proxy measure to indicate the level of upper limb involvement in maintaining stability.

Implementation

To measure upper limb vertical loading kinetics, 4 single-axis button load cells (SLB-250, Transducer Technologies, USA) were vertically mounted into each leg of the rollator frame (Figure 1, left panel). These sensors were selected for their small size factor, to maintain the form and appearance of a typical rollator, and 250 lb load capacity to ensure seated activities were captured. In-line strain gage amplifiers (LCV-U5-CAB, Lorenz Messtechnik GmbH, Germany) were mounted underneath the seat (Figure 1, middle). Each load cell was calibrated by placing one wheel of the rollator (i.e., load cell) on top of a forceplate and loading the iWalker using standard weights. Linear regression results between the load cell output and forceplate signals were used to transform recorded signals into force units. COP was calculated from the relative difference in vertical load between the front and rear (sagittal plane), and left and right legs (frontal plane) using the following equations (free-body diagrams are shown in Appendix A):

$$COP_{sagittal} = \frac{(F_{FrLeft} - F_{ReLeft}) + (F_{FrRight} - F_{ReRight})}{F_{FrLeft} + F_{FrRight} + F_{ReLeft} + F_{ReRight}} \qquad (1)$$

$$COP_{frontal} = \frac{D_{front} . (F_{FrLeft} - F_{FrRight}) + D_{rear} x (F_{ReLeft} - F_{ReRight})}{F_{FrLeft} + F_{FrRight} + F_{ReLeft} + F_{ReRight}} \qquad (2)$$

where F_{FrLeft}, $F_{FrRight}$, F_{ReLeft}, and $F_{ReRight}$, represent vertical forces from the four load cells and D_{front} and D_{rear} are distances of the load cells to the midline of the walker (22.3 cm and 26.6 cm, respectively). The total vertical load (F_z) was measured by the sum of the four load cell outputs.

Validation

To validate the iWalker vertical loading (F_z) and COP measurements, data were collected from two healthy young participants under 2 conditions: 1) quiet and, 2) perturbed standing. In the quiet standing condition, the participants (1 female (21 years); 1 male (32 years)) stood while holding the iWalker with feet placed pelvis-width apart. In the perturbed standing condition, lateral pushes to the shoulder were randomly delivered by the experimenter to elicit a transient stabilizing response. For all trials, the iWalker was placed on top of a forceplate (AMTI, OR-6) and trials lasted 60 s. The resulting iWalker records were directly compared to those collected by the forceplate placed directly underneath the iWalker. The correlation between iWalker and forceplate measurements, indicated by mean R^2 value across all trials, was high across COP and F_z measures (medial-lateral (M/L) COP: mean $R^2 = 0.973$; anterior-posterior (A/P) COP: mean $R^2 = 0.932$; F_z: mean $R^2 = 0.921$). Once the sensor readings were validated, we conducted an initial study to relate the measures to upper limb involvement to assist in maintaining stability.

An initial study conducted with the iWalker was conducted to characterize the upper limb contribution to frontal plane balance control during rollator-assisted walking. To assess their role in assisting frontal plane stability, 11 young adults walked under: 1) normal conditions; and 2) simulating balance impairment by walking along a narrow (5 cm wide) wooden beam. Every participant demonstrated increases in both the magnitude of M/L upper limb COP and mean vertical loading (F_z) under balance challenged conditions compared to normal walking (Figure 2) [132]. These results supported the hypothesis that the upper limbs are used to compensate for restrictions to lower limb capabilities to produce stabilizing torques and validates the iWalker upper limb kinetics approach to measure their contribution.

2.3 Spatiotemporal foot placement

Design/measurement rationale

To achieve a comprehensive understanding of postural control, lower limb involvement must also be measured to complement measures of the upper limb contributions. While our previous studies have demonstrated a general reduction in lower limb involvement in quiet standing and steady-state walking with rollator use, there may exist circumstances necessitating lower limb involvement. For example, we have observed a combined upper and lower limb response to stabilize against lateral perturbations that may have otherwise exceeded the upper limb capabilities [136]. Furthermore, information of the phase of gait has been shown to be an important determinant of balance recovery reactions such as stepping over obstacles [39].

The most common methods of measuring lower limb gait parameters in an ambulatory fashion utilize kinematics sensors, such as accelerometers, angular velocity sensors (gyroscopes), or pressure-sensitive switches. Temporal parameters related to gait stability, including double support time, step time variability and asymmetry, have been extracted from kinematics sensors attached to the lower limbs [6], or by instrumenting footwear with pressure-sensitive switches or insoles [119]. Temporal and spatial parameters of foot placement are complementary indicators of balancing behaviour during walking, which indicate different strategies of maintaining stability. While ambulatory methods of capturing temporal stepping parameters are available, extracting spatial measurements, such as step length and step width, have been less reliable. Estimating distances using kinematics sensors, such as accelerometers, have been problematic due to drift issues [10]. Furthermore, we opted to explore methods of measuring lower limb behaviour that did not require instrumenting participants in this initial work.

Implementation

Video was captured using a portable digital video system camera (Archos Helmet Camcorder/404 Media Player, Archos, Inc., France) featuring digital video streaming (sampling rate 30 Hz) to an on-board hard drive and integrated battery power supply. Mounted to the front stabilizer bar of the rollator frame and aimed back at the feet, the video camera was outfitted with a wide-angle lens (1.7mm, Edmund Optics, USA) to maximize the field of view. After correcting for lens distortion, reflective markers fixed to participants' shoes above the 1st and 5th metatarsals were tracked offline with video capture software (Peak Motus 7.0, Vicon, UK) which extracts 2D image coordinates of each marker. Image coordinates were scaled according to a calibrated grid system to estimate true position, and step width was calculated as the average mediolateral distance between the toe markers on consecutive footfalls.

Validation

The current step width algorithm was tested on two healthy young adults (1 male, 24 years; 1 female, 25 years) ambulating across an in-lab walkway with varying step widths: preferred, narrow, and wide. Preliminary results indicate that the step width estimates correlated strongly with those computed by a gold-standard Vicon optical motion capture system in the lab (Figure 3.3, right, R^2=0.9588) [23].

2.4 iWalker motion

Design/measurement rationale

Walking kinematics (distance, velocity, and acceleration) are key outcome measures that have been used to measure global walking competency. Improvements in gait velocity has been used to assess the effectiveness of exercise interventions to prevent falling [91], as an indicator of the consequences of improved balance control [33]. Gait velocity also has a strong influence on many gait and balance measures, such as step length [35] and step width variability [16]. Walking distances, typically measured by the distance covered over a 6 minute timed walk, is a common outcome measure reflecting gait efficiency and endurance limits [124]. Measuring displacements and velocity is relatively straightforward on wheeled platforms using potentiometers, optical or magnetic encoders to count wheel rotations.

Implementation

To measure distances, the iWalker was fitted with optical encoders quantifying wheel rotation and direction (i.e., forward and reverse). The optical encoders pulsed as high-contrast wheel markings painted onto the wheels passed the sensors (Figure 1, right). The encoders were calibrated by moving the rollator over a fixed 10m distance and recording the number of encoder pulses. The resulting spatial resolution of the encoders was 6.28 mm/pulse. Distances were time-differentiated to provide instantaneous estimates of iWalker velocity.

Validation

To assess the validity of using the iWalker velocity as an indicator of gait speed, a sample of walking was collected from three participants (2 male stroke survivors, 50/63 years; 1 female traumatic brain injury patient, 66 years) who required the use of a rollator for all independent mobility. Participants walked over a 4m long pressure-sensitive mat (GaitRite 3.9, CIR Systems, USA) that measures spatiotemporal measures of foot placement. Since the GaitRite system captures footfall information, the average gait speed calculated by the distance and time between the first and last steps on the mat. The average iWalker velocity was taken as the mean of the instantaneous velocity recorded over the time taken to traverse the mat. The corresponding gait and iWalker velocities (Figure 4), matched well (R^2=0.98) indicating that the iWalker velocity is reflective of the walking speed.

2.5 Spatial surroundings (Obstacles, terrain)

Design/measurement rationale

The primary environmental influences considered to influence balance control were physical features of the immediate spatial surroundings. The majority of environmental factors in homes that have been identified as potential tripping hazards are physical features, such as vertical thresholds, low lighting levels, carpets and slippery surfaces [116]. Possible sensing capabilities to record the environmental surroundings include global positioning satellite (GPS), infrared, ultrasound and video. While widely available, we chose not to include GPS as a sensor due to the low spatial resolution and limitations to indoor locations. Ultrasound and infrared sensors can effectively measure distances from static and dynamic targets using reflected sound and infrared light energy, respectively. However, they cannot provide information regarding the nature of the target (e.g., size, shape or movement), and are subject to noise from other sources of energy [37]. Video is a promising sensing modality, offering the most flexibility and richness of information. Image processing algorithms can extract useful features (e.g., movement, colour, shape) and estimate distances with multiple cameras (e.g., stereo vision).

Information regarding the vertical and frontal plane movements was desired to complement the video record of the environment. Specifically, vertical accelerations of test vehicles have been used to measure of surface roughness [57], and would assist in characterizing the effects of transitions, such as vertical thresholds. There may also be important features drawn from accelerations in the frontal plane, such as collisions or tilting. Two types of sensors were considered to capture 3D kinematics: angular (gyroscopes), or linear (accelerometers). Since the majority of desired information was most likely to produce linear movements, a linear accelerometer sensor was chosen for implementation.

Implementation

A second portable digital video recording unit (see Spatiotemporal Foot Placement section for specification) was employed to continuously capture the spatial surroundings of the iWalker. The camera unit was mounted underneath the seat of the rollator (Figure 1, middle) and was aimed at the ground to sample (30 Hz) the immediate and upcoming environment (Figure 5). The camera was positioned to include the front wheels of the rollator as a frame of reference. The triaxial accelerometer sensor unit (SEN-00847, Sparkfun Electronics, USA) selected to record 3D linear accelerations was also mounted to the custom-built case housed underneath the seat of the rollator. Each axis was calibrated by rotating the device to capture the gravitational pull in the positive and negative directions and correlating the minimum and maximum signals to -1 and +1 g, respectively.

Validation

To assess the feasibility of using the video record to identify environmental features, pilot testing was conducted using data from a healthy young adult (male, 20 years) walking freely in and around the vicinity of an urban rehabilitation hospital with the iWalker. Manual inspection of the video record identified negotiation of terrain transitions (e.g., thresholds, carpeting, sidewalks), changes in lighting intensity, stationary obstacles (e.g., furniture, doorways) and pedestrian traffic. These features were validated against a physical inspection of the hospital space traveled by the participant conducted a posteriori.

The validity of using 3D accelerations of the rollator to capture vertical and frontal plane movements was assessed using the same dataset. As shown in Figure 3.6, clear increases in the vertical acceleration magnitude were observed when walking over uneven sidewalk compared to a smooth hospital floor. The magnitude of the vertical acceleration time series was clearly larger on the rough sidewalk (RMS value = 0.057 g; blue trace) compared to the hallway (RMS value = 0.012 g; red), indicating a relative roughness of the outdoor terrain as 4.75 times greater than smooth terrain. The 3D sensors accelerometers also captured peaks associated with elevator acceleration and decelerations, further validating their utility in capturing physical features of the spatial surroundings.

2.6 Support systems (Acquisition, Storage, Power)

Design/measurement rationale

While the video systems included integrated data storage and power supply, the load cells, optical encoders and triaxial accelerometer sensors required external support. Combined, these sensors required a minimum of 8 channels of analog data for acquisition with sufficient storage for 4+ hrs. To match the temporal resolution of the video capture equipment, the minimum sampling rate was 30 Hz. Finally, the power needs of the system required a minimum of 15 V DC output at 600mA for 4+ hrs. Furthermore, the criteria to minimize weight and size was considered.

Implementation

Analog signals were converted to digital (16 bit, sampling rate 50 Hz) and transmitted wirelessly via Bluetooth radio (BlueSentry-AD, Roving Networks, USA) to a PDA device (iPaq hx2190, HP Inc., USA). The PDA received and stored the data using acquisition software developed in LabView (National Instruments, USA). The PDA was chosen for its small size, microprocessor capabilities and expandable memory capacities (SD card). Importantly, the touchscreen interface and familiar Windows-based operating system facilitated setup times and maintained flexibility in organizing files. All on-board electronics were powered by a high-capacity rechargeable 18.5V lithium-ion battery pack (BatterySpace, USA). Finally, a custom-built ABS plastic case was designed and fabricated to house all electronic components underneath the seat of the rollator (Figure 1, middle).

3 Clinical examples

In this section, representative data collected from individuals who use rollators for independent walking are presented to demonstrate the utility of the iWalker for identification of everyday environmental factors that influence balance control. Two very specific examples of walking with the iWalker are described from two different patient cases that demonstrate the challenges of everyday environmental factors on maintaining balance control. The events include: 1) apparent recovery from an unexpected elevation change from uneven terrain, and 2) a collision between the foot of the user and the rollator.

3.1 Walking course

The featured examples were taken from a larger dataset where participants walked along a pre-defined course within and outside a rehabilitation hospital setting designed for the individuals to encounter typical challenges to mobility reflective of everyday, common activities. The course comprises of a path that included walking in hallways, up and down ramps, entering/exiting doors, negotiating around an oncoming pedestrian and taking elevators within an rehabilitation in-patient hospital in an urban setting. The course distance was approximately 300m in length, took less than 30 minutes to complete and included two designed opportunities for seated rests. Participants were also permitted to sit on the iWalker seat at any time during the course, if needed. Both participants rested 1-2 times during the course, with total rest times of < 2 minutes for both individuals over the entire course.

3.2 Recovery from external perturbations on uneven terrain

This example describes an apparent balance recovery reaction in response to a terrain conditions from a 46 yr old female with multiple sclerosis who self reported she used a rollator to address balance impairments. Within the pre-defined course, participants walked along 65m long outdoor walking section of sidewalk adjacent to the hospital. The terrain over the sidewalk was characterized as being 4.75 times rougher than smooth hallway terrain, as measured by the iWalker RMS vertical accelerations (Figure 6). A particularly large vertical acceleration during this section was observed (Figure 7, top left panel), potentially indicating a large transient perturbation. Inspection of the video record from the foot placement camera (Figure 3.7, right) confirmed a drop in the sidewalk. Following the large perturbation by 200 ms, an upper limb balance response was observed in the mediolateral iWalker COP on the same side of the foot stepping down onto the sidewalk. This example demonstrates the potential benefit of rollator use in facilitating upper limb involvement in maintaining frontal plane balance over uneven terrain.

3.3 Foot-device collision

In this example, a collision event between the foot of the user and the rollator was captured from a 63 yr old male stroke in-patient. The patient demonstrated mild-to-moderate impairments to the right side, indicated by Chedoke-McMaster Stroke Assessment scores of 5/7 for the right foot and leg with spasticity at the extremes of range of motion and rapid movement. The right hand demonstrated greater impairment (4/7) with spasticity throughout the range. To compensate for the right side hemiplegia, the participant displayed a consistent foot placement bias to the paretic side during walking tasks.

While walking down a hallway, the patient navigated around another patient in a wheelchair revealed from the scene camera (Figure 8, upper right panel). Following a change in direction, a collision between the right foot and the rear wheel of the rollator was observed in the foot camera video (Figure 8, lower right) and confirmed by the iWalker accelerometer revealing a sharp responses to the collision (Figure 8, upper left) in the vertical (black trace) and anterior-posterior (red) planes. Peaking at 60 ms following the foot collision, an increase in the iWalker vertical load (F_z) was observed (Figure 8, lower left), indicating a rapid upper limb loading behaviour in response to the apparently unexpected perturbation to the lower limbs. Notably, a decrease in F_z (green arrow) was observed 1.2 s prior to the foot collision, occurring during the 1-2 steps prior to the collision. Inspection of the foot camera video demonstrated the unloading behaviour occurred during execution of the navigation task.

This example supports the argument that rollators may introduce a tripping hazard in certain circumstances that may increase the potential risk to falling though the individuals did not fall in this specific circumstance. This individual case demonstrated repeated incidents of foot contact with the rollator frame, suggesting that individuals who present with strong lateral foot placement bias may be at higher risk for foot-device collision. In particular, the interaction between increased risk and turning behaviour is highlighted.

4 Discussion

The iWalker was designed to be able to permit evaluation of the effectiveness of using rollators to facilitate walking and prevent falling by providing the ability to assess balance in everyday environments over extended durations. Taking advantage of a relatively stable platform to mount sensors and instrumentation, the iWalker provides a detailed record of both intrinsic balance behaviour (i.e., foot placement, upper limb COP) and immediate spatial surroundings arising from mobility during navigation. One of the initial applications of the iWalker, to observe interactions between environmental factors and balance behaviour, was briefly described to illustrate the utility of the device and provide an everyday perspective of rollator-assisted walking.

The iWalker involved development of two aspects of movement control that we believe informative about balance control: foot placement and upper limb kinetics. To date, the iWalker foot placement method is the only alternative to one other known ambulatory method of measuring step width [11], an important measure of frontal plane stability. A key advantage the iWalker's video-based system, compared to a method based on inertial movement sensors (e.g., accelerometer/gyroscope), is a visual record of the footfalls providing visual confirmation of behaviour. For example, a foot-device collision may be inferred from stepping kinematics alone, but could be confirmed by inspection of captured video in combination with the accelerometer record. While the described method requires markers to be placed on the feet, efforts are being conducted for markerless lower limb extraction [94] and to provide further spatial stepping parameters (e.g., step length, toe clearance). To complement the spatial data, temporal stepping parameters can be resolved by coupling wireless accelerometers worn on the ankles of the individual to capture foot-contact and foot-off times. This approach has been used previously in non-walker studies looking at everyday walking behaviour among stroke patients [108].

Instrumenting walking frames with force transducers to measure upper limb kinetics has been conducted previously for standard (pickup) [9, 40], wheeled [5], and robotic walkers [112]. However, measurement and use of the upper limb forces transmitted through the walker specifically to assess balance control, expressed as COP and F_z, is an important distinction from previous efforts. Experimental studies conducted by our group have validated these measures of upper limb involvement in balance control during static (standing) [134] and dynamic (gait initiation [135], steady-state walking [137]) tasks. In both of the clinical examples presented in this chapter, the significant role of the upper limb involvement further supports the need for their inclusion in assessing rollator use in everyday life circumstances.

Capturing details about the environmental context with the iWalker, particularly the immediate physical characteristics, is also highlighted. The environmental context is an important perspective to consider for interpretation of data collected from ambulatory measurement protocols assessing everyday behaviour. For example, higher step width variability measured from everyday activities could be interpreted as increased risk for falling [16], or explained by increased obstacle avoidance from natural circumstances (e.g., pedestrian traffic). The iWalker captures information of the immediate spatial surroundings from the combination of walker motion and a video record of the immediate upcoming environment.

A key advantage that the iWalker provides is sufficient temporal resolution (30 Hz) that can be examined to reconstruct the sequence of events, necessary to observe the timing between balancing behaviour and the environmental factors. Not only is identification of the relevant environmental details essential, but the specific combination and the timing in which they are encountered are key factors determining behavioural strategies. For example, the phase of walking in which the obstacle is encountered is a key determinant in balance recovery from a trip [12, 39] and which foot is used to execute a compensatory step in response to an unexpected perturbation [105]. In the clinical examples presented, the close timing of upper limb reactions in response to the foot collision (60ms) and sudden drop in elevation (200ms) demonstrate the need for such temporal resolution.

4.1 Limitations

Initial evaluations of the iWalker demonstrated several limitations to the current methodology. The major limitation involved the intensive time required to inspect the video records. Manual inspection of video records was used to identify environmental features that influence balance control. Given the extended duration of records, automated algorithms to extract relevant environmental features (e.g., upcoming pedestrians) are needed to reduce the analysis load. Furthermore, methods to provide more detailed information of the environment (e.g., distance, speed and direction of movement) can be developed. For example, applying a stereo camera system could provide the desired spatial data of the upcoming environment [95].

Since the analog and two video acquisition systems captured data asynchronously, reconstructing the combined data record relied on accurate timestamp information. Although the clocks used in the equipment were fairly accurate, slight differences in timekeeping accumulated which introduced relevant phase lag over long periods. A regular (e.g., every 5 minutes) synchronization pulse to the acquisition equipment is proposed to remedy this issue.

The iWalker is limited to upper limb kinetics related to the vertical loading applied to the rollator frame, and does not measure transverse plane (shear) forces. While shear forces can act over the large moment arm of the height of the handles to generate substantial stabilizing torques, an assessment of their contributions from forceplate data in quiet standing indicated that they play a small role compared to the contributions originating from shifts in vertical loading (see Appendix A, section 2 for details). Hence, we consider the shifts in vertical loading measured by the iWalker to be an indicator of the level of involvement of the upper limbs in maintaining stability. A full evaluation of the biomechanical contribution of the upper limbs to maintaining balance would require consideration of both the shear and vertical forces applied to the rollator frame.

5 Conclusions

While the iWalker was primarily designed as a research tool to investigate the breadth and complexity of environmental factors challenging mobility, the platform may lead to future clinical applications and inform future design of assistive devices. The utility of using the iWalker to automatically collect and evaluate balance in patients for clinicians is being explored. The unique combination of intrinsic balance and extrinsic environmental information provided can assist in identifying rehabilitation targets. For example, training users to effectively execute negotiate uneven terrain. Patient assessments collected with the iWalker can also provide a basis to define key issues (e.g., avoiding collisions) including the specific circumstances in which they occurred (e.g., turning to the stroke-affected side). Prototypes of new design features, such as automated braking, fall risk alerts, or adaptive suspension, can be then be field tested for effectiveness in facilitating mobility and minimizing safety risks.

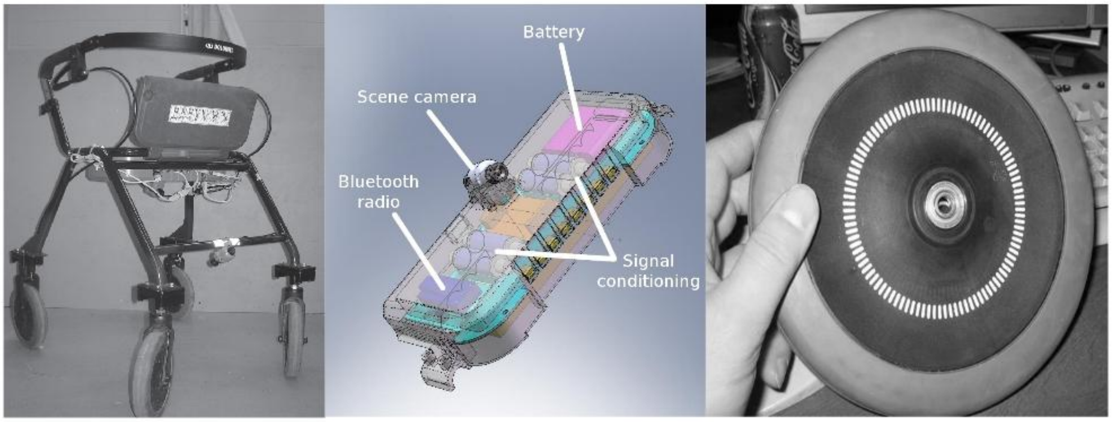

Figure 1: iWalker implementation. Profile view (left panel). Under-seat components including signal conditioning, acquisition, power, and scene camera mount (middle). Wheel marks for encoder tracking (right).

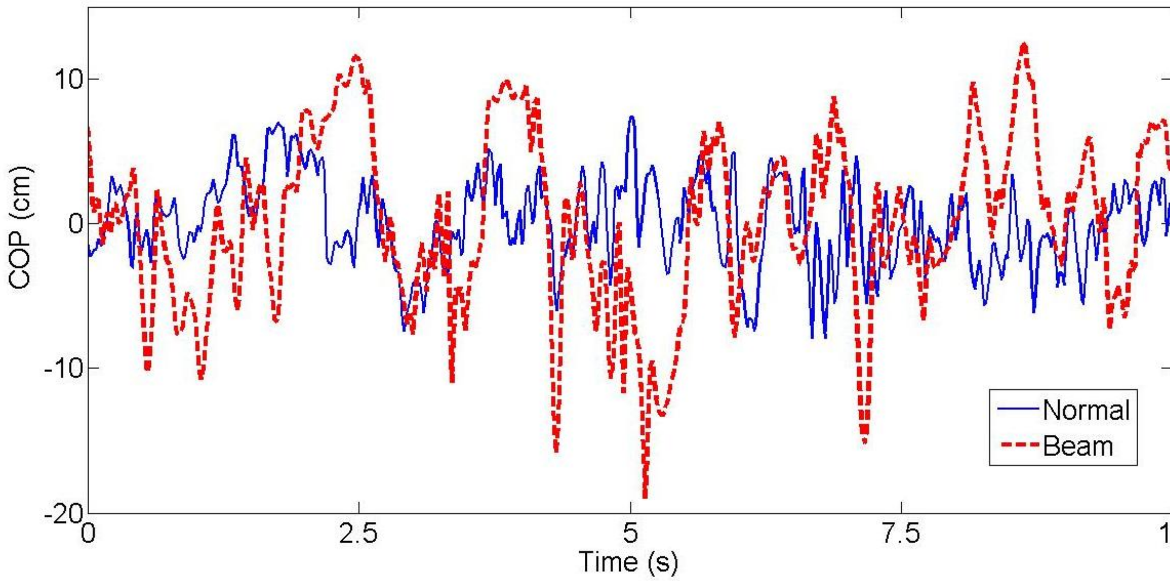

Figure 2: Sample 10s timeseries of unfiltered mediolateral upper limb COP from a healthy adult walking over a normal, level surface (blue trace) and under balance challenge conditions (beam, red trace).

