Perturbation Evoked Balance Control Reactions in Individuals with Stroke

by

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Abstract

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Individuals with stroke suffer from impaired balance that increases their risk of falling. Controlling reactive balance is essential to maintaining stability. The objective of the first study was to identify the role of pre-perturbation stance asymmetry on limb preference for reactive stepping in healthy young adults. This study demonstrated that steps taken with a pre-loaded limb are short, directed laterally and have a rapid swing time. The objective of the second study was to investigate the challenges of reactive stepping among individuals with stroke. This study demonstrated that participants primarily execute reactive stepping with their non-paretic limb, although those steps are highlighted by delays in timing and increased incidence of multiple stepping compared to healthy controls, even though all participants had very good clinical balance scores. Outcomes from this thesis present the need for improved clinical assessment of reactive balance control to help reduce the incidence of falling following stroke.
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List of Abbreviations

ABC – Activity-specific Balance Confidence scale
AP – Antero-Posterior
APA – Anticipatory Postural Adjustment
ANOVA – Analysis of Variance
BBS – Berg Balance Scale
BOS – Base of Support
BW – Body Weight
CIS – Change-in-Support
CNS – Central Nervous System
CMSA – Chedoke-McMaster Stroke Assessment
COM – Centre of Mass
COP – Centre of Pressure
EMG – Electromyography
FO – Foot Off
MG – Medial Gastrocnemius
ML – Medio-Lateral
MOCA – Montreal Cognitive Assessment
NIHSS – National Institutes of Health Stroke Scale
PC – Patient Case (x)
TA – Tibialis Anterior
1.0 Introduction

In Canada, stroke is the leading cause of neurological disability (Heart & Stroke Foundation of Canada, 2009). Despite increasing survival rates following stroke, the incidence has remained relatively unchanged, resulting in an increased prevalence of stroke survivors and further need for rehabilitative care. The majority of stroke survivors are left with some persisting disability (Nyberg & Gustafson, 1996). More generally, many stroke survivors often have difficulty in conducting activities of daily living and are limited in their mobility and ability to functionally ambulate in their environment (Nyberg & Gustafson, 1997).

In particular, these individuals are at an increased risk of falling due to the use of unsafe gait patterns and the presence of increased sway during quiet stance, which are caused by an array of secondary deficits following stroke (Mion et al., 1989; Nyberg & Gustafson, 1995; Nyberg & Gustafson, 1996). Due to their increased fall risk and decreased bone density on the paretic side, individuals with stroke have a four-fold increase of hip fracture compared to healthy older adults (Ramnemark, Nyberg, Borssen, Olsson, & Gustafson, 1998). However, while a number of studies have focused on predictors of fall risk, far fewer have detailed the underlying causes of falls and the specific challenges to the control of fall prevention strategies that might be used by individuals with stroke.

Falls arise from some source of instability; either an internal error or an external perturbation that requires a balance response (Maki & McIlroy, 2003). Reactive balance control strategies are the primary means of defense in response to unpredictable perturbations to balance and are often utilized in the everyday environment, where balance
disturbances can frequently occur (Maki & McIlroy, 1997; Maki & McIlroy, 1999). As a result, the ability to control such reactive responses is essential to maintaining upright stability in everyday life. There are two primary classes of reactive response: fixed support and change in support (CIS) reactions. Both classes are characterized by exceptionally rapid onset times (i.e. significantly faster than volitional movement) while preserving accurate muscle activation patterns and scaling of muscle activity (Horak & Nashner, 1986; Maki & McIlroy, 1997; McIlroy & Maki, 1995b; Nashner, 1977; Nashner, Woollacott, & Tuma, 1979).

Fixed support strategies, in which the base of support (BOS) remains stable and where generation of torques at the ankles and hip counteract the perturbation (Horak & Nashner, 1986; Nashner, 1977; Nashner et al., 1979), have been studied in individuals with stroke. These studies have demonstrated challenges in control, highlighted by delayed paretic limb muscle onset latencies and reduced torque generation on the paretic side compared to the non-paretic limb, as well as altered timing in the normal distal to proximal muscle activation pattern that is typically associated with fixed support responses (Dickstein, Hocherman, Dannenbaum, & Pillar, 1989; Dickstein, Dvir, Jehousa, Rois, & Pillar, 1994; Dietz & Berger, 1984; Garland, Stevenson, & Ivanova, 1997; Holt, Simpson, Jenner, Kirker, & Wing, 2000; Horak, Esselman, Anderson, & Lynch, 1984). Compromised fixed-support reactions result in increased instability and a consequent reliance on CIS responses (Maki & Whitelaw, 1993).

CIS strategies are characterized by control of the upper or lower limbs to rapidly establish a new base of support (Maki & McIlroy, 1997). For example, the use of a lower limb for a stepping response requires sophisticated lower limb control in order to rapidly unload the stepping limb, promote stability on the stance limb and safely guide the swing
limb trajectory in a rapid timeframe. Despite the potential importance of compensatory stepping reactions for individuals with stroke as a means to promote upright stability, there have been no studies that have attempted to investigate the challenges of controlling these responses after stroke.

The overarching objective of this thesis is to identify the limb preference as well as the temporal and spatial deficiencies that individuals with stroke may face when attempting to execute compensatory stepping reactions. Successful recognition of risk factors and abnormalities associated with the selection of compensatory reactive balance strategies could help provide the basis for clinical testing. In turn, this information can assist in identifying individuals with stroke who are at an increased risk for falling. In addition, this research has the potential to guide novel clinical therapies that attempt to specifically improve reactive, dynamic, balance control reactions.
2.0 Background

2.1 Epidemiology of stroke

Stroke is the fourth leading cause of death and the leading cause of neurological disease in Canada (Heart & Stroke Foundation of Canada, 2009). Every year, there are between 40,000 and 50,000 new strokes reported and close to 300,000 stroke survivors in Canada, with an annual economic cost of approximately $2.7 billion (Heart & Stroke Foundation of Canada, 2009). Improvements in acute care following stroke have aided in increasing the survival rate, but, in turn, has also increased the number of individuals requiring rehabilitation following stroke. Of every 100 Canadians who have a stroke, 15 die, 10 recover completely, 25 recover with minor impairment, 40 are left with moderate impairment and 10 are so severely disabled that they require long-term care (Heart & Stroke Foundation of Canada, 2009). Stroke is the largest primary diagnosis for rehabilitation in Canada and requires the third longest length of stay in hospital (18 days: Canadian Institute for Health Information, 1999). Rehabilitation following stroke requires a tremendous contribution from the health care system, considering that over half of stroke survivors will live for at least five years after their stroke. In addition, multiple factors relate to the prognosis of balance recovery following stroke, including: age, lesion size/location, and severity of neurological deficits & neglect (Hyndman, Ashburn, & Stack, 2002; Mayo, Korner-Bitensky, & Kaizer, 1990; Nyberg & Gustafson, 1997; Sackley, 1991).

2.2 Post-stroke rehabilitation

Effective rehabilitation programs are a critical component in providing stroke survivors with the opportunity to regain independence and functional abilities. Stroke rehabilitation involves the re-learning of previously learned skills as well as the integration
of new skills and movements to compensate for lost function as a result of the stroke. This process is multifaceted and includes physical, occupational, speech-language and vocational therapies in order to assist individuals proceed with activities of daily life. Following stroke, individuals can present with sensory disturbances, aphasia, anosognosia, neglect, depression, anxiety and motor control deficits (Nyberg & Gustafson, 1997). In particular, inefficient balance control and patient falls have been reported as a significant health risk in the stroke rehabilitation population (Nyberg & Gustafson, 1995). Falls have also been identified as having the greatest potential to lead to secondary injury and decreased balance confidence which can result in diminished overall activity as well as cardiovascular and musculoskeletal health problems (Dromerick & Reding, 1994; Nyberg & Gustafson, 1995).

### 2.3 Post-stroke fall risk is elevated

As a result of the cognitive and physical deficits caused by stroke, individuals often suffer from unsafe gait patterns, confusion and wheelchair confinement – all of which are significant risk factors for falling (Nyberg & Gustafson, 1995). In addition, increased postural sway and increased motor response time to visual stimuli experienced by individuals post-stroke increases fall risk (Nyberg & Gustafson, 1995). Fall rates in stroke rehabilitation wards have been reported between 25-39%, and the number of hospital falls has been shown to be a significant predictor of falls following discharge (Dromerick & Reding, 1994; Nyberg & Gustafson, 1995). Among community dwelling stroke survivors, between 23-73% suffer a fall within a four to six month period following the stroke (Forster & Young, 1995; Jorgensen, Engstad, & Jacobsen, 2002). There is a general overrepresentation of patients with previous strokes among hip fracture patients who have sustained falls. Stroke patients
have also been shown to experience a four-fold increase of hip fractures compared to healthy older adults, associated with their high incidence of falls as well as a loss of bone density on the stroke affected limb (Ramnemark et al., 1998). Individuals who do regain the ability to stand following stroke are typically characterized by having increased postural sway during quiet stance and a weight bearing asymmetry that favours the non-paretic limb, which might be the result of pain, spasticity, sensory loss, muscle weakness or perceptual deficits (Dickstein et al., 1989; Dickstein et al., 1994; Dietz & Berger, 1984; Garland et al., 1997; Holt et al., 2000; Horak et al., 1984). Much of the research investigating fall prevention post-stroke has been focused on predictors of fall risk rather than on the particular strategies used to prevent falling. The following sections will discuss postural control followed by an overview of response strategies used to maintain upright stability in healthy young adults, older adults and individuals with stroke.

2.4 Postural control

Postural control depends on an interaction between musculoskeletal components (joint range of motion and properties of muscle) and neural components (motor control, sensory integration and central nervous system processing) (Shumway-Cook & Woollacott, 2006). Overall body posture involves a coordinated system comprised of external forces acting upon the body, the mechanical forces of the body and neuromuscular forces (Massion, 1994). Integration of these body systems ensures that postural stability is maintained. Postural maintenance, or ‘static’ stability, requires that the centre of mass (COM) either remain within, the base of support (BOS) and is the result of a contribution of reactive and predictive balance control reactions (Maki & McIlroy, 1999; Pai, Rogers, Patton, Cain, & Hanke, 1998). In contrast, postural stabilization, or ‘dynamic’ stability, refers
to the control of upright balance in response to a perturbation, in which the end result is postural maintenance. A system is considered stable if the internal forces are sufficient to oppose the perturbing ones (Bouisset & Do, 2008). Control of the COM is regulated by the visual, vestibular and sensorimotor systems which act synergistically to detect errors between the predicted and actual locations of body position in the environment (Massion, 1994).

In order to maintain stability, the body utilizes both predictive and reactive mechanisms during the course of daily activities. Predictive (i.e. anticipatory) balance control strategies are utilized to compensate for destabilization caused by volitional movement, such as walking, whereas reactive (i.e. compensatory) balance control strategies are utilized when destabilization is caused by an unexpected disturbance to balance (Maki & McIlroy, 1997). Due to the unpredictable nature of the environment, the control of reactive balance control strategies are a critical component to maintaining stability in everyday life.

2.5 Reactive balance control responses

Evaluating the integrity of balance control reactions requires perturbation of the control systems and identification of the strategies utilized to maintain upright stability. Previous research has identified two important classes of balance recovery strategies in response to an unexpected disturbance to balance: fixed BOS reactions, in which the BOS remains constant and change-in-support (CIS) reactions, in which the BOS changes by use of a stepping or reaching response.

