Université de Montréal

Post-effets et rééducation à la marche chez le sujet hémiparétique

Locomotor after-effects and rehabilitation of gait in individuals with hemiparesis

par

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Résumé

L'asymétrie de longueur de pas est une caractéristique du patron de marche fréquemment observée chez des personnes qui ont subi un accident vasculaire cérébral (AVC). Très peu d'interventions conventionnelles en réadaptation ont démontré leur efficacité sur ce paramètre de marche. Une approche novatrice utilisant un tapis roulant à double courroie (DC) a récemment présenté des effets prometteurs en réduisant, à court et long termes, l'asymétrie de longueur de pas chez des personnes post-AVC. Cependant, une meilleure compréhension des mécanismes sous-jacents aux changements induits par cette intervention est nécessaire avant que l'utilisation de cette intervention soit recommandée en clinique. Ce projet doctoral visait à améliorer les connaissances sur la contribution musculaire et les facteurs biomécaniques impliqués dans les changements immédiats (c.-à-d. les post-effets) et à long terme de l'asymétrie de la longueur de pas chez des personnes post-AVC. Les objectifs principaux étaient 1) d'analyser l'activité musculaire associée aux changements de longueur de pas après la marche sur un tapis à DC avec des vitesses de courroies inégales, 2) d'évaluer les effets d'un entraînement sur un tapis à DC sur l'asymétrie de longueur de pas et sur la capacité de marche au sol et 3) de quantifier la contribution musculaire et les stratégies sous-jacentes aux changements du patron de marche résultant de l'entraînement. L'hypothèse générale était que les muscles distaux des membres inférieurs seraient majoritairement impliqués dans les changements de longueur de pas induits par la marche sur le tapis à DC avec des vitesses de courroies inégales (ratio 2:1) chez les personnes post-AVC. L'étude transversale (article #1) a quantifié l'activité musculaire des membres inférieurs associée aux post-effets observés au niveau de la longueur de pas après six minutes de marche sur le tapis à DC (ratio 2:1) chez 16 personnes post-AVC et 10 personnes saines. Les résultats ont confirmé que les muscles distaux, c.-à-d. les fléchisseurs plantaires et dorsaux, étaient associés aux post-effets de la longueur de pas. Ces effets ont été observés, quel que soit le membre inférieur (c.-à-d. parétique ou non parétique) qui était sur la courroie rapide. La deuxième étude, a démontré que six séances d'entraînement sur le tapis à DC conduisaient à une réduction de l'asymétrie de longueur de pas et amélioraient la vitesse de marche sur le sol chez 12 personnes post-AVC (article #2). Les changements ont persisté un mois après l'entraînement. En outre, les résultats de cette étude pilote ont suggéré une bonne faisabilité de ce protocole d'entraînement dans un environnement clinique (données supplémentaires de l'article #2). Dans l'article #3, il a été démontré que l'amélioration post-entraînement de la symétrie et de la vitesse de marche était associée à une variété de stratégies et de contributions musculaires chez nos participants. Cependant, une contribution prédominante a été observée au niveau du membre inférieur entraîné sur la courroie rapide avec des tailles d'effet modérées obtenues surtout pour les changements de moment et d'activité musculaire des fléchisseurs plantaires pendant la phase d'appui de la marche. Suite à ces résultats et ceux de l'article #1, il a été suggéré que ce groupe musculaire jouait un rôle principal dans l'adaptation locomotrice et la réduction à long terme de l'asymétrie de la longueur de pas chez des personnes post-AVC. Néanmoins, des études cliniques contrôlées avec une population plus importante sont nécessaires afin de préciser la pertinence de l'entraînement sur le tapis à DC ainsi que les différentes stratégies musculaires associées aux changements de l'asymétrie de longueur du pas à long terme chez des personnes post-AVC.