Figure 3: Development of spatiotemporal foot placement measurement method. View of feet and markers from foot placement camera (left panel). Step width calculation (middle). Comparison of step widths calculated by the algorithm and Vicon (right).

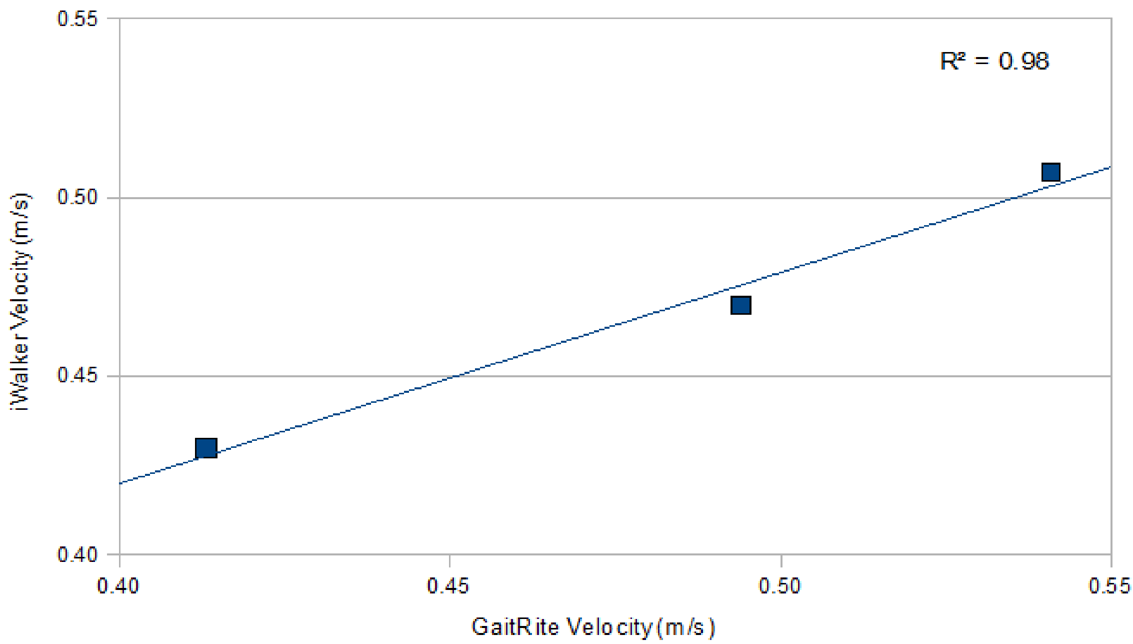

Figure 4: Comparison of mean iWalker velocity to gait speed (as measured by GaitRite) in three regular rollator users. Blue line indicates linear regression fit.

Figure 5: Screen capture from a video record of the scene camera aimed at the upcoming environment. Note that the front two wheels of the rollator are captured in the view to maintain a frame of reference.

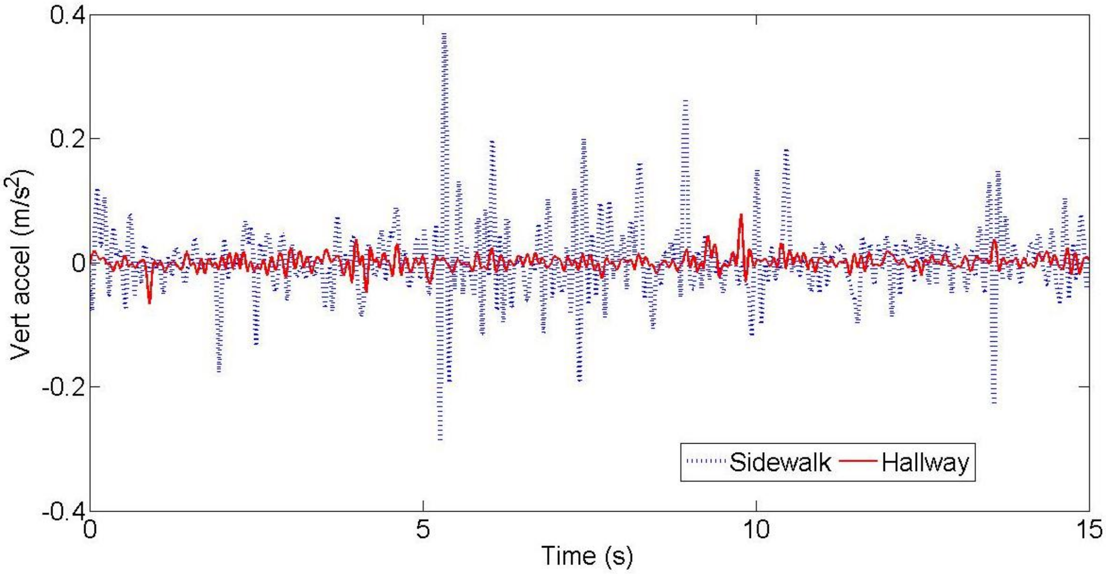

Figure 6: Larger vertical accelerations were observed while walking over uneven terrain (sidewalk, RMS value = 0.057 m/s2, blue trace) compared to smooth terrain (hallway, RMS value = 0.012 m/s2 , red trace).

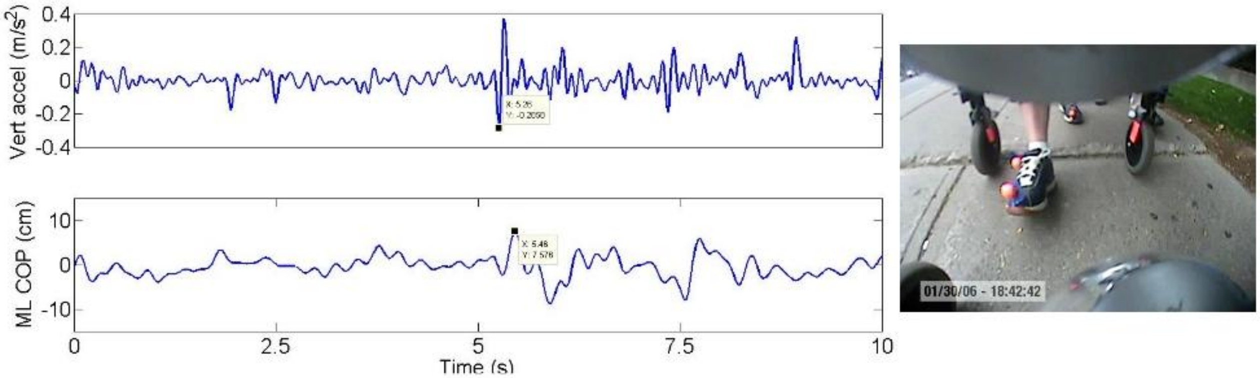

Figure 7: Clinical example 1: Recovery from external perturbations on uneven terrain. A sudden vertical acceleration (top left panel) was observed at 5.26s of the record, indicating a large drop due to uneven terrain (confirmed in the foot placement camera, right)

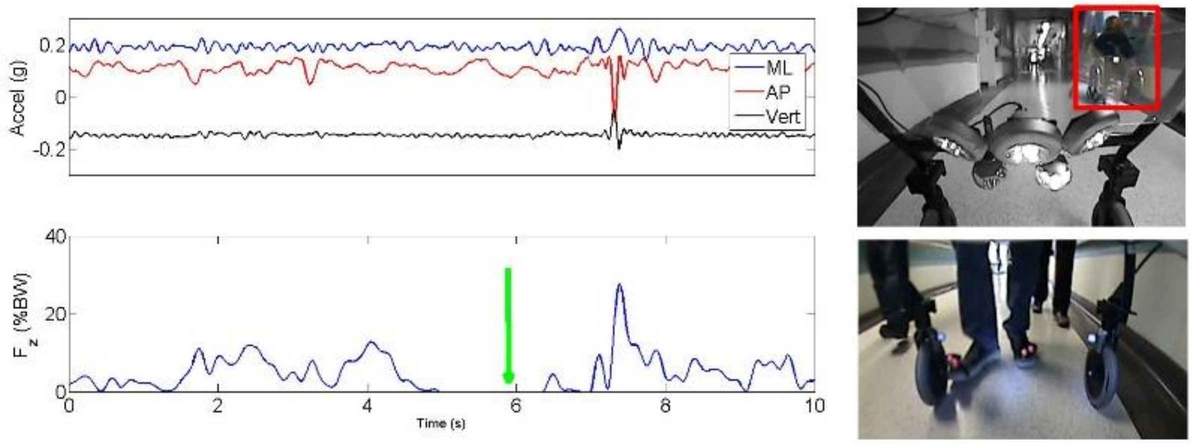

Figure 8: Clinical example 2: Foot-device collision. The patient executed a direction change to avoid a wheelchair pedestrian (boxed, top right), and subsequently collided his right foot against a rear wheel of the walker (bottom right). Timeseries data from the iWalker are plotted left, with accelerometer (upper left) and vertical load (lower left) plotted. The green arrow indicates a reduction in upper limb loading 1-2 steps prior to the collision.

Chapter 4
Bilateral Integration of Upper Limbs in Standing Balance Control with an Assistive Device

1 Introduction

Numerous studies have examined how the central nervous system (CNS) controls lower limb muscles to maintain upright balance, yielding insight into the role of specific affector (sensory) [60, 88], effector (muscle) [53, 76] and integrative (sensorimotor/cognitive) contributions [96, 80] to balance control for standing and walking. In contrast, there are far fewer studies that have investigated the role of upper limbs in maintaining upright stability. An important justification for a focus investigating the contribution of the upper limbs to stability control is the potentially important role of the upper limbs for users of ambulatory aids. Ambulatory aids, such as canes, crutches and walkers, play an integral role for millions of people worldwide by involving the upper limbs to address balance and mobility impairments [13]. For example, the rollator (or 'four-wheeled walker') is a walking frame frequently prescribed to facilitate standing and walking, supported by studies that have demonstrated use of these devices increased walking distances [125] and improved efficiency [109] in cardiopulmonary populations.

Despite widespread acceptance of clinical benefits, little empirical evidence is available to indicate the advantages and limitations of walker use in the control of balance [13]. Rollator use has been shown to provide some benefit to frontal plane balance control [132, 134]; however, those benefits may be offset by limitations in compensatory stepping reactions due to additional tripping hazards [12]. In light of the potentially important role played by the upper limbs (i.e., through the use of assistive aids), the current study was conducted to investigate how the CNS integrates upper- and lower-limb control for maintenance of standing balance when using an ambulatory aid, specifically when using a rollator.

The few studies that have focused on upper limb contributions to balance control have investigated unilateral contributions and have typically focused on a specific aspect of balance control (e.g., sensory contributions or reactive responses). For example, when only light fingertip touch is applied to a stationary object, the upper limbs provide tactile sensory input, which can lead to a reduction in center of pressure (COP) displacement during quiet standing [60, 61].

Under walking conditions, walking sticks and guide canes can provide haptic cues regarding features of the spatial surroundings, such as obstacles and terrain changes [106]. However, the role of the upper limbs in generating stabilizing forces as an effector of postural or balance control is less clear. Cordo and Nashner compared the perturbation-evoked EMG responses of the arm and leg, with and without the availability of a hand support [26]. With hand support, postural reactions from anterior support-surface translations shifted from lower limb to upper limb muscles with latencies comparable to the automatic postural responses in the leg. Elger et al. [38] investigated the coordination of hand and hip EMG responses to a lateral push while gripping a handhold in front of the participant. They observed large upper limb EMG bursts temporally coupled with persisting, but smaller amplitude, lower limb responses suggesting a coupled upper and lower limb response.

The focus of the present chapter was to advance understanding of the potential coupling of upper and lower limb control of upright stability under conditions that simulate to the use of assistive devices. Specifically, the current work was intended to develop understanding of the role of bilateral upper limb control in upright standing during quiet stance and in response to balance perturbations. An emphasis was placed on task situations, such as use of assistive aids, in which the upper limbs were permitted to provide mechanical contributions to balance control (in contrast with sensory paradigms which limit the loading [60, 61]). Considering the inverted pendulum model, the CNS must select appropriate strategies to control the position and motion of its center-of-mass (COM) with respect to the stability limits defined by the base of support (BOS). During bipedal standing tasks (no upper limb involvement), stability is maintained in the frontal plane predominantly through moments generated by the hip muscles, reflected by a shift in COP to the falling side [142]. In the sagittal plane, bipedal standing balance is maintained primarily through the moments generated through the ankle musculature.

With the introduction of a walking frame, the upper limbs become available to the CNS to provide mechanical as well as sensory influences on balance control. In contrast to sensory studies, the specific focus of this work is the nature of the mechanical contributions of the upper limbs and the impact to lower limb contributions. The focus of this study is to address the questions, ‘Do the upper limbs play a significant role in maintaining standing balance with an assistive device? If so, what are the strategies and control characteristics of upper limb contributions to balance control? ’

Overall we hypothesize that, even among healthy adults, the presence of hand supports will lead to reliance on mechanical support through the upper limbs. This would be reflected by measurable force and COP excursion associated with involvement of upper limb activity on control of balance and a reduction in COP activity from the lower limbs. More importantly, the upper limb contribution would increase with increased challenge to balance control, evidenced by increase in vertical load and COP excursion measured under the upper limbs. In parallel, we anticipate a reduction in COP excursion measured from the lower limbs when upper limb support is available reflecting a shift in control from the lower limbs to the upper limbs (due to the benefits afforded by upper limb contributions). We believe that this reliance on upper limbs and attenuation of lower limb contributions will be present in both static (stationary standing) and dynamic (transient perturbation) balance tasks when support is available through the upper limbs. The rationale for a ‘preferred’ reliance on the upper limbs may be due to the mechanical advantages afforded by the placement of the hands, including a larger BOS and height from the axis of rotation (i.e. feet) as depicted in Figure 9.

Furthermore, we explored the temporal synchrony of the development of torque and movement executed by the upper and lower limbs, reflected by COP, to provide insight into the potential link in central CNS control. Independent of the relative contribution to stability control (i.e., amplitude of COP excursion), we proposed that the timing of upper and lower limb stabilizing torque development would be temporally synchronized potentially reflecting a common control.

2 Methods

The contribution of the upper limbs to balance control while holding the walker was investigated in two task conditions conducted in two different studies: 1) Static, a stationary standing task, and 2) Dynamic, a standing task with applied transient lateral perturbations. The same assistive aid (an instrumented rollator), which individuals were permitted to load/use, was employed in all tasks. The rollator was fixed with the brakes locked in the ‘engaged’ position, which prevented movement during and between trials. These methods were approved by the local research ethics committee.

2.1 Study 1: Stationary standing

Eleven (11) healthy, young adults (4 male, 7 female, 20-35y, mass 50-100 kg (mean 68.2 kg)) provided informed consent to participant in the study. Participants reported no musculoskeletal or neurological impairment that might affect their balance. Participants performed the stationary standing task in 4 different task conditions: 1) normal stance (feet pelvis-width apart) without touching the rollator [Hands OFF]; 2) normal stance while holding the handles of the rollator [Hands ON]; 3) increased balance challenged (IC) stance (feet together, eyes closed, on a 1.9 cm thick medium-density foam) , without touching the rollator [IC-Hands OFF]; and 4) increased challenge stance while holding the handles of the rollator [IC-Hands ON]. Participants wore shoes throughout testing. In the Hands OFF condition, participants stood with their eyes fixated straight ahead at a target located 5m away, elbows flexed 20 degrees and palms down to approximate the arm position and posture used while gripping the rollator. In the Hands ON conditions (i.e., while holding the rollator), the handles were adjusted to the height of the radial styloid with arms hanging straight.

Prior to data collection, participants performed a single training trial (30 seconds) to acclimate to each of the task conditions, and to determine optimal foot position. During this trial, foot placement was adjusted such that the ankles were in line with the axis of rotation of the rear wheels of the rollator and participants were instructed to do whatever came naturally to stay upright. Following the acclimatization trial, participants were instructed to not shift the position of their feet. The order of conditions was randomized for each participant.

Four force plates (AMTI, BP-250-500), embedded within a wooden platform, were used to measure the ground reaction forces (GRF) underneath the feet and the rollator (Figure 10). A single force plate was located under the feet and three force plates were used to collect the GRF underneath the rollator. A resultant COP was calculated from the moments and forces measured from the 3 force plates under the rollator (Eqs. 1, 2) using distances shown in Figure 10. Free-body diagrams are shown in Figure 32, Appendix A.

$$COP_{frontal} = \frac{(COP_{y1}.F_{z1}) - (C + COP_{x2}).F_{z2} + (C - COP_{x4}).F_{z4}}{(F_{z1} + F_{z2} + F_{z4})} \quad (1)$$

$$COP_{sagittal} = \frac{[(COP_{x1} + A).F_{z1} + COP_{y2}.F_{z2} + COP_{y4}.F_{z4}] + B}{(F_{z1} + F_{z2} + F_{z4})} \quad (2)$$

For each trial of 60s, force plate records were acquired at 250 Hz and digitally filtered with a zero-lag 2nd order low-pass Butterworth filter (F_c= 10 Hz). The mean COP position was removed and root-mean-square (RMS) values of COP displacement were calculated for both frontal (M/L) and sagittal (A/P) planes. A 2-way ANOVA was conducted to analyze the effects of Hands (OFF/ON) and Challenge (Normal/Increased) for using SAS (SAS Inc., version 8.0). Separate analyses were conducted for the frontal and sagittal planes. Since upper limb COP measurements were only available for Hands ON conditions, a 1-way ANOVA was conducted to assess the effect of Challenge (Normal/Increased) on upper limb RMS COP and upper limb vertical forces. Upper limb COP analyses were also conducted separately for frontal and sagittal planes.

Temporal synchrony between the upper and lower limbs, for both M/L and A/P planes, was assessed by determining the proportion of participants with a significant mean square coherence between the COP responses of the upper and lower limbs. Coherence provides a measure of the linear correlation between two signals as a function of frequency [70, 120]. Values of close to 1 at a given frequency indicate a strong linear correlation between the signals, even in the presence of phase lag, whereas values near zero indicate no linear relationship at that frequency. Statistically, the threshold value to be significantly different from zero depends on the parameters used to estimate coherence [70, 120]. Using a 212 sample FFT length, 1000 point Hamming window, and 75% overlap, the critical value for a significant coherence is 0.44. The proportion of participants exhibiting any coherence value greater than this threshold was determined over the full (0-10Hz) range. Low and high frequency bands were identified based on reported ankle muscle rise times of 250-380ms [60, 88] during quiet standing. As one quarter of a full oscillation, these rise times correspond to a frequency range from 0.657-1.25 Hz is calculated. A cutoff of 0.625 Hz was adopted to differentiate between low (< 0.625 Hz) and high (> 0.625 Hz) response frequencies.

2.2 Study 2: Perturbed standing

Eight (8) of the 11 participants who completed the stationary standing task also participated in the perturbed standing task (4 male, 4 female, 21-34y, mass 45-98 kg, mean=71.4 kg). A magnet-release perturbation system was used to elicit a balance recovery response in the mediolateral direction. A belt around the chest was attached to a weight (2.0 kg) by an electromagnet located to the left of the body through a pulley system. The participant leaned laterally against the weight until a steady state position was attained. The magnet released the weight at a random time after the steady-state position was attained, and the participant reacted to the sudden change in equilibrium. Perturbation trials were recorded without a rollator [Hands OFF] and with a rollator [Hands ON]. A further 10 trials were performed with a rollator and a heavier weight (2.72 kg) to increase the balance challenge [IC-Hands ON]. As a secondary objective, to assess to effect of preloading the assistive device, a fourth condition where the palms are lightly touching the sides of the rollator using the heavier weight [IC-Light Touch] was conducted. Trials were presented in blocks of 10 perturbations for each condition, with the order of blocks randomized across participants.

Lower-limb COP data were acquired directly from a force plate (AMTI, AccuSway) situated beneath the feet of the participant. Data was sampled at 50 Hz. The forces applied by the upper-limbs were sampled using a custom built instrumented walker (iWalker). The iWalker measured the vertical loads in each leg of the frame using uniaxial force transducers (Transducer Technologies, SLB-250). An estimate of the movement of the COP of the walker frame, associated with the forces applied by the upper limbs, was calculated by resolving the vertical loads from each of the four legs of the frame into a single resultant force; the position of the resultant force was defined as the upper limb COP. Analog signals from the iWalker, sampled at 50 Hz, were converted through an on-board 16 bit analog-to-digital conversion unit (Roving Networks Inc., BlueSentry) and sent to a collection computer wirelessly via Bluetooth. To synchronize the iWalker and force plate records, a short pulse to the handles of the iWalker was performed at the beginning of each trial. The peak vertical force on the iWalker data record and corresponding valley in the force plate record was marked as time zero for the trial.

Baseline COP position values (mean value from 250-500 ms prior to onset of the perturbation) were subtracted from trial records and the peak COP excursion values were calculated. To test the main effects (Hands ON/Hands OFF; Normal weight/IC), contrasts were performed against the pooled and within-condition variance. For example, to assess the hypothesized reduction in lower limb peak COP excursion with the availability of the upper limbs, the pooled variance from all Hands ON trials was compared to the variance from Hands OFF trials. A paired t-test comparing the IC-Hands ON and IC-Light Touch conditions was performed to assess the effect of preloading.

As part of the evaluation of temporal synchrony between the upper and lower- limbs, an analysis of the onset latency times was conducted. Onset times for each participant were calculated from the ensemble average of the upper and lower limb COP trials. Onset of the COP response was defined as the sample prior to a sudden change in the lower limb COP rate greater than 10 cm/s. Trials in which upper limb COP did not meet the onset criterion were excluded from the analysis. Upper limb onset times were reported relative to lower limb times (i.e., negative values were prior to onset of the lower limbs). Paired t-tests were used to evaluate differences between task conditions.

3 Results

3.1 Study 1: Stationary standing

Table 2 summarizes the descriptive statistics (mean ±± SE) of RMS COP and vertical loading observed in the quiet standing study. Figure 11 presents time-series COP data for a representative participant. Figure 12 presents the group RMS values across the different task conditions (varying hand support and balance challenges). In the presence of a greater challenge to stability (IC condition) it was anticipated there would be an increase in the COP excursion from the upper limbs compared to normal challenge (NC). Consistent with this idea there was an increase in upper limb COP excursion in the challenged condition in the M/L direction ($F = 17.81$, $p = 0.0018$) but not in the A/P direction ($F = 1.29$, $p = 0.28$, Figure 12).

The increased reliance on upper limb control was also predicted to be associated with a decrease in lower limb COP. This was supported by the smaller lower limb RMS COP with Hands ON in the A/P ($F = 222.61$, $p > 0.0001$) and M/L planes ($F = 160.79$, $p > 0.0001$). In Hands ON conditions, no differences were observed with increased challenge (NC-Hands ON versus IC-Hands ON, A/P: $F = 0.06$, $p = 0.81$, M/L: $F = 0.09$, $p = 0.77$, Figure 12). Confirming the increased challenge associated with the IC condition, an increase in lower limb COP in the IC-Hands OFF condition was observed (NC-Hands OFF versus IC-Hands OFF, A/P: $F = 13.76$, $p = 0.004$, M/L: $F = 58.23$, $p > 0.0001$, Figure 12).

In contrast to the differences in the COP measures, no significant task challenge differences in the mean levels of vertical upper limb loading (F_z) were observed (NC versus IC, $F = 0.44$, $p = 0.523$). A wide range of upper limb F_z, as a percentage of body weight (%BW), was observed (0.92-8.59% BW, Table 2).

To evaluate the temporal synchrony between the upper and lower limb COP responses, we assessed the proportion of participants who exhibited significant coherence between the upper and lower limbs. Figure 13 plots an example of coherence within a single participant in the IC condition (top), and the percentage of participants with significant coherence (i.e. values > 0.44) in the M/L and A/P planes (bottom). A majority of participants (9 of 11) displayed significant coherence in the full frequency band for both NC-Hands ON and IC-Hands ON conditions in the A/P plane. In the M/L plane, 3/11 and 9/11 of participants exhibited significant coherence across the full frequency band for the NC-Hands ON and IC-Hands ON conditions, respectively. Further examination revealed that the coherence in the M/L plane is attributed primarily to the low frequency content (< 0.625 Hz). Only 1 participant demonstrated coherence in the high frequency band (> 0.625 Hz).

3.2 Study 2: Perturbed standing

Figure 14 presents representative lower limb COP data from lateral perturbation trials in the Hands OFF (solid lines) and Hands ON (dashed) conditions from one participant, with the ensemble average superimposed (thick). Descriptive statistics for peak COP displacements, upper limb vertical forces and upper limb COP onset times (relative to lower limb onset) are presented in Table 3.

Similar to static standing, we predicted that increased challenge in perturbed standing would result in an increase in upper limb COP displacement. The results support this prediction as evidenced by a significant effect of the Hands ON in upper limb peak COP displacement ($F = 4.93$, $p = 0.043$, Table 3). The complementary prediction that availability of upper limb support would reduce lower limb responses was also supported. In Hands ON conditions, peak lower limb COP displacement was significantly smaller ($F = 29.03$, $p < 0.0001$). However, the prediction that increased challenge would not affect the lower limb contribution, indicated by an unchanged peak lower limb COP was not supported. A significant increase in peak lower limb COP displacement was observed in the IC conditions ($F = 22.76$, $p = 0.0001$).