Fixed BOS reactions, as a result of a perturbation in the sagittal plane, consist of generating torque at the hip or ankles (but typically a combination of both) to counteract
the direction of perturbation (Horak & Nashner, 1986; Nashner, 1977). Typically the muscle activation is synergistic and sequential depending on the location and direction of perturbation. Nashner (1977) and Horak & Nashner (1986) characterized the sequential activation of lower limb and trunk musculature in response to a posterior platform translation in which gastrocnemius activates 90-100ms following the perturbation, followed by the hamstrings 30ms later and concluding approximately 25ms later with activation of the paraspinal muscles. A similar distal-to-proximal muscle activation pattern is seen in response to anterior platform translations. The use of the ankle strategy to respond to perturbations is typically restricted to conditions in which perturbation amplitude is small and in populations with the ability to generate adequate force at the ankles (Nashner, 1977; Nashner et al., 1979). The hip strategy is typically utilized in situations where the perturbation amplitude is larger or faster and in situations where the support surface is smaller than the length of the feet. Similar to the ankle strategy, following a perturbation that causes forward sway, muscle onset for the hip strategy occurs in approximately 90-100ms in the abdominal muscles, followed shortly after by quadriceps activation (Horak & Nashner, 1986). Fixed support reactions are adaptable to task conditions, such as perturbation velocity, and are not simply ‘all or none’ responses, but rather a continuum from the ankle to the hip strategy (Runge, Shupert, Horak, & Zajac, 1999). For example, as individuals increase their anterior lean angle prior to an anteriorly directed balance perturbation, they gradually switch to using more of a hip strategy and less of an ankle strategy, indicating the adaptability of the response depending on the task conditions (Horak, Shupert, & Mirka, 1989). The sophisticated manner in which each of these responses are activated and executed suggests some central level of organization.
Change-in-support strategies, specifically the compensatory stepping strategy, were previously viewed as only being utilized when the perturbation magnitude forced the COM outside the BOS (Horak & Nashner, 1986; Horak et al., 1989; Shumway-Cook & Woollacott, 2006). However, evidence has demonstrated the CIS reactions are utilized even when the perturbation magnitude is low and the COM remains well within the limits of the BOS (Maki & Whitelaw, 1993; McIlroy & Maki, 1993a; Mille et al., 2003). Importantly, McIlroy & Maki (1993a) demonstrated that the specific instructions given to participants prior to a balance disturbance might alter the response strategy. If participants are told refrain from stepping in response to the perturbation, they utilize a stepping response significantly less frequently than a condition in which they are given no particular instructions about which strategy to use (McIlroy & Maki, 1993a). The importance of compensatory stepping is further highlighted when considering that video footage in geriatric health care facilities demonstrated compensatory stepping behaviour in 32-45% of falls or near-falls (Connell, 1996; Holliday, Fernie, Gryfe, & Griggs, 1990). While compensatory stepping reactions are utilized when perturbation magnitude is low, they are the only means of fall prevention when the perturbation magnitude causes the COM to move outside the BOS.

2.6 Control of change-in-support reactions

CIS reactions are characterized by both the sophisticated spatial control of the limb trajectory as well as the rapid timing of the phases of the response. The ability to control compensatory stepping responses in a rapid manner is critical in ensuring that the stepping response itself does not induce destabilizing forces upon the body. Control of these reactions is regulated by both feedforward and feedback postural strategies.
Feedforward strategies are utilized when a balance disruption can be anticipated (i.e. during volitional movement that is destabilizing, such as gait). Anticipatory postural adjustments (APAs) are one such feedforward strategy that prepares the COM for impending instability by preemptively activating musculature to counteract the balance disturbance. With regards to stepping, APAs are defined as an initial centre of pressure (COP) trajectory directed towards the swing limb in order to propel the COM towards the stance limb, typically accompany volitional stepping responses in the lower limb (Jian, Winter, Ishac, & Gilchrist, 1993). Earlier research suggested that APAs were only utilized during volitional movement and were indicators only of pre-planned movement (McIlroy & Maki, 1993a; McIlroy & Maki, 1995a). However, more recent work has demonstrated that APAs can be present during reactive stepping when demanded by the task conditions (Zettel, McIlroy, & Maki, 2002). This is in accordance with previous studies of upper extremity reactive balance reactions (reach and grasp) which demonstrated that the earliest trajectory of upper limb movement in response to perturbation can be altered according to handhold location (Maki & McIlroy, 1997; McIlroy & Maki, 1995a). In almost all cases, the anticipatory phase is not present when the perturbation is novel (first presented) and tends to appear with repeated exposure to the same stimulus (McIlroy & Maki, 1995a). Even in situations where perturbation direction is unpredictable, participants generally step less frequently and take shorter steps once they gain familiarity with the perturbation (Maki & Whitelaw, 1993). The ability to improve performance following repeated exposure to perturbations could provide the basis for a perturbation-based training intervention to assist with reactive balance control in at risk populations (Jobges et al., 2004; Mansfield, 2007).
Feedback strategies utilize sensory information from visual, vestibular and somatosensory systems to generate an adequate response to the given balance disturbance. Of critical importance for reactive balance control is the speed of response. Reactive stepping is distinguished from volitional stepping by difference in onset latency (approximately 100ms more rapid in reactive stepping) (Burleigh, Horak, & Malouin, 1994; McIlroy & Maki, 1993a; McIlroy & Maki, 1996b). Although the exact mechanisms that control reactive compensatory stepping are not known, the ability to abort a stepping response prior to foot-off suggests that there must be some post-perturbation feedback that allows for ‘on-line’ control of the stepping response (Maki & Whitelaw, 1993; McIlroy & Maki, 1993a). Somatosensory information from the soles of the feet play a critical role in providing feedback regarding the degree of loading on each limb during weight transfer, foot off and foot contact. Perry et al. (2000) used a hypothermic anesthetic to attenuate somatosensory feedback from the soles of the feet and found that, despite instructions not to step, participants increased the frequency of stepping responses as well as the incidence of multiple step responses compared to the condition in which no anesthetic was used. The specific parameters of the central nervous system that control these responses remain unclear but further investigation in at risk populations help to provide further details of the associated mechanisms.

2.7 Compensatory stepping in older adults

Studies that have investigated compensatory stepping responses in older adults have revealed that torque generation at the ankle joint is slower and that compensatory steps are typically executed at lower levels of instability compared to healthy young adults (Jensen, Brown, & Woollacott, 2001; Mille et al., 2003). Of particular importance are the
increased frequency of multiple stepping responses, the use of upper extremity responses and the loss of balance, which indicate difficulty controlling the initial compensatory step (Luchies, Alexander, Schultz, & Ashton-Miller, 1994; McIlroy & Maki, 1996a; Wolfson, Whipple, Amerman, & Kleinberg, 1986). The incidence of multiple stepping typically indicates an insufficient initial step which failed to capture the COM within the BOS, thus resulting in secondary lateral instability that requires a second step (McIlroy & Maki, 1996a). In addition, the onset timing associated with the response is similar in healthy young adults and older adults, suggesting that difficulty is typically associated with the swing and landing phase of the compensatory response rather than the onset (McIlroy & Maki, 1996a; Thelen, Wojcik, Schultz, Ashton-Miller, & Alexander, 1997). Given that stroke primarily effects elderly adults, the elevated fall risk presented by this population only enhances the need for research investigating compensatory response in those with stroke.

2.8 Reactive balance responses in individuals with stroke

To date, research has only attempted to investigate fixed support strategies in individuals with stroke. This work has demonstrated that individuals with stroke tend to have delayed initial muscle onset latency (120-380ms following perturbation, compared to 90-100ms following perturbation in healthy young adults), reduced paretic muscle activity, co-contraction within the lower limb that leads to increased instability, and reliance on non-paretic limb musculature to compensate for the inability to generate sufficient levels of torque in the paretic limb when compared to healthy older adults (Dickstein et al., 1989; Dickstein et al., 1994; Dietz & Berger, 1984; Garland et al., 1997; Holt et al., 2000; Horak et al., 1984). Ikai et al. (2003) demonstrated that in response to anterior and posterior platform translations, individuals with stroke tended to have delayed response onset
latency as well as diminished response strength on the paretic side. The timing between distal to proximal activation of lower limb muscles also appears to be disturbed following stroke (Badke & Duncan, 1983; Di Fabio, Badke, & Duncan, 1986). Marigold & Eng (2006) attempted to distinguish between the kinematic and temporal profiles in response to platform perturbations of fallers and non-fallers in the post-stroke population. They found that, immediately following a platform translation, intralimb coupling duration (distal to proximal muscle activation) was delayed in fallers and that fallers demonstrated a higher trunk velocity as well as a greater change in paretic ankle angle compared to non-fallers. These investigations have demonstrated that individuals with stroke present with a number of abnormalities in response to a balance disturbance that may be associated with their increased risk of falling.

2.9 Rationale and objectives

Despite the importance of CIS responses, there have been no studies to date that have attempted to characterize them in individuals with stroke. Previous work has demonstrated great challenges to reactive balance control when utilizing fixed support strategies, but there has only been anecdotal evidence that simply noted that stepping responses are executed by individuals with stroke (Marigold & Eng, 2006). Harburn et al. (1995) recognized, in their postural stress test, that balance reactions that require a multiple step response in individuals with stroke put them at an increased risk of falling, but no further research was done to characterize the associated response strategies. Given that balance control in individuals with stroke is impaired and fall risk in this population is very high, greater emphasis must be placed on understanding the reactive control strategies associated with their balance responses to perturbations.
The main objective of this thesis was to investigate and characterize the potential challenges of executing compensatory stepping responses among individuals with stroke. This thesis is presented in a chapter format which presents two specific studies.

The objective of the first study was to identify the roles of pre-perturbation stance symmetry and limb dominance on limb preference for compensatory stepping in healthy young adults. Due to the stance asymmetry found in individuals with stroke, we attempted to simulate a similar degree of pre-perturbation stance asymmetry by forcing healthy participants to step with an asymmetrically loaded lower limb. It was hypothesized that: 1) as loading of the preferred stepping limb increases, the frequency of stepping with that limb would decrease; 2) when the preferred stepping limb is loaded and the non-preferred stepping limb has an environmental constraint placed around it, compensatory steps would be executed by the loaded limb and would be characterized by delayed time to foot-off, faster swing time, larger APA magnitude, longer APA duration and shorter step length when compared with the equal weight distribution, no constraint condition.