Mots-clés : Accident vasculaire cérébral, Adaptation locomotrice, Tapis à double courroie, Longueur de pas, Entraînement

Abstract

Step length asymmetry is a common characteristic of post-stroke gait, but considered as particularly resistant to conventional gait interventions. A recent novel approach using a split-belt treadmill (SBT) showed promising results in short- and long-term reduction of step length asymmetry post-stroke. However, the underlying mechanisms of this intervention and its effects must be better understood before recommending its use in clinical settings. This project aimed to improve our current knowledge about muscular and biomechanical factors contributing to immediate and long-term changes in step length asymmetry in chronic stroke survivors. The main objectives were to: 1) analyze muscle activity associated with changes in step length after walking at unequal belt speeds on a SBT; 2) test the effects of repeated exposure to SBT walking on step length asymmetry and gait ability during walking over ground; and 3) investigate the muscular contributions and strategies involved in these changes. To attain these objectives, a cross-sectional analysis was conducted followed by a pilot training study. The general hypothesis was that distal lower limb muscles are the main contributors to SBT-induced changes in step length asymmetry in chronic stroke survivors using a SBT protocol where two belts were set at unequal speeds with a ratio of 2:1 for a period of time (split-belt configuration). The cross-sectional study analyzed the immediate changes in muscle activity and step length after six minutes of SBT walking in a group of 16 individuals post-stroke and in 10 healthy controls. The findings confirmed that regardless of the side (paretic or non-paretic) walking on the fast belt during split-belt configuration, changes in muscle activity of the ankle plantar- and dorsiflexors were mainly associated with changes in step length symmetry (paper #1). The pilot training study demonstrated that repeated exposure to SBT protocol reduced step length asymmetry and improved walking speed over ground in 12 individuals post-stroke (paper #2). Improvements persisted at least one month post-training. Findings also indicated that from a therapist's viewpoint the training protocol was easy to use and practical in a clinical environment (supplementary data paper #2). Paper #3 showed that these consistent improvements in gait parameters were achieved by a variety of muscular contributions and strategies which involved both lower limbs with a predominant contribution on the side that was trained on the faster belt. Large effect sizes were found in the plantarflexor group during late stance of gait for both net joint moments and muscle activity in the training study. These results combined with the findings of paper #1 indicate that overall, step length asymmetry post-stroke can be successfully reduced with repeated exposure to the tested SBT protocol with distal lower limb muscles appearing to be strong contributors to locomotor adaptation post-stroke and long-term changes in step length asymmetry. Larger control trials are necessary to confirm the relevance of the use of SBT protocols and to further understand the role of the distal lower limb muscles in improvements in step length symmetry post-stroke.

Keywords: Cerebral stroke, Locomotor adaptation, Split-belt treadmill, Step length symmetry, Training

Table of Contents

List of Tables

List of Figures

Chapter 2: General Introduction

<u>Supplementary Results Paper #2:</u>

<u>Paper #3:</u>

Chapter 6: General Discussion

Appendix I:

List of Abbreviations and Glossary of Terms

6MWT	6-Minute Walk Test
10MWT	10 Meter Walk Test
%max	Percentage of the maximum
ACSM	American College of Sports Medicine
ADL	Activities of daily life
AVC	Accident vasculaire cérébral
BBS	Berg Balance Scale
CC	Cross-correlation
CLRC	Constance-Lethbridge Rehabilitation Center
CRLB	Centre de réadaptation Lucie-Bruneau
cm	Centimeter
CMRR	Common Mode Rejection Ratio
CMSA	Chedoke McMaster Stroke Assessment
CPG	Central pattern generators
DC	Double courroie
D	Dominant
EMG	Electromyography
FAC	Functional ambulation category
F_{A-P}	Anteroposterior force
F_{M-L}	Mediolateral force
F_V	Vertical force
GRF	Ground reaction force
GL	Gastrocnemius Lateralis
GM	Gastrocnemius Medius
GLM	Gluteus Medius
GMAX	Gluteus Maximus
H-reflex	Hoffman reflex
HAMS	Hamstrings

