The use of haptic forces in the control of gait and posture in chronic stroke and elderly individuals

Gianluca Sorrento, MSc

Faculty of Medicine

School of Physical and Occupational Therapy

McGill University Montreal, Quebec, Canada

Date of initial submission: December 15, 2017

Date of final submission: April 15, 2018

A thesis submitted to the Faculty of Graduate and Postdoctoral Studies in partial fulfillment of the requirements of the degree of Doctor of Philosophy

(Rehabilitation Sciences)

DEDICATION

I dedicate this thesis to my mother, father, and brother. Their love, support, and patience throughout my PhD studies was the source of strength needed to accomplish this thesis.

ACKNOWLEDGEMENTS

My PhD experience was a truly gratifying journey accompanied by many wonderful people. Indeed, they have kept me company throughout a long, arduous and beautiful path. This thesis is a testament to that journey.

My deepest gratitude goes foremost to my supervisors Dr. Philippe Archambault and Dr. Joyce Fung for providing me with guidance, wisdom and belief over the years. Their role in making this thesis possible was absolutely essential at every level. I am eternally grateful to them both.

I was also fortunate to cross paths with a vastly talented and committed research team at the Jewish Rehabilitation Hospital. This thesis was also made possible with their contributions. I am sincerely thankful and truly indebted to Valeri Goussev for his original analytical and intellectual contributions to the project. I am also grateful to Christian Beaudoin who created the virtual scene, and Igor Sorokin who created the leash system. By offering their talents and innovation, the virtual dog and leash became a reality. I also want to thank Gevorg Chilingaryan for his expertise in statistical analysis, Michael Trevino for sharing his clinical insights, Vira Rose for her tireless efforts in recruitment, and Louise Veronneau and Melissa Chartrand for their administrative support. I also thank my lab mates, Nahid Norouzi, Gordon Tao, Anuja Darekar, Amir Vafadar, Samir Sangani and fellow students at the JRH and SPOT, past and present, with whom I shared many graduate study experiences and milestones.

I am also grateful to the McGill School of Physical and Occupational Therapy, including Dr. Eva Kehayia and Dr. Isabelle Gélinas for their guidance through the different phases of my PhD studies, and Maria Ruocco and the rest of administrative team at the McGill SPOT for all of their valuable assistance.

Many thanks are also extended to the CRIR ethics committee and the CRIR scientific committee, especially the latter who collectively entrusted me with the responsibility of student representation during most of my PhD studies.

I would also like to thank members of my protocol committee, Dr. Anouk Lamontagne and Dr. Cyril Duclos, who both when called upon contributed valuable scientific ideas.

I am also infinitely thankful to all of my course instructors earlier on in my PhD experience. The rich education provided by the likes of Dr. Nancy Mayo, Dr. Robert Dykes, Dr. Sara Ahmed, and Dr. Sharon Wood-Dauphnee, among others, has provided me with a solid foundation of research knowledge and critical scientific thought.

I would also like to thank the funding organizations whose belief and generosity shown towards my PhD project have helped sustained it throughout. My gratitude goes out to the Richard and Edith Strauss Foundation, the Fonds de Recherche en Santé de Québec (FRSQ), the St. Justine / REPAR-MENTOR fellowship, and the INTER doctoral fellowship.

Lastly, I would like to acknowledge the participants of this study. I am thankful for the time and effort that they have graciously contributed. This thesis would not exist without them. May the fruits of this work benefit them most.

TABLE OF CONTENTS

ABSTRACT

The high prevalence of stroke among the elder population is of particular concern and poses serious challenges to daily functions such as locomotion. In light of these challenges, the use of haptic stimuli has been developed as a strategy to help stabilize gait and posture. These stimuli come in difference forms - from earth-fixed surfaces to elaborate wearable devices promoting stability. Many methods involve some form of contact or weight bearing to bring on a somatosensory stimulus, while much fewer investigate external forces brought on to the body. The present study offered a novel approach by applying haptic tensile forces on the hand, in the direction of locomotion, to study changes in gait outcomes. Changes in these gait parameters would support the hypothesis that haptic tensile forces bring on locomotor adaptation and post-adaptation effects in chronic stroke and healthy non-stroke individuals. These adaptation and post-adaptation effects could eventually develop into an effective rehabilitation strategy for clinicians. To investigate this prospect, a proof-of-concept study was first designed involving 13 healthy young adults who walked on a self-paced treadmill in a virtual environment with a robotic-powered leash in hand to investigate haptic tensile effects of the leash on steady-state walking outcomes, during and after a 10 or 20N force application, relative to baseline walking. The main findings were significant increases in gait velocity with accompanying stride length and double limb support time changes. These findings suggested that the application of force was enough to bring on transient changes in spatiotemporal gait parameters of healthy individuals during and after force exposure.

In light of the evidence found in the first study, the question remained if an older chronic stroke population also had the ability to change gait parameters in a similar way and if these changes promote dynamic stability from a kinematic perspective. To address this question, 14

older chronic stroke and healthy age-matched control subjects were recruited. Similar to the pilot study, both chronic stroke and control groups increased gait velocity by over 0.1 m/s when 10 or 15N forces were introduced for one minute. Changes in gait velocity remained above baseline for an additional minute after force removal. Also, stride lengths increased as stride times decreased for both groups. In terms of dynamic stability, double limb support time was significantly decreased for both groups during and after force exposure, while step width did not seem to be significantly altered by tensile forces. In terms of postural control, most subjects also tended to shift their center of mass towards the paretic side for stroke subjects and non-dominant side for controls. In order to develop a rehabilitation strategy that may prove beneficial to the stroke population, it would be necessary to compare how the leash strategy changes gait spatiotemporal, postural and coordination parameters compared to a very common walking aid such as the cane. Hence, for the third study, both stroke and control groups completed walking trials with the leash and with an instrumented cane. The stroke group was stratified into lower functioning stroke and higher functioning stroke. As in the previous two studies, gait velocity was then maintained above baseline after force removal to an extent comparable to walking with a cane, suggesting a post-adaptation effect. The lower limb coordination of the paretic leg (stroke) revealed increases of both hip and foot flexion and greater angular velocity of the limb during leash walking relative to baseline levels. In light of all findings, future work is warranted to maximize the efficacy of the haptic leash strategy for possible use in the clinical setting.

ABRÉGÉ

L'importante prévalence d'accidents vasculaires cérébraux (AVC) chez les personnes âgées est préoccupante et pose de grands défis à la fonction, notamment la locomotion. Compte tenu de ces défis, l'utilisation de stimuli haptiques a été développée comme stratégie pour stabiliser la démarche et la posture. Ces stimuli peuvent être présentés sous plusieurs formes, comme par exemple, des surfaces fixes ou bien de technologies portables qui promeuvent la stabilité. De nombreuses méthodes ont utilisés un point de contact physique ou de soutien afin de créer un stimulus somatosensoriel. Cependant, on connait moins les effets des forces extérieures apportées au corps. Par conséquent, ce projet propose une nouvelle approche, impliquant des forces de traction haptiques au niveau de la main, en direction de la locomotion, pour mesurer les changements dans les paramètres de la démarche. Ces changements soutiennent l'hypothèse selon laquelle les forces de traction haptique provoquent des effets d'adaptation et de post-adaptation locomotrice chez les personnes qui ont subi un AVC et les individus sains (sans AVC). Ces effets d'adaptation et de post-adaptation pourraient mener à une stratégie de réadaptation efficace utilisée dans le milieu clinique. Pour approfondir cette perspective, un projet pilote a été effectué, impliquant 13 adultes jeunes et en bonne santé qui ont marché sur un tapis roulant, à leur rythme, dans un environnement virtuel et avec une laisse robotique à la main. On a mesuré les effets haptiques de la laisse pendant la marche à rythme constant, pendant et après une application de force (10 ou 20N), par rapport à la marche de base. Les résultats principaux ont dévoilé une vitesse de la démarche significativement élevée. La longueur de la foulée a aussi été élevée par rapport à la marche de bas, alors que le temps en double appui a été diminué. Ces résultats suggèrent que l'application de force a été suffisante pour causer un changement transitoire des paramètres spatiotemportaux de la marche chez les

personnes en bonne santé, pendant et après l'application de la force. Cependant, il restait à savoir si une population plus âgée, qui a subi un AVC, pourrait également modifier ses paramètres de démarche d'une manière similaire, et si ces changements promeuvent une meilleure stabilité dynamique. Pour aborder cette question, 14 sujets âgées atteints d'AVC et des sujets témoins de même âge ont été recrutés. Comme dans l'étude pilote, les deux groupes ont augmenté la vitesse de la démarche de plus de 0.1 m/s lorsque des forces de 10 ou 15N ont été activées pendant une minute. Les changements de la vitesse de la démarche sont demeurés au-dessus du niveau de base, même une minute après le retrait de la force. De plus, les longueurs des foulées ont augmenté alors que les temps des foulées ont diminué pour les deux groupes. Quant à la stabilité dynamique, le temps en double-appuie des membres inférieurs a significativement diminué chez les deux groupes, pendant et après l'exposition à la force, alors que la largeur de pas n'a pas été réduite significativement. En ce qui concerne le contrôle postural, la plupart des sujets expérimentaux (AVC) et témoins ont également eu tendance à déplacer leur centre de masse vers le côté parétique ou non dominant. Afin d'élaborer une stratégie de réhabilitation qui pourrait être avantageuse pour la population qui a subi un AVC, il faudrait comparer comment cette stratégie de laisse robotisée modifie les paramètres de démarche (spatiotemportaux, posturaux et de coordination) par rapport à un aide à la marche assez commune comme la canne. Donc, pour la troisième étude, les deux groupes ont complété des essais de marche avec la laisse et avec une canne adaptée. Le groupe expérimental (AVC) a été stratifié selon leur niveau fonctionnel (faible ou élevé). Comme dans les deux études précédentes, la vitesse de la démarche a été maintenue au-dessus du niveau de base après l'élimination de la force au niveau comparable à la marche avec une canne, ce qui suggère un effet de post-adaptation. La coordination de membre inférieur parétique (AVC) et non dominant (témoin) a montré une augmentation de flexion au niveau de la hanche et du pied, ainsi qu'une augmentation de la vitesse angulaire du membre parétique pendant la marche avec la laisse, par rapport au niveau de base. Ces résultats justifient une poursuite de la recherche afin de maximiser l'efficacité des stimuli haptiques comme éventuelle stratégie dans le milieu clinique.

STATEMENT OF AUTHORSHIP

I, Gianluca Sorrento, hereby confirm primary authorship of this dissertation and claim all responsibility for its scientific accuracy and content.

CONTRIBUTIONS OF AUTHORS

Study concept and methodology was a joint effort between Gianluca Sorrento, Dr. Joyce Fung and Dr. Philippe Archambault. Gianluca Sorrento collected, processed and analyzed data under the guidance of Dr. Archambault and Dr. Fung. Gianluca Sorrento also composed all three manuscripts included in the thesis with review and contributions provided by Dr. Philippe Archambault and Dr. Joyce Fung.

STATEMENT OF ORIGINALITY

The following dissertation is of original scientific work composed in a manuscript-based format. The two of three manuscripts included have been published, or have been submitted for publication, in peer-reviewed scientific journals at the time of thesis submission. The third is in preparation to be submitted for winter 2018. While slight variations to the manuscripts have been made to thesis and article versions, the contents and data presented are fundamentally similar. Hence, considerable overlap of these works is expected. A review of literature and discussion sections have been added to the dissertation according to the requirements specified by McGill Graduate and Postdoctoral Studies.

All data and results presented in this dissertation have been conceptualized, collected, analyzed and reported at the Feil Oberfeld Research Centre of the Jewish Rehabilitation Hospital (JRH) located in Laval, Québec, Canada. The JRH is affiliated with the McGill University School of Physical and Occupational Therapy and the Centre for Interdisciplinary Research in Rehabilitation of Greater Montréal (CRIR). The CRIR ethics board approved of the protocol for this thesis.

The works presented in this dissertation contribute to the field of rehabilitation in gait and posture with the aim of understanding the efficacy of forward-leading tensile haptic forces imposed on the body during gait in older stroke and healthy populations. The first study established evidence of gait parameter changes among healthy young adults using the haptic force. The second study replicated the first with older chronic stroke and healthy age-matched controls. The third study employed both groups and compared spatiotemporal, postural and coordination gait parameters between haptic forces and an instrumental cane.

LIST OF TABLES

LIST OF FIGURES

LIST OF ABBREVIATIONS

6MWT - Six Minute Walk Test

AVC - Accident Vasculaire Cérébrale

BA - Brodmann Area

BBS - Berg Balance Scale

BOS - Base of Support

BWS – Body Weight Support

CI - Confidence Interval

CNS - Central Nervous System

COM - Centre of Mass

COP - Centre of Pressure

CPG - Central Pattern Generators

CPU - Central Processing Unit

CRIR - Centre de Recherche Interdisciplinaire en Réadaptation du Montréal Métropolitain

EEG - Electroecnephalography

FES - Falls Efficacy Scale

fMRI - Functional Magnetic Resonance Imaging

fNIRS - Functional Near- Infrared Spectroscopy

GEE - Generalized Estimating Equation

HbO_2 - Oxygenated Hemoglobin

HSF - Heart and Stroke Foundation

ICF - International Classification of Function Disability and Health

PDI - Proportional Derivative Integration

SAC - Somatosensory Association Cortex

SCI - Spinal Cord Injury

SMA - Supplementary Motor Area

VE - Virtual Environment

VR - Virtual Reality

WHO - World Health Organization

CHAPTER ONE: OVERVIEW AND OBJECTIVES

1.1 Overview

The high prevalence of stroke in Canada is concerning. The Canadian Heart and Stroke Foundation (HSF) estimates that approximately 50 000 strokes occur yearly (Heart and Stroke Foundation, 2011). This translates to an overall national prevalence of stroke of ~1.3% as of 2013 estimates (Krueger et al., 2015). These rates are problematic to many individuals and their families as both cope with the various physical and psychosocial challenges of stroke. It also places a considerable strain on the public health system. This very system must provide a myriad of services that aim to rehabilitate individuals back into daily living and provide long-term care for those who are the most severely afflicted. Healthcare costs covering morbidity and long-term care alone are well over 3 billion dollars annually (Cameron et al., 2016; Heart and Stroke Foundation, 2011). With the growing uncertainty whether such costs are sustainable to health care (Sepehri & Chernomas, 2004), it is clear that stroke survivors reacquire the functional independence to alleviate some of the burden - for the sake of the patient, family (Lutz et al., 2011) and indeed healthcare system (Di Carlo, 2009). Among the rehabilitation objectives assumed, perhaps one of the most widely cited goals of post-stroke patients is the ability to regain the static and dynamic functional stability necessary to reintegrate safely into community dwelling (Baer & Smith, 2001; Eng & Tang, 2007)

While the stroke mortality rate in recent years has been fortunately decreasing (Lackland et al., 2014), functional walking in the chronic stroke stage continues to prove a challenge for many stroke survivors as falling remains quite commonplace and tends to have a strong effect on post-stroke morbidity (Hesse, 2003; Hong, 2015). Even so, post-stroke rehabilitation provides a unique challenge, as virtually all stroke patients are functionally dissimilar, and yet, most survivors with lower limb paretic involvement exhibit some walking dysfunction in the terms of gait velocity, asymmetry and energy efficiency (Balaban & Tok, 2014; Belda-Lois et al., 2011; Lindsay et al., 2008). Estimates in the literature state that walking constraints are found in at least 50% of the stroke population (Baer & Smith, 2001) and 66% of those over the age of 65 (Granger et al., 1992). Studies show that these shortcomings tend to predict important outcomes related to overall quality of life, such as social participation and anxiety of falling (A. G. Andersson et al., 2008). Consequently, it is not surprising that a large focus of stroke rehabilitation is centered towards restoring independence in these individuals as safe pedestrians in the community (Barclay et al., 2015; Fulk et al., 2010).

While there is a fundamental physical component that contributes to safe and functional community dwelling, which is also the main focus of this thesis, it is by no means the only factor. If it were just a physical matter, then walking safely in the community would be matter of putting one foot in front of the other and avoiding a fall. The task calls for attending to other factors in order to be achieved. These challenges can be brought into a complete construct that outlines what a stroke individual must overcome in order to regain participation into the community setting (Shumway-Cook et al., 2002). With this in mind, research that directly addresses the physical rehabilitative strategies for clinicians aims to mainly address the physical aspect of a construct that also involves the environmental and task contexts (Eng & Tang, 2007).

Many researchers have taken to this task with technologically innovative approaches that incorporate the fundamentals of sensorimotor integration and engineering (Bastian, 2008). Principles in motor control and skill acquisition can be applied to change motor behaviours for safer outcomes (Shadmehr & Mussa-Ivaldi, 1994). More specifically, some researchers have used the effects of sensorimotor adaptations to redefine how the stroke patient can actively learn and unlearn previously hindered motor patterns in the CNS. Researchers have also approached this idea using different sensory modalities, such as visual (Finley et al., 2014; Lamontagne & Fung, 2009), auditory (Turchet et al., 2013), and somatosensory cues (Afzal et al., 2015), often presented in virtual environments (Bermúdez i Badia S., 2016), to induce changes in gait behaviour, thereby potentially changing or reshaping motor programs that drive locomotion for the betterment of functional walking (Dobkin, 2004).

By drawing from the exciting innovation in this field, the present dissertation is dedicated to suggesting a novel way of approaching functional gait rehabilitation. In doing so, this work aims to eventually provide an outlet to clinicians in terms of a potentially cost-effective strategy that can be implemented in the clinical milieu with simple modifications. Namely, the use of forward tensile forces applied on the hand, in the direction of walking, can potentially draw on the principles of environmental constraints and sensorimotor integration to drive motor programs responsible for generating functional and effective locomotion in stroke patients, and potentially those who suffer from other neuromuscular disorders. The following objectives are presented as individual studies in the dissertation, but nonetheless progress cumulatively by building principles from one study to the next.

1.2 Research Questions and Objectives

1. In a healthy young population, to what extent are haptic tensile forces applied to the hand, in the direction of walking, compared to no force, facilitate adaptation changes evident in spatiotemporal gait outcomes, and to what extent do the adaptations persist after the force is removed.

2. In an elder post-stroke and age-matched healthy populations, to what extent do haptic tensile forces applied to the hand, in the direction of walking, compared to no force, facilitate adaptation changes evident in spatiotemporal gait and postural outcomes, and to what extent do these persist after the force is removed.

3. In an elder post-stroke and age-matched healthy populations, to what extent do haptic tensile forces applied to the hand in the direction of walking, compared to an instrumented cane, facilitate adaptation changes as evidenced by spatiotemporal, postural, and kinematic outcomes, when the force is applied and after it is removed.

The following chapter consists of a comprehensive, yet concise, review of literature pertinent to addressing the objectives of this dissertation. Considerations will be given to the challenges that stroke patients, clinicians, and researchers face with returning to community ambulation and the multifaceted challenges that the individual must confront in doing so. A dissertation in the field of gait analysis will require reference point outlining the aspects of healthy gait in order to compare and contrast with chronic stage post-stroke gait by outlining the general constraints of hemiparetic ambulation. Once the specifics of post-stroke gait are outlined, common gait measurement strategies found in the literature will be presented, as well as measurement tools to quantify them.

After establishing a rationale as to why post-stroke gait must be improved, how it differs from healthy locomotion, and how it is assessed, the different strategies taken by researchers and clinicians in addressing functional gait recovery in post-stroke are reviewed and discussed. Lastly, having gained a thorough understanding of the state of the literature, the three studies presented in the dissertation aim to address both the specific objectives previously outlined and the knowledge gap identified in the literature review.

CHAPTER TWO: REVIEW OF LITERATURE

2.1 The scope and goals of stroke rehabilitation

2.1.1 Impact of stroke

According to the World Health Organization (WHO) stroke can be defined as a cerebrovascular event causing disturbances of cerebral function beyond 24 hours, or even death, even if the exact vascular origin is unknown (Thorvaldsen et al., 1995). They can be ischemic, whereby a brain lesion is created due to an occlusion of oxygenated blood, as in the majority of cases, or by hemorrhaging of a blood vessel in the brain. According to the Canadian Heart and Stroke Foundation (HSF), there are over 300,000 thousand individuals living with stroke, which make up 1.3% of the national population. In addition, there are approximately 50,000 cases of stroke occurring annually in Canada (Cameron et al., 2016; Heart and Stroke Foundation, 2011). These figures account for approximately 3.5B $ (Heart and Stroke Foundation, 2011) annually in health care costs, not to mention the social and economic burden placed on the individual and family members.

While the general incidence of stroke mortality has been steadily in decline (Lackland et al., 2014), believed to be in large part due to reducing risk factors such as hypertension, smoking (Bogousslavsky et al., 1988; Tuomilehto et al., 1991), taking on healthy activity (C. D. Lee et al., 2003) and diet choices (Goldstein et al., 2006), there is still a need for those who have suffered a stroke to regain as much function in daily living as possible. Some estimates suggest that half of those living with stroke deficits are dependent on others to some degree to accomplish daily tasks (Mercier et al., 2001). With the extent to which the post-stroke individual, their kin, and health care system are involved (Lindsay et al., 2008), it becomes clear that effective interventions, such as the one the current dissertation evaluates, are needed to address this problem.

2.1.2 Risk of Falling Post-stroke

Adding to the problem of high stroke rates is the ensuing risk of falling post-stroke, which has been firmly established in the literature (F. Batchelor et al., 2010; Nyberg & Gustafson, 1995a). Falls have been identified by some authors as the leading cause of medical complication for the post-stroke community (Davenport et al., 1996; Vivian Weerdesteyn et al., 2008b) and disability (Ma et al., 2014). In fact, falls occurring as a result of post-stroke are enough commonplace for researchers to consider the event of falling itself as a complication of the acute stroke during the rehabilitation process (Nyberg & Gustafson, 1997) and after home discharge during the chronic phase (Simpson et al., 2011; Tsur & Segal, 2010). Reduced muscular tone and limb coordination, as manifested in paralysis in both static and dynamic stability phases are some of the significant physical factors contributing post-stroke falls (Tsur & Segal, 2010). One must also consider a general fear of falling as a common outcome for post-stroke survivors. It has been reported that a fear of falling can lead the post-stroke survivor to assume less activity and participation (V. Weerdesteyn et al., 2008a). A physical deconditioning (Watanabe, 2005) can ensue, leaving the post-stroke survivor more prone to falling. As a potential factor influencing walking performance, many studies take to measuring a subject's aversion to falling through various measures. Once such responsive measure (Hellstrom et al., 2002) is the Falls Efficacy Scale (FES) featured in the current project (see Appendix A). Further, subsequent insults is a cause for concern among populations such as post-stroke, and in particular elderly populations typically over the age of

65 (Tinetti, 2003). One notable study (Tinetti et al., 1994; Tinetti et al., 1990) found that elderly populations were more prone to falling with increases in risk factors.

Consequently, the objective of clinicians and researchers is how to help chronic stroke individuals return safely to the community dwelling and diminish the chances of falling (F. A. Batchelor et al., 2012). However, we will see there are factors that define the challenges and needs associated with bringing individuals back to a level of walking that is functionally independent, or even close to the levels of function prior to the stroke. From this starting point, researchers continue to work within the clinical scope of rehabilitative science in an effort to offer clinical interventions that facilitate and maximize an individual's rehabilitation potential and ultimately regain an important degree of pre-injury function and quality of life. A significant step in reacquiring function, and central to this dissertation is the use and development of assistive devices (Bateni & Maki, 2005; Jutai et al., 2007) to facilitate the gait rehabilitation process.

2.1.3 Use of assistive devices in post-stroke.

The use of assistive devices is commonplace for fall prevention and in the rehabilitation process for the post-stroke population (Allet et al., 2009). The extent of use depends on the severity of a stroke and the degree of resulting functional impairment (Jutai et al., 2007). They can be employed in a wide array of daily functional activities, supporting both dynamic and static posture, such as a task requiring stable walking and standing. Common assistive devices for locomotion include walking canes (including quad canes, and Nordic sticks), walkers and wheelchairs. Assistive devices geared towards providing postural stability for more static daily tasks (e.g. bathing, dressing and cooking) include adaptive support bars and shelves. In the scope of gait rehabilitation, the use of assistive devices can often indicate the extent in which functional ability has been adapted, reacquired, or possibly maladapted (Bateni et al., 2004; Bateni & Maki, 2005).

2.1.4 Goal and scope of gait rehabilitation

An important distinction has emerged in the field of rehabilitation that discerns the ideas of functionally based goal setting and recovery or restoration of function (Scobbie et al., 2013). A functionally based goal involves regaining the ability to achieve a functional ability or task, even if some level adaptation is required. For instance, we can imagine someone, who after surviving a stroke, is constrained to a wheelchair in the acute stroke phase. Fortunately, this individual does regain some locomotor ability, despite the hemiparesis evident on the affected side of the body. Among other considerations affecting gait, which will be discussed in depth, this will compromise the individual's ability to be proactive and reactive to walking in a typical community setting (Muir et al., 2015; Nieuwenhuijzen & Duysens, 2007).

In contrast, researchers and clinicians are mindful of the mounting evidence of neural plasticity (Hallett, 2001) and its role in functional sensorimotor recovery and motor learning in neurological patients such as stroke (Murphy & Corbett, 2009; Nudo & Friel, 1999; Teasell, 2003; Teasell et al., 2006). It is also encouraging to note that evidence of such recovery may be possible well beyond the sub-acute stage (Kwakkel et al., 2004). Consequently, researchers pursue ways to achieve the best rehabilitation gains possible given the physical potential of the patient and reduce adapted function. To accomplish this, both clinicians and researchers investigate innovative strategies, such as the one proposed in the current project, to advance rehabilitation methods that can potentially restore as much recovery of function possible, given the severity and functional level of the individual (Teasell, 2003).

While post-stroke survivors ideally aim to restore pre-stroke autonomy, the challenge of walking independently, or with a walking aid for that matter, is not entirely physical in nature (Shumway-Cook et al., 2002). Functionally safe locomotion at home or in the community, with or without assistive devices, requires a series of other factors that are also non-physical and cognitive in nature (Nanninga et al., 2015). The International Classification of Functioning Disability and Health (ICF) developed for the physical therapy domain a broader scoped contextual model accounting for these factors. They can be contextual (i.e. environmental or personal) factors as well physical and participation ones. As one can imagine, most gait and posture studies will aim to address the physical aspect of the ICF model, since the strategies used are often designed to change the kinematic and spatiotemporal aspects of walking (Silva et al., 2015). However, researchers and clinicians often bear in mind this framework when considering how any physical intervention, such as the assistive devices or devices designed to optimize motor recovery, such as those presented in this dissertation, potentially align with the interdisciplinary scope accounting for these factors. To meet this challenge, innovative techniques and technologies, such as robotically driven assistive devices and platforms, as well as and virtual reality (VR) provide an invaluable means to achieving novel interventions (Deutsch et al., 2004; Hallett, 2001). As we will see, the current project also abides by this viewpoint as it employs the use of virtual environments (VE) and robotics to provide a safe and ecologically valid context to the rehabilitation strategy proposed.

2.1.5 Virtual Reality in stroke rehabilitation

The study of locomotion has often brought on innovative strategies that not only accurately assess patient function, but also serve as a rehabilitation tool for community ambulation (Yang et al., 2008). The use of virtual reality (VR) technologies has proven useful in achieving both objectives (Keshner, 2004). Indeed, the integration of VR into rehabilitation science paradigms has even proved at times as an advantageous adjuvant strategy to compliment conventional therapy (Dobkin, 2004; Luque-Moreno et al., 2015). Unsurprisingly, the field of rehabilitation science has been a significant contributor to the implementation of VR technologies for research purposes. Given a population such as post-stroke poses different functional challenges with each patient, a challenge for VR is to remain generalized enough to service many functional levels, but to also provide a tailored experience to the user according to their functional level and therapeutic needs. A good example was the current project as VR enabled the simulation of walking a dog in an urban setting (Sorrento et al., 2013) (see fig 3.1).

The tailored use of VR, in contrast of real world rehabilitation, indeed has several potential advantages. First, it allows for the control over many physical variables that influence motor behavior, while also enabling the recording of precise spatiotemporal and kinematic data (Hollman et al., 2006; Sveistrup, 2004). Second, is the potential benefit of movement repetition, the control of intensity, the frequency of movement, and the sheer motivation factor for participants to attain personal achievements based on the performance feedback, often made available to them in VR paradigms (Adamovich et al., 2009; Jack et al., 2001; Sveistrup, 2004). In addition, VR scenes provide a safe environment for these individuals to be assessed or trained (Holden, 2005). This is an important factor when studies like those in the current project recruit neurological populations, such as stroke, who often have compromised stability.

A number of studies have established some measure of validity and reliability with the use of VR for stroke rehabilitation (Henderson et al., 2007; Merians et al., 2002). Specific to gait and posture, the effects VR in terms of evaluation and potential rehabilitation seem to have encouraging outcomes (Fluet & Deutsch, 2013; Henderson et al., 2007). Moreover, it has been shown that replicating an immersive environment as defined by a self-paced treadmill that is synchronized with a virtual scene can elicit similar behaviours to when walking over ground (Deutsch & Mirelman, 2007; Fung et al., 2006). The capacity to improve can also arise with repeated use of such systems, which can even translate to empirical estimates of motor learning, which are often evident in adaptation behaviours in activities such as walking (Olney, 1996). Further, the increasing accessibility due to reduced costs and increasingly commercially available products also favours the VR approach. The virtual environment typically provides a platform of applications that range from training to recreation that can be just as dynamic and innovative. Furthermore, VR can display simulated real-life functional activities, such as shopping (Rand et al., 2007) and crossing the street (Katz et al., 2005; J. Kim et al., 2007) in a safe and often entertaining manner.

Important questions remain regarding the challenges of applying and measuring changes brought on as a result of using VR technology and the relevance or transfer of virtual training to the real world (Bermúdez i Badia S., 2016; N. Kim, Park, et al., 2015; Savin et al., 2014). Some studies suggest that VR is a valid alternative to the over ground real-life methods of gait analysis with the added advantage of being able to measure precisely in real-time (Gates et al., 2012). The current study employs the use of VR for the very reasons mentioned (Sorrento et al., 2013). Stroke patients in the process of rehabilitation can benefit from VR as it is ideal for adapting to tasks, making them safer and set at the appropriate functional level of a post-stroke individual (Saposnik et al., 2016). This can be made evident, for example, when a patient is fitted into a body weight support (BWS) harness in order to walk with visual feedback of the patient walking in an augmented VR scene (Mehrholz et al., 2017; Moseley et al., 2005). The virtual scene provides goal-oriented repetition that can prove valuable to the individual receiving the exposure. The use of VR to supplement conventional therapy has improved mobility outcomes for the post-stroke population, as evidenced in several studies reporting improved functional walking outcomes (McEwen et al., 2014; Saposnik et al., 2011). One study specifically highlighted the use of balance training (e.g. simulating sport techniques) as a factor for improved functional outcomes, such as improvements in lower limb impairment scores post training according to the Chedoke-McMaster assessment (McEwen et al., 2014). Further results support the idea that using VR technology can promote the recovery outcomes needed for reacquiring functional gait (Mirelman et al., 2010). The following sections aim to underline the requirements of such gait, in contrast the compromised chronic stroke gait found in this project.

2.2 The analysis of posture and gait in healthy and stroke

2.2.1 Biomechanical considerations for static and dynamic posture

An important feature in human movement is our capacity to maintain an erect posture while performing functional bipedal gait. Bipedal gait has proven advantageous to humans in several respects such as enabling greater prehensile abilities (Kroker, 1999) and staying upright to see further away for potential food or danger (Harcourt-Smith & Aiello, 2004). The ability to maintain static and dynamic stability does, however, create a different set of kinematic and dynamic challenges to the body. Namely, we must overcome the biomechanical challenge of maintaining stability on two limbs instead of four, while controlling our center of mass (COM) relative to the base of support (BOS) (Capaday, 2002; Lugade et al., 2011). The BOS, which is the area immediately around the weight bearing limbs, will be generally narrower in bipedal stance relative to a four-point BOS. The limits of BOS become an important factor as the ability to ambulate effectively requires a coordinated effort of COM displacement relative to the limits of the BOS (Hof et al., 2005). Further, the biomechanical properties of bipedalism also imply that COM excursions within the base of support tend to be less than quadrupedal walking. Functionally speaking, community dwelling in a safe and functional manner must do with how well one controls their COM relative to the BOS (Hof et al., 2005; Lugade et al., 2011).

To many, locomotion may seem like a means of getting from point A to point B. A closer look into how this is actually accomplished suggests it may not be so trivial. Locomotion is achieved by placing the body's COM outside, preferably in the case of walking, just in front of the BOS. In this way, the body's COM is in a state of transient 'fall' in that particular moment. Essentially, we can think of walking as merely a series of well-coordinated and rhythmic 'falls' and 'catches' that the body performs in order to propel itself forward (Bancroft & Day, 2016; Lyon & Day, 1997). Specifically, once the COM is shifted forward beyond the BOS, the body propels itself forward with a step as to 'catch' the COM and return it within the confines of the BOS. This process repeated in succession renders the effect of normal locomotion in the forward direction. While it is understood that displacing the COM is a prerequisite of the locomotion process, a description of the gait pattern has yet to be defined. The next sections will outline the requirements of the gait cycle and compare the features typically characterising normal and chronic stroke gait.

2.2.2 Requirements and phases of the normal gait

In order for the 'fall and catch' mechanism to propel locomotion, there are well-defined guidelines outlining the physical requirements for gait to occur. First, a prerequisite for any mobile individual is the ability to maintain a stable static posture during quiet standing. According to the physical principle of Newton's third law, the upright body must keep a static equilibrium with ground reaction forces countering the gravitational forces exerted on the body. Anatomically, this can be accomplished mainly by the control of forces of moments acting on joints such as the ankle and knee to maintain a net zero body force (Mochizuki & Aliberti, 2017). Secondly, from a dynamic perspective, is the ability to physically unload and propel one foot past the other and create a net force in the anterior-posterior direction (Lacquaniti et al., 2012). This propulsion must be supported by the ground reaction forces projected against the feet within the BOS, which gives the necessary support during locomotion (Pai & Patton, 1997). In order to achieve the gait described, not only must the body be able to control its excursions with respect to the BOS to create locomotion, but it must also perform such functions in an environment that usually calls for the proactive and reactive functions to keep the body moving safely. There are many instances when dysfunction renders this ability difficult or nearly impossible depending on the severity. Loading of the foot requires the not only the ability to support the body's weight, which is comprised of mass and gravity on the narrowed surface of the foot step, but also the loading and unloading of the body's mass from one side to the other as the body is propelled forward

(Della Croce et al., 2001). From a biomechanical perspective, this requires that the body must be able to manage the normal forces brought on from the walking surface to the foot and overcome the mass and gravitational forces to create a net force in the direction of motion and also have a musculature that is stable enough to avoid slipping and falling.

The series of loading and unloading actions combine to form the phases of gait. In fact, each phase of gait has underlining physical requirements that will be discussed in this section. Saunders, Inman and Eberhart (1953) initially summarized these phases in what are known as the six determinants of gait (Kuo, 2007; Kuo & Donelan, 2010; Saunders et al., 1953). A complete gait cycle begins during the instant when the foot's heel initially strikes the ground and lasts until the next heel strike of the same foot. Within this single gait cycle, bilateral gait events occur in two main phases - stance and swing. The stance phase begins with the initial heel strike. During this initial contact, the quadriceps and hamstrings are activated as are the dorsiflexors of the lower leg. This configuration allows for the body to shift its weight onto the foot beginning with the heel and rolling onto more foot contact with the ground in the early stance phase. During this loading phase, co-activation of the quadriceps and hamstrings is still present. However, as the stride progresses into the mid-stance phase, this co-activation becomes mostly inactive. At this point, weight is shifted over the physical properties of the skeletal alignment. This leads to the heel lift phase, which is still in the loading aspect of the gait cycle (Kerrigan et al., 2000). The late stages of mid-stance to terminal stance is where an increased activation of the triceps surae occurs for plantarflexion in preparation of terminal stance. This coincides with the end of double limb support, as the contralateral foot prepares for the swing phase with the toe lift. As such, the leg is bearing most of the body weight with persisting contractions of the triceps surae, while some slight flexion of the hip, mainly from the iliopsoas muscle during the preswing phase. (Gard & Childress, 2001; Kerrigan et al., 2000; Saunders et al., 1953).

The swing phase of the gait cycle, which accounts for approximately 40% of the healthy gait cycle, is first comprised of the transition of toe off to the mid swing phase, where the contralateral limb is primarily responsible for bearing the body's weight. This is also the acceleration component of the leg swing. At this point, hip flexion is achieved with contractions of the iliopsoas, while some dorsiflexion is seen at the ankle. The mid swing marks the beginning of the deceleration of the swing phase where greater co-contraction of quadriceps and hamstrings can be seen. Dorsiflexion is also present as the hip extends to help with the landing of the foot. The terminal phase of the swing is once again the terminal heel strike, which returns the gait cycle to its beginning. It is important to note that these processes occur mainly in the sagittal plane, but also in the frontal and vertical planes as well. While slight physical deviations can occur from person to person, these guidelines are useful when compared to the deviations found in pathological gait, such as chronic stroke. The next sections will outline the characteristics of pathological gait with a focus on the features of stroke.

2.2.3 Approach to gait analysis chronic stroke

Pathological gait is often characterized by mechanical inefficiencies and restricted motion (Mena et al., 1981; Waters & Mulroy, 1999). Consequently, neurological deficits or dysfunction can interfere with the 'fall and catch' mechanism. Many chronic stroke individuals, for example, are faced with such challenges (Eng & Tang, 2007). We recall that in the gait pattern of a healthy individual, coordination and movement of body segments will fall within ‘normal’ limits. In the case of gait in chronic stroke, an experienced clinician should be able to notice even slight deviations in sagittal, frontal and transverse planes that would fall outside the limits of healthy gait (Eng & Winter, 1995). These deviations often stem from the neurological deficits, which in the case of stroke are often evident with hemiparesis, often preclude the physical requirements driving the determinants of gait (do Carmo et al., 2015; Olney, 1996). Once a description of such deficits is identified, such as the one that will be provided below, the clinician can then focus on what needed to restore greater symmetry in the gait cycle. This process should take into account quantifiable measures such as postural reaction, limb coordination and spatiotemporal parameters which help predict favorable community reintegration (Kondoh et al., 1995). These measures are discussed in greater detail in section 2.3.

For many post-stroke individuals, the hallmark circumduction gait pattern of the hemiplegic leg is typically brought on by an hypertonic proximal muscle synergy coupled with extensor hypertonia of the weak distal limb musculature (Singer et al., 2013). The resulting gait pattern of the affected leg, in reference to normal gait would likely vary in the following manner – during heel contact, the stroke pattern would exhibit less hip flexion and greater knee flexion (Lucareli & Greve, 2006) and plantarflexion (C. M. Kim & Eng, 2003). The toe lift phase would engage the opposite synergy, namely, more hip flexion and less knee and plantar flexion are present relative to normal gait (Sheffler & Chae, 2015). Fundamentally, the distal weakness of the ankle often seen in stroke patients leads the weak dorsiflexion that makes foot clearance difficult. This consequently leads to the compensating strategies performed by the more proximal joints. For example, a common strategy is the semicircular movement, or circumduction of the extended paretic leg (Sulzer et al., 2010). It is attained by changes in hip and knee flexion synergies of both the affect and non-affected sides (Chen et al., 2005). From a kinematic perspective, the locomotor deficits often exhibited with the hemiplegic gait pattern, such as circumduction will have most likely exhibit deviations in the sagittal, coronal and transverse planes described in the previous section (Allen et al., 2011).

From a spatiotemporal perspective this would likely entail a decrease in gait velocity and a stride length in both legs (S. L. Patterson et al., 2007). From a dynamic point of view this would probably lead to a weakness (Bowden et al., 2006) or decoupling of the right and left sides in terms of activation magnitudes and timing. Temporally, post-stroke individuals typically spend less time in the stance phase per given gait velocity and also tend to spend more time in the double limb support phase as to find greater stability (K. K. Patterson et al., 2010; von Schroeder & Coutts, 1995). The researcher or clinician can both measure and prescribe what changes and exercise to be implemented to achieve greater alignment and symmetry thereby contributing to better functional mobility.

Often found with physical impairments are variability or deviation of movements and the lack of synchronicity in muscle contractions. This lack of coordination and symmetry, in relation to lesion site (Alexander et al., 2009), can render the cycle more variable across the certain aspects of the gait cycle, if not the entire cycle. For example, a post stroke individual with a hallmark hemiplegic gait may have difficulty after the unloading phase during dorsiflexion to allow toe lift for the affected limb (Greene & Granat, 2000). This may be manifested with the mentioned circumduction or ‘’hip-hike’’ of that affected side in order to have the foot clearance and extra loading of the healthy contralateral side. Interestingly, there is evidence that enabling more weight bearing on the paretic side can lead to greater gait symmetry (Nam et al., 2017).

An important goal for clinicians, and indeed this dissertation, is to determine where such deviations occur and to prescribe the appropriate rehabilitation techniques to restore as much regularity and symmetry to the gait cycle as possible, thus rendering it as functional and efficient as possible. To achieve this, the brain must accommodate for walking adaptation effects (Krakauer et al., 2005; Lam et al., 2006; Malone et al., 2011; Patla, 2003). In other words, it requires the plasticity needed to reshape the motor patterns that drive gait. Once more, the adaptability of gait is a central component of this project, and the next few sections will highlight how researchers believe this is achieved.

2.2.4 Adaptability of gait

Humans have the ability to forge adaptations in their walking ability. The capacity and the extent in which both healthy and neurological populations can do this is central to this study. There are number of reasons why humans can and do adapt their gait. When people age, for example, a series of physiological reductions including bone and muscle density (Mochizuki & Aliberti, 2017), as wells as mild degradation in motor learning via corticospinal pathways (Bruijn et al., 2012) play a significant role in the way the individual walks over a lifetime. Individuals who have sustained or are recovering from musculoskeletal injuries, would be another example of those who change their walking parameters. In short, the physical condition that one finds themselves is also an important means for producing gait adaptations. These adaptations are made possible when the motor programs that shape and control gait are altered. Perhaps transiently in the beginning, but eventually with repeated exposure to the physical constraints may go on to become acquired changes in the motor program. From a neuroanatomical perspective, changes in the motor program can occur in several areas of the central nervous system (CNS). Some researchers believe that a catalyst for such changes lay in the cerebellum (Bastian, 2011). Other areas of integrating the sensorimotor learning or reshaping of a learned motor task are cortical based, such as the prefrontal cortex and parietal located just rostral to the occipital lobe.

2.2.5 Cortical centers driving gait and locomotor adaptations

Seminal studies (Sherrington, 1910) describe the role of the central pattern generators (CPG) as a spinal cord circuitry of motor neuron and interneuron synapses driving gait in animal models. This caudal neuronal circuitry in the CNS has been shown to reflexively generate and even coordinate locomotor movement in the absence of supraspinal inputs from the brain (Guertin, 2012). The circuitry is maintained with dynamic interactions of excitatory and inhibitory feedback (e.g. type 1a and 1b reflex pathways) mechanisms. While there to date evidence of CPG mechanisms occurring in humans is debatable, some studies exhibit similar circuitries with spinal cord injury (SCI) patients who demonstrated continued bipedal movements (Nadeau et al., 2010). However, in the context of generating the movement necessary for goal-directed locomotion, contributions of higher level CNS areas, such as several cortical areas (Barthelemy et al., 2011) via pyramidal (i.e. supraspinal) tracts, are required. Additionally, there is evidence of subcortical (Luft et al., 2008) involvement including cerebellar activity (Martino et al., 2014) as influences in goal-directed gait and gait adaptation (Thach & Bastian, 2004).