The objective of the second study was to investigate compensatory stepping responses in individuals with stroke. In particular, the intention was to inspect whether individuals with post-stroke limb dyscontrol utilize compensatory steps, the limb choice for those steps (paretic or non-paretic) and the temporal and spatial features of those reactions. It was hypothesized that: 1) there would be a reliance on the non-paretic limb for stepping, despite the limb loading asymmetry that typically favours the non-paretic limb; 2) steps would be delayed due to the large anticipatory phase required to maintain lateral stability when stepping with a loaded limb.
Currently, reactive balance control is not a specific measure during clinical assessment of individuals with stroke and, consequently, rehabilitative therapies may not adequately reflect the need for reactive balance training. The outcomes of this thesis may have implications on both the way that balance control is assessed as well as on the clinical therapies used to improve reactive balance control in individuals with stroke.
3.0 Experiment 1: Characterizing the determinants of limb preference for compensatory stepping in healthy young adults

3.1 Introduction

Compensatory balance control is a critical component of the balance recovery system in response to a perturbation. Specifically, change-in-support (CIS) reactions, or reactions that involve movement of the limbs to adjust the base-of-support (BOS), are frequently observed in response to balance disturbances (Maki & McIlroy, 1997). It was previously hypothesized that CIS reactions such as stepping and reaching were responses of last resort, used only after feet-in-place reactions would no longer maintain the centre of mass (COM) within the BOS (Horak & Nashner, 1986; Horak et al., 1989). However, more recent evidence has shown that CIS responses are used frequently in situations where perturbations are of low magnitude and the limits of the COM are well within the initial BOS (Maki & Whitelaw, 1993; McIlroy & Maki, 1993a; Mille et al., 2003).

CIS stepping responses are distinguished from volitional stepping by their rapid onset latencies and time to foot-off (Maki & McIlroy, 1997). In order to achieve more rapid response times, compensatory steps are often, but not always, executed in the absence of an anticipatory postural adjustment (APA) (McIlroy & Maki, 1993a; McIlroy & Maki, 1995b). Such APAs are characterized by an initial medio-lateral centre of pressure (COP) shift towards the swing limb that acts to drive the COM towards the stance limb and promote stability at foot-off (Jian et al., 1993). Pre-planned steps, such as volitional steps, are frequently accompanied by an APA (Brunt et al., 1991; Carlsoo, 1966) and evidence has demonstrated that humans have a dominant limb, to execute these steps (Gentry &
Gabbard, 1995). However, more recent work has suggested that APAs may be utilized by the CNS during compensatory stepping to adapt for increased task challenge and act to harness the COM in conditions with increased ML instability (Zettel et al., 2002). There is currently little understanding of the determinants of limb preference for reactive compensatory stepping. Gaining further insight into limb preference for compensatory stepping is particularly important when attempting to apply knowledge to individuals with unilateral limb dyscontrol.

The focus of the current study is to understand the determinants of limb preference for compensatory stepping. Unlike the upper limb, where the limbs can function independent of each other, lower limb control must be integrated between both limbs to maintain stability. For the lower limb, limb dominance refers to the limb that is used for mobilization (i.e. stepping, kicking a ball) whereas non-limb dominance is the limb used for support (Sadeghi, Allard, Prince, & Labelle, 2000). In healthy individuals, we contend that the main factors for task-related evidence of limb preference for compensatory stepping are limb dominance and pre-perturbation limb loading. We suggest that while limb dominance would influence limb preference for a compensatory step while standing with equal weight distribution over each limb, stance loading will be the primary determinant of stepping limb preference when loading is asymmetric.

If, as suggested, unloading a limb leads to increased reliance on performing a compensatory step with that limb, then it should also be associated with a faster time to foot-off because less unloading time is associated with the step. Unloading one limb should either reduce the frequency of APA occurrence or attenuate the APA size because the COM
is laterally shifted towards the stance limb prior to the perturbation. In contrast, forced stepping with the loaded limb, which counters limb preference, would make rapid compensatory stepping more challenging and is a feature we would expect to see in populations with unilateral limb deficits. Previous work has demonstrated that clearance of an obstacle when completing a compensatory step generates a large amplitude APA to promote medio-lateral stability compared to the control condition in which the APA amplitude was either attenuated or non-existent (Zettel et al., 2002). Stepping with the loaded limb would be highlighted by greater challenges to control due to the need to regulate the displacement of the medio-lateral COM to maintain lateral stability and would likely result in a delayed time to foot-off.

It is hypothesized that: 1) the dominant limb will be the preferred limb for stepping when pre-perturbation stance loading is symmetrical; 2) as loading of the preferred stepping limb increases, the frequency of stepping with that limb will decrease; 3) when the preferred stepping limb is loaded and stepping with the preferred (loaded limb) is ‘forced’, reactions will be characterized by larger APA magnitude, longer APA duration, delayed time to foot-off, faster swing time and shorter step length when compared with the equal weight distribution, no constraint condition.

### 3.2 Methods

#### 3.2.1 Participants

Ten healthy participants, characterized in Table 1 volunteered to complete this study. Informed consent was obtained from all participants and the protocol was approved by the institution’s research ethics board. All participants reported that they were free of neuromuscular and neurological impairments. Participants completed the Waterloo
Handedness and Footedness Questionnaire (Elias, Bryden, & Bulman-Fleming, 1998) to determine their dominant lower limb.

Table 1. Participant characteristics. Values shown are frequency counts or means ± standard deviation (range)

<table>
<thead>
<tr>
<th>Characteristic</th>
<th>Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Men/Women</td>
<td>5/5</td>
</tr>
<tr>
<td>Age (years)</td>
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<tr>
<td>Weight (kg)</td>
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<td>Height (cm)</td>
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Lower-limb dominance

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<td>Right</td>
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Preferred stepping limb

<table>
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<tr>
<td>Right</td>
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</table>

*Limb dominance was assessed using the Waterloo Handedness and Footedness Questionnaire (Elias et al., 1998).

3.2.2 Protocol

Participants stood in a standardized foot position with one foot on each force plate [heel centers 0.17m apart with and a 14° angle between the long axes of the feet (McIlroy & Maki, 1997)]. They wore earplugs so they were unable to hear cues to the upcoming perturbation, and were provided with visual feedback of their weight distribution over each limb. Anterior compensatory steps were evoked using a lean & cable release system similar to those used in previous studies (Do, Breniere, & Brenguier, 1982; Hoshiyama, Watanabe, Kaneoke, Koike, & Takahashi, 1993; Thelen et al., 1997) (Figure 1).
Figure 1. Lean and cable-release perturbation set-up. Participants leaned forward approximately 10cm from neutral stance. Participants stood on two separate force plates prior to the perturbation and foot-fall was captured by a force plate located in front of the participant. A load cell mounted in the platform was used to detect onset timing of the perturbation.

The participant wore a harness with a cable attached posteriorly at the height of the xyphoid process. When attached to the cable, participants were instructed to lean forwards from their ankle joint by approximately 10cm (the slack in the cable) from their neutral stance position so that support for the forward lean was maintained by the cable.
Perturbations were evoked by manual release of a pin which attached the cable to the balance platform and participants were informed that they “may or may not receive a perturbation when leaning forwards” and in the event of a perturbation to “respond as safely as possible to maintain balance”. Participants wore a full body safety harness fixed to the ceiling to prevent a fall to the floor in the event of a failure to recover balance effectively. There were three separate task conditions that were presented as separate blocks of trials: 1) symmetric weight distribution; 2) asymmetric weight distribution; 3) constrained, asymmetric weight distribution. There were 10-30s between each perturbation and a short rest break between each trial block. Catch trials (25% of total trials), in which the participant was required to lean forward, but no perturbation was delivered, were also included in the protocol to increase unpredictability.

In the symmetric weight distribution task, five perturbation trials were completed in which participants were instructed to maintain an equal weight distribution over each limb prior to the perturbation. The ‘preferred’ stepping limb for subsequent trials was defined as the limb that the participant stepped with most frequently out of the five trials; the other limb is referred to as the ‘non-preferred’ limb. In the asymmetric weight distribution task, participants vertically loaded their preferred limb from 30-70% body weight in 5% increments and the order of trials was randomly presented. Participants completed three trials for each vertical load less than 50% body weight on the preferred stepping limb and five trials for each load greater than 50% body weight on the preferred stepping limb. In the constrained, asymmetric weight distribution task a foam barrier was placed in front of the non-preferred limb. Participants stood with either 30% or 70% body weight over their preferred limb in a randomized order and five perturbation trials were completed in each of
these conditions. The barrier was 55cm high and was used to promote stepping with the preferred limb. Participants were instructed to “respond as safely as possible to maintain balance without hitting the barrier.” For safety, the barrier was easily displaced if contacted.

3.2.3 Data acquisition

Two force plates (50cm long and 25cm wide; Advanced Medical Technology Inc., Watertown, MA, USA) were mounted in parallel within the platform and a third force plate (51cm long and 46cm wide; Advanced Medical Technology Inc., Watertown, MA, USA) was located in front of the participant to capture footfall. Motion analysis was conducted using a 6-MX camera, three-dimensional capture system sampled at 60Hz (VICON, Motion Analysis, Santa Rosa, CA, USA). Reflective markers were affixed to the skin with double-sided tape at anatomic landmarks bilaterally on the greater trochanter, lateral knee, lateral malleolus, heel and 1st distal metatarsal. An in-line load cell was fixed within the release cable and recorded the participant’s lean force and provided the indication of perturbation onset time. Surface electromyography (EMG) was recorded bilaterally from the medial gastrocnemius (MG) and tibialis anterior (TA). EMG signals were pre-amplified at a gain of 500 and then amplified by 1000. Silver/silver chloride (Ag/AgCL) electrodes were fixed 1cm apart over the muscle belly. Electrode sites were cleaned with an alcohol wipe and abraded with abrasive cream. Sites were shaved if necessary. Real-time EMG signals from MG and TA were monitored prior to each perturbation to ensure that participants had no anticipatory muscle activity (e.g. participants were solely relying on the cable to maintain their forward lean). Force plate and EMG signals were all sampled at 1000Hz.
Figure 2. Summary of analysis method for locating characteristics of the APA (if present) and foot-off timing. The upper MLCOP tracing outlines how APA onset, duration and magnitude were determined and the lower Fz (vertical load) tracing for the stepping limb demonstrates how foot-off time was located.

3.2.4 Data analysis

Frequency of limb choice for a stepping response was determined by using visual inspection and confirmed with measures of vertical loading from the force plates. Step-onset was defined as the time when the medio-lateral (ML) COP deviated by 4mm from the pre-perturbation mean ML COP (calculated for the 200ms prior to the perturbation)
(McIlroy & Maki, 1999). Step onset either represents an APA if the initial MLCOP deviation is
towards the swing limb, or the onset of limb unloading if the initial MLCOP deviation is
towards the stance limb. If present, APA duration was determined as the time between APA
onset and the time at which the MLCOP was equal to the MLCOP at APA onset (represented
in Figure 2). APA magnitude was the area under the MLCOP curve for the duration of the
APA. Foot-off time was the time from perturbation onset at which the vertical load on one
force plate was less than 1% of body weight and was expressed relative to perturbation-
onset time (i.e. when the load on the load cell was < 2% of body weight). Swing time was
the foot off time subtracted from the time that the first foot made contact on the anterior
force plate (i.e. > than 2% of body weight)

Step trajectories were determined from the Vicon motion capture system. Onset of
movement was defined as the point in which the heel marker of the stepping limb
advanced anteriorly two standard deviations from its resting neutral position. Step
completion was defined as the first point in which the anterior stepping trajectory
plateaued for greater than 100ms. Antero-posterior (AP) & ML step trajectories were
determined by the calculating the distance traveled in each of the AP & ML planes between
step-onset and step completion times.