The prediction that increased challenge would be associated with an increased vertical loading with the upper limbs was not supported. No significant effects were observed in either the upper limb peak vertical load ($F = 0.4$, $p = 0.56$, Table 3) or the mean loading levels recorded prior to the perturbation ($F = 1.29$, $p = 0.27$, Table 3). However, average preloading levels were higher in these dynamic perturbation studies compared to those collected during stationary standing tasks (perturbed vs stationary: 7.42 ±± 1.10 vs 2.74 ±± 0.49 %BW, $p < 0.0001$). There was concern that the increased upper limb COP could potentially arise from the increased vertical pre-perturbation loading applied during perturbation trials. To assess the impact of preloading, we assessed the effect of the Light Touch condition on COP displacement and perturbation-evoked peak vertical load. The desired effect of limiting pre-perturbation loading with the Light Touch condition was observed ($F = 31.05$, $p = 0.0001$, Table 3). There was a significant increase in perturbation-evoked peak lower limb COP displacement with the Light Touch condition ($F = 13.54$, $p = 0.0014$, Table 3). However, there was no significant effect of Light Touch on peak upper limb COP displacement ($F = 2.02$, $p = 0.18$) or peak upper limb vertical loading ($F = 0.72$, $p = 0.41$).

Temporal synchrony was examined by comparing the COP onset times of the upper and lower limb reactions. Across participants, the onset of lower limb COP movement occurred 146-249 ms after the perturbation with individual standard deviations ranging from 9 to 36 ms (n=40 trials). Given the consistency of the lower limb onset timing, we compared the upper limb times relative to the lower limbs across task conditions. Overall, there were no significant task differences in upper limb onset times relative to the lower limb onsets largely due to considerable variability (Table 3). To describe the source of variability, Figure 15 plots the number of participants demonstrating upper limb onset times (relative to lower limb COP movement onset) in three bins: 50ms prior to lower limb onset (< -50ms), within 50ms of the lower limb onset (-50 to 50ms), or 50 after lower limb onset (>50ms). In both Hands ON and IC-Hands ON conditions, the majority of participants displayed response times prior to -50 ms or after 50 ms. In contrast, in IC-Light Touch condition, a majority of participants displayed response times within 50 ms of the lower limbs response.

4 Discussion

Overall the study supported the hypothesis that availability of hand supports, in the form of a rollator, led to individuals mechanically relying on upper limbs to control both static and dynamic stability. In addition, the lower limb contributions to control appeared to be reduced as a possible reflection of the added contribution associated with the control of the upper limbs with potential benefits to stability control. While this may appear as a somewhat intuitive observation, it emphasizes the important role that upper limbs may play in stability control in the presence of hand support such as that provided by assistive devices. This reliance on upper limbs was not simply associated to passive events linked to vertical loading since these loads were relatively modest, even under challenged conditions, and the COP dynamics were characterized by amplitude and time lags reflective of active control. In addition, the analysis of temporal synchrony appeared to support the idea that the upper and lower limb control may share some common central regulation. Overall, we believe the study helps to redirect our attention to the capacity and complications of stability control using the upper limbs when they are made available. This focus is specifically relevant to understanding the control of stability when individuals are reliant on assistive devices.

While previous work has emphasized the haptic contributions of upper limbs to balance control [61, 71], such studies experimentally limited the potential mechanical force the upper limbs are permitted to contribute. In the present study participants were permitted to use hand support in any way they chose. In all cases, individuals generated significant loads and active control through the hands reflected by both upper limb vertical load and COP. The usefulness of the upper limbs in effecting support is likely due to the mechanical advantage provided by the placement of the hands as a product of the assistive device configuration. The influence of the effective mechanical advantage on the control strategies has potentially important implications to the design of hand supports. For example, increasing the effective mechanical advantage by placing the hands more anteriorly (relative to the feet) would likely result in larger upper limb involvement in the sagittal plane.

A consequence of the reliance on the upper limbs was that lower limb involvement was attenuated when hand support was available. Such attenuation may have occurred for two possible reasons: 1) the inclusion of hand support improved stability (smaller COM excursions) reducing the requirement of lower limb COP; and/or 2) the CNS reduced the reliance on the lower limb COP as a shift in strategy from the lower to upper limbs. An exception occurred in the perturbation task where we found that larger amplitude lateral perturbations resulted in significant increases in both upper and lower limbs. We attribute the greater lower limb COP responses as a reflection of the relative amplitude of perturbation. During pilot testing, 2/5 participants tested with the more challenging perturbation without hand support (not shown) required compensatory steps to prevent falling, indicating that the perturbation amplitude was close to or exceeded the limits for a feet-in-place strategy. Both upper and lower limbs were likely recruited for greater stabilizing torques than could be rapidly generated by either the upper or lower limbs in isolation.

One parameter of interest in the present study was the vertical loading though the upper limbs. Overall, we interpret greater amplitude of vertical loading as an index of the increased reliance on upper limbs. Vertical loading during stationary standing was relatively low, even during the more challenged stance conditions (2-3% BW) and increased in the dynamic task conditions (6-9% BW). However, the observed loading levels are smaller than the few studies that have reported loading levels on walking frames used for stability. Studies investigating compensatory stepping responses have instructed participants to simulate reliance on pick-up walkers by preloading with 20% BW [12, 14]. A case study of a progressive supranuclear palsy patient with balance impairments recorded loading of 30-35% BW on a pick-up walker during gait [40]. In contrast, the current study demonstrates a relatively small level of loading (< 10% BW) is needed to facilitate upper limb stability control in quiet and perturbed standing.

In the perturbation conditions there were two components of upper limb control that could influence responses: 1) (iso)tonic loading prior to onset of perturbation and 2) phasic reaction evoked by the perturbation. We found an important link between upper limb tonic loading (i.e., maintaining tension) and the evoked reactions in the current study. In the light touch condition, upper limb contribution was significantly reduced and lower limb responses returned to the magnitudes observed in the Hands OFF condition. This suggests that tonic loading, prior to the onset of perturbation, may facilitate active upper limb generation of stabilizing torques in response to perturbations. As a result, we interpret the higher vertical load prior to perturbation was specifically related to the size of the perturbation to generate appropriately amplified balance reactions.

The issue of shared, common control of upper and lower limbs was assessed by comparing the temporal coupling between the two sets of limbs. Overall, the findings support a shared CNS control for the upper and lower limbs for reactive balance corrections. In quiet standing, the hypothesized coherence between the upper and lower limbs was exhibited by the majority of participants. Interestingly, only a small proportion of participants exhibited coupled behaviour in the frontal plane for the IC task (Figure 13, C), the condition in which the observed upper limb contribution was largest. This finding could be attributed to a relative lack of signal recorded under the feet, but potentially indicates a shift from common control to a decoupled control strategy arising from peripheral sources within the upper limbs. In contrast to the quiet standing task, the perturbation results indicate temporal synchrony only when preloading was restricted. When preloading was permitted, only one participant demonstrated upper limb timing within 100 ms of lower limb onset. However, the majority of participants did exhibit temporal synchrony when preloading was not permitted (light touch condition). Elger et al. [38] reported synchronized upper and lower limb force onset times in response to lateral perturbations using conditions that also restricted preloading the handhold. We interpret these findings as an association between preloading levels and increased passive stiffness in the upper limbs. Stiffness arising from tonic muscle activity would result in passive COP responses to perturbation occurring earlier than active CNS control.

The current study lacked direct measurements of upper limb kinetics applied to the assistive device, using proxy measures collected at the level of (or close to) the ground, limiting the ability to interpret the nature of the interaction between the hands and the device. In particular, the relative contribution of transverse plane forces (i.e. shear forces orthogonal to vertical loading) to the upper limb COP signals cannot be distinguished from those produced by shifts in vertical loading. Furthermore, the lack of direct shear force measurements may underestimate the theoretical COP developed by the upper limbs (assessed in Appendix A). Further study in determining these contributions may be valuable in assistive device design. For example, characterizing frontal plane shear forces for stability could influence wheelbase width standards to minimize the risk for tipping.

The key finding that the upper limbs play a major role in effecting stabilizing moments in the frontal plane has potentially important clinical implications for populations who rely on assistive devices (e.g., elderly, stroke, traumatic brain injury). Balance control executed by the upper limbs may provide an important means to assess safe device use, training methods, and prescription criteria. Considering that preloading facilitates active upper limb responses, it may be an important indicator of an individual's reliance on assistive devices. Whether the findings from the current study from healthy controls can be generalized to regular assistive device users remains to be evaluated. Investigation of behaviour during dynamic activities (e.g., walking, turning) is also needed to deepen our understanding upper limb integration.

	Lower limb RMS COP (cm)		Upper limb RMS COP (cm)		Mean Upper limb F_z (%BW)	
Condition	ML	AP	ML	AP	Mean ±SE	Range
Hands OFF	0.23 ±0.03	0.43 ±0.04				
IC-Hands OFF	0.55$^+$ ±0.04	0.55$^+$ ±0.05				
Hands ON	0.10* ±0.02	0.10* ±0.02	0.27 ±0.03	0.50 ±0.08	2.51 ±0.52	1.21-7.12
IC-Hands ON	0.11* ±0.03	0.10* ±0.02	0.66$^+$ ±0.08	0.55 ±0.08	3.03 ±0.76	0.92-8.59

*Means significantly ($p<0.05$) different from Hands OFF. $^+$Increased Challenge (IC) means different from Normal challenge condition

Table 2: Experiment 1: Static standing. Summary of descriptive statistics (mean±SE) of RMS COP and vertical loading under the hands (F_z) (n=11).

Condition	Lower Limb Peak COP (cm)	Upper Limb Peak COP (cm)	Upper Limb Peak F_z (%BW)	Upper Limb Mean Preload (%BW)	Upper Limb Onset Time (ms)
Hands OFF	4.47* ±0.29				
Hands ON	3.03 ±0.13	8.61 ±2.03	4.96 ±7.89	6.55 ±1.33	-20.00 ± 34.36
IC-Hands ON	3.64^{+} ±0.20	13.34^{+} ±2.61	6.55 ±8.40	8.29 ±1.79	-42.50 ± 39.54
IC-Light Touch	$4.64^{+\wedge}$ ±0.22	10.56 ±3.52	4.41 ±8.08	$0.25^{\wedge}$ ±0.16	-6.67 ± 13.33

*Means significantly ($p<0.05$) different from Hands ON. $^{+}$Increased Challenge (IC) means different from Normal Challenge (NC) condition. $^{\wedge}$IC-Light Touch condition means different from IC-Hands ON

Table 3: Experiment 2: Perturbed standing. Descriptive statistics (mean±SE) of perturbed standing peak COP excursion for lower and upper limbs, peak upper limbs vertical loading, mean vertical preloading (prior to perturbation) under the hands, and reaction time onset relative to the lower limbs (n=8)

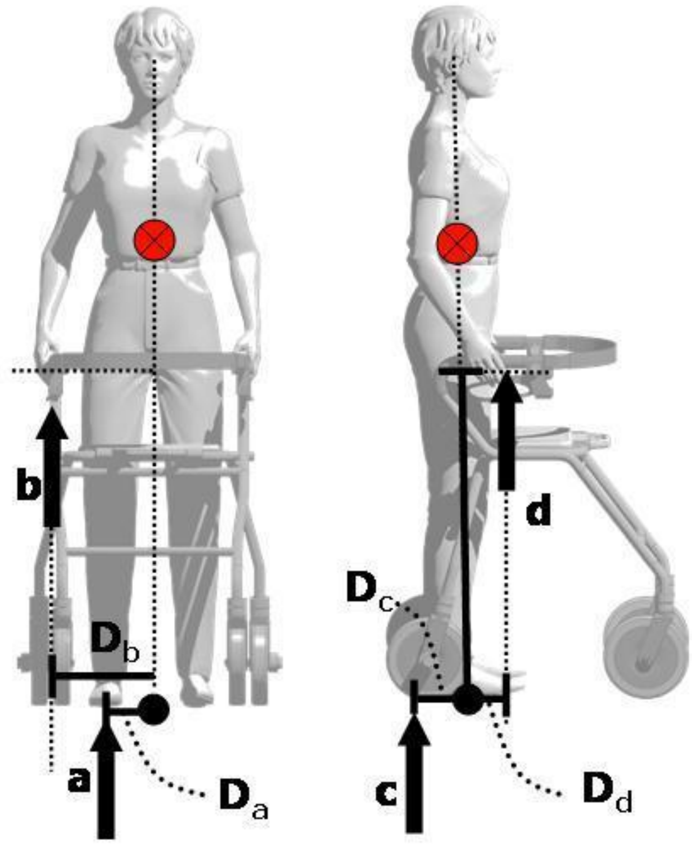

Figure 9: Diagram illustrating body position during standing with a rollator assistive device in frontal (left) and sagittal (right) planes. Lower-limb vertical forces (a, c) and corresponding moment arms (D_a,D_c) are shown with upper-limb vertical forces (b, d) and moment arms (D_b,D_d).

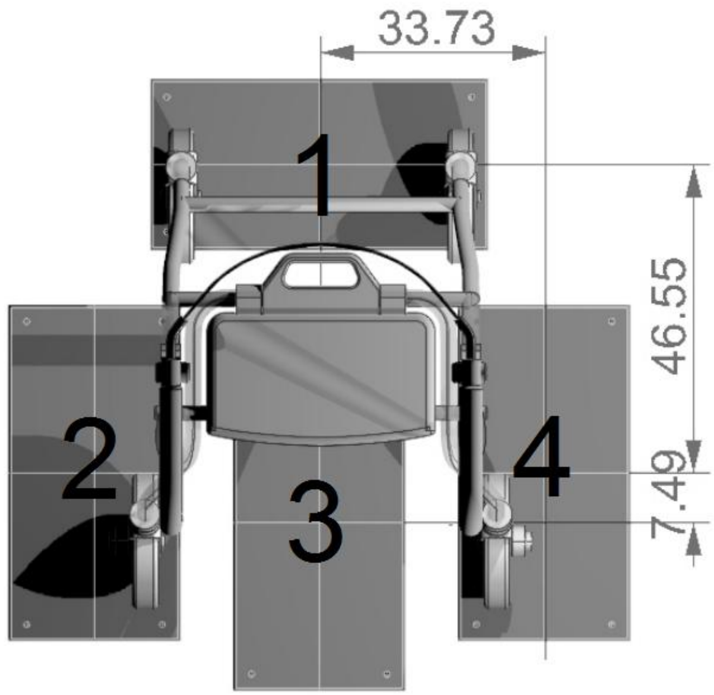

Figure 10: Four force plate setup with relative distances (in cm).

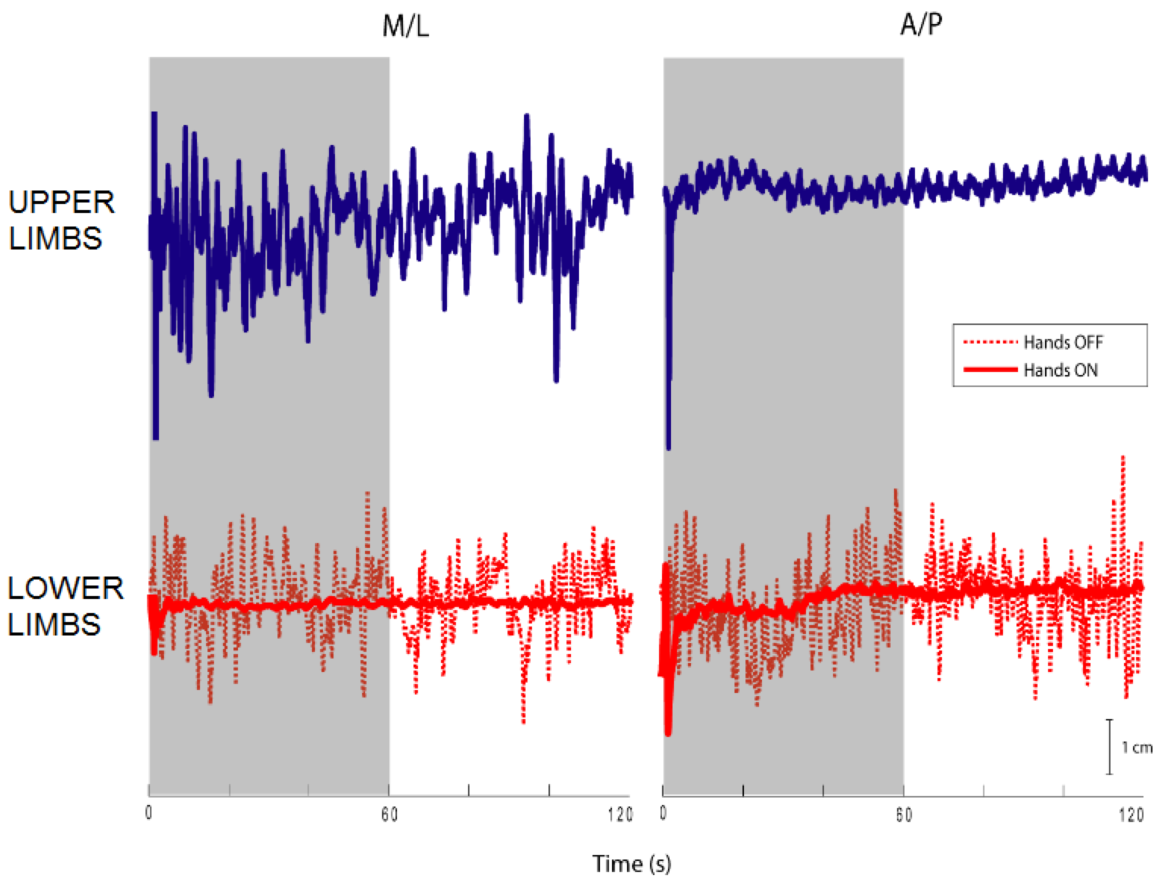

Figure 11: Experiment 1: Representative time-series of upper limbs (blue) and lower limbs (red) COP during quiet standing in M/L (left) and A/P (right) planes. Vertical axis scale is the identical for upper and lower limbs. Note the reduced amplitude of the lower limb COP from the Hands OFF (dotted line) to the Hands ON (solid line) conditions.

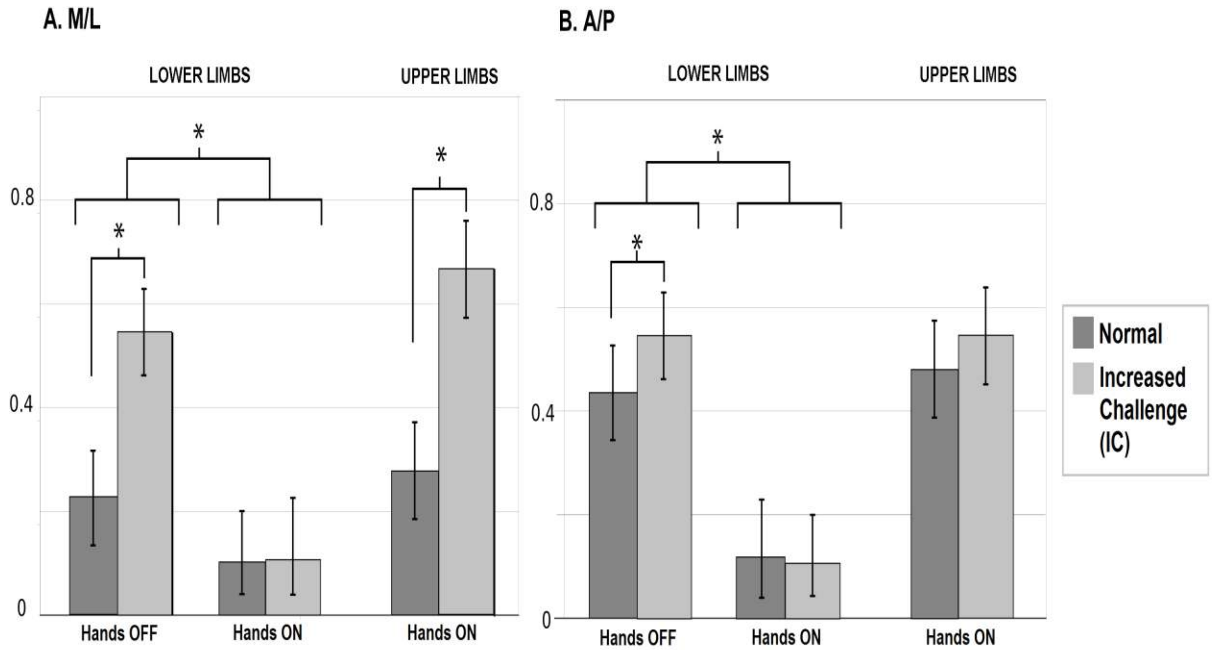

Figure 12: Experiment 1: Group mean RMS COP results in the M/L (A) and A/P (B) planes during quiet standing. Normal stance means indicated by the dark bars, and challenged stance (IC) is indicated by light bars. Asterisks (*) indicate statistically significant results ($p < 0.05$). Error bars indicate standard error.

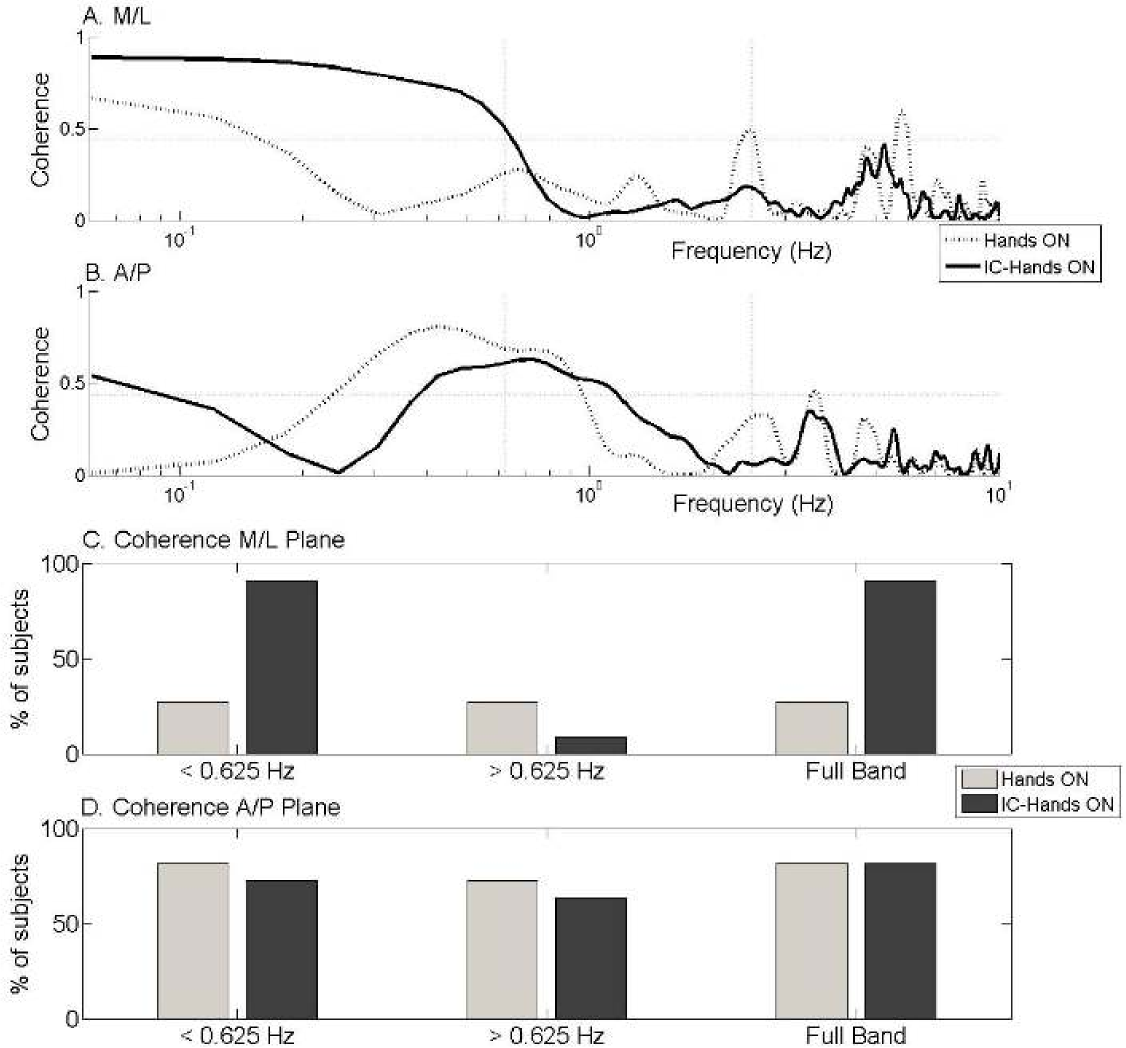

Figure 13: Experiment 1: Example coherence plot for a single participant in the M/L (A) and A/P (B) planes. Horizontal dotted line represents critical value for statistically significant coherence. Vertical dotted line indicates cutoff (0.625 Hz) for low and high frequency bands. Solid traces represent Hands ON trial. Bold traces indicate IC-Hands ON trial. Percentage of participants exhibiting significant coherence in the M/L (C) and A/P (D) planes. Light bars indicate Hands ON trials and dark bars represent IC-Hands ON trials.