With a focus on how sensory information is used in shaping and altering motor patterns for gait, such as somatosensory input delivered in this project, two of the more prominent cortical areas are the projections from parietal cortex to the prefrontal and primary motor cortices. In specific relation to the haptic and proprioception information necessary for locomotion presented in this project, such inputs are believed to be integrated specifically in Brodmann areas (BA) 5 and 7 of the superior parietal cortex. These areas are considered part of the somatosensory association cortex (SAC). Recent findings suggest that this area is involved in both the temporal and spatial organization aspects of walking (Knaepen et al., 2015). Other findings were reported in VR immersive settings that parietal areas were involved motor planning and intention of premotor-parietal networks (Wagner et al., 2014).

The use of recent imaging techniques such as EEG (Bulea et al., 2015) and functional near-infrared spectrometry (fNIRS) have been useful in quantifying the ongoing excitability of relevant cortical areas during locomotor training. For example, studies have found that other areas relating to the ongoing learning and adaptation process of locomotion are the supplementary motor cortex (SMA), premotor cortex, and prefrontal cortex (H. Y. Kim, Yang, et al., 2016). Studies have showed that these areas are active during the adaptation process of self-selected gait speeds during treadmill and robot-assisted walking, similar to what was done in the current project (H. Y. Kim, Yang, et al., 2016). Other studies found that in healthy individuals, changes in prefrontal cortex and premotor cortex were seen in speed associated changes and therefore adaptation (Suzuki et al., 2004) Additionally, several studies using the fINRS technique how reported most excitability in the prefrontal cortex as evidenced by hemodynamic changes (HbO_2) (H. Y. Kim, Yang, et al., 2016; Sangani et al., 2015). The significance of this area's excitability may be due to the cognitive demands required to maintain voluntary postural and locomotor control (H. Y. Kim, Yang, et al., 2016). In relation to stroke, pilot work on a chronic stroke subject established that similar changes in the sensorimotor functions driving locomotor training and adaptation (Sangani et al., 2015). The following sections provide an overview of measures used in the gait mechanisms described as they pertain to this project.

2.3 Measurement of gait outcomes

2.3.1 Kinematic and dynamic gait measures

While a strong understanding of generating both stroke and non-impaired gait is fundamental in the current project, equally important is how gait is measured. There are two main steams of empirical gait analysis; dynamics quantifies forces acting on the body (Eng & Winter, 1995) causing movement, and kinematics focuses on the measurement of body movement in space (McGinley et al., 2009). Scientists have long been interested in both aspects of the gait profile. In the 19th century, electrophysiological dynamic properties were believed to be introduced by Duchenne (Parent, 2005). Other notable scientists such as Saunders, Inman and Eberhart have laid out a modelling or framework for effectively quantifying gait (Della Croce et al., 2001; Saunders et al., 1953). More recently, gait analysis for neurological disorders ensued. For example, Perry and colleagues took to classifying the features and challenges specifically relevant for rehabilitating the stroke population into community dwelling (Perry et al., 1995).

The current thesis approaches gait analysis, both healthy and stroke from a predominantly kinematic perspective, as spatiotemporal measures of movement and posture can be derived from kinematic information. Effective quantification of spatiotemporal qualities includes instantaneous gait velocity, stride length and double or single limb support time. The same kinematic methods can be used to report measures of postural stability and coordination by means of COM excursion, limb segment range of movement (ROM) and angular momentum. The advantage of measuring such outcomes is that many of the ones described are sensitive to minimal change (Rábago et al., 2015), thus giving some insight to the efficacy of a potential rehabilitation strategy. The following sections will cover the use the most pertinent measurement outcomes according to the literature and of relevance to the current thesis.

2.3.2 Gait Velocity

A kinematic analysis using spatiotemporal outcomes often includes a measure of gait velocity. Gait velocity in its simplest form can be defined as the amount of distance travelled per unit time. It is a measure cited as a main predictor of community ambulation (Perera et al., 2006) since a strong association has been established between gait velocity changes and improved functional gait (Schmid et al., 2007), dynamic balance and coordinated function (Lamontagne & Fung, 2004). These outcomes can hold especially true for populations, such as the elderly (Studenski et al., 2011), and populations that are challenged with bilateral stability such as high-risk fallers and post-stroke (Hars et al., 2013; Hsiao et al., 2017). Gait velocity is also regarded as a contributing factor to other spatiotemporal and dynamic outcomes (Wonsetler & Bowden, 2017). To further highlight its importance, gait velocity serves as an effective evaluation (Wagenaar & Beek, 1992) clinical screening tool (Allet et al., 2009; S. L. Patterson et al., 2007) and a predictor of health and community dwelling (Hornyak et al., 2012; S. L. Patterson et al., 2007; van Kan et al., 2009). Alternatively, it is common for researchers to report instantaneous velocity within the progression of a stride. It is defined as the derivative of position with respect to time. The advantage with this approach is that it provides more information regarding the variation of speed during a task or during different events of the gait cycle.

For a real-life example of how gait velocity can serve as functional outcome, we can return to the scenario of an individual crossing an intersection before the traffic light turns red. Clinically, it is fundamental to determine velocity benchmarks accurately predicting the minimum required gait speed to accomplish this or similar tasks. Research addressing this point has established a 0.8 m/s threshold for functional community dwelling (Schmid et al., 2007) in order to manage real life situations such as the one described. Also reported was that independent home dwelling has been found to be sustainable above the 0.4 m/s mark (Schmid et al., 2007). Perhaps just as important to clinicians and researchers are the clinically meaningful changes in gait velocity that can result from a rehabilitation intervention. Authors report minimal clinically important differences of 0.13 m/s and beyond (Richard W. Bohannon et al., 2013), while smaller changes can be reported in the 0.04 m/s – 0.06 m/s range (Perera et al., 2006) . These values can be used as a benchmark for significant changes seen in this project.

Such gait velocity changes can be measured in both clinical and laboratory settings using several methods. Motion capture is an increasingly common and reliable way to measure gait in a laboratory setting. It is often implemented in tandem with VR and employs one or more cameras to record movement of body segments for offline analysis. For example, the Vicon system used in the current project can record gait velocity by tracking the position of the heel within a stride at a rate of 120 Hz, with a calibrated spatial accuracy of approximately 1.5 mm (Merriaux et al., 2017).While gait velocity can be calculated precisely offline from experimental setups, such as the case in this project, other valid gait velocity measures can be simply taken from a pre-marked area and a stopwatch. Tests that commonly evaluate gait speed in the clinical setting, such as the 10 meter and 6-minute walk test have shown to be strongly correlated to gait speeds used in community, particularly when used in tandem (Dean et al., 2001). Both are reported to have high reliability (Flansbjer et al., 2005; Fulk & Echternach, 2008). For example, interclass correlation scores greater than 0.9 have been reported for normal and fast-paced 10-meter walking tests (H. J. Kim, Park, et al., 2016). Moreover, such walking tests seem to provide favourable concurrent validity and responsiveness (Mehrholz et al., 2007) particularly with walking distances greater than 10 meters comfortable walking speed (Dean et al., 2001). Both stopwatch methods and motion capture are used in the current thesis. Namely, the 10m walk test is used to establish both normal and face pace baseline walking velocities of all participants as a reference point for functional level, while motion capture was used during walking trials enabling a more precise empirical evaluation of gait velocity for both spatial and temporal parameters.

2.3.3 Stride length

Stride length is defined as the distance from one initial foot contact to the next contact of the same foot. It can vary in coordination and symmetry for both the intralimb and interlimb depending not only on walking requirements but also walking ability (Daly et al., 2007; Junho Kim, Oh, et al., 2015; K. K. Patterson et al., 2010). While individuals free of neurological deficits tend to have more symmetrical bilateral stride lengths, notable differences can be seen in younger and older (i.e. >65 y.o.) populations, possibly due to reduce strength and flexibility (Kang & Dingwell, 2008).

In comparison, post-stroke individuals are more prone to asymmetries in stride length (Hoogkamer et al., 2014). This typically comes as a result of compensatory strategies that arise due to weakness or poor motor control of the paretic side. In reference to components of the gait cycle, researchers report such compensations as strategies to offset weakness of the plantar flexors (Allen et al., 2011). For example, it has been demonstrated that post-stroke individuals are more likely to modify their stride length. This is generally achieved with shorter or irregular strides during velocity increases in an attempt to maintain stability during comfortable walking speed (Bayat et al., 2005; Hak et al., 2015). A notable benefit to improving the symmetry of stride lengths, including propulsion ability, between the paretic and healthy limb during an intervention process is a reduction of the physiological cost of walking (Awad et al., 2015). Thus, stride length symmetry is an aim for potential interventions such as the one described in the thesis.

Potential changes in stride length can also be classified as clinically meaningful and have been benchmarked in the literature. Researchers report that such changes can fall within the 25 mm to 60 mm range (Brach et al., 2010). In this project, if such significant changes in stride length occur in tandem with increased gait velocities, then it is also important address what effect this may have on walking economy. Outcomes in study 3 aim to suggest such changes.

2.3.4 Double limb support time

Walking by definition requires the both feet making contact simultaneously at some point of the gait cycle. In healthy gait this includes the time between the heel contact of one foot and the toe lift of the opposite foot. This occurs in two sub-phases during initial heel contact and toe off phases. As the double limb support phase roughly accounts for 20% of the gait cycle, there can be marked variations in post-stroke gait (Sudarsky, 1990). Again, these variations depend on the strength imbalances and coordination strategies of the paretic and non-paretic side. Most researchers, however, agree that any intervention or training that contributes to reducing double limb support time is beneficial to the individual, as the time one spends on both feet is a biomarker for a normalized gait (Goldie et al., 2001). It would be of particular interest if any intervention, such as the one introduced in the thesis, would significantly reduce double limb support times, in a similar manner suggesting a more stable dynamic posture.

The significance of reducing double limb support times may also address the efficiency and coordination of limb co-contraction during gait (Raja et al., 2012). Researchers have also found that among post-stroke individuals, the double limb support sub phases of the gait cycle tends to involve the excessive co-activation of the non-paretic lower leg in order to summon more postural stability (Lamontagne et al., 2000). Moreover, it is found that those who most fear falling, such as an elderly population, tend to increase double limb support time (Chamberlin et al., 2005) by shortening step length (Cromwell & Newton, 2004).

2.3.5 Single limb stance time

Single limb support time relates to the double limb time as they give an idea of how stable a particular limb is during the full loading phase of the gait cycle. In symmetrical gait, there tends to be very little inter-limb variability between legs in terms of single limb support time. However, the presence of a hemiparetic weakness in a limb can greatly affect the single limb support phase. This can also affect the loading ability of the paretic leg. Studies have shown that post-stroke individuals tend to spend on average less time on the paretic leg (Chen et al., 2005), placing the onus on the non-paretic limb to bear more of the single limb support. Further, the non-paretic single limb support is usually performed in tandem with a typical hip circumduction strategy that many post-stroke individuals adapt to achieve locomotion. Adding to this point, it has been shown that the non-paretic limb tends to increase in kinetic force relative to the paretic side, as it uses relatively more power to drive the gait process (Farris et al., 2015). In fact, one study reports that the mechanical work on the paretic side could be as low as 16% for high paretic severity and still less than 50% for low severity (Bowden et al., 2006).

Researchers suggest that increases in single limb support time on the paretic can represent an important contribution to a more symmetrical gait. An notable feature of this from a dynamic aspect would be the propulsion of paretic leg from the stance phase with sufficient leg extension (Peterson et al., 2011). Interestingly, it is also possible for the paretic side to regain more dynamic engagement with the use of canes and handrails and thus increased step time (Bingenheimer et al., 2015). This last note is of particular importance to the current thesis, as an external force or somatosensory (i.e. haptic) cueing may also lead to greater increases in relative step time and ultimately greater kinetic power of the paretic leg.

2.3.6 Centre of Mass (COM) displacement

In addressing the postural quality of locomotion, the measurement the centre of mass (COM) displacement can give valuable information about the displacement and energy costs of an individual's movement through space. Early models presented by Saunders, Inman, and Eberhart postulated how six determinants of gait can drive the COM forward in an efficient manner by spending the least amount of energy possible (Kuo, 2007). Specifically, these determinants contribute to reductions in vertical and COM displacement rendering the gait process more efficient. Other researchers however, suggest that limiting vertical displacement might be metabolically costly (Ortega & Farley, 2005). Alternative views have since been presented such as the inverted pendulum model, which accounts for the metabolic cost of stance (Gage et al., 2004; Yu et al., 2008) and weight shifting on the stance leg from one limb to another with the COM trajectory, including the vertical component being displaced as a result. Proponents of this view would argue that reducing the vertical displacement may in fact increase metabolic costs (Gordon et al., 2009). While the two views differ in opinion, both keep in mind energetic cost reductions as an implicit goal (Kuo, 2007). Since then, authors have added that it can be beneficial to understanding this outcome in pathological gait (Della Croce et al., 2001).

Maintaining a stable COM is a challenge to many post-stroke individuals. In fact, when comparing a post-stroke individual's COM trajectory to that of a healthy individual, the asymmetry of the COM in the stroke individual would often feature less displacement on the paretic side, relative to the non-paretic side. Specifically, during hemiparetic single limb support, there would be less forward displacement of the CoM accompanied with larger lateral displacement (Stanhope et al., 2014). In the swing phase, the hemiparetic side tends to show a higher vertical and lateral displacement with less forward displacement (do Carmo et al., 2015) and higher metabolic cost relative to the unaffected side (Farris et al., 2015). This is yet another description of the 'hip hike' walking strategy used to achieve foot clearance of the hemiparetic leg. These observed walking patterns along with empirical measures help disambiguate stroke gait patterns from healthy ones. Other studies report symmetrical COM displacement velocities in the healthy individual with increases in COM velocity during the swing phase. For the current thesis, it was important to measure any COM mediolateral displacements favouring paretic side involvement, which is indicated by CoM displacements in the frontal and sagittal planes, as a result of the exposure to the haptic force intervention.

2.3.7 Limb coordination in Gait

While we have seen that several models attempt to describe the gait process, a main factor in any description is energy efficiency. To achieve this, gait as a complex process requires movement in the sagittal plane as a main contributor to forward progression. However, as researchers point out, movement is also generated in the frontal and vertical planes as well (Eng & Winter, 1995). Deviations in gait in all three planes can be seen in healthy individuals, while the compensatory movements of hemiparetic post-stroke gait only serve to increase deviations (Allen et al., 2011). Hence contributions of movement in each plane may underline the inefficiencies of movement in a population such as post-stroke and healthy elder individuals (Farris et al., 2015; Winter et al., 1990). This may serve as a guide to improve specific aspects of the gait profile in terms of coordination. The underlying difficulties expressed in hemiparetic gait can contribute in large part to the loss of physical function. Different populations are faced with different physical enablers and barriers in achieving dynamic stability. It’s been shown that even within healthy populations the effect of age can have varying effects on the spatiotemporal factors that govern gait, whether due to loss peak muscle strength or fear of falling (Chamberlin et al., 2005).

Studies have attempted to illustrate a composite of lower limb coordination by combining lower limb segments such as the thigh, leg, and foot as a single expression or trajectory of movement in space (Bianchi et al., 1998; Ivanenko et al., 2002). Under such analysis, researchers can track not only the movements of each limb segment, but also the relative contributions of each given the walking conditions and how they relate to the velocity, mechanical and coordination factors already mentioned. The evaluation of coordination, such as the one conducted in this thesis, can serve to further elucidate the gait strategies used when, for example, an intervention such as walking with a force applied to the body or under different sensory conditions.

2.4 Rehabilitation strategies using haptic information for gait adaptation

2.4.1 Perturbations in locomotor rehabilitation

The use of perturbations to better understand mechanisms of balance (Freyler et al., 2015), walking (Mawase et al., 2013) and why falls occur (Bhatt et al., 2013) is commonplace in the research setting. Such perturbations are also analogous to many situations encountered in community dwelling. For example, they may come in the form of surface instabilities, such as travelling by bus or metro, walking across uneven surfaces, or even riding an escalator. In research, examples used to simulate these perturbations include moving platforms (Kaski et al., 2012), adapted treadmills (Bruijn et al., 2012) direct body perturbations (e.g. ankle perturbation) (Krasovsky et al., 2014), slippery surfaces (J. J. Jeka & Lackner, 1995) and uneven surfaces (Marigold & Patla, 2008). From a sensorimotor perspective, perturbations have been employed in gait analysis to observe both reactive and proactive (Bhatt et al., 2013) strategies of muscle recruitment and postural reactions in response to the unexpected disturbances applied to the body or encountered in the environment.

Many of the spatiotemporal measures mentioned in section 2.3 are also used to quantify responses to perturbations. These include measurement outcomes related to gait velocity and coordination. This is an important point as recovery of functional walking is not usually a matter of regaining locomotion in an isolated condition. In controlling posture, the brain must receive and integrate sensorimotor information to meet the mechanical demands of normal posture, whether standing, sitting, or walking. This must be accomplished in terms of both stability and efficiency in a gravitational environment that is often dynamic and subject to perturbations. Recent findings from Chvatal & Ting (2012) suggest that surface perturbations while walking elicit locomotor muscle synergies that originate for various cortical and subcortical areas of the brain in parallel descending pathways (Chvatal & Ting, 2012). These locomotor synergies account for atypical gait (e.g. from perturbations) for both proactive and reactive strategies (Nieuwenhuijzen & Duysens, 2007). In the event of perturbations, authors also suggest that balance control mechanisms tend to be mediated by brainstem structures, while motor planning of gait derives mainly from cortical areas such as the premotor-parietal tract (Wagner et al., 2014). Neurophysiological evidence provided from such studies may provide insight when behavioural outcomes (e.g. postural reactions) in gait are observed, which serve to prevent instability and falling.

The senses in question include, but are not limited to, visual, vestibular proprioceptive, and somesthetic inputs. They have been thoroughly studied in the fields of posturography and gait analysis due to their well-known roles in standing and walking (Peterka, 2002; Peterka & Loughlin, 2004). However, the manner and extent to which each modality contributes depending on to the control of posture and gait in the healthy adult is still being investigated. Moreover, challenges can potentially arise to these sensory integration processes and neural networks as they can be compromised in the case of neurological deficits, which commonly occur in the post-stroke population. It is still unclear whether a sensory modality such as somatosensation is reweighted based on the availability of other sensory modalities (i.e. visual, vestibular). Thus, the role of an essential sensory modality such somatosensation should be elucidated from both neurophysiological and functional biomechanical perspectives. To address this, a common approach is to compare and contrast healthy and neurologically impaired posture and gait.

2.4.2 Use of surface perturbations

Adding to the growing role of somatosensory inputs in gait and posture, researchers have made innovative use of physical perturbations in order to change walking outcomes in patient populations such as post-stroke. To accomplish this, projects often pair VR technology (Mirelman et al., 2010) with the perturbation cues (Punt et al., 2017). For example, researchers have used the constraining effects of changing surfaces to evoke changes in both static (Freyler et al., 2015) and dynamic posture while walking in a virtual environment. This can be through the use of moving platforms, such as multi degrees of freedom treadmills, to introduce perturbations for the purpose of changing walking conditions and orientations. The advantage of such strategies is that they can measure both the proactive and reactive aspects of functional gait and detail how the body reacts from both kinematic and dynamic perspectives.

Similar paradigms can be found, for example, when individuals walk along a split-belt treadmill or is suddenly met with a surface perturbation in the pitch orientation (Bugnariu & Fung, 2007; Fung & Perez, 2011). In such instances, one study reported that the healthy individual will display a muscle reaction response that is considered biomechanically efficient (Winter et al., 1990). Most notably, the muscles crossing the ankle will flex leaving the rest of the body relatively unreactive to the perturbation. Such a reaction would most likely ensure postural stability with a capacity to meet the challenges of the environment and task as a result. The healthy individual's reaction can be contrasted this with stroke individual's, as they tend to show a different postural response to the same perturbation. Specifically, the post-stroke individual's responses can be characterized as delayed onsets of muscle activations over many more body segments (Fung et al., 2003).

Such response mechanisms in maintaining stability during surface perturbations uncover unique challenges for clinicians and researchers, alike. The clinical challenge is how to mitigate, or even change these muscle synergies in response to postural destabilization to render the post-stroke individual with more stable and efficient postural reactions. Without the appropriate reactions, the body is left with the difficult task of maintaining an unstable dynamic posture while negotiating its environment. Such studies illustrate the importance of quantifying the differences in those of healthy static and dynamic posture compared to those with impaired stability such as a post-stroke individual.

2.4.3 Adaptation and post-adaptation – 'Broken Escalator effect'

An important concept in changing locomotor ability, which is central to the objective of the current thesis, is to elicit gait adaptations by means of walking constraints or alterations in the environment, to achieve positive effects on postural stability of the individual even after the adaptation process. A notable example of this is the 'Broken elevator effect' (Bronstein et al., 2009; Reynolds & Bronstein, 2003) reported originally by Reynolds and Bronstein (2003).

The authors found that as individuals habituated their locomotion to meet the needs of a sliding surface, the individual is likely to continue walking in the same adapted manner after surface suddenly remains still. This is the case even if the individual is pre-emptively aware that the surface will come to a halt. What researchers believe is taking place is that the regular motor program that would be responsible for generating the efficient cyclic movement of locomotion and keeping dynamic stability on a fairly regular surface, has now been transiently altered to meet the needs of stable locomotion on a sliding surface. This alteration, in response the perturbation brought on by the sliding surface has constrained the CNS to shape a motor plan to account for and mitigate the perturbing effects of the altered surface thereby maintaining postural control. To achieve this, the CNS must integrate the sensory information available, including somatosensory, visual and vestibular inputs (Bunday & Bronstein, 2009) to shape an altered motor program by using a feedforward loop that compares what the body must be doing to reduce errors and the efferent copy of predictive feedback of the normal movement (Afzal et al., 2015). This feedback loop will eventually be responsible for narrowing the error caused by the altered walking surface, ultimately changing the way the person walks. This principle of walking adaptability has a very important role in the studies presented in this thesis.

As dynamic stability adapts to different walking conditions, such as a sliding surface, researchers find that the CNS does not immediately revert to the ‘normal’ walking pattern program once the environment returns to a stable state - prior to a ground perturbation, for example. Rather, the individual tends to continue employing the altered walking strategy for a transient period in what is considered a state of post-adaptation. In fact, during post-adaptation the individual is in a state of actively unlearning the adapted program if they are resuming locomotion under original conditions.

What can be seen here are two important phenomena - one is that once the individual adapted to the altered environment the CNS had to alter the motor program that did not simply disappear immediately after the surface perturbation disappeared. This phenomenon is an example of post-adaptation (Patel et al., 2014). The second is that the post-adaptation effect is present, even if the individual is cognitively aware of the perturbation onset, thus reinforcing the notion that the post-adaptation is rooted in the changes made in motor control. Authors also underlined that the relative ease with which a post-adaptation effect can be elicited suggests a strong dissociation between the knowledge of altered conditions and the action that is taken at the level of motor control (Bronstein et al., 2009).

Researchers believe that action taken as a result of changes in motor control takes place in both cortical and cerebellar structures (Bronstein et al., 2009) among other brain regions. Additionally, researchers find that when transcranial current stimulation is presented to the motor (M1) and premotor cortices, post-adaptation effects are prolonged during similar broken escalator paradigms (Kaski et al., 2012). This would provide evidence that the brain is plastic to reshape the way someone interacts with a particular environment. What is also pertinent to the current thesis, as far as neurological populations such as post-stroke are concerned, is that the impaired integration somatosensory or proprioceptive input that the individual could experience may not affect the possibility of exhibiting a post-adaptation effect. Rather, the loss of somatosensory input can affect the manner in which the adaptation and overall motor program is shaped and maintained (J. J. Jeka & Lackner, 1994; Martinelli et al., 2015). It is believed that this can come as a consequence of a modified process in the feedforward process previously described. In light of these findings, follow-up paradigms, such as the one presented in the current thesis should compare and contrast any differences in the adaptation and post-adaptation parameters for both the target and control populations.

The adaptation and post-adaptations reported in the broken elevator effect and in related paradigms potentially provide a new perspective on the potential to rehabilitate post stroke and other populations by enforcing adaptations and pushing the limits of post-adaptation, as therein lies a potentially efficacious strategy for longer term rehabilitative changes such as functional walking outcomes to the level of safe and functional community dwelling. In fact, the underlying paradigm across studies in the present thesis follows the pre-condition, condition, post-condition formula to investigate the extent to which locomotor adaptations and post-adaptations are present as a result of exposure to a haptic tensile force applied to the hand.

2.4.4 Split-belt treadmill

Also emerging from research are other novel ways of putting the body in a position to manage an altered walking environment (Tyrell et al., 2015). A prime example is the use of the split-belt treadmill. Such treadmills feature two belts powered with separate motors, capable of going at separate speeds (Mawase et al., 2013). This allows for training limb discrepancies as the individual walks. A typical paradigm may have an individual walking on the treadmill at an even pace between belts, after which the belts can be programmed to run at different paces. While the individual walks on the platform with two belts at different paces, the two limbs must train with different limb coordination (Malone & Bastian, 2014). This adaptation process bears similarities to the broken escalator paradigm (G. Torres-Oviedo et al., 2011). It is needed to change the motor program that is otherwise dedicated to walking under normal circumstances. If after a certain amount of time (i.e. to the order of minutes) the belts are then to return back to even pace, then the potential for post-adaptation can be measured. This process could have very interesting effects on individuals with asymmetrical imbalances of left and right sides of the body, such as post-stroke individuals exhibiting hemiparetic gait.

It is reported that post-stroke individuals walking on the split-belt treadmill with a shorter step on a faster paced belt can create an adaptation of step length symmetry (Nadeau et al., 2015). It is well to note that the cadence of the paretic leg, that is the number of strides per minute, is typically lower than that of the non-paretic side. On the split-belt treadmill, the effect achieved is that the paretic side is now constrained to move as fast as with the non-affected side, effectively adapting to the disparity of the belts speeds (Mawase et al., 2013; Mawase et al., 2014). As the stroke individual is exposed to this adaptation phase, the motor program in the CNS is now reshaping as a result of the disparity detected in the sensorimotor feedback loop (Mawase et al., 2014; Reisman et al., 2005; Reisman et al., 2007; Vasudevan & Bastian, 2010). The CNS must deal with this disparity if the individual is to maintain their dynamic stability on the treadmill and avoid potentially unstable postures or even falling.

Perhaps even more interesting is that when the is split-belt pacing is restored to equal speeds, the stroke individual does not return immediately to original paretic walking patterns prior to the split-belt paces changes (Reisman et al., 2005). Just as in the broken elevator effect, Bastian et al. (2011) reports that the post-adaptation effect of the paretic leg can be clearly seen. The newly adapted paretic leg keeps the same dynamic and kinematic qualities during the gait stride even when the belt paces have been equalled (Bastian, 2011). This effect does improve walking asymmetry (Reisman et al., 2013), particularly over repeated exposures, even if to date it is transient and diminishes over a short period of time. For example, one study found that a deadaptation of the lower limb can be seen after 20 to 40 steps (Lam et al., 2006), after adapting to walking with resistance on a treadmill. Lastly, a case study reported considerable changes in self-selected walking speeds of up to 0.15 m/s after one month of training on the split-belt (Reisman et al., 2010).

Despite the impermanent nature of this effect, such studies demonstrate alterations in the motor program for walking can be achieved to the potential benefit of paretic gait rehabilitation. Also encouraging is that follow-up studies have reported the ability for post-stroke individuals to transfer split-belt post-adaptations to over ground walking (Reisman et al., 2009b; Gelsy Torres-Oviedo & Bastian, 2012). Repeated exposure protocols have been devised to better understand what the potential benefits can be with reinforcing these post-adaptations to have individual take away a more permanent alteration of the motor program as a consolidated motor program (Helm & Reisman, 2015). As mentioned, perhaps a challenge to this field and potentially very insightful avenue for research would be the use of imaging techniques to underline at the cortical activations present during these adaptation and post-adaptation phases. Both researchers and clinicians can use this insight to better understand the areas of the brain responsible for such adaptation and post-adaptation effects. This in turn can lead to well-devise therapeutic strategies that specifically target and reinforce desired movement patterns by challenging these CNS areas. The split-belt can indeed provide valuable insight into how the brain can reshape motor programs that even address the intra-limb symmetry aspect of paretic and non-paretic limb, which otherwise would have been deemed difficult to alter given the stroke individual’s condition and functional level.

2.4.5 Haptic stimuli

Another important approach involves altering or enhancing the sensory modalities, such as vision, somatosensation and auditory to better understand the function of static and dynamic postural stability, and ultimately improve it. This dissertation places a special focus on the effects of somatosensation, and in particular haptics, which refers to the sense of touch. It should be noted that while an emphasis is placed on somatosensation, the effects of visual, auditory as well as vestibular inputs, when available, continue to play an important role in the sensorimotor continuum of movement and should be considered when evaluating locomotor behaviours. As such, it has been repeatedly found in the literature that somatosensory input contributes to maintain balance and posture (Bolton et al., 2011). For the purpose of this thesis, the term somatosensation can be divided into two components. Firstly, haptic information, which includes the detection of tactile inputs, such as touch, pressure, vibration, temperature, or pain. The second being proprioception, or kinesthesia in the context of motion, which is responsible for detecting the spatial position and movement velocity using receptors grouped in joints and muscle tissue.

Researchers have shown that haptic information is essential for maintaining static and dynamic posture (Dickstein & Laufer, 2004; John J. Jeka, 1997; Kodesh et al., 2015) and influence both spatial and temporal aspects of the gait cycle (Karunakaran et al., 2014). For example, an earlier study describes the role of proprioceptive inputs in regulating postural tone (Franzen et al., 2011) as well as underlining the muscular torques and energy demands imposed on the body, while level or inclined (Oates et al., 2008). Thus, a major concept for the current project emerges from the literature relating the use of tactile cues to the control of normal posture. For example, Jeka et al. (1997), and Jeka & Lackner (1994), found that the body effectively counters postural sway reflexively when light touch and proprioceptive inputs are available (J. A. Barela, Jeka, J.J., Clark, J.E. & more, 1999; J. J. Jeka & Lackner, 1994; J. J. Jeka et al., 1997; Lackner et al., 1999). Researchers have also shown that light touch can reduce postural sway to a similar extent that vision does, in the absence of vision (Rabin et al., 2013). Vaugoyeau et al. (2008) also found that healthy adults tend to use proprioceptive cues while moving in support or dynamic postural conditions (Vaugoyeau et al., 2008). Even when tactile cues originating from interpersonal touch, such as finger-to-finger contact (e.g. a non-fixed contact), generally reduced sway compared to standing alone (Johannsen et al., 2007, 2012). Moreover, it has even been shown that the direction of sway will be orientated in the same direction as the stimulus (J. J. Jeka et al., 1997) , just as Lee et al. (1975) found with optic flow and vision (D. N. Lee, 1980; D. N. Lee & Lishman, 1975). Most recently, research with similar findings related to greater postural stability and haptic feedback suggest that older individuals may require slightly greater forces to reduce sway variability (A. Barela et al., 2017).

From a functional standpoint, haptic cues are processed relatively quickly (~50 ms), and thus are generally effective in stabilizing posture in neurologically impaired populations such as stroke (S. Lee et al., 2015) despite their compromised stability. There is evidence suggesting that for stroke individuals, deficits in joint movement, body proprioception, and other somatosensory stimuli can have the same debilitating effects on quiet stance as vision does. When discussing the use of proprioception with respect to the neurological population, specifically in the post-stroke population, studies report that post-stroke individuals show larger centre of pressure (COP) areas compared to healthy control subjects. However, the stroke subjects also exhibit attenuated postural sway patterns similar to those shown by the control group suggesting that post-stroke individuals have the capacity to employ somatosensory inputs when using light touch in a similar manner as healthy controls (Afzal et al., 2015). This robust effect has be replicated in different populations such as those with Parkinson's (Rabin et al., 2013) when they performed both light and firm finger contact. Haptic information has a similar role in facilitating both normal posture (J. A. Barela et al., 1999; Clapp & Wing, 1999) and gait (Fung & Perez, 2011). For example, Dickstein et al. (2004) found that light finger contact with a stable object during locomotion had a similar affect to firm finger contact, or vision, on attenuating sway and reducing COM deviation (Dickstein & Laufer, 2004).

The current thesis aims to add these findings by investigating the effects non-fixed, active tensile forces, such as the ones presented in all three studies could achieve such stability. Hence, the next section will focus on the effects of non-fixed haptic stimuli, such as the cane and the rehabilitation dog can facilitate dynamic stability. In all, there is sufficient evidence from these findings that a broad range of tactile cues effectively attenuate posture to merit such investigation.

2.4.6 Use of Cane and Rehabilitation Dog

Among neurological populations, researchers also show that greater stability and muscle activation in the paretic limb are exhibited when post-stroke subjects walk while holding a cane (Buurke et al., 2005). Reports show that post-stroke individuals, particularly those with stabilized gait walking just below the 0.8 m/s community dwelling threshold, tend to benefit from walking with a cane (Nascimento et al., 2016). A study Perez & Fung (2011) also reported that post-stroke subjects using an instrumented cane as a haptic cue and showed improvements in walking velocity (Perez & Fung, 2011). These results are promising results since positive changes in velocity have proven difficult outcomes for post-stroke subjects.

Lastly, an innovative pilot study investigated the changes of walking outcomes when an individual post-stroke walked with a specially trained rehabilitation dog (Abbud et al., 2014; Rondeau et al., 2010). The dog was equipped with a harness leash and guided the individual as they walked together. The guided support by the dog meant that the post-stroke individual walked on average faster over 10 meters compared to walking alone with a cane. The 10 meter time was further reduced over subsequent walking sessions and post measures (Rondeau et al., 2010). Despite these encouraging findings, it remains to be seen if walking under similar conditions, using a similar haptic input can cause adaptations and ultimately post-adaptations to the gait cycle post stroke. Such findings could prove a promising finding in clinical circles. The current dissertation aims to fill this knowledge gap.

2.5 Hypotheses

In light of the mounting evidence of adaptation and post-adaptation effects present in gait parameters, resulting from exposure to haptic inputs in a VR setting. The three studies detailed in this dissertation were designed to test the hypothesis that adaptation and post-adaptation effects come as a result of being exposed to a forward-leading tensile haptic force. The hypotheses for each of the three studies in this project are listed below.

Study 1 A) In a healthy young adult population, walking with tensile forward-leading haptic force applied to the hand for up to a minute, compared to no force, will change spatiotemporal outcomes such as increased gait velocity and stride length as well as decreased double limb support time.

B) These changes will be maintained for up to 60 seconds *after* force removal.

Study 2 A) In both elderly post-stroke and elderly healthy age-matched populations, a tensile forward-leading haptic force applied to the hand, compared to no force, increased gait velocity and stride length while decreased stride time and double limb support times, with accompanying centre of mass excursions, both during and the onset and offset of force.

B) These changes will be maintained for up to 60 seconds *after* force removal.

Study 3 A) In elderly post-stroke and elderly healthy populations, walking with tensile forward-leading haptic force applied to the hand, compared to the use of an instrumented cane, will change spatiotemporal, postural and kinematic gait outcomes.

The hypotheses for three distinct studies aim to progressively explore and ultimately establish growing evidence of the efficacy of using tensile forward-leading forces to promote functionally changed walking in chronic-stroke. In doing so, the global aim of this dissertation is two-fold. First, to contribute to strong evidence that motor learning can be achieved, as evidenced adaptations and post-adaptation effects, in healthy and post-stroke populations. Second, is to compare its efficacy to a widely used walking aid, such as the cane. Ultimately, the result obtained would suggest evidence of an innovative and promising alternative for gait rehabilitation that is functional in the context of community dwelling. Thus, the objective of the following chapter is to address the first hypothesis. Namely, to establish proof of concept that haptic forces change gait parameters in a healthy young population.

CHAPTER 3: MANUSCRIPT 1

ADAPTATION AND POST-ADAPTATION EFFECTS OF HAPTIC FORCES ON LOCOMOTION IN HEALTHY YOUNG ADULTS

Gianluca U. Sorrento, MSc, PhD (c) [§1-2],

Philippe S. Archambault, OT, PhD[1-2],

Joyce Fung, PT, PhD[1-2]

[1] School of Physical & Occupational Therapy, McGill University, Montréal, Québec, Canada;

[2] Centre for Interdisciplinary Research in Rehabilitation of Greater Montreal (CRIR), Jewish Rehabilitation Hospital, site of CISSS-Laval, Québec Canada.

[§]Corresponding author

GS: gianluca.sorrento@mail.mcgill.ca

Reprinted with permission from the Journal of Neuroengineering and Rehabilitation

J Neuroeng Rehabil. 2018 Mar 13;15(1):20. doi: 10.1186/s12984-018-0364-0.

Article accepted for publication on March 13, 2018.

3.1 Abstract

Background: Developing rehabilitation strategies to improve functional walking and postural control in patients is a priority for rehabilitation clinicians and researchers alike. One possible strategy is the use of sensory modalities to elicit adaptive locomotor gait patterns. This study aimed to explore to what extent haptic inputs, in the form of forward-leading tensile forces delivered to the hand, compared to no force, may lead to adaptation and post-adaptation effects on gait parameters, during and after the haptic exposure, respectively.

Methods: Thirteen healthy young individuals were recruited for this study. We developed an innovative system combining virtual reality and haptic tensile forces in the direction of locomotion to simulate walking with a dog. A robotic arm generated forces via an adapted leash to the participant's hand while they walked on a self-paced treadmill immersed in a virtual environment with scene progression synchronized to the treadmill.

Results: All participants showed significant increases in instantaneous gait velocity and stride length, with accompanying decreases in double limb support time ($p<0.05$) when walking with a haptic tensile force of either 10 or 20N, relative to pre-force epoch levels, indicating an adaptation effect. When the 10 or 20N force was removed, gait measures generally remained changed relative to baseline pre-force levels ($p<0.05$), providing evidence of a post-adaptation effect.

Conclusions: Changes in spatiotemporal outcomes provide evidence that both adaptation and post-adaptation effects were present in response to the application and removal of a haptic force. Future studies will investigate whether similar changes in elderly and post-stroke populations can be actualized during steady-state walking.

Keywords: Gait; Locomotor adaptation; Haptics; Virtual reality; Sensorimotor integration

3.2 Introduction

In physical rehabilitation, independent walking is an important outcome for reintegration into the community (R. W. Bohannon et al., 1991; Lord & Rochester, 2005; Pang et al., 2007). To achieve this, both gait and postural reactions must be performed in a safe and functional manner. This dynamic stability depends not only physical but environmental and task dependent circumstances (Fung et al., 2006; Haley et al., 1994). Safely crossing a busy street intersection, for example, involves all these factors. Maintaining dynamic stability requires responses that are both proactive for adapting to the environment and reactive to perturbations and obstacles that may be present in the surroundings (Muir et al., 2015; Nieuwenhuijzen & Duysens, 2007). This dynamic stability can be achieved when the individual's gait and posture are recovered well enough to produce the required gait speed (Fulk et al., 2017; Fulk et al., 2010; Verghese et al., 2011), symmetry and postural reactions (de Oliveira et al., 2008) needed to walk in an effective and safe manner to avoid collisions and falls (Cheng et al., 2001). These spatiotemporal parameters help predict falling in populations with reduced mobility, such as older adults and stroke survivors (Fulk et al., 2010). In fact, these two populations have alarmingly high incidences of falling: annually, falls can occur in 35% - 45% of individuals older than 65 years (Soriano et al., 2007) and increases with health risk factors (Tinetti et al., 1994), while other studies report a similar incidence in post-stroke populations (Nyberg & Gustafson, 1995b; Teasell et al., 2002).

Researchers and clinicians are addressing this challenge with strategies that focus on training functional gait adaptations (Malone et al., 2011; Wutzke et al., 2013). These adaptations, in the context of procedural learning, occur as an error-driven learning process that alters well-established motor patterns in the brain (Malone et al., 2011; Shadmehr & Mussa-Ivaldi, 1994; G. Torres-Oviedo et al., 2011). This process is useful when an individual must adjust to different physical, environmental and task demands. In the context of gait, there is evidence to suggest such adaptations may be driven to restore the dynamic stability (Cajigas et al.,

2017) necessary to safely reintegrate individuals into community dwelling. Two important adaptation strategies are relevant to the present study. One involves creating spatiotemporal constraints, such as modified walking surfaces (Bronstein et al., 2009; Reisman et al., 2007; Tang et al., 2013). The other employs multisensory stimuli (Finley et al., 2014; G. Torres-Oviedo & Bastian, 2010; Turchet et al., 2013), such as somatosensory cues (Afzal et al., 2015; Dickstein & Laufer, 2004; Kodesh et al., 2015).

Perhaps more significant to functional recovery is the extent to which locomotor adaptations can be prolonged after the exposure is removed (Bastian, 2008). This particular retention can be conceived as a post-adaptation, or a 'carry-over' response in locomotion. Adaptation and post-adaptation, once used to describe the 'broken elevator effect' by Reynolds and Bronstein (2003), have been investigated in subsequent studies (Bronstein et al., 2009; Kaski et al., 2012; Reynolds & Bronstein, 2003). Researchers have often used the split-belt treadmill to investigate this phenomenon. Essentially, the split-belt treadmill has two separate belts that can move both legs in tandem and at different speeds or directions. These changes can constrain or change bilateral locomotion. For example, Choi et al. (2007) suggest that the functional circuitry of both legs can be trained separately (Choi & Bastian, 2007), potentially mitigating the effects of asymmetric gait (Malone & Bastian, 2014). Keeping with this approach, researchers examined post-adaptation changes in gait on a split-belt treadmill for healthy individuals and for those with neurological deficits (Reisman et al., 2013; Reisman et al., 2007). For example, Reisman et al. (2007) reported a reduction in spatiotemporal discrepancies between the paretic leg and non-paretic leg in post-stroke individuals during post-adaptation responses (Reisman et al., 2007). These changes were found after participants adapted to different treadmill speeds. This post-adaptation has also been successfully transferred to overground walking (Reisman et al., 2009b; Savin et al., 2014). Yet, it is less clear as to what extent this strategy could induce longer lasting post-adaptation effects (Bastian, 2008). Researchers believe that access to adapted motor patterns remains available in the CNS, despite transient effects and eventual unlearning (Krakauer et al., 2005). Hence, the potential for lasting post-adaptations is possible, particularly with repeated exposure to a given stimulus (e.g. as part of a rehabilitation intervention).

Haptic cues may constitute another approach for retraining dynamic gait stability. For example, there is evidence that light finger contact with an earth-fixed object during locomotion can have an attenuating effect on sway and reduce postural deviations (Dickstein & Laufer, 2004; J. J. Jeka & Lackner, 1994). Researchers also reported similar effects of light finger contact with a static external object in terms of greater postural stability in the lower limb (Kodesh et al., 2015). Moreover, Fung & Perez (2011) reported that the use of an instrumented cane for both post-stroke and healthy participants holds ecological validity with the potential of carry over to overground walking (Perez & Fung, 2011). Yet another strategy for providing helpful haptic cues to chronic stroke individuals is the use of a specially trained rehabilitation dog. In one study, chronic stroke survivors increased gait velocity and decreased gait variability when walking with a trained rehabilitation guide dog (Rondeau et al., 2010). In light of these previous studies, the current study uses an innovative and novel approach combining haptic tensile forces in the form of a ‘virtual leash’ within a virtual environment in order to investigate the possibility of adaptation and post-adaptation effects in human gait.

Specifically, the research question is: in young healthy individuals, to what extent does a haptic tensile force applied to the hand during steady-state walking, compared to no force, change spatiotemporal gait outcomes? To answer this question, two hypotheses have been presented to address the possibility of both adaptation and post-adaptation effects. The first hypothesis is that exposure to a haptic tensile force results in gait adaptation effects, manifested as increases in gait velocity and stride length and a decrease in double limb support time, relative to baseline levels. The second hypothesis is that the adaptation will be maintained even after removal of the tensile force, resulting in persisting post-adaptation changes in spatiotemporal gait parameters.