3.2.5 Statistical Analysis

Statistical analysis focused on conditions in which pre-perturbation vertical loading
on the preferred stepping limb was greater than 50% of body weight. Repeated measures
analysis of variance (ANOVA) was used to analyze the effect of increased load on the
preferred limb (55% body weight, 60% body weight, 70% body weight with barrier blocking
non-preferred limb) compared with symmetrical loading, on the primary dependent
variables (step onset time, time to foot-off, swing time, APA duration, APA magnitude, ML step trajectory and anterior step trajectory). Tukey-Kramer post-hoc analysis was conducted to determine level of significance between tasks conditions. Statistical significance was set at $\alpha = 0.05$.

### 3.3 Results

All participants completed the protocol with no complications. The preferred compensatory stepping limb (the limb used to step with more frequently in the symmetric stance trials) was the dominant limb, as measured by the Waterloo Footedness and Handedness Questionnaire, for 7/10 participants. Of the seven participants who used their dominant limb as their preferred stepping limb, five stepped with their dominant limb in 100% of trials and the remaining two stepped with their dominant limb in 60% of trials. All three participants whose preferred compensatory stepping limb was not their dominant limb stepped with their non-dominant limb in 100% of the trials from symmetric stance. Of these three participants, one self-reported that martial arts might have trained the dominant limb/preferred stepping limb discrepancy, one had a limb dominance discrepancy between upper and lower limbs (i.e. right hand/left leg dominant) and one did not present with any features that might explain the divergence between preferred compensatory stepping limb and dominant limb.

During the anterior lean, the mean loading on the cable across all participants and conditions was $11.1\pm1.5\%$ body weight and mean angle at the ankle joint was $9.2\pm1.6$ degrees. Conditions in which steps with the preferred limb were taken in less than 20% of the trials were not analyzed further (65% & 70% body weight over the preferred limb, with no constraint which had 7/50 & 6/50 trials with a preferred limb step, respectively). Values
Table 2: Mean spatiotemporal characteristics for steps with the preferred limb across limb loading conditions. Loading conditions in which the preferred limb was used to step in less than 20% of trials (65% & 70%, no constraint) are not included. Values presented are means ± standard deviation

<table>
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<th>30%</th>
<th>40%</th>
<th>45%</th>
<th>50%</th>
<th>55%</th>
<th>60%</th>
<th>70%*</th>
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<tr>
<td>Actual vertical load on preferred limb (%BW)</td>
<td></td>
<td></td>
<td></td>
<td>49.9±0.01</td>
<td>54.4±0.01</td>
<td>59.4±0.01</td>
<td>69.8±0.01</td>
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<td>Frequency of non-preferred limb step (%)</td>
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<td></td>
<td>17±4</td>
<td>8±3</td>
<td>55±5</td>
<td>71±5</td>
<td>8±3</td>
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<td>40</td>
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<td>Step onset time (ms)</td>
<td>53±30</td>
<td>89±30</td>
<td>90±18</td>
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<td>Foot-off time (ms)</td>
<td>282±42</td>
<td>293±53</td>
<td>299±44</td>
<td>337±104</td>
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<td>304±27</td>
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<td>Swing time (ms)</td>
<td>189±33</td>
<td>158±32</td>
<td>144±35</td>
<td>155±37</td>
<td>154±40</td>
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<td>APA duration (ms)</td>
<td>86±42</td>
<td>123±42</td>
<td>123±44</td>
<td>129±75</td>
<td>116±57</td>
<td>143±50</td>
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<td>APA magnitude (mm∙ms)</td>
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<td>0.84±0.92</td>
<td>1.40±1.41</td>
<td>1.46±1.66</td>
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<td>Anterior step displacement (mm)</td>
<td>508±106</td>
<td>494±111</td>
<td>469±99</td>
<td>511±102</td>
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<td>466±109</td>
<td>378±144</td>
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APA = anticipatory postural adjustment; *The preferred limb was constrained in this condition

presented in the following sections are means ± standard deviation. Repeated measures ANOVA revealed significant main effect differences between tasks on medio-lateral step trajectory ($F_{3,18} = 32.38$, $p<0.0001$) and anterior step distance ($F_{3,18} = 8.35$, $p=0.0011$).
3.3.1 Limb preference during symmetric weight distribution

During the symmetric stance condition, the mean vertical loading across participants on the preferred limb was 49.9±0.008% body weight. Under such conditions, participants stepped in 97% (57/59) of trials and of the trials with a step, 91% (52/57) of trials were taken with the preferred limb. Due to the limited frequency of steps with the non-preferred limb, the analysis of the stepping characteristics for this task condition was limited to the 52 steps taken with the preferred limb.

Mean step onset time was 102±33ms. Mean time to foot-off and swing time was 337±104ms and 155±37ms, respectively. APAs were present in 44% (23/52) of steps and mean APA magnitude and APA duration were 1.46±1.6mm·ms and 129±75ms, respectively. Mean AP and ML (+ = lateral, - = medial) step placement locations were 511±102mm and 16±18mm, respectively.

3.3.2 Limb preference during asymmetric weight distribution

Participants were able to accurately adapt to the asymmetric limb loading conditions from 30%-70% body weight over the preferred limb (Actual loading values at each task condition are displayed in Table 2). Overall, the frequency of stepping with the preferred limb decreased as vertical loading on that limb increased (Figure 3). In all conditions where preferred limb loading was either 50% body weight or less, step frequency with the preferred limb was greater than 80%. In conditions where preferred limb loading was greater than 50% body weight, the frequency of stepping with the preferred limb decreased and frequency of stepping with the non-preferred limb increased (Table 2). Conditions in which steps with the preferred limb were taken in less than 20% of the trials were not analyzed further (65% & 70% body weight over the preferred limb, with
Figure 3. Relationship between increasing vertical load and frequency of steps taken by the preferred limb averaged across participants. Values shown are means with standard error bars. Dashed line indicates symmetric stance position. When loading over the limbs was symmetric (vertical dashed line) there was a preferred limb for compensatory stepping. In general, as vertical loading on the preferred limb increased, the frequency of stepping with the preferred limb decreased.

no constraint which had 7/50 & 6/50 trials with a preferred limb step, respectively). To compare step characteristics associated with different pre-perturbation stance loadings, step characteristics were compared between preferred limb steps from the symmetric weight distribution condition to each of the asymmetric weight distribution loads in which the preferred limb was loaded at greater than 50% body weight. There were no significant differences between primary outcome measures in the comparison of steps taken from
symmetric stance to those taken when the preferred limb was loaded at 55% body weight. Steps taken with the preferred limb when loaded at 60% body weight had non-significant trends towards an increased lateral step trajectory \((p = 0.065)\) and faster swing time \((p=0.071)\) than steps taken with the preferred limb from symmetric stance.

**Figure 4** highlights the increasing lateral step trajectory as vertical loading on the stepping limb increased across task conditions. There were no significant differences in characteristics of the APA from the asymmetric task conditions to the symmetric task condition. The mean time to foot-off and swing time, mean AP and ML step trajectories and APA characteristics for the preferred limb, by weight distribution are located in **Table 2**.

Figure 4. Lateral foot placement for steps taken only with the preferred limb compared across amplitudes of pre-perturbation vertical load. Values shown are means with standard error bars. Dashed box indicates condition in which non-preferred limb was obstructed. As vertical loading on the preferred limb increased, steps were directed more laterally.

3.3.3 Forced stepping with the loaded limb
When the non-preferred limb was constrained with a barrier and the preferred limb loaded to 70% body weight, a compensatory step was taken in 98% (48/49) of trials. Of those steps, 90% (43/48) were taken with the preferred, loaded limb. The actual mean vertical load under the preferred limb in this condition was 69.8±0.01%body weight. Step-onset time, foot-off time and swing time for steps with the preferred limb in this condition were 93±49ms, 295±69ms and 120±43ms, respectively.

Figure 5. Footfall location for steps taken with the preferred limb (right limb, in figure). Values shown are means with standard error bars. Pre-perturbation vertical loading over each limb is indicated by the number (in %body weight) displayed on each foot. Steps taken from symmetric stance (centre) had a long step length and were directed anteriorly. Steps taken with an unloaded limb (left) had a long step length and were directed slightly medially. Steps taken with a loaded limb, with the unloaded limb obstructed (right) had a short step length and lateral step trajectory.

Compared to steps taken with the preferred limb when the limbs were symmetrically loaded (50% condition), there were significant differences in anterior step
distance (p=0.0005) and ML step distance (p<0.0001) and a non-significant trend towards faster swing time (p=0.057) and shorter time to foot-off (p=0.081). Figure 5 highlights the different mean preferred limb stepping locations when it was loaded at 30%, 50% and 70% of body weight. Mean temporal and spatial values for steps taken in this condition with the preferred limb can be found in Table 2.

3.4 Discussion

The findings from this study emphasize that the self-perceived dominant limb, as measured by the Waterloo Footedness Questionnaire is likely to be the preferred stepping limb when executing compensatory balance reactions. The limb dominance effect on compensatory stepping limb choice persisted even when the preferred limb was loaded at 55% body weight and to a lesser extent when the preferred limb was loaded at 60% body weight, despite these steps having a shorter swing time and a more lateral step trajectory than steps taken from symmetric stance. Although steps were not taken by every participant in these conditions, this is interesting to note because it would presumably be less challenging to the system (decreased demand for an APA and rapid response timing) to step with the unloaded non-preferred limb. Despite the shorter swing time, the overall foot contact time from perturbation onset does not change, suggesting that the healthy balance control system has the ability to adapt characteristics of the compensatory response to account for modest changes in stance loading position - all within normal limits of the central nervous system (CNS).

As hypothesized, the frequency of stepping with the preferred limb decreased as loading on the preferred limb increased. Contrary to our hypothesis, there was no trend associating the frequency of APAs and increased loading on the preferred limb. In general,
as stance loading became more symmetric, APA onset time and APA duration increased. In addition, APA magnitude increased as loading on the preferred limb increased. The lack of APAs across all conditions is consistent with previous work which demonstrated that APAs are often absent or truncated during true compensatory stepping responses (McIlroy & Maki, 1993b; McIlroy & Maki, 1995a). However, more recent work has suggested that APAs may be utilized by the CNS during compensatory stepping to adapt for increased task challenge and act to harness the COM in conditions with increased ML instability (Zettel et al., 2002) and although not significant, the increased APA magnitude as stance loading on the stepping limb increased reflects these findings. This suggests a decreased urgency when stepping with the unloaded, preferred limb and the ability of the CNS to modulate the size of the response to the task demand, which has been demonstrated previously in other research (Zettel et al., 2002).