HSF	Heart and Stroke Foundation
Hz	Hertz
IRGLM	Institut de réadaptation Gingras-Lindsay de Montréal
ISEK	International Society of Electrophysiology and Kinesiology
L/S	Longer step length / shorter step length
m	Meter
ms	Milliseconds
MEG	Magnetoencephalography
MUR	Muscular utilization ratio
MVC	Maximal voluntary contraction
N	Number
ND	Non-dominant
NP	Non-paretic
NSLD	Normalized step length difference
P	Paretic
p.	Page
RF	Rectus Femoris
RMS	Root-mean-square
Rxy	Correlation coefficient of the cross-correlation analysis
s	Second
SBT	Split-belt treadmill
SL	Step length
SOL	Soleus
ST	Semitendinosus
TA	Tibialis Anterior
TUG	Timed Up and Go test
UMN	Upper motoneuron
VL	Vastus Lateralis
VM	Vastus Medialis
WHO	World Health Organisation

Glossary of terms

NP-fast	NP-fast refers to the condition when the non-paretic leg walked on the belt which was set at fast speeds during the period of adaptation.
P-fast	P-fast refers to the condition when the paretic leg was walking on the belt which was set at fast speeds during the period of adaptation.
Fast-belt	Denotes that the results presented refer to the side that was walking on the faster belt during the period of adaptation and vice versa for the expression "slow-belt".
Fast SL	Refers to the step length of the leg that was walking on the faster belt during the period of adaptation; vice versa for "slow SL"; the same concept applies to all parameters described with "slow" or "fast" (e.g., slow TA, or fast GL).
Longer SL	This expression indicates that the step length of this side was longer compared to the step length of the other side. Vice versa for "shorter SL"; These terms were also used for the equation of the symmetry ratios in papers #2 and #3.

List of Equations

Dedication

À mes parents qui ont toujours crû en moi, qui m'ont soutenue et qui ont toujours été un ilôt de stabilité dans n'importe quelle situation de ma vie et malgré l'océan entre nous.

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Chapter 1. Introduction

1.1 General Problem

Recent statistics estimate around 50 000 new cases of stroke occur each year in Canada, with 315 000 Canadians living with the consequences of a stroke (Heart and Stroke Foundation, 2016). These individuals are left with persistent deficits including muscle weakness (Arene & Hidler, 2009; Heart and Stroke Foundation, 2016), contralesional spasticity (Arene & Hidler, 2009; Watkins et al., 2002), impaired balance control (Hayes, Donnellan, & Stokes, 2016), as well as secondary health complications such as fatigue and depression (Robinson-Smith, Johnston, & Allen, 2000; Van de Port et al., 2006). These deficits reduce the quality of life of stroke patients, impairing the performance of daily living activities such as self-care or walking (Harris & Eng, 2004). The Public Health Agency of Canada reported that 80% of stroke survivors are restricted in their daily activities and 60% are not independent in everyday living (Heart and Stroke Foundation, 2016). For example, the analysis of 205 individuals three years post-stroke in the Netherlands revealed that non-independent stroke survivors for activities of daily life (ADL) have 1.93 times the odds of having declined mobility compared to independent individuals (Van de Port et al., 2006). Evidence suggests that improving walking ability by means of intensive physical training can increase mobility status (Kwakkel et al., 2004) and is a key factor for the ability to "get out and about" in the community (Salbach et al., 2013). Indeed, this is in agreement with the report that 74% (97/115) of stroke survivors report that the ability to "walk in the community" is essential to their quality of life (Lord et al., 2004). Therefore, it is not surprising that improving walking function is considered a major goal for recovery in individuals post-stroke at a chronic stage (Harris & Eng, 2004).