3.3 Methods

3.3.1 Participants

A total of 13 healthy young adults (18-38 years old, 7 male and 6 female) participated in the study. All participants were free of any musculoskeletal, neurological, or cognitive deficits (self-report) and did not require walking aids. All participants were right-handed, thus they all held the leash in the right hand. The ethics review board of the Center for interdisciplinary research in rehabilitation of Greater Montreal (CRIR) approved the study and all participants provided their informed consent.

3.3.2 Apparatus

Participants were fit into a safety harness and walked on a self-paced treadmill (0.8 x 1.7 m) while immersed in a virtual environment that was rear-projected onto a large screen, placed 1.5 m in front of the participant. They were given time to habituate their walking ability on the self-paced treadmill before data recording. The self-paced treadmill was custom-made on site and its motor speed was PID-controlled by an algorithm in a micro-controller using the distance signals obtained with an electro-potentiometer tethered to the back of the walking participant, as well as the first derivative of the distance (speed). The treadmill's speed is adjusted instantaneously to keep the participant positioned at the center while walking (Fung et al., 2006). The virtual scene's progression was in turn controlled and synchronized with the treadmill's speed by the CAREN-3 system (Motek BV, Netherlands). A latency between the treadmill and the CAREN system was marked at 30 ms. The virtual scene was kept contextually relevant with the physical task by featuring a sidewalk in an urban setting, with an animated dog. The dog moved in the same direction and pace as the participant (Fig. 3.1).

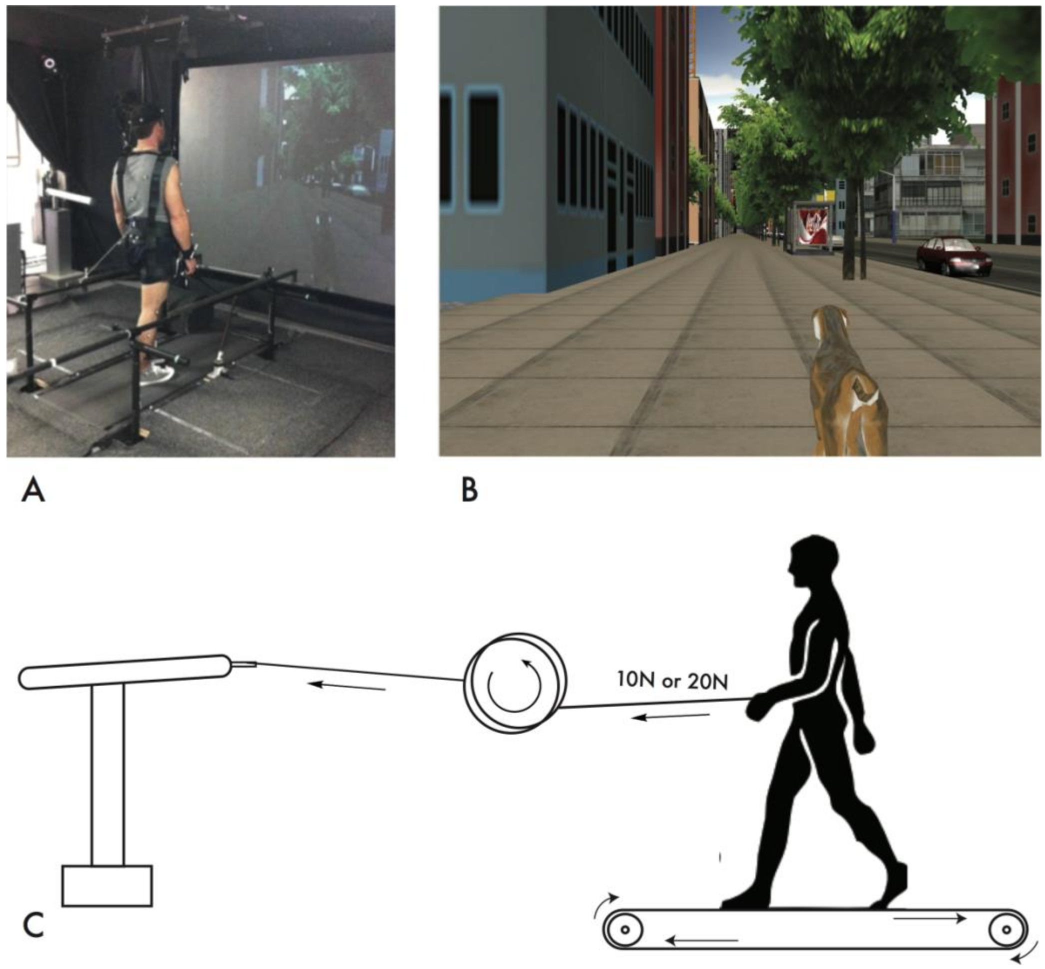


Fig. 3.1: Experimental setup.

A: Virtual reality setup with HapticMaster, self-paced treadmill and virtual environment back-projected onto a screen in front of the treadmill. B: Synchronized virtual reality scene featuring dog and city-scape. C: Schematic of HapticMaster robotic arm, pulley, leash, and self-paced treadmill.

Participants were unexpectedly exposed to a step change in forward pulling force from 0N to either 10 or 20N applied to the hand. This was achieved via a leash and pulley system anchored to a force-controlled haptic robotic arm (HapticMaster, Moog BV, Netherlands) (See Fig. 3.1 and 3.2). To take into account any possible friction or stretch in the system, force levels were pre-calibrated at the hand and were transmitted through to a steel, stretch-resistant leash.

A camera-based motion capture system (six cameras, Vicon, UK) was used to record three-dimensional (3D) positions of reflective markers firmly attached to the body landmarks necessary to measure gait analysis outcomes in this study. All reflective markers were placed on the participant in accordance with the Vicon plug-in-gait© 42 marker set-up. Extra markers were placed on the leash and treadmill frame. Kinematic data were recorded at a frequency of 120 Hz in the Workstation gait analysis program (Vicon, UK).

3.3.3 Constant Force Paradigm

The paradigm used to generate a step change in force is illustrated in Fig. 3.2. The paradigm was divided into three distinct gait epochs; pre-force, force, and post-force (see Fig. 3.2). Prior to the experiment, participants were briefly habituated on the self-paced treadmill and were instructed to walk at a comfortable, self-selected pace throughout the experimental paradigm. The pre-force epoch consisted of the participant walking with a slack leash (i.e. no tension) for 30 seconds. For analysis purposes, this epoch corresponded to baseline walking levels. An instantaneous step increase of force from 0N to either 10 or 20N was then transmitted to the participant's hand via the robot-controlled leash. Participants walked with this force for 60 seconds. This corresponded to the force epoch. The force on the hand was then removed and participants continued to walk with a slack leash for up to another 60 seconds. This was regarded as the post-force epoch. Each force condition (10 and 20N) was repeated twice, in a random order. The data analyzed within these epochs consisted of periods when steady-state walking was maintained at a pace consistent with daily overground walking. Hence, portions of walking data excluded from steady-state walking included the beginning and end of the trial, when participants accelerated into steady-state walking and when they started to decelerate to a stop. A period of one full gait cycle before and after the perturbation of force onset and offset was also excluded from steady-state walking.

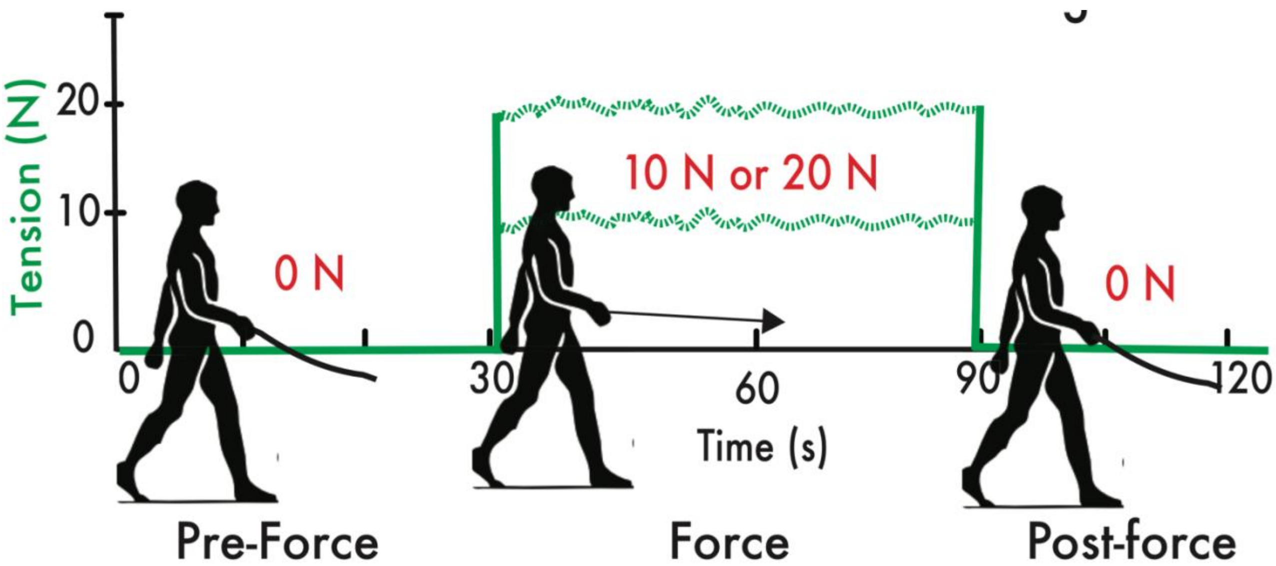


Fig. 3.2: Schematic of the force step change paradigm.

The green trace represents the haptic force controlled by the HapticMaster. Participants walk with 0N during the pre-force epoch between 0s – 30s. Tension is then delivered to the hand during the force epoch. This tension could be either 10 or 20N of force. The force is then released at the 90 s mark as walking continues in the post-force epoch.

3.3.4 Data Recording and Analysis

Customized control software was developed to control the HapticMaster (Moog, Netherlands) robotic arm. For the beginning and ending of each walking trial, an instantaneous analog pulse was sent from the CAREN system, and was simultaneously received by both Vicon and HapticMaster systems. This enabled the synchronization of trial events across hardware platforms. This synchronization was necessary to define the critical time points of step-change force onset and offset that define pre-force, force, and post-force epochs. Motion capture from the six camera Vicon © set-up recorded a full plug-in-gait 42 reflective marker setup (see index II). These 3D kinematic data were recorded at 120 Hz. After data collection, marker trajectories were reconstructed and processed offline using the Vicon Workstation software package.

Instantaneous gait velocity was derived from the 3D kinematics data and averaged over the course of a stride, using custom Matlab (MathWorks) routines. Specifically, it was calculated via the reflective marker position of the ankle by taking the stride length divided by the stride time, based on the 120 Hz frame rate. Stride length was defined by the distance between toe lifts of the same foot. Bilateral stride lengths were recorded and paired t-tests were conducted to compare stride lengths of the two legs. Double limb support time was defined as the segment of the stride between the ipsilateral initial contact and the contralateral toe lift.

Descriptive statistical analysis for each of the three outcomes involved percent change relative to pre-force levels. Absolute mean values were then used for statistical comparisons. We used a mixed model 2 x 3 repeated measures ANOVA taking into account force (10 and 20 N) and epoch (pre-force, force, post-force) condition levels. The covariance structure used was the variance components model, as it was found to be the most appropriate structure given the sample size (Guo et al., 2013). Post hoc comparisons with Bonferroni adjustments were then used to investigate changes between the three epochs and two force conditions. Significance was accepted at $p<0.05$. Also, a separate paired t-tests were conducted to compare stride lengths between the two feet.

3.4 Results

3.4.1 Instantaneous Gait Velocity

Average instantaneous gait velocity for the 10 and 20N conditions depicted in Fig. 3.4 reveals all 13 participants walked faster with tensile forces during the force epoch relative to the pre-force epoch. Specifically, instantaneous velocity jumped from 1.0 ± 0.2 m/s during the pre-force epoch to 1.1 ± 0.2 m/s during the force epoch for the 10N condition and 1.0 ± 0.2 m/s to 1.2 ± 0.2 m/s for the 20N (see blue and red horizontal traces in Fig. 3.4A-B). When comparing the post-force and pre-force epochs, the average gait velocity across participants tended remain above baseline levels, despite an initial decrease of 4-5 strides in response to the force offset perturbation (Fig. 3.3C). On average, this corresponded to 1.1 ± 0.2 m/s after exposure to the 10N tensile force and 1.1 ± 0.2 after the 20N tensile force (see horizontal green traces in Fig. 3.4 A-B). A mixed model repeated measures ANOVA was used to assess the effects due to epochs and force conditions, where significant main effects were found ($F_{(2,24)} = 9.31$, $p<0.01$). Post hoc comparisons revealed significant increases in gait velocity in the force ($p<0.01$) and post-force ($p<0.05$) epochs compared to the pre-force epoch, as well as a significant increase in gait velocity from the 10N to the 20N conditions ($p<0.01$).

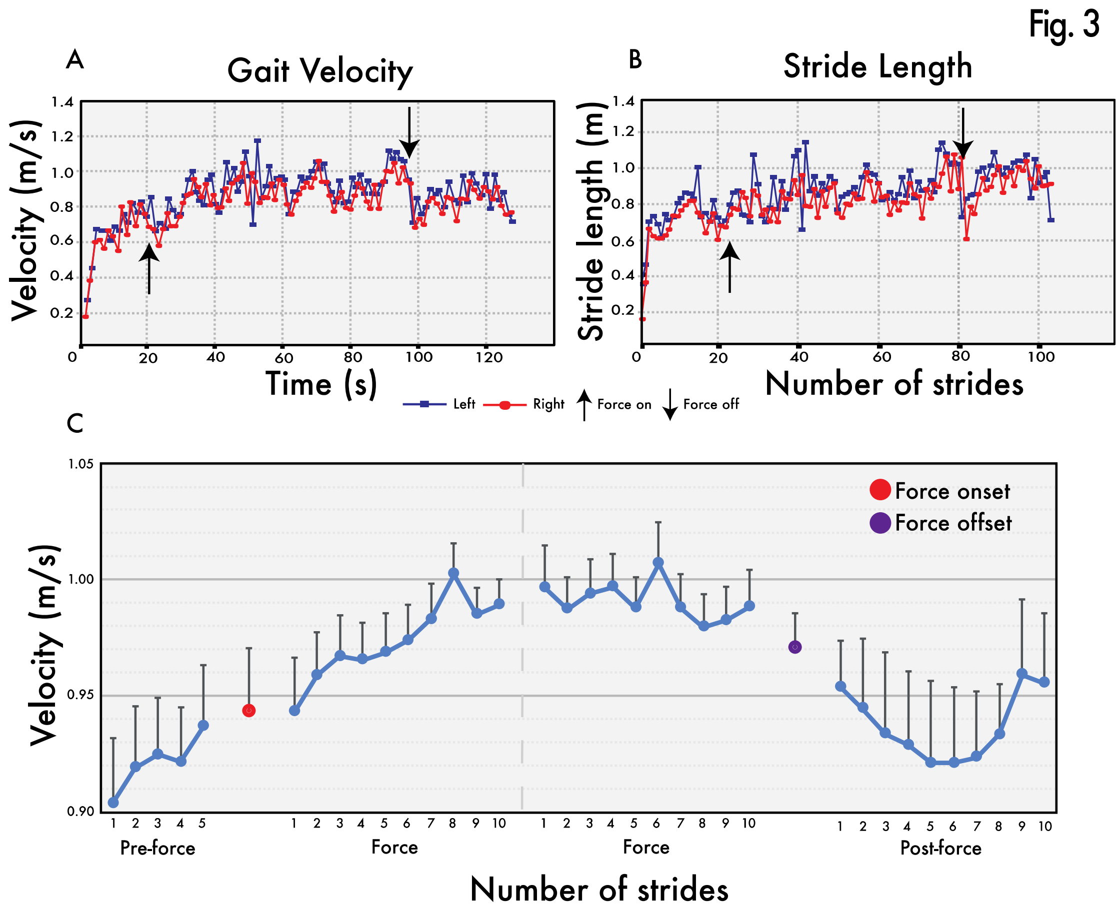


Fig. 3.3: Per stride gait velocity and stride length of a single participant and average gait velocity across participants.

A: Average instantaneous gait velocity and B: stride length, per stride, for both left and right limbs of a single participant. C: Average gait velocity across participants for the last five strides before (pre-force epoch) and first ten strides after 10N force onset (force epoch), as well as the final ten strides before (force epoch) and after force offset (post-force epoch).

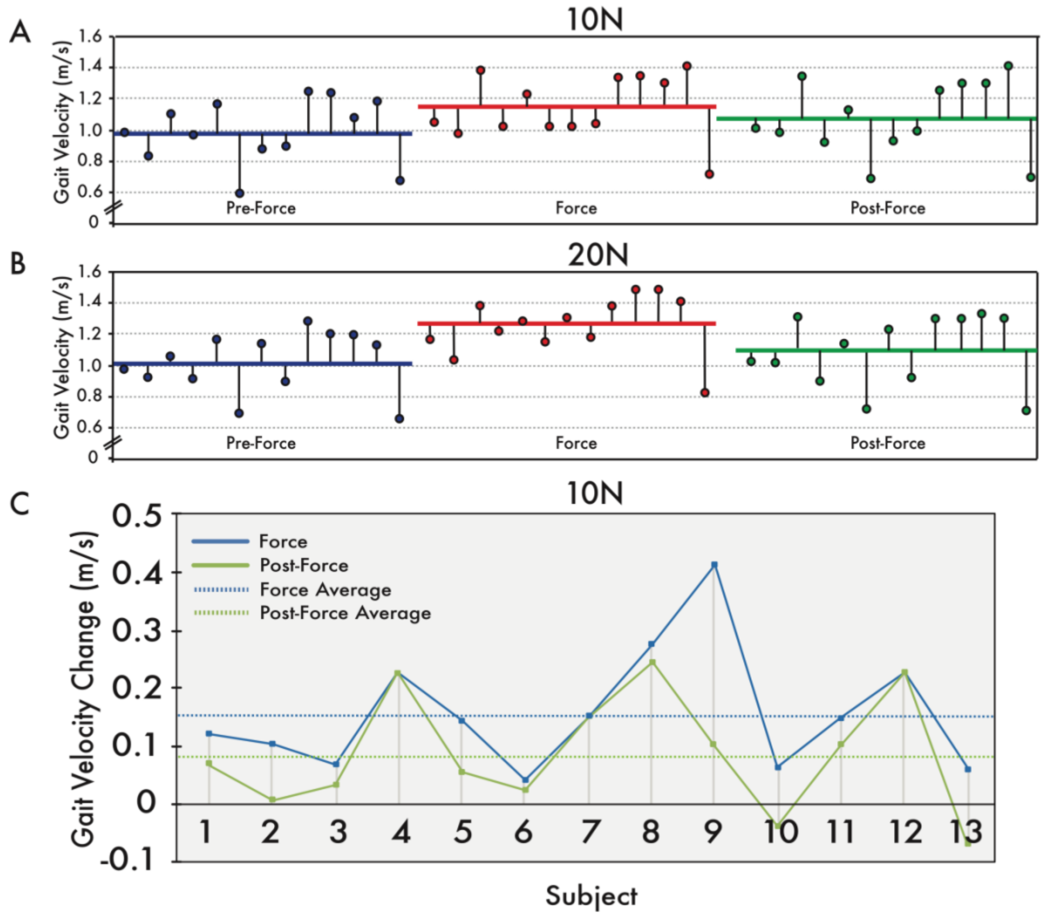


Fig. 3.4: Average gait velocity per epoch and average gait velocity change for each participant.

A-B: Mean instantaneous gait velocity (m/s) for each participant during the pre-force (blue), force (red) and post-force (green) epochs in the 10N and 20N conditions. The horizontal colored bars illustrate average group gait velocity. C: Average gait velocity changes in the 10N condition, for force and post-force epochs, relative to the pre-force epoch. Horizontal colored bars represent average velocity changes for force and post-force epochs across participants.

The boxplots in Fig. 3.5 A-B illustrate the interquartile median percent change of gait velocity during the force and post-force epochs relative to pre-force levels for participants in both 10 and 20N conditions. Relative to baseline levels, all 13 participants increased their instantaneous gait velocities when walking with tension in the leash in the force epoch with 17.8% and 25.2% changes in the 10 and 20N conditions, respectively (see Fig. 3.5 A-B). For the post-force condition, velocity still generally remained above baseline levels with an average increase of 9.0% and 7.0% in the 10 and 20N conditions, respectively, relative to pre-force levels (Fig. 3.5 A-B). A majority of the participants (11/13) demonstrated such post-adaptation changes.

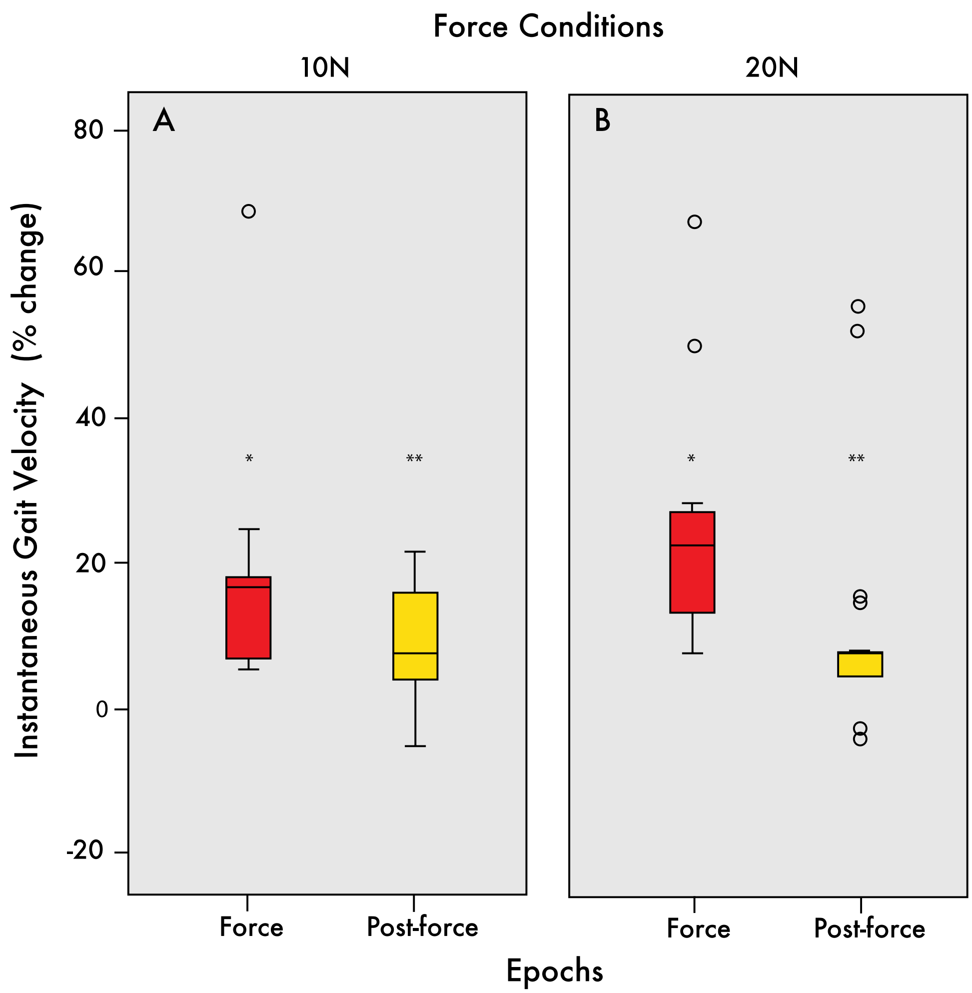


Fig. 3.5: Median percent change of gait velocity.

Box and whisker plots showing interquartile ranges and medians of instantaneous gait velocity percent change in 10N (A) and 20N (B) conditions, relative to pre-force. Gait velocity generally increased and maintained above pre-force baseline levels in 10 and 20N conditions. $^{**}p < 0.01$; $^{*}p < 0.05$.

3.4.2 Stride Length

An initial paired samples T-test revealed significant changes in stride length between legs ($p< 0.05$). As such, stride length was analyzed bilaterally according to the dominant leash side (right leg) and non-dominant side (left leg). The results of bilateral stride length seen in Fig. 3.6 A-B were similar to those of instantaneous gait velocity in that all 13 participants showed changes in stride length during the force epoch relative to pre-force baseline levels in both the 10 and 20N force conditions (Fig. 3.6A-B). These changes corresponded to approximately 10.6% increases for the left and right legs relative to pre-force stride lengths for the 10N condition and 13.0% and 13.6% differences in the 20N condition. Changes were maintained for the post-force epoch relative to pre-force levels for 10 of the 13 participants. Fig. 3.6A-B boxplots depict average changes of 6.5% and 6.6% in the left and right legs across all participants for the 10N condition relative to pre-force. In the 20N condition, post-force stride length percent change was maintained at 4.5% and 4.8% for the left and right legs. For the right leg, the mixed-model repeated measures ANOVA revealed a significant interaction of epoch and force conditions ($F_{(2,24)} = 6.17$, $p<0.01$). Post hoc comparisons showed significant changes in force epoch stride lengths ($p<0.01$) compared to pre-force stride lengths in both the 10 and 20N conditions. Only the 20N condition showed significant changes ($p<0.01$) in stride length between pre-force and post-force epochs. Similar post hoc comparison results were found for the left stride length ($F_{(2,24)} = 6.28$, $p<0.01$).

Stride length inter-stride variability for both legs was then computed with the coefficient of variation for all participants according to epochs in both 10 and 20N conditions. During the force epoch, the variability was reduced by 1.5% on the non-dominant left leg and 1.6% on the dominant right leg for the 10N condition relative to pre-force. Variability was also reduced by 1.2% and 1.4% on the left and right legs respectively in the 20N condition. During the post-force, smaller reductions were seen in the variability, with 0.1% and 0.6% for the left and right legs respectively in the 10N condition and a 1.2% increase and 1.0% decrease respectively in the 20N condition. Despite the changes in stride length variability, a one-way ANOVA revealed no statistical significance in stride length variability across epochs and conditions for either leg ($F_{(5,77)} = 0.25$, $p>0.05$).

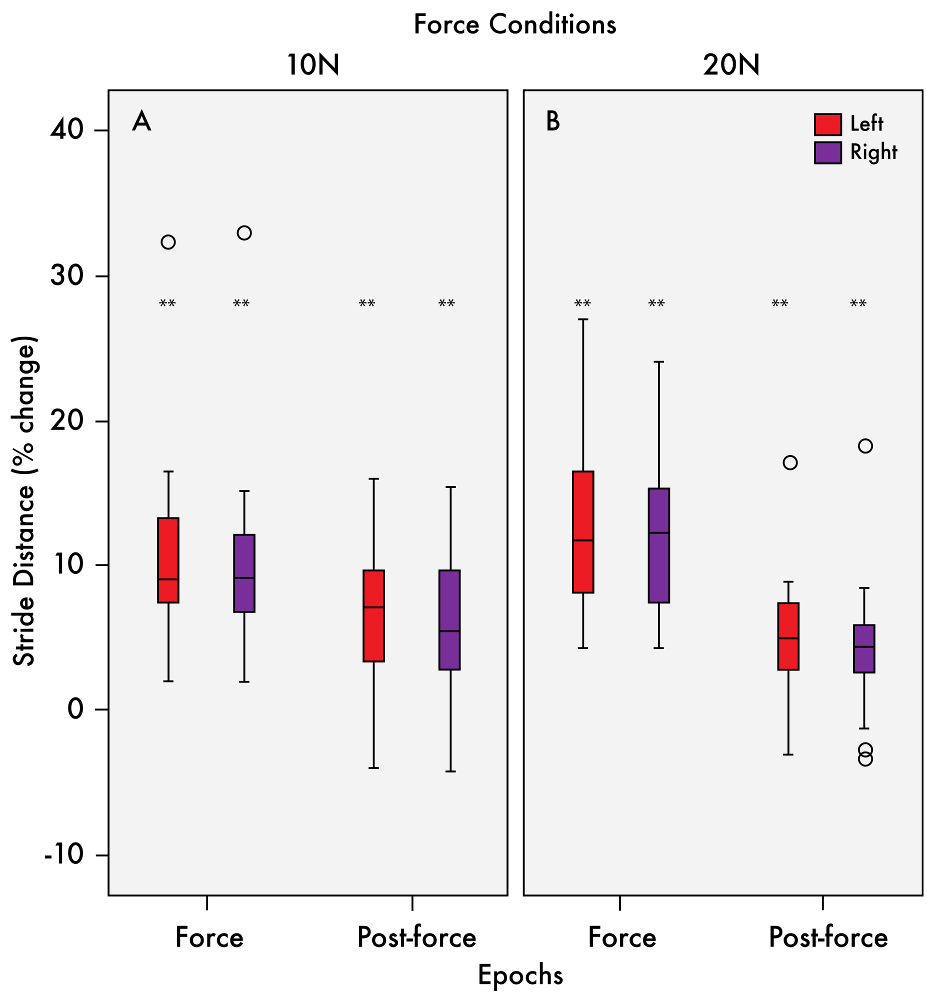


Fig. 3.6: Median percent change of stride length.

Box and whisker plots showing interquartile ranges and medians of stride length for the left (red) and right (purple) leg during the force and post-force epochs in the 10N (A) and 20N (B) conditions, relative to pre-force. Stride length increased in both epochs with respect to the baseline. **$p < 0.01$.

3.4.3 Double limb support time

Time spent in double limb support can be seen in Fig 3.7A-B. Double limb support times expressed as a percentage of change during the force epoch showed decreases relative to baseline pre-force levels; indicating less time spent in double limb support in either the 10 or 20N conditions. For example, when participants walked during the force condition, all 13 of the participants decreased their double limb support time. These differences correspond to an average decrease of 15.6% and 22.7% relative to pre-force values, for the 10 and 20N conditions respectively (Fig. 3.7A-B). Similarly, during the post-force epoch, almost all participants showed a decrease in double limb support times relative to the pre-force epoch. These changes in double limb support times were maintained at 7.5% and 7.9% below baseline in the 10 and 20N conditions, respectively, across participants. The same mixed-model ANOVA approach was used and showed once again a significant interaction between epochs and force conditions ($F_{(2,24)} = 3.85$, $p<0.05$). Post hoc comparisons revealed significant changes in double limb support time in the force ($p<0.001$) and post-force epochs ($p<.05$) relative to the pre-force epoch, in both the 10 and 20N force conditions.

An accompanying investigation of inter-stride variability was conducted on double limb support time. When comparing the force epoch to the pre-force, the variability was reduced by 6.5% in the 10N condition and by 7.1% in the 20N condition. The variability then remained lower in the post-force by 1.1% and 1.3% in 10 and 20N conditions, respectively. A one-way ANOVA revealed a significant decrease in variability ($F_{(5,77)} = 2.96$, $p<0.02$), where double limb support time during the 20N force epoch proved significantly changed ($p<0.05$).

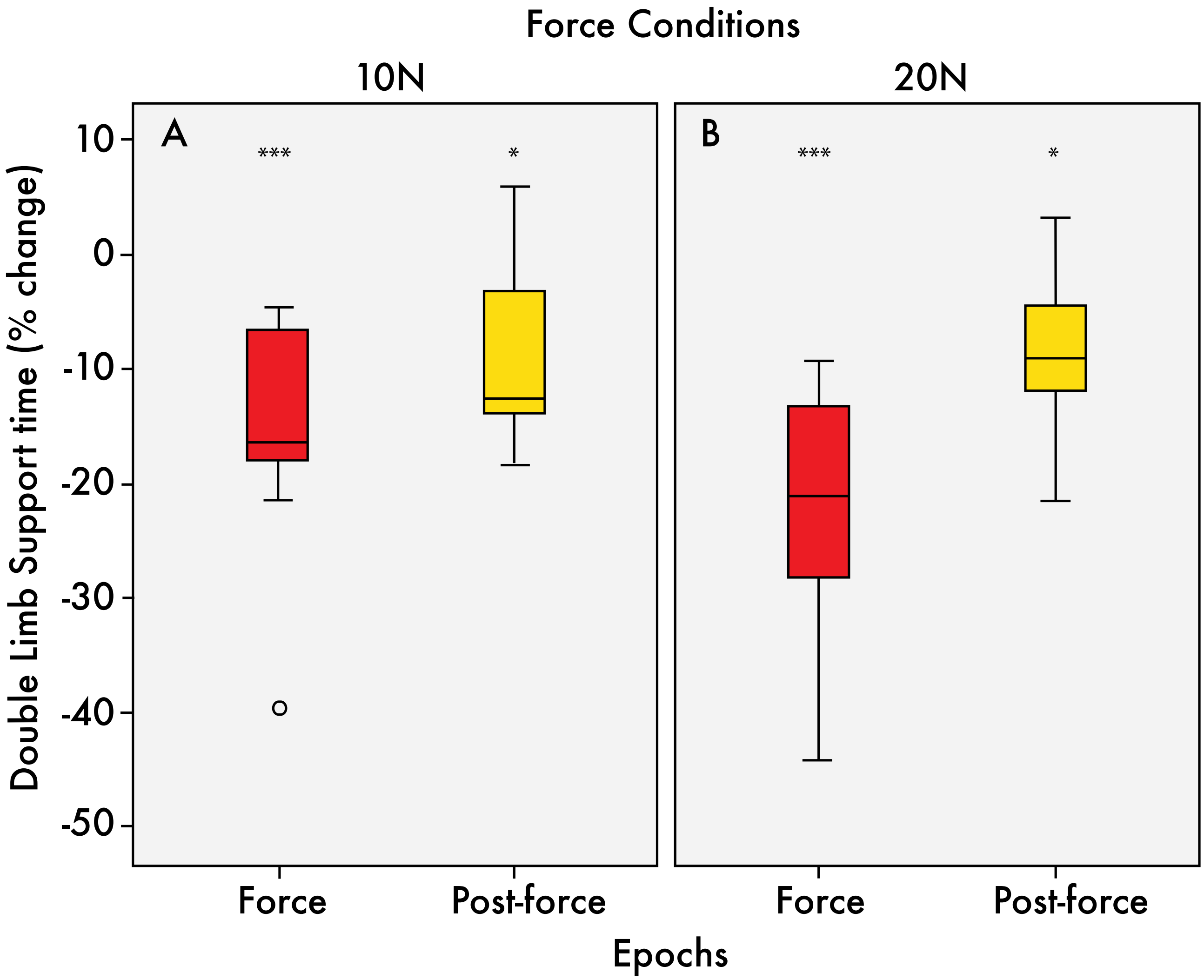


Fig. 3.7: Median percent change of double limb support time.

Box and whisker plots showing interquartile ranges and medians of instantaneous double limb support time in 10N (A) and 20N (B) conditions, relative to pre-force. Double limb support times generally decreased below the pre-force baseline levels during both force and post-force epochs. ***p < 0.001; *p < 0.05.

3.5 Discussion

We have investigated the extent to which haptic tensile forces applied to the hand, in the direction of walking, cause both an adaptation and a sustained post-adaptation response. As a first step, healthy young participants were recruited to establish a proof of concept by walking with and without haptic forces. In reference to the first hypothesis, walking with either 10 or 20N of force resulted in gait adaptation as evidenced by changes in spatiotemporal outcomes. Specifically, both gait velocity and stride length increased, while double limb support time decreased relative to pre-force baseline values. The second hypothesis on post-adaptation effects was also accepted, as the spatiotemporal gait changes were maintained relative to pre-force levels when the force was removed during the post-force epoch. The results suggest the evidence of both adaptation and post-adaptation effects in gait parameters due to haptic forces. It is likely that a transient, altered motor program formed during the adaptation epoch, as a consequence of the haptic tensile force, leads to lingering post-adaptation after-effects. It is not likely that the adaptation is due to a biomechanical change caused by a hauling effect on the hand. If this were the case, the adaptation effect would disappear immediately after force removal. Given that adaptations and post-adaptations were in fact present, such a sensorimotor enhancement strategy may be applied to neurological rehabilitation, such as gait enhancement post-stroke.

To this end, the spatiotemporal outcomes chosen for the study were intended to outline changes that are functionally relevant to the gait profile. Combining these outcomes can give some valid measures of both the quantity and quality of locomotion. Some fundamental findings have been highlighted in the results. The results are encouraging in that they seem to be consistent with findings from previous studies. For example, the present study provided evidence of adaptations and post-adaptations similar to those described by the broken escalator effect and the split-belt treadmill (Malone & Bastian, 2014). In addition, the use of a haptic force increased the capacity of dynamic stability and mobility, much like walking with a haptic strip or the instrumented cane to improve gait (Dickstein & Laufer, 2004; Perez & Fung, 2011).

The increase in gait velocity in this study is a clear example of an outcome exhibiting gait changes resulting from haptic forces. Namely, during the force and post-force epochs, all of the participants walked faster compared to pre-force baseline velocities. This can be illustrated with gait velocity changes, relative to the pre-force epoch, of 0.16 m/s and 0.10 m/s for force and post-force in the 10N condition and 0.23 m/s and 0.10 m/s for the 20N condition. From a clinical perspective, if such results were to be reported among a post-stroke population, a velocity increase of approximately 0.1 m/s would indicate clinically meaningful changes (Brach et al., 2010; Hornyak et al., 2012; Tilson et al., 2010). In the present study, healthy young participants increased gait velocity by approximately 18-25% and 9-8% in force and post-force epochs, respectively in 10 and 20N conditions (Fig. 3.5A-B). Given this rather robust finding, future studies should indeed investigate such changes in the elderly and post-stroke populations.

From a clinical perspective, changes above 0.1 m/s may represent clinically meaningful changes (Brach et al., 2010; Hornyak et al., 2012; Tilson et al., 2010), but it remains to be seen if post-stroke individuals are also able to increase gait velocity in excess of 0.1 m/s, either during or after force exposure, bearing in mind that they generally walk slower than healthy individuals. It is also notable that the average baseline gait velocity of 1.0 m/s in this study is slower than the normal overground walking speed, which may be attributed to walking on a self-paced treadmill in a virtual environment, as shown previously (Bayat et al., 2005; Powell et al., 2009).

Differences in stride length were seen between epochs and force epochs, and between the two legs. Bilateral stride length was a measure intended to describe the spatial and symmetrical qualities of locomotion. Evidence was found for changes in stride length between epochs. Specifically, all participants elongated their stride during the force epoch, while most maintained their stride above baseline during the post-force epoch. This was the case in both the 10 and 20N conditions. The findings corresponded to ~11-14% and 5-7% changes in both legs in the force and post-force epochs (Fig. 6A-B). These changes may be translated to clinically meaningful gait outcomes in stroke and older populations (Richard W. Bohannon & Glenney, 2014). Further analysis also revealed that the inter-stride variability also generally, although not significantly, decreased during the application of 10 and 20N forces.

Less time spent in the double limb support phase is a biomechanical consequence of increased gait speed. Per stride, this often translates to a faster, more secure dynamic posture that spends more time in the swing phase and less in stance phase, often leading to greater medial-lateral stability (LaRoche et al., 2014). Given the increases in gait velocity seen in this study, reductions in double limb time support may lend evidence for a changed gait, addressing both the quality and quantity aspects of the gait cycle. In fact, during the force condition, the percent time of gait cycle duration of double limb support was reduced by 16% to 23% in the force epoch in 10 and 20N conditions. These double limb support time reductions remained ~8% below pre-force levels during the post-force epoch. Further analysis of inter-stride variability showed that double limb support time variability significantly decreased, as gait speed increased during force and post-force epochs. It would also be of interest if such changes translate into greater dynamic stability with the possibly of attenuating medial-lateral sway in force and post-force conditions. While investigating postural outcomes was beyond the proof-of-concept approach to this study, it would be interesting if similar findings can be observed in an elder chronic stroke and age-matched population.

The measures chosen were intended to test the hypotheses offered in this proof-of-principle study. As such, the reported outcomes were limited to spatiotemporal gait measures a priori. However, in light of the encouraging evidence found in gait velocity, stride length and double limb support times, future study can focus on a more comprehensive gait analysis including kinematic and postural measures, such as joint angle changes and center of mass displacement in elder post-stroke and healthy controls. Replicating spatiotemporal findings in addition to outcomes addressing postural control and coordination in a neurological population would also provide valuable information regarding quality and efficiency of gait changes brought on by the haptic force. Ultimately, such an analysis may help discern whether such changes in gait promote dynamic stability for functional walking in post-stroke, or other neurological conditions.

A decision was made to use 10 and 20N force conditions in light of the eventual implementation in a post-stroke population. Lighter tensile forces may be sufficient enough to alter gait in a way to promote dynamic stability rather than hinder it. Nevertheless, it might be a limitation to use a constant tensile force rather than calibrating forces based on the participants' body mass. Indeed, the mass was quite variable ranging from 53 to 89 kg across participants. Lastly, further study can also compare the use of visual stimuli, with particular focus on how increasing the speed of the virtual scene may affect a post-adaptation effect.

3.6 Conclusion

This study provided evidence to suggest that the use of haptic tensile forces applied to the hand during steady-state walking elicit adaptation and post-adaptation changes to spatiotemporal components of gait in healthy young adults. The changes were evident in all measurement outcomes - instantaneous gait velocity, stride length and double limb support time. Future studies are warranted to provide further evidence of kinematic and postural outcomes consistent with adaptations and post-adaptations in elderly and chronic stroke individuals. The long-term potential for using haptic tensile forces includes devising a training protocol that would evoke abiding locomotor changes, particularly in the post-adaptation phase.

AUTHORS’ CONTRIBUTIONS

GS contributed to participant recruitment, data analysis, interpretation of results, and manuscript production. JF and PSA participated in the study design, data analysis, and manuscript review. All authors read and approved the final manuscript.

ACKNOWLEDGMENTS

This study was from FRQS (Canada), CIHR (Canada), CFI (Canada) and the Richard and Edith Strauss Foundation. We would like to thank Gevorg Chilingaryan for statistical support, Valeri Goussev for programming assistance, Christian Beaudoin for VR support and Igor Sorokin for technical support.

CHAPTER 4: MANUSCRIPT 2

4.1 Preface

Previously, a pilot study involved healthy young subjects who walked on a self-paced treadmill in a virtual environment while forward-leading haptic force were applied to the hand (Sorrento et al., 2013; Sorrento et al., 2018). This study was designed to simulate the effect of walking a dog with a leash as gait measures were recorded. Evidence of spatiotemporal changes in gait emerged in response to the haptic force. The most notable change was an increase in gait velocity, deriving from increased stride length and decreased double limb support times (Sorrento et al., 2013; Sorrento et al., 2018). These findings follow up earlier reports linking somatosensory stimuli in the form of earth-fixed tactile surfaces such as haptic strips (Dickstein & Laufer, 2004; Fung & Perez, 2011; Perez & Fung, 2011) and walking aids such as the use of a cane (Perez & Fung, 2011) or rehabilitation dog (Rondeau et al., 2010) with improvements in dynamic stability.

Two main phenomena emerged from the study. First, tensile forces delivered by the haptic leash caused gait changes relative to baseline walking, thus revealing adaptation effects to the gait motor pattern. Second, and perhaps more importantly, was the presence of a post-adaptation effect as spatiotemporal parameters remained changed relative to baseline levels – even after the force was removed. This would suggest that transient modifications in the motor pattern driving gait were achieved. This also excluded the possibility that the initial adaptation effects were purely based on mechanical pull on the body. The adaptation and post-adaptation effects in response to forward-tensile forces reported were found in healthy young adults. However, it remains unknown whether the same effects can be seen in neurological populations with mobility deficits, such as an elder chronic stroke population. Thus, an important next step in the evaluation of haptic tensile forces is to measure their effects on chronic stroke subjects.