In order to examine the characteristics of a compensatory stepping response taken with a loaded limb, stepping was forced with the preferred limb while vertically loaded at 70% body weight by constraining the unloaded limb. Firstly, it is important to recognize that all participants were able to successfully maintain stability without assistance from the safety harness or spotter during this challenging task. However, in order for participants to complete these steps, they adopted a stepping strategy that resulted in extremely laterally directed, short, fast steps when compared to the steps taken from the symmetric loading position. While healthy individuals were able to safely respond to perturbation that required them to step with a loaded limb, individuals with stroke may face a variety of challenges when attempting to respond to external balance perturbations. Numerous studies have demonstrated during quiet standing, individuals with stroke present with a
weight-bearing asymmetry favouring the non-paretic limb (Bohannon & Larkin, 1985; Dickstein, Nissan, Pillar, & Scheer, 1984; Mizrahi, Solzi, Ring, & Nisell, 1989). Compensatory stepping responses have not been widely investigated in individuals with stroke, but some evidence suggests that the loaded non-paretic limb is the preferred limb for compensatory stepping in this population (Lakhani, Mansfield, Sibley, Mochizuki, & McIlroy, 2008; Marigold et al., 2005). The current study highlights problems that individuals with stroke may face when forced to used a compensatory stepping response with a loaded limb to regain their balance. There are many variables that can determine whether a compensatory step will be successful. Once the foot is off the ground, a successful step is the result of an optimal combination of swing time, step length and step trajectory. In the current study, we identified that when stepping with a loaded limb, swing time is decreased, steps are directed more laterally and step distance is decreased. All of these factors suggest that stepping with a loaded limb leads to a high degree of instability upon weight acceptance. If the step distance is too short and the swing time too quick, the new BOS created by the step might not be large enough to prevent the COM from continuing to fall and may result in multiple stepping or even falls in individuals with a stroke.

The current study was limited by the perturbation system used. Since the cable was attached at a fixed point on the centre of the platform, the cable was slightly off axis, rather than perpendicular, to the participant when they were asymmetrically pre-loaded on either side. This may have introduced a slight ML perturbation that exaggerated the amplitude of lateral instability arising from the challenges from lifting the pre-loaded limb. Systems that allow the attachment point to the platform to move laterally with the participant, thus ensuring that the cable is always attached at a right angle between the participant and the
platform, would solve this problem. In addition, a small group of participants were also able to step with the unloaded limb when it was obstructed by the constraint. Future studies that attempt to block limb use might benefit from using a more rigid barrier that is both visually imposing and does not move easily when contacted to ensure steps are taken with the loaded limb.

3.4.1 Conclusions

The current study demonstrates that there is a limb dominance effect on limb choice for compensatory stepping when stance loading is symmetrical and that this effect persists even when stance loading slightly favours the stepping limb. In addition, when the unloaded limb is constrained and stepping with the loaded limb is forced, the steps are shorter, directed more laterally and have a faster swing time than steps taken from symmetric stance, while the overall foot contact time is kept constant. This study demonstrates that the CNS is able to accommodate for altered task challenge by modulating response characteristics to maintain stability in healthy individuals. Additionally, this study illustrates the challenges that individuals with stroke may face if attempting to step with a dominant, loaded limb. The increased ML instability introduced by creating a short, laterally directed step could increase fall risk in this susceptible population. Future research should attempt to investigate and train the stepping strategy as well as the temporal and spatial characteristics of compensatory stepping individuals with stroke.
4.0 Experiment 2: Compensatory stepping responses in individuals with stroke: case reports

4.1 Introduction

The ability to control upright stability is a critical challenge during recovery from stroke. Loss of postural control is associated with a high incidence of falls, a reduced willingness to walk independently and reduced overall activity (Ramnemark et al., 1998; Vellas, Cayla, Bocquet, de Pemille, & Albarede, 1987). The incidence of falls has been reported as high as 73% in the first six months post-stroke (Forster & Young, 1995; Nyberg & Gustafson, 1995). Consequently, improved understanding of post-stroke changes in balance control has the potential to direct future therapeutic interventions to target these specific deficits.

Detriments to reactive balance control are exposed by examining responses to transient perturbations (Maki & McIlroy, 1997). Such perturbation evoked reactions, while rarely assessed, are thought to be important in the clinical assessment of balance (Harburn et al., 1995; Horak, Wrisley, & Frank, 2009). Perturbation studies have uncovered two classes of these responses: feet-in-place and change-in-support (CIS) strategies (Maki & McIlroy, 1997; Shumway-Cook & Woollacott, 2006). Earlier studies have revealed that, following stroke, feet-in-place reactions are often compromised. Individuals with stroke tend to have a delayed muscle onset latency (2-4 times slower than healthy adults) and co-contraction within the lower limb that leads to increased instability (Badke & Duncan, 1983). In addition, studies have reported reliance on non-paretic limb muscle activation to compensate for the inability to generate sufficient torques in the paretic limb (Di Fabio et al., 1986; Dickstein et al., 1989), as well as delayed and reduced paretic muscle activation
If feet-in-place reactions are compromised, instability increases when faced with a postural disturbance and, as a result, elevates the potential reliance on CIS reactions such as stepping (Maki & Whitelaw, 1993).

CIS strategies involve stepping or reach-to-grasp reactions that increase the size of the BOS to regain stability (Maki & McIlroy, 1997). CIS reactions are the only defense against large-magnitude postural perturbations, but are evoked at low thresholds of instability when there are no instructional or environmental constraints (Maki & McIlroy, 1997). Sophisticated control is required to generate rapid compensatory steps in a spatially precise manner (McIlroy & Maki, 1995b). Compensatory steps can be preceded by anticipatory postural adjustments (APAs) (McIlroy & Maki, 1993b; McIlroy & Maki, 1995b). APAs are characterized by an initial medio-lateral centre-of-pressure (COP) shift towards the swing limb that drives the centre of mass towards the stance limb and promotes stability at foot-off (Jian et al., 1993). The foot (stepping) or hand (grasping) must be placed both precisely and at remarkably rapid speeds; CIS reactions are initiated and executed much faster than voluntary movements (Gage, Zabjek, Hill, & McIlroy, 2007; McIlroy & Maki, 1996b). Older adults frequently rely on CIS reactions (Luchies et al., 1994; McIlroy & Maki, 1996a), although control of these reactions is impaired with aging, which is suggested to be linked to an increased risk of falling (Maki, Holliday, & Topper, 1991; Maki et al., 2001).

Despite the importance of CIS reactions, no studies have described these reactions following stroke. Previous research has highlighted the importance of number of steps taken in response to perturbation (Harburn et al., 1995). Although they emphasized the importance of such behaviours, the study did not detail the specifics of the responses. After
stroke, weakness and dyscoordination of a lower limb can have a profound impact on the strategies and approaches utilized to preserve stability in response to a perturbation. The slowing of movement and challenge of precise foot placement may limit the reliance on the paretic limb; however the challenge in bearing weight through the paretic limb while stepping with the non-paretic limb may also be challenging.

This case series aims to examine specific characteristics of compensatory steps in individuals with stroke. It is hypothesized that: 1) there will be a reliance on the non-paretic limb for stepping, despite the limb loading asymmetry that typically favours the non-paretic limb; 2) steps will be delayed due to the large anticipatory phase required to maintain lateral stability when stepping with a loaded limb.

4.2 Methods

4.2.1 Participants

Four male hemiparetic patients with a recent stroke (>1-6 months post-stroke, see Table 3) and 11 healthy young adults (six women; mean age ± standard deviation: 27 ± 6 years) participated in this study. Informed consent was obtained and the study was approved by the institution’s Research Ethics Board.

Individual patient cases

Participants were recruited from in- and out-patient stroke rehabilitation programs. Those who could stand independently for 60 seconds without an assistive device were eligible. Patients were excluded if they presented with musculoskeletal injury, dementia, or other neurological impairments not caused by the stroke that could affect balance control.
Table 3. Patient case information for each of the four individuals tested. Clinical profile of the four patient cases. CMSA = Chedoke-McMaster Stroke Assessment, NIHSS = National Institutes of Health stroke score, MOCA = Montreal Cognitive Assessment Score, ABC = Activity-specific Balance Confidence score, BBS = Berg Balance Scale.

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</table>

Stroke type and location information was gathered from medical records. Standard clinical measures were obtained to describe the participants’ clinical profile (Table 3). The Chedoke-McMaster Stroke Assessment (CMSA: Gowland et al., 1993) provides information regarding the stage of motor recovery at the time of testing. CMSA scores were obtained for the upper extremities, leg and foot. The National Institutes of Health Stroke Scale (NIHSS: Goldstein, Bertels, & Davis, 1989) describes the severity of the stroke based on the domains of consciousness, vision, sensation, movement, speech and language. The
Montreal Cognitive Assessment (MOCA: Nasreddine et al., 2005) is a cognitive screening test. The Activity-specific Balance Confidence scale (ABC: Powell & Myers, 1995) is a questionnaire assessing confidence in performing various activities without falling or losing balance. The Berg Balance Scale (BBS: Berg, Wood-Dauphinee, Williams, & Gayton, 1989) is a performance based assessment tool that evaluates standing balance during a number of functional activities.

Healthy young adults

Healthy young adult participants were different than those included in experiment 1 and were free of any neurological or musculoskeletal illness or injury that adversely affected balance or mobility. Data from the healthy young adults was used as a reference of the ‘ideal’ characteristics of evoked reactions in response to the applied perturbations. Limb dominance was assessed using the Waterloo Footedness & Handedness Questionnaire (Elias et al., 1998).

4.2.2 Protocol

Perturbation

Forward and backward compensatory steps were evoked using a weight drop system similar to those used in previous studies (Harburn et al., 1995; Luchies et al., 1994; Mansfield & Maki, 2009; Mansfield & Maki, 2009; Mansfield & Maki, 2009; Wolfson et al., 1986). Weights were attached to the front and back of a harness worn around the height of the xyphoid process via pulleys. Weight drops were triggered by a manual button press which released an electromagnet, allowing the load to fall 10cm (Figure 6).
Figure 6. Balance perturbation set-up. Weights (not shown) were attached to the front and back of participants via a system of cables and pulleys. Accelerometers were used to detect the onset timing of the perturbation, and the force plates recorded ground reaction forces.

For patient cases, high-magnitude (3% body-weight backward, 4.5% body-weight forward) and low-magnitude (0.5% body-weight backward, 0.75% body-weight forward) perturbations were applied. Pilot testing revealed that the high-magnitude perturbation consistently evoked stepping responses, whereas the low-magnitude perturbation evoked feet-in-place reactions. To account for the increased instability in those with stroke, perturbation magnitudes used for individuals with stroke were half of those used in healthy young adults (see Table 4). Participants wore a safety harness fixed to the ceiling to prevent a fall to the floor in the event of a failure to recover balance effectively. Spotters stood nearby during testing individuals with stroke.
Measurements

Two force plates (25cm x 50cm; Advanced Medical Technology Inc., Watertown, MA) were mounted in parallel within the platform under each foot. A third force plate (46cm x 51cm) was located behind the participant to capture footfall. Linear accelerometers fixed to each cable recorded perturbation-onset time, which was the time when the acceleration was greater than three standard deviations from the pre-perturbation value. Two digital video cameras provided sagittal and overhead views of stepping responses and were used to record stepping patterns and other behaviours.

Tasks

Participants stood in a standardized foot position with one foot on each force plate (heel centers 17cm apart with and 14° between the long axes of the feet (McIlroy & Maki, 1997)). Participants were informed that they would “receive a pull in either direction at any time” and to “respond as safely as possible to maintain balance”. There were 10-30s between each perturbation, and a short rest break between every block of five trials.