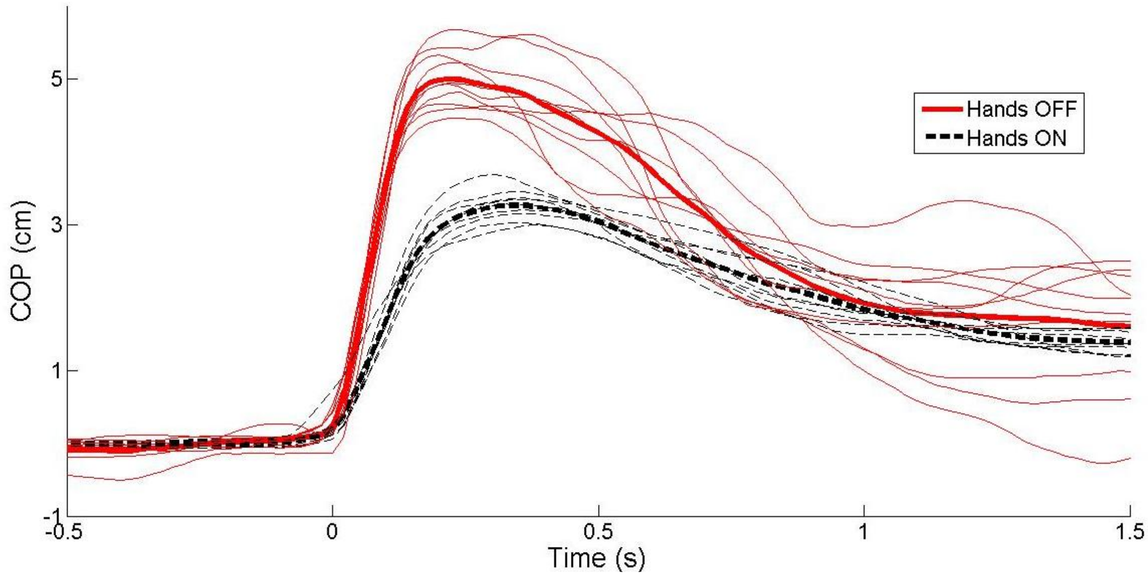

Figure 14: Experiment 2: Representative data of lower limb COP excursion for a single participant in M/L perturbed standing. Thin traces indicate single trials and thick traces represent ensemble averages for Hands OFF (solid) and Hands ON (dotted) conditions. Positive values represent COP shifts laterally (to left leg). Time zero is set to foot COP reaction onset.

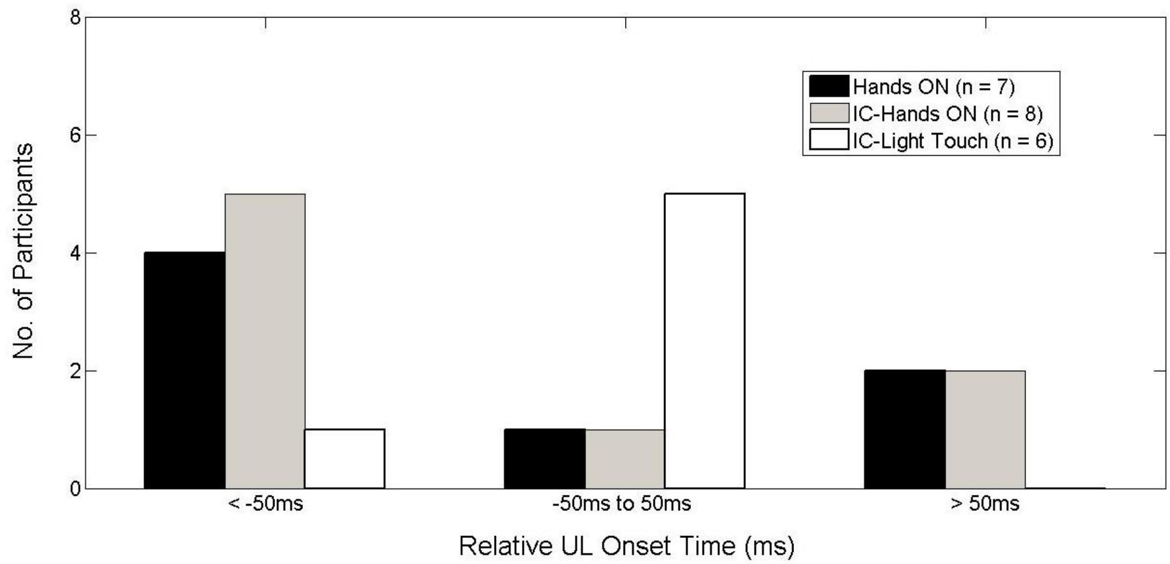

Figure 15: Experiment 2: Histogram of Upper Limb COP response onset (relative to Lower Limb onset) for all conditions with upper limbs available. Negative times indicate upper limb onsets preceding lower limb times.

Chapter 5
Upper Limb Involvement for Frontal Plane Balance Control in Rollator-Assisted Walking

1 Introduction

Mobility aids, such as canes, crutches and walkers, play an integral role in addressing mobility impairments for individuals with compromised cognitive and motor capabilities. An estimated 1.8 million adults use a walker in the United States [65], while nearly 1 in 10 seniors use walkers across Canada [111]. In particular, the rollator (or four-wheeled walker) is an increasingly popular aid prescribed to facilitate standing and walking activity. In the province of Ontario (Canada) alone, an estimated 10,000 new rollators are publicly subsidized each year [100] and many more are purchased privately without a prescription. Several studies have demonstrated the short term biomechanical and metabolic consequences of walking with rollators. Rollators can reduce knee and ankle joint loading [3] during steady-state gait compared to unassisted walking in healthy controls by transferring weight to the upper limbs to relieve the lower limbs of gravitational loads. For example, individuals with rheumatoid arthritis benefit from reduced pain associated with joint loading [32]. Patients with chronic obstructive pulmonary disorder (COPD) reported increased timed walking distances [125], improved walking efficiency [109] and faster walking velocities [125, 109, 48]. However, over 50% of the population who use assistive devices in Canada use them to prevent falls or to 'make me feel steady/balanced' indicating the potential importance of assistive devices in facilitating balance control [111].

Our group previously studied the upper limb control contribution to maintain balance in quiet and perturbed standing using a stationary rollator. The upper limbs were selected preferentially over the lower limbs to effect balancing moments for frontal plane control, likely due to the wider base of support afforded by the position of the hands [134, 136]. Other studies have demonstrated potential balance risks associated with walker use. Bateni et al. found that walker frames limit the available range of lateral stepping movements, restricting the capacity for compensatory stepping movements [12]. Finkel et al. demonstrated the potential risk of falling during transitions from sitting on a rollator seat to standing (and vice versa) due to slippage of the wheels [41]. However, there is little empirical data to establish how the upper limbs are used to maintain balance during walking. As the most common form of mobility, older adults reported falling during level walking (44%), compared to ascending or descending stairs (26%) and falling due to snowy/icy surface conditions (20%) [99].

Considering the inverted pendulum model applied to unassisted standing, stability is maintained by controlling the position and motion of the body's centre-of-mass (COM) within the base of support (BOS) defined by the placement of the feet. During locomotion, stability is challenged because the positions of the COM and BOS are constantly changing relative to one another according to the gait cycle. Indeed, the COM is only within the BOS during double stance, which comprises only 20% of the gait cycle [105]. However, the CNS is able estimate changes in COM and BOS from ongoing movements and produce anticipatory postural adjustments in a feedforward manner (e.g., gait initiation) [113]. Feedback control mechanisms, or reactive control, produce stabilizing behaviour (e.g., compensatory stepping) in response to unpredictable perturbations to stability [87]. Torques generated by the body to maintain stability comprise of two components: 1) the moment arm, defined as the distance between the point of application of ground reaction forces (center of pressure (COP)) and COM, and 2) and vertical load (F_z), with the net righting moment as a product of the vertical load and moment arm [76].

With rollators, balance control can be facilitated by permitting the upper limbs to be involved in sensing body position and generating stabilizing torques. In standing, the placement of the hands on the walking frame provides a wider effective BOS in the frontal plane compared to the feet. During gait, when the BOS afforded by foot placement varies within the gait cycle, the upper limbs have the advantage of maintaining the BOS width continuously. Previous studies investigating the kinetics applied to walking frames provide some indirect insight into how the upper limbs are involved. Alwan et al. studied the use of kinetics measured from the handles of an instrumented rollator to capture gait events, such as toe-off and heel strike [4]. A characteristic frontal plane moment oscillation linked to the gait cycle was found in healthy individuals (who do not use a mobility aid) suggesting upper limb behaviour associated with the gait cycle [4]. Fast et al. also reported similar, cyclical upper limb use in a patient with a lower limb fracture with a pickup walker [40]. In contrast, more random, asynchronous frontal plane loading pattern was observed in a patient using a walker primarily 'to enhance their stability and balance' [40].

The goals of the current study were to: 1) characterize the manner in which the upper limbs may be used for balance control during walking with a rollator, and 2) investigate the consequences of using the upper limbs for balance control on walking with a rollator performance. We hypothesized that use of the upper limbs for stability during walking would increase when balance is challenged or impaired. An increased challenge task was tested in young adults by imposing a greater demand on mediolateral balance control and comparing the task-related changes in upper limb use of the rollator. The comparison of balance impairment was conducted between two groups of older adults; one that relied on rollators for balance control and an elderly control group that walked independently without a mobility aid. Specifically, we predicted that increased balance challenge or impairment would demonstrate: 1) increased mean vertical loading (mean F_z) through the upper limbs, and 2) increased variation in the reactive loading of the rollator, indicated by an increased magnitude of high frequency COP oscillation generated by the upper limbs. We also predict the increased reliance on the upper limbs would be associated with improved walking competency. Specifically, individuals with the greatest reliance on upper limbs (greater mean F_z and upper limb COP variability) would show the greatest improvements in gait speed, step width and step width variability when using the rollator compared to non-rollator trials.

2 Methods

The contribution of the upper limbs to assist frontal plane balance control during steady-state walking was investigated in two different studies. The first was a test of the influence of challenge balance that was conducted in healthy, young subjects comparing: (a) normal conditions and (b) simulation of frontal plane balance impairment (narrow beam walking). The second was a test of the influence of impairment comparing: (a) older adults with clinically measurable balance impairments who regularly use rollators and (b) older adult controls who do not use mobility aids under normal conditions. In all cases the same assistive aid, an instrumented rollator (iWalker), was used to measure upper limb involvement. These methods were approved by the local research ethics committee.

2.1 Experiment 1: Influence of rollator on challenged balance walking versus normal walking in young adults

Participants and Tasks

Eleven (11) healthy, young adults (6 female, 5 male, 20-39 y, 56.7-83.9 kg) were recruited to participate in the experiment. In the normal walking (NW) task, participants walked across a 6m long level walkway at their preferred gait speed. To challenge mediolateral balance subjects performed a beam walking (BW) task. In this task participants walked along a 5 cm wide wooden beam (5 cm high x 6 m long) with the instruction to 'walk across the beam at your preferred speed without stepping off'. Participants walked under two conditions: 1) with a rollator assistive device (ROL) and 2) unaided (NOROL) for both tasks. Six (6) blocks of 4 trials, randomized by condition, were performed for a total of 24 walking trials. The rollator handles were adjusted by the height of the beam in BW trials.

Measures

Gait speed (VEL) was measured by determining the time elapsed in the middle 5m of the walk using optical beam switches. The number of foot contacts to the floor to recover balance in the beam walking condition (nMISS) and cadence were determined by observation from a video record of each trial. Upper limb kinetics were recorded using a custom-built instrumented 4-wheeled walker (iWalker), which has been previously described in detail [20]. The iWalker includes 4 single-axis load cells mounted vertically into each leg of the walking frame to measure the vertical loading (F_z) and COP generated by the upper limbs through the rollator frame. According to general handle height adjustment guidelines [107], the handles were adjusted to the height of the radial styloid with arms hanging straight for each participant.

Analysis

All signals were sampled at 50Hz by an on-board analog-to-digital converter and stored on an external personal digital assistant. Load cell signals were converted to force units, low-pass filtered (2nd order Butterworth; cutoff = 10 Hz), and resolved into a single mediolateral estimate, $COP_{(raw)}$, using the following relation:

$$COP_{(raw)} = \frac{(F_{FrLeft} - F_{ReLeft}) + (F_{FrRight} - F_{ReRight})}{F_{FrLeft} + F_{FrRight} + F_{ReLeft} + F_{ReRight}} \quad (1)$$

where F_{FrLeft}, $F_{FrRight}$, F_{ReLeft}, and $F_{ReRight}$, represent vertical forces from the four load cells and D_{front} and D_{rear} are distances of the load cells to the mediolateral midline of the walker (22.3 cm and 26.6 cm, respectively). The total vertical load (F_z) was calculated as the sum of the four load cell outputs. Note that this approach does not fully consider the effects of shear, or transverse, forces on the theoretical COP (see Appendix A for an assessment).

While in standing studies one can rely on overall COP as an index of balance control, the COP during walking is more complex. There is a significant source of variability that is linked to the oscillations of the COM associated to the cadence of walking. In order to disentangle the reactive balance control component, we presently separated slower cadence-related COP displacement from the faster reactive balance control responses in the $COP_{(raw)}$ signal. Two frequency sub-bands were determined: 1) a low-frequency band (0.5-1.25 Hz) encompassing the observed range of cadence values (85.8-119.8 steps/min, or 0.71-1.00 Hz), and 2) a high frequency band (1.25-5 Hz) reflecting upper limb response rise times to unpredictable perturbations to standing [10]. The observed transient rise times (one quarter of a full cycle) in response to perturbations ranged between 50 and 200 ms, corresponding to a frequency range of 1.25-5 Hz. To compute the upper limb kinetics measures, $COP_{(raw)}$ and F_z records from each trial were compiled into a single time series for each condition. To compute the desired high-band COP measure, the compiled $COP_{(raw)}$ time series was bandpass filtered (10th order Butterworth) using the high band (1.25-5 Hz) as cut-off frequencies and COP RMS values were computed from the filtered output. For the remainder of the chapter, COP is defined as the high-band component of $COP_{(raw)}$. The mean upper limb vertical loading was calculated from the full F_z record.

To test the hypotheses predicting increased upper limb use in the presence of an increased balance challenge, paired t-tests were used to assess the effect of Task (NW/BW) on the two primary dependent measures (RMS COP and F_z). To test the impact of upper limb use on walking performance, a 2-way ANOVA was conducted to analyze the effects of Walker (ROL/NOROL) and Task (NW/BW) on gait speed and number of missteps (nMISS).

2.2 Experiment 2: Upper limb use among chronic older adult rollator users and older adult controls

Participants and Tasks

In the second experiment, 10 older adults who regularly used a rollator (RU) were recruited from a local retirement residence to participate. Ten (10) community-dwelling older adults (CTL) who did not use an assistive device were recruited as comparison group. Inclusion criteria for the RU group were a need of their assistive device to walk independently and perform daily mobility activities (e.g., sitting, standing, going through doors), and ability to follow two-step commands, assessed by the residence kinesiology staff. Potential participants were excluded if they were in palliative care, had uncorrected vision, experienced significant pain in standing or moving for brief periods, had diagnosed pathology which severely affect physical function within the past 6 months (principally stroke, knee/hip arthroplasty, cardiac disease), or had been identified by the long term care home staff as a frequent faller. This latter group, defined as having fallen more than twice in previous months, was excluded for safety reasons.

Testing was conducted at the retirement residence. All participants walked with (ROL) and without (NOROL) the iWalker over a 4 m pressure-sensitive mat (GaitRite) to record spatiotemporal measures of foot placement. Two (2) trials of each condition were performed for most participants, with additional walks performed to ensure at least 15 steps per condition. The handles of the iWalker were adjusted to the height of their regular device for RU participants. Standard handle heights, adjusted to the height of the radial styloid (with arms hanging straight) [107], were used for CTL participants. Spotters walked to the side of the participant during all trials to minimize chance of falling to the ground.

Measures

COP and F_z measures collected from the iWalker were acquired and processed in the same manner as described in Experiment 1. Spatiotemporal measures collected from the mat were gait speed (VEL), cadence (CAD), step length (SL), and step width (SW). Step width variability (SWV) was computed as the standard deviation of step width. A 2-way ANOVA was performed to assess the effects of the device (factor: device [NOROL/ROL]) and group (factor: group [RU/CTL]) on spatiotemporal walking parameters (VEL, CAD, SL, SW, SWV). Differences in the characteristics of gait were determined by calculating difference between no rollator and rollator trials (e.g., $dVEL = VEL_{unaided} - VEL_{rollator}$). A clinical measure of balance, Berg Balance Score (BBS) [20], was also used to measure balance capabilities of each individual.

Analysis

Correlations were performed to assess potential relationships between iWalker measures and gait performance characteristics to assess whether covariates should be included in analyses. VEL was significantly correlated to upper limb measures COP ($r = 0.4429$, $p = 0.051$) and mean F_z ($r = 0.5642$, $p = 0.011$). Hence, VEL was used as a covariate in all subsequent analyses involving upper limb measures.

To test the first hypothesis that upper limb use is higher in the balance-impaired group (RU) compared to controls (CTL), one-way ANOVAs (factor: group [RU/CTL]) were performed on iWalker measures (COP and F_z). To test the second hypothesis that increased involvement of the upper limbs for balance would be associated with improved walking parameters; separate one-way ANOVAs were conducted using COP and mean F_z as the main factors on changes in stepping parameters related to using a rollator (dVEL, dSW, dSWV).

The potential effect of arthritis affecting the fingers and/or wrists on upper limb use behaviour was assessed in a post-hoc analysis. In this analysis, the participants reporting arthritis in the upper limbs (4 from RU group; 2 in CTL) were excluded and a one-way ANOVA (factor: group [RU/CTL]; independent variables COP and F_z) was conducted on the remaining subset of participants.

3 Results

3.1 Experiment 1: Influence of rollator on challenged balance vs. normal walking in young adults

Table 4 summarizes the descriptive statistics (mean ±± SE) of gait speed, missteps, RMS COP, and vertical loading (F_z) observed in Experiment 1. Figure 16 provides a 10s sample of frontal plane upper limb $COP_{(raw)}$ (top) and F_z (bottom) time series data for a single subject using the rollator for level ground (solid trace) and beam walking trials (dashed). Figure 17 provides the corresponding frequency spectra for the $COP_{(raw)}$ signal (top), and high-pass filtered COP signal (bottom).

As predicted by the first hypothesis, an increase in RMS COP in the frontal plane ($p < 0.001$) and increase in mean vertical load ($p < 0.001$) was observed when comparing the beam walking (balance challenge, BW) condition to level ground walking (NW). Note there were also increases in RMS magnitude of the raw, unfiltered $COP_{(raw)}$ (p=0.001) and low-frequency, cadence-related $COP_{(low)}$ (p=0.002). The tasks were distinguished by a significant decrease in gait speed associated with the BW condition (p<0.001) and the effect of the assistive device on reducing gait speed was close to significance (p=0.089). However there was a statistically significant interaction, specifically that the task-related reduction in gait speed was significantly smaller when using the assistive device (p<0.001). Also, the stabilizing benefit of the rollator was observed in the number of missteps in the BW condition: missteps occurred in only 0.4% of trials with the assistive device (ROL) compared to 27.3% of trials without the rollator (NOROL).

3.2 Experiment 2: Upper limb use among balance impaired older adults and older adult controls

Group differences in clinical profile and gait without rollator

The sex, age, Berg balance scores, and self-reported medical history of the participants studied in Experiment 2 are presented in Table 5. The rollator user group was significantly older (p = 0.001) and had significantly poorer balance, indicated by lower Berg balance scores ($p < 0.001$). Note that one control participant had a Berg score that fell within the range of the RU group but did not use a rollator (or other devices) for independent mobility. Table 6 summarizes the descriptive statistics (mean ±± SE) of gait speed, spatiotemporal walking parameters, upper limb COP, and mean vertical loading observed. Consistent with the age and degree of balance impairment, the RU group exhibited gait performance (without a rollator) that was slower ($p < 0.001$) with shorter strides ($p < 0.001$), and lower cadence (p = 0.007) when compared to the CTL group. No significant group difference in mean step width (p = 0.810), or step width variability (p = 0.244) was observed.

Group differences during walking with rollator

The first hypothesis, which predicted an increased upper limb dependency in the balance-impaired group (RU) was not supported by the results. Contrary to our prediction, upper limb COP (p=0.299), and mean F_z (p=0.590) were similar in both the balance impaired (RU) and older adult (CTL) groups (Table 6). In the post-hoc analysis excluding participants reporting arthritis in the upper limbs, the subset of RU (n=6) exhibited significantly more impaired balance than the subset of CTL participants (n=8) ($p < 0.001$). Comparing the groups without upper limb arthritis, no significant differences were observed in upper limb COP ($p =0.370$) and mean F_z ($p = 0.565$). However, a significant group difference was observed in the cadence-related low frequency component $COP_{(low)}$ ($p=0.041$).

Mean frontal plane spatiotemporal stepping parameters with (ROL) and without (NOROL) the assistive device are presented by group in Figure 18. Use of the rollator resulted in significantly reduced mean step width ($p = 0.018$) and step width variability ($p < 0.001$). As shown in Figure 19, no statistically significantly effects associated with using the rollator for step length ($p = 0.338$) or gait speed ($p = 0.945$) were observed. A significant decrease in cadence ($p = 0.048$) associated with using the device was found. The overall range of cadence values was 76.0-129.6 steps/min corresponded to a frequency range of 0.63-1.08 Hz, which is within the 0.5-1.25 Hz frequency band used to filter COP associated with cadence

Effect of upper limb use on walking parameters

The second hypothesis predicting that greater dependency on the upper limbs, indicated by COP and mean F_z, would be associated with greater change in walking competency, indicated by change in stepping parameters with a rollator, was partially supported. Table 8 provides the resulting p values of the linear models using upper limb usage measures (COP and mean F_z) to predict change in gait parameters (dSW, dSWV, dVEL). In support of our hypothesis, higher amplitudes of upper limb COP (adjusted for gait speed) were significantly related to a greater reduction in step width variability (dSWV) ($p = 0.023$; Figure 8, left panel). However, no significant associations between mean F_z and change in stepping parameters were observed (Table 8). Note that increased unfiltered ($COP_{(raw)}$) and cadence-related band (COP(low)) was significantly related with reduced mean step width ($p = 0.018$ and $p = 0.014$, respectively). Increased amplitudes of $COP_{(low)}$ were also significantly related to increased walking speeds ($p = 0.011$).

4 Discussion

The goals of the current study were to: 1) characterize the upper limb involvement in maintaining balance during walking with a rollator, and 2) examine the consequences of using the upper limbs for balance control on walking. The findings of the current study supported the hypotheses that the upper limbs play a significant role in effecting frontal plane balance control during rollator-assisted walking. Increased contribution from the upper limbs in maintaining stability corresponded with reduced lower limb involvement, indicating that the upper limbs can compensate for lower limb limitations. While these findings are not surprising, this study deepens our understanding of assistive device use by providing a basis for assessing the capacity for upper limb involvement when they are made available through assistive devices.

One of the main objectives of the study was to examine the relationship between upper limb measures, COP and mean F_z, and frontal plane stability during overground walking. Experiment 1 provides the strongest evidence of this relationship by comparing upper limb use under unchallenged and balance challenged (beam) walking conditions. Beam-walking has been used previously to impose a frontal plane challenge to stability, with narrower beams eliciting larger head displacements and increasing frequency of handrail use [21]. In the current study, we observed a remarkably smaller rate of missteps in beam-walking trials with a rollator compared to without (0.4% vs 27% of trials), demonstrating the use of the upper limbs for improved stability. Corresponding to the improved stability, amplitudes of upper limb COP excursion increased significantly. This association between increased upper limb COP excursion and increased balance challenge conditions was also found in quiet and laterally-perturbed standing [9,10]. Hence, we consider Experiment 1 to provide further evidence that COP excursion as measured through a rollator frame is reflective of upper limb involvement in maintaining stability.

Similarly, we consider increased mean F_z as an index of reliance on the upper limbs for stability. Overall, increases in mean vertical loading increased with task-related balance difficulty in healthy controls. In comparison to other tasks, the lowest levels of mean F_z were reported during quiet standing (2-3% BW) [9] and rose to higher levels during perturbed standing (6-9% BW) [10]. The results from Experiment 1 continue the upwards progression from 9.1% BW in level ground walking to 13.7% BW under the challenged balance condition. In our previous study examining perturbed standing, higher loading levels facilitated active upper limb generation of stabilizing torques in response to standing perturbations [10]. Increased mean vertical loading may also be associated with increased passive upper limb stiffness, which has the advantage of providing stabilizing responses at faster latencies than active CNS control [10]. Hence, increased vertical loading can be interpreted as an indicator of increased reliance on upper limb stabilizing torque-generation capabilities. However, vertical loading may indicate reliance on the upper limbs for purposes other than for stability reasons. A study examining the effect of rollators on dyspnea (shortness of breath) in COPD patients found low levels of vertical loading (7% BW). While loading levels have yet to be collected, arthritis patients have reported reduced lower limb pain while walking with rollators due to a shift in weight-bearing to the upper limbs [22]. Hence, increased vertical loading may reflect upper limb reliance to compensate for multiple impairments.

The consequences of upper limb use to assist in maintaining walking stability on lower limb behaviour were also examined. In quiet and perturbed standing, a consequence of using the upper limbs for support was attenuation of lower limb contributions [9, 10]. This reciprocal relationship was observed in Experiment 2 of the current study, associating individuals exhibiting high upper limb COP to the greatest reduction in step width variability. Step width variability (SWV) measures the use of mediolateral foot placement strategy to reactively stabilize against unpredictable perturbations [21, 23]. Hence, the correlation between the increased COP and reduced step width variability suggests that COP indicates reactive stability control effected by the upper limbs through the rollator. There was also a significant association between increased amplitudes of cadence-related $COP_{(low)}$ (0.5-1.25 Hz) and reductions in mean step width. Increasing step width extends the lateral borders of the BOS, which facilitates larger amplitude postural adjustments effected by the lower limbs [20]. This finding suggests that the low frequency band ($COP_{(low)}$) potentially measures upper limb involvement for stability arising from self-initiated perturbations associated with the gait cycle.

Having confirmed the link between upper limb measures and frontal plane control, we evaluated whether a balance-impaired older adult group were more reliant on the upper limbs for stability. The prediction that balance-impaired older adults who use rollators in everyday life (RU) would be distinguished from a group of older adults who were not routine rollator users (CTL) by greater upper limb involvement was not supported. The lack of group differences could be attributable to large differences in gait speed, and/or co-morbidities. The control group walked at nearly double the average rollator user gait speed (1.05 vs 0.55 m/s) and speed was found to be related to both upper limb COP ($r = 0.4429$, $p = 0.051$) and mean F_z ($r = 0.5642$, $p = 0.011$).