The effects of robot-controlled haptic tensile forces on gait and posture in older chronic stroke and older adults

Gianluca U. Sorrento, Philippe S. Archambault, Joyce Fung

School of Physical and Occupational Therapy, McGill University, Montréal, Canada

Feil/Oberfeld/CRIR Research Centre, Jewish Rehabilitation Hospital site of CISSS-Laval, Laval, Canada

Manuscript in preparation for submission to Gait & Posture in winter 2018

4.2 Abstract

A previous pilot study demonstrated young healthy adults changed their spatiotemporal gait parameters during and after walking with a haptic tensile leash. Given the potential clinical implications, we wanted to investigate the same haptic strategy in the older post-stroke population. Both older post-stroke (n=14) and healthy age-matched adults (n=14) where recruited and walked with haptic tensile forces in the direction of locomotion in a virtual environment to replicate the effects of walking a dog in an urban setting. The adaptation effects from the force epoch, when either a 10 or 15N force was applied to the hand via the leash, and post-adaptation effects during the post-force epoch, were both compared to baseline walking (pre-force epoch). For both groups, walking velocity increased and remained above baseline during both the force and post-force epochs. These velocity changes were found to be a product of a significant increase in stride length and significant decreases in stride and double limb support times. While there was little evidence of step width change, postural analysis revealed subjects shifted their body's center of mass (COM) towards the paretic (stroke) and non-dominant (control) side in the mediolateral plane during force and post-force

epochs, relative to baseline. This suggests that increases in gait velocity with the use of a haptic forces may lead to increased use of the paretic lower limb.

4.3 Background

The prevalence rate of stroke in Canada is approximately 1.15% (Krueger et al., 2015). Only 25% of stroke survivors will return to a community-dwelling functional level comparable to the non-stroke population (Dobkin, 2004; Duncan et al., 2005) owing to declining postural stability during gait. The effects of aging alone can require more attentional resources to meet the challenge of dynamic stability (Hollman et al., 2006; Lajoie et al., 1996). Investigating both aging and stroke, in tandem, often reveals reduced gait speeds resulting from both spatial (Bongers et al., 2015; Callisaya et al., 2010) and temporal gait parameters (Callisaya et al., 2010), particularly when lower extremity muscle strength is reduced (R.W. Bohannon & Andrews, 1998). This potential for injury facing the chronic stroke elders, such as falling rates of up to 50%, are some of the highest rates in any population (Jorgensen et al., 2002). These adverse outcomes contribute to increased morbidity and decreased quality of life (F. A. Batchelor et al., 2012; Divani et al., 2009).

Researchers suggest that an important strategy for improving postural stability is the use of somatosensory stimuli. In fact, elder individuals tend to use and react to somatosensory stimuli with different strategies when compared to younger counterparts (Qiu et al., 2012), as these stimuli have been shown to improve both static (J. J. Jeka & Lackner, 1994) and dynamic stability (Dickstein & Laufer, 2004). For example, the effect of light touch of a fingertip on an earth-fixed surface has been shown to attenuate sway while walking on a treadmill both for healthy elders (Dickstein & Laufer, 2004; J. J. Jeka & Lackner, 1994; Lackner et al., 2001) and chronic stroke individuals (Boonsinsukh et al., 2011; Fung & Perez, 2011), while also walking on challenging terrain (Oates et al., 2008). Other studies in stroke have employed specially trained rehabilitation dogs to improve walking outcomes such as improved kinematics, spatiotemporal outcomes and (Rondeau et al., 2010), and walking dynamics (Oates et al., 2008).

Chronic stroke individuals can regain motor recovery, particularly when training is task-specific and intense in nature (Langhorne et al., 2009). Yet, an important question remains - are stroke individuals capable of reshaping motor patterns related to gait and posture, just like the healthy population, as evidenced by adaptation and post-adaptation effects (Bronstein et al., 2009). Some evidence suggests that stroke individuals walking on a split-belt treadmill with belts at varying speeds as to constrain the paretic leg speed to match the speed non-paretic side (Reisman et al., 2005). These adaptations can then be carried over to over ground walking (Reisman et al., 2009a) and even during more functional-based locomotion (Gelsy Torres-Oviedo & Bastian, 2012; G. Torres-Oviedo et al., 2011).

In light of the above findings and the proof of concept presented in study 1, this study aims to measure adaptation and post-adaptation effects in the chronic stroke population from the 'virtual leash' haptic input. Further, the prospect of integrating robotics with virtual reality for gait rehabilitation in post-stroke has shown promise when compared to robotic training alone (Mirelman et al., 2010). The two main objectives can be defined as follows: 1) to determine whether chronic stroke individuals when walking with a haptic tensile force, compared to no force baseline walking, change both spatiotemporal and postural outcomes; and 2) to determine if the chronic stroke individuals are able to retain spatiotemporal and postural gait changes after the force is removed, compared to baseline walking.

4.4 Methods

4.4.1 Participants

Fourteen post-stroke survivors (mean age: 70.6 +/- 3.0 y.o.) and 14 age-matched controls (mean age: 71.8 +/- 2.7 y.o.) between the ages of 65 and 80 years old were recruited for this study. Chronic stroke subjects were all post 6 months and were ambulatory with or without assistance for a minimum of 3 minutes with controlled balance and posture. Other inclusion criteria were included Chedoke-McMaster Stroke Assessment (Gowland et al., 1993) scores between 3 and 6 for the leg and foot; and a Berg Balance Scale (BBS) above 34. Additionally, subjects required a score above 23 on the Falls Efficacy Scale (FES) (Tinetti et al., 1990), as evidence supports fear of falling as a considerable factor in spatiotemporal outcomes (A. G. Andersson et al., 2008; Chamberlin et al., 2005). Any individual with a history of orthopedic, cardiovascular or neurological disorder (other than stroke), or with walking discomfort that would obscure the analysis of post-stroke gait, was excluded. Subjects were invited for a single walking habituation session on the treadmill prior to the experiment. All subjects were recruited and provided informed written consent according to the specifications of the ethics review board of the Center for interdisciplinary research in rehabilitation of the greater Montreal (CRIR).

4.4.2 Apparatus

Subjects walked on a self-paced treadmill wearing a safety harness. An electro-potentiometer tethered to the back of the subject's harness enabled a self-selected walking pace on the treadmill. Using a derivative-based algorithm, a micro-controller then accounted for treadmill velocity based on the forward displacement of the individual on the treadmill relative to initial position (See figure 1a). The self-paced treadmill's velocity was used to synchronize a virtual environment projected onto a large screen, approximately 1.5 meters from the subject. This virtual scene consisted of a city sidewalk and a dog avatar (See figure 1). The scene was back projected in 2D by a projector located 2.5 meters behind the screen and was constantly updated relative to the individual's gait speed. The virtual scene was powered by CAREN-3 software (Motek, Netherlands) that ran on a separated CPU from the one that operated the treadmill. Subjects held the leash comfortably with the arm extended (anatomical position) level to the hip. The leash vector from the hand was projected in line with the pulley, thus minimizing any mediolateral force components. The vector also pointed slightly downward with respect to the pulley, making it almost parallel to the ground (see figure 4.1). Stroke subjects held the leash in their non-affected hand, while controls held it in their dominant hand. The leash was powered by a robotic arm with three degrees of freedom (HapticMaster, Moog BV, Netherlands). A steel reinforced stretch-resistant cord was attached to the robot and was feed through a series of pulleys to reach the subject's hand in the form of a leash (See fig. 1). On cue, the robot could instantly apply a force to the hand via an in-home program.

4.4.3 Constant Force Paradigm

The paradigm used to address the objectives of the study was the constant force paradigm (Figure 4.1C). After a verbal "start" cue, the subject proceeded to walk at a baseline comfortable self-pace for thirty seconds (pre-force epoch). A tensile force of either a 10 or 15N was then applied to the leash by the robotic device for 60 seconds (force epoch). Finally, the force was then removed, and the subject continued to walk for a up to a final 60 seconds (post-force epoch). A pause was taken between trials to avoid fatigue. The 10 and 15N force magnitudes were determined a priori given the results of study 1 and were intended to simulate the tension one would experience with real tension on a dog leash.

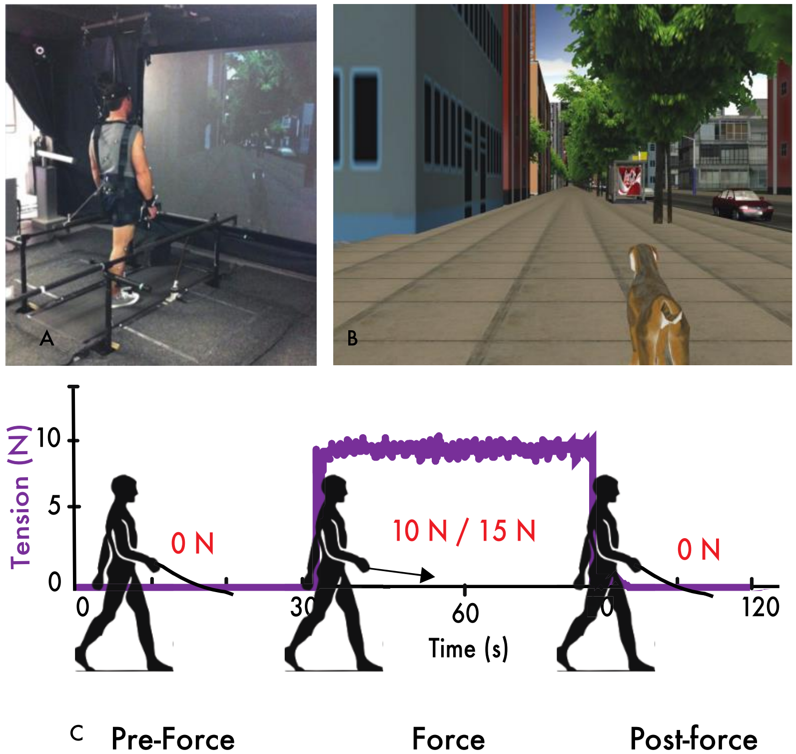


Fig. 4.1: Subject walking on self-paced treadmill synchronized to rear-projected screen.

A: Subject walking on self-paced treadmill synchronized to rear-projected screen, HapticMaster robot. B. Virtual scene projected on large screen consisting of an urban setting and dog avatar C: Schematic of the constant force paradigm consisting of pre-force, force and post-force epochs. The purple trace represents the either 10 or 15N leash tension delivered to the hand.

4.4.4 Data recording and analysis

Six Vicon motion capture cameras recording at 120Hz recorded positions of reflective markers placed on body landmarks (see index II). The cameras were pre-calibrated to record the area of the treadmill where subjects walked (see Figure 1A). An analogue pulse signal was used to synchronize the onsets of treadmill movement and robot force, which also served as time markers for the pre-force, force and post-force epoch analyses. Vicon data were processed offline in the Workstation (Vicon) gait analysis program.

A lab-developed Matlab (the Matworks, USA) routine calculated the study outcomes from the processed data. Spatiotemporal and postural outcomes were analyzed by epoch and per group. Gait velocity, stride time and stride length were calculated based on the displacement of foot markers over time during a gait cycle. Stride length and time were analyzed bilaterally. A stride was defined as all gait events between heels strikes of the same foot. Double limb support time encompassed the double limb phases before and after the swing phase of the same leg. Step width was calculated according to the lateral distance between the midpoint of each foot. Finally, mediolateral center of mass (COM) displacement was calculated by taking the average change in position of COM during force and post-force epochs relative to pre-force.

Statistical analysis was done with SPSS statistical software. Listed in Table 1 are the descriptive and inferential results of each outcome. Descriptive results included means, standard deviation and 95% confidence intervals. Varying functional levels in the stroke group meant that the data was not normally distributed. Hence, a general estimating equation (GEE) was chosen as an appropriate model for inferential analysis. Comparisons of interest were within subject comparisons of epoch. Specifically, the force and post-force epochs compared to the pre-force epoch. Between subject comparisons of stroke and control groups were also made. A priori descriptive analysis indicated that 10 and 15N conditions were similar, and thus were not compared. Significance levels were set to $p<.05$.

4.5 Results

4.5.1 Gait Velocity, Stride Length and Stride Duration

The boxplots in Figure 2 A-B depict pre-force, force and post-force epoch velocity values for both the 10 and 15N conditions. For the 10N force (Fig. 4.1A), stroke subjects on average increased gait velocity from

0.45m/s during the pre-force epoch to 0.52 m/s during the force epoch, corresponding to a 0.07m/ change. Post-force gait velocity also remained above the pre-force baseline velocities and corresponded to a 0.04 m/s and 0.07 m/s increase, for the 10 and 15N force levels, respectively. In comparison, healthy controls also increased gait velocity from 1.00 m/s during the pre-force epoch to 1.13 m/s during the 10N force epoch by 0.13 m/s. When the force was then removed, average gait velocity was at least 0.12 m/s above baseline for both the 10 and 15N conditions.

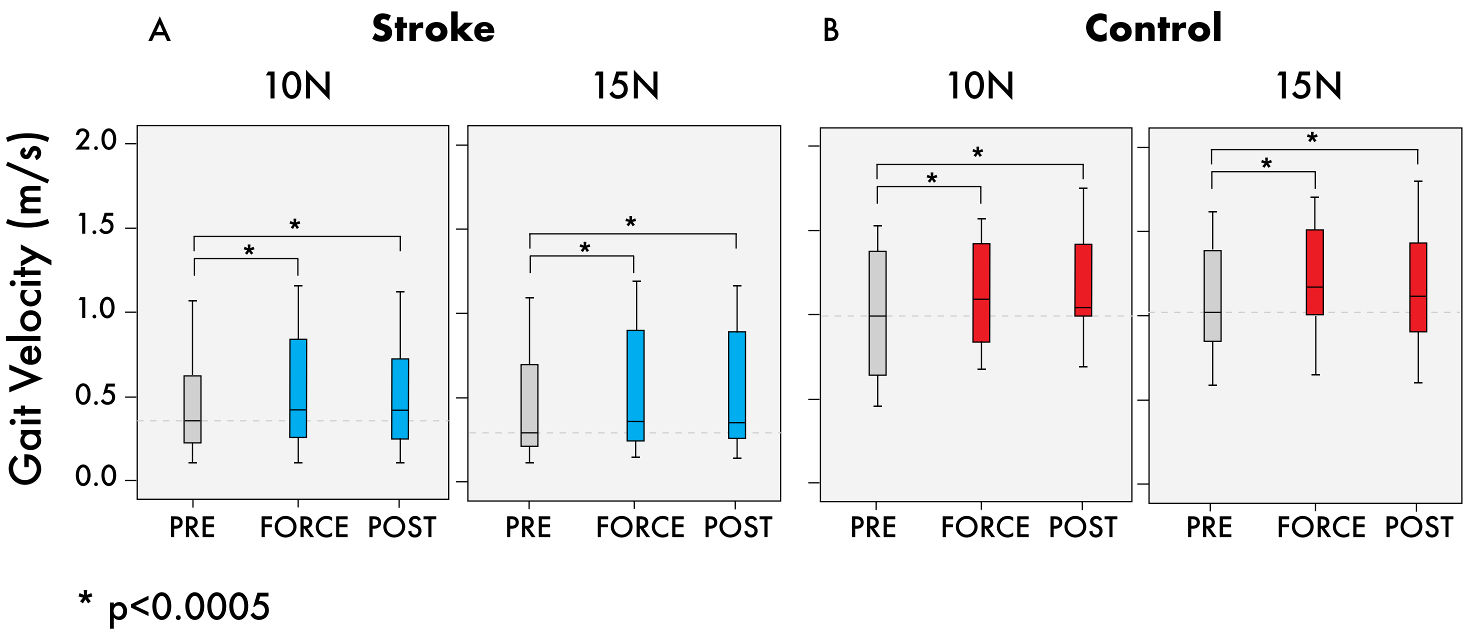


Fig. 4.2: Median and interquartile values for gait velocity in stroke and control groups.

Boxplot and whisker plots for gait velocity in stroke (A) and control (B) groups during pre-force, force and post-force epochs across 10 and 15N force conditions. The grey dotted line in each graph denotes median pre-force gait velocity levels. Both groups significantly increased gait velocity during force and post-force epochs relative to pre-force ($p<0.0005$). Significant changes in gait velocity were also measured between groups ($p<0.0005$).

Stride length also increased by 0.06 m and 0.05 m for the non-paretic (leash) and paretic (non-leash) limbs for the force epoch when compared to baseline. This increase was maintained at 0.05m above baseline during the post-force epoch across 10 and 15N trial conditions (See Table 4.1). The control group also increased stride length on average by 0.11m during force and post-force epochs. Stride duration, in contrast, tended to decrease during force and post-force epochs for both stroke and control groups. Bilateral stride time reductions for the stroke group were 0.08s for the force and 0.06s for the post-force epochs, while controls reduced stride time by 0.04s for both the force and post-force epochs (see Table 1).

General estimate equation (GEE) parametric testing confirmed significant changes for gait velocity (Wald X^2 = 74.6, $p<0.0005$) epoch, as post-hoc comparisons found significant interactions for both force and post-force epochs relative to the pre-force ($p<0.0005$). Significant changes were also seen for stroke and control groups (Wald X^2 = 74.6, $p<0.0005$). Stride length epochs were also significantly different (Wald X^2 = 74.6 $p<0.0005$), as post-hoc investigation showed significance for both force ($p<0.0005$) and post-force ($p<0.001$) epochs. An effect was also found for non-paretic and paretic sides (Wald X^2 = 6.0 $p<0.05$). A significant effect was also seen for stroke and control groups (Wald X^2 = 28.2 $p<0.0005$). Step time epochs were also significantly changed (Wald X^2 = 42.9 $p<0.0005$) with significant force ($p<0.0005$) and post-force ($p<0.0005$) interactions. There was no significant effect between non-paretic and paretic sides, however there was between groups (Wald X^2 = 16.4 $p<0.0005$).

4.5.2 Double limb support time and stride width

The boxplots in figure 4.3A-B reveal notable reductions in double limb support times in the force and post-force epochs relative to pre-force for the stroke and control subjects. Stroke subjects on average reduced double limb support times by 0.08 s and 0.06 s relative to baseline for the force and post-force epochs, respectively. The control group also tended to reduce double limb support times by 0.04 s and 0.03 s relative to baseline for force epoch and post-force epoch. Step width on the other hand remained relatively constant across epochs for both conditions and groups (see table 4.1).

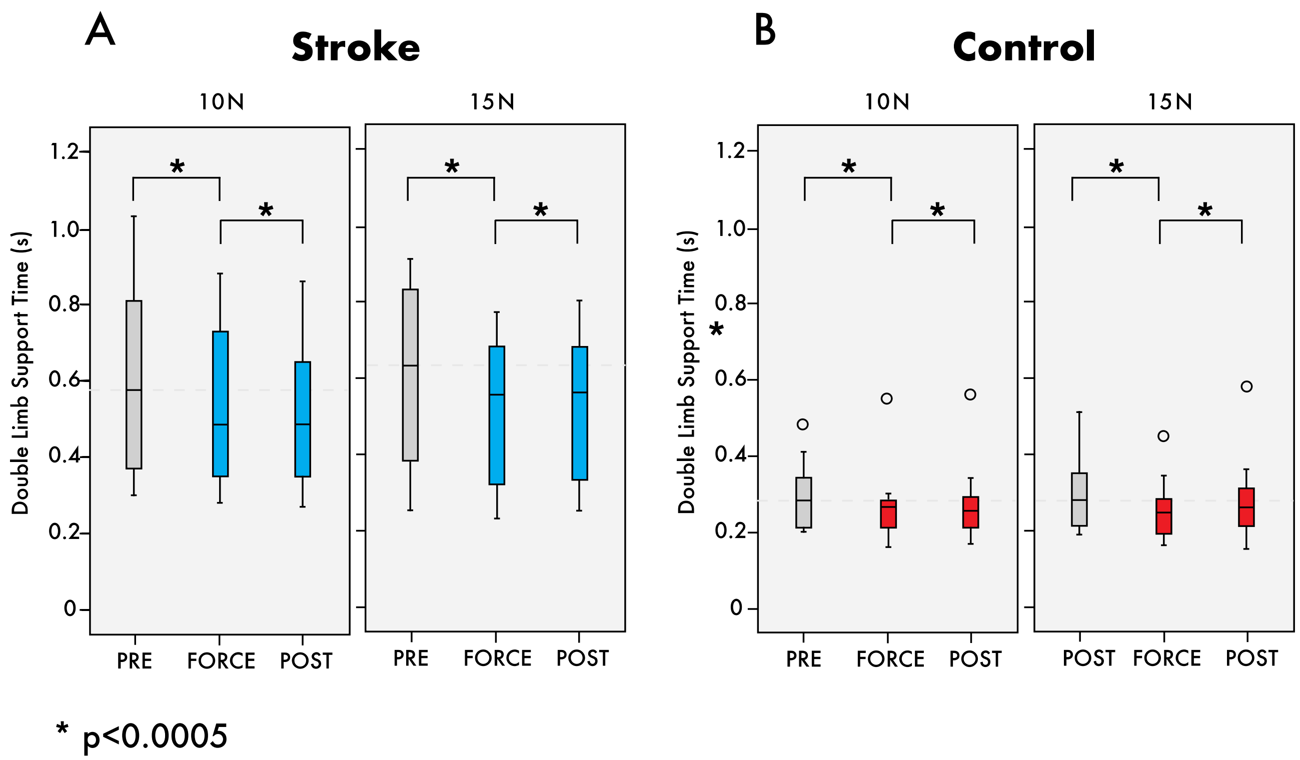


Fig. 4.3: Median and interquartile values for double limb support time.

Boxplot and whisker plots for double limb support time show (A) stroke and (B) control groups significantly decreased support times for force and post-force epochs relative to pre-force (p<0.0005). There was also a significant increase measured between groups (p<0005).

GEE analysis confirmed significant changes for double limb support time (Wald X^2 = 51.8 = p<0.0005) epoch. Post-hoc comparisons found significant interactions for both force and post-force epochs relative to the pre-force ($p<0.0005$). Significant changes were also seen for stroke and control groups (Wald $X^2 = 23.3$ $p<0.0005$). A similar analysis revealed no significant effects for step width.

4.5.3 Mediolateral Center of Mass displacement

Figure 4.4 illustrates a stroke and healthy control subject showing a distinct mediolateral shift of the COM gait cycle trajectory during force and post-force epoch from the non-paretic to paretic side for stroke (i.e. dominant side to the non-dominant side for controls) relative to pre-force.

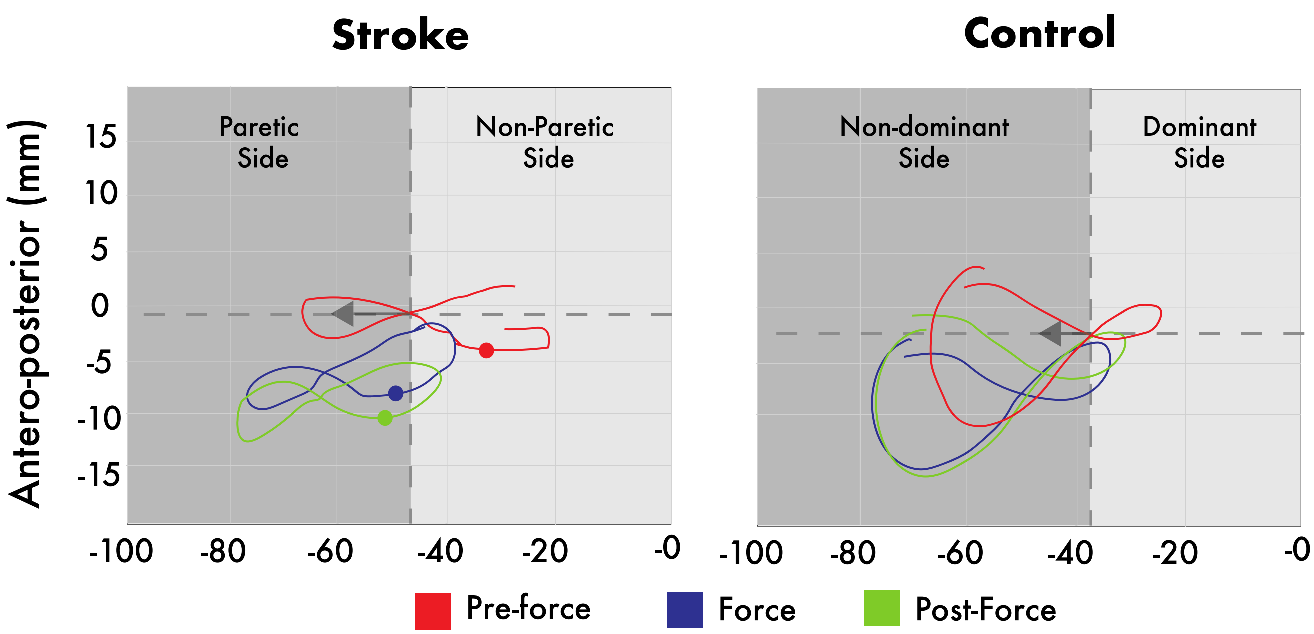


Fig. 4.4: Average mediolateral COM displacement for both a stroke and control subject across epochs. Average mediolateral CoM displacement during pre-force, force and post-force epoch for a stroke and control subject during a single trial. The grey arrows demarcate the magnitude of mediolateral COM shift during force and post-force epoch towards the paretic (stroke) and non-dominant (control) sides.

Figure 4.5 depicts boxplots for mediolateral COM displacement during force and post-force epochs, relative to pre-force both 10 and 15N force conditions for both stroke and control subjects. Across subjects, there was a tendency for the mediolateral CoM to displace to the paretic (or non-dominant) side during force and post-force epochs relative to pre-force by as much as 0.01m for both stroke and control groups. However, the GEE analysis revealed that no significant changes were seen between epochs or groups (p>0.05).

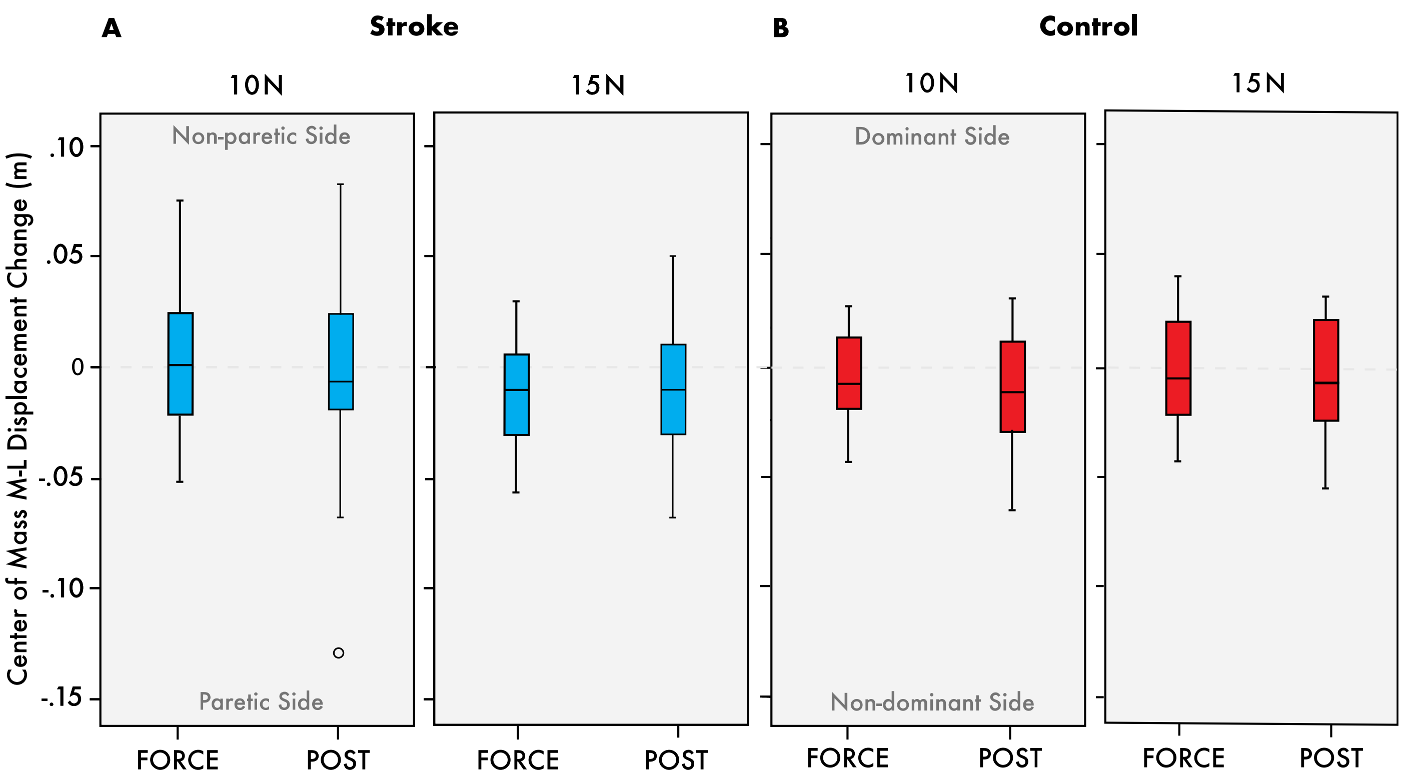


Fig. 4.5: Median and interquartile values for mediolateral COM displacement.

Boxplot and whisker plots for mediolateral CoM displacement show both (A) stroke and (B) control groups have a tendency to shift mediolateral CoM to the paretic or non-dominant side. There was no within group effect for epoch (p>0.05).

TABLE 4.1							
	EPOCH			95% CONFIDENCE INTERVAL			
STROKE N=14	PRE	FORCE	POST	FORCE - PRE		POST - PRE	
10N	x̄ / σ	x̄ / σ	x̄ / σ	Lower bound	Higher bound	Lower bound	Higher bound
Gait Velocity (m/s)	0.45 / 0.30	0.52 / 0.34	0.49 / 0.32	35.20	105.55	23.08	61.03
Stride Length (m)	0.58 / 0.31	0.63 / 0.33	0.60 / 0.32	21.85	81.67	2.25	49.71
Stride Duration (s)	1.39 / 0.27	1.31 / 0.21	1.33 / 0.22	-0.15	-0.02	-0.12	-0.01
Double Limb Support time (s)	0.59 / 0.22	0.51 / 0.19	0.54 / 0.21	-0.13	-0.02	-0.07	-0.01
Step width (m)	0.13 / 0.03	0.12 / 0.03	0.13 / 0.03	-8.96	3.79	-10.57	13.05
CoM ML displacement (m)	0.01 / 0.15	0.01 / 0.15	0.01 / 0.16	-9.47	24.47	-22.61	34.21
15N							
Gait Velocity (m/s)	0.45 / 0.33	0.53 / 0.37	0.51 / 0.35	49.13	113.75	29.30	103.31
Stride Length (m)	0.55 / 0.32	0.62 / 0.33	0.61 / 0.32	36.44	100.50	22.24	95.71
Stride Duration (s)	1.40 / 0.25	1.31 / 0.23	1.33 / 0.23	-0.12	-0.06	-0.10	-0.04
Double Limb Support time (s)	0.61 / 0.24	0.51 / 0.19	0.53 / 0.19	-0.14	-0.06	-0.12	-0.04
Step width (m)	0.13 / 0.03	0.13 / 0.03	0.13 / 0.03	-7.45	0.45	-8.67	3.59
CoM ML displacement (m)	0.00 / 0.14	0.02 / 0.15	0.01 / 0.16	-11.51	17.65	-18.22	19.12
HEALTHY N=14							
10N							
Gait Velocity (m/s)	1.00 / 0.34	1.13 / 0.3	1.14 / 0.30	79.51	180.68	50.68	227.93
Stride Length (m)	1.05 / 0.28	1.15 / 0.22	1.15 / 0.19	55.21	152.39	32.45	182.80
Stride Duration (s)	1.08 / 0.16	1.04 / 0.17	1.04 / 0.17	-0.05	-0.03	-0.06	-0.02
Double Limb Support time (s)	0.30 / 0.10	0.26 / 0.08	0.27 / 0.09	-0.02	-0.01	-0.03	-0.01
Step width (m)	0.11 / 0.03	0.10 / 0.03	0.11 / 0.03	-12.78	3.82	-8.76	4.31
CoM ML displacement (m)	0.00 / 0.04	0.01 / 0.04	0.01 / 0.04	-5.72	16.13	-11.42	19.58
15N							
Gait Velocity (m/s)	1.07 / 0.34	1.19 / 0.34	1.15 / 0.35	85.60	150.47	24.79	125.69
Stride Length (m)	1.11 / 0.25	1.19 / 0.23	1.15 / 0.22	53.44	104.64	13.07	72.10
Stride Duration (s)	1.07 / 0.16	1.03 / 0.16	1.04 / 0.17	-0.05	-0.03	-0.05	-0.01
Double Limb Support time (s)	0.29 / 0.09	0.25 / 0.08	0.27 / 0.09	-0.04	-0.02	-0.03	-0.01
Step width (m)	0.11 / 0.03	0.10 / 0.03	0.11 / 0.03	-3.87	3.07	-0.58	8.99
CoM ML displacement (m)	0.02 / 0.04	0.02 / 0.04	0.02 / 0.03	-13.58	12.74	-15.21	12.28

Table 4.1: Average, standard deviation and 95% confidence intervals for each outcome in 10 and 15N condition for stroke and control groups.

TABLE 4.2		
GEE MODEL		
Outcome	**Wald Chi -Square value (df)**	**P Value**
Gait Velocity (m/s)	**Group** = 27.5 (1) **Epoch** = 74.6 (2)	**Group** p<0.0005 **Force & Post-Force** p<0.0005
Stride Length (m)	**Group** = 28.2 (1) **Epoch** = 48.5 (2) **Side** = 6.0 (1)	**Group** p<0.0005 **Force & Post-Force** p<0.005 **Side** p<0.014
Stride Duration (s)	**Group** = 16.4 (1) **Epoch** = 42.9 (2)	**Group** p<0.0005 **Force & Post-Force** p<0.0005
Double Limb Support Time (s)	**Group** = 23.3 (1) **Epoch** = 51.8 (2)	**Group** p<0.0005 **Force & Post-Force** p<0.0005
Step Width (m)	**No Effects**	p>0.5
CoM ML Displacement (m)	**No Effects**	p>0.5

Table 4.2: Results of GEE analysis for each outcome. Factors include epoch, side, group and force condition.

4.6 Discussion

This study built on pilot findings of changed gait under tensile force in healthy young adults. On that occasion, spatiotemporal changes in walking lead to adaptation and post-adaptation effects, both during and after force application. The results begged the question if a post-stroke population was also capable of changing their gait profile as a result of a tensile force, and if so, to what extent both spatiotemporally and posturally.

4.6.1 Spatiotemporal outcomes

The spatiotemporal trends among stroke and control subjects largely replicated those found in the pilot study. For example, gait velocity during the force epoch systematically increased relative to the baseline pre-force walking. Some changes reached above the 0.07 m/s mark during force and post-epochs for the stroke group, which approach the 0.1 m/s minimally important changes reported in the literature (Bohannon & Glenney, 2014). The control group displayed even greater changes (0.1 m/s) during the force and post-force epochs.

The ensuing question was which spatiotemporal parameters account for the gait speed increases. From a spatial perspective, stride length was a factor in the gait velocity increase. Stroke and control subjects were able to increase stride length by at least 0.06 m, which approximate some reported minimal detectable changes for post-stroke of 0.06 - 0.07 m (Kesar et al., 2011). This effect was seen both on the non-paretic and paretic sides, suggesting adaptation and post-adaptation effects from altering the original walking motor command was bilaterally present during the gait cycle. From a temporal perspective, both groups also tended to reduce bilateral stride times as the trials progressed to force and post-force epochs, showing that the temporal component was also changed with respect to increased gait velocity.

4.6.2 Gait stability

Measures addressing gait stability when walking with and without tension also showed encouraging results. For example, double limb support times were significantly reduced by as much as 0.08s for the force and post-force epochs for the stroke group and by 0.02 – 0.03 s in the control group. While step width did have a tendency to reduce ever slightly, there was no evidence of it being a significant factor in the gait changes found for either the stroke or healthy control groups.

4.6.3 Postural Control

There was a tendency for both stroke and controls to shift their COM to the paretic side during force and post-force epochs. Given the shift to the paretic side, it could be informative to understand whether the haptic force caused the individuals to change their kinematic strategy. The possibility of engaging the paretic side during increased gait velocity could approximate the kinematics of a healthy stride and achieve a safer functional gait. Evidence in the literature suggests that the main factors affecting gait velocity in mild to moderate chronic stroke patients are the hip, knee flexion as well as plantarflexion spasticity of the affected leg (Hsu et al., 2003). Hence, further analysis should be carried out regarding the coordination of limb segments and how it can affect the activation of the paretic limb given the presence of the mediolateral CoM shift. Such changes could underpin the kinematic mechanisms driving the adaptation and post-adaptation changes occurring in both spatiotemporal and postural outcomes.

4.6.4 Limitations

There were several limitations to the study. For example, though stroke subjects were ambulatory, there was a wide array of walking performance, accounting for a significant variability across walking performance within the stroke group in particular. Future work may look to further stratify the chronic stroke group according to functional level with criterion such as functional gait velocity clinical testing. Furthermore, the walking endurance and fatigue of chronic stroke patients was a factor in limiting the walking paradigm to ~2.5 minutes. Thus, both the force and post-force epochs were limited to one minute each. Lastly, the target sample size number was 14 per group. This calculation was determined a priori based on desired alpha and beta levels and effect size. This along with a statistical analysis accounting for a wider biological variability should also be considered when interpreting the results.

4.7 Conclusion

This study provided encouraging evidence that haptic tensile forces in the direction of walking can change the way chronic stroke and older individuals walk. This was demonstrated with spatiotemporal changes that contribute to increased gait velocity, as well as postural shifts in the mediolateral COM that promote greater usage of the paretic side. The potential clinical benefits of the adaptation and post-adaptation effects elicited should be evaluated further. Specifically, a follow-up study is to be designed to compare the adaptation and post-adaptation effects to walking with a cane. The significance of changed walking outcomes relative to a common walking aid could provide insight into the potential clinical impact this strategy may offer.

CHAPTER 5: MANUSCRIPT 3

5.1 Preface

So far, we have investigated the effects of haptic tensile forces in the healthy and elder stroke populations. It has been established that in healthy young individuals, such forces tend to change gait patterns while walking with the force in hand. With further investigation elder populations, both healthy and post-stroke, also showed similar effects. From a spatial temporal perspective, we can say that the haptic tensile force, whether 10 or 15N in magnitude, was responsible for the increase of gait velocity. These gait velocity changes, could either have been as a result of increased stride length, decreased stride time. It was shown that the velocity change was indeed a combination of reduced stride time and stride distance both. These changes were also clear when subject walked with the haptic leash and when it was removed, relative to baseline walking. Moreover, equally interesting findings were found in postural measures. For instance, we see that relative to pre-force baseline walking, the post-stroke and healthy age-matched controls tend to favor the paretic or non-dominate side as evidenced in a shift of COM. Specifically, the COM shifts to the paretic side in post-stroke and non-dominant side in healthy controls. In the following study, we aim to establish evidence suggesting that the leash modality can be just as effective, if not, more effective as a commonly used haptic mobility device - the cane. An instrumented version of a cane has been developed in the virtual environment (Perez & Fung 2011) and in comparing these modalities, the we can better understand the clinical potential of the haptic leash as employed in a training or rehabilitation protocol.

The Effects of a Robot-Controlled Haptic Leash Compared with an Instrumented Cane on Gait and Posture in Post-Stroke and Older Adults

Gianluca U. Sorrento, Philippe S. Archambault, Joyce Fung

School of Physical and Occupational Therapy

McGill University

Montréal, Canada

Feil/Oberfeld/CRIR Research Centre

Jewish Rehabilitation Hospital site of CISSS-Laval

Laval, Canada

gianluca.sorrento@mail.mcgill.ca; philippe.archambault@mcgill.ca; joyce.fung@mcgill.ca

Published as Conference Proceedings of the International Conference on Virtual Rehabilitation (ICVR) and available in ieeexplore.ieee.org (June 2017)

5.2 Abstract

Haptic tensile forces in the direction of locomotion were used in a virtual environment to investigate the adaptation and post-adaptation effects on steady-state walking in older post-stroke (n=14) and healthy age-matched adults (n=14). Increased walking velocity was observed in both post-stroke and control groups by as much as 0.11 m/s and 0.14 m/s, respectively, as subjects adapted to walking with a haptic force. Gait velocity was then maintained above baseline after force removal to an extent comparable to walking with a cane, suggesting a post-adaptation effect. These velocity changes were explained by both decreases in stride time and increases in stride length. Further investigation revealed that, with increased haptic forces, the

body’s center of mass in post-stroke participants tended to shift towards the paretic side in the mediolateral plane compared to walking with a cane and pre-force walking. This effect was most robust in lower functioning post-stroke as compared to higher functioning post-stroke and healthy subjects. Results suggest that increases in velocity with the use of a haptic force may be accompanied by increased usage of the paretic side. Further study will assess the coordination of bilateral limbs to further ascertain these findings.

Keywords—Virtual reality; rehabilitation robotics; haptics; elderly; stroke

5.3 Introduction

Improving the functional walking capacity of a post-stroke individual is often a major goal in rehabilitation (Baer & Smith, 2001; Craig LE, 2011; Eng & Tang, 2007). In fact, clinicians often explore possibilities of innovative techniques that potentially facilitate positive changes in mobility post-stroke. When assessing locomotion in chronic stroke survivors, what is commonly observed is some degree of paresis. For post-stroke individuals, this is evident in the compromised spatiotemporal and kinematic patterns during movement. Bilateral asymmetry is often a consequence of hemiparesis (Alexander LD, 2009). Such asymmetries are rooted in changes in muscle tone of the affected limb, making it difficult for the limb to control movements necessary for efficient gait and posture. This can often lead to functionally slower and less stable locomotion (P. Andersson & Franzen, 2015). Ultimately, there is an increased chance of falling and further injury. This poses a challenge to dynamic stability to the point of compromising the functional autonomy of the post-stroke individual (Nyberg & Gustafson, 1995b). Moreover, there is evidence that these spatiotemporal and postural outcomes help predict falling in post-stroke (Nyberg & Gustafson, 1997) and older adults (Palmer, 2001; Tinetti, 2003) and are essential in maintaining dynamic stability that is both proactive in adapting to the environment and reactive in response to perturbations and obstacles (Lamontagne & Fung, 2004).

Clinicians and researchers are taking to this challenge by investigating potential strategies that can promote improvements in dynamic postural equilibrium. In the process, they also aim to better understand the motor control properties involved in such changes. These changes, triggered by an intervention, may lead to long-term adaptations (Kahn & Hornby, 2009). Clinicians and researchers alike are interested in the extent to which spatiotemporal constraints or haptic cues brought on by an intervention can cause these adaptations (G. Torres-Oviedo et al., 2011). Perhaps more importantly, as far as functional recovery in stroke is concerned, is the extent to which such adaptations in locomotion are retained even when the exposure is no longer present. This particular retention can be thought of as a post-adaptation effect. The idea of adaptation and post-adaptation described as the ‘broken elevator effect’ (Bronstein et al., 2009; Patel et al., 2014; Reynolds & Bronstein, 2003) has been investigated in various locomotor adaptation studies involving the split-belt treadmill (Reisman et al., 2007). Essentially, the split-belt treadmill features two separate belts that can move either in tandem, or at different speeds in order to constrain locomotion. The result that researchers have reported is a reduction in spatiotemporal discrepancies between the paretic leg and non-paretic leg. Other studies (Choi & Bastian, 2007; Reisman et al., 2009b) show that stroke participants are able to adapt their walking with increasing gait symmetry between the paretic and non-paretic legs (Reisman et al., 2007) and that post-adaptation effects may be transferrable to overground walking (Reisman et al., 2009b)

Less understood, however, is how or to what extent does this sort of training bring about a longer lasting post-adaptation effect, particularly in neurological populations. Suggestions as to how this potential could be maximized may come from comparing locomotion between chronic stroke and age-matched controls, as it is believed that stroke individuals may exhibit different motor learning behavior (Bastian, 2008). For example, one study using the split-belt treadmill reported that when such walking constraints are administered to healthy individuals, an adaptation during the exposure is brought on quicker than for the stroke patient. However the stroke patient shows longer retention in the post-adaption phase. From a motor control perspective, this may prove important when considering how adaptations could be used to reinforce motor patterns, and maximize the potential benefits from a post-adaptation phase (Kahn & Hornby, 2009).