Preferred stance loading for patients was determined by calculating the mean percentage of body weight over each limb while the participant stood quietly for 60 seconds. Participants were given verbal feedback to ensure the pre-perturbation stance loading remained consistent across trials. Patients completed two separate test sessions within two days of each other (summary of number of trials by condition and test session is found in Table 4). The first test session focused on: 1) determining preferred stance weight distribution (to be used as feedback to ensure this remained constant across tasks); 2) determining the response characteristics in response to a perturbation without limb
Table 4. Summary of test protocol for healthy controls and patient cases. A summary of task conditions, perturbation magnitude and number of trials in each condition is presented. Patients completed perturbations in two test sessions on two separate days. Healthy young adults completed all trials in a single session. Only high-magnitude backward-directed perturbations were analyzed (bolded); additional trials were included with varying task conditions so that the direction and magnitude of the perturbation would be unpredictable. High- and low-magnitude perturbations for individuals with stroke were half the magnitude of those for controls.

<table>
<thead>
<tr>
<th>Participant group</th>
<th>Testing session</th>
<th>Condition</th>
<th>Magnitude</th>
<th>Trials</th>
</tr>
</thead>
<tbody>
<tr>
<td>Healthy young adults</td>
<td>1</td>
<td>No constraint</td>
<td>9% body weight</td>
<td>5 backward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>6% body weight</td>
<td>2 forward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>1.5% body weight</td>
<td>4 backward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>1% body weight</td>
<td>2 forward</td>
</tr>
<tr>
<td>Patient cases</td>
<td>1</td>
<td>No constraint</td>
<td>4.5% body weight</td>
<td>5 backward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>3% body weight</td>
<td>3 forward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.75% body weight</td>
<td>3 backward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.5% body weight</td>
<td>3 forward</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>Paretic limb constraint</td>
<td>4.5% body weight</td>
<td>5 backward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>3% body weight</td>
<td>2 forward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.75% body weight</td>
<td>2 backward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.5% body weight</td>
<td>2 forward</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Non-paretic limb constraint</td>
<td>4.5% body weight</td>
<td>5 backward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>3% body weight</td>
<td>2 forward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.75% body weight</td>
<td>2 backward</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>0.5% body weight</td>
<td>2 forward</td>
</tr>
</tbody>
</table>

constraints. In the second test session, participants were again tested without constraints (unconstrained) prior to completing two other test conditions: 1) constrained paretic limb (encourage non-paretic use); 2) constrained non-paretic limb (encourage paretic use).

Cardboard obstacles were placed around either limb to prevent stepping with the
constrained limb. The order of limb constraint tasks was counter-balanced between participants.

Healthy young adults underwent one test session in which the unconstrained task condition was completed. The pre-perturbation stance loading weight distribution position was standardized at 50% body weight over each limb for controls. This was accomplished by presenting participants with visual feedback of the vertical force over each limb.

4.2.3 Data analysis

The current study characterized the responses from individual cases (case series) and compared the values to data from the healthy young adults. Each trial and response was characterized by: 1) pre-perturbation stance loading; 2) limb used for stepping response; 3) onset of stepping; 4) foot-off time; 5) swing time; 6) anticipatory response duration; 7) number of steps.

Information regarding the limb used for compensatory stepping and the number of steps taken were obtained from the videos and confirmed with the force plates. The onset of compensatory stepping was identified as the time when the medio-lateral (ML) COP deviated by 4mm from the mean ML COP 200ms prior to the perturbation (McIlroy & Maki, 1999). The onset of ML shifts could be associated with an initial anticipatory postural adjustment (APA), if an APA was executed, or the onset of limb unloading. Distinguishing the presence of an APA was determined by the direction of the initial ML COP excursion (i.e. towards the swing limb). As a result, duration of an APA, if present, was defined as the difference between the response onset time and the time of peak ML COP displacement towards the swing limb. Unloading phase time was the difference between the time to foot-
off and onset time of swing limb unloading leading to a step. Time to foot-off was the time when vertical load on one force plate was less than 1% of body weight (McIlroy & Maki, 1995a). Swing time was the difference between foot-off and foot-contact (>1% body weight on the posterior force plate) time.

4.3 Results

Results for the healthy young adults and patient cases are highlighted below. Values presented are means ± standard deviation. All four patient cases presented with high clinical balance (BBS) scores (range 50-56 out of 56), and posses leg /foot impairment scores (CMSA) ranging from moderate to good (2 - 6 out of 7). Frequency of stepping, multiple stepping and timing of responses for trials in the unconstrained condition are presented in Table 5. The duration of each of the phases of the stepping response are displayed in Figure 7. ML COP displacements for each participant in the unconstrained condition are displayed in Figure 8.

4.3.1 Healthy young adults

Of the 11 participants, eight were right leg dominant. Controls stepped in 89% (49/55) of total trials in the no constraint condition and 96% (47/49) of steps were taken with the dominant limb. Nine participants stepped in response to the very first perturbation. Foot-off times for steps taken with the dominant and non-dominant limbs were 515±127ms and 540±219ms respectively.

4.3.2 Patient Case 1 (PC1)

PC1 had an excellent clinical balance score (BBS 56/56), mild foot impairment (CMSA 6/7) but more impaired lower limb control (CMSA 3/7). In spite of his high BBS score his
balance confidence was low (ABC 44/100). During quiet stance, his preferred stance loading was 53% body weight over the non-paretic limb.

Unconstrained condition

PC1 appeared reluctant to step following the perturbations. He stepped in 3/5 trials, and in the other two trials he relied on assistance from the spotters during a prolonged posterior lean. He did not step in the very first trial. Due to a technical error, step timing data were not available for PC1; however, his stepping reactions appeared to be slow in the three trials where he stepped. All steps were taken with the non-paretic limb. This strategy of delayed or absent compensatory stepping reactions persisted in session two; he stepped in one of the three trials in the unconstrained condition, and the step was taken with the non-paretic limb.

Limb constraint condition

Attempts to encourage use of the paretic limb by blocking the non-paretic limb were unsuccessful. PC1 stepped in only one of five trials and the step was taken with the non-paretic limb, resulting in a collision with the obstacle and a reliance on the spotters and harness to ensure stability. An additional trial was attempted in which PC1 was explicitly asked to take attempt to take a step with the paretic limb when perturbed with the non-paretic limb constrained; the participant refused to do so.

4.3.3 Patient Case 2 (PC2)

PC2 had a high clinical balance score (BBS 50/56), severely impaired foot and leg control (CMSA Foot 2/7, CMSA Leg 2/7) and low balance confidence (ABC 49/100). During quiet stance, his preferred loading was 47% body weight over the non-paretic limb.
Table 5. Step frequency and phase timing average across trials for each patient case. Data presented is mean ± SD from trials in the unconstrained condition in session one for the patient cases and the unconstrained condition in healthy young adults. Unless otherwise indicated, timing is presented for steps taken with the non-paretic limb (or dominant limb in controls).

<table>
<thead>
<tr>
<th></th>
<th>PC1</th>
<th>PC2</th>
<th>PC3</th>
<th>PC4</th>
<th>Controls</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stepping Frequency (%)</td>
<td>60</td>
<td>100</td>
<td>100</td>
<td>100</td>
<td>89</td>
</tr>
<tr>
<td>Frequency of Steps</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>with Non-Paretic/</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Dominant Limb* (%)</td>
<td>100</td>
<td>100</td>
<td>100</td>
<td>40</td>
<td>96</td>
</tr>
<tr>
<td>Onset Time (ms)</td>
<td>N/A</td>
<td>297±57</td>
<td>256±58</td>
<td>137±33</td>
<td>227±52</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>125±61 (P)</td>
<td></td>
</tr>
<tr>
<td>Foot-Off Time (ms)</td>
<td>N/A</td>
<td>891±96</td>
<td>681±59</td>
<td>515±33</td>
<td>515±127</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>558±118 (P)</td>
<td></td>
</tr>
<tr>
<td>Swing Time (ms)</td>
<td>N/A</td>
<td>147±11</td>
<td>211±30</td>
<td>109±41</td>
<td>129±42** (McIlroy &amp; Maki, 1996a)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>253±66 (P)</td>
<td></td>
</tr>
<tr>
<td>Multiple Step Frequency</td>
<td>0</td>
<td>100</td>
<td>0</td>
<td>100</td>
<td>2%</td>
</tr>
</tbody>
</table>

P=Steps taken with paretic limb.
*Dominant limb refers to control subjects.
**Swing time for controls was not available from this study. This is a referenced value.

Unconstrained condition

PC2 stepped in all trials and all steps were taken with the non-paretic limb. For unconstrained steps, time to foot off was slow compared to young adults, though swing time was rapid. PC2 used multi-step responses in all trials and the second step was always taken with the paretic limb. PC2 had no anticipatory phase in the very first trial, but the anticipatory phase was apparent for all other trials and became longer in duration with
each trial. Observations from video show that both the initial step with the non-paretic limb and the second step with the paretic limb were always directed postero-medially, resulting in limb collisions on the second step in three of the five trials. The unconstrained condition in session two revealed a similar strategy to that used in session one. PC2 stepped in all three trials in this condition and all steps were taken with non-paretic limb. Time to foot off for these steps was 879±76ms and swing time was 191±20ms.

*Limb constraint condition*

During trials in which the non-paretic limb was constrained, PC2 stepped in all five trials with the non-paretic limb and collided with the obstacles. Time to foot off was long at 891±94ms and swing time was 161±12ms for these steps.

4.3.4 *Patient Case 3 (PC3)*

PC3 had an excellent clinical balance score (BBS 55/56), mild foot and leg impairment (CMSA Foot 6/7, CMSA Leg 5/7) and relatively high balance confidence (ABC 72/100). During quiet stance, PC3 had a preferred stance loading of 38% body weight over the non-paretic limb (i.e. more weight on the paretic limb).

*Unconstrained condition*

PC3 stepped in all trials in the unconstrained condition and all steps were taken with the non-paretic limb. Immediately following each perturbation, PC3 exhibited a small posterior lean prior to foot off and all steps were directed postero-medially but he did not have any multiple step responses. In contrast to session 1, during session 2, PC3 did not step in any trials in the unconstrained condition.
Figure 7. Duration of response phases for each individual trial in the unconstrained condition in testing session one. All trials represent steps taken with the non-paretic limb, except for those surrounded by a dashed line, which indicate paretic limb steps. PC2 & PC3 demonstrated long anticipatory/unloading phases which contributed the long delay in time to foot-off. PC2 and PC4 had rapid swing times for steps taken with the non-paretic limb that may have been associated with the occurrence of multiple stepping responses in these conditions. When stepping with the paretic limb, PC4 had a delayed foot-off time, slower unloading phase and no anticipatory response compared to steps taken with the non-paretic limb.
Limbs constraint condition

Similarly, PC3 did not step in any trials with the limb constraint but it appeared as though a step was not necessary to safely maintain stability.

4.3.5 Patient Case 4 (PC4)

PC4 had a high clinical balance score (BBS 52/56), severe foot impairment (CMSA 2/7) and mild leg impairment (CMSA 4/7). In spite of his foot and leg impairment scores, his balance confidence was the highest of the patients (ABC 79/100). During quiet stance, PC4 had a stance loading of 52% weight over the non-paretic limb.

Unconstrained condition

In the unconstrained condition, PC4 stepped in all trials; three of the steps were taken with the paretic limb and two were with the non-paretic limb. All five trials resulted in multi–step responses and the second step was taken with the limb opposite to the first step. The first perturbation resulted in a step with the non-paretic limb. Similar to the first session, PC4 stepped in all three trials in the unconstrained condition in session two. One of the three steps was taken with the non-paretic limb and the other two with the paretic limb. Foot-off and swing times for the step taken with the non-paretic limb were 459ms and 160ms, respectively. Foot-off and swing times for steps taken with the paretic limb were 627±49ms and 369±239ms, respectively.