While it is clear that improving walking function is critical for the well-being of stroke survivors, it is difficult to attain. After discharge from rehabilitation, 60.7% of 115 stroke survivors reported that they were able to walk in the community, with 7.6 - 16.9% limited to walking within their immediate environment or requiring supervision for leaving home (Lord et al., 2004). Although many stroke survivors do regain some walking function and independence, their gait capacity is often characterized by reduced walking speed (Ada, Dean, & Lindley, 2013; Mulroy et al., 2003; Perry et al., 1995), endurance (Ada et al., 2013; Dunn et al., 2015) and symmetry (Lauzière et al., 2014b; Patterson et al., 2008a) when compared to unimpaired individuals. Reported mean walking speed in individuals post-stroke ranges from 0.10 - 1.08 m/s (Ada et al., 2013; Perry et al., 1995; Taylor et al., 2006). In terms of endurance post-stroke, individuals can cover a distance ranging from 171 - 318 meters during a six-minute walk test (Dunn et al., 2015). Unfortunately, these values are below the suggested speed (0.9 - 1.2 m/s) and distance (600 m) required for full community ambulation, including passing crosswalks or going to shopping malls (Andrews et al., 2010; Salbach et al., 2013). Moreover, around 60% of individuals post-stroke walk with temporal asymmetries (stance time, double support and swing time) and about 33.3% have interlimb differences in step length (Patterson et al., 2010a; Patterson et al., 2008a). Such gait asymmetry can have major consequences on the structural organization of the body. In particular, gait asymmetry has been associated with a loss of bone density, increased neuromuscular pain on the non-paretic limb (Jorgensen et al., 2000), greater challenges in balance control during walking (De Bujanda et al., 2003) and gait inefficiencies (Detrembleur et al., 2003; Ellis, Howard, & Kram, 2013). Beyond the physical consequences of gait asymmetry, it is important to the psychological well-being of stroke patients, as regaining a “normal” walking pattern is “important” to “very important” for individuals post-stroke (Bohannon, Horton, & Wikholm, 1991).

There are multiple interventions that have been developed to improve walking function in stroke patients. In particular, strong evidence exists for treadmill interventions (with or without body-weight support) in improving walking speed and endurance in individuals with chronic stroke (Macko et al., 1997; Mehrholz, Pohl, & Elsner, 2014; Patterson et al., 2008b; Pohl et al., 2002; Polese et al., 2013). In addition, the use of auditory feedback and increasing walking speeds have resulted in successful improvements in temporal aspects of symmetries such as double stance or cycle duration (Afzal et al., 2015; Cha, Kim, & Chung, 2014; Lamontagne & Fung, 2004; Yen, Schmit, & Wu, 2015). However, these interventions did not improve spatial features of gait such as step length. Indeed, step length asymmetry tends to be more resistant to physical interventions compared to other gait parameters and functional outcomes. A longitudinal evaluation of individuals post-stroke showed that only 14% (5/35) of the patients improved step length symmetry, compared to 30% and 62% who improved gains in walking speed and balance, respectively (Patterson et al., 2015).

In an attempt to address the lack of successful interventions for step length symmetry, the development of a novel and unconventional therapeutic approach consisting of the use of split-belt treadmill (SBT) walking led to a reduction of step length asymmetry which was maintained over three months post-training in 12/12 individuals with chronic stroke (Reisman et al., 2013). The mechanisms underlying split-belt induced improvements in step length symmetry have been studied with cross-sectional investigations using an error-augmentation strategy (Choi & Bastian, 2007; Malone & Bastian, 2010; 2013; Ramas et al., 2007; Reisman, Block, & Bastian, 2005). Essentially, the SBT disrupts the gait pattern by increasing the speed of one of the belts relative to the other. These unequal belt speeds initially lead to an asymmetry (error) in different gait parameters. This creates a need for the gait pattern to be corrected, in order to overcome and adjust to the new walking demand. As a consequence, the patient's gait pattern becomes more symmetrical through adaptation of the interlimb gait parameters such as step length. When the belts are set back at equal speeds (ratio 1:1) after-effects can be observed where the individuals continue to walk with their newly adapted step lengths for some steps. Whereas in healthy individuals these after-effects are characterized by an asymmetrical step length, individuals post-stroke with initial step length asymmetry display a reduced asymmetry.

In the last decade, the response to SBT walking has been studied extensively in both healthy controls and individuals post-stroke by using spatiotemporal gait parameters (Dietz, Zijlstra, & Duysens, 1994a; Lauzière et al., 2014a; 2016; MacLellan et al., 2014; Malone & Bastian, 2013; Ogawa et al., 2014; Reisman et al., 2005; Reisman et al., 2007; Vasudevan & Bastian, 2010). However, only a few of these studies have assessed biomechanical parameters that could explain the observed changes in spatiotemporal parameters (Dietz et al., 1994a; Lauzière et al., 2014a; 2016; MacLellan et al., 2014; Ogawa et al., 2014). For example, the analysis of ground reaction forces (GRF) and muscle activity led to the hypothesis that ankle stiffening is a major control strategy used to adjust the gait pattern observed during SBT adaptation in healthy individuals (Ogawa et al., 2014). Recently, lower limb joint moments and effort after SBT walking were studied in individuals post-stroke (Lauzière et al., 2014a). The results of this study demonstrated that plantarflexion moments play a major role in the SBT-induced step length after-effects in both healthy controls and individuals post-stroke. These results suggest that improving plantarflexor muscle function could facilitate improvements in step length symmetry post-stroke.