Using various platform perturbations to induce spatiotemporal constraints on gait is not the only strategy for inducing changes in walking performance. Researchers have also investigated the role of haptic cues for training dynamic stability. For example, there is evidence suggesting that light finger contact with an earth-fixed object during locomotion can have an attenuating effect on sway, thereby reducing postural deviation (Dickstein & Laufer, 2004). Researchers also report similar effects in terms of greater postural stability and muscle activation in the paretic limb when light finger contact was used in a virtual reality (VR) setting. We have previously shown that VR can be effectively used for locomotor training by simulating different cognitive and environmental demands on the individual adaptations (Fung et al., 2006). An instrumented cane was added to the VR setup with ecological validity, as evidenced by the ability of both post-stroke and healthy subject subjects in adapting their gait velocity similar to overground walking (Perez & Fung, 2011). Another strategy for providing helpful haptic cues to chronic stroke individuals is the use of a specially trained rehabilitation dog. In one pilot study, chronic stroke survivors increased gait velocity and decreased gait variability when being guided by the trained rehabilitation dog during locomotion (Rondeau et al., 2010).

Drawing from what has been measured and observed in these previous studies, a novel approach was designed combining haptic cues in the form of a 'virtual leash' and virtual reality in order to investigate the possibility of adaptations and post-adaptations in gait. A proof-of-principle study showed that healthy individuals are capable of changing their gait profile when exposed to these haptic stimuli (Sorrento et al., 2018). However, it remains unclear whether post-stroke individuals, who have recovered some walking ability despite the persistence of sensorimotor impairments, can adapt their gait patterns in a similar manner as healthy participants. The aim of this study was to assess the extent of gait and postural changes when post-stroke individuals walked with and without haptic tensile forces induced by a robot controlling a hand-held leash, in a virtual environment with a dog avatar, as compared to older adults. Once evidence of changes are established, the secondary aim was to compare the gait outcomes obtained with the haptic leash in relation to a commonly used walking aid, such as a cane.

5.3 Methods

5.3.1 Subjects

Fourteen chronic older post-stroke survivors (65-80 years old) and 14 healthy age-matched control subjects were recruited upon written informed consent approved by the Centre interdisciplinaire de recherche du Montréal (CRIR) ethics committee. Post-stroke participants were stratified based on their overground 10m gait speed into 2 sub-groups, higher (0.8 m/s or above) or lower functioning (<0.8 m/s), respectively (see Table 1). The threshold of 0.8 m/s is generally accepted as a marker of community ambulation (Schmid et al., 2007). All stroke subjects were at least 6 months' post-stroke at the time of participation and had attained a motor score of 3-6 out of 7 for the leg and foot components of the Chedoke-McMaster impairment inventory (Alexander LD, 2009; R. W. Bohannon, 2009), as well as a Berg Balance Scale score above 35 out of 56 to be considered for the study. Also, stroke subjects had to be free of any pre-existing neurological, skeletomuscular or cardiovascular conditions that could have adverse or otherwise shrouding effects on the study of gait post-stroke. Healthy age-matched control subjects were also free of pre-existing conditions. Subjects were also required to walk independently for a minimum of three minutes at a comfortable pace.

Subjects LS1-LS7	**Lower functioning post-stroke**						
	Gender	***Age***	***Leash/Cane Hand (in VR)***	***Walking Aid***	***Orthosis***	***Chronicity***	***Overground Walking Speed (m/s)***
LS1	**M**	**71**	**Left**	**Cane**	**Yes**	**>1 year**	**0.60**
LS2	**M**	**73**	**Left**	**Cane**	**No**	**>0.5 years**	**0.58**
LS3	**M**	**67**	**Left**	**Cane**	**No**	**>0.5 years**	**0.66**
LS4	**F**	**72**	**Left**	**Cane**	**No**	**0.5 years**	**0.23**
LS5	**F**	**75**	**Left**	**No**	**No**	**>1 year**	**0.66**
LS6	**M**	**67**	**Left**	**Cane**	**No**	**>5 years**	**0.75**
LS7	**M**	**68**	**Right**	**Cane**	**Yes**	**>0.5 years**	**0.37**
MEAN (STDEV)	**5M/2F**	**70.4 (3.2)**					**0.55 (0.17)**
Subjects HS1-HS7	**Higher functioning post-stroke**						
	Gender	***Age***	***Leash/Cane Hand (in VR)***	***Walking Aid***	***Orthosis***	***Chronicity***	***Overground Walking Speed (m/s)***
HS1	**F**	**73**	**Right**	**No**	**No**	**>5 years**	**0.81**
HS2	**F**	**70**	**Left**	**No**	**No**	**0.5 years**	**1.30**
HS3	**M**	**67**	**Right**	**No**	**No**	**>1 year**	**0.79**
HS4	**M**	**67**	**Right**	**No**	**No**	**>1 year**	**1.40**
HS5	**M**	**73**	**Right**	**No**	**No**	**>1 year**	**1.20**
HS6	**M**	**72**	**Right**	**No**	**No**	**>1 year**	**0.92**
HS7	**M**	**74**	**Right**	**Walker/ Cane**	**No**	**0.5 years**	**0.83**
MEAN (STDEV)	**5M/2F**	**70.9 (2.9)**					**1.04 (0.26)**
Subjects H1-H14	**Healthy age-matched controls**						
	Gender	***Age***	***Leash/Cane Hand (in VR)***	***Walking Aid***	***Orthosis***	***Chronicity***	***Overground Walking Speed (m/s)***
MEAN (STDEV)	**7M/7F**	**71.8 (2.7)**	**5L/9R**				**1.33 (0.23)**

Table 5.1: Demographics of subjects.

Post-stroke subjects were stratified into higher functioning, those who walked >0.8 m/s over ground (community dwelling pace), and lower functioning (<0.8 m/s). Subjects recruited based on Chedoke-McMaster leg and foot scores between 3-6 and Berg balance scores above 35.

5.3.2 Apparatus

Subjects were fitted into a harness and walked on a self-paced treadmill. The treadmill was PID-controlled with an algorithm using the subject's real-time distance and velocity signals. The distance signal was monitored by an electro-potentiometer via an extensible cord tethered to the back of the subject's harness. The potentiometer could sense the cord's displacement, relative to its initial position, and instantaneously send the distance and velocity signals computed by a micro-controller to the virtual environment (VE), powered by the CAREN-3 system (Motek BV) system. The VE featured a street scene with a dog avatar that progressed at the same speed as the walking subject. The synchronized scene was rear-projected onto a 2.5m by 3m screen located 1.5 m in front of the subject's starting point on the treadmill (see Fig. 4.1), providing a sense of immersion within the VE. The starting point of the treadmill was delineated on the frame of the treadmill to ensure that the subject started each trial in the same location. The step-change in forward tensile force from 0N to 10 or 15N on the hand was provided via a leash and pulley system anchored to a force-controlled haptic robotic arm with 3 degrees of freedom (HapticMaster, Moog BV) (see Fig.4. 2 foreground). Custom-made software was used to control the step-wise changes of the robotic arm.

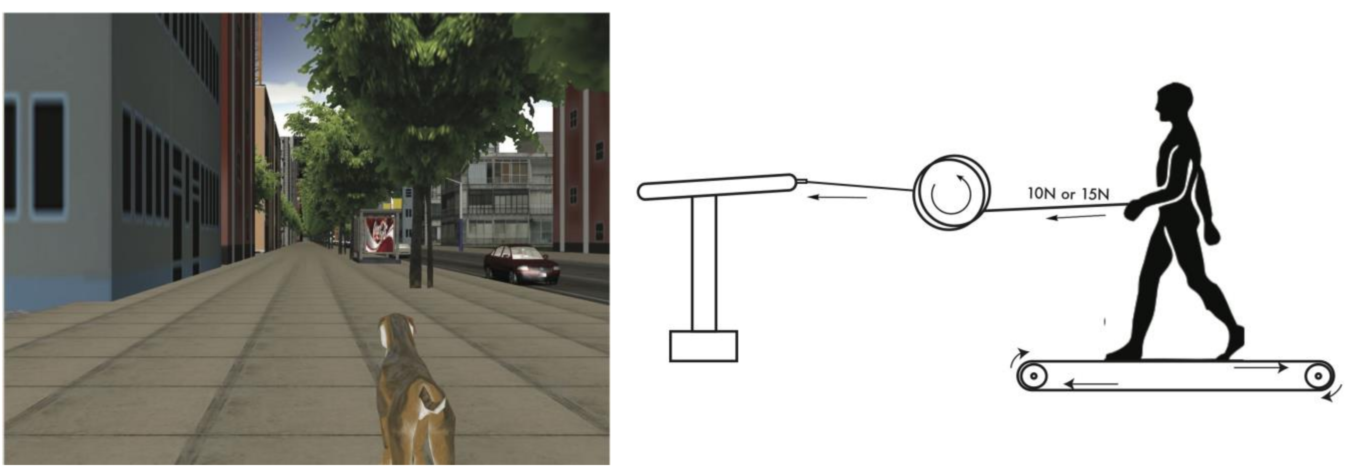


Fig. 5.1: Virtual reality scene used during the constant force paradigm.

Virtual reality scene used during the constant force paradigm and schematic of the Hapticmaster robotic arm, pulley system, leash and self-paced treadmill.

5.3.3 Virtual Leash and Cane Paradigms

Before data recording, participants were given the opportunity to habituate their walking ability on the self-paced treadmill until their perceived stability and walking speed were comparable to overground walking. In order to achieve this, some post-stroke subjects may have even required separate habituation sessions. During experimentation, subjects walked with a robot-controlled leash or an instrumented cane held by the non-paretic (post-stroke) or dominant (control) hand. Subjects began a leash trial by walking for 30 seconds with a slack leash in the pre-force epoch (see Fig. 5.2, right). This epoch was intended to provide a walking baseline to which outcomes obtained during subsequent epochs could be compared. After 30 seconds, the force epoch would begin when a forward pulling force, from 0N to either 10 or 15N, was rapidly applied to the hand. This force remained constant for the duration of one minute. Finally, the force was then rapidly released and subjects continued to walk with a slack leash for up to an additional minute in the post-force epoch (see Fig. 5.2). All subjects also walked with an instrumented cane fixed to the base of the treadmill during a separate cane walking trial. For comparison, a free walk trial where the subject walked completely independently was conducted and compared with the pre-force baseline walk.

5.3.4 Measurement and analysis

A total of 42 reflective markers were attached to body landmarks as suggested by the Vicon plug-in-gait© marker set-up (See Appendix B). An additional two markers were placed on the leash and three on the instrumented cane to track their respective positions. Body motions were captured at 120 Hz by 6 MX-Vicon cameras.

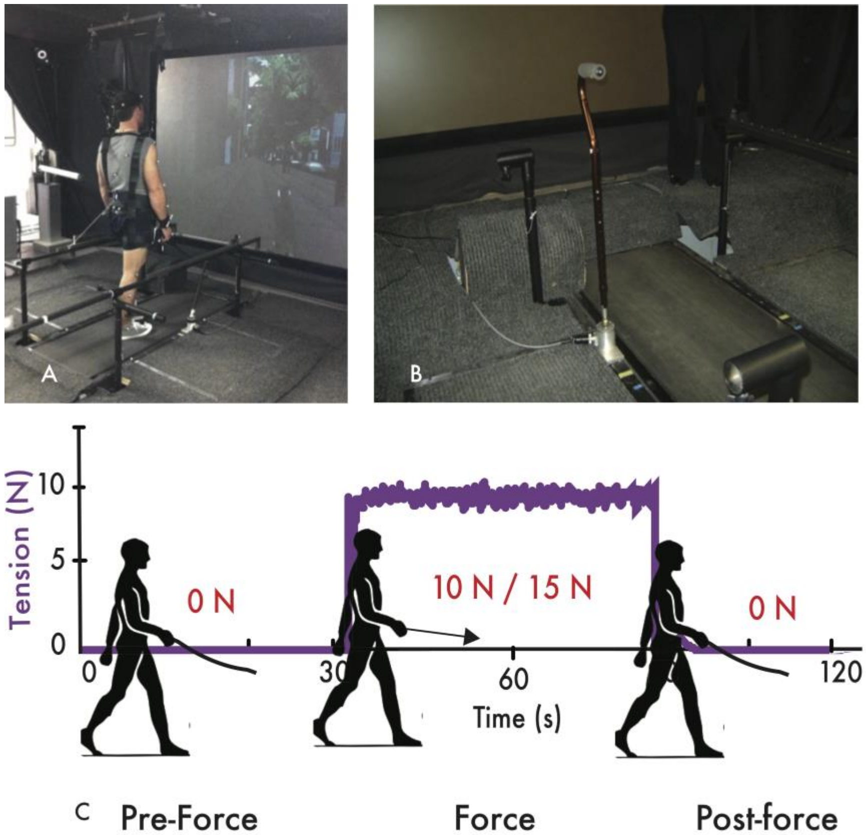


Fig. 5.2: Virtual environment, instrumented cane and schematic of the constant force paradigm.

A: Virtual environment, including rear-projected scene, robot (foreground) and treadmill. B: Instrumented cane mounted on the treadmill. C: Schematic of the constant force paradigm illustrating pre-force, force and post-force epochs. The purple trace represents the leash tension brought on by the robotic arm.

Acquired kinematic data were used to compute all outcomes processed offline using an in-house routine programmed in Matlab (MathWorks). Synchronization across Vicon and Hapticmaster controllers was necessary to define the critical time points of step-change force delineating pre-force, force, and post-force epochs. To address this, the beginning and ending of each walking trial was marked by an analog pulse from the CAREN system to both Vicon and the Hapticmaster platforms and was indexed in their respective data files for analysis. The instantaneous gait velocity was derived from the kinematic data by dividing the stride length with the stride duration for each gait cycle (see Fig. 5.3 A-B). In order to estimate the adaptation effects, the instantaneous gait velocity was compared between force and pre-force epochs. Post-adaptation effects were then estimated by comparing the post-force and pre-force epoch. Average differences in gait velocity across subjects with 95% confidence intervals were taken in reference to the paretic leg in post-stoke subjects and the corresponding non-dominant leg in the healthy controls. The same process was used to estimate changes in stride duration and stride length when walking with the instrumented cane were taken relative to pre-force conditions recorded during 10N and 15N leash trials. Data for all outcomes fell under a normal distribution, thus a mixed model repeated measures ANOVA was conducted taking into consideration group, epoch, and limb side levels.

Kinematic data recorded from all landmark reflective markers in Vicon plug-in gait formation (see Appendix B) and offline anthropometric records were required to calculate average mediolateral and vertical CoM displacement. Markers of the lower limb where also used to calculate average x, y and z angle orientations of the thigh, leg and foot coordination per stride cycle for pre-force, force and post-force epochs. Finally, to measure changes in angular momentum of the lower limb, an average vector product calculation was calculated for each gait cycle within the three epochs. This vector was obtained by multiplying the embedded thigh, leg and foot orientations, with their corresponding angular velocities.

5.4 Results

5.4.1 Gait Velocity

For the two groups of post-stroke subjects and healthy controls, all but one subject increased their gait velocity when walking with 10N of tension, compared to baseline walking. The same stroke subjects and all healthy controls increased gait velocity with 15N of tension. A mixed model repeated measures ANOVA confirmed this result for epoch $F_{(2,136)} = 4.4$, $p<0.014$) and group $F_{(2,327)} = 4.3$, $p<0.015$). Specifically, this corresponded to an average 0.05 m/s, 95% CI [0.03 - 0.08 m/s] increase, as compared to baseline, when walking with either 10 or 15N conditions in the lower functioning stroke subjects and 0.10 m/s, 95% CI [0.07 - 0.14 m/s] and 0.11 m/s, 95% CI [0.08m/s-0.14m/s] changes, respectively, in the higher functioning subjects (Fig. 5.4). Healthy controls increased by 0.14 m/s 95% CI [0.09-0.2m/s] and 0.11 m/s 95% CI [0.10-0.18 m/s] during the 10 and 15N conditions, respectively. When the force was then removed for the post-force epoch, subjects maintained gait velocity above pre-force pace. This corresponded to 0.04 m/s 95% CI [0.03 - 0.07m/s] and 0.06 m/s 95% CI [0.03 - 0.09m/s] in lower functioning stroke and 0.05 m/s 95% CI [0.03 - 0.07m/s] and 0.07 m/s 95% CI [0.02 - 0.12m/s] for higher functioning stroke, following 10N and 15N tension, respectively. Healthy controls maintained gait velocity above pre-force levels by 0.13 m/s 95% CI [0.06 - 0.21m/s] and 0.08 m/s 95%CI [0.05 - 0.12m/s] for the 10N and 15N conditions in the post-force epoch.

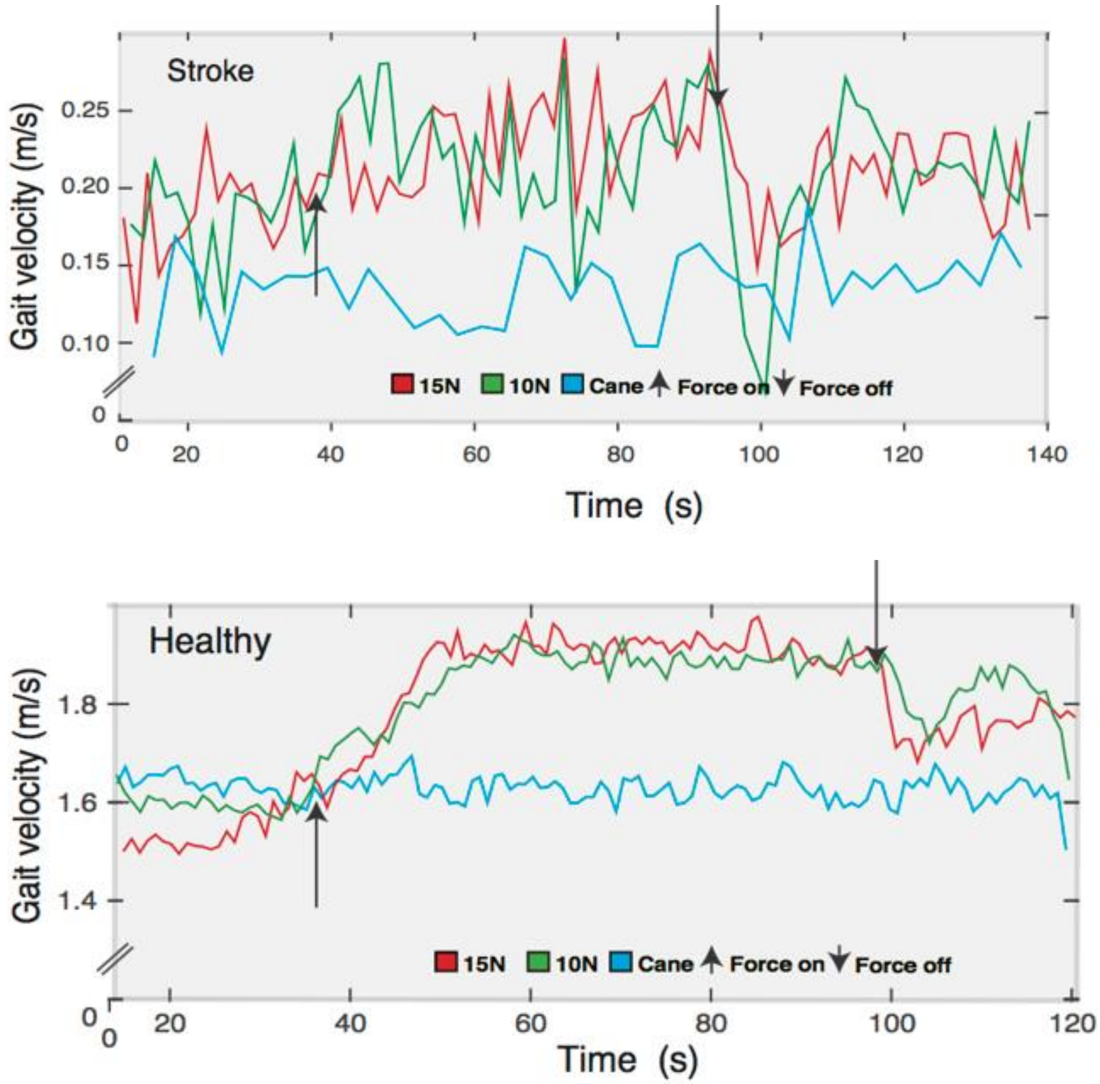


Fig. 5.3: Instantaneous gait velocity traces for both a post-stroke and control subject.

Instantaneous gait velocity traces for age-matched controls (top) and post-stroke (bottom) participants during 10N, 15N and cane trials. The arrows demarcate the onset and removal of the haptic force.

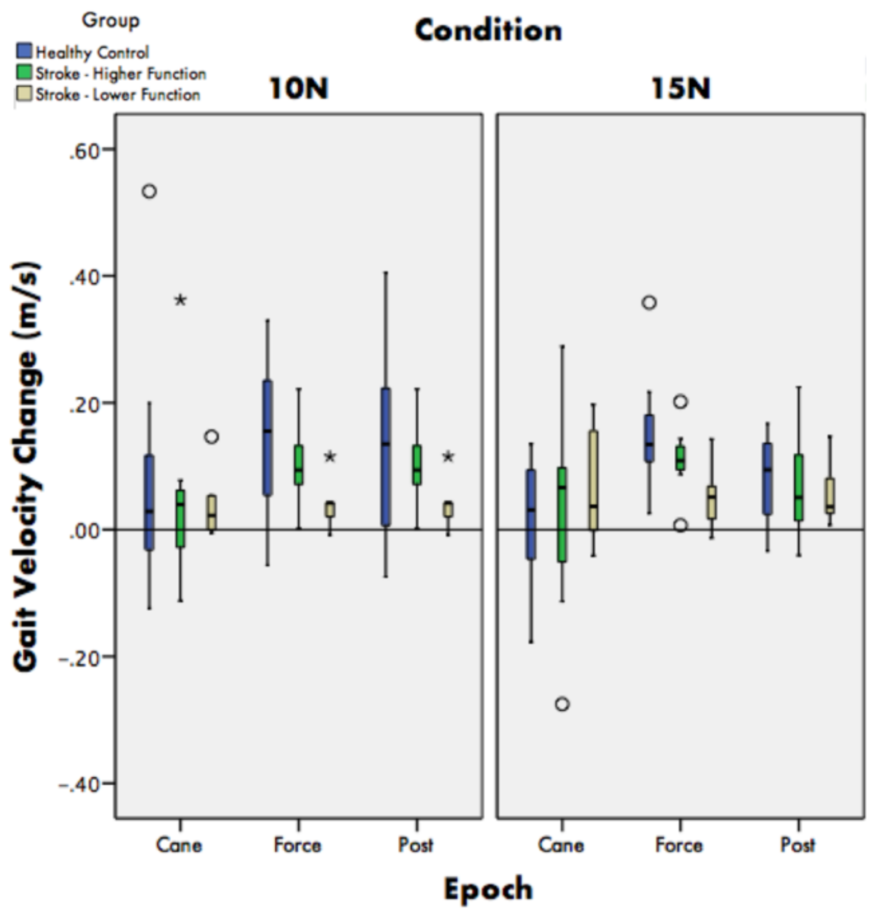


Fig. 5.4: Median change in walking velocity of stroke and controls.

Boxplot and whisker plots for walking velocity of stroke and age-matched controls in force, and post-force epochs for 10N and 15N and cane conditions, compared to baseline (pre-force).

Both lower and higher functioning stroke subjects generally walked faster with a cane compared to pre-force levels, with changes from 0.03 m/s to 0.07 m/s, relative to pre-force baselines. A similar average change of 0.07 m/s 95% CI [0.02m/s - 0.15m/s] to 0.02 m/s 95%CI [-0.03m/s - 0.07m/s] was found in healthy controls (Fig. 5.3).

Figure 5.4 illustrates gait velocity in 10N and 15N force trials (green and red traces), as well as walking with the cane (blue trace) for a healthy and post-stroke subject. A typical occurrence across post-stroke and healthy controls during force onset (upward arrow) was a considerable increase in gait velocity well above pre-force levels. What can be noted is that upon the release of force (downward arrow), velocity levels tended to decrease, but remained above pre-force levels. The participant's velocity while walking with a cane remained fairly constant throughout the trial.

5.4.2 Stride Time and Stride Length

Strides times decreased during the force epoch for the lower functioning stroke subjects, relative to the pre-force epoch, with averages of 0.07s 95% CI [0.01 – 0.13s] and 0.10s [0.06 – 0.13s] for 10 and 15N conditions, respectively. Stride times were also reduced during the post-force epoch by 0.05s 95% CI [0.03 – 0.08s] and 0.07s [0.02 - 0.12s] for 10 and 15N conditions (Fig. 5.5). Higher functioning stroke subjects also reduced stride times during the force epoch as average changes were 0.11s 95% CI [-0.01 – 0.22s] and 0.08s [0.05 – 0.12s] for 10 and 15N conditions. Post-force epoch stride times remained below baseline levels by 0.09s 95% CI [-0.01 – 0.19s] and 0.07s [0.03 - 0.12s] for the 10 and 15N conditions. Healthy controls also reduced their stride times by 0.04s [0.02s – 0.05s] for the force epoch and maintained reduced stride times during the post-force epoch by 0.04s [0.02 -0.06s] for the 10 and 0.03s [0.01-0.05] for 15N condition. During cane trials, lower functioning post-stroke and healthy control subjects increased stride time by 0.01s 95% CI [0.01 – 0.16s] 0.05s 95% CI [0.01 – 0.09s] , while only higher functioning stroke subjects on average lowered stride times by 0.01s. Slight increases were seen among control subjects during cane walking with an average of 0.02s 95% CI [-0.03 – 0.05] and 0.01s 95% CI [-0.02 – 0.06] compared to 10 and 15N force condition levels, respectively. The mixed model revealed a significant interaction between epoch and group $F(1,318) = 5.6$, $p<0.005$.

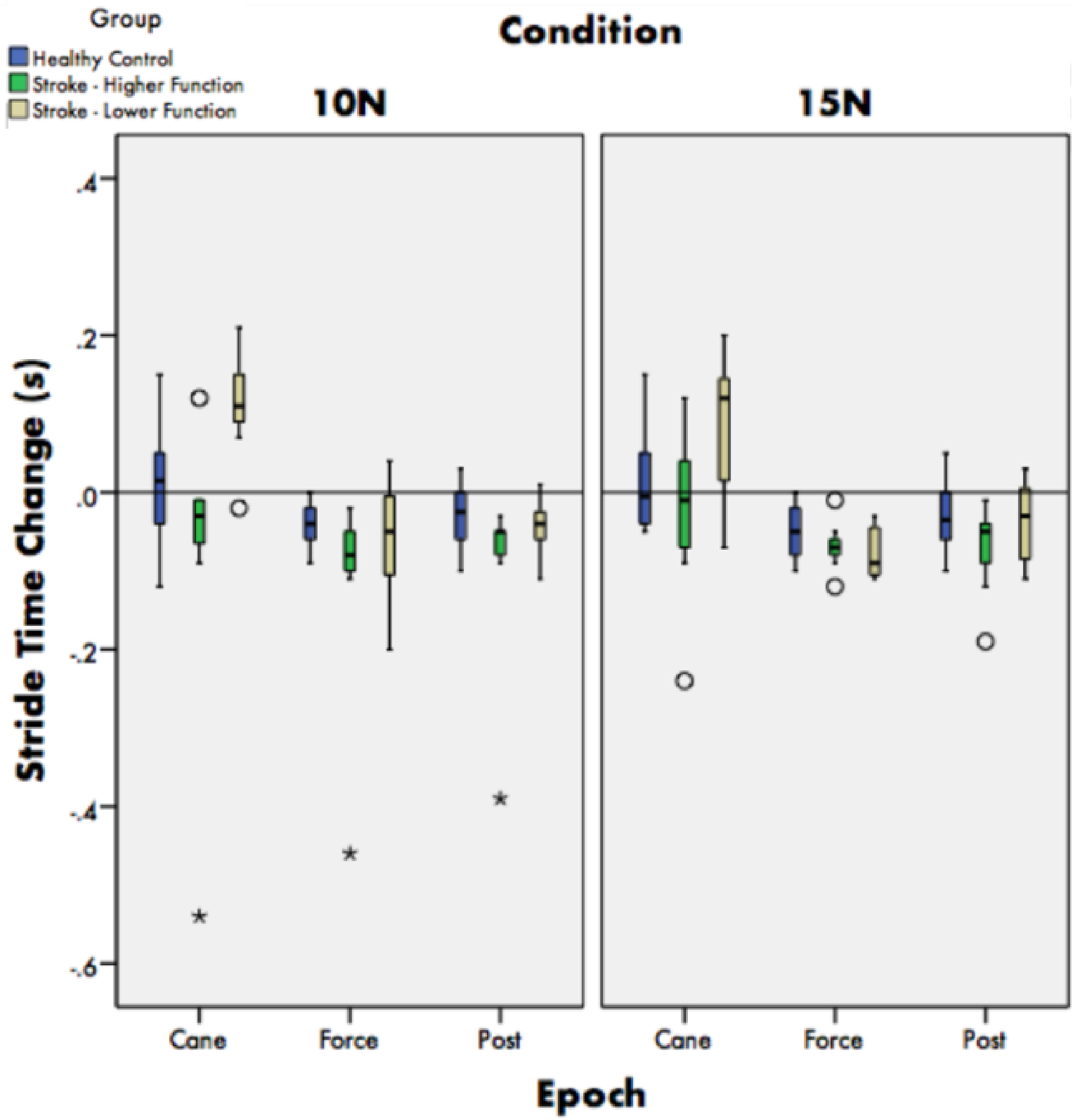


Fig. 5.5: Median change in stride time of the paretic and non-dominant limb.

Box and whisker plots for stride time of the paretic (stroke) and non-dominant (healthy) side, relative to pre-force levels for stroke and healthy controls during force, post-force and cane epochs.

Both lower and higher functioning stroke subjects on average increased stride length during the force epoch, relative to the pre-force epoch, by 0.04m [0.02 – 0.05m] and 0.07m [0.02 – 0.13m] for 10N and 0.05m [0.01

– 0.10] and 0.09 [0.05 – 0.13] 15N conditions, respectively (Fig. 5.6). When the force was removed, lower and higher functioning stroke subjects maintained increased stride lengths by 0.04m 95% CI [0.02 - 0.08m] and 0.01 [-0.02 – 0.05m] in the 10N condition and 0.07m 95% CI [0.03 – 0.12m] for 0.05m [-0.01 – 0.10m] for the 15N condition. Healthy controls also increased stride lengths by 0.10m [0.60 – 0.15m] during 10N and 0.08m [0.05 – 0.10m] during 15N force conditions, relative to pre-force levels. During the post-force epoch, healthy subjects maintained increases in stride lengths ranging from 0.11m [0.03m – 0.18m] for the 10N and 0.04m and [0.01 – 0.07m] for the 15N condition. Stride length increases relative to pre-force were also seen when subjects walked with the cane. Lower functioning stroke subjects increased stride length by 0.08m 95% CI [0.02 – 0.15m] and 0.12m [0.02 – 0.22m] compared to 10N and 15N pre-force levels, while higher functioning stroke subjects had smaller increases of 0.03m relative to 10N levels 95% CI [-0.08 – 0.14m] and 15N level [-0.08-0.15m]. Control subjects also increased stride length by 0.4m 95% CI [0.02 – 0.05m] and 0.07m [0.02 – 0.13m] relative to 10N and 15N force levels during cane walking. A mixed model was used to evaluate stride time $F(2,318) = 29.5$, $p<0.0005$ for epoch and $F(1,318) = 21.0$, $p<0.0005$.

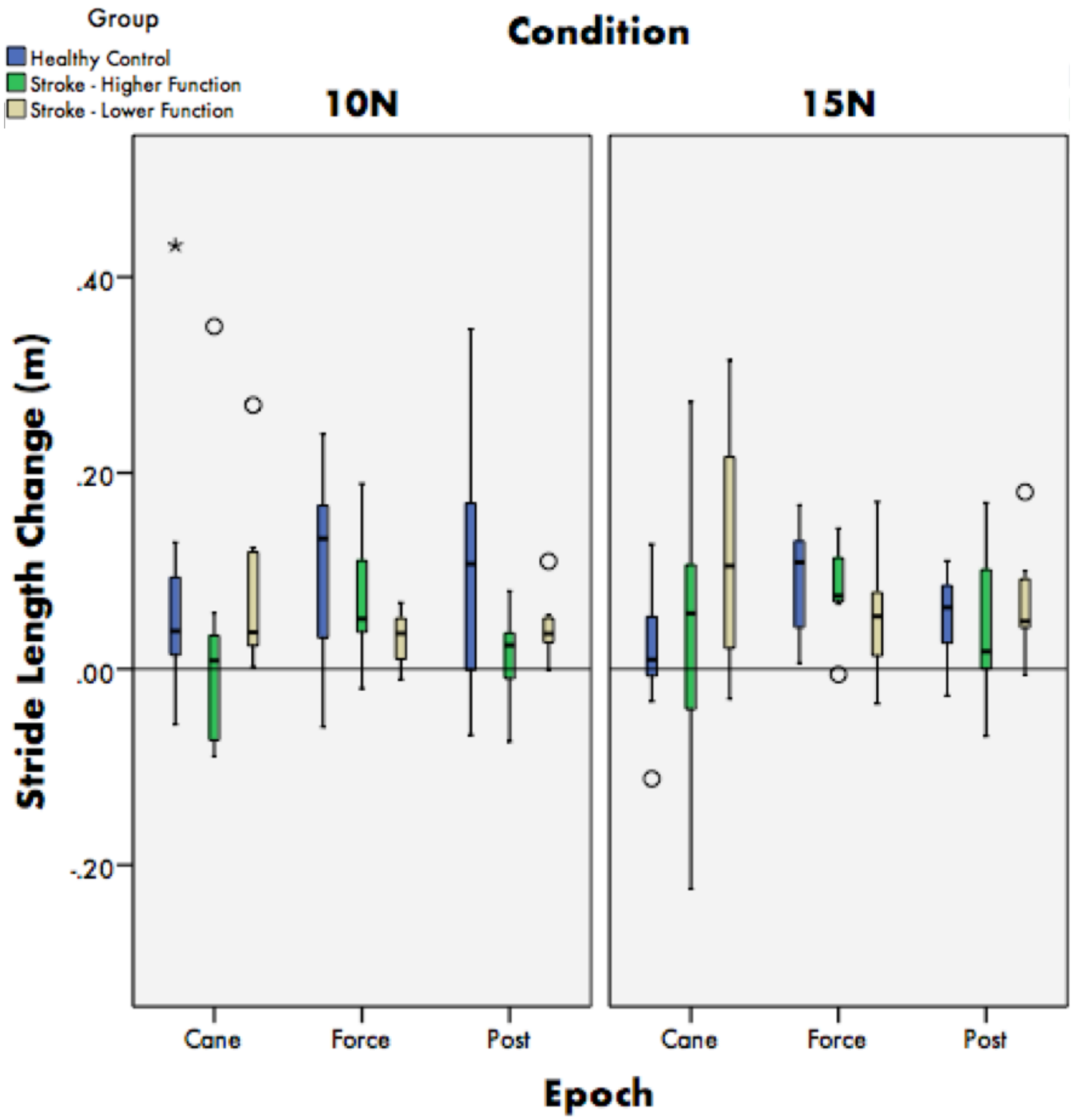


Fig. 5.6: Median change in stride length of the paretic and non-dominant limb.

Box and whisker plots of stride length of the paretic (stroke) and non-dominant (healthy) side, relative to pre-force levels for stroke and healthy controls during force, post-force and cane epochs.

5.4.3 Centre of Mass

The average center of mass (COM) excursions for the 10N force epoch, cane trial and free walk (no leash in hand) for a representative healthy control subject, a higher functioning stroke subject and lower functioning stroke subject are illustrated in Figure 5.7. All three subjects depicted in Figure 5.7 held both the leash and cane with the left hand. When these subjects walked with a cane, the COM tended to stay toward the left (non-paretic) on the medial-lateral plane. However, when the post-stroke subjects walked with a leash vector of 10 or 15N tension applied directly forward with regard to the hand during the force epoch, and later in the post-force epoch, the center of mass tended to shift towards the paretic side in the mediolateral plane. This mediolateral shift to the paretic side was found in most subjects, particularly during the 15N exposure. Specifically, both the lower and higher functioning post-stroke subjects shifted their CoM shifted by 0.01 95% CI [-0.03 – 0.01m] and 0.02 [0.03 – 0.01m] during the 15N force epoch. These subjects would maintain this shift to the paretic side on average by 0.01m 95% CI [-0.03 – 0.01m] and 0.01m [-0.04 – 0.01m] during the post-force epoch during 15N force conditions. Both lower and higher subjects had minimal changes of COM during the 10N force condition with changes in force and post-forces ranging from -0.01m – 0.01m relative to the leash hand.

The healthy control depicted in Figure 5.7 showed the tendency to shift the COM more to the side of the cane. The COM displacement seen when post-stroke subjects are exposed to 15N of force was similar in pattern and magnitude to the 10N condition. Most notably, in Figure 5.8 (right panels), the lower functioning post-stroke subject kept a mediolateral displacement of 0.06m relative to the COM position when walking with a cane. The higher functioning post-stroke subject shifted approximately 0.04m relative the side holding the leash. As in the 10N condition, the control subject also shifted posture mediolaterally, but only slightly away from the dominant hand.

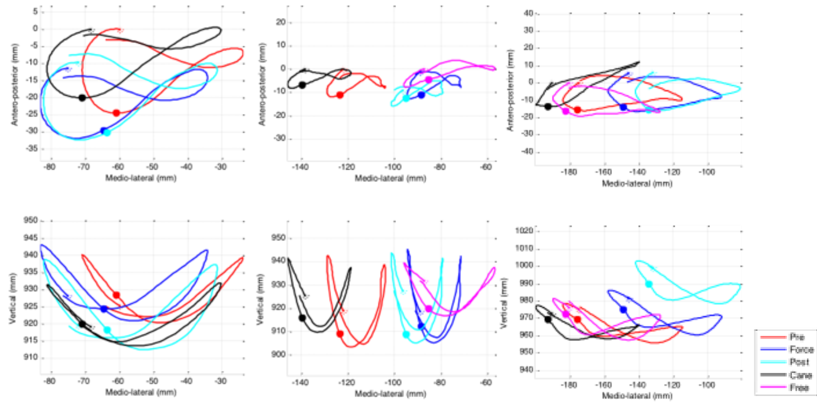


Fig. 5.7: Average center of mass excursion per epoch for the 10N condition.

Average mediolateral (top row) and vertical (bottom row) center of mass excursions (mm) per epoch for the 10N condition compared to cane and free walk for healthy controls, higher functioning and lower functioning stroke (left to right).

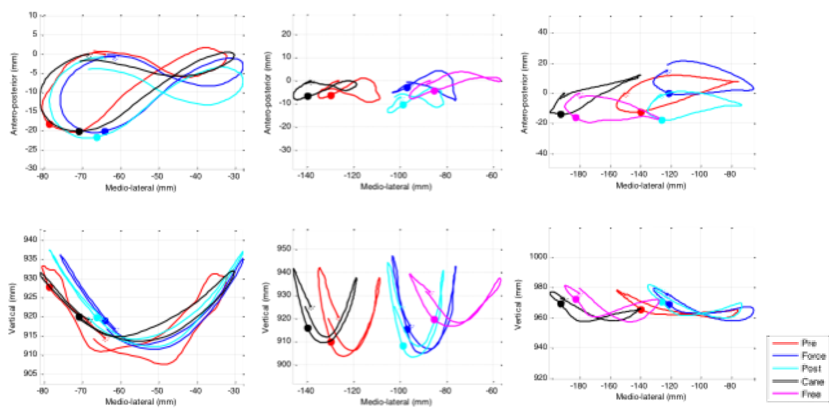


Fig. 5.8: Average center of mass excursions per epoch for the 15N condition.

Average mediolateral (top row) and vertical (bottom row) center of mass excursions (mm) per epoch for the 15N condition compared to cane and free walk for healthy controls, higher fuctioning and lower functioning stroke (left to right).

5.4.4 Paretic limb coordination

Average segment angles depicted in fig. 5.10 were calculated by adding together the minimum perpendicular lengths from traces to best fit plane for the entire gait cycle. Using the total sum of squares approach the sum was then divided by the number of trajectories to give the average limb segment orientations of the thigh, leg and foot. Illustrated in figure 5.9 are the average angle changes of the paretic foot, leg and thigh limb segments over the entire gait cycle for force and post-force epochs during 10, 15N and cane conditions. Changes are also depicted for healthy controls during the same conditions. For stroke subjects, the effect of walking with either a 10 or 15N force, on average, increased dorsiflexion and hip flexion across the gait cycle. For example, with respect to the pre-force epoch the 10N force brought on an average change of 9.1° 95% CI [-3.2 – 21.3°] positive ankle flexion (dorsiflexion) and an 8.6° 95% CI [-4.5 – 21.8°] change in hip flexion for the lower functioning stroke group. The same dorsiflexion and hip flexion trend occurred for the higher functioning group with smaller angle changes of 1.7° 95% CI [-4.5 – 21.8] and 2.5° [-3.1 – 8.2°] for the foot and thigh. When the 10N force was removed, both lower and higher functioning stroke groups maintained 6.8° 95% CI [2.2 – 11.4°] and 1.7° [-4.2 – 7.6°] dorsiflexion and 4.3° 95% CI [-3.7 – 12.3°] and 2.5° [-3.1 – 8.2°] of hip flexion changes, respectively. In regard to the 15N force epoch, the lower functioning stroke group showed a change of -2.4° 95% CI [-6.1 – 1.3°] dorsiflexion and 2.0° [-1.3 - 0.2°] hip flexion. The higher functioning group showed a more robust effect with the 15N force with changes of 11.2° 95% CI [0.3 – 22.1°] dorsiflexion and 9.7° [-0.5 – 19.9] hip flexion. The effect was found even after the 15N removal with 2.8° 95% CI [-0.4 – 6.0°] and 4.5° [-.2 – 9.1°] dorsiflexion and 4.4° 95% CI [-0.8 – 9.5°] and 4.9° [-2.4 – 12.1°] hip flexion, for lower and higher functioning stroke groups, respectively. In contrast, knee flexion tended to decrease relative to baseline across epoch and force levels for the lower functioning group. The higher functioning group in contrast tended to show slight changes knee flexion (see fig. 5.9).

Walking with a cane, with respect to the leash, proved to change some factors in paretic lower limb coordination. For example, the lower functioning stroke group tended to also dorsiflex the paretic foot by -4.8° 95% CI [-1.9 - 11.5°] but favor hip extension with a -3.4° 95% CI [-20.1 – 13.4°] relative to 10N pre-force levels and 2.1° 95% CI [-6.0 - 10.2°] and 2.8° [-19.0 – 13.4°] for the foot and hip extension with respect to 15N force. Interestingly, higher functioning tended to favor a foot plantarflexion change of 4.34° 95% CI [-15.5 – 6.8°] and -3.3° [-11.6 – 5.0°] relative to 10N and 15N forces, respectively. Hip flexion remained almost unchanged with a 0.6° 95% CI [-6.7 – 7.9°] and 0.9° [-7.9 – 9.7°] change relative to 10N and 15N epochs.