Of the four cases, PC4 was the only participant who was able to step with the paretic limb. The steps taken with the paretic limb in the unconstrained condition had no anticipatory phase as well as a markedly slower unloading time & swing time prior to foot off compared with the steps taken with the non-paretic limb. The MLCOP tracing for the
Figure 8. Medio-lateral centre of pressure plots for each individual trial in the unconstrained condition of the first testing sessions for the patient cases and one representative trial from a healthy young adult. The first trial in the testing session is highlighted in black and the subsequent trials in light grey. The first vertical tick represents the onset of response and the second vertical tick represents time of foot-off. Following response onset, PC2 & PC3 demonstrated a ‘hesitation’ phase, denoted by the perseveration in the MLCOP tracing, which delayed foot-off time. In the first and second trials, PC4 seemed to demonstrate an aborted paretic limb step due the abnormally large MLCOP shift towards the non-paretic side prior to stepping with the non-paretic limb. Steps taken in subsequent trials for PC4 were taken with the paretic limb and anticipatory MLCOP shifts were absent in these trials.
very first trial (step with non-paretic limb) for PC4 displays an abnormally large anticipatory 
phase that resulted in nearly 50% unloading of the vertical force under the paretic limb 
approximately 400ms following the perturbation, prior to stepping with the non-paretic 
limb. This degree of unloading suggests that the original intention was to step with the 
paretic limb, but the step appeared to be aborted in favour of stepping with the non-paretic 
limb.

Limb constraint condition

When the non-paretic limb was constrained, PC4 stepped in all five trials with the 
paretic limb. Foot-off and swing times for these steps were 546±225ms and 252±158ms, 
respectively.

4.3.6 Summary of Cases

Overall, all patients demonstrated a great degree of instability in response to 
applied perturbation (reliance on other support, multiple stepping etc.). Stepping was 
occasionally avoided and commonly much slower even when involving the non-paretic limb. 
Of the four cases, three were nearly symmetric in their stance loading position and one 
bore more weight on his paretic limb. Three of the four PC took compensatory steps in the 
very first trial and these steps were all taken with the non-paretic limb (Figure 1). The foot-
off time was generally long and the onset time and swing time were generally comparable 
to healthy controls. Three of the four cases stepped in every trial in the unconstrained 
condition and all participants stepped in at least three of the trials. Only one participant, 
PC4, executed compensatory steps with the paretic limb. Swing times for the three steps
taken with the paretic limb were delayed compared to steps taken with the non-paretic limb.

4.4 Discussion

The four individuals with stroke in this study were all able, in some trials, to execute stepping responses to attempt to regain balance following postural perturbations. However, in spite of conventional clinical indices indicating very good to excellent recovery of standing balance after stroke, all cases were characterized by profound challenges to the control of rapid compensatory stepping. Three primary characteristics of compensatory stepping were observed: 1) predominant stepping with the non-paretic limb, and an inability or unwillingness to step with the paretic limb; 2) multi-step responses; 3) delayed execution of compensatory stepping. Furthermore, the individuals with stroke recruited for this study were relatively young (<55 years old) and had no pre-morbid conditions that are likely to have affected their postural control; therefore, the impairments in the control of compensatory stepping observed are likely primarily associated with the stroke. This is the first study to attempt to quantify impairments in compensatory stepping responses in individuals with stroke. The present results highlight the need for further study of the underlying control problems, the consequences of such control challenges, and suggest specific patient characteristics that can relate to the problems with such reactions.

In general, there was a reluctance to step with the paretic limb in three of the four cases. This has been anecdotally reported previously (Marigold et al., 2005). PC4 was the only participant who stepped with the paretic limb. He also had a high balance confidence score but, interestingly, had the greatest impairments in foot and leg control. Conversely
PC1 had the lowest balance confidence score in the group and was the only participant who resisted stepping with either limb in the first test session. PC1’s attempted use of compensatory grasping reactions (grasping the cable) is consistent with previous studies that have demonstrated attempts to use upper limb reactions to avoid stepping (Maki & McIlroy, 1997). Therefore, it is possible that balance confidence is associated with an individuals’ capacity to execute a compensatory stepping responses with either limb (paretic or non-paretic). It is interesting to note that the reluctance or capacity to step with the paretic limb was not simply associated with the degree of limb impairment for these cases. However, conventional clinical measures do not reveal limitations in speed of movement, which may be a more critical determinant for rapid compensatory behavior.

Multi-step responses were frequently observed in the patient cases presented here. These responses are among the most frequently-reported age-related impairments in compensatory stepping responses and the tendency to respond to postural perturbations with more than one step has been suggested to be linked with an increased risk of falling (Luchies et al., 1994; Maki, Holliday, & Topper, 1994; Mansfield & Maki, 2009; McIlroy & Maki, 1996a; Rogers, Hedman, Johnson, Cain, & Hanke, 2001). While there is debate regarding the role of multi-step responses as either a measure of instability or a pre-planned strategy to prevent a fall, the pattern of stepping should be considered when attempting to decipher between the two possible outcomes. In this study, the second step was always directed medially, suggesting that there was some lateral instability during weight acceptance of the initial response and previous research has demonstrated that extra steps may be needed if the first compensatory step was insufficient to prevent a fall.
(Maki & McIlroy, 1999; Pai et al., 1998). Therefore, despite the fact that patients predominantly stepped with their ‘better’ (i.e. non-paretic) limb, a single step with this limb was ineffective in completely recapturing the falling centre of mass following the perturbation. Consequently, a second step, most commonly taken with the paretic limb, was needed to fully recover from the perturbation. As such, patients were often unable to completely avoid having to step with the paretic limb.

The timing of the stepping response phases varied across cases, but in general the overall time to foot contact was slower than in healthy young adults. Delays could be associated with onset timing, time to unload (and generate an APA) and swing time. It was hypothesized that such delays would exist due to a need to unload the more-loaded non-paretic limb in order to step with that limb (Lakhani et al., 2008). However, three of the four patients adopted a symmetric stance position (47-53% of body weight over the non-paretic limb) and the other patient bore more weight on the paretic limb. The delay in foot-contact was not attributable to the onset of the stepping response.

The onset time of stepping was generally preserved, which appears to counter previous studies revealing delayed paretic limb activation in response to perturbations. For example, previous research has demonstrated that EMG response onset time in the paretic limb is delayed in individuals with stroke (Ikai et al., 2003). In such studies the focus is on the generation of limb specific responses (e.g. fixed support reactions). In contrast, the measurement of the onset of stepping is commonly done from the ML COP (McIlroy & Maki, 1999). This was the case in the present study and, therefore, the intact onset time of stepping may well represent the fact that the initial ML COP excursion could be initiated by
the non-paretic side. Delays in foot-contact, then, arise from delays to the initiation of unloading, which could be due to need for large ML APA, and/or the time to unload the limb.

Unloading time (time to foot-off), was delayed compared to healthy young adults. All participants showed abnormal anticipatory control of ML COP prior to foot-off, and two participants showed perseveration in the COP time series, with multiple APAs apparent. Such perseveration may have contributed to the delays in foot-off time. Multiple APAs have been observed in a previous study where participants were prevented from pre-selecting their stepping foot (Jacobs & Horak, 2007). The possible presence of multiple APAs in the current study could indicate indecision regarding the choice of stepping limb. Indeed, it is unclear if the large APAs observed prior to PC4’s steps were indeed APAs or if they were aborted attempts to step with the paretic limb.

Swing time was generally very fast in PC2 & PC4 when steps were taken with the non-paretic limb. However, PC2 & PC4 also had multiple stepping responses in every trial which suggests that the initial step with the non-paretic limb might not have been large enough to stabilize the falling COM. The inability to adequately prepare for the impending loss of stability caused by a stepping response could result in an accelerating COM towards the swing limb, requiring a rapid, short step to prevent a fall. Conversely, responses with slow swing times were also observed. PC3 had slow swing times with steps taken with the non-paretic limb, but he did not have any multiple stepping responses. Similarly, PC4, when stepping with the paretic limb, had a much slower swing time compared to his steps with the non-paretic limb. The slow swing time in paretic steps could be associated with previous
research that has identified reduced paretic muscle activation as well as co-contraction in lower limb muscles when responding to perturbations (Badke & Duncan, 1983; Dietz & Berger, 1984). These steps with the paretic limb also needed multiple stepping responses that may have been the result of the COM trajectory falling too far posteriorly during the long swing time so that the initial step was not enough to harness the COM.

Despite the challenges faced by the stroke patients to execute these balance reactions (e.g. movement delays, multiple stepping), all four cases had very good to excellent clinical balance scores (BBS greater than 50 out of a maximum score of 56). Studies have shown that BBS scores less than 49 are associated with an increased risk of falls (Berg, Wood-Dauphinee, Williams, & Maki, 1992; Bogle Thorbahn & Newton, 1996; Riddle & Stratford, 1999; Shumway-Cook, Baldwin, Polissar, & Gruber, 1997). Thus, initial observations from the present study, acknowledging the small sample size, might indicate that current clinical assessments of balance control post-stroke do not adequately reflect limitations in reactive balance control. While the BBS includes measures of dynamic balance control, such control is volitional and not exclusively reflective of reactive control. This highlights the need for measurement tools that specifically characterize reactive balance control. New tools, such as the BestTest, include a component that specifically assesses compensatory stepping reactions (Horak et al., 2009). Reactive control is required to generate appropriately timed and scaled responses that serve as the last line of defense to a perturbation, and is essential to independent mobility (Maki & Whitelaw, 1993; McIlroy & Maki, 1995a; McIlroy & Maki, 1995b).
This study attempted to characterize compensatory stepping responses in individuals with stroke. Due to the high degree of variability in stroke patients, a case series approach was adopted to allow for the examination a small number of cases in greater detail. However, this limits our ability to generalize these results to a larger population. Further studies will be required with larger samples to verify these results in wider range of individuals. This study was further limited by technical difficulties with the accelerometers that arose during testing of PC1 which resulted in a lack of temporal indices for steps taken by that participant. In addition, the perturbation device created for this study might not have delivered as brisk a perturbation as desired. The use of a weight drop system to deliver the perturbation requires the weight to gradually accelerate to its maximal velocity, thus decreasing the overall impact force generated on the person (Mansfield & Maki, 2009). Finally, constraining the non-paretic limb in order to force stepping responses with the paretic limb was unsuccessful in this group. Despite being unable to force stepping with the paretic limb, we were able to demonstrate that overriding either the physical or perceived restrictions of stepping with the paretic limb might be a more difficult endeavor than originally anticipated. Future attempts at forcing stepping via use of a constraint may benefit from using a more rigid barrier that simulates actual environmental barriers that would not simply move out of the way.