While gait speed was used as a covariate in the group analysis, the magnitude of the speed difference renders group comparisons difficult to interpret.

Although the participants were recruited on the basis of balance impairment, age and co-morbidities may have contributed to the large upper limb COP variability in the RU group (SE = 0.22 cm) compared to the older adult controls (SE = 0.10 cm) and beam-walking young controls (SE = 0.07 cm). Aging impacts upper limb muscle strength, reduced vibrotactile sensitivity, and slowed movement speeds [145]. In the current study, the lower levels of upper limb F_z observed in the significantly older RU group (89.1 vs 83.1 y) may reflect an aging effect related to muscle weakness. In the post-hoc analysis excluding participants reporting arthritis in the fingers or hands, the subset of balance-impaired participants (n=6) demonstrated significantly larger amplitudes in the cadence-related low frequency band $COP_{(low)}$. This finding indicates that co-morbidities influence assistive device loading strategies and suggests that the RU group may adopt a slower (i.e., low frequency) upper limb strategy to compensate for lower limb balance impairments during walking.

An alternative interpretation of the lack of differences is the groups possess similar frontal plane balance capabilities. While the upper limb COP measures demonstrated sensitivity to frontal plane stability, the group balance capabilities were assessed using a global balance instrument (i.e. Berg Balance Scale) that may not be specific to frontal plane balance control. Furthermore, straight line overground walking may not be sufficiently challenging to elicit upper limb usage for frontal plane stability. Examining reliance on the upper limbs in more challenging tasks could provide stronger insight into the need for rollators for stability. For example, turning requires dynamic mediolateral balance control [23], may be an appropriate task to examine the potential reliance on upper limbs for frontal plane stability. Overall, this study provides evidence to demonstrate the importance of the upper limbs in maintaining frontal plane balance control in rollator-assisted walking. The findings from this work will impact guidelines for prescription, fitting, and specification of future assistive devices. While it is clear that stability control can be effected by the upper limbs to compensate for lower limb dysfunction, the need to assess upper limb functioning should be emphasized. For example, hemiparetic stroke survivors may require minimum hand function levels for effective rollator use. Further examination of the relationship between upper limb function (e.g., strength, range of motion, dexterity) and balance control would help provide such guidelines. One of the limitations of the study is the lack of information regarding the temporal coordination between upper limb measures and the gait cycle. Future studies examining the timing between the upper and lower limbs could inform about the upper limb contributions to effect reactive and anticipatory stabilizing moments.

	No rollator (NOROL)		With rollator (ROL)		p	
	Level Ground	Beam Walking	Level Ground	Beam Walking	Task NW/BW	Device NOROL/ROL
Gait speed (m/s)	1.30±0.06	0.88±0.07	1.22±0.06	1.01±0.06	<0.001*#	0.089#
Mean F_z (%BW)			9.12±2.38	13.70±1.99	<0.001*	
COP (cm)			1.07±0.06	1.35±0.07	<0.001*	
$COP_{(raw)}$ (cm)			2.14±0.12	2.93±0.15	0.001*	
$COP_{(low)}$ (cm)			0.92±0.06	1.39±0.09	0.002*	

Table 4: Experiment 1: Normal versus Beam walking statistics summary. Gait speed and COP values are reported as mean±SE. *indicates significant difference (p<0.05). # indicates significant task x device interaction (p<0.05)

Rollator Users (RU)				Older Adult Controls (CTL)			
Sex	Age	Berg	History	Sex	Age	Berg	History
M	92	41	ST	M	82	55	
F	86	45	ART↑, OP	M	82	54	
F	85	49	CD, DIA	M	80	56	CD
M	86	44	VIS, KA	F	80	52	OP
F	86	38	ST, BT, ART↓	M	87	56	
F	87	38	VIS, ART↑	M	82	52	ART↓, VER
M	88	42	ART↑, ART↓, ST, CD	F	80	46	ART↑, ART↓, CD, OP, HA, TBI
F	97	38	VIS, HA, OP	M	89	39	ART↑, ART↓, HA, CD, DIA
F	93	42	ART↑, KA	F	86	52	
F	91	42	OP	F	82	56	
4M, 6F	89.1* ±1.3	41.9* ±1.1		6M, 4F	83.1* ±1.1	51.4* ±1.9	

Legend ST=stroke, ART↑=Arthritis (upper limbs), ART↓=arthritis (lower limbs), OP=osteoporosis, CD=cardiac disease, VIS=low vision, KA=knee arthroplasty, HA=hip arthroplasty, VER=vertigo, DIA=diabetes mellitus, BT=brain tumour

Table 5: Experiment 2: Participant sex, age, Berg balance scores, and history; * indicates significant group differences ($p \leq 0.05$)

	Full group			Excluding Upper Limb Arthritis		
	RU (n=10)	CTL (n=10)	p	RU (n=6)	CTL (n=8)	p
BBS	41.9±1.1	51.4±1.9	<0.001*	42.0±1.7	53.4±1.2	<0.001
Mean F_z	12.40±1.46	16.44±2.12	0.640^+	12.8±2.4	16.7±2.5	0.565^+
COP	0.92±0.22	1.10±0.10	0.299^+	1.10±0.32	1.04±0.12	0.370^+
$COP_{(raw)}$	2.42±0.39	2.06±0.16	0.400	2.66±0.58	0.83±0.11	0.041*

$^+$ with velocity as covariate

Table 6: Experiment 2: Upper limb usage group comparison statistics. Means±SE and p values from ANOVA for group effect (CTL/RU). * indicates significant effect (p ≤ 0.05)

	Without rollator		With Rollator		p	
	RU	CTL	RU	CTL	Group	Device
Step Width (cm)	12.9±0.9	12.2±1.3	9.9±0.8	10.2±0.9	0.810	0.018*
Step Width Variability (cm)	2.0±0.3	2.4±0.3	1.3±0.1	1.3±0.1	0.244	<0.001*
Step Length (cm)	32.9±2.5	57.5±3.9	39.6±2.8	56.7±2.6	<0.001*	0.338
Cadence (steps/min)	98.8±5.2	108.3±3.6	89.2±3.8	102.0±3.9	0.007*	0.048*
Gait speed (cm/s)	55.1±5.2	104.8±8.6	60.8±5.3	98.1±7.2	<0.001*	0.945

Table 7: Experiment 2: RU and CTL spatiotemporal gait statistics summary. Means±SE and p values from ANOVA for group (CTL/RU) and device (NOROL/ROL) are tabulated. * indicates significant group effect (p ≤ 0.05)

	dSW	dSWV	dVEL
Mean F_z^+	0.199	0.901	0.903
COP^+	0.928	0.023*	0.938
$COP_{(raw)}$	0.018*	0.454	0.160
$COP_{(low)}$	0.014*	0.724	0.011*

Table 8: Experiment 2: p values of linear regression results associating changes in gait parameters and walker usage measures; * indicates significant result ($p \leq 0.05$)

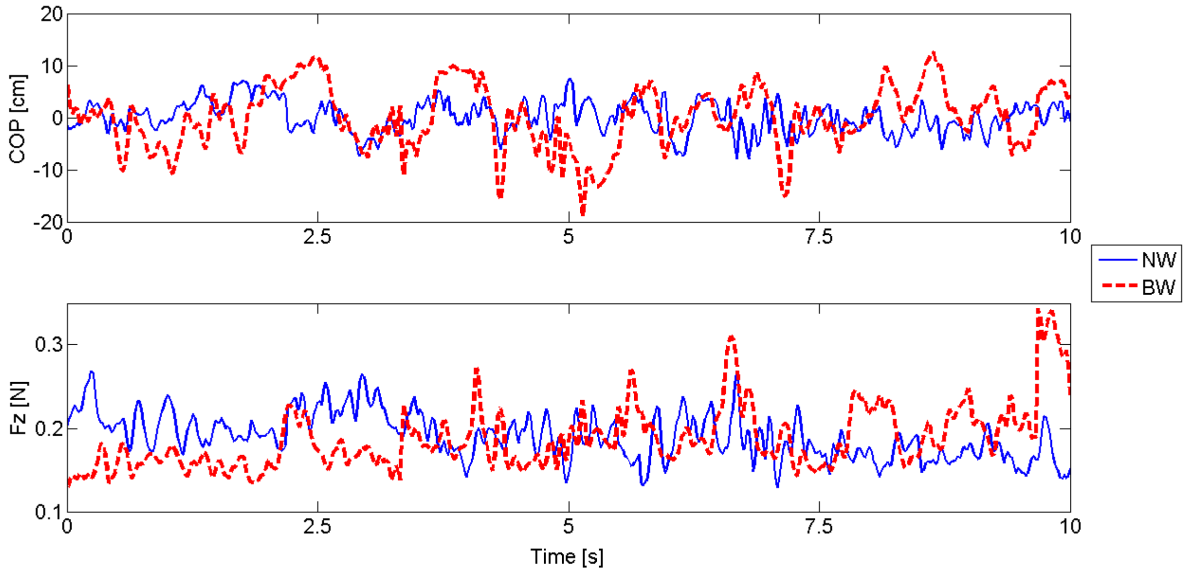

Figure 16: Experiment 1: Sample 10s time series of unfiltered mediolateral upper limb $COP_{(raw)}$ (upper plot) and mean vertical loading, F_z (lower) from a representative young, healthy participant. Normal walking (NW) condition is plotted (continuous blue trace) against balance challenged (beam) walking (BW) condition (dashed red).

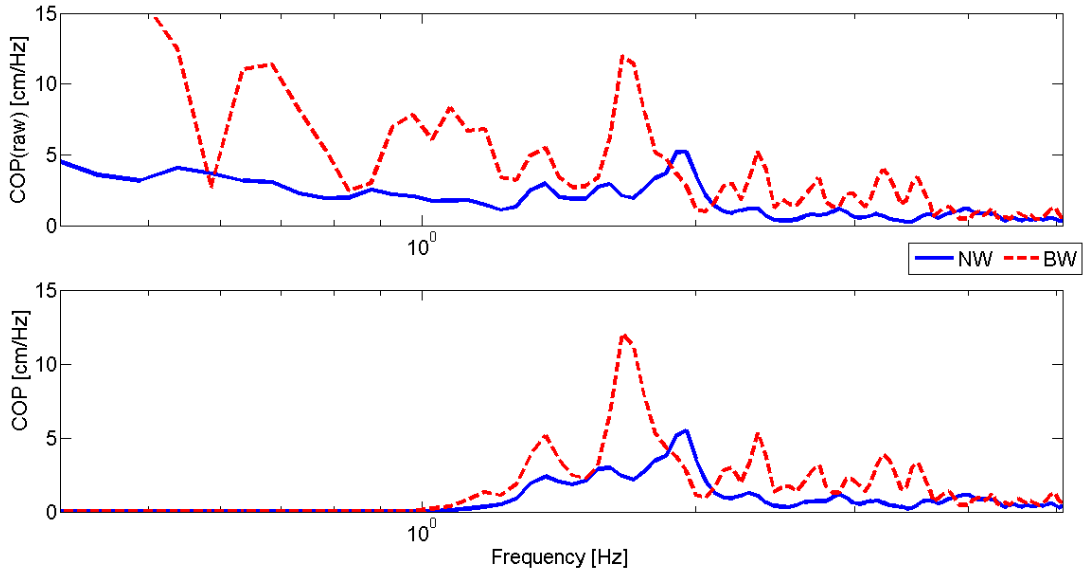

Figure 17: Experiment 1: Sample frequency spectrum plots from a representative subject. Spectral estimates from the raw signal (COP$_{(raw)}$; 0-10 Hz) and high frequency components (COP; 1.25-10 Hz) are plotted in the upper and lower plots, respectively. Normal walking (NW) condition is plotted (continuous blue trace) against balance challenged (beam) walking (BW) condition (dashed red).

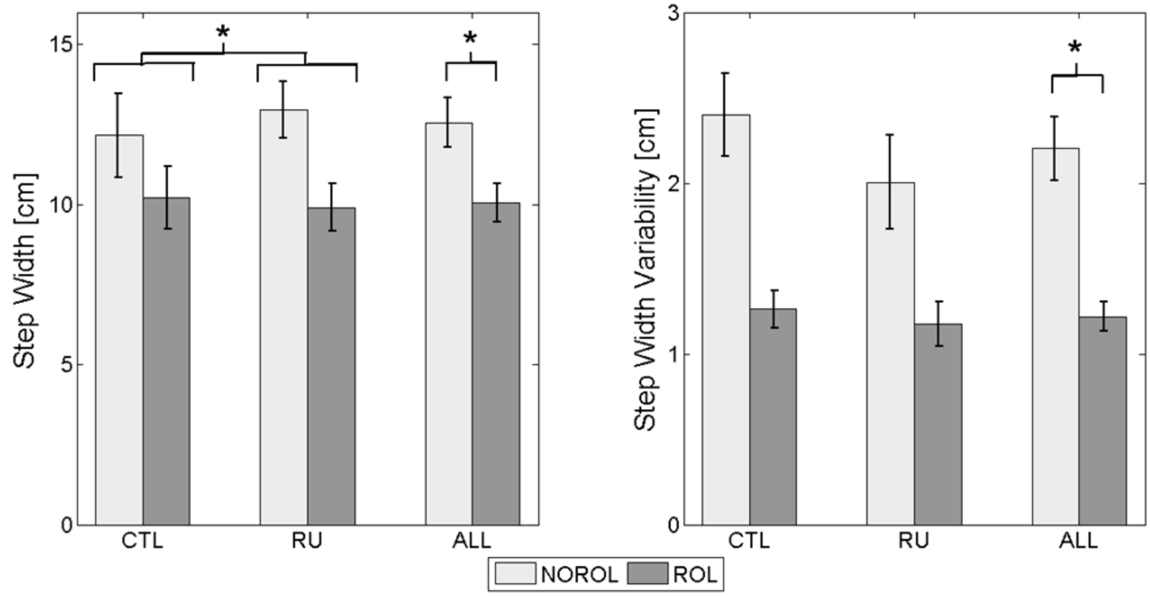

Figure 18: Experiment 2: Mean step width (SW; left plot) and step width variability (SWV; right plot) gait parameters with (ROL; dark bars) and without (NOROL; light) rollator assistance. Rollator users (RU), older adult controls (CTL), and the pooled (ALL) means are plotted. * indicates statistically significant differences ($p < 0.05$). Error bars indicate standard errors.

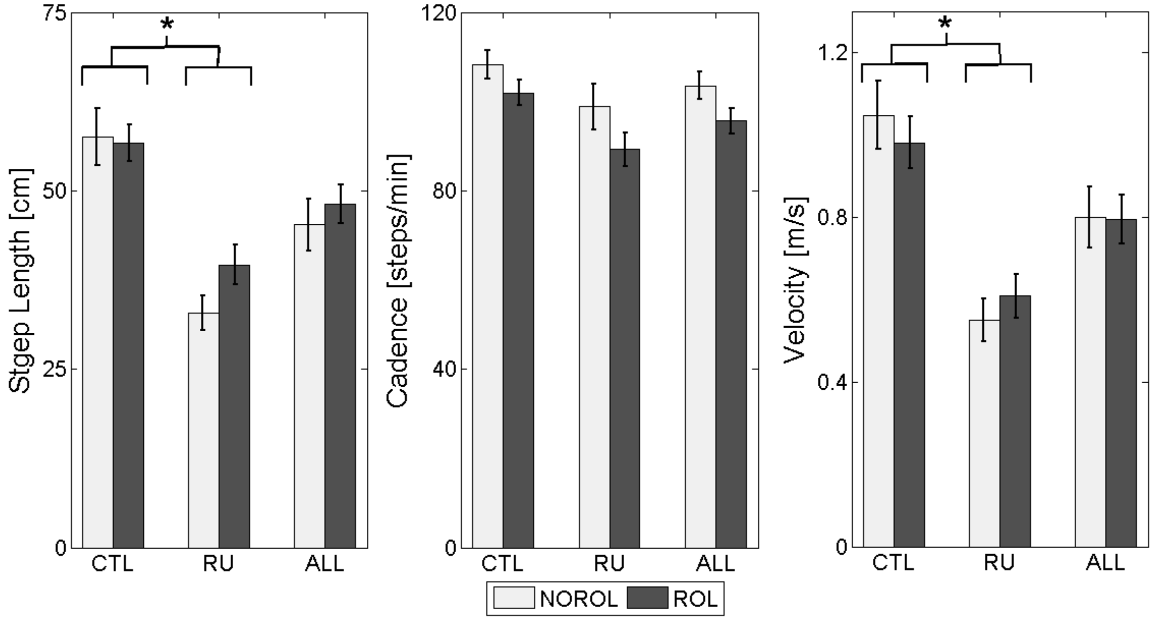

Figure 19: Experiment 2: Step length (SL; left plot), cadence (CAD; middle), and gait speed (VEL; right plot) gait parameters with (ROL; dark bars) and without (NOROL; light) rollator assistance. Rollator users (RU), older adult controls (CTL), and the pooled (ALL) means are plotted. * indicates statistically significant differences (p < 0.05). Error bars indicate standard errors.

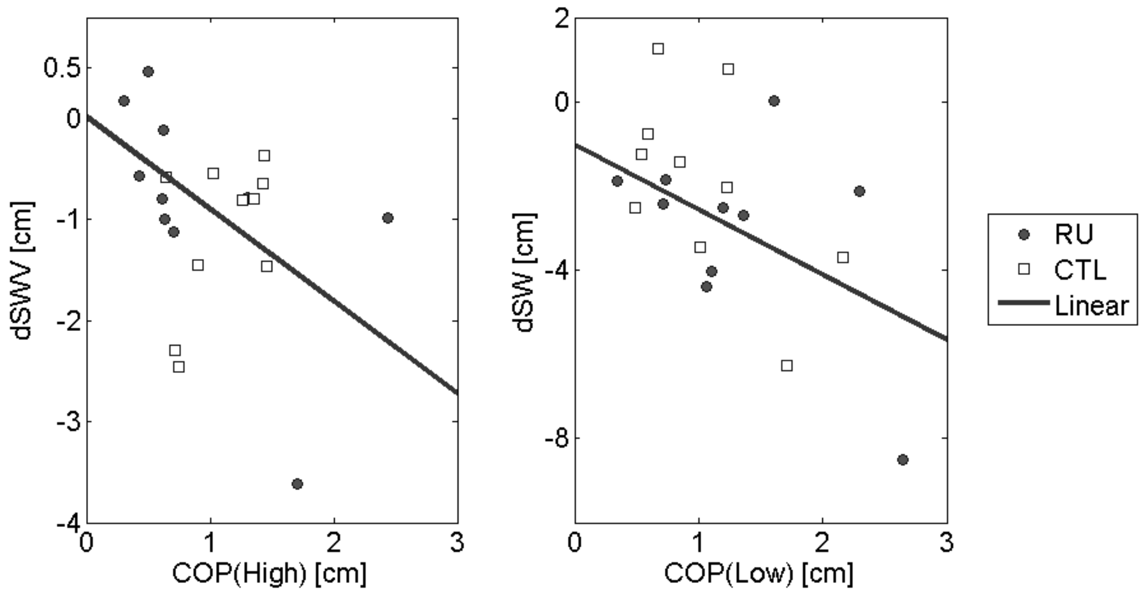

Figure 20: Experiment 2: Linear regressions between upper limb measures to change in spatiotemporal gait measures (relative to without rollator assistance). Upper limb COP shows a significant relation (p = 0.023) to change in step width variability (dSWV) (left panel). The right panel shows the significant relationship (p = 0.014) between upper limb COP$_{(low)}$ and change in mean step width (dSW). Individual RU (filled circles) and CTL participants (open squares) are shown.

Chapter 6
Everyday Balance Related to Rollator-Assisted Mobility: Case Studies in Neurologic Rehabilitation In-Patients

1 Introduction

When mobility is compromised due to motor and/or cognitive impairment from neurological injury, individuals suffer due to declines in activity and participation. As the primary form of physical activity, walking promotes maintenance of healthy musculoskeletal, cardiovascular and tissues and systems [63]. Mobility also facilitates important social interactions [31] and continued independence in activities of daily living [55], both factors related to maintaining mental health [1]. Hence, restoring and maintaining the ability to move from one place to another without the supervision or assistance of another person is a primary goal of rehabilitation following neurological injury, such as stroke [15].

Assistive devices for mobility are often prescribed to facilitate independent mobility during rehabilitation and following discharge. In particular, rollators (4-wheeled walkers) are a class of mobility assistive devices widely used to address gait and balance issues [13]. These devices assist patients with chronic obstructive pulmonary disorder (COPD) by extending walking distances, improving walking efficiency [125] and providing a convenient seat for resting. In the arthritis population, rollators alleviate pain associated with lower limb joint loading by permitting body weight to be shifted to the upper limbs [32]. However, over 50% of the older adult (> 65 y) population who use assistive devices in Canada use them to prevent falls or to 'make me (them) feel steady/balanced' [111] indicating the importance of assistive devices in addressing balance control impairments.

Despite the critical importance of maintaining upright balance for safe mobility, the literature examining the effects of rollators on balance control is relatively sparse [13]. Considering the inverted pendulum model, upright balance is maintained by controlling the position and motion of the body's center of mass (COM) within the base of support (BOS), typically defined by the placement of the feet. Under this model, stabilizing torques comprise of two components: 1) the moment arm, defined as the distance between the point of application of ground reaction forces (center of pressure [COP]) and COM, and 2) vertical load (F_z). The net righting torque acting on the COM is the product of the vertical load and moment arm. With the introduction of a rollator, the upper limbs can be involved in effecting stabilizing torques in addition to the lower limbs. The implications of these additional effectors on maintaining whole body balance has only recently been the subject of investigation [134, 136, 137]. Previously, we studied the involvement of the upper limbs in overall body balance control during standing [134] and walking with a rollator [137]. Overall, an increased reliance on the upper limbs and decreased reliance on the lower limbs was found when balance was challenged. In standing, when the available BOS under the feet was limited (i.e., feet together), the contribution of the upper limbs (indicated by COP measured under the rollator) increased in young adults [134]. Coupled with a corresponding decrease in the contribution of the lower limbs, we concluded that a shift in execution of balance control was attributable to the biomechanical advantage of a wider BOS afforded by the hands on the rollator. During walking, the increased reliance on the upper limbs under increased balance challenge to generate frontal plane stabilizing torques was preserved [137]. In a group of older adults with and without balance impairments the magnitude of upper limb use (indicated by COP measured under the rollator) correlated negatively with changes in stepping parameters, step width and step width variability [137]. Hence, we confirmed that stabilizing moments can be generated by the upper limbs to potentially compensate for lower limb dysfunction in standing and straight-line walking.

While our previous work has highlighted the role of the upper limbs, other studies have demonstrated potential risks associated with walker use. Stevens et al. examined United States emergency room admissions and found that walker-related injuries were 7 times higher than the incidence of cane-related incidents [127]. However, since walker users typically have higher levels of impairment than cane users, the potential fall hazard associated with using walkers is confounded by the intrinsic balance deficits of the user group. One of the challenges for falls research is a lack of data examining the circumstances surrounding a fall due to their relative rarity of occurrence. To illustrate, the annual rate of falls in Canadian seniors is 47.7 per 1000 [99], roughly translating to 1 fall every 2 years. Due to their rarity, data on falls has relied on self-reporting instruments, (e.g. calendars [33], diaries [90], or emergency room admission records [127]) that are limited in providing details of the circumstances leading to the fall. Another approach to assessing fall risk is to examine near-fall incidents, or events that may precipitate a fall. For example, Bateni et al. found that walking frames limit mediolateral foot clearance resulting in collisions between the feet of the user and walking frame, potentially resulting in a greater risk of falling [12].

The focus of the present work is directed to advancing understanding of rollators during everyday activities outside of the conventional laboratory/clinical setting. Everyday environments are often complex, dynamic environments that can pose unpredictable permutations of factors that challenge balance and mobility. While controlled laboratory or clinical assessments are useful in determining balance capabilities under conditions that are relatively simple, whether these capabilities are reflected in natural behaviour remains to be examined. For example, do collisions between the feet and the rollator, revealed under laboratory settings [12] occur in everyday activities? Perhaps more importantly, key behaviours that influence balance control (e.g., position during standing) are relatively unconstrained under everyday conditions leading to the possibility that individuals behave differently when faced with such challenges.

To address these questions, we have developed a unique ambulatory measurement tool based on an instrumented rollator. The iWalker was designed to collect data outside of the traditional laboratory/clinical environment, over extended periods, and with as few constraints as possible towards capturing behaviour more reflective of everyday tasks. Specifically, the iWalker measures rollator motion, stabilizing torques generated by the upper limbs (through the rollator) and simultaneously records video of the upcoming environment and position of the feet [133]. An advantage that an ambulatory measurement protocol potentially provides is the ability to observe behaviour that may not be easily observed or reproduced in a lab or clinic, potentially leading to new hypotheses.

The scope of the current study is to compare rollator-assisted balance behaviour collected under typical laboratory protocols with measurements obtained from a protocol reflective of everyday circumstances in select cases of neurologic patients undergoing in-patient rehabilitation. The secondary goal of this study is to identify events in everyday activity, and the associated environmental conditions, that may be precursors to falling (e.g., collisions) that are difficult to observe and reproduce in a controlled lab setting or clinic.

2 Methods

To compare standing and walking balance with a rollator evaluated in the lab to evaluations conducted over a walking course reflecting everyday challenges, select cases of neurological rehabilitation in-patients were assessed under 2 protocols: 1) in a modern gait and balance laboratory, and 2) performing an extended walking course within and outside a rehabilitation hospital comprising of everyday mobility tasks and situations.