Healthy controls when exposed to the 10N condition showed very light changes in foot, leg and thigh coordination during force and post-force epochs (fig. 5.9). Changes were on average 0.2° 95% CI [-1.8 – 3.2°] dorsiflexion and 2.3° [-2.0 – 2.3] hip flexion during the 10N force and 0.7° 95% CI [-6.8 – 5.5°] plantarflexion and 0.7° [-5.4 – 4.0] hip extension change during the post-force epoch. The 15N condition brought on changes of 3.1° 95% CI [-8.3 – 2.1°] plantarflexion and 3.3° [-7.1 - 0.4°] of hip extension. When the 15N force was removed changes of 3.6° 95% CI [-10.4 – 3.2°] plantarflexion and 3.9 [-7.5 - -0.3°] hip extension were maintained during the post-force.

During cane walking, healthy controls showed, foot plantarflexion changes of 8.7° 95% CI [-17.7 – 0.3°] and minimal hip extension of 0.4° [-3.4 – 2.5°] relative to the 10N pre-force values. Changes of 9.4° 95%

CI [-17.5 – 1.4°] plantar flexion and 1.2° [-3.9 – 1.5°] hip extension were also found during cane walking with respect to 15N baseline values.

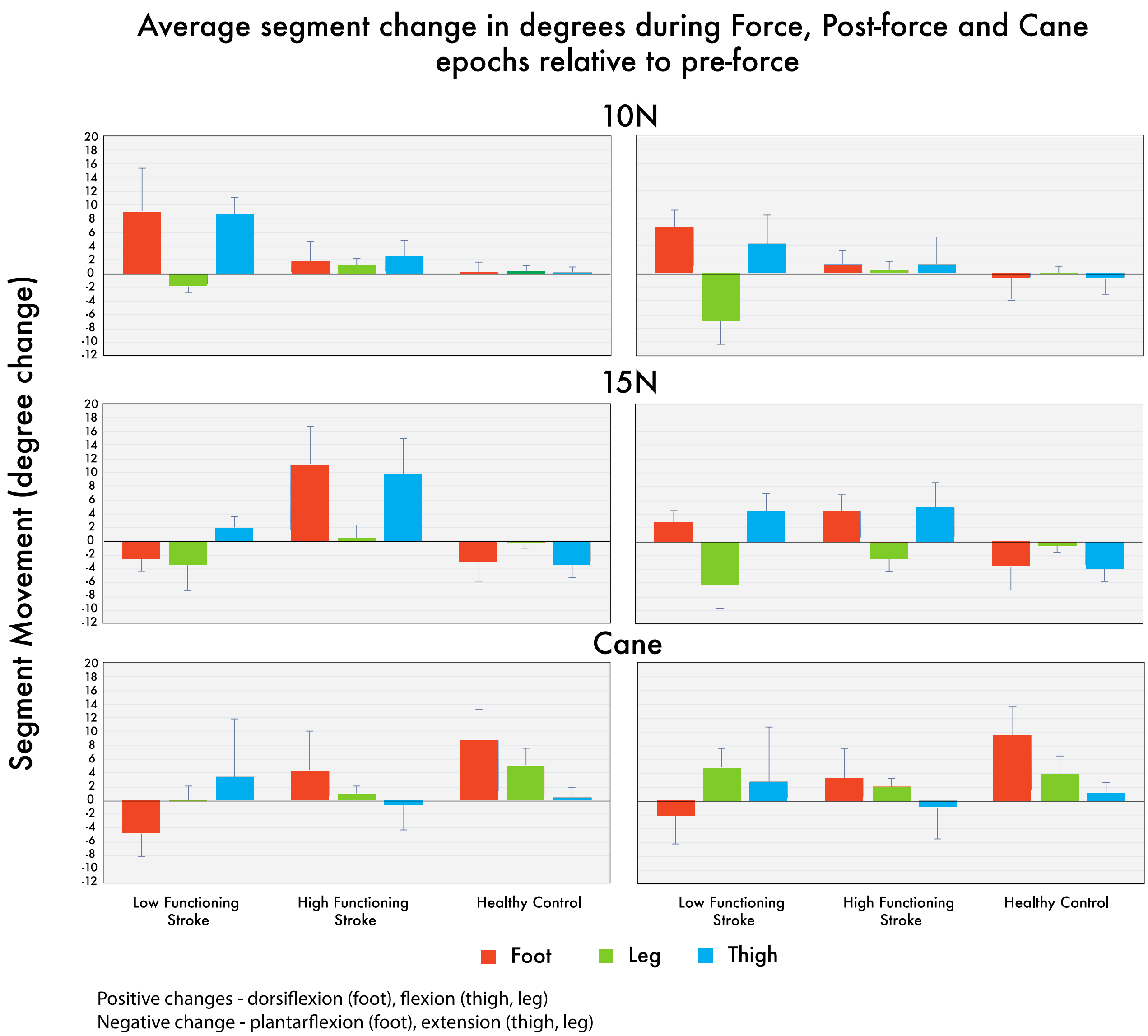


Fig. 5.9: Average lower limb segment changes per relative to the pre-force epoch for the leash and cane. Average thigh, leg and foot segment changes (in degrees) across the entire gait cycle for the force and post-force epochs for the 10N, 15N and Cane trials relative to pre-force baseline.

5.4.5 Lower limb angular momentum

The angular momentum combining thigh, leg and foot segments was calculated by attaining the vector product of the three angles of each segment (thigh, leg and foot) and the angular velocity of these angles over the entire course of the gait cycle. The resulting angular momentum vectors of both legs can be seen in figure 5.11 and are represented by blue (right) and red (left) arrows. The blue and red planes are exactly perpendicular to the angular momentum vectors.

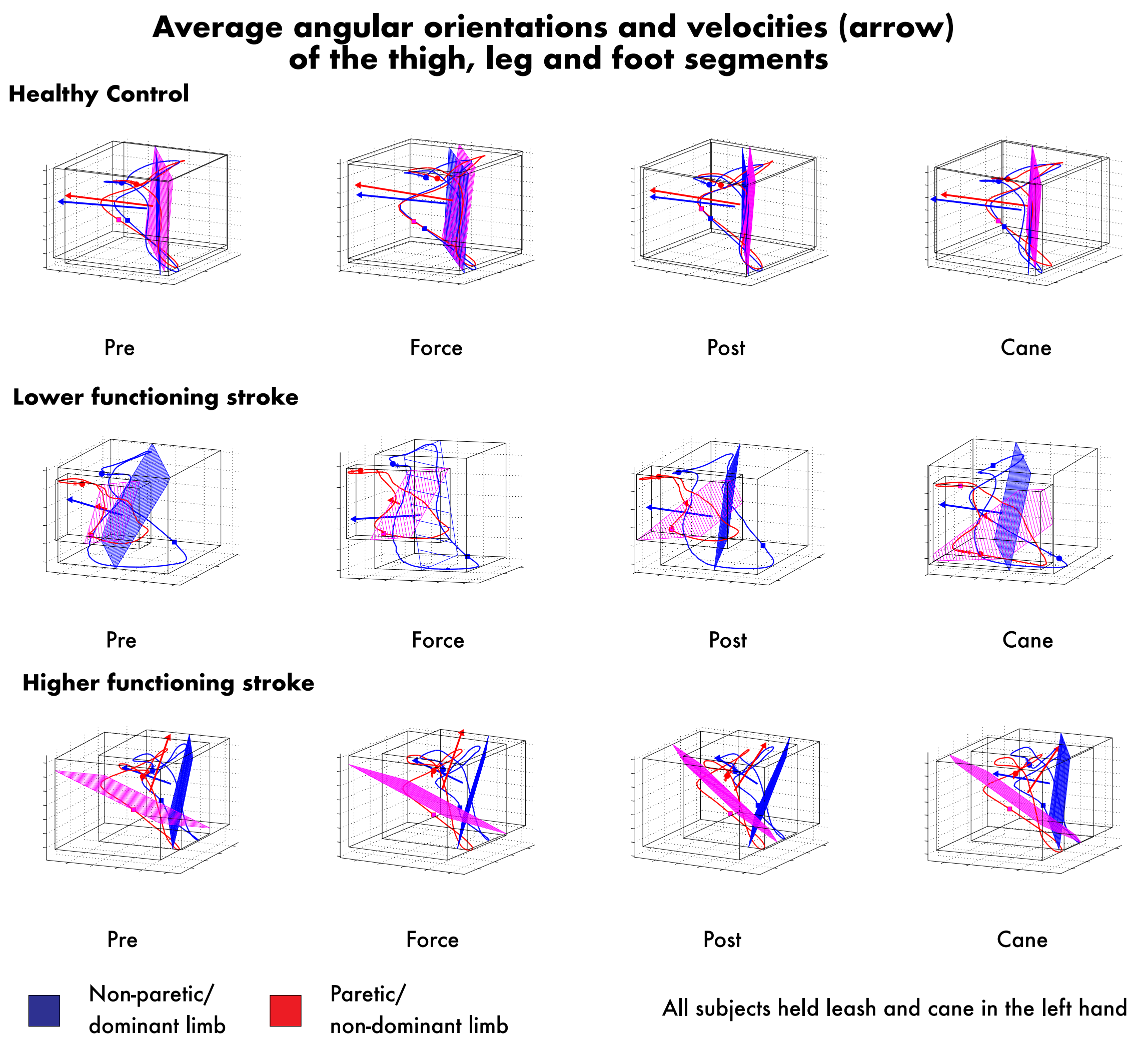


Fig. 5.10: Average lower limb segment angular orientations and angular momentum in the pre-force epoch.

Average angular orientation (traces) and momentum (arrows) of the thigh, leg and foot segments of the paretic/non-dominant (red) and non–paretic/dominant (blue) legs for control, higher and lower functioning post-stroke subjects during pre-force, force, post-force epoch of a 10N trial and cane trial.

A distinct trend can be seen in terms of angular momentum changes during force and post-force during leash walking and cane walking. Relative to pre-force levels, angular momentum of the leg on the paretic side for lower and higher functioning stroke was 0.3 kg • m^2/s 95% CI [0.0 - 0.6 kg • m^2/s] and 0.9 kg • m^2/s [0.1 – 1.8 kg • m^2/s], respectively, during the 10N force epoch. Angular momentum remained changed by 0.3 kg • m^2/s 95% CI [-0.1 - 0.7 kg • m^2/s] and 0.6 kg • m^2/s [0.1 – 1.4 kg • m^2/s] during the 10N post-force epoch. The non-paretic leg also increased its angular momentum on average 1.4 kg • m^2/s 95% CI [0.5 – 2.3 kg • m^2/s] and 1.4 kg • m^2/s [0.3 – 2.4 kg • m^2/s], for lower and higher functioning stroke subjects, during the 10N force epoch. Angular momentum stayed above pre-force levels by 1.5kg • m^2/s 95% CI [0.4 – 2.6 kg • m^2/s] and 0.5 kg • m^2/s [-0.6 – 1.5 kg • m^2/s]. Similar angular momentum changes were seen for the 15N force condition for both stroke subject groups across epochs (see fig. 5.11). Cane walking also showed similar trend in angular momentum change for the paretic limb. Specifically, both lower and higher functioning stroke subjects changed angular momentum values by 0.3 kg • m^2/s 95% CI [-0.1 – 0.8 kg • m^2/s] and 0.8 kg • m^2/s [-1.6 – 3.2 kg • m^2/s] for the paretic leg with respect to 10N pre-force condition and 0.4 kg • m^2/s 95% CI [-0.2 – 0.9 kg • m^2/s] and 0.3 kg • m^2/s [-1.9 – 3.3 kg • m^2/s] with respect to the 15N.

Healthy controls also increased their angular momentum during 10 and 15N force and post-force epochs. For example, subjects changed angular momentum values by 2.3 kg • m^2/s 95% CI [1.3 – 3.3 kg • m^2/s] for the non-dominant leg during the 10N force epoch. Angular momentum would remain changed by 2.7 kg • m^2/s [0.7 – 4.7 kg • m^2/s]. The dominant side also showed the same effect with changes of 1.7 kg • m^2/s

95% CI [-0.1 – 3.5 kg • m^2/s] and 3.2 kg • m^2/s 95% CI [1.1 – 5.3 kg • m^2/s] for the 10N force and post-force epochs, respectively. Similar results were found for the 15N force condition (fig. 5.11). Walking with a cane showed increases in angular momentum for both non-dominant and dominant legs when compared to 10N and 15N pre-force levels. For example, average angular momentum increases of the non-dominant leg were 2.4 kg • m^2/s 95% CI [0.5 – 4.3 kg • m^2/s] and 2.7 kg • m^2/s 95% CI [-0.5 – 4.9 kg • m^2/s] for the dominant leg, relative to the 10N pre-force levels (fig. 5.11).

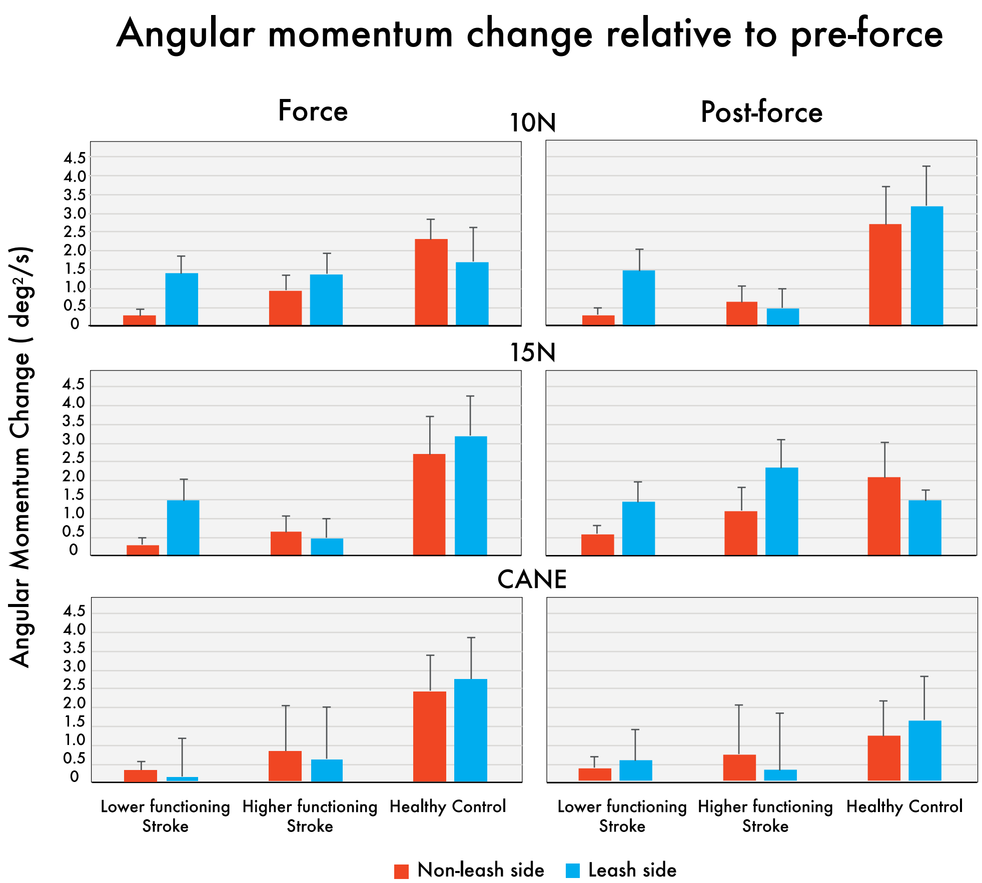


Fig. 5.11: Average bilateral angular momentum changes of lower limb segments.

Average changes in angular momentum, for stroke and control subjects during force and post-force epochs for 10N, 15N and cane conditions relative to pre-force levels.

5.5 Discussion

The first objective of this study was to assess the extent to which haptic forces applied to the hand, in the direction of walking, could cause an adaptation behavior in gait, followed by a post-adaptation response, in post-stroke individuals compared to healthy controls. A previous study (Sorrento et al., 2013) revealed that the effect was seen in healthy young adults and could have potential clinical implications if post-stroke individuals show similar adaptations. In that study, it was believed that the walking adaptations were achieved as a result of a transient change in the normal motor program that was brought on by the haptic force exposure. The presence of a post-adaptation effect suggests that the transient motor program can be retained, even after the haptic forces are removed. It remains to be seen if the post-adaptation effects can be consolidated into an abiding motor program by rest and subsequent sessions on different days.

The outcomes chosen for the study were intended to depict not only changes in spatiotemporal gait parameters, but also postural stability, as measured by COM displacement and limb segment coordination, that can impact functional locomotion (Lamontagne & Fung, 2009). In this respect, important findings have been measured as far as adaptations and post-adaptation in response to force onset and removal. For example, this was clear with both post-stroke and healthy age-matched controls in relation to gait velocity. Just as previously seen with healthy young subjects, gait velocity changes proved to be robust across conditions and epochs. Almost every subject in the study increased their gait velocity when being pulled with 10N of force, while all but one participant managed a velocity increase with 15N of force. This participant's gait velocity decreased as the 15N leash trial progressed, perhaps in this case suggesting a fatigue effect.

The force epoch showed particularly robust results for the higher functioning post-stroke patients and healthy controls. These groups on average increased their gait velocity by at least 0.1 m/s for the 10N and 15N conditions. It should be noted that in functional terms, such values would potentially meet the significant 0.1m/s clinically meaningful threshold for gait velocity described in the literature (R. W. Bohannon, 2009). The lower functioning post-stroke group had more modest, yet promising changes for both 10 and 15N conditions. While the magnitude of change was smaller, the tendency to adapt to the force and increase velocity was just as systematic. Nevertheless, a change of 0.05m/s is consistent with acceptable moderate clinical improvements (R. W. Bohannon, 2009). These findings address an important question on whether post-stroke individuals can adapt their gait pattern in response to changes of haptic forces. Although the higher and lower functional post-stroke individuals adapted their gait patterns to different extents, as compared to healthy controls, the presence of a post-adaptation effect in all subjects suggests gait training with VR and haptic forces can be potentially beneficial. However, the effects of long-term practice remain to be investigated.

The post-adaptation effects were also evident from the increase in walking velocity above baseline levels, as the haptic force was removed. As seen in the previous study with younger healthy adults, the presence of a post-adaptation suggests the carry-over effect of an altered motor program (Sorrento et al., 2013). Specifically, both lower and higher functioning post-stroke subjects maintained a moderate increase of 0.04 to 0.07 m/s for the 10 and 15N conditions, while the healthy controls also maintained robust changes of 0.08 to 0.13 m/s. Perhaps the significance of these changes is brought into perspective when compared to the gait velocities with the instrumented cane. Walking with the cane did tend to increase gait velocity slightly, as post-stroke subjects walked on average 0.03m/s to 0.07m/s for 10 and 15N conditions, while the healthy controls similarly increased velocity by 0.02m/s to 0.07m/s. However, comparing the changes in gait velocity between the leash and the cane revealed that in most cases, walking with a 10 or 15N force lead to greater increases in gait velocity. In addition, gait velocity changes for the higher functioning post-stroke and control subjects tended to remain comparable or greater than during cane walking in post-force epoch. This suggests that walking independently with post-adaptation effects after exposure to a haptic force is just as fast, if not faster than when walking with the instrumented cane.

Changes in gait velocity can result from either a decrease in stride duration or an increase in stride length. But, it is also plausible that a combination of both can contribute to gait velocity increases. For both groups of post-stroke and healthy subjects both factors indeed contribute to the increase in gait velocities when walking with increased tensile forces. In terms of stride duration, both lower and higher functioning post-stroke subjects reduced stride duration of the paretic side by a range of 0.05s to 0.12s when a 10N or 15N force was present. Controls also reduced stride time by a similar margin. During the post-force epoch, both post-stroke subjects and controls maintained a reduced stride time by an average of 0.01s to 0.07s for the 10N and 15N conditions. In contrast to the cane, both higher functioning post-stroke subjects and healthy controls reduced stride time to a similar extent as the post-force epoch. However, the lower functioning stroke group on average increased stride time. In this case, perhaps there was a tendency to spend more time with weight shifted on the cane, which would consequently lead to increased stride time on the paretic leg. Stride length also changed relative to pre-force levels during the force epoch. Post-stroke subjects increased stride length an average of 0.04m and 0.07m for 10 and 15N, respectively. Controls increased on average 0.11m. Both groups maintained changes in stride length, which during the post-force epoch were very comparable with the changes brought on by using the cane.

The next step in assessing the effects of haptic forces in comparison to an instrumented cane was to understand how these changes affect posture. Using the average COM trajectory to determine postural reactions to steady-state walking during the three epoch trial and cane trial, it was evident the dynamic postural reactions of the post-stroke population differed to some degree relative to the control group. Specifically, when the post-stroke population walks with the leash, relative to the cane, this population had the general tendency to shift the CoM medial-laterally to the paretic side. This effect was also maintained during the post-force epoch. This was especially the case during the 15N condition where up to 9 of 14 subjects showed this trend with changes of ~0.01 m for force and post-force epochs. The healthy controls generally had some tendency to favor the non-dominant side (analogous to the paretic side in stroke), but the difference in mediolateral shift of the COM towards the non-dominant side during force and post-force epochs in 10 and 15N condition relative to the cane position is much less in magnitude, as compared to post-stroke individuals. Post-stroke individuals might have over-compensated by weight-bearing more with the cane, as commonly seen during overground walking. Such over-compensation may be remediated by a low level haptic tensile force, directly in the direction of locomotion, afforded by the leash. Such a strategy can provide a secure means to facilitate forward progression.

The objectives of this study were intended to build on the findings of a previous proof of principle study, which reported the adaptations followed by post-adaptations in response to a haptic tensile force applied to the hand while walking in a virtual environment. The potential clinical implication is the sensorimotor enhancement afforded by such intervention, as post-stroke individuals of varying functional levels are capable of improving their gait outcomes with a haptic leash. Perhaps more interesting is that many post-stroke individuals exhibited the ensuing post-adaptation that was on par with or even better than walking with a cane.

A favourable short-term assessment of the haptic force strategy was present, but it remains to be seen if longer term effects can be harnessed from training with the adaptation and post-adaptation process with haptic forces over multiple sessions. It should be noted that while in other studies such as split-belt treadmill walking, there is yet to establish evidence of longer lasting post-adaptation effects, particularly with overground transfer (Reisman et al., 2009b). To address the perspective of the haptic leash inducing more abiding post-adaptations, one may want to consider the differences in adaptations derived from a split-belt treadmill, which uses set-paced speed, compared to a self-paced treadmill. In fact, one study found that for able-bodied and transtibial amputee populations, spatiotemporal and range-of-motion changes where more consistent with natural gait compared to set-paced treadmill speeds. Set-paced walking tended to bring on gait compensations to cope with changes in gait speed (Sinitski et al., 2015). Interestingly, this study also coupled the self-paced treadmill with the CAREN virtual environment. From this perspective, there might be an encouraging advantage to the haptic leash strategy when compared to the split-belt technique as it may promote more natural post-adaptation gait profile. This coupled with the knowledge of haptic inputs and their role in gait control (Knaepen et al., 2015) may contribute to longer lasting post-adaptations. Consequently, this may be beneficial to longer lasting changes in the event of a training protocol or overground walking. To investigate this question, a future direction in the development of a training protocol is to investigate the optimal adaptation and post-adaptation duration that will lead to maximal gait improvements, and to determine the efficacy of such an intervention through a randomized clinical trial in acute stroke patients.

The average lower limb segment changes provided an interesting and insightful evaluation of leash walking in comparison to walking the cane. The comparison was not intended to evaluate the leash's efficacy over that of the cane, rather it was to shed light on the characteristic changes that a post-stroke individual could benefit from with an eventual leash training protocol. The leash definitely did change the coordination of the paretic leg upon exposure during the force epoch, and in keeping with other outcomes in this study, maintained changes during the post-force epoch. Both the lower and higher functioning stroke groups tended to increase dorsiflexion and hip flexion during the force and post-force epochs. The evidence of stroke individuals using an increase in foot strategy, as evidence by dorsiflexion increases, is an encouraging finding especially since the effect is more robust in the lower functioning group. If we were to compare these subjects' coordination with cane walking, on can see that during the cane condition the increase in dorsiflexion is almost absent. In fact, it can be seen that the higher functioning stroke and control groups tend to take on more of a foot and knee strategy during cane walking. The effects of increased dorsiflexion and hip flexion may be conducive to mitigating, or even correcting to some extent the effect of paretic leg circumduction, which is present in many post-stroke individuals. Also of note is the apparent decrease in knee flexion as evidenced in the negative change of the paretic leg segment. This may in fact counter the positive effects seen in dorsiflexion and hip flexion increases.

Finally, angular momentum changes reported served to substantiate the foot dorsiflexion and hip flexion strategies measured during leash walking with respect to cane walking. In fact this change in paretic coordination may have been driven by the increase in angular momentum measured in both limbs. It is plausible that the forward pull of the leash could, to some extent, engage the lower limb to stabilize and counter the forward force by increasing joint torques and utilizing more rotational kinetic energy. This in turn, may well have driven the observations of COM shift to the paretic side and increased dorsiflexion and hip flexion, which were mainly absent during cane walking.

5.6 Conclusions

Post-stroke individuals with recovered walking ability adapted to increased haptic forces by increasing their gait speeds. In fact, those walking above the threshold of community ambulation (0.8m/s) adapted similarly to healthy subjects. Increased gait velocity was also maintained after haptic force removal, yielding a post-adaptation effect. The post-adaptation changes seen in post-stroke, particularly higher functioning subjects, were comparable to walking with a cane. The adaptation and post-adaptation effects resulting in increased gait velocity can be explained by both decreases in stride duration and increases in stride length in response to haptic force changes.

Along with the changes in spatiotemporal parameters, post-stroke subjects walking with increased tensile forces also tend to favor a greater use of the paretic side, based on average CoM displacement. Lastly, these spatiotemporal and postural changes are accompanied with increases in dorsiflexion and hip flexion among stroke subjects during force and post-force epochs. In terms of lower limb coordination the leash strategy proved to differ from the cane strategy. A global description can be made of a stroke individual increasing gait velocity by shifting the COM to the paretic side, while engaging more dorsiflexion and hip flexion. Little evidence was found, however, for increased knee flexion. It is likely that this change in coordination was driven in part by increases in lower limb angular momentum. Angular momentum of the paretic leg was comparable, if not at times greater during force and post-force epochs relative to the cane. This description would most likely enable the stroke individual to move away from compensation strategies, such as leg circumduction, and improve movement economy. Further study should examine the extent to which prolonged exposure to leash forces, perhaps in the form of a training protocol, aimed at altering the motor programs that bring on gait adaptation and post-adaptation effects.

CHAPTER 6: DISCUSSION AND CONCLUSIONS

6.1 Summary of Results

To recall the objectives of this three-part dissertation, the global research question was framed as the following: *in stroke and healthy populations, to what extent do handheld haptic tensile forces, pulling in the direction of walking, compared no force or a cane, change spatiotemporal, postural, coordination and gait symmetry outcomes*. This was achieved by means of an innovative combination of virtual reality and robotics in a mixed-reality system. This mixed-reality system was designed to simulate the effects of walking with a dog in an urban setting. Combining the haptic input and ecological context of this simulation could potentially lead to direct real-world transfer and implementation into a clinical set-up. The first step in achieving this was the development of a proof of concept pilot trial, establishing evidence for the efficacy of this haptic-based strategy among healthy young subjects. Any changes in gait parameters in the initial study would make way for the main phases of the project involving an older chronic stroke population.

6.1.1 Study 1: Healthy young pilot study

The first of a three-part project was a pilot study to test the hypothesis that haptic tensile forces in the form of a leash can change gait parameters in a young healthy population. Such changes would be seen both during and after force exposure, relative to baseline walking. The study was intended to establish proof of concept that such tensile forces could cause changes in gait parameters. Therefore, results were limited to spatiotemporal outcomes. Indeed, gait velocity proved to increase when subjects walked with the haptic force. Gait velocity was also maintained above pre-force velocity levels after the force was removed. The two phenomena observed were referred to as adaptation and post-adaptation effects, respectively. The former proved fairly robust, with gait velocity changes reaching significant clinical changes above thresholds stated in the literature of ~0.1 m/s (Perera et al., 2006). What was also very interesting was post-adaptations remaining well-above baseline gait velocity. This significant observation duly ruled out the possibility of gait changes due to the haptic leash arising purely from a biomechanical hauling effect placed on the body. Measuring changes in gait velocity and the underlying spatiotemporal factors contributing to this change provided the evidence necessary to evaluate the same technique with an older chronic stroke population, adding a more in-depth analysis of kinematic and postural outcomes.

6.1.2 Study 2: Elder chronic stroke vs healthy controls

For the second and main phase of the project, both an elder stroke group and an age-matched control group were recruited to walk in a similar paradigm as that of study 1. The specific objectives of study 2 were aimed towards replicating study 1 findings, and more importantly, addressing the changes in postural control in response to the tensile force in elderly individuals, with and without stroke. There were significant differences in the spatiotemporal outcomes reported during force epochs and robust changes held over during post-force walking. These findings were then coupled with interesting postural changes in the mediolateral plane. Specifically, a tendency to shift COM occurred on the mediolateral plane away from the leash side and towards the paretic leg, or non-dominant leg for the control group. This potentially meant that the paretic side was engaged more than under normal walking conditions in the stroke group when gait velocity increased during the force and post-force epochs. This encouraging effect of mediolateral CoM shift was seen in a majority of stroke subjects. To what extent this would have on coordination and how it changes with respect to another walking aid, such as a cane, was subject to further investigation in study 3. Changes in coordination, especially in contrast to cane walking, could provide insight into the quality of gait that is achieved with the 'virtual leash' given the changed gait parameters. Further, any changes seen are conducive to functional walking would likely encourage further development of the haptic force and potential transfer to clinical application.

6.1.3 Study 3: Leash vs Cane

The last phase of the project was intended to explore the phenomena of spatiotemporal parameters, postural changes, and lower limb coordination by comparing the leash strategy with cane walking. Since both strategies have shown to increase gait velocity in chronic stroke (Allet et al., 2009; Rondeau et al., 2010; Sorrento et al., 2018), the significance of this study was to further advance the evaluation of the haptic leash strategy by measuring lower limb coordination during force and post-force walking relative to baseline, and of course, cane walking, in elderly individuals with and without stroke. While the objective was clearly not to suggest an alternative to the cane, the results did serve to illustrate the potentially advantageous benefits to being exposed to the 'virtual leash'. For example, changes in spatiotemporal outcomes tended to be larger when walking with the virtual leash compared to the cane, thus being more conducive to greater adaptation effects. Post-force effects also remained comparable to cane levels suggesting a post-adaptation effect. The significance of this last point is that chronic stroke subjects, such as those who daily walk with a cane, walk with comparable gait parameters while walking unaided after just 60 seconds of force adaptation. Additionally, the effect of mediolateral shifting of CoM found previously, was accompanied by increases in dorsiflexion and hip flexion of the paretic limb also featured in the 'virtual leash' strategy but was largely absent with the cane. The study concluded that spatiotemporal changes were more robust with the 'virtual leash' compared to the instrumented cane, and while in achieving these changes, both CoM and paretic limb coordination that are seemingly favorable to functional and symmetrical gait.

6.2 Findings related to the haptic forces and walking adaptation

6.2.1 The haptic leash as a somatosensory stimulus

A central premise of the thesis is that somatosensory stimuli have a positive effect on the walking abilities of stroke and non-stroke individuals, alike. The significance of a haptic input as an anchoring point to maintain dynamic stability has been investigated in the literature and in this project. This anchoring point can be either fixed or non-fixed (Eliane Mauerberg-deCastro, 2004; E. Mauerberg-deCastro & Moraes, 2014) haptic input and has been outlined in the literature (Oates et al., 2017) for orienting oneself in space has been shown to help stabilize both static (J. A. Barela et al., 1999) and dynamic posture in healthy (Costa et al., 2015) and stroke (S. Lee et al., 2015). Achieving a more stable gait by means of an anchoring point, such as a cane for instance, often leads to improved gait parameters conducive to improving functional mobility in a community setting. Thus, studies have used somatosensory inputs through various means, including stable earth-fixed surfaces, such as fixed haptic strips (Oates et al., 2008) and have evolved to more active non-fixed forms of haptic stimuli such as the cane (Albertsen et al., 2010; Perez & Fung, 2011) and the rehabilitation guide dog (Rondeau et al., 2010). This study adds a new and important component to previous research. For the first time to our knowledge, the application of forward-leading external forces in the form of a leash were applied to the body to elicit gait changes.

6.2.2 The role of VE with the haptic leash

We know that the use of VE when processing sensory stimuli can facilitate motor recovery (Adamovich et al., 2009; Bermúdez i Badia S., 2016; Fluet & Deutsch, 2013). The advantages of the VE alone have been shown to improve clinical gait and posture outcomes (Shema et al., 2014). Specifically, when creating a mixed-reality system coupling somatosensory information with a VE, such as in this project, researchers can create an effective lifelike and safe task or environment. This project along with previous works found that the coupling of sensory modalities such as haptics and VE provided the necessary challenge that can be ultimately conducive to motor recovery (Sveistrup, 2004). An example of this was the change in joint range and force of concur found in study 3, which was previously reporting in the literature (Jack et al., 2001).

Expanding further, we recall goal-oriented functional walking encompasses individual (i.e. physical and cognitive), task and environmental aspects of the locomotor task (Lamontagne et al., 2014). This requires both proactive and reactive capabilities, to anticipate the dynamic environment and to use the appropriate postural responses to maintain stability and avoid falling (Lamontagne & Fung, 2004), respectively. With this in mind, the 10, 15, and 20N forces were intended to closely simulate the experience of walking with a dog providing tension on a leash. The 10 and 20N forces were decided a priori with this simulation effect in mind. Then based on the similar results of 10 and 20N conditions in study 1, a decision was made to reduce the 20N force to 15N for the post-stroke and elderly adults.

Achieving this stability requires adapting to a changing environment. The use of VE was the medium we presented to stroke individuals to recreate such dynamic environment, and it was central to this thesis to establish whether haptic forces can facilitate these adaptive changes. Indeed, we established walking with the haptic leash in VE did change result in adaptations. This was evidenced by changes in gait spatiotemporal, postural and coordination gait parameters.

6.2.3 The effects of the haptic leash on spatiotemporal parameters

Along with increased gait velocity in all three studies were also increases in bilateral stride length mirroring other findings reported in the literature (Shawnna L. Patterson et al., 2008). Interestingly, it was found in study 2 (fig. 4.2) and study 3 (fig. 5.3) that the lower functioning stroke group – those who walked slower than 0.8 m/s over ground, tended to take larger strides on the paretic side relative to the non-paretic side during the 15N force condition. If one were to examine the post-stroke gait cycle it would be apparent that a stroke individual would typically favor single limb bearing on the non-affected side. This compensation for the paretic side would allow for a comparatively longer paretic-sided step if the individual has the intent of walking at a comfortable pace. Conversely, the post-stroke individual would favor less the single limb support on the paretic side, leading to hasty steps generated on the non-paretic side. Additionally, authors reported that paretic leg propulsion was inversely related to the step length (Balasubramanian et al., 2007). In contrast, our third study demonstrated that angular velocity of the paretic leg as a function of propulsion increased relative to baseline, and even cane walking. While this is a promising finding, further studies should be conducted to confirm that this trend is conducive to training a more economical gait pattern.

6.2.4 The effect of the haptic leash on gait stability

The 'virtual leash' has also provided some indirect evidence of changes in stability measures such as decreased double limb support times and step width. For example, double limb support times did decrease significantly in all three studies during force and post-force epochs suggesting that while the perturbation was present, it did not serve to destabilize or interrupt the individual's stride rhythm. In fact, the results of this study support the finding of other studies suggesting that reductions in double limb support times are favorable to gait stability in stroke (Junho Kim, Oh, et al., 2015). Step width did show some evidence of slight decreases for the stroke and control groups in study 2. However, these changes were not (see table 2.1) shown to the same extent as what other studies reports in stroke populations (Bateni et al., 2004) , particularly those walking with a cane (Perez & Fung 2011), which reduced to the order of ~5mm relative to free walking. While it is uncertain as to why leash walking did not reduce step width to the same extent, this could be explained in part by the larger three-point (i.e. lower limbs and cane) BOS offered by the cane.

6.2.5 The effect of the haptic leash on posture and lower-limb coordination

The changes in the kinematic, postural and coordination outcomes in Studies 2 and 3 are important in the context of achieving the adaptations and post-adaptation effects described in the literature, as the broken escalator effect (Reynolds & Bronstein, 2003) or split-belt treadmill (Helm & Reisman, 2015) effects come to mind. These effects would suggest that the cortical mechanisms that drive gait patterns are modified, at least transiently, in order to cope with the altered environment, which for the purpose of this study was the exposure to haptic force. Such abilities for the stroke population to adapt their coordination was reported in other studies, such as Reisman et al. (2007) where stroke subjects adapted convergent stride velocities for paretic and non-paretic limbs. Just as in that study, both Studies 2 and 3 included participants who had either cortical and subcortical brain lesions and were indeed able to show evidence of altering their gait pattern with exposure to the haptic force.

6.2.6 Cortical centers possibly driving adaptations and post-adaptation effect

The ability to reshape transient motor commands, as reported in the literature, calls to task various cortical, subcortical and cerebellar areas. While the main focus of the current study was not to single out which brain centers are responsible for the adaptation and post-adaptation processes exhibited behaviorally on the treadmill, one could speculate on the influences of the haptic forces on the gait and postural changes seen. At the cortical level for instance, authors have highlighted that gait adaptation effects could occur in several areas to achieve an altered motor command that drives gait under the force condition (H. Y. Kim, Yang, et al., 2016; Sangani et al., 2015). Furthermore, the cortical processes that shape motor commands driving gait are wide spread. For instance, we know that parietal, supplementary motor area (SMA) and premotor areas help create, shape and alter gait commands. This becomes evident in the gait abnormalities present when lesions affect these cortical areas. The prefrontal cortex is also active when contextual factors are taken into account and is believed to help alter the motor commands powering gait and is also involved in the gait adaptation process.

In relation to this project, it has been found that both premotor and prefrontal areas are associated to gait speed changes (Suzuki et al., 2004), such as those found in all three studies. For example, techniques such as near-infrared spectrometry have been used to measure changes in oxygenated hemoglobin (HbO_2), and thus brain activation during walking paradigms (Sangani et al., 2015). Studies using this technique also confirm that the prefrontal cortex is more active during velocity change and adaptation to gait velocity changes, while less activated during steady-state walking. A greater understanding of how these processes can be optimized may be critical for turning transient post-adaptations into prolonged training effects. Given the relevance of the cortical centers mentioned previously, it would be informative from a clinical perspective, to measure the contribution of each through imaging techniques. Perhaps a large step in understanding the role of somatosensory information in the adaptation and post-adaptation phases would be to employ a similar experimental protocol to this project that focus on visual or auditory inputs and measure differences in brain activation of these sensory modalities. From a somatosensory perspective, understanding the extent of its involvement in the adaptation and post-adaptation gait processes may be essential for discerning the appropriate quantities of haptic forces required to drive changes.

6.2.7 Adaptation and competing theories

The adaptation and post-adaptation effects reported in all three studies concur with observations related to split-belt and haptic modality paradigms presented in the review of literature. It was proposed that such adaptations involve transiently reshaped motor commands that adhere to walking conditions. These newly shaped motor commands are held over as aftereffects until post-adaptation eventually ceases and returns to the original motor command. However, one should also consider alternative explanations as to how changes in measures, such as gait speed, can be explained by means other than the adaptation process. For example, one possible view is that cortical activation is not necessarily required to create a synergy for increased gait speed if a pulling force is present. for example, the human central pattern generators (CPGs) described in section 2.2.4, may explain the walking behaviour seen during force and post-force epochs. There are several points that suggest otherwise. First, the very presence of CPGs in humans is still debatable (Minassian et al., 2017). Second, while there is some evidence of adaptation (MacKay-Lyons, 2002), as a result of CPGs, that the possibility of aftereffects remains to be examined. Another possibility would be the dynamic effect of being exposed to the external tension in force condition. From a classic dynamic perspective and external force that promotes forward momentum and thus increases in gait speed during the force epoch would not necessarily carry over into the post-adaptation epoch. In fact, it is quite possible that any gait changes brought on by the leash may return to baseline level in the post-force epoch. Of course, we see that this is not the case with leash walking as post-adaptations resisted the force offset perturbations and lasted the duration of the post-force epoch.

6.2.8 The haptic leash and related studies

The effects of these haptic forces on gait indeed have shown to corroborate with many of the findings that have used somatosensory input. For example, the instrumented cane used in study 3 and in other studies (Perez & Fung, 2011) showed evidence of increased gait speed and decreases in stride length and width variability. What was perhaps even more unexpected was the difference in increase of gait velocity with the leash compared to the cane. For example, when comparing force condition walking with cane walking the average velocity increase for using the leash was greater than 0.05 m/s compared to the cane. What was also interesting was once the force was removed, both stroke and healthy individuals walked at a pace that was similar to cane walking. From a clinical perspective, this is very encouraging since stroke individuals who normally used a cane in their daily lives showed the ability to walk at the same pace in a virtually independent state with a slack leash in hand, after exposure to the haptic forces. We also know among the stroke population, that walking with a cane improves other gait kinematics, showing more symmetrical spatial outcomes and a smoother COM mediolateral shift (Kuan et al., 1999), while from a dynamic perspective the use of a cane facilitates more symmetrical muscle activation synergies (Buurke et al., 2005). The use of the 'virtual leash' in the second and third studies, also provided some evidence of kinematic and postural changes and to a lesser extent dynamic (via angular acceleration) changes that would also promote greater symmetry. The effect of having increases in dorsiflexion and hip flexion, for instance, during force and post-force walking for the stroke group was also an interesting result. We do know that an exaggerated workload is typically placed on the paretic leg (Chen et al., 2005) as it often induces a hip-hiking strategy to achieve toe clearance of the plantar flexed toe. It would be interesting to investigate further whether these dynamic outcomes such as EMG could show favourable muscle synergies upon exposure to the leash.

6.3 Project considerations and limitations

While encouraging results have emerged from the three-part project, the process was not exempt from limitations. First, it should be taken into account that stroke individuals recruited in this study suffered from lesions in different areas of the brain. Given the diversity of the brain lesions, each lesion location potentially compromises the ability to functionally walk and ultimately to alter their walking pattern. Several measures have been taken, within the reasonable limits of the study to admit a more 'uniform' or homogeneous post-stroke population in order to render more specific results. This meant excluding cerebellar strokes, as it has been debated in the literature that cerebellar lesions, particularly those in the midline cerebellum have adverse effects to motor learning and thus adaptation (Bastian, 2011). This could potentially affect the behavioral performance of gait in terms of being able to adapt to the haptic stimulus. Another point to consider was the inclusion of subjects who sustained a single stroke was eventually opened to subjects who had multiple strokes. The justification for this inclusion may be debatable since there is little evidence to suggest multiple stroke lesions may alter motor learning (Vidoni et al., 2008) and thus the ability to adapt and post-adapt. This could be noted as a potential selection bias. While the vast majority of subjects where single stroke survivors, our results suggested that the two stroke subjects that sustained more than one stroke did not show notable changes in walking outcomes with respect to the other single stroke subjects. Additional analysis deemed that having multiple lesions did not prove detrimental to adaptation and post-adaptation effects.

Taking into account rather rigorous inclusion and exclusion criteria, the minimal target sample size number of 14 per group was determined a priori for post-stroke and healthy elderly groups in studies 2 and 3. This number was determined based on achieving a moderate effect size of 0.7 with alpha and beta levels set at $p < 0.05$ and 0.8, respectively. Thus, caution should be taken with when generalizing the results with external validity. It should also be noted that further stratification of seven higher and seven lower functioning participants was made in study 3. The 0.8 m/s community dwelling benchmark was used for stratification given the high biological variability of the population. This was especially the case for study 3 involving comparisons between leash and cane walking. Most individuals in the lower functioning strata did indeed use a cane for daily walking, while most of the higher functioning individuals did not. Despite the reduction in sample sizes imposed by this process, both statistically and clinically significant spatiotemporal results were attained for stroke and control groups in both studies 2 and 3, supporting the main hypothesis of the thesis.