4.4.1 Conclusions

This current study describes the challenges individuals with stroke may face when attempting to execute compensatory stepping responses. A general reluctance to initiate steps with the paretic limb and that steps taken with the non-paretic limb have abnormal MLCOP displacement prior to swing limb unloading, are slow and frequently require
multiple stepping responses were also observed. Even though this case series tested a small sample of individuals with stroke we were able to emphasize that, although the stroke population is extremely heterogeneous, similarities in the response strategy used to respond to a balance disturbance may exist. No studies to date have investigated the characteristics of these responses. Further investigation with a larger cohort is needed to identify spatio-temporal deficits in steps taken with the paretic limb. These studies could attempt to target specific areas of the response (time to foot off, foot contact time, limb trajectory) in an attempt to identify modifiable characteristics, which could be the basis for perturbation-based balance training.
5.0 General Discussion

The purpose of this thesis was to explore the potential challenges that might arise when individuals with stroke execute a perturbation evoked compensatory stepping responses. In order to first understand the limitations that individuals with stroke might encounter, an attempt to simulate one of the post-stroke challenges (stance asymmetry and stepping with a loaded limb) was made and the responses were tested in healthy young adults. It was discovered that increasing stance loading on the preferred stepping limb and forcing stepping with that limb results in steps that are shorter length and directed more laterally than steps taken from symmetric stance. In addition, healthy individuals were able to alter their limb choice for the stepping response depending on the environmental conditions of the task and make ‘online’ adjustments to their balance response in a similar manner as was reported previously in other studies.

When individuals with stroke responded to stance perturbations they were often reluctant to step and, when necessary, they tended to rely on their non-paretic limb to execute compensatory stepping reactions, as healthy controls rely on their dominant limb. In particular, individuals with stroke were characterized by long delays to the onset of limb unloading and time to foot-off, during which time there were medio-lateral centre of pressure oscillations. This might have been the result of a hesitation associated with executing the stepping response because of either anxiety or a genuine inability to complete a step with the paretic limb. Interestingly, the patient case with the lowest balance confidence was also the case that was most resistant (greatest delays) to executing a stepping response, while the patient case with the greatest confidence was the only case that was able to step with both his non-paretic and paretic limb.
The short swing phase times found in both healthy young adults when loaded on their preferred stepping limb and in individuals with stroke may be linked to the altered trajectory of stepping responses as well as the increased incidence of multiple stepping responses, an index of overall instability. Multiple stepping responses have been associated with an increased risk of falling in older adults and are likely the result of an initial first step that is insufficient in controlling the trajectory of the COM (Luchies et al., 1994; Maki et al., 1994; McIlroy & Maki, 1996a; Rogers et al., 2001). Importantly, this work revealed specific challenges to the control of compensatory stepping that were not revealed in the more traditional clinical assessment of balance. For example, all patient cases had Berg Balance Scores greater than 50, indicating no increased fall risk and very good balance control. In reality, our study revealed that in the face of a balance disturbance, the balance recovery reaction is significantly compromised after stroke, as represented by reluctance to use the paretic limb, delays in foot-off time, rapid swing time and a high incidence of multiple stepping. This suggests that current clinical methods of assessing reactive balance control may not be sufficient to capture deficiencies in balance responses in this population.

This work highlights the importance of better assessing the challenges in reactive control after stroke to both inform about the potential increased ‘fall risk’ and, more importantly, to help guide therapies specifically directed at reactive balance control. Our current work, while featuring a small sample size, clearly highlights the concern that those with good clinical balance scores are at a high risk of falling when faced with an external perturbation. This is an essential determinant of their safety and capacity for independence when discharged into the community.
5.1 Limitations

There are some limitations to the present study. The three most significant are: 1) characteristics of perturbation and generalizability to other balance conditions; 2) absence of kinematic data for patient cases; 3) small sample sizes for stroke collection and the potential impact on generalizability to wider range of stroke patients.

The methods used to deliver perturbations delivered in each of the two studies were different and each had potential limitations in regards to the direction, amplitude and particular task conditions. The original bi-directional weight drop perturbation device, which was used for the stroke study, was limited by its inability to deliver a brisk onset perturbation. The weight drop resulted in a more gradual increase of pull force on the participant. This allowed participants to delay their time to foot-off or completely resist stepping altogether by using a fixed support balance recovery strategy. To accommodate for this, a lean & release perturbation device was designed similar to those used in previous studies (Do et al., 1982; Hoshiyama et al., 1993; Thelen et al., 1997) The lean & release system required an extremely rapid response, better challenging the balance control system and requiring a change-in-support reaction (with the appropriate lean angle). Despite the rapid release of the lean & release system, this device may have been limited by its ability to only deliver a unidirectional (anterior) perturbation. In addition, previous research has also reported the potential for participants to learn and adapt to particular characteristics a perturbation; improving their ability to respond with repeated exposures to perturbation trials (McIlroy & Maki, 1995a). In patients, novel first trial responses were separated from subsequent trials for further analysis and attempts randomize the timing of perturbation as well as overall expectation of perturbation (by including catch trials), but...
future investigations would benefit from a perturbation device that can deliver brisk perturbations in multiple directions. Perturbations directed medio-laterally would decouple the response characteristics for each limb and further highlight difficulties associated with paretic limb stepping responses.

The selection of perturbation amplitude is challenging because delivering a perturbation that is too small may not evoke the desired response and a perturbation that is too large can impose potential risks for individuals with stroke. In order to maximize safety, a safe standardized weight-drop load lean distance was selected that could encompass even the lowest functioning stroke patient that met the eligibility criteria. In order to make the perturbations more challenging, perturbation magnitude could be gradually increased (via the lean angle in future models) to find the maximal safe amplitude for each individual participant. Customization of the perturbation amplitude may be necessary in future studies to ensure that the perturbation is both challenging as well as safe for individuals with stroke.

The introduction of an environmental barrier to block stepping with the non-paretic limb in individuals with stroke was unsuccessful. In the interest of safety, the cardboard barrier placed around the non-paretic limb was easily displaced upon contact. However, in order to force stepping with the paretic limb, participants should either be unaware of the ease of barrier displacement or the barrier should be rigid and more visibly imposing. More recent testing of individuals with stroke has utilized a strategy of simply having the therapist place an arm in front of the limb. This has likely been successful because of the perception experienced by the participants that they may cause physical harm to the therapist if they
attempt to step with the blocked limb. If this method of limb blocking continues to be successful, it will prove to be useful due to its ease of clinical implementation.

Many previous studies characterizing balance reactions have relied on measures of body movement (e.g. kinematics). In the current study kinematic analysis was limited to observational analysis of video data to minimize collection time for individuals with stroke. The result may have limited the ability to characterize the details of the step trajectories, which may have provided an indication of how successful the initial stepping response was compared to healthy controls. In a study such as this, where the task itself can be fatiguing on the participant, kinematic analysis is not a feasible option. Future investigations that attempt to quantify step placement following a perturbation may benefit from the use of pressure sensitive mats to capture footfall location.

Finally, despite the case series nature of experiment 2, it was limited by its small number of participants. The study was revealing in that it highlighted the challenge of reactive stepping control among a small series of relatively high functioning individuals with stroke. This work drew attention to reactive balance control challenges and revealed the potential to conduct such perturbation testing among individuals with stroke. Certainly future studies, following this initial work, will need to focus on larger groups of patients to establish links between reactive control and clinical characteristics. This study contributed to the development of a clinic that is presently routinely assessing reactive balance control responses in individuals with stroke. The clinic will assist in establishing the links between response strategies to participant characteristics as well as provide each patient with an individualized treatment plan that focuses specifically on their balance deficits.
5.2 Future Directions

The results from this thesis provide the groundwork for future investigations that would be required so that outcomes can be directly applied to the advancement of clinical assessments and rehabilitation therapies for individuals with stroke. In particular, future research in this population should attempt to: 1) investigate the role of different features perturbation elements (direction, magnitude, pre-perturbation stance) on response strategy; 2) establish the relationship between clinical indices of balance control and the strategy used to respond to an external perturbation; 3) explore the potential for perturbation based training to improve reactive balance control.

Firstly, varying the characteristics of the perturbation could uncover additional balance control demands that have yet to be revealed. In our initial study, perturbations were applied only in the sagittal plane. However, because falls in individuals with stroke typically occur as a result of insufficient lateral stability, an investigation of the response strategy to perturbations in the coronal plane would be beneficial. In addition, medio-lateral perturbations would uniquely capture the specific interlimb response differences due to the asymmetry caused by the stroke. Additionally, given that falls frequently occur during gait, future studies may benefit by investigating the effect of delivering a balance perturbation during gait on the response strategy and characteristics of that response. Varying measures of ‘central set’ (affect, arousal, attention, expectations, prior experience) as well as perturbation characteristics (timing, magnitude, direction) on the balance response could all help to uncover aspects of the balance control system that are affected by stroke.
Secondly, establishing relationships between characteristics of reactive balance control responses and balance confidence, incidence of falling and clinical balance scores could help disentangle some of the underlying causes for balance dyscontrol in this population. Results from this thesis suggest that there might be a relationship between patient characteristics (ABC score, CMSA grade, BBS score) and the characteristics of the compensatory response strategy. Identification of predictors of increased fall risk as a result of impaired reactive balance control strategy is an important step to enhancing rehabilitative therapy and reducing the risk of falls in this population. In addition, current clinical measures of balance control may not adequately assess reactive reactions, as demonstrated by this thesis. All four cases in this study had BBS greater than 50, indicating no increased fall risk (Berg et al., 1992; Bogle Thorbahn & Newton, 1996; Riddle & Stratford, 1999; Shumway-Cook et al., 1997), but all cases had clear deficiencies in their response compared to healthy controls. Thus, the development of a new clinical assessment, which includes a reactive balance test, is a critical step in helping to successfully identify high-fall risk individuals. Future investigations could define participant characteristics of interest (i.e. balance confidence, stroke severity, cognitive abilities, clinical balance scores) and attempt to correlate them with balance response markers of interest (limb choice for stepping, phase response latencies, limb trajectories) using a larger cohort of individuals with stroke.

Finally, perturbation based balance training has lead to improvements in response latency and step length in individuals with Parkinson’s disease (Protas et al., 2005), but the extent to which this type of training could be useful in individuals with stroke is unknown. Previous work has demonstrated that individuals with stroke are able to adapt and improve fixed support strategies in response to balance training but no research to date has
investigated the trainability of compensatory stepping responses (Marigold et al., 2005). Conventional clinical rehabilitation for stroke survivors currently focuses on training rapid voluntary stepping but, due to longer response latency and the associated APAs that typically accompany volitional stepping, this method of training might not effectively transfer to reactive, perturbation evoked stepping responses. Reactive balance control training, that specifically attempts to decrease the length of the unloading phase of the response, improve limb trajectory, and encourage stepping with the paretic limb, should be a major focus during rehabilitation for individuals with stroke to help decrease the number of falls experienced by this at-risk population. The effect of positive spatio-temporal changes following either single session or long term perturbation training could have important clinical implications.

5.3 Conclusions

This work was the first to investigate compensatory stepping reactions in individuals with stroke and the findings demonstrate the spatial and temporal deficits associated with these responses. Healthy individuals, when forced to step with an asymmetrically loaded limb to mimic a potential post-stroke response, have short swing phase times as well as laterally directed steps and short step lengths. Individuals with stroke, however, seem reluctant to step with their paretic limb and steps taken with the non-paretic limb are characterized by slow time to foot-off, short swing times and increased incidence of multiple stepping when compared to healthy individuals. Ongoing investigation of the underlying mechanisms of the impaired stepping response and subsequent translation of those findings into perturbation based rehabilitative training are required in order to reduce the risk of falling and improve the overall quality of life in this population.
6.0 References


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