2.1 Participants

Three (3) in-patients undergoing intensive rehabilitation from a traumatic neurological condition were recruited to participate in the study. Inclusion criteria were: i) acquired brain injury, specifically stroke or traumatic brain injury (TBI), survivors who are frequent users of rollators to address balance impairments, ii) the ability to walk independently, with or without walking aid for 200 m (assessed by a physical therapist), and iii) comprehension of the consent form instructions, and ability to follow verbal instruction. Details of their neurological injury, clinical Berg Balance Scores assessed on admission, and gait and balance measures taken the day of testing are summarized in Table 9. All procedures were approved by the local ethics committee.

2.2 iWalker

All trials involving the use of a rollator employed an instrumented rolling walker, iWalker, developed as an ambulatory measurement tool capable of simultaneously measuring variables related to the control of balance (i.e., spatiotemporal foot placement, upper limb kinetics) and the associated environmental context (e.g., spatial surroundings). Full details of complete iWalker tool have been previously described [133]. The iWalker includes 4 single-axis load cells mounted vertically into each leg of the walking frame, a 3-axis accelerometer, and wheel encoder that captures the distance covered. Analog sensor data were converted digitally at 50 Hz (BlueSentry, Roving Networks, Inc.) and sent wirelessly to a personal digital assistant (PDA; iPAQ 2600, HP, Inc.) for storage. Two on-board digital video cameras (Archos Media Player 404, Archos, Inc.) were used to record: 1) placement of the feet, and 2) the immediate spatial context at a rate of 30 Hz. The video and sensor records were synchronized in time by a common visual trigger (i.e., light on the PDA).

Load cell signals were converted to force units, low-pass filtered (2nd order Butterworth; cutoff = 10 Hz), and resolved into a single mediolateral $COP_{iWalker}$ estimate using the following relation:

$$COP_{iWalker} = \frac{(F_{FrLeft} - F_{ReLeft}) + (F_{FrRight} - F_{ReRight})}{F_{FrLeft} + F_{FrRight} + F_{ReLeft} + F_{ReRight}} \qquad (1)$$

where F_{FrLeft}, $F_{FrRight}$, F_{ReLeft}, and $F_{ReRight}$, represent vertical forces from the four load cells and D_{front} and D_{rear} are distances of the load cells to the mediolateral midline of the walker (22.3 cm and 26.6 cm, respectively). The total vertical load (F_z) was calculated as the sum of the four load cell outputs. Note that this approach does not fully consider the effects of shear, or transverse, forces on the theoretical COP (see Appendix A for an assessment).

2.3 In-Lab Assessment

The in-lab component comprised of a standing balance assessment, followed by a straight-line walking assessment. In the standing evaluation, participants were asked to stand on a forceplate (Accusway, AMTI, Inc.) for 60 s in three conditions: 1) without the iWalker in an open stance, and 2) with the iWalker (placed in isolation from the forceplate) in open stance. The mean position was removed and root-mean-square (RMS) of the COP_{feet} record from the forceplate was calculated by taking the standard deviation of the final 30s of the timeseries data in both ML and AP planes. Mean COP_{feet} velocities were calculated by dividing the total sway path by the trial time. In trials involving rollator use, upper limb contribution to balance was measured using RMS values of $COP_{iWalker}$ (Equation 1) and the mean F_z from the last 30 s of the trial.

For the walking assessment, the participant walked along a 4m pressure-sensitive mat (GaitRite, CIR Systems, Inc.) with and without the iWalker at their preferred pace. Walks were repeated until at least 15 steps from each condition were captured, typically 2-3 times. Gait speed was calculated using the time taken to traverse the length of the mat and asymmetry ratio was calculated as the ratio of the average step time of the left leg divided by the average step time of the right leg. In rollator-assisted walking trials, balance was measured using RMS values of $COP_{iWalker}$ and the mean F_z over the walk. In addition, two additional frequency bands of $COP_{iWalker}$ were assessed: 1) a low band (0.5-1.25 Hz; $COP_{(Lo)iWalker}$), and 2) a high band (1.25-5

Hz; $COP_{(Hi)iWalker}$). These two bands were used previously to separate anticipatory and reactive components of upper limb control, respectively [137]. Bandpass filters (Butterworth, 10th order) were employed to separate the bands and RMS values were computed from the filtered signals.

2.4 Walking Course Assessment

Following the in-lab assessments, participants walked along a pre-defined course designed to introduce typical challenges to mobility reflective of everyday, common activities. Illustrated in Figure 21, the course comprised of a path including walking in hallways, up and down ramps, reaching for buttons, entering/exiting doors and taking elevators. On one hallway walk, one of the experimenters (unknown to the patient) walked towards the patient to simulate a pedestrian navigation task. The course distance was approximately 300m in length and took less than 30 minutes to complete. Two opportunities for seated rests were incorporated and participants were permitted to sit on the iWalker seat at any time during the course, if needed.

2.5 Analysis

To address the first objective of comparing balance behaviour collected in the laboratory with everyday situations, periods of standing and walking were identified from the walking course data record by manual inspection of the video record. Standing was defined as a period lasting 5 s or longer in which the participant was standing upright with the iWalker and without foot movement, typically identified while waiting for and riding in elevators. Standing balance measures collected by the iWalker, RMS $COP_{iWalker}$ and mean F_z, were computed for each period. Walking was defined as a period of straight-line movement of 10 s or longer, not including the first and final 3 steps, typically observed in hallways. For each bout of walking, RMS values of $COP_{iWalker}$, $COP_{(Lo)iWalker}$, and $COP_{(Hi)iWalker}$, were computed along with mean F_z and mean gait speed. To compare in-lab and everyday balance behaviour, a Z-test was used to assess whether in-lab assessment scores (e.g., mean walking F_z) were similar to assessments conducted within the walking course.

The second objective, identifying potential precursors to falling, was addressed by first inspecting the video records. The video record provided the opportunity to observe potentially hazardous events such as trips, stumbles, and collisions. Other behaviours that may not have been observed during the standard assessment, such as lifting the rollator, were also noted. The video record aimed at the feet was used as the primary source for observing events and the record of the camera aimed at the upcoming environment was used to detail the immediate and upcoming environmental context (e.g., elevator doors, pedestrians). Following identification of the events, the iWalker sensor timeseries data (e.g., $COP_{iWalker}$, mean F_z, accelerations) was inspected to reconstruct the behavioural sequence leading up to and following each incident.

3 Results

3.1 Patient 1 (P1)

P1 was a 50 yr old male undergoing rehabilitation from his second stroke (left pons), mildly affecting his right leg and foot (Chedoke-McMaster Stroke Assessment (CMSA): foot = 6/7, leg = 6/7). From a previous stroke (2 years prior), P1 had residual impairments to the left side (CMSA: foot = 4/7, leg = 4/7). On admission, the patient was assessed with a Berg Balance score of 44/56 which is within the transition level for an increased risk for multiple falls [93]. The laboratory assessments revealed a strong dependency on the rollator for gait and standing balance. Unaided, the patient walked with a very slow gait speed (0.164 m/s) but displayed a 201% increase when using the rollator (0.494 m/s). Surprisingly, using a rollator increased the patient's asymmetry ratio from 1.14 (unassisted) to 1.23 (assisted). This increase in asymmetry, coupled with such a dramatic increase in walking speed, indicates a heavy reliance on the rollator to facilitate forward progression. Measures of standing balance yielded an increased standing sway velocity without the rollator (COP_{feet} velocity ML = 0.90; AP =1.07 cm/s). When the rollator was made available for support, the patient displayed a reduction of 54% and 39% in frontal plane forceplate measures, RMS COP_{feet} and mean COP_{feet} velocity, respectively. In the sagittal plane, P1 demonstrated a 44% and 48% reduction in sagittal plane RMS COP_{feet} and mean COP_{feet} velocity, respectively.

Objective 1 (In-Lab versus Walking Course Balance)

Figure 22 plots the iWalker measures time series over the entire walking course. The periods of standing are illustrated as yellow bars and indicated by a label Sn, where n is the bout number. Six standing periods were observed, with a mean duration ± SD, 8.7 ± 5.3 s. Six identified walking periods are shown as pink bars (mean duration ± SD: 43.9 ± 21.9 s), and indicated by the label Wn (where n is the period number). The two seated rest periods (Rn) are indicated by green bars; once lasting 62 s followed by a second instance for 8 s. Combined, the identified periods (standing, walking, rest) represented only 31% of the total time elapsed to complete the walking course. In the remaining 69% of time, the patient engaged in movement behaviour that did not fit the standing or walking criteria. For example, while waiting for and riding in elevators, the patient shifted their feet, reached for buttons, and rocked the rollator forwards and backwards. Intermediate movements lasting only a few steps did not meet the minimum 10s criterion for walking. Furthermore, turning and navigating obstacles did not meet the straight-line walking criterion.

The comparison between the in-lab and walking course assessments is plotted in Figure 23, with the standing measures ($COP_{iWalker}$, mean F_z) plotted on the left and the walking parameters ($COP_{iWalker}$, mean F_z, mean gait velocity) plotted on the right. Comparing the standing assessments, no significant differences were found between the in-lab (blue bars) and walking course (yellow) measures. While a large difference between the RMS $COP_{iWalker}$ measured from walking course activities (mean ± SD = 2.40 ± 1.57 cm) and in the lab (Open stance (OP): 0.23 cm; Closed (CL): 0.50 cm) was observed, large between-bout variability in the walking course measures contributed to the lack of statistical significance (OP versus course: $p = 0.168$; CL vs course: $p = 0.227$). For walking, no statistically significant differences were found between in-lab (blue bars) and walking course (red bars) measures. Upper limb measures of balance, mean vertical loading (8.95 vs 9.64 ± 0.68 %BW, $p = 0.309$) and RMS $COP_{iWalker}$ (5.53 vs 5.44 ± 0.49 cm, $p = 0.872$), were similar across in-lab assessment and walking course activities. Mean walking velocity (0.470 vs 0.473 ± 0.05 m/s, $p = 0.946$) was also similar between the in-lab and walking course assessments.

Following completion of the course, the patient experienced fatigue near the end of the walk. A post-hoc analysis was conducted to assess whether the physical demand, measured by the distance covered, contributed to changes in the walking parameters by conducting linear regressions between distance covered (independent variable) and iWalker measures (dependent variable). Significant negative relationships were found between distance and mean gait speed (Figure 24, left, $p = 0.001$) and RMS $COP_{(Lo)iWalker}$ (Figure 24, right; $p = 0.0003$), indicating that fatigue influenced walking performance.

Objective 2 (Potentially Destabilizing Events)

Upon inspection of the video record, 4 potentially destabilizing events were observed. Two (2) instances of collisions between the rollator and elevator door were observed (Figure 22, a marks). In both cases, the elevator door began closing as the patient was exiting and the patient was unable to prevent a collision. Both collision events were associated with large peaks in the iWalker acceleration (Figure 22, top trace), and F_z (Figure 22, third from top) records. Two (2) episodes of stumbling were also observed (Figure 22, b marks) during the walking course. The first event was observed while executing a tight turn, and was associated with a large transient vertical load (F_z) on the iWalker following the stumble. The second episode occurred during preparation to walk down a ramp, and was also marked by a large transient increase in F_z after the stumble.

3.2 Patient 2 (P2)

P2 was 66 yr old female undergoing rehabilitation from head trauma resulting from fall. With a history of cardiac disease (valve replacement surgery pending) and stroke (14 y), coupled with continued episodes of dizziness, this patient was assessed as a high fall risk. Patient P2 demonstrated balance impairments on admission to the rehabilitation hospital, indicated by Berg Balance Score (43/56), further supporting the increased fall risk. This patient demonstrated a 14% increase in walking speed (0.616 versus 0.541 m/s), but demonstrated a substantially more asymmetric gait (Symmetry = 1.13 versus 1.02) when walking with the rollator compared to unassisted. On the forceplate, the patient demonstrated a 72 and 74% decrease in RMS COP_{feet} measures in the frontal and sagittal planes, respectively. An 18 and 21% decrease in mean COP_{feet} velocity was observed in the frontal and sagittal planes.

Objective 1 (In-Lab versus Walking Course Balance)

Over the walking course, four periods of standing (mean duration ± SD = 10.8 ± 5.2 s), 7 periods of walking (mean duration ± SD = 28.0 ± 13.9 s), and 1 period of resting (duration = 110 s) was identified (Figure 6.5). The combined identified periods represented 31% of the total time taken for the walking course. This patient demonstrated fewer standing periods than the other cases due to repeated instances of foot movement while waiting for and riding in elevators. Numerous intermediate movements lasting only a few steps did not meet the minimum duration criteria for walking. Figure 26 plots the standing (left plots) and walking (right) measures for both the in-lab (blue bars) and walking course (yellow and red bars). Overall, standing balance measures assessed in the lab were significantly smaller than those measured over the walking course. Upper limb use measured by RMS $COP_{iWalker}$ was significantly smaller under the in-lab open stance condition compared to the walking course estimates (0.47 versus 0.81 ± 0.07 cm , $p < 0.0001$). Despite a high level of variability, mean vertical load assessed in the lab was significantly smaller compared to the walking course in both open (2.0% versus 13.9 ± 6.0 %BW) and closed stance (1.4% versus 13.9 ± 6.0 %BW). Inspection of the video record demonstrated that the foot placement adopted in each of the standing periods, shown in Figure 27, varied considerably over the walking course. In the final assessment (S4), the patient stood in a manner most comparable to the position used in the laboratory assessment session and exhibited the most comparable loading behaviour (S1 versus in-lab OP: mean F_z = 4.95 versus 2.04 %BW).

For the walking measures, vertical loading (5.35 versus 6.36 ± 0.46 %BW, $p = 0.047$), mean velocity (0.507 versus 0.655 ± 0.035 m/s, $p < 0.0001$) were significantly lower when assessed in-lab compared to walking course behaviour. However, no statistically significant differences were found between the in-lab and walking course $COP_{iWalker}$ (4.68 versus 3.80 ± 0.62 cm, $p = 0.816$).

Objective 2 (Potentially Destabilizing Events)

Manual inspection of the video record yielded 3 observations of lifting and sliding the rollator (Figure 25, a marks), and 1 instance leaving the mobility aid to perform a reaching task (Figure 25, b). Most of the instances occurred as a result of needing to manoeuvre in a confined space. For example, the first (at time-point 212 s) and third (time-point = 675 s) lift-and-slide manoeuvres were observed while moving through a doorway and inside the elevator, respectively. In the second (time-point = 232 s) and fourth (time point = 940 s) observations, the patient was moving backwards and turning.

3.3 Patient 3 (P3)

P3 was a 63 yr old male undergoing rehabilitation from stroke affecting the pons, left midbrain and thalamus. Combined with mild residual impairments from a previous stroke (1 yr, cerebellum), this patient displayed both upper and lower limb motor impairments to the right side (CMSA foot = 5/7, leg = 5/7, hand = 4/7). This patient demonstrated severely impaired balance on admission, scoring only 15/56 on the Berg Balance Scale. P3 displayed a slow gait speed without the rollator (0.385 m/s), which increased marginally with the device (0.413 m/s, 7% increase). No effect on gait symmetry was observed with or without the rollator (Symmetry = 1.10 versus 1.10). During assisted walking, the patient consistently placed the feet closer to the to the hemiplegic (right) side of the rollator frame, identified by inspection of the foot camera video record. In the standing balance measures, using the rollator led to lower COP sway amplitudes compared to standing without a rollator (RMS COP_{feet}: ML= 0.17 versus 0.28 cm; AP = 0.27 versus 0.30 cm). However, an increased mean sway velocity associated with using a rollator was observed in the ML direction (with rollator versus without: 0.49 versus 0.42 cm/s) and a very small decrease in the AP direction (with rollator versus without: 0.85 versus 0.93 cm/s) in this patient.

Objective 1 (In-Lab versus Walking Course Balance)

Four standing periods (mean duration ± SD = 13.8 ± 3.4 s) and 6 walking periods (mean duration ± SD = 34.2 ± 27.0 s) were assessed in the walking course for P3, indicated in yellow and red bars, respectively, on Figure 28. The identified periods of standing, walking and rest periods comprised of 27% of the total time. As the patient with the slowest walking speed, P3 spent more time executing intermediate movements that did not meet the walking criterion, such as manoeuvring into and out of elevators or through doorways. There were also numerous incidents of collisions between the foot and rollator frame that interrupted potential walking periods. P3 relied more heavily on the upper limbs for standing behaviour during the walking course compared to the assessments conducted in the lab. The comparison between the in-lab and walking course assessments is illustrated in Figure 6.9. For standing, this patient loaded the walker frame less in the in-lab open stance condition compared to everyday measurement (4.1 versus 13.6 ± 5.1 %BW, $p = 0.06$), and also exhibited smaller RMS $COP_{iWalker}$ amplitudes (0.43 versus 2.51 ± 1.1 cm, $p = 0.06$). For walking, patient P3 walked significantly slower in the lab compared to the walking course (0.430 versus 0.612 ± 0.026 m/s, $p < 0.01$) and exhibited significantly larger RMS ML $COP_{iWalker}$ (5.87 versus 4.74 ± 0.21 cm, $p = 0.03$). Comparing the vertical loading during walking, no significant differences were observed between the in-lab and walking course assessments (11.25 versus 10.26 ± 1.07 %BW, $p = 0.36$).

Objective 2 (Potentially Destabilizing Events)

This patient exhibited considerable difficulty in negotiating doorways at the beginning and end of the walking course, indicated by collisions between the rollator and a door (Figure 6.8, a, time-points = 165 and 1372 s). This patient also exhibited four episodes of foot collisions with the rollator frame (Figure 6.8, b, time-points = 648, 650, 788, and 1320 s) all during turning manoeuvres. Three of 4 collisions occurred with the affected (right) foot during turning manoeuvres away from affected side (to the left). As an example, a reconstruction of one collision is shown in Figure 30. The patient navigated around another patient in a wheelchair shown in the periphery of the scene camera (Figure 30, upper left panel). Following a change in direction, the right foot collided with the rear wheel of the rollator observed in the foot camera (Figure 30, upper right). The resulting iWalker record shows sharp acceleration responses to the collision (Figure 30, upper plot) in the vertical (black trace) and anterior-posterior (red) plane. An increase in the iWalker vertical load, F_z (Figure 30, lower plot) was observed to peak 60ms following the collision, indicating a rapid upper limb loading behaviour in response to the unexpected perturbation to the lower limbs. Notably, a decrease in F_z (red arrow) was observed 1.2 s prior to the foot collision, likely during the 1-2 steps preceding the event.

4 Discussion

The key finding from this study is that balance and walking assessments conducted in a controlled laboratory environment differed significantly from behaviour observed in everyday conditions. While it is not surprising that short snapshot laboratory assessments may not fully reflect the breadth of challenges needed for mobility in the everyday life, few studies have investigated balance and gait performance in everyday conditions over extended durations. In particular, the task durations in the everyday conditions were different from those typically performed in a lab or clinical assessment and the postures adopted in standing were highly variable. The second key finding is that repeated instances of collisions, stumbling and unloading behaviours that may precipitate falls were measured during the performance of the walking course. These observations of behaviour reflective of everyday situations provide evidence to complement clinically-assessed impairments of fundamental capabilities towards developing more comprehensive individual mobility profiles.

One of the important aspects of the present study is the comparison of behaviour measured in the clinic versus in everyday environments. Two specific differences worthy of attention are highlighted: 1) dependence on the assistive aid (rollator) and 2) gait and balance task characteristics. The difference between unassisted and assisted walking and standing indicates the benefit of using a rollator. P1 demonstrated the greatest benefit from using the rollator for walking, indicated by a 201% increase in gait speed, and standing balance, indicated by a 40-60% decrease in all sway measures. P1 also demonstrated the most consistent use of the upper limbs (mean F_z and COP) in the laboratory and over the walking course. We interpret the combination of high benefit associated with using a rollator and the consistency of upper limb use in the walking course as evidence of P1's dependence on the rollator. Conversely, P2 and P3 displayed more moderate benefit from using the rollator for walking and standing, and demonstrated significant differences between in-lab and walking course behaviour. Considering both the benefit and inconsistency of upper limb use, these findings suggest that these patients do not rely on the upper limbs but are able to use the lower limbs for support.

Gait and standing balance measured in the lab generally underestimated the need for upper limb involvement compared to behaviour in the walking course. Two potential reasons for these differences are considered: 1) task durations and 2) environmental conditions. Overall, standing durations were shorter in duration in the walking course compared to the lab. For the purposes of the study, the definition of standing was loosely defined as an absence of lower limb movement for at least 5 s. The longest observed standing period in the walking course (18 s long) falls well short of the recommended 60 [19] and 120 s [72] trial durations for in-lab assessments. While the rationale for in-lab assessment durations is primarily methodological, our findings do not support their ecological validity. The differences in patient behaviour observed in the laboratory and over the walking course were influenced by the opportunities and constraints of the environment. For example, cases P2 and P3 demonstrated higher average gait velocity during the everyday walking course than in the lab while patient P1 was limited in ability to increase speed due to a strong dependence on the upper limbs for support. We rationalize this difference to the need to cover the greater distances in the hallways (ranging between 30-65 m) in the course compared to the GaitRite mat (4 m) in the lab. Findings of increasing speed to meet environmental demands have been reported in stroke survivors to meet distance goals (e.g., on sidewalk [36]) or temporal constraints (e.g., crosswalk [128]). However, the differences between in-lab and everyday walking speed found in the current study (P2: +0.147 m/s, P3: +0.181 m/s) were considerably larger than those reported in previous studies (e.g., + 0.01 m/s [36]). The consistency of the walking speeds measured in the walking course suggests that patients behaved more cautiously in the lab.

The freedom to move in the everyday environment without task constraints resulted in a range of behaviours. The high variability of behaviours and task situations challenged the definitions of standing and walking. The criterion of static lower limbs to define standing and straight-line steady speed walking rejected numerous potential periods. For example, steady-state standing times while waiting for and riding elevators were shortened or interceded by limb movements (e.g., button presses) and postural adjustments (e.g., shifting foot placement). Slowing down and navigating in response to environmental obstacles, such as pedestrian traffic, were the two most common behaviours rejecting walking periods. As the only other study reporting the frequency of transitions under free-living conditions, Shumway-Cook et al. tracked older adults as they shopped for supplies in their community. They found that seniors performed 2-3 gait initiation and termination events over a 10-minute interval [122], which is comparable to the frequency of gait transitions in the current study (6-7 walking periods in 20-30 minutes). However, the combined standing, walking, and rest periods represented only 27-31% of the total time to complete each walk. This relatively low percentage of identifiable tasks reflects the breadth and flexibility of movement needed for everyday mobility.

Another important aspect of assessing mobility in everyday activities is the potential to reveal events and circumstances that may be a barrier to mobility or increase risk of injury (e.g. fall). We believe that exposure to the complexity of everyday challenges and extended recording durations were critical contributors to capturing adverse events that did not occur during the laboratory assessment protocol, including collisions with the rollator frame, interactions with moving obstacles, and lifting the device for manoeuvring. All of the adverse events captured during the walking course were associated with dynamic tasks, characterized by changing direction and/or speed, and constrained by the everyday context. For example, the lift-and-slide events observed were a result of the need to change direction within a confined physical space, such as an elevator or doorway. Turning with the rollator was a common factor that precipitated events from all three cases and warrants further investigation. The fact that 2 of 3 cases exhibited collisions supports the idea that rollator use increases the risk for potentially destabilizing perturbations [12].

Surprisingly, Shumway-Cook et al. observed no instances of collision avoidance behaviour in adults with mobility disability throughout their study observing participants video records while they went grocery shopping [122]. The most likely contributor to the lack of observed avoidance behaviour is traffic density of the environment. Patients in the current study exhibited numerous direction changes to avoid pedestrians or obstacles, likely as a result of increased frequency of pedestrian encounters in the hallways of a busy hospital environment. Furthermore, Shumway-Cook rationalized that the walking speed of the adults with disabilities was slower than those around them and the faster walkers tended to walk around the participants rather than vice versa [122]. This rationale may also explain the higher frequency observed in the current study. Many of the pedestrians encountered in the walking course moved more slowly than the participants (e.g., pedestrians in wheelchairs), and the participants were forced to avoid them.

Since the rate of falls is low, capturing a fall event would require further extension of recording durations. With estimated walker-related fall rates at 1.2% of users per year (US estimates) [22], recording durations of weeks and months are needed. However, we believe that recording near falls, or potentially destabilizing events (e.g., collisions) is an effective means of examining the everyday circumstances linked to increased fall risk. In the current study, we were able to profile 2-4 repeats of potentially destabilizing events in each patient case over approximately 25 minutes of recording. An important limitation of the protocol reflective of everyday activities also emerged. The events are associated with challenges of a live environment that is unpredictable, affecting the frequency and type of behaviour observed. To address this uncertainty, more detailed measurements of the exposure to environmental conditions can be used to quantify the type and level environmental challenges encountered.

Overall, behaviour observed from a naturalistic protocol provides strong evidence to complement clinical assessments. By combining the basic walking and balance capabilities measured in the lab with everyday performance, individual profiles of the challenges to balance emerged from each of the cases studied. These profiles can be used to target specific training, assess suitability for assistive devices, and recommend rehabilitation goals. The iWalker tool facilitated the walking course protocol reflective of everyday activity by providing the ability to record and reconstruct events with high temporal precision. Further methodological development is needed to extract potentially destabilizing events such collisions, stumbling, and lifting, in a time-efficient manner. For example, using the sensor data to extract such features to reduce the video data required for visual inspection is currently being explored. Given the strong influence of environmental factors on balance and gait measurements, including a record of the associated environmental context is recommended for future ambulatory monitoring methods.