Another limitation was walking endurance and its effect on gait performance. Since we know that walking endurance is restricted in the stroke population, the experimental protocol had to take this into account. To address this, a maximal of time possible for adaptation and post-adaptation epochs to occur was allocated before the effects of fatigue began to influence walking performance. Quantifying the potential effect of fatigue within gait parameters, particularly towards the later stages of the protocol, would not be feasible by any reliable measure available to the study. While this should be considered a form of procedural bias as subjects potentially exerted beyond normal walking effort, steps were taken to reduce the effect of fatigue. This included resting periods between trials and possible return dates between treadmill habituation and data recording sessions. This last point of treadmill habituation should also be addressed. The fact that some stroke individuals required longer habituation sessions on the self-paced treadmill could potentially impose a learning effect prior to data recording. To avoid this, exposure to treadmill was limited only to habituating to natural steady-state walking and not to training in an experimental protocol.

The endurance restrictions also prevented the study from investigating to what extent adaptations effects be achieved in the event of a prolonged force epoch. In other words, it remains unclear if changes in gait parameters reached a plateau within the 60 second force condition period, or if changes may still arise if the post-adaptation epoch was prolonged. Maybe even more crucial in a clinical context is understanding how long the post-force effects seen would eventually last since they largely remained at the end of the post-force epoch. Perhaps prolonging this epoch or employing multiple training sessions may address this limitation inherent to the project protocol.

6.4 Future directions and possible impact

Despite the limitations mentioned, the results largely replicated across the three studies build a very promising picture with regard to using haptic forward-leading forces to the hand. From here, there are many exciting directions that this strategy can take to further develop, both in terms of research and clinical practice. It would be critical to measure how well these results continue over the span of a training protocol and to what extent this strategy can translate to overground walking. The protocol may be as simple as having subjects return for multiple training sessions and measure whether adaptation and post-adaptation effects become more robust and if these effects translate to changes in everyday walking with post training measures. From a clinical standpoint, this may mean prolonging the post-adaptation epoch to see if more abiding changes are seen over time. The prospects of an efficacious result are promising, considering the results seen with a one-time participation used in this protocol. Furthermore, with the development and increasing commercialization of VR and robotic technologies, providing a similar mixed reality set-up including forward-leading forces is a rather feasible study to recreate.

The use of constant forces does not have to be confined of VE’s. A strong case can also be made for the feasibility of using such a strategy in the clinical setting. For example, simply applying a tensile force via a tensor band or towel while leading the stroke patient forward could potentially replicate to a large extent what was seen in this study. What can make such a strategy effective is that if applied carefully by a clinician, it can provide a cost and time-effective means of providing intensive gait therapies to both chronic and post-acute stroke individuals. If we recall the strategy is composed of both force and post-force phases, hence exposing the patient to the possibility of adaptation and post-adaptation effects in the clinic can help address both the physical and neurorehabilitation aspects, which were touched on in this thesis. However, further study would be necessary in order to determine the efficacy and transferability of the haptic force method through other mediums (e.g. tensor bands) typically found in the clinical setting. In addition, the strategy is also scalable, meaning the force can be modified (i.e. increased or decreased) according to the needs of the patient. While the 10 to 15N force range was sufficient for the post-stroke and elder healthy in this study, a further understanding of different force level effects would serve to prescribe the appropriate force level based on the individual's functional capacity. Clinically, this could ensure safe and incremented increases in haptic force designed to constantly challenge dynamic stability.

6.4 Summary of findings and conclusion

To summarize, the hypothesis proposed in this thesis postulated that haptic forward-leading forces delivered to the hand would change walking outcomes, both before and after exposure, was supported based on the evidence of three studies. In the first study, we established that young healthy individuals are capable of changing gait during steady-state walking when holding on to a haptic forward leading force delivered by a mechanical leash in a virtual environment. Gait velocity increased during the force epoch and post-force epoch, providing evidence of both adaptation and post-adaptation effects. This was accompanied by stride time increases and double limb support decreases. The very same findings were replicated in the main study involving post-stroke and age-matched older individuals. Additionally, stride time decreased.

Mediolateral stride width was marginally decreased without establishing significant effects. Moreover, a trend occurred with COM tending to shift mediolaterally to the paretic side during both force and post-force epochs as the gait velocity increased. The last study showed that the spatiotemporal and postural changes rendered by the leash allowed for greater dorsiflexion and hip extension of the paretic side during force and post-force epochs, compared to walking with the cane. A greater angular momentum of the lower paretic limb was also discovered, suggesting an increase in force generated with the affected leg during force and post-force conditions. The overall picture has the 'virtual leash' improving spatiotemporal outcomes while engaging the paretic side owing to a shift in the COM and a greater ankle and hip flexion.

To conclude, the prospect of both investigating and developing research protocols and clinical strategies based on haptic forward-leading forces shows a fairly optimistic outlook. Development of this technique could and should be employed in both spheres as it seems to be an efficient way to address the underpinning motor aspects of rehabilitation at the level of walking and posture in neurological and healthy aging populations. More work is warranted to develop this exciting strategy further and hopefully contribute to improving patient locomotion, and quality of life as a result.

REFRENCES

Abbud, G., Janelle, C., & Vocos, M. (2014). The Use of a Trained Dog as a Gait Aid for Clients with Ataxia: A Case Report. *Physiotherapy Canada, 66*(1).

Adamovich, S. V., Fluet, G. G., Tunik, E., & Merians, A. S. (2009). Sensorimotor training in virtual reality: a review. *NeuroRehabilitation, 25*(1), 29-44. doi:10.3233/nre-2009-0497

Afzal, M. R., Byun, H.-Y., Oh, M.-K., & Yoon, J. (2015). Effects of kinesthetic haptic feedback on standing stability of young healthy subjects and stroke patients. *Journal of NeuroEngineering and Rehabilitation, 12*(1), 20. doi:10.1186/s12984-015-0020-x

Albertsen, I. M., Temprado, J. J., & Berton, E. (2010). Effect of haptic supplementation on postural stabilization: A comparison of fixed and mobile support conditions. *Human Movement Science, 29*(6), 999-1010. doi:10.1016/j.humov.2010.07.013

Alexander, L. D., Black, S. E., Patterson, K. K., Gao, F., Danells, C. J., & McIlroy, W. E. (2009). Association Between Gait Asymmetry and Brain Lesion Location in Stroke Patients. *Stroke, 40*(2), 537-544. doi:10.1161/STROKEAHA.108.527374

Alexander LD, B. S., Patterson KK, Gao F, Danells CJ, McIlroy WE. (2009). Association between gait asymmetry and brain lesion location in stroke patients. *Stroke, 40*(2), 537-544.

Allen, J. L., Kautz, S. A., & Neptune, R. R. (2011). Step length asymmetry is representative of compensatory mechanisms used in post-stroke hemiparetic walking. *Gait & Posture, 33*(4), 538-543. doi:10.1016/j.gaitpost.2011.01.004

Allet, L., Leemann, B., Guyen, E., Murphy, L., Monnin, D., Herrmann, F. R., & Schnider, A. (2009). Effect of different walking aids on walking capacity of patients with poststroke hemiparesis. *Arch Phys Med Rehabil, 90*(8), 1408-1413. doi:10.1016/j.apmr.2009.02.010

Andersson, A. G., Kamwendo, K., & Appelros, P. (2008). Fear of falling in stroke patients: relationship with previous falls and functional characteristics. *Int J Rehabil Res, 31*(3), 261-264. doi:10.1097/MRR.0b013e3282fba390

Andersson, P., & Franzen, E. (2015). Effects of weight-shift training on walking ability, ambulation, and weight distribution in individuals with chronic stroke: a pilot study. *Top Stroke Rehabil, 22*(6), 437-443. doi:10.1179/1074935715Z.00000000052

Awad, L. N., Binder-Macleod, S. A., Pohlig, R. T., & Reisman, D. S. (2015). Paretic Propulsion and Trailing Limb Angle Are Key Determinants of Long-Distance Walking Function

After Stroke. *Neurorehabil Neural Repair*, *29*(6), 499-508. doi:10.1177/1545968314554625

Baer, G., & Smith, M. (2001). The recovery of walking ability and subclassification of stroke. *Physiother Res Int*, *6*(3), 135-144.

Balaban, B., & Tok, F. (2014). Gait disturbances in patients with stroke. *PM R, 6*(7), 635-642. doi:10.1016/j.pmrj.2013.12.017

Balasubramanian, C. K., Bowden, M. G., Neptune, R. R., & Kautz, S. A. (2007). Relationship Between Step Length Asymmetry and Walking Performance in Subjects With Chronic Hemiparesis. *Archives of Physical Medicine and Rehabilitation, 88*(1), 43-49. doi:10.1016/j.apmr.2006.10.004

Bancroft, M. J., & Day, B. L. (2016). The Throw-and-Catch Model of Human Gait: Evidence from Coupling of Pre-Step Postural Activity and Step Location. *Front Hum Neurosci*, *10*, 635. doi:10.3389/fnhum.2016.00635

Barclay, R., Ripat, J., & Mayo, N. (2015). Factors describing community ambulation after stroke: a mixed-methods study. *Clin Rehabil, 29*(5), 509-521. doi:10.1177/0269215514546769

Barela, A., Caporicci, S., & Barela, J. (2017). Locomotion and Posture in Older Adults. *springer*, 209-218. doi:10.1007/978-3-319-48980-3_14

Barela, J. A., Jeka, J. J., & Clark, J. E. (1999). The use of somatosensory information during the acquisition of independent upright stance. *Infant Behavior and Development*.

Barela, J. A., Jeka, J.J., Clark, J.E., & more, S. (1999). The use of somatosensory information during the acquisition of independent upright stance. *Infant Behavior and Development, 22*(1), 87-102.

Barthelemy, D., Grey, M. J., Nielsen, J. B., & Bouyer, L. (2011). Involvement of the corticospinal tract in the control of human gait. *Prog Brain Res, 192*, 181-197. doi:10.1016/B978-0-444-53355-5.00012-9

Bastian, A. J. (2008). Understanding sensorimotor adaptation and learning for rehabilitation. *Curr Opin Neurol, 21*(6), 628-633. doi:10.1097/WCO.0b013e328315a293

Bastian, A. J. (2011). Moving, sensing and learning with cerebellar damage. *Current Opinion in Neurobiology, 21*(4), 596-601. doi:10.1016/j.conb.2011.06.007

Batchelor, F., Hill, K., Mackintosh, S., & Said, C. (2010). What works in falls prevention after stroke?: a systematic review and meta-analysis. *Stroke, 41*(8), 1715-1722. doi:10.1161/STROKEAHA.109.570390

Batchelor, F. A., Mackintosh, S. F., Said, C. M., & Hill, K. D. (2012). Falls after stroke. *Int J Stroke, 7*(6), 482-490. doi:10.1111/j.1747-4949.2012.00796.x

Bateni, H., Heung, E., Zettel, J., McLlroy, W. E., & Maki, B. E. (2004). Can use of walkers or canes impede lateral compensatory stepping movements? *Gait & Posture*.

Bateni, H., & Maki, B. E. (2005). Assistive devices for balance and mobility: benefits, demands, and adverse consequences. *Arch Phys Med Rehabil, 86*(1), 134-145.

Bayat, R., Barbeau, H., & Lamontagne, A. (2005). Speed and Temporal-Distance Adaptations during Treadmill and Overground Walking Following Stroke. *Neurorehabilitation and Neural Repair, 19*(2), 115-124. doi:10.1177/1545968305275286

Belda-Lois, J. M., Mena-del Horno, S., Bermejo-Bosch, I., Moreno, J. C., Pons, J. L., Farina, D., . . . Rea, M. (2011). Rehabilitation of gait after stroke: a review towards a top-down approach. *J Neuroeng Rehabil, 8*, 66. doi:10.1186/1743-0003-8-66

Bermúdez i Badia S., F. G. G., Llorens R., Deutsch J.E. . (2016). Virtual Reality for Sensorimotor Rehabilitation Post Stroke: Design Principles and Evidence *Virtual Reality for Sensorimotor Rehabilitation Post Stroke: Design Principles and Evidence* (pp. pp 573-603): Springer, Cham.

Bhatt, T., Wang, T. Y., Yang, F., & Pai, Y. C. (2013). Adaptation and generalization to opposing perturbations in walking. *Neuroscience, 246*, 435-450. doi:10.1016/j.neuroscience.2013.04.013

Bianchi, L., Angelini, D., Orani, G. P., & Lacquaniti, F. (1998). Kinematic coordination in human gait: relation to mechanical energy cost. *Journal of Neurophysiology, 79*(4), 2155-2170.

Bingenheimer, K., Temprado, J., Harnagea, M., Bricot, N., Villani, P., & Berton, E. (2015). Effects of a light touch on fixed or mobile supports on gait parameters in visually restricted young adults. *Neuroscience Letters, 589*, 176-180. doi:10.1016/j.neulet.2015.01.054

Bogousslavsky, J., Van Melle, G., & Regli, F. (1988). The Lausanne Stroke Registry: analysis of 1,000 consecutive patients with first stroke. *Stroke, 19*(9), 1083-1092.

Bohannon, R. W. (2009). Measurement of gait speed of older adults is feasible and informative in a home-care setting. *J Geriatr Phys Ther, 32*(1), 22-23.

Bohannon, R. W., & Andrews, A. W. (1998). Relationships between impairments in strength of limb muscle actions following stroke. *Perceptual and Motor Skills, 87*(3), 1327-1330.

Bohannon, R. W., Andrews, W. A., & Glenney, S. S. (2013). Minimal Clinically Important Difference for Comfortable Speed as a Measure of Gait Performance in Patients Undergoing Inpatient Rehabilitation after Stroke. *Journal of Physical Therapy Science, 25*(10), 1223-1225. doi:10.1589/jpts.25.1223

Bohannon, R. W., & Glenney, S. S. (2014). Minimal clinically important difference for change in comfortable gait speed of adults with pathology: a systematic review. *Journal of Evaluation in Clinical Practice, 20*(4), 295-300. doi:10.1111/jep.12158

Bohannon, R. W., Horton, M. G., & Wikholm, J. B. (1991). Importance of four variables of walking to patients with stroke. *Int J Rehabil Res, 14*(3), 246-250.

Bolton, D. A. E., McIlroy, W. E., & Staines, R. W. (2011). The impact of light fingertip touch on haptic cortical processing during a standing balance task. *Experimental Brain Research, 212*(2), 279-291.

Bongers, K. T., Schoon, Y., Graauwmans, M. J., Schers, H. J., Melis, R. J., & Olde Rikkert, M. G. (2015). The predictive value of gait speed and maximum step length for falling in community-dwelling older persons. *Age Ageing, 44*(2), 294-299. doi:10.1093/ageing/afu151

Boonsinsukh, R., Panichareon, L., Saengsirisuwan, V., & Phansuwan-Pujito, P. (2011). Clinical identification for the use of light touch cues with a cane in gait rehabilitation poststroke. *Topics in stroke rehabilitation, 18*(sup1), Pages 633-642.

Bowden, M. G., Balasubramanian, C. K., Neptune, R. R., & Kautz, S. A. (2006). Anterior-Posterior Ground Reaction Forces as a Measure of Paretic Leg Contribution in Hemiparetic Walking. *Stroke, 37*(3), 872-876. doi:10.1161/01.STR.0000204063.75779.8d

Brach, J. S., Perera, S., Studenski, S., Katz, M., Hall, C., & Verghese, J. (2010). Meaningful change in measures of gait variability in older adults. *Gait Posture, 31*(2), 175-179. doi:10.1016/j.gaitpost.2009.10.002

Bronstein, A. M., Bunday, K. L., & Reynolds, R. (2009). What the "broken escalator" phenomenon teaches us about balance. *Ann N Y Acad Sci, 1164*, 82-88. doi:10.1111/j.1749-6632.2009.03870.x

Bruijn, S. M., Impe, A., Duysens, J., & Swinnen, S. P. (2012). Split-belt walking: adaptation differences between young and older adults. *Journal of Neurophysiology, 108*(4), 1149-1157. doi:10.1152/jn.00018.2012

Bugnariu, N., & Fung, J. (2007). Aging and selective sensorimotor strategies in the regulation of upright balance. *J Neuroeng Rehabil, 4*, 19. doi:10.1186/1743-0003-4-19

Bulea, T. C., Kim, J., Damiano, D. L., Stanley, C. J., & Park, H. S. (2015). Prefrontal, posterior parietal and sensorimotor network activity underlying speed control during walking. *Front Hum Neurosci*, *9*, 247. doi:10.3389/fnhum.2015.00247

Bunday, K. L., & Bronstein, A. M. (2009). Locomotor Adaptation and Aftereffects in Patients With Reduced Somatosensory Input Due to Peripheral Neuropathy. *Journal of Neurophysiology, 102*(6), 3119-3128. doi:10.1152/jn.00304.2009

Buurke, J. H., Hermens, H. J., Erren-Wolters, C. V., & Nene, A. V. (2005). The effect of walking aids on muscle activation patterns during walking in stroke patients. *Gait & Posture, 22*(2), 164-170. doi:10.1016/j.gaitpost.2004.09.003

Cajigas, I., Koenig, A., Severini, G., Smith, M., & Bonato, P. (2017). Robot-induced perturbations of human walking reveal a selective generation of motor adaptation. *Sci. Robot., 2*, 1-9.

Callisaya, M. L., Blizzard, L., Schmidt, M. D., McGinley, J. L., & Srikanth, V. K. (2010). Ageing and gait variability--a population-based study of older people. *Age Ageing*, *39*(2), 191-197. doi:10.1093/ageing/afp250

Cameron, J. I., O'Connell, C., Foley, N., Salter, K., Booth, R., Boyle, R., . . . Stroke Foundation Canadian Stroke Best Practice, C. (2016). Canadian Stroke Best Practice Recommendations: Managing transitions of care following Stroke, Guidelines Update 2016. *Int J Stroke*, *11*(7), 807-822. doi:10.1177/1747493016660102

Capaday, C. (2002). The special nature of human walking and its neural control. *Trends Neurosci*, *25*(7), 370-376.

Chamberlin, M. E., Fulwider, B. D., Sanders, S. L., & Medeiros, J. M. (2005). Does Fear of Falling Influence Spatial and Temporal Gait Parameters in Elderly Persons Beyond Changes Associated With Normal Aging? *The Journals of Gerontology: Series A, 60*(9), 1163-1167. doi:10.1093/gerona/60.9.1163

Chen, G., Patten, C., Kothari, D. H., & Zajac, F. E. (2005). Gait differences between individuals with post-stroke hemiparesis and non-disabled controls at matched speeds. *Gait & Posture, 22*(1). doi:10.1016/j.gaitpost.2004.06.009

Cheng, P. T., Wu, S. H., Liaw, M. Y., Wong, A. M., & Tang, F. T. (2001). Symmetrical body-weight distribution training in stroke patients and its effect on fall prevention. *Arch Phys Med Rehabil*, *82*(12), 1650-1654. doi:10.1053/apmr.2001.26256

Choi, J. T., & Bastian, A. J. (2007). Adaptation reveals independent control networks for human walking. *Nat Neurosci, 10*(8), 1055-1062. doi:10.1038/nn1930

Chvatal, S. A., & Ting, L. H. (2012). Voluntary and reactive recruitment of locomotor muscle synergies during perturbed walking. *J Neurosci, 32*(35), 12237-12250. doi:10.1523/JNEUROSCI.6344-11.2012

Clapp, S., & Wing, A. M. (1999). Light touch contribution to balance in normal bipedal stance. *Exp Brain Res, 125*(4), 521-524.

Costa, A. A., Manciopi, P. A., Mauerberg-deCastro, E., & Moraes, R. (2015). Haptic information provided by the "anchor system" reduces trunk sway acceleration in the frontal plane during tandem walking in older adults. *Neurosci Lett, 609*, 1-6. doi:10.1016/j.neulet.2015.10.004

Craig LE, W. O., Bernhardt J, Langhorne P. (2011). Predictors of poststroke mobility: systematic review. *International Journal of Stroke., 6*(4), 321-327.

Cromwell, R. L., & Newton, R. A. (2004). Relationship between balance and gait stability in healthy older adults. *Journal of Aging and Physical Activity, 12*(1), 90-100. doi:10.1123/japa.12.1.90

Daly, J. J., Sng, K., Roenigk, K., Fredrickson, E., & Dohring, M. (2007). Intra-limb coordination deficit in stroke survivors and response to treatment. *Gait Posture, 25*(3), 412-418. doi:10.1016/j.gaitpost.2006.05.007

Davenport, R. J., Dennis, Wellwood, I., & Warlow, C. P. (1996). Complications After Acute Stroke. *Stroke, 27*(3), 415-420. doi:10.1161/01.STR.27.3.415

de Oliveira, C. B., de Medeiros, I. R., Frota, N. A., Greters, M. E., & Conforto, A. B. (2008). Balance control in hemiparetic stroke patients: main tools for evaluation. *J Rehabil Res Dev, 45*(8), 1215-1226.

Dean, C. M., Richards, C. L., & Malouin, F. (2001). Walking speed over 10 metres overestimates locomotor capacity after stroke. *Clin Rehabil, 15*(4), 415-421. doi:10.1191/026921501678310216

Della Croce, U., Riley, P. O., Lelas, J. L., & Kerrigan, D. C. (2001). A refined view of the determinants of gait. *Gait Posture, 14*(2), 79-84.

Deutsch, J. E., Merians, A. S., Adamovich, S., Poizner, H., & Burdea, G. C. (2004). Development and application of virtual reality technology to improve hand use and gait of individuals post-stroke. *Restor Neurol Neurosci, 22*(3-5), 371-386.

Deutsch, J. E., & Mirelman, A. (2007). Virtual reality-based approaches to enable walking for people poststroke. *Top Stroke Rehabil, 14*(6), 45-53. doi:10.1310/tsr1406-45

Di Carlo, A. (2009). Human and economic burden of stroke. *Age Ageing, 38*(1), 4-5. doi:10.1093/ageing/afn282

Dickstein, R., & Laufer, Y. (2004). Light touch and center of mass stability during treadmill locomotion. *Gait Posture, 20*(1), 41-47. doi:10.1016/S0966-6362(03)00091-2

Divani, A. A., Vazquez, G., Barrett, A. M., Asadollahi, M., & Luft, A. R. (2009). Risk factors associated with injury attributable to falling among elderly population with history of stroke. *Stroke, 40*(10), 3286-3292. doi:10.1161/STROKEAHA.109.559195

do Carmo, A. A., Kleiner, A. F., & Barros, R. M. (2015). Alteration in the center of mass trajectory of patients after stroke. *Top Stroke Rehabil, 22*(5), 349-356. doi:10.1179/1074935714Z.0000000037

Dobkin, B. H. (2004). Strategies for stroke rehabilitation. *Lancet Neurol, 3*(9), 528-536. doi:10.1016/S1474-4422(04)00851-8

Duncan, P. W., Zorowitz, R., Bates, B., Choi, J. Y., Glasberg, J. J., Graham, G. D., . . . Reker, D. (2005). Management of Adult Stroke Rehabilitation Care: a clinical practice guideline. *Stroke, 36*(9), e100-143. doi:10.1161/01.STR.0000180861.54180.FF

Eng, J. J., & Tang, P. F. (2007). Gait training strategies to optimize walking ability in people with stroke: a synthesis of the evidence. *Expert Rev Neurother, 7*(10), 1417-1436. doi:10.1586/14737175.7.10.1417

Eng, J. J., & Winter, D. A. (1995). Kinetic analysis of the lower limbs during walking: what information can be gained from a three-dimensional model? *J Biomech, 28*(6), 753-758.

Farris, D., Hampton, A., Lewek, M. D., & Sawicki, G. S. (2015). Revisiting the mechanics and energetics of walking in individuals with chronic hemiparesis following stroke: from individual limbs to lower limb joints. *Journal of NeuroEngineering and Rehabilitation, 12*(1), 1-12. doi:10.1186/s12984-015-0012-x

Finley, J. M., Statton, M. A., & Bastian, A. J. (2014). A novel optic flow pattern speeds split-belt locomotor adaptation. *Journal of Neurophysiology, 111*(5), 969-976. doi:10.1152/jn.00513.2013

Flansbjer, U. B., Holmback, A. M., Downham, D., Patten, C., & Lexell, J. (2005). Reliability of gait performance tests in men and women with hemiparesis after stroke. *J Rehabil Med, 37*(2), 75-82. doi:10.1080/16501970410017215

Fluet, G. G., & Deutsch, J. E. (2013). Virtual Reality for Sensorimotor Rehabilitation Post-Stroke: The Promise and Current State of the Field. *Curr Phys Med Rehabil Rep, 1*(1), 9-20. doi:10.1007/s40141-013-0005-2

Foundation, C. H. a. S. (2011). Tracking heart disease and stroke in Canada. Retrieved from http://www.phac-aspc.gc.ca/cd-mc/cvd-mcv/sh-fs-2011/pdf/StrokeHighlights_EN.pdf

Franzen, E., Gurfinkel, V. S., Wright, W. G., Cordo, P. J., & Horak, F. B. (2011). Haptic touch reduces sway by increasing axial tone. *Neuroscience, 174*, 216-223. doi:10.1016/j.neuroscience.2010.11.017

Freyler, K., Gollhofer, A., Colin, R., Brüderlin, U., & Ritzmann, R. (2015). Reactive Balance Control in Response to Perturbation in Unilateral Stance: Interaction Effects of Direction, Displacement and Velocity on Compensatory Neuromuscular and Kinematic Responses. *PLOS ONE, 10*(12). doi:10.1371/journal.pone.0144529

Fulk, G. D., & Echternach, J. L. (2008). Test-retest reliability and minimal detectable change of gait speed in individuals undergoing rehabilitation after stroke. *J Neurol Phys Ther, 32*(1), 8-13. doi:10.1097/NPT0b013e31816593c0

Fulk, G. D., He, Y., Boyne, P., & Dunning, K. (2017). Predicting Home and Community Walking Activity Poststroke. *Stroke, 48*(2), 406-411. doi:10.1161/STROKEAHA.116.015309

Fulk, G. D., Reynolds, C., Mondal, S., & Deutsch, J. E. (2010). Predicting home and community walking activity in people with stroke. *Arch Phys Med Rehabil, 91*(10), 1582-1586. doi:10.1016/j.apmr.2010.07.005

Fung, J., Boonsinsukh, R., & Rapagna, M. (2003). *Postural responses triggered by surface perturbations are task-specific and goal directed*.

Fung, J., & Perez, C. F. (2011). Sensorimotor enhancement with a mixed reality system for balance and mobility rehabilitation. *Conf Proc IEEE Eng Med Biol Soc, 2011*, 6753-6757. doi:10.1109/IEMBS.2011.6091666

Fung, J., Richards, C. L., Malouin, F., McFadyen, B. J., & Lamontagne, A. (2006). A treadmill and motion coupled virtual reality system for gait training post-stroke. *Cyberpsychol Behav, 9*(2), 157-162. doi:10.1089/cpb.2006.9.157

Gage, W. H., Winter, D. A., Frank, J. S., & Adkin, A. L. (2004). Kinematic and kinetic validity of the inverted pendulum model in quiet standing. *Gait Posture, 19*(2), 124-132. doi:10.1016/S0966-6362(03)00037-7

Gard, S. A., & Childress, D. S. (2001). What Determines the Vertical Displacement of the Body During Normal Walking? *JPO Journal of Prosthetics & Orthotics, 13*(3), 64-67.

Gates, D. H., Darter, B. J., Dingwell, J. B., & Wilken, J. M. (2012). Comparison of walking overground and in a Computer Assisted Rehabilitation Environment (CAREN) in individuals with and without transtibial amputation. *J Neuroeng Rehabil, 9*, 81. doi:10.1186/1743-0003-9-81

Goldie, P. A., Matyas, T. A., & Evans, O. M. (2001). Gait after stroke: Initial deficit and changes in temporal patterns for each gait phase. *Archives of Physical Medicine and Rehabilitation, 82*(8), 1057-1065. doi:10.1053/apmr.2001.25085

Goldstein, L. B., Adams, R., Alberts, M. J., Appel, L. J., Brass, L. M., Bushnell, C. D., . . . American Academy of, N. (2006). Primary prevention of ischemic stroke: a guideline from the American Heart Association/American Stroke Association Stroke Council: cosponsored by the Atherosclerotic Peripheral Vascular Disease Interdisciplinary Working Group; Cardiovascular Nursing Council; Clinical Cardiology Council; Nutrition, Physical Activity, and Metabolism Council; and the Quality of Care and Outcomes Research Interdisciplinary Working Group: the American Academy of Neurology affirms the value of this guideline. *Stroke, 37*(6), 1583-1633. doi:10.1161/01.STR.0000223048.70103.F1

Gordon, K. E., Ferris, D. P., & Kuo, A. D. (2009). Metabolic and mechanical energy costs of reducing vertical center of mass movement during gait. *Arch Phys Med Rehabil, 90*(1), 136-144. doi:10.1016/j.apmr.2008.07.014

Gowland, C., Stratford, P., Ward, M., Moreland, J., Torresin, W., Van Hullenaar, S., . . . Plews, N. (1993). Measuring physical impairment and disability with the Chedoke-McMaster Stroke Assessment. *Stroke, 24*(1), 58-63.

Granger, C. V., Hamilton, B. B., & Fiedler, R. C. (1992). Discharge outcome after stroke rehabilitation. *Stroke, 23*(7), 978-982.

Greene, P. J., & Granat, M. H. (2000). The effects of knee and ankle flexion on ground clearance in paraplegic gait. *Clin Biomech (Bristol, Avon), 15*(7), 536-540.

Guertin, P. A. (2012). Central pattern generator for locomotion: anatomical, physiological, and pathophysiological considerations. *Front Neurol, 3*, 183. doi:10.3389/fneur.2012.00183

Guo, Y., Logan, H. L., Glueck, D. H., & Muller, K. E. (2013). Selecting a sample size for studies with repeated measures. *BMC Med Res Methodol, 13*, 100. doi:10.1186/1471-2288-13-100

Hak, L., Houdijk, H., van der Wurff, P., Prins, M. R., Beek, P. J., & van Dieën, J. H. (2015). Stride frequency and length adjustment in post-stroke individuals: influence on the margins of stability. *Journal of rehabilitation medicine, 47*(2), 126-132. doi:10.2340/16501977-1903

Haley, S. M., Coster, W. J., & Binda-Sundberg, K. (1994). Measuring physical disablement: the contextual challenge. *Phys Ther, 74*(5), 443-451.

Hallett, M. (2001). Plasticity of the human motor cortex and recovery from stroke. *Brain Res Brain Res Rev, 36*(2-3), 169-174.

Harcourt-Smith, W. E., & Aiello, L. C. (2004). Fossils, feet and the evolution of human bipedal locomotion. *J Anat, 204*(5), 403-416. doi:10.1111/j.0021-8782.2004.00296.x

Hars, M., Herrmann, F. R., & Trombetti, A. (2013). Reliability and minimal detectable change of gait variables in community-dwelling and hospitalized older fallers. *Gait & Posture, 38*(4), 1010-1014. doi:10.1016/j.gaitpost.2013.05.015

Hellstrom, K., Lindmark, B., & Fugl-Meyer, A. (2002). The Falls-Efficacy Scale, Swedish version: does it reflect clinically meaningful changes after stroke? *Disabil Rehabil, 24*(9), 471-481. doi:10.1080/09638280110105259

Helm, E. E., & Reisman, D. S. (2015). The Split-Belt Walking Paradigm: Exploring Motor Learning and Spatiotemporal Asymmetry Poststroke. *Phys Med Rehabil Clin N Am, 26*(4), 703-713. doi:10.1016/j.pmr.2015.06.010

Henderson, A., Korner-Bitensky, N., & Levin, M. (2007). Virtual reality in stroke rehabilitation: a systematic review of its effectiveness for upper limb motor recovery. *Top Stroke Rehabil, 14*(2), 52-61. doi:10.1310/tsr1402-52

Hesse, S. (2003). Rehabilitation of Gait After Stroke: Evaluation, Principles of Therapy, Novel Treatment Approaches, and Assistive Devices. *Topics in Geriatric Rehabilitation, 19*(2), 109. doi:10.1097/00013614-200304000-00005

Hof, A. L., Gazendam, M. G., & Sinke, W. E. (2005). The condition for dynamic stability. *J Biomech, 38*(1), 1-8. doi:10.1016/j.jbiomech.2004.03.025

Holden, M. K. (2005). Virtual environments for motor rehabilitation: review. *Cyberpsychol Behav, 8*(3), 187-211; discussion 212-189. doi:10.1089/cpb.2005.8.187

Hollman, J. H., Brey, R. H., Robb, R. A., Bang, T. J., & Kaufman, K. R. (2006). Spatiotemporal gait deviations in a virtual reality environment. *Gait Posture, 23*(4), 441-444. doi:10.1016/j.gaitpost.2005.05.005

Hong, E. (2015). Comparison of quality of life according to community walking in stroke patients. *J Phys Ther Sci, 27*(7), 2391-2393. doi:10.1589/jpts.27.2391

Hoogkamer, W., Bruijn, S. M., & Duysens, J. (2014). Stride length asymmetry in split-belt locomotion. *Gait & Posture, 39*(1), 652-654. doi:10.1016/j.gaitpost.2013.08.030

Hornyak, V., VanSwearingen, J. M., & Brach, J. S. (2012). Measurement of Gait Speed. *Topics in Geriatric Rehabilitation, 28*(1), 27-32. doi:10.1097/TGR.0b013e318233e75b

Hsiao, H., Gray, V. L., Creath, R. A., Binder-Macleod, S. A., & Rogers, M. W. (2017). Control of lateral weight transfer is associated with walking speed in individuals post-stroke. *Journal of Biomechanics*. doi:10.1016/j.jbiomech.2017.06.021

Hsu, A. L., Tang, P. F., & Jan, M. H. (2003). Analysis of impairments influencing gait velocity and asymmetry of hemiplegic patients after mild to moderate stroke. *Archives of Physical Medicine and Rehabilitation*.

Ivanenko, Y. P., Grasso, R., Macellari, V., & Lacquaniti, F. (2002). Two-thirds power law in human locomotion: role of ground contact forces. *NeuroReport, 13*(9), 1171. doi:10.1097/00001756-200207020-00020

Jack, D., Boian, R., Merians, A. S., Tremaine, M., Burdea, G. C., Adamovich, S. V., . . . Poizner, H. (2001). Virtual reality-enhanced stroke rehabilitation. *IEEE Trans Neural Syst Rehabil Eng, 9*(3), 308-318. doi:10.1109/7333.948460

Jeka, J. J. (1997). Light Touch Contact as a Balance Aid. *Physical Therapy, 77*(5), 476-487. doi:10.1093/ptj/77.5.476

Jeka, J. J., & Lackner, J. R. (1994). Fingertip contact influences human postural control. *Exp Brain Res, 100*(3), 495-502.

Jeka, J. J., & Lackner, J. R. (1995). The role of haptic cues from rough and slippery surfaces in human postural control. *Experimental Brain Research, 103*(2), 267-276. doi:10.1007/BF00231713

Jeka, J. J., Schoner, G., Dijkstra, T., Ribeiro, P., & Lackner, J. R. (1997). Coupling of fingertip somatosensory information to head and body sway. *Exp Brain Res, 113*(3), 475-483.

Johannsen, L., Wing, A. M., & Hatzitaki, V. (2007). Effects of Maintaining Touch Contact on Predictive and Reactive Balance. *Journal of Neurophysiology, 97*(4), 2686-2695. doi:10.1152/jn.00038.2007

Johannsen, L., Wing, A. M., & Hatzitaki, V. (2012). Contrasting effects of finger and shoulder interpersonal light touch on standing balance. *Journal of Neurophysiology, 107*(1), 216-225. doi:10.1152/jn.00149.2011

Jorgensen, L., Engstad, T., & Jacobsen, B. K. (2002). Higher incidence of falls in long-term stroke survivors than in population controls: depressive symptoms predict falls after stroke. *Stroke, 33*(2), 542-547.

Jutai, J., Coulson, S., Teasell, R., Bayley, M., Garland, J., Mayo, N., & Wood-Dauphinee, S. (2007). Mobility assistive device utilization in a prospective study of patients with first-ever stroke. *Arch Phys Med Rehabil, 88*(10), 1268-1275. doi:10.1016/j.apmr.2007.06.773

Kahn, J. H., & Hornby, T. G. (2009). Rapid and long-term adaptations in gait symmetry following unilateral step training in people with hemiparesis. *Phys Ther, 89*(5), 474-483. doi:10.2522/ptj.20080237

Kang, H., & Dingwell, J. B. (2008). Separating the effects of age and walking speed on gait variability. *Gait & Posture, 27*(4), 572-577. doi:10.1016/j.gaitpost.2007.07.009

Karunakaran, K., Abbruzzese, K., Xu, H., Ehrenberg, N., & Foulds, R. (2014). Haptic proprioception in a virtual locomotor task. *Conference proceedings : ... Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE Engineering in Medicine and Biology Society. Annual Conference, 2014*, 3594-3597. doi:10.1109/embc.2014.6944400

Kaski, D., Quadir, S., Patel, M., Yousif, N., & Bronstein, A. M. (2012). Enhanced locomotor adaptation aftereffect in the "broken escalator" phenomenon using anodal tDCS. *J Neurophysiol, 107*(9), 2493-2505. doi:10.1152/jn.00223.2011

Katz, N., Ring, H., Naveh, Y., Kizony, R., Feintuch, U., & Weiss, P. L. (2005). Interactive virtual environment training for safe street crossing of right hemisphere stroke patients with unilateral spatial neglect. *Disabil Rehabil, 27*(20), 1235-1243. doi:10.1080/09638280500076079

Kerrigan, D. C., Della Croce, U., Marciello, M., & Riley, P. O. (2000). A refined view of the determinants of gait: significance of heel rise. *Arch Phys Med Rehabil, 81*(8), 1077-1080.

Kesar, T. M., Binder-Macleod, S. A., Hicks, G. E., & Reisman, D. S. (2011). Minimal detectable change for gait variables collected during treadmill walking in individuals post-stroke. *Gait & Posture, 33*(2), 314-317. doi:10.1016/j.gaitpost.2010.11.024

Keshner, E. A. (2004). Virtual reality and physical rehabilitation: a new toy or a new research and rehabilitation tool? *J Neuroeng Rehabil, 1*(1), 8. doi:10.1186/1743-0003-1-8

Kim, C. M., & Eng, J. J. (2003). The relationship of lower-extremity muscle torque to locomotor performance in people with stroke. *Phys Ther, 83*(1), 49-57.

Kim, H. J., Park, I., Lee, H. J., & Lee, O. (2016). The reliability and validity of gait speed with different walking pace and distances against general health, physical function, and chronic disease in aged adults. *J Exerc Nutrition Biochem, 20*(3), 46-50. doi:10.20463/jenb.2016.09.20.3.7

Kim, H. Y., Yang, B. I., S.P.I, P., Kim, G. L. I., Jooc, E., You, I., & Hyunb, J. S. (2016). Best facilitated cortical activation during different stepping, treadmill, and robot-assisted walking training paradigms and speeds: A functional near-infrared spectroscopy neuroimaging study. *NeuroRehabilitation, 38*(2), 171-178.

Kim, J., Kim, K., Kim, D. Y., Chang, W. H., Park, C. I., Ohn, S. H., . . . Kim, S. I. (2007). Virtual environment training system for rehabilitation of stroke patients with unilateral neglect: crossing the virtual street. *Cyberpsychol Behav, 10*(1), 7-15. doi:10.1089/cpb.2006.9998

Kim, J., Oh, S.-I., Cho, H., Kim, H., Chon, J., Lee, W., . . . Kim, M. (2015). Gait patterns of chronic ambulatory hemiplegic elderly compared with normal Age-Matched elderly. *International Journal of Precision Engineering and Manufacturing, 16*(2), 385-392. doi:10.1007/s12541-015-0051-z

Kim, N., Park, Y., & Lee, B. H. (2015). Effects of community-based virtual reality treadmill training on balance ability in patients with chronic stroke. *J Phys Ther Sci, 27*(3), 655-658. doi:10.1589/jpts.27.655

Knaepen, K., Mierau, A., Tellez, H. F., Lefeber, D., & Meeusen, R. (2015). Temporal and spatial organization of gait-related electrocortical potentials. *Neurosci Lett, 599*, 75-80. doi:10.1016/j.neulet.2015.05.036

Kodesh, E., Falash, F., Sprecher, E., & Dickstein, R. (2015). Light touch and medio-lateral postural stability during short distance gait. *Neuroscience Letters, 584*, 378-381. doi:10.1016/j.neulet.2014.10.048

Kondoh, S., Araki, S., & Kaji, K. (1995). Relationship among Walking Speed, Stride Length, and Cadence in Hemiplegic Patients. *Rigakuryoho Kagaku, 10*(1), 11-14. doi:10.1589/rika.10.11

Krakauer, J. W., Ghez, C., & Ghilardi, M. F. (2005). Adaptation to visuomotor transformations: consolidation, interference, and forgetting. *J Neurosci, 25*(2), 473-478. doi:10.1523/JNEUROSCI.4218-04.2005

Krasovsky, T., Lamontagne, A., Feldman, A. G., & Levin, M. F. (2014). Effects of walking speed on gait stability and interlimb coordination in younger and older adults. *Gait Posture, 39*(1), 378-385. doi:10.1016/j.gaitpost.2013.08.011

Kroker, P. (1999). The problem of remaining upright. *BMJ, 319*.

Krueger, H., Koot, J., Hall, R. E., O'Callaghan, C., Bayley, M., & Corbett, D. (2015). Prevalence of Individuals Experiencing the Effects of Stroke in Canada: Trends and Projections. *Stroke, 46*(8), 2226-2231. doi:10.1161/STROKEAHA.115.009616

Kuan, T.-S., Tsou, J.-Y., & Su, F.-C. (1999). Hemiplegic gait of stroke patients: the effect of using a cane. *Archives of Physical Medicine and Rehabilitation, 80*(7), 777-784. doi:10.1016/S0003-9993(99)90227-7

Kuo, A. D. (2007). The six determinants of gait and the inverted pendulum analogy: A dynamic walking perspective. *Hum Mov Sci, 26*(4), 617-656. doi:10.1016/j.humov.2007.04.003

Kuo, A. D., & Donelan, J. M. (2010). Dynamic principles of gait and their clinical implications. *Phys Ther, 90*(2), 157-174. doi:10.2522/ptj.20090125

Kwakkel, G., Kollen, B., & Lindeman, E. (2004). Understanding the pattern of functional recovery after stroke: facts and theories. *Restor Neurol Neurosci, 22*(3-5), 281-299.

Lackland, D. T., Roccella, E. J., Deutsch, A. F., Fornage, M., George, M. G., Howard, G., . . . Translational, B. (2014). Factors influencing the decline in stroke mortality: a statement from the American Heart Association/American Stroke Association. *Stroke, 45*(1), 315-353. doi:10.1161/01.str.0000437068.30550.cf

Lackner, J. R., DiZio, P., Jeka, J., Horak, F., Krebs, D., & Rabin, E. (1999). Precision contact of the fingertip reduces postural sway of individuals with bilateral vestibular loss. *Exp Brain Res, 126*(4), 459-466.

Lackner, J. R., Rabin, E., & DiZio, P. (2001). Stabilization of posture by precision touch of the index finger with rigid and flexible filaments. *Exp Brain Res, 139*(4), 454-464.