	Patient 1	Patient 2	Patient 3
Age (sex)	50 (M)	66 (F)	63 (M)
Neurological condition	Stroke	TBI	Stroke
Lesion location (type)	Left pons	Frontal (hemorrhage)	Left pons, thalamus
Time from incident	57	48	41
Comorbidites	Stroke (2 yrs)	Stroke (14 yrs) Bilateral knee arthroplasty Cardiac disease	Stroke (1 yr) Diabetes mellitus
Berg Balance Score	44/56	43/56	15/56
Gait speed Unassisted (Assisted) [m/s]	0.164 (0.494)	0.541 (0.616)	0.385 (0.413)
Symmetry ratio $^{+}$ Unassisted (Assisted)	1.14 1.23	1.02 1.14	1.10 1.10
Standing RMS COP_{feet} Unassisted (Assisted) [cm]	ML: 0.37 (0.17) AP: 0.27 (0.15)	ML: 0.26 (0.07) AP: 0.23 (0.06)	ML: 0.28 (0.17) AP: 0.30 (0.27)
Mean COP_{feet} Velocity Unassisted (Assisted) [cm/s]	ML: 0.90 (0.55) AP: 1.07 (0.56)	ML: 0.40 (0.33) AP: 0.65 (0.52)	ML: 0.42 (0.49) AP: 0.93 (0.85)

RMS COP = root-mean-square of center-of-pressure; *Mean COP Velocity calculated as total sway path/trial time; $^{+}$ Symmetry ratio calculated as ratio of left leg step time/ right leg step time

Table 9: Participant profiles including type of neurological condition, co-morbidities, Berg Balance Score, time since injury, gait, and forceplate standing measures.

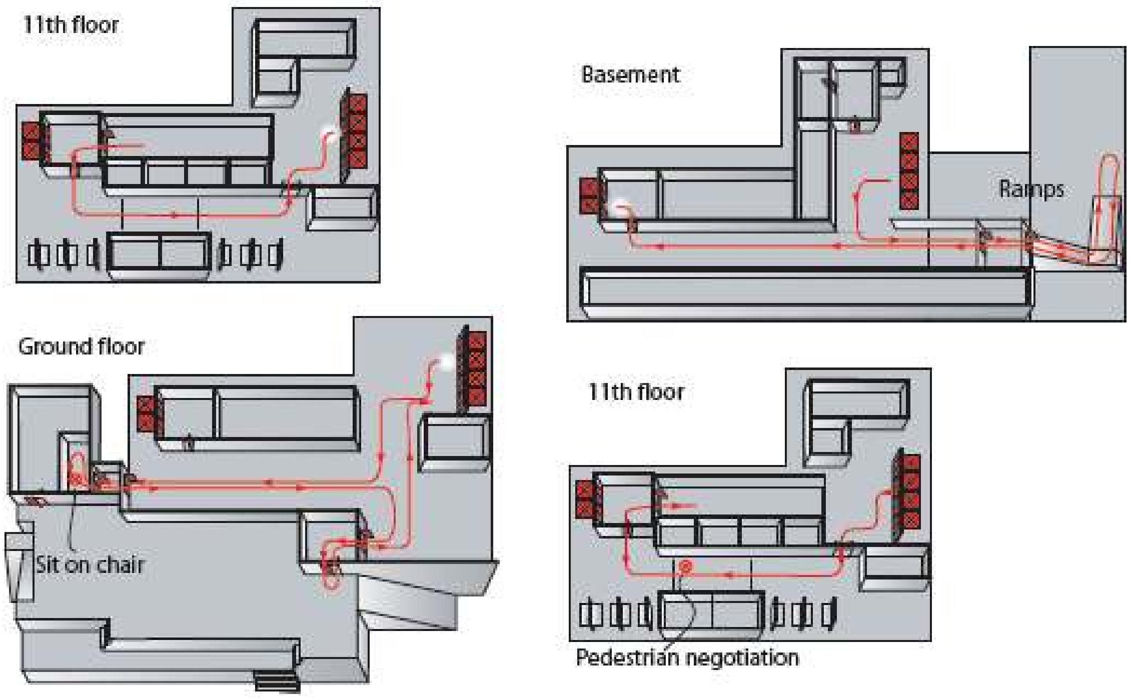

Figure 21: Schematic of pre-defined walking course designed to reflect everyday mobility challenges. The course takes place inside of a rehabilitation hospital and includes negotiating doorways, ramps, elevators, terrain changes, and pedestrian traffic.

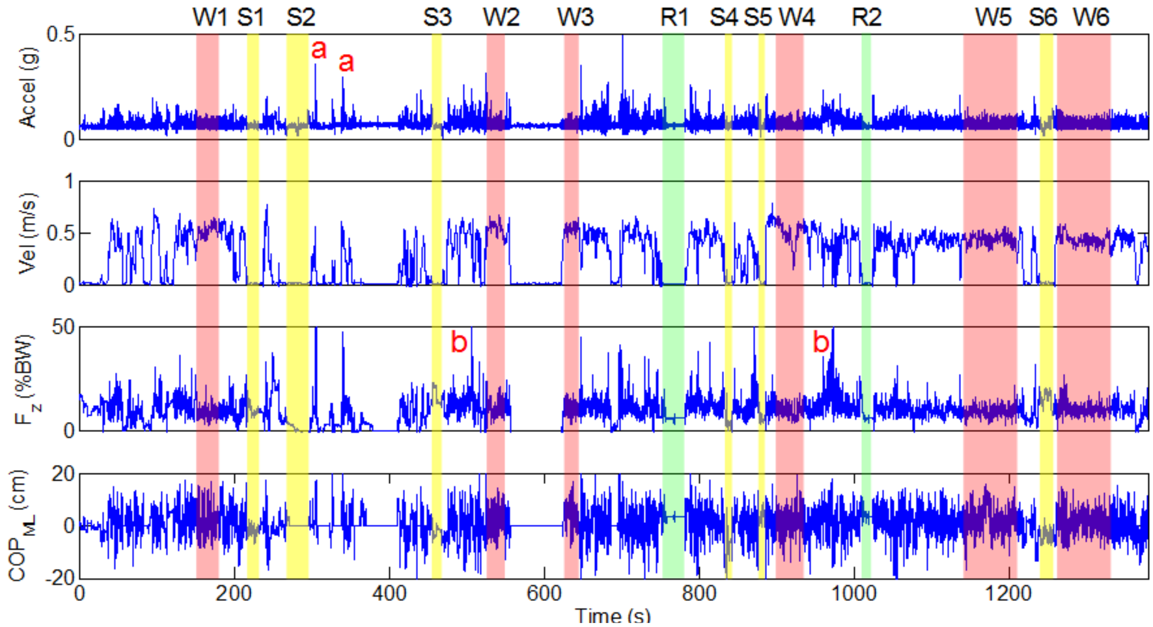

Figure 22: Walking course activity data for P1. iWalker acceleration (top), velocity (second from top), vertical load (third), and mediolateral COP (bottom) timeseries are plotted. Standing assessment periods are marked by yellow bars and labeled Sn, where n is the assessment number. Walking and rest periods are marked by red and green bars with Wn and Rn labels, respectively. a = elevator door collision; b = stumble.

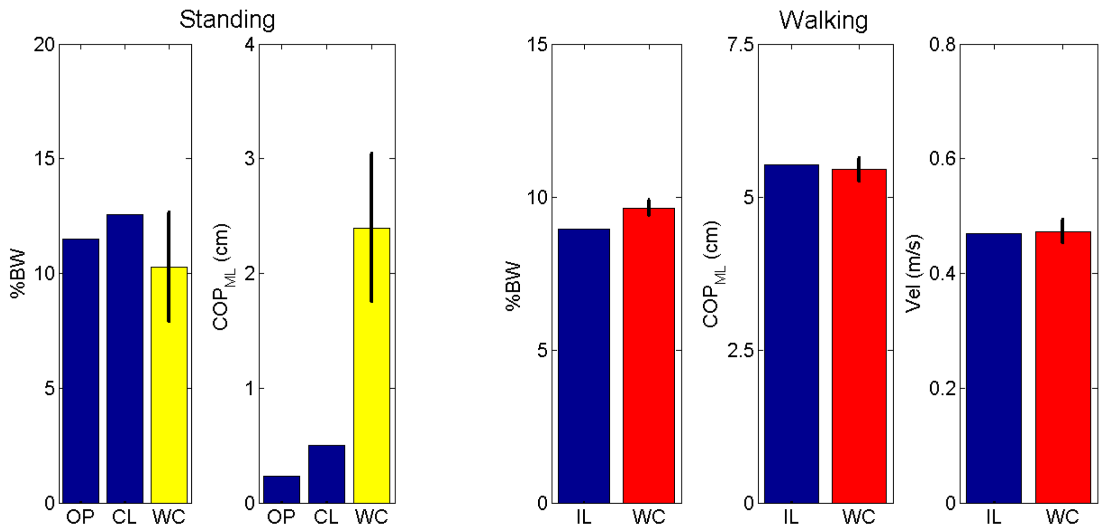

Figure 23: In-lab and walking course activity comparison for P1. Standing (mean F_z, $COP_{iWalker}$) and walking measures (mean F_z, $COP_{iWalker}$, mean gait velocity) are plotted from left to right. Blue bars indicate in-lab assessments, and yellow and red bars indicate everyday measurements for standing and walking, respectively. Error bars represent standard deviation. OP = open stance; CL = closed stance; IL = in-lab; WC = walking course.

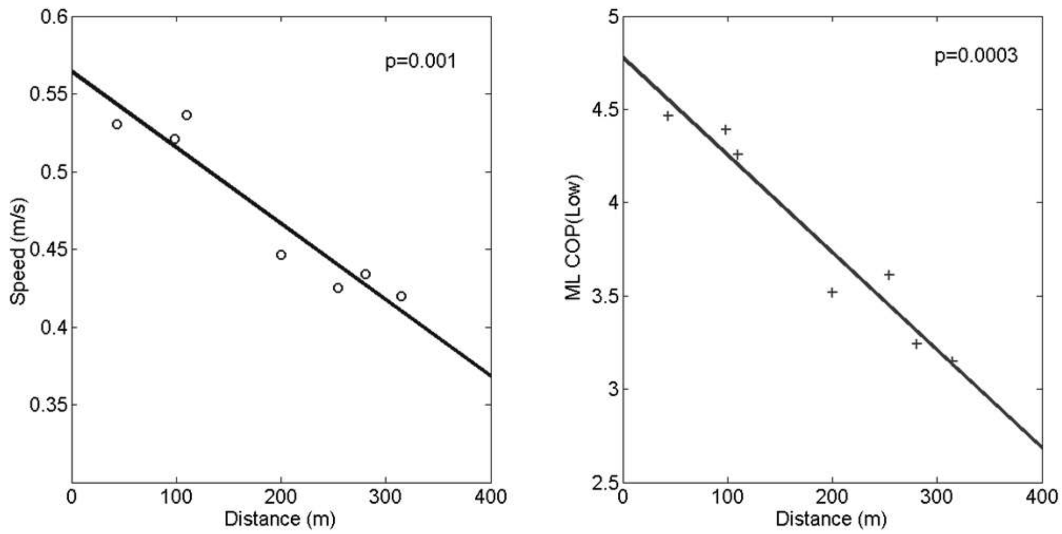

Figure 24: Effect of fatigue on walking and upper limb use in P1. Walking course measurements of gait speed (left panel) and RMS COP(Lo)$_{iWalker}$ (right) are plotted over the distance covered. Linear regressions are plotted as solid lines.

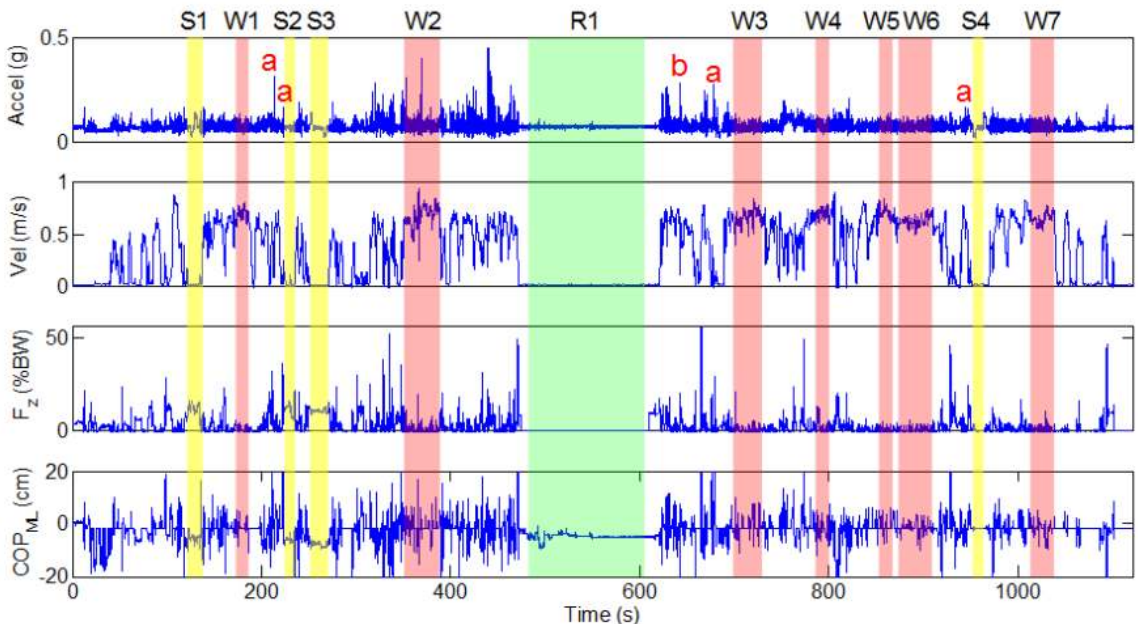

Figure 25: iWalker walking course timeseries for P2. Acceleration (top), velocity (second from top), vertical load (third), and mediolateral COP (bottom) are plotted. Standing (yellow bars), walking (red bars) are marked by yellow bars and labeled Sn, where n is the assessment number. Walking and rest periods are marked by red and green bars with Wn and Rn labels, respectively. a = lifting rollator; b = leaving rollator

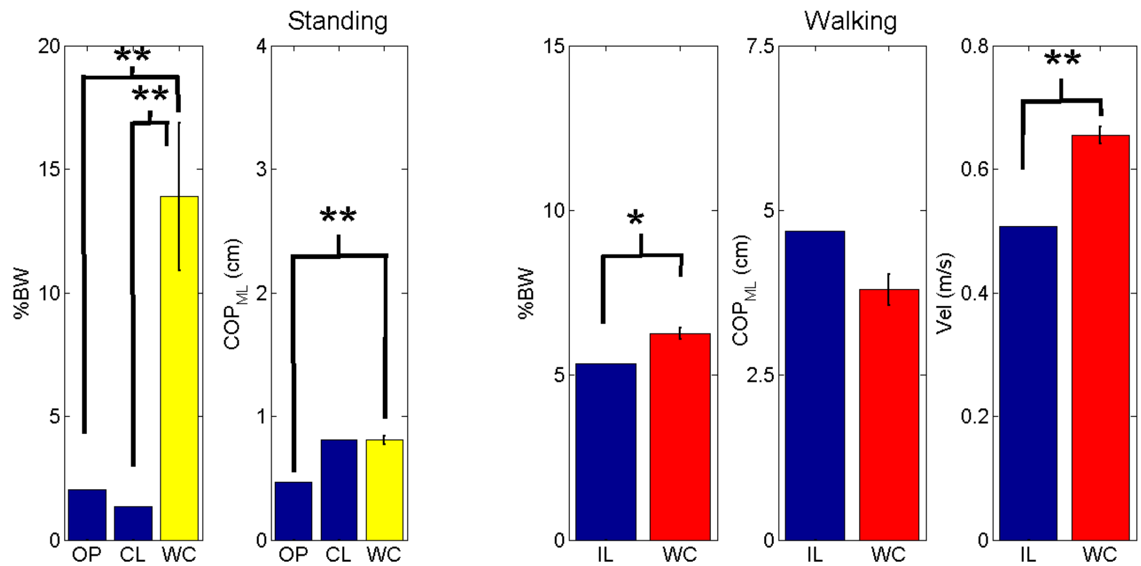

Figure 26: In-lab and walking course activity comparison for P2. Standing (mean F_z, $COP_{iWalker}$) and walking measures (mean F_z, $COP_{iWalker}$, mean gait velocity) are plotted from left to right. Blue bars indicate in-lab assessments, and yellow and red bars indicate everyday measurements for standing and walking, respectively. Significant differences indicated by * ($p < 0.05$) and ** ($p < 0.01$). Error bars represent standard deviation. OP = open stance; CL = closed stance; IL = in-lab; WC = walking course.

S1 S2 S3 S4

Figure 27: Variability in stance position during periods of standing measurements during the walking course. Pictures from iWalker foot placement camera captured during the 4 standing assessments (S1-S4, yellow bars in Figure 5.5) are shown left to right.

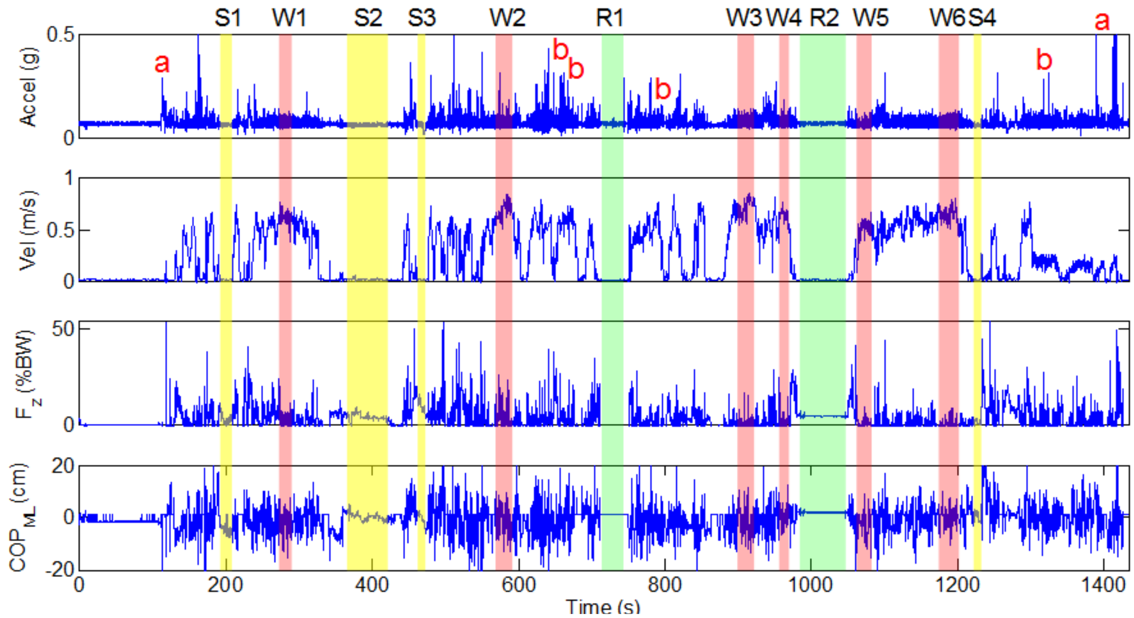

Figure 28: iWalker walking course timeseries for P3. Acceleration (top), velocity (second from top), vertical load (third), and mediolateral COP (bottom) are plotted. Standing (yellow bars), walking (red bars) are marked by yellow bars and labeled Sn, where n is the assessment number. Walking and rest periods are marked by red and green bars with Wn and Rn labels, respectively. a = collision with door; b = foot-device collision.

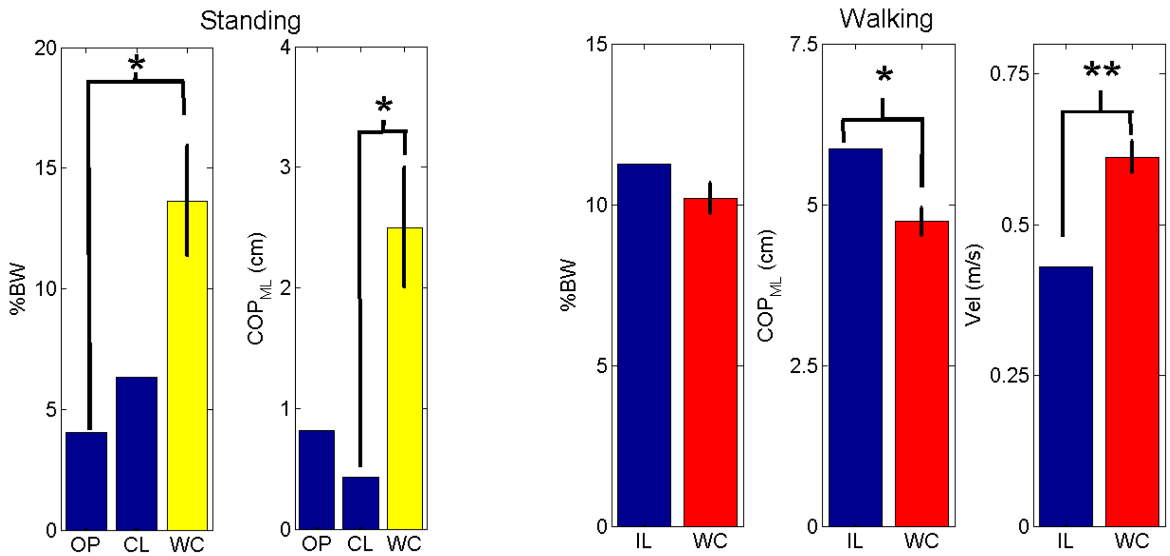

Figure 29: In-lab and walking course activity comparison for P3. Standing (mean F_z, $COP_{iWalker}$) and walking measures (mean F_z, $COP_{iWalker}$, mean gait velocity) are plotted from left to right. Blue bars indicate in-lab assessments, and yellow and red bars indicate walking course measurements for standing and walking, respectively. Significant differences indicated by * ($p < 0.05$) and ** ($p < 0.01$). Error bars represent standard deviation. OP = open stance; CL = closed stance; IL = in-lab; WC = walking course.

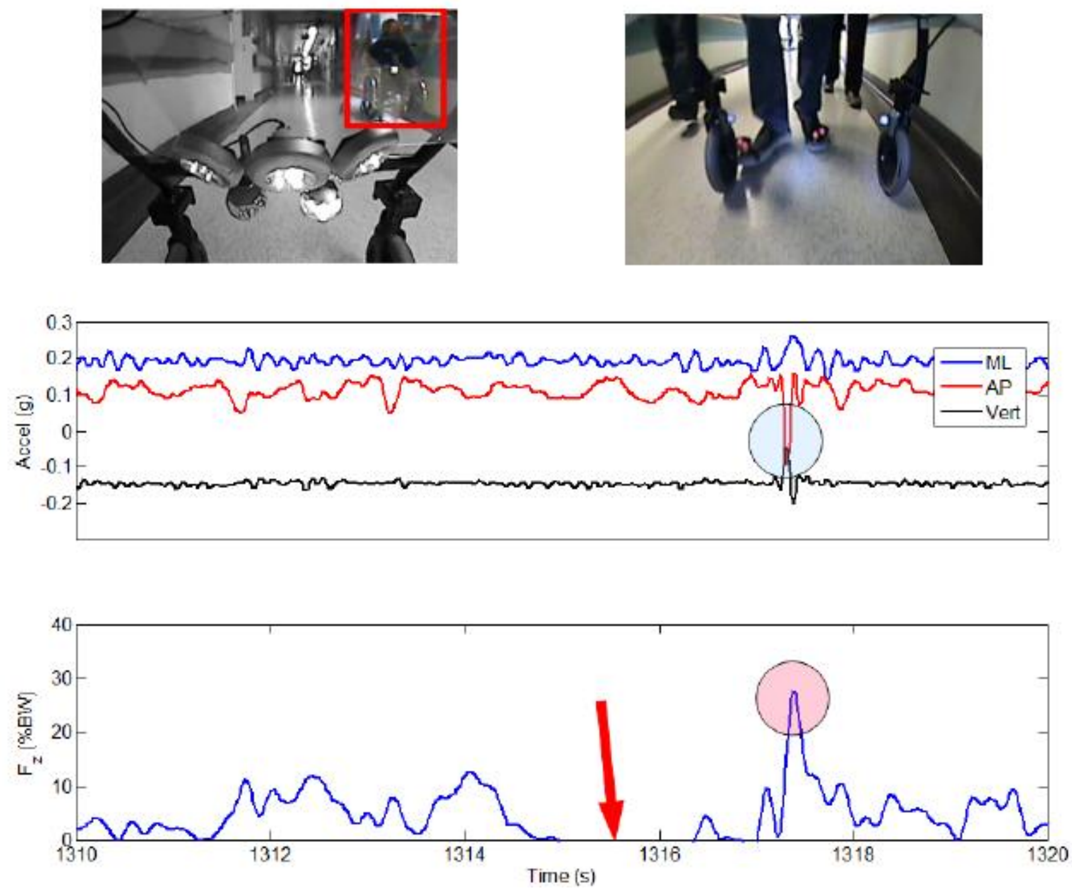

Figure 30: iWalker data from foot-device collision. The patient executed a direction change to avoid a wheelchair pedestrian (red box, top left panel), and subsequently collided his right foot against a rear wheel of the walker (top right). iWalker timeseries accelerometer (lower panel) and vertical load (F_z, lower plot) are shown. Blue and red circles indicate accelerometer and upper limb loading responses to the collision. Red arrow indicates a reduction in upper limb loading 1-2 steps prior to the collision.

Chapter 7
General Discussion

In this concluding chapter, the key findings of the laboratory and walking course studies are summarized and their significance in addressing the benefits and adverse consequences of using rollators is discussed. Furthermore, the issues and lessons learned from the current body of work are discussed towards clinical assessment, designing more effective assistive device designs, developing studies of larger scale, and the utility of an ambulatory assessment approach.