Lacquaniti, F., Ivanenko, Y. P., & Zago, M. (2012). Patterned control of human locomotion. *The Journal of Physiology, 590*(10), 2189-2199. doi:10.1113/jphysiol.2011.215137

Lajoie, Y., Teasdale, N., Bard, C., & Fleury, M. (1996). Upright standing and gait: are there changes in attentional requirements related to normal aging? *Exp Aging Res, 22*(2), 185-198. doi:10.1080/03610739608254006

Lam, T., Anderschitz, M., & Dietz, V. (2006). Contribution of feedback and feedforward strategies to locomotor adaptations. *J Neurophysiol, 95*(2), 766-773. doi:10.1152/jn.00473.2005

Lamontagne, A., & Fung, J. (2004). Faster is better: implications for speed-intensive gait training after stroke. *Stroke, 35*(11), 2543-2548. doi:10.1161/01.STR.0000144685.88760.d7

Lamontagne, A., & Fung, J. (2009). Gaze and Postural Reorientation in the Control of Locomotor Steering After Stroke. *Neurorehabilitation and Neural Repair, 23*(3), 256-266. doi:10.1177/1545968308324549

Lamontagne, A., Keshner, E. A., Bugnariu, N., & Fung, J. (2014). Virtual Reality Reveals Mechanisms of Balance and Locomotor Impairments. *Virtual Reality for Physical and Motor Rehabilitation*, 169-202.

Lamontagne, A., Richards, C. L., & Malouin, F. (2000). Coactivation during gait as an adaptive behavior after stroke. *Journal of Electromyography and Kinesiology, 10*(6), 407-415. doi:10.1016/S1050-6411(00)00028-6

Langhorne, P., Coupar, F., & Pollock, A. (2009). Motor recovery after stroke: a systematic review. *Lancet Neurol, 8*(8), 741-754. doi:10.1016/S1474-4422(09)70150-4

LaRoche, D. P., Greenleaf, B. L., Croce, R. V., & McGaughy, J. A. (2014). Interaction of age, cognitive function, and gait performance in 50-80-year-olds. *Age (Dordr), 36*(4), 9693. doi:10.1007/s11357-014-9693-5

Lee, C. D., Folsom, A. R., & Blair, S. N. (2003). Physical activity and stroke risk: a meta-analysis. *Stroke, 34*(10), 2475-2481. doi:10.1161/01.STR.0000091843.02517.9D

Lee, D. N. (1980). The optic flow field: the foundation of vision. *Philos Trans R Soc Lond B Biol Sci, 290*(1038), 169-179.

Lee, D. N., & Lishman, J. R. (1975). Visual proprioceptive control of stance. *Journal of Human Movement Studies, 1*(2), 87-95.

Lee, S., Lee, D., Lee, Y., Jee, Y., Lee, G., & Park, D.-s. (2015). Influence of light touch using the fingertips on postural stability of poststroke patients. *Journal of Physical Therapy Science, 27*(2). doi:10.1589/jpts.27.469

Lindsay, P., Bayley, M., McDonald, A., Graham, I. D., Warner, G., & Phillips, S. (2008). Toward a more effective approach to stroke: Canadian Best Practice Recommendations for Stroke Care. *CMAJ, 178*(11), 1418-1425. doi:10.1503/cmaj.071253

Lord, S. E., & Rochester, L. (2005). Measurement of community ambulation after stroke: current status and future developments. *Stroke, 36*(7), 1457-1461. doi:10.1161/01.STR.0000170698.20376.2e

Lucareli, P. R., & Greve, J. M. (2006). Alteration of the load-response mechanism of the knee joint during hemiparetic gait following stroke analyzed by 3-dimensional kinematic. *Clinics (Sao Paulo), 61*(4), 295-300.

Luft, A. R., Macko, R. F., Forrester, L. W., Villagra, F., Ivey, F., Sorkin, J. D., . . . Hanley, D. F. (2008). Treadmill exercise activates subcortical neural networks and improves walking after stroke: a randomized controlled trial. *Stroke, 39*(12), 3341-3350. doi:10.1161/STROKEAHA.108.527531

Lugade, V., Lin, V., & Chou, L. S. (2011). Center of mass and base of support interaction during gait. *Gait Posture, 33*(3), 406-411. doi:10.1016/j.gaitpost.2010.12.013

Luque-Moreno, C., Ferragut-Garcías, A., Rodríguez-Blanco, C., Heredia-Rizo, A. M., Oliva-Pascual-Vaca, J., Kiper, P., & Oliva-Pascual-Vaca, Á. (2015). A Decade of Progress Using Virtual Reality for Poststroke Lower Extremity Rehabilitation: Systematic Review of the Intervention Methods. *BioMed research international, 2015*, 342529. doi:10.1155/2015/342529

Lutz, B. J., Young, M. E., Cox, K. J., Martz, C., & Creasy, K. R. (2011). The crisis of stroke: experiences of patients and their family caregivers. *Top Stroke Rehabil, 18*(6), 786-797. doi:10.1310/tsr1806-786

Lyon, I. N., & Day, B. L. (1997). Control of frontal plane body motion in human stepping. *Exp Brain Res, 115*(2), 345-356.

Ma, V. Y., Chan, L., & Carruthers, K. J. (2014). Incidence, prevalence, costs, and impact on disability of common conditions requiring rehabilitation in the United States: stroke, spinal cord injury, traumatic brain injury, multiple sclerosis, osteoarthritis, rheumatoid arthritis, limb loss, and back pain. *Arch Phys Med Rehabil, 95*(5), 986-995 e981. doi:10.1016/j.apmr.2013.10.032

MacKay-Lyons, M. (2002). Central pattern generation of locomotion: a review of the evidence. *Phys Ther, 82*(1), 69-83.

Malone, L. A., & Bastian, A. J. (2014). Spatial and temporal asymmetries in gait predict split-belt adaptation behavior in stroke. *Neurorehabil Neural Repair, 28*(3), 230-240. doi:10.1177/1545968313505912

Malone, L. A., Vasudevan, E. V., & Bastian, A. J. (2011). Motor adaptation training for faster relearning. *J Neurosci, 31*(42), 15136-15143. doi:10.1523/JNEUROSCI.1367-11.2011

Marigold, D. S., & Patla, A. E. (2008). Visual information from the lower visual field is important for walking across multi-surface terrain. *Experimental Brain Research, 188*(1). doi:10.1007/s00221-008-1335-7

Martinelli, A., Coelho, D., Magalhães, F., Kohn, A., & Teixeira, L. (2015). Light touch modulates balance recovery following perturbation: from fast response to stance restabilization. *Experimental Brain Research*, *233*(5). doi:10.1007/s00221-015-4214-z

Martino, G., Ivanenko, Y. P., Serrao, M., Ranavolo, A., d'Avella, A., Draicchio, F., . . . Lacquaniti, F. (2014). Locomotor patterns in cerebellar ataxia. *Journal of Neurophysiology, 112*(11), 2810-2821. doi:10.1152/jn.00275.2014

Mauerberg-deCastro, E. (2004). Developing an "anchor" system to enhance postural control. *Motor control, 8*(3), 339-358.

Mauerberg-deCastro, E., & Moraes, R. (2014). Haptic anchoring and human postural control. *Psychology & ...*.

Mawase, F., Haizler, T., Bar-Haim, S., & Karniel, A. (2013). Kinetic adaptation during locomotion on a split-belt treadmill. *J Neurophysiol, 109*(8), 2216-2227. doi:10.1152/jn.00938.2012

Mawase, F., Shmuelof, L., Bar-Haim, S., & Karniel, A. (2014). Savings in locomotor adaptation explained by changes in learning parameters following initial adaptation. *J Neurophysiol, 111*(7), 1444-1454. doi:10.1152/jn.00734.2013

McEwen, D., Taillon-Hobson, A., Bilodeau, M., Sveistrup, H., & Finestone, H. (2014). Virtual reality exercise improves mobility after stroke: an inpatient randomized controlled trial. *Stroke, 45*(6), 1853-1855. doi:10.1161/STROKEAHA.114.005362

McGinley, J. L., Baker, R., Wolfe, R., & Morris, M. E. (2009). The reliability of three-dimensional kinematic gait measurements: a systematic review. *Gait Posture, 29*(3), 360-369. doi:10.1016/j.gaitpost.2008.09.003

Mehrholz, J., Thomas, S., & Elsner, B. (2017). Treadmill training and body weight support for walking after stroke. *Cochrane Database Syst Rev, 8,* CD002840. doi:10.1002/14651858.CD002840.pub4

Mehrholz, J., Wagner, K., Rutte, K., Meissner, D., & Pohl, M. (2007). Predictive validity and responsiveness of the functional ambulation category in hemiparetic patients after stroke. *Arch Phys Med Rehabil, 88*(10), 1314-1319. doi:10.1016/j.apmr.2007.06.764

Mena, D., Mansour, J. M., & Simon, S. R. (1981). Analysis and synthesis of human swing leg motion during gait and its clinical applications. *J Biomech, 14*(12), 823-832.

Mercier, L., Audet, T., Hebert, R., Rochette, A., & Dubois, M. F. (2001). Impact of motor, cognitive, and perceptual disorders on ability to perform activities of daily living after stroke. *Stroke, 32*(11), 2602-2608.

Merians, A. S., Jack, D., Boian, R., Tremaine, M., Burdea, G. C., Adamovich, S. V., . . . Poizner, H. (2002). Virtual reality-augmented rehabilitation for patients following stroke. *Phys Ther, 82*(9), 898-915.

Merriaux, P., Dupuis, Y., Boutteau, R., Vasseur, P., & Savatier, X. (2017). A Study of Vicon System Positioning Performance. *Sensors (Basel), 17*(7). doi:10.3390/s17071591

Minassian, K., Hofstoetter, U. S., Dzeladini, F., Guertin, P. A., & Ijspeert, A. (2017). The Human Central Pattern Generator for Locomotion. *Neuroscientist*, 1073858417699790. doi:10.1177/1073858417699790

Mirelman, A., Patritti, B. L., Bonato, P., & Deutsch, J. E. (2010). Effects of virtual reality training on gait biomechanics of individuals post-stroke. *Gait Posture, 31*(4), 433-437. doi:10.1016/j.gaitpost.2010.01.016

Mochizuki, L., & Aliberti, S. (2017). Locomotion and Posture in Older Adults. *springer*, 45-54. doi:10.1007/978-3-319-48980-3_4

Moseley, A. M., Stark, A., Cameron, I. D., & Pollock, A. (2005). Treadmill training and body weight support for walking after stroke. *Cochrane Database Syst Rev*(4), CD002840. doi:10.1002/14651858.CD002840.pub2

Muir, B. C., Haddad, J. M., Heijnen, M. J., & Rietdyk, S. (2015). Proactive gait strategies to mitigate risk of obstacle contact are more prevalent with advancing age. *Gait Posture, 41*(1), 233-239. doi:10.1016/j.gaitpost.2014.10.005

Murphy, T. H., & Corbett, D. (2009). Plasticity during stroke recovery: from synapse to behaviour. *Nat Rev Neurosci, 10*(12), 861-872. doi:10.1038/nrn2735

Nadeau, S., Jacquemin, G., Fournier, C., Lamarre, Y., & Rossignol, S. (2010). Spontaneous motor rhythms of the back and legs in a patient with a complete spinal cord transection. *Neurorehabil Neural Repair, 24*(4), 377-383. doi:10.1177/1545968309349945

Nadeau, S., Lauziere, S., Mieveille, C., & Betschart, M. (2015). Locomotor asymmetry in post-stroke individuals: Why and how to use the split-belt treadmill to improve the gait performance. *Annals of Physical and Rehabilitation Medicine, 58*(1), e82.

Nam, S. H., Son, S. M., & Kim, K. (2017). Changes of gait parameters following constrained-weight shift training in patients with stroke. *J Phys Ther Sci, 29*(4), 673-676. doi:10.1589/jpts.29.673

Nanninga, C. S., Meijering, L., Schonherr, M. C., Postema, K., & Lettinga, A. T. (2015). Place attachment in stroke rehabilitation: a transdisciplinary encounter between cultural geography, environmental psychology and rehabilitation medicine. *Disabil Rehabil, 37*(13), 1125-1134. doi:10.3109/09638288.2014.955136

Nascimento, L. R., Ada, L., & Teixeira-Salmela, L. F. (2016). The provision of a cane provides greater benefit to community-dwelling people after stroke with a baseline walking speed between 0.4 and 0.8 metres/second: an experimental study. *Physiotherapy, 102*(4), 351-356. doi:10.1016/j.physio.2015.10.005

Nieuwenhuijzen, P. H., & Duysens, J. (2007). Proactive and reactive mechanisms play a role in stepping on inverting surfaces during gait. *J Neurophysiol, 98*(4), 2266-2273. doi:10.1152/jn.01226.2006

Nudo, R. J., & Friel, K. M. (1999). Cortical plasticity after stroke: implications for rehabilitation. *Rev Neurol (Paris), 155*(9), 713-717.

Nyberg, L., & Gustafson, Y. (1995a). Patient falls in stroke rehabilitation. A challenge to rehabilitation strategies. *Stroke; a journal of cerebral circulation, 26*(5), 838-842. doi:10.1161/01.STR.26.5.838

Nyberg, L., & Gustafson, Y. (1995b). Patient falls in stroke rehabilitation. A challenge to rehabilitation strategies. *Stroke, 26*(5), 838-842.

Nyberg, L., & Gustafson, Y. (1997). Fall prediction index for patients in stroke rehabilitation. *Stroke, 28*(4), 716-721.

Oates, A. R., Hauck, L., Moraes, R., & Sibley, K. M. (2017). The effects of haptic input on biomechanical and neurophysiological parameters of walking: A scoping review. *Gait & Posture*.

Oates, A. R., Perez, C. F., & Fung, J. (2008). A VR-haptic locomotor system to retrain anticipatory postural adjustments post stroke. *IEEE Virtual Rehabilitation, 2008, September*.

Olney, S. J. R., C. (1996). Hemiparetic gait following stroke. Part I: Characteristics. *Gait & Posture, 4*(2), 136-148.

Ortega, J. D., & Farley, C. T. (2005). Minimizing center of mass vertical movement increases metabolic cost in walking. *Journal of Applied Physiology, 99*(6), 2099-2107. doi:10.1152/japplphysiol.00103.2005

Pai, Y.-C., & Patton, J. (1997). Center of mass velocity-position predictions for balance control. *Journal of Biomechanics, 30*(4), 347-354. doi:10.1016/S0021-9290(96)00165-0

Palmer, R. (2001). Falls in elderly patients: predictable and preventable. *Cleve Clin J Med, 68*(4), 303-306.

Pang, M. Y., Eng, J. J., & Miller, W. C. (2007). Determinants of satisfaction with community reintegration in older adults with chronic stroke: role of balance self-efficacy. *Phys Ther, 87*(3), 282-291. doi:10.2522/ptj.20060142

Parent, A. (2005). Duchenne De Boulogne: a pioneer in neurology and medical photography. *Can J Neurol Sci, 32*(3), 369-377.

Patel, M., Kaski, D., & Bronstein, A. M. (2014). Attention modulates adaptive motor learning in the 'broken escalator' paradigm. *Exp Brain Res, 232*(7), 2349-2357. doi:10.1007/s00221-014-3931-z

Patla, A. E. (2003). Strategies for dynamic stability during adaptive human locomotion. *IEEE Eng Med Biol Mag, 22*(2), 48-52.

Patterson, K. K., Gage, W. H., Brooks, D., Black, S. E., & McIlroy, W. E. (2010). Evaluation of gait symmetry after stroke: A comparison of current methods and recommendations for standardization. *Gait & Posture, 31*(2), 241-246. doi:10.1016/j.gaitpost.2009.10.014

Patterson, S. L., Forrester, L. W., Rodgers, M. M., Ryan, A. S., Ivey, F. M., Sorkin, J. D., & Macko, R. F. (2007). Determinants of walking function after stroke: differences by deficit severity. *Arch Phys Med Rehabil, 88*(1), 115-119. doi:10.1016/j.apmr.2006.10.025

Patterson, S. L., Rodgers, M. M., Macko, R. F., & Forrester, L. W. (2008). Effect of treadmill exercise training on spatial and temporal gait parameters in subjects with chronic stroke: a preliminary report. *Journal of rehabilitation research and development, 45*(2), 221-228.

Perera, S., Mody, S. H., Woodman, R. C., & Studenski, S. A. (2006). Meaningful change and responsiveness in common physical performance measures in older adults. *J Am Geriatr Soc, 54*(5), 743-749. doi:10.1111/j.1532-5415.2006.00701.x

Perez, C. F., & Fung, J. (2011). An Instrumented Cane Devised for Gait Rehabilitation and Research. *Journal of Physical Therapy Education Vol. 25*(No. 1 , Winter 2011).

Perry, J., Garrett, M., Gronley, J. K., & Mulroy, S. J. (1995). Classification of walking handicap in the stroke population. *Stroke, 26*(6), 982-989.

Peterka, R. J. (2002). Sensorimotor integration in human postural control. *J Neurophysiol, 88*(3), 1097-1118.

Peterka, R. J., & Loughlin, P. J. (2004). Dynamic regulation of sensorimotor integration in human postural control. *J Neurophysiol, 91*(1), 410-423. doi:10.1152/jn.00516.2003

Peterson, C. L., Cheng, J., Kautz, S. A., & R.R., N. (2011). LEG EXTENSION IS AN IMPORTANT PREDICTOR OF PARETIC LEG PROPULSION IN HEMIPARETIC WALKING. *Gait & Posture, 32*(4), 451-456.

Powell, W., Stevens, B., & Simmonds, M. (2009). Treadmill Interface for Virtual Reality vs. Overground Walking: A Comparison of Gait in Individuals with and without Pain. *Stud Health Technol Inform, 144*, 198-203.

Punt, M., Bruijn, S. M., Roeles, S., van de Port, I. G., Wittink, H., & van Dieen, J. H. (2017). Responses to gait perturbations in stroke survivors who prospectively experienced falls or no falls. *J Biomech, 55*, 56-63. doi:10.1016/j.jbiomech.2017.02.010

Qiu, F., Cole, M. H., Davids, K. W., Hennig, E. M., Silburn, P. A., Netscher, H., & Kerr, G. K. (2012). Enhanced somatosensory information decreases postural sway in older people. *Gait Posture, 35*(4), 630-635. doi:10.1016/j.gaitpost.2011.12.013

Rábago, C. A., Dingwell, J. B., & Wilken, J. M. (2015). Reliability and Minimum Detectable Change of Temporal-Spatial, Kinematic, and Dynamic Stability Measures during Perturbed Gait. *PLOS ONE, 10*(11). doi:10.1371/journal.pone.0142083

Rabin, E., Chen, J., Muratori, L., DiFrancisco-Donoghue, J., & Werner, W. G. (2013). Haptic feedback from manual contact improves balance control in people with Parkinson's disease. *Gait & Posture, 38*(3), 373-379. doi:10.1016/j.gaitpost.2012.12.008

Raja, B., Neptune, R. R., & Kautz, S. A. (2012). Quantifiable patterns of limb loading and unloading during hemiparetic gait: Relation to kinetic and kinematic parameters. *J Rehabil Res Dev, 49*(9), 1293-1304.

Rand, D., Katz, N., & Weiss, P. L. (2007). Evaluation of virtual shopping in the VMall: comparison of post-stroke participants to healthy control groups. *Disabil Rehabil, 29*(22), 1710-1719. doi:10.1080/09638280601107450

Reisman, D. S., Block, H. J., & Bastian, A. J. (2005). Interlimb Coordination During Locomotion: What Can be Adapted and Stored? *Journal of Neurophysiology, 94*(4), 2403-2415. doi:10.1152/jn.00089.2005

Reisman, D. S., McLean, H., & Bastian, A. J. (2010). Split-Belt Treadmill Training Poststroke: A Case Study. *Journal of Neurologic Physical Therapy, 34*(4), 202. doi:10.1097/NPT.0b013e3181fd5eab

Reisman, D. S., McLean, H., Keller, J., Danks, K. A., & Bastian, A. J. (2013). Repeated Split-Belt Treadmill Training Improves Poststroke Step Length Asymmetry. *Neurorehabilitation and Neural Repair, 27*(5), 460-468. doi:10.1177/1545968312474118

Reisman, D. S., Wityk, R., Silver, K., & Bastian, A. J. (2007). Locomotor adaptation on a split-belt treadmill can improve walking symmetry post-stroke. *Brain, 130*(Pt 7), 1861-1872. doi:10.1093/brain/awm035

Reisman, D. S., Wityk, R., Silver, K., & Bastian, A. J. (2009a). Split-belt treadmill adaptation transfers to overground walking in persons poststroke. *Neurorehabilitation and Neural Repair, 23*(7), 735-744. doi:10.1177/1545968309332880

Reisman, D. S., Wityk, R., Silver, K., & Bastian, A. J. (2009b). Split-belt treadmill adaptation transfers to overground walking in persons poststroke. *Neurorehabil Neural Repair, 23*(7), 735-744. doi:10.1177/1545968309332880

Reynolds, R. F., & Bronstein, A. M. (2003). The broken escalator phenomenon. Aftereffect of walking onto a moving platform. *Exp Brain Res, 151*(3), 301-308. doi:10.1007/s00221-003-1444-2

Rondeau, L., Corriveau, H., Bier, N., Camden, C., Champagne, N., & Dion, C. (2010). Effectiveness of a rehabilitation dog in fostering gait retraining for adults with a recent stroke: A multiple single-case study. *NeuroRehabilitation, 27*(2), 155-163. doi:10.3233/nre-2010-0592

Sangani, S., Lamontagne, A., & Fung, J. (2015). Cortical mechanisms underlying sensorimotor enhancement promoted by walking with haptic inputs in a virtual environment. *Prog Brain Res, 218*, 313-330. doi:10.1016/bs.pbr.2014.12.003

Saposnik, G., Cohen, L. G., Mamdani, M., Pooyania, S., Ploughman, M., Cheung, D., . . . Stroke Outcomes Research, C. (2016). Efficacy and safety of non-immersive virtual reality exercising in stroke rehabilitation (EVREST): a randomised, multicentre, single-blind, controlled trial. *Lancet Neurol, 15*(10), 1019-1027. doi:10.1016/S1474-4422(16)30121-1

Saposnik, G., Levin, M., & Outcome Research Canada Working, G. (2011). Virtual reality in stroke rehabilitation: a meta-analysis and implications for clinicians. *Stroke, 42*(5), 1380-1386. doi:10.1161/STROKEAHA.110.605451

Saunders, J. B., Inman, V. T., & Eberhart, H. D. (1953). The major determinants in normal and pathological gait. *J Bone Joint Surg Am, 35-A*(3), 543-558.

Savin, D. N., Morton, S. M., & Whitall, J. (2014). Generalization of improved step length symmetry from treadmill to overground walking in persons with stroke and hemiparesis. *Clinical Neurophysiology, 125*(5), 1012-1020. doi:10.1016/j.clinph.2013.10.044

Schmid, A., Duncan, P. W., Studenski, S., Lai, S. M., Richards, L., Perera, S., & Wu, S. S. (2007). Improvements in speed-based gait classifications are meaningful. *Stroke, 38*(7), 2096-2100. doi:10.1161/STROKEAHA.106.475921

Scobbie, L., McLean, D., Dixon, D., Duncan, E., & Wyke, S. (2013). Implementing a framework for goal setting in community based stroke rehabilitation: a process evaluation. *BMC Health Serv Res, 13*, 190. doi:10.1186/1472-6963-13-190

Sepehri, A., & Chernomas, R. (2004). Is the Canadian health care system fiscally sustainable? *Int J Health Serv., 32*(4), 229-243.

Shadmehr, R., & Mussa-Ivaldi, F. A. (1994). Adaptive representation of dynamics during learning of a motor task. *J Neurosci, 14*(5 Pt 2), 3208-3224.

Sheffler, L. R., & Chae, J. (2015). Physical Medicine and Rehabilitation Clinics of North America. *Physical Medicine and Rehabilitation Clinics of North America, 26*(4), 611-623. doi:10.1016/j.pmr.2015.06.006

Shema, S., Brozgol, M., Dorfman, M., Maidan, I., Sharaby-Yeshayahu, L., Malik-Kozuch, H., . . . Mirelman, A. (2014). Clinical Experience Using a 5-Week Treadmill Training Program With Virtual Reality to Enhance Gait in an Ambulatory Physical Therapy Service. *Physical Therapy, 94*(9), 1319-1326. doi:10.2522/ptj.20130305

Sherrington, C. S. (1910). Flexion-reflex of the limb, crossed extension-reflex, and reflex stepping and standing. *J Physiol, 40*(1-2), 28-121.

Shumway-Cook, A., Patla, A. E., Stewart, A., Ferrucci, L., Ciol, M. A., & Guralnik, J. M. (2002). Environmental demands associated with community mobility in older adults with and without mobility disabilities. *Phys Ther, 82*(7), 670-681.

Silva, S. M., Correa, F. I., Faria, C. D., Buchalla, C. M., Silva, P. F., & Correa, J. C. (2015). Evaluation of post-stroke functionality based on the International Classification of Functioning, Disability, and Health: a proposal for use of assessment tools. *J Phys Ther Sci, 27*(6), 1665-1670. doi:10.1589/jpts.27.1665

Simpson, L. A., Miller, W. C., & Eng, J. J. (2011). Effect of Stroke on Fall Rate, Location and Predictors: A Prospective Comparison of Older Adults with and without Stroke. *PLOS ONE, 6*(4). doi:10.1371/journal.pone.0019431

Singer, J. C., Mansfield, A., Danells, C. J., McIlroy, W. E., & Mochizuki, G. (2013). The effect of post-stroke lower-limb spasticity on the control of standing balance: Inter-limb spatial and temporal synchronisation of centres of pressure. *Clinical biomechanics (Bristol, Avon), 28*(8), 921-926. doi:10.1016/j.clinbiomech.2013.07.010

Sinitski, E. H., Lemaire, E. D., Baddour, N., Besemann, M., Dudek, N. L., & Hebert, J. S. (2015). Fixed and self-paced treadmill walking for able-bodied and transtibial amputees in a multi-terrain virtual environment. *Gait Posture, 41*(2), 568-573. doi:10.1016/j.gaitpost.2014.12.016

Soriano, T. A., DeCherrie, L. V., & Thomas, D. C. (2007). Falls in the community-dwelling older adult: a review for primary-care providers. *Clin Interv Aging, 2*(4), 545-554.

Sorrento, G., Archambault, P. A., & Fung, J. (2013). A research protocol exploring the use of haptic forces for stroke rehabilitation. *Virtual Rehabilitation (ICVR), 2013 International Conference on*.

Sorrento, G., Archambault, P. S., & Fung, J. (2018). Adaptation and post-adaptation effects of haptic forces on locomotion in healthy young adults. *J Neuroeng Rehabil, 15*(1), 20. doi:10.1186/s12984-018-0364-0

Stanhope, V. A., Knarr, B. A., Reisman, D. S., & Higginson, J. S. (2014). Frontal plane compensatory strategies associated with self-selected walking speed in individuals post-stroke. *Clinical Biomechanics, 29*(5), 518-522. doi:10.1016/j.clinbiomech.2014.03.013

Studenski, S., Perera, S., Patel, K., Rosano, C., Faulkner, K., Inzitari, M., . . . Guralnik, J. (2011). Gait Speed and Survival in Older Adults. *JAMA, 305*(1), 50-58. doi:10.1001/jama.2010.1923

Sudarsky, L. (1990). Geriatrics: gait disorders in the elderly. *N Engl J Med, 322*(20), 1441-1446. doi:10.1056/NEJM199005173222007

Sulzer, J. S., Gordon, K. E., Dhaher, Y. Y., Peshkin, M. A., & Patton, J. L. (2010). Preswing knee flexion assistance is coupled with hip abduction in people with stiff-knee gait after stroke. *Stroke, 41*(8), 1709-1714. doi:10.1161/STROKEAHA.110.586917

Suzuki, M., Miyai, I., Ono, T., Oda, I., Konishi, I., Kochiyama, T., & Kubota, K. (2004). Prefrontal and premotor cortices are involved in adapting walking and running speed on the treadmill: an optical imaging study. *NeuroImage, 23*(3), 1020-1026. doi:10.1016/j.neuroimage.2004.07.002

Sveistrup, H. (2004). Motor rehabilitation using virtual reality. *J Neuroeng Rehabil, 1*(1), 10. doi:10.1186/1743-0003-1-10

Tang, K. S., Kaski, D., Allum, J. H., & Bronstein, A. M. (2013). The effect of gait approach velocity on the broken escalator phenomenon. *Exp Brain Res, 226*(3), 335-346. doi:10.1007/s00221-013-3438-z

Teasell, R. (2003). Stroke recovery and rehabilitation. *Stroke, 34*(2), 365-366.

Teasell, R., Bayona, N., Salter, K., Hellings, C., & Bitensky, J. (2006). Progress in clinical neurosciences: stroke recovery and rehabilitation. *Can J Neurol Sci, 33*(4), 357-364.

Teasell, R., McRae, M., Foley, N., & Bhardwaj, A. (2002). The incidence and consequences of falls in stroke patients during inpatient rehabilitation: factors associated with high risk. *Arch Phys Med Rehabil, 83*(3), 329-333.

Thach, T. W., & Bastian, A. J. (2004). Role of the cerebellum in the control and adaptation of gait in health and disease. *Progress in brain research, 143,* 353-366.

Thorvaldsen, P., Asplund, K., Kuulasmaa, K., Rajakangas, A. M., & Schroll, M. (1995). Stroke incidence, case fatality, and mortality in the WHO MONICA project. World Health Organization Monitoring Trends and Determinants in Cardiovascular Disease. *Stroke, 26*(3), 361-367.

Tilson, J. K., Sullivan, K. J., Cen, S. Y., Rose, D. K., Koradia, C. H., Azen, S. P., . . . Locomotor Experience Applied Post Stroke Investigative, T. (2010). Meaningful gait speed improvement during the first 60 days poststroke: minimal clinically important difference. *Phys Ther, 90*(2), 196-208. doi:10.2522/ptj.20090079

Tinetti, M. E. (2003). Clinical practice. Preventing falls in elderly persons. *N Engl J Med, 348*(1), 42-49. doi:10.1056/NEJMcp020719

Tinetti, M. E., Baker, D. I., McAvay, G., Claus, E. B., Garrett, P., Gottschalk, M., . . . Horwitz, R. I. (1994). A multifactorial intervention to reduce the risk of falling among elderly people living in the community. *N Engl J Med, 331*(13), 821-827. doi:10.1056/NEJM199409293311301

Tinetti, M. E., Richman, D., & Powell, L. (1990). Falls efficacy as a measure of fear of falling. *J Gerontol, 45*(6), P239-243.

Torres-Oviedo, G., & Bastian, A. J. (2010). Seeing is believing: effects of visual contextual cues on learning and transfer of locomotor adaptation. *J Neurosci, 30*(50), 17015-17022. doi:10.1523/JNEUROSCI.4205-10.2010

Torres-Oviedo, G., & Bastian, A. J. (2012). Natural error patterns enable transfer of motor learning to novel contexts. *Journal of Neurophysiology, 107*(1), 346-356. doi:10.1152/jn.00570.2011

Torres-Oviedo, G., Vasudevan, E., Malone, L., & Bastian, A. J. (2011). Locomotor adaptation. *Prog Brain Res, 191,* 65-74. doi:10.1016/B978-0-444-53752-2.00013-8

Tsur, A., & Segal, Z. (2010). Falls in stroke patients: risk factors and risk management. *Isr Med Assoc J, 12*(4), 216-219.

Tuomilehto, J., Bonita, R., Stewart, A., Nissinen, A., & Salonen, J. T. (1991). Hypertension, cigarette smoking, and the decline in stroke incidence in eastern Finland. *Stroke, 22*(1), 7-11.

Turchet, L., Serafin, S., & Cesari, P. (2013). Walking Pace Affected by Interactive Sounds Simulating Stepping on Different Terrains. *Acm Transactions on Applied Perception, 10*(4), 14. doi:10.1145/2536764.2536770

Tyrell, C. M., Helm, E., & Reisman, D. S. (2015). Locomotor adaptation is influenced by the interaction between perturbation and baseline asymmetry after stroke. *Journal of Biomechanics, 48*(11), 2849-2857. doi:10.1016/j.jbiomech.2015.04.027

van Kan, G. A., Rolland, Y., Andrieu, S., Bauer, J., Beauchet, O., Bonnefoy, M., . . . Vellas, B. (2009). GAIT SPEED AT USUAL PACE AS A PREDICTOR OF ADVERSE OUTCOMES IN COMMUNITY-DWELLING OLDER PEOPLE AN INTERNATIONAL ACADEMY ON NUTRITION AND AGING (IANA) TASK FORCE. *Journal of Nutrition Health & Aging, 13*(10), 881-889.

Vasudevan, E. V. L., & Bastian, A. J. (2010). Split-Belt Treadmill Adaptation Shows Different Functional Networks for Fast and Slow Human Walking. *Journal of Neurophysiology, 103*(1), 183-191. doi:10.1152/jn.00501.2009

Vaugoyeau, M., Viel, S., Amblard, B., Azulay, J. P., & Assaiante, C. (2008). Proprioceptive contribution of postural control as assessed from very slow oscillations of the support in healthy humans. *Gait Posture, 27*(2), 294-302. doi:10.1016/j.gaitpost.2007.04.003

Verghese, J., Wang, C., & Holtzer, R. (2011). Relationship of clinic-based gait speed measurement to limitations in community-based activities in older adults. *Arch Phys Med Rehabil, 92*(5), 844-846. doi:10.1016/j.apmr.2010.12.030

Vidoni, E. D., Tull, A., & Kluding, P. (2008). Use of three gait-training strategies in an individual with multiple, chronic strokes. *Journal of Neurologic Physical Therapy, 32*(2), 88-96.

von Schroeder, H. P., & Coutts, R. D. (1995). Gait parameters following stroke: a practical assessment. *Journal of ...*.

Wagenaar, R. C., & Beek, W. J. (1992). Hemiplegic gait: A kinematic analysis using walking speed as a basis. *Journal of Biomechanics, 25*(9), 1007-1015. doi:10.1016/0021-9290(92)90036-Z

Wagner, J., Solis-Escalante, T., Scherer, R., Neuper, C., & Gernot Müller-Putz, G. (2014). It's how you get there: walking down a virtual alley activates premotor and parietal areas. *Front Hum Neurosci., 8*(93).

Watanabe, Y. (2005). Fear of falling among stroke survivors after discharge from inpatient rehabilitation. *Int J Rehabil Res, 28*(2), 149-152.

Waters, R. L., & Mulroy, S. (1999). The energy expenditure of normal and pathologic gait. *Gait & Posture, 9*(3), 207-231. doi:10.1016/S0966-6362(99)00009-0

Weerdesteyn, V., de Niet, M., van Duijnhoven, H. J., & Geurts, A. C. (2008a). Falls in individuals with stroke. *J Rehabil Res Dev, 45*(8), 1195-1213.

Weerdesteyn, V., de Niet, M., van Duijnhoven, H. J. R., & Geurts, A. C. H. (2008b). Falls in individuals with stroke. *Journal of rehabilitation research and development, 45*(8), 1195-1213.

Winter, D. A., Patla, A. E., Frank, J. S., & Walt, S. E. (1990). Biomechanical walking pattern changes in the fit and healthy elderly. *Phys Ther, 70*(6), 340-347.

Wonsetler, E. C., & Bowden, M. G. (2017). A systematic review of mechanisms of gait speed change post-stroke. Part 1: spatiotemporal parameters and asymmetry ratios. *Top Stroke Rehabil, 24*(6), 435-446. doi:10.1080/10749357.2017.1285746

Wutzke, C. J., Mercer, V. S., & Lewek, M. D. (2013). Influence of lower extremity sensory function on locomotor adaptation following stroke: a review. *Topics in stroke rehabilitation, 20*(3), 233-240. doi:10.1310/tsr2003-233

Yang, Y.-R., Tsai, M.-P., Chuang, T.-Y., Sung, W.-H., & Wang, R.-Y. (2008). Virtual reality-based training improves community ambulation in individuals with stroke: A randomized controlled trial. *Gait & Posture, 28*(2), 201-206. doi:10.1016/j.gaitpost.2007.11.007

Yu, E., Abe, M., Masani, K., Kawashima, N., Eto, F., Haga, N., & Nakazawa, K. (2008). Evaluation of postural control in quiet standing using center of mass acceleration: comparison among the young, the elderly, and people with stroke. *Arch Phys Med Rehabil, 89*(6), 1133-1139. doi:10.1016/j.apmr.2007.10.047

APPENDICES

Appendix A:

Falls Efficacy Scale used in studies 2 and 3 (FES I both English and French versions)

FES-I

Now we would like to ask some questions about how concerned you are about the possibility of falling. Please reply thinking about how you usually do the activity. If you currently don't do the activity (e.g. if someone does your shopping for you), please answer to show whether you think you would be concerned about falling IF you did the activity. For each of the following activities, please tick the box which is closest to your own opinion to show how concerned you are that you might fall if you did this activity.

		Not at all concerned 1	*Somewhat concerned 2*	*Fairly concerned 3*	*Very concerned 4*
1	Cleaning the house (e.g. sweep, vacuum or dust)	*1* ☐	*2* ☐	*3* ☐	*4* ☐
2	Getting dressed or undressed	*1* ☐	*2* ☐	*3* ☐	*4* ☐
3	Preparing simple meals	*1* ☐	*2* ☐	*3* ☐	*4* ☐
4	Taking a bath or shower	*1* ☐	*2* ☐	*3* ☐	*4* ☐
5	Going to the shop	*1* ☐	*2* ☐	*3* ☐	*4* ☐
6	Getting in or out of a chair	*1* ☐	*2* ☐	*3* ☐	*4* ☐
7	Going up or down stairs	*1* ☐	*2* ☐	*3* ☐	*4* ☐
8	Walking around in the neighbourhood	*1* ☐	*2* ☐	*3* ☐	*4* ☐
9	Reaching for something above your head or on the ground	*1* ☐	*2* ☐	*3* ☐	*4* ☐
10	Going to answer the telephone before it stops ringing	*1* ☐	*2* ☐	*3* ☐	*4* ☐
11	Walking on a slippery surface (e.g. wet or icy)	*1* ☐	*2* ☐	*3* ☐	*4* ☐
12	Visiting a friend or relative	*1* ☐	*2* ☐	*3* ☐	*4* ☐
13	Walking in a place with crowds	*1* ☐	*2* ☐	*3* ☐	*4* ☐
14	Walking on an uneven surface (e.g. rocky ground, poorly maintained pavement)	*1* ☐	*2* ☐	*3* ☐	*4* ☐
15	Walking up or down a slope	*1* ☐	*2* ☐	*3* ☐	*4* ☐
16	Going out to a social event (e.g. religious service, family gathering or club meeting)	*1* ☐	*2* ☐	*3* ☐	*4* ☐

FES-I: Prof Lucy Yardley and Prof Chris Todd

FES-I

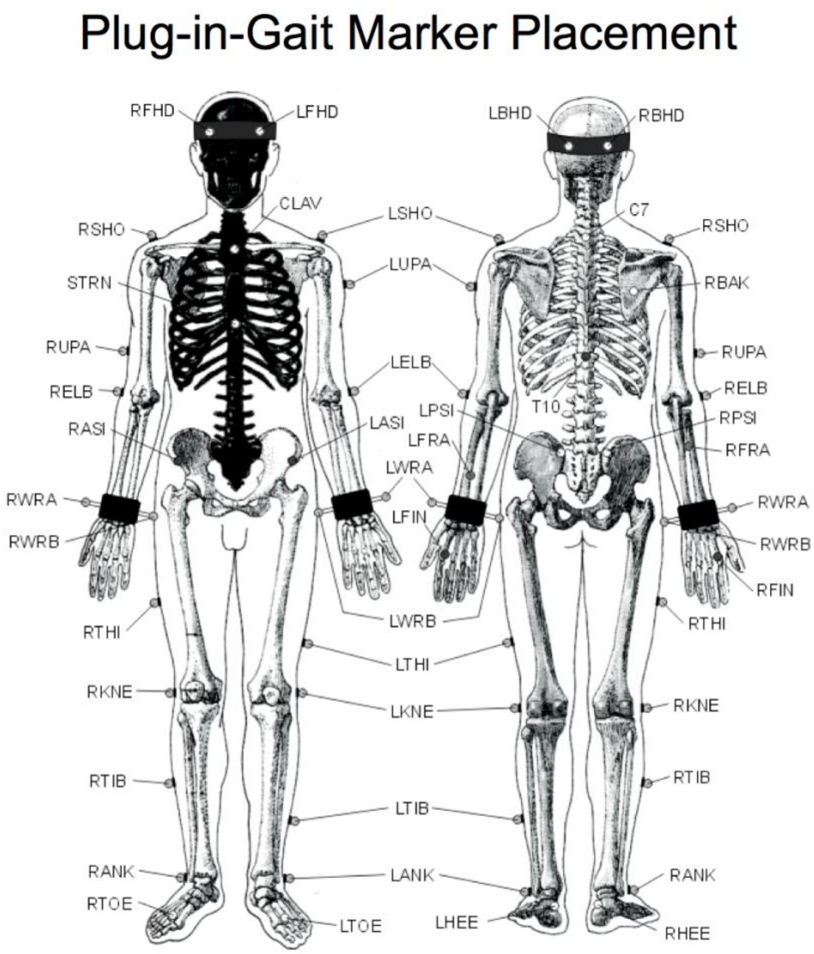
Nous aimerions vous poser quelques questions qui ont pour but de déterminer si vous ressentez de l'inquiétude face à la possibilité de tomber. Répondez en pensant à la manière dont vous effectuez habituellement cette activité. Si actuellement vous ne faites pas cette activité (par exemple si quelqu'un fait les courses à votre place), répondez à la question en imaginant votre degré d'inquiétude **SI** vous réalisiez en réalité cette activité. Pour chacune des activités suivantes, mettez une croix dans la case qui correspond le plus à votre opinion et qui montre le degré d'inquiétude que vous ressentez face au fait de pouvoir tomber lors de la réalisation de cette activité.

		Pas du tout inquiet *1*	*Un peu Inquiet* *2*	*Assez Inquiet* *3*	*Très Inquiet* *4*
1	Faire votre ménage (par ex : balayer, passer l'aspirateur, ou la poussière)	*1* ☐	*2* ☐	*3* ☐	*4* ☐
2	Vous habiller et vous déshabiller	*1* ☐	*2* ☐	*3* ☐	*4* ☐
3	Préparer des repas simples	*1* ☐	*2* ☐	*3* ☐	*4* ☐
4	Prendre une douche ou un bain	*1* ☐	*2* ☐	*3* ☐	*4* ☐
5	Aller faire des courses	*1* ☐	*2* ☐	*3* ☐	*4* ☐
6	Vous lever d'une chaise ou vous asseoir	*1* ☐	*2* ☐	*3* ☐	*4* ☐
7	Monter ou descendre des escaliers	*1* ☐	*2* ☐	*3* ☐	*4* ☐
8	Vous promener dehors dans le quartier	*1* ☐	*2* ☐	*3* ☐	*4* ☐
9	Atteindre quelque chose au-dessus de votre tête ou par terre	*1* ☐	*2* ☐	*3* ☐	*4* ☐
10	Aller répondre au téléphone avant qu'il s'arrête de sonner	*1* ☐	*2* ☐	*3* ☐	*4* ☐
11	Marcher sur une surface glissante (par ex : mouillée ou verglacée)	*1* ☐	*2* ☐	*3* ☐	*4* ☐
12	Rendre visite à un ami, ou à une connaissance	*1* ☐	*2* ☐	*3* ☐	*4* ☐
13	Marcher dans un endroit où il y a beaucoup de monde	*1* ☐	*2* ☐	*3* ☐	*4* ☐
14	Marcher sur un sol inégal (route caillouteuse, un trottoir non entretenu)	*1* ☐	*2* ☐	*3* ☐	*4* ☐
15	Descendre ou monter une pente	*1* ☐	*2* ☐	*3* ☐	*4* ☐
16	Sortir (par ex : service religieux, réunion de famille, rencontre d'une association)	*1* ☐	*2* ☐	*3* ☐	*4* ☐

FES-I Swiss French translated by Prof Chantal Piot-Ziegler

Appendix B:

Marker placement for kinematic gait recording in Vicon.



The following describes in detail where the Plug-in-Gait markers should be placed on the subject. Where left side markers only are listed, the positioning is identical for the right side.