1 Summary of key findings

The overall purpose of the thesis was to assess the potential effectiveness and challenges associated with using rollators to maintain balance in everyday conditions. The main benefit found was that rollators enabled the upper limbs to play an important role in maintaining stability, particularly when the capabilities of lower limb involvement are compromised. This benefit was observed in both the laboratory studies investigating upper limb involvement in standing and walking tasks, and in environments relective of everyday life. By adopting an ambulatory measurement approach, the specific conditions in which rollators assisted the user in preventing a fall were precisely reconstructed. This approach also exposed the circumstances leading up to the occurrence of potentially destabilizing events associated with rollator use, particularly related to the physical footprint introduced by the device.

1.1 Benefits related to upper limb contributions to stability

As shown in the laboratory studies (Chapters 4 & 5), the availability of hand supports, in the form of a rollator, led to individuals mechanically relying on upper limbs to control stability in (quiet and perturbed) standing and straight-line walking when balance was experimentally challenged. As suggested by the existing literature examining the effect of walkers on lower limb behaviour [12, 13, 74], stability control can be effected by the upper limbs to compensate for lower limb limitations in addition to providing haptic feedback. Previous work examining the role of the upper limbs when grasping a handhold in whole body balance control have also found that the upper limbs play a significant, even dominant, role in maintaining stability. The current work extends the importance of the upper limbs in two ways relevant to the use of an assistive device: 1) using a bilateral upper limb configuration, and 2) tonic vertical loading through the upper limbs.

The configuration of the hands, defined by the rollator handles, influenced the directional differences in upper limb involvement observed in the static standing study. Compared to the greater mechanical advantage in the frontal plane provided by the bilateral configuration of the hands, the support base in the sagittal plane is limited to the grip surface. Hence, we attribute the directional differences to distinctions in hand placement and mechanical leverage. The influence of larger moment arms afforded by the upper limbs (relative to the lower limbs) is supported by previous studies examining upper limb integration into balance control when grasping single handholds. In response to support surface translations, Cordo and Nashner found a dominant shift in control to the upper limb while grasping a single handle at chest height [27]. More recently, Elger et al [38] observed a shared upper and lower limb response when gripping a single handhold at waist height. We interpret the larger moment arm provided by the height of the handle in the former study (i.e., chest height) compared to shorter moment arm in the latter study (i.e., waist height) as an important factor influencing the relative contribution of the upper limbs. In these studies, the stabilizing forces were generated principally through horizontal (shear) forces and the moment arm is defined by the height of the handholds. In contrast, this thesis demonstrates that frontal plane stabilizing moments are effected primarily through shifts in vertical loading applied through the rollator frame.

An important contribution of the thesis was examining the influence of tonic upper limb vertical loading, expressed as % body weight applied, on maintaining balance. Overall, an increase in vertical loading was observed as the level of balance challenge increased in healthy controls. In comparison to other tasks, the lowest levels of vertical loading were reported during quiet standing (2-3% BW) and rose to higher levels during perturbed standing (6-9% BW). The walking results continue the upwards progression from 9.1% BW in level ground walking to 13.7% BW under the challenged balance condition. The influence of tonic vertical loading was directly addressed in the perturbed-standing experiment (Chapter 4) by controlling the ability to load the upper limbs prior to the perturbation (i.e., Light Touch condition). When tonic loading was permitted prior to the perturbation, lower limb involvement (i.e., peak COP excursion) decreased and upper limb response onset times tended to be faster. These results indicate that tonic loading facilitates both upper limb response magnitude and faster timing, potentially through increased passive stiffness. Increased passive stiffness from increased tonic contraction has been shown in the ankle [Weiss 1986], and hand [Carter 1990]. When using a rollator, the increased stiffness associated with preloading resulted in passive responses to perturbation occurring at latencies earlier than active CNS control.

Overall, the studies in this thesis support a common balance control system integrating the upper and lower limbs. The relative upper and lower limb control contributions scaled according to the stability demands and demonstrated the ability to shift between sets of limbs to compensate for restrictions or impairments. Furthermore, the temporal coupling of the upper and lower limb COP dynamics was generally consistent in both static and perturbed standing. Further insight into upper limb integration to whole body balance control could be elucidated by assessing the directional-dependence of upper limb control. As one of the hallmarks of balance control when only lower limb control is available [146], examining the influence of perturbation direction could shed further light on upper limb control mechanisms. For example, appropriate unilateral upper limb EMG responses to opposing mediolateral perturbations (i.e., left and right) would indicate directional-dependency. Alternatively, a bilateral EMG response could indicate a generalized upper limb stiffening strategy. In Chapter 6, the upper limb involvement to maintain stability observed in the laboratory was generalized to conditions more reflective of everyday life. In the walking course protocol that exposed participants to such conditions, upper limb activity was observed in response to potentially destabilizing events, such as perturbations from collisions, terrain changes, and stumbling. These initial findings observed under everyday life conditions present new evidence to link the mechanisms for stability benefits examined in the laboratory to everyday life circumstances.

1.2 Adverse consequences associated with rollator use

In the walking course study (Chapter 6), adverse consequences related to rollator use were observed. In particular, repeated instances of collisions and unloading behaviours that could be precursors to falls were observed. The repeated instances of collisions between the user and the rollator frame provides direct observations from everyday life situations to link the increased risk for collisions demonstrated in the laboratory [12] with reports from epidemiological sources (i.e., emergency room admission data) [22]. Importantly, the work presented captured the specific circumstances surrounding the potentially destabilizing event. All of the adverse events captured from the walking course study were associated with dynamic tasks, characterized by changing direction and/or speed.

In particular, turning and navigating with a rollator was a common characteristic of the circumstances for potentially destabilizing events from all three cases. This finding provides empirical evidence to support the perceived difficulty of manoeuvrability when using rollators [17]. Manoeuvrability issues were likely due to the increased physical footprint associated with using a rollator, similar to the issues observed in adapting to the wider base of wheelchairs [68, 54]. Furthermore, the increased physical footprint translated into a number of collisions with environmental obstacles (e.g., doorways and elevators) that potentially introduce destabilizing perturbations.

2 Implications

In light of the new evidence supporting the balance benefit and specific circumstances underlying an increased fall risk, the implications for rollator use for fall prevention are discussed in terms of clinical use, device design improvements, and future research directions.

2.1 Clinical implications

In contrast to wheelchair mobility [67, 68, 69], there are few guidelines for safe and effective rollator use. Having a stronger understanding of the role of the upper limbs, the factors relating to their involvement, has implications for rollator prescription and training. Furthermore, the initial examination of rollator use in everyday life has identified underlying circumstances that pose problems which may indicate specific areas for targeted training intervention.

The effect of preloading levels on recovery responses emphasized the need to consider upper limb capabilities. One of the main benefits associated with rollator use was the ability to generate stabilizing moments with the upper limbs. While there may be potential benefit to stability associated with increased haptic information through the upper limbs [60, 38], reduced preloading was associated with larger requirement of lower limb involvement to stabilize against perturbations. The implications of upper limb impairments (i.e., arthritis) influencing the ability to load the rollator suggests the need to consider upper limb assessments in prescribing rollators.

For example, the impact of reduced hand grip strength on the capacity to use a rollator for stability should be considered.

Manoeuvring the rollator, such as turning and navigation, was a common factor involved in the majority of the potentially destabilizing events (i.e., collisions, stumbling). While turning strategies without mobility aids have been suggested as a potential fall risk indicator [2, 51, 35], further investigation of the impact on balance control during turning with a rollator is warranted. In particular, guidelines for training safe turning can be evaluated to mitigate the potential increase in fall risk associated with manoeuvrability tasks.

As a clinical assessment tool, the iWalker could be an efficient way to evaluate potential sources of fall risk associated with using rollators towards developing performance targets. The aim of tailored assessment and intervention programs for fall prevention is to target interventions to the risks based on individual assessments [66, 47, 21]. In the case series study, a profile of an individual's balance and mobility encompassing a wide range of typical tasks emerged over the 25-30 minute walking course protocol. For example, a patient case (P1) in the walking course study demonstrated repeated stumbling events at vertical transition points (i.e., ramps, thresholds). Targeted training programs can then be developed towards negotiating these environmental features that pose a fall risk.

2.2 Rollator design recommendations

Considering the importance of upper limb involvement to assist in maintaining stability, the implications of the key findings on rollator design should be highlighted. As the interface between the device and user, handle design plays an important role in the stability benefit provided by the rollator. Specifically, designing rollators to facilitate vertical loading and optimize the leverage (i.e., moment arm) are recommended.

Understanding that upper limb stabilizing moments are effected primarily by applying vertical loads to the rollator frame leads to potential design recommendations to facilitate such loads. In Chapter 5, we observed that 4 of 10 balance-impaired rollator users suffered from upper limb arthritis, often in the fingers and wrist, which likely limited their ability to apply vertical loads to the handles. Designing ergonomic handle interfaces are recommended to reduce joint stresses to facilitate vertical loading. For example, designing wrist supports to maintain a neutral angle and increase surface area to distribute loading may be an effective means to facilitate upper limb involvement. The height of the handles may also an important parameter influencing the ability to vertically load the upper limbs. Recently, Takanokura examined the effect of handle height on upper limb and back muscle loads using a model simulating the arm and trunk positions of standing with a rollator [147]. The simulation results indicated a critical handhold height of 48% (of body height) in which vertical loading was maximized and upright posture was maintained. Increasing the height resulted in reduced vertical loading and lowering the handles introduced trunk flexion. Although the model findings present initial indications, the effect of handle height on vertical loading behaviour, and the implication for upper limb balance control, in human subjects remains to be confirmed.

One of the key benefits provided by the rollator is the large moment arm defined by the width of the handles. Current design standards for rollators, such as those developed by the International Standards Organization (ISO) [59], have focused on the width of the wheels as a design standard and not on the handle width. There have been rollator models that feature the ability to lock the handles into a horizontal bar position similar to a shopping cart handle allowing the hands to be placed together to improve manoeuvrability. Considering the direct impact of hand width placement on the ability to generate frontal plane torques, defining and adopting minimum handle width standards is recommended. Although brake use was not the focus of the present work, difficulty using the brake feature is one of the more common complaints about rollator use [148]. In our experience with rollator users (i.e., balance-impaired older adults and neurological rehabilitation in-patients), the brakes were used only in 2 situations: 1) during chair transfers; and 2) ramp descent. In the former case, the brakes were locked to maintain a stable support to assist with transfers to and from a seated position. In the latter case, brake were applied prior to and during ramp decent to assist with controlling forward progression. We also observed a dichotomy in brake use behaviour between in-lab standing and standing in everyday conditions. While standing in the laboratory assessment, the brakes were locked to prevent rollator movement. In contrast, the rollator users did not apply the brakes during periods of standing in the walking course. Overall, we interpret these observations as support for the self-reported difficulty in using brake mechanism.

As an ambulatory measurement device, the iWalker tool was used to identify and define specific issues for new mobility assistive device designs and features. For example, a comparison of alternative braking systems may be useful in assessing the potential balance benefit of less demanding braking mechanisms. The iWalker could provide a method of recording brake usage behaviour and the specific conditions in which they were applied to help define new design criteria (e.g., limiting speed during turns). Prototypes of new designs can then be field tested using the iWalker system for effectiveness in facilitating mobility and minimizing fall risks.

2.3 Key lessons for further studies

The findings from these studies, and experience in conducting the protocols, provide input to guide future studies. One of the main limitations of the described works is the small number of participants in the walking course study. Hence, issues related to scaling the study to a larger number of participants and longer durations should be highlighted. Scaling a study similar to the protocol used in the case series study (Chapter 5) presents a data management and processing challenge. Video data capturing details of the environmental context, particularly the immediate physical environment, was a key component of the iWalker record used to reconstruct the sequence of events. Since manual inspection of the video records was time-intensive, developing methods to reduce this load are recommended. Automated machine vision algorithms to extract and code relevant environmental features (e.g., upcoming pedestrians) could be employed to reduce the analysis load.

In general, the studies detailed in this thesis demonstrated the significance of the upper limb contributions, particularly when lower limbs contributions are limited. However, when the lower limbs were not compromised and the challenge to balance was manageable by the individual, there were large variances in upper limb behaviour. Hence, a clear picture of the overall balance behaviour requires a combination of both upper and lower limb contributions. The iWalker has the potential to extract step width from a single lens video camera mounted inside the well of the rollator. Further testing of the algorithm on a larger test group, and generalization to a clinical population are needed to evaluate its accuracy. While the current algorithm requires light-reflecting markers to be placed on the 1st and 5th metatarsals, development efforts for markerless feature extraction are on-going [94]. An alternative approach could be to incorporate inertial sensors (e.g., accelerometers) or footswitches to capture foot contacts [108].

The iWalker is limited to upper limb kinetics related to the vertical loading applied to the rollator frame, and did not collect transverse plane, or shear, forces. While shear forces can play a substantial role in maintaining stability, estimates of their contributions collected from forceplate data in quiet standing indicates that they play a smaller role compared to the vertical loading shifts (see Appendix A, section A.2 for details). While the work described in the thesis employs the vertical load shifts as an indicator of upper limb involvement in maintaining stability, the full mechanical contribution of the upper limbs cannot be determined without the shear force measurements.

In principle, the ambulatory measurement approach was successful in identifying and describing the interactions between intrinsic capabilities and extrinsic factors that dictate the task. One limitation of the iWalker as the sole collection tool was that any activity conducted without the rollator would not be captured. However, one of the advantages of the iWalker is a relatively stable platform for mounting sensors and acquisition equipment. There remain significant technical barriers to capture features of the spatial surroundings and corresponding balance behaviour without the advantage of a stable platform. However, the benefit of developing such techniques is the ability to generalize the ambulatory measurement approach to unaided mobility.

Another outstanding challenge to larger scale studies using everyday life collection protocols is selecting appropriate analysis techniques. While the principal advantage of utilizing ambulatory measurement protocols (e.g., walking course) is the ability to observe behaviour in its natural context, there is an important lack of control over the frequency of activities and associated environmental conditions. The implication of limited ability to manipulate the relevant factors is the likely difficulty in replicating the results, which renders traditional laboratory analysis techniques based on randomized-assignment-to-treatment designs (e.g., ANOVA) inapplicable to everyday life protocols. A case study (i.e., n=1) approach may be used to explore the interactions between intrinsic body factors (e.g., gait asymmetry) and extrinsic factors (e.g., obstacles) that may underlie specific behaviours (e.g., collisions) to form initial hypotheses that may be tested more rigorously with controlled laboratory experimentation. Another potential avenue draws from epidemiological studies examining fall rates that also face similar challenges with replication due to the rarity of falls. Drawing from this work, ambulatory measurement studies may consider a relative risk approach to help account for uncontrolled factors, such as activity levels and range of environmental conditions encountered. An important advantage of the iWalker tool is the capacity to describe relevant events in high detail, such as characterizing and quantifying near-falling events from everyday life. Measuring these near-fall events have not been feasible in epidemiological research and could be more sensitive outcomes reflective of fall risk.

3 Conclusions

This thesis advances a methodological approach, symbolized by the iWalker, towards developing a deeper understanding of the benefits and risks associated with rollator use. By facilitating the capture of a combination of upper limb behaviour and information about the environmental context, the iWalker enables the possibility of conducting ambulatory measurement protocols. Standing and walking studies provided demonstrations of the significance of the upper limb measures for maintaining stability, insight into factors influencing their control, and a basis for interpretation of data collected from everyday activity protocols. As an initial evaluation, the case series study (Chapter 5) demonstrates the feasibility and utility of using the iWalker in larger scale studies involving longer durations and a greater number of participants.

Generally, rollator users benefit from the availability of upper limb support to generate stabilizing forces as observed both under laboratory and during a 30-minute walking assessment similar to everyday life conditions. The biomechanical factors (e.g., preloading, upper limb impairment) influencing the coupling the upper and lower limbs for maintaining stability were explored. However, rollators introduced an increased risk of foot collisions with the device and were difficult to manoeuvre. The circumstances surrounding repeated adverse events (e.g., collisions, stumbling) were observed in patients using rollators during recovery from neurological injury (i.e., stroke, traumatic brain injury). Implications of this work on clinical assessment, rollator assistive device design, and falls research are discussed.

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Appendix A: Forceplate Calculations and Free Body Diagrams

1 Resolving 3 forceplate setup into a single upper limb COP

This section describes the method used to resolve the output of 3 forceplates used in Chapter 4, Experiment 1: Quiet Standing into a single COP calculation. Figure A.1 illustrates the setup of the 3 forceplates used to capture upper limb kinetics (Forceplates 1, 2, and 4; light boxes), and the 1 forceplate used to measure lower limb COP (Forceplate 3, dark box).

The rollator was placed with the 2 front wheels on Forceplate 1, the left rear wheel on Forceplate 2, and the right rear wheel on Forceplate 4 (see Figure 4.2). To derive the overall resultant upper limb COP, the individual forceplates were first resolved into separate vertical force and COP vectors for each forceplate. Equations A.1 and A.2 provide example formulae for COP calculations at Forceplate 2.

$$COP_{frontal,1} = -\frac{M_{y,2} + F_{x,2} \cdot z_{offset,2}}{F_{z,2}} \qquad (1)$$

$$COP_{sagittal,1} = \frac{M_{x,2} - F_{y,2} \cdot z_{offset,2}}{F_{z,2}} \qquad (2)$$

where $M_{y,2}$ and $M_{x,2}$ are the measured moments about frontal and sagittal planes, respectively. Due to the difference between the geometric and actual (measurement) origin of the forceplate, the second terms in Equations A.1 and A.2 are corrections for the shear force contributions to the COP. The moment produced by the shear forces, $F_{x,2}$ and $F_{y,2}$, acts through the moment arm defined by the distance of the surface to the actual (or measurement) origin of the forceplate.

Once the COP local to the individual forceplates were calculated, a single resultant COP with a global origin was calculated using an equivalent sum of moments.

$$M_{y,origin}+(COP_{y,1}+(46.55+7.49)).F_{z,1}-(COP_{y,2}+7.49).F_{z,2}+(COP_{y,4}+7.49).F_{z,4}=0 \quad (3)$$

Since the local COP from each forceplate is on the same vertical plane as the global origin, $COP_{origin} = M_{origin}/F_{z,total}$. Hence,

$$COP_{x,origin} = \frac{[COP_{x,1}.F_{z,1}+(COP_{x,2}-33.75).F_{z,2}+(COP_{x,4}+33.75).F_{z,4}]}{(F_{z,1}+F_{z,2}+F_{z,4})} \quad (4)$$

$$COP_{y,origin} = \frac{[(COP_{y,1}+(46.55+7.49)).F_{z,1}+(COP_{y,2}+7.49).F_{z,2}+(COP_{y,4}+7.49).F_{z,4}]}{(F_{z,1}+F_{z,2}+F_{z,4})} \quad (5)$$

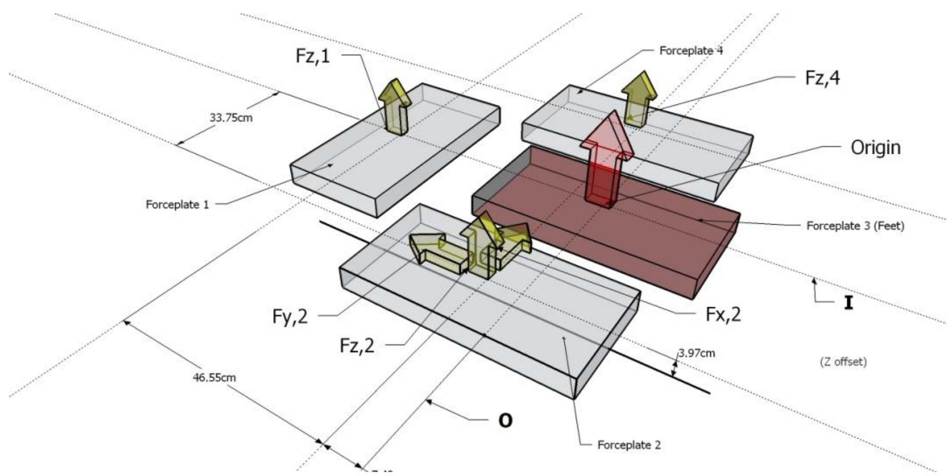

Figure 31: Four force plate setup with relative distances (in cm).

2 Shear force contribution to upper limb COP as measured by forceplate

This section describes the vertical load and shear components to upper limb COP as measured by the ground reaction force.

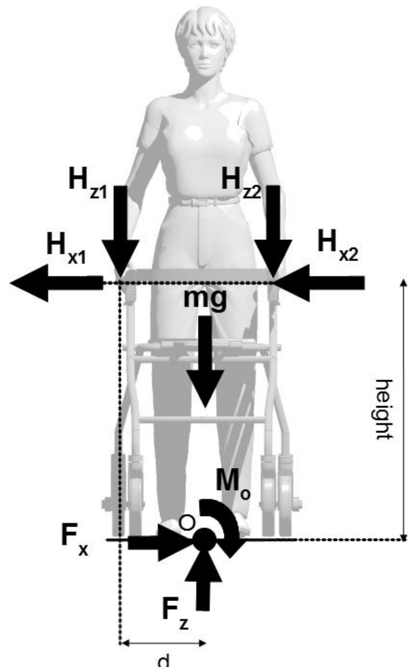

Figure 32: Free body diagram of upper limb forces exerted on walker and ground reaction force.

Figure 32 illustrates free body diagrams using the forceplate setup to measure COP viewed from the frontal plane. The diagram depicts the upper limb forces acting at the handles of the rollator having vertical H_{z1},H_{z1} and shear H_{x1},H_{x2} components. The forceplate reaction forces at the ground underneath the wheels of the rollator are represented by shear (F_x), vertical (F_z), and moment (M_o) components. The weight of the rollator, acting at the midline is shown by the vector, mg.

Using equations of motion for static equilibrium:

$$\sum F_x : -H_{x1} - H_{x1} + F_x = 0 \qquad (6)$$

$$\sum F_z : -H_{z1} - H_{z2} + F_z - mg = 0 \qquad (7)$$

$$\sum M_O : M_O - H_{x1}.height - H_{x2}.height - H_{z1}.d + H_{z2}.d = 0 \qquad (8)$$

Rearranging,

$$M_O = (H_{x1} + H_{x2}).height + (H_{z1} - H_{z2}).d = 0 \tag{9}$$

and substituting into Equation A.1,

$$COP_x = \frac{(H_{z1} - H_{z2}).d}{F_z} - \frac{(H_{x1} + H_{x2}).(height + z_{offset})}{F_z} \tag{10}$$

The resulting COP_x has a vertical load and shear component. To estimate the relative contribution of the shear component to the overall COP_x (Equation A.1), sample data from standing balance experiments (Chapter 4) were calculated.

The upper limb shear forces, $H_{x1} + H_{x2}$, are equal to the shear forces measured by the plate, F_x, and act at $height + z_{offset} = 75.0 + 3.97cm$ from the plane of the floor. The vertical load component, $(H_{z1} - H_{z2}).d/F_z$, was calculated by taking the difference if the total COP_x, measured by the 3 forceplate setup (as described in the previous section (A.1)), and the shear force component. RMS values were calculated over the final 30 seconds of quiet standing with the rollator under Normal (Norm) and Increased Challenged (IC) conditions (for details, see Chapter 4.2.1).

Across the 11 participants tested, the mean (±SE) RMS shear contribution was calculated to be 0.174±0.028 cm for the Norm condition, and 0.349±0.036 cm for the IC condition (see Figure A.3). In comparison, the RMS values of the vertical load contribution were 0.315±0.045 cm and 0.938±0.128 cm for the Norm and IC conditions, respectively. Considering the task-related effects, it appears that the vertical loading (+0.5892) plays a larger role than the shear component (+0.1416).

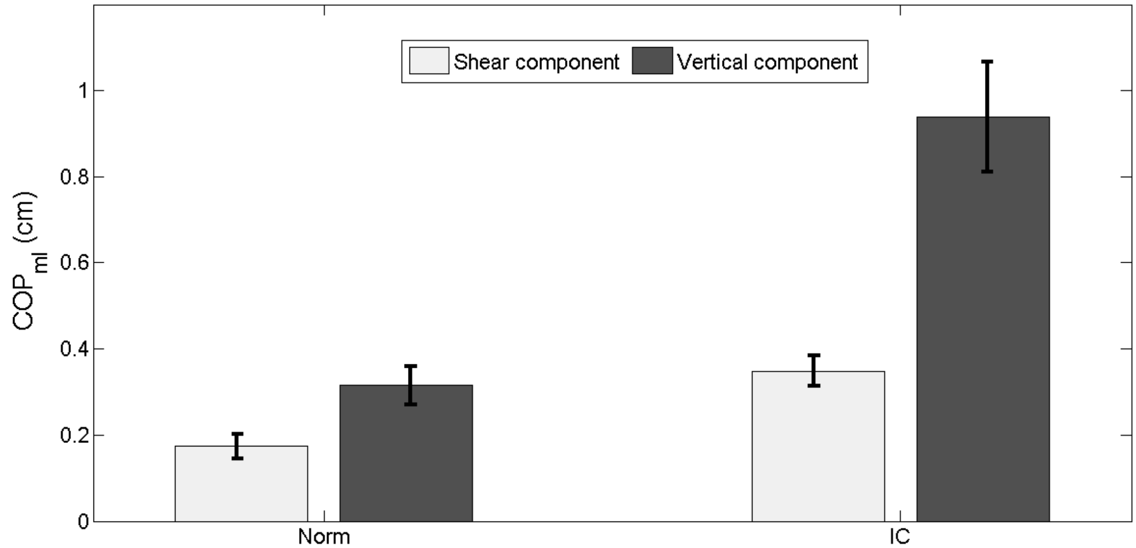

Figure 33: Task-related effect on shear (light bars) and vertical loading (dark bars) components of COP_{ml} in quiet standing. Norm = Normal stance condition; IC = Increased Challenge condition (for details, see Section 4.2.1)