Unfortunately, to date lower limb muscle activity in individuals post-stroke has not yet been studied in relation to SBT-induced spatiotemporal changes. This is surprising, considering that impaired muscle function is among the most common deficits following a stroke (Arene & Hidler, 2009; English et al., 2010; Hafer-Macko et al., 2008), with a large impact on gait symmetry, speed and performance (Daly et al., 2011; Hall et al., 2011; Hsu, Tang, & Jan, 2003; Lin et al., 2006). Furthermore, the effects of repeated exposure to SBT walking on muscle activity, joint moments, and other biomechanical aspects have not yet been investigated. As such, the analysis of lower limb muscle activity and biomechanical parameters could contribute to a greater understanding of the mechanisms underlying SBT-induced improvements in individuals who have suffered a cerebral stroke. Furthermore, such analyses could shed light on the strategies used by individuals post-stroke to change their step length symmetry over the long term and how these strategies are related to deficits in muscle function. From a clinical point of view, knowledge about long-term changes in muscle function can facilitate the development of specific and evidence-based treatment approaches targeting improvements in step length symmetry. The results on SBT-induced short- and long-term reduction of step length asymmetry post-stroke are promising. However, the marginal knowledge about the underlying contribution of lower limb muscles affected by the stroke emphasizes the need for more research on this topic.

The main goal of this doctoral thesis was two-fold: 1) to investigate the short-term changes in muscle activity during SBT walking in individuals post-stroke and its associations with the modification of the gait pattern induced by SBT walking; 2) to analyze the effect on gait biomechanics and muscle activity during repeated SBT walking with the aim of improving step length asymmetry in individuals post-stroke.

1.2 Organisation of the Thesis

The thesis begins with a review of the literature before presenting the general hypotheses and objectives. The literature review starts with an overview of the characteristics and biomechanics of gait in healthy individuals to provide a reference for the interpretation of stroke-induced impairments in gait. The characteristics of post-stroke gait are described in a subsequent section which begins with a short overview of epidemiological facts about cerebral stroke. This is followed by a description of the main deficits and disorders relevant for the understanding of impairments in gait post-stroke. A main focus is set on impaired muscle function secondary to a cerebral stroke. Furthermore, studied changes in gait are presented within the framework of our current understandings of error-augmentation theories. The literature review section is followed by the objectives of the present work in line with the three scientific papers included in this thesis. The subsequent method section provides a description of the participants and methods used to approach the objectives. The first paper presents the results of a cross-sectional study on the effects of SBT walking on muscle activity in individuals post-stroke (paper #1). The second (paper #2) and third (paper #3) papers present the clinical and biomechanical changes of a repeated exposure to SBT walking on individuals post-stroke. Finally, in the general discussion section, the results of the three papers are synthesized and their impact discussed relative to what has been shown in the literature. Supplementary results are presented at the end of the corresponding papers. The appendices include a detailed description of the electromyography (EMG) treatment and analysis (Appendix I), questionnaires used in paper #2 (Appendix II) and documents regarding study ethics (e.g., consent forms) (Appendix III - VII) and published abstracts (Appendix VIII).

Chapter 2. Literature Review

2.1 Locomotion in Healthy Individuals

Walking is a complex task requiring the coordination of trunk with upper and lower limbs (Dietz, 2002) that needs to be constantly adapted to the demands of the environment (Lam, Dietz, & Anderschitz, 2006; McFadyen & Carnahan, 1997; Patla et al., 1991) in order to provide controlled forward progression of the body without falling. Smooth progression requires the interaction of different neuronal systems controlling the coordination of limb muscles (Dietz, 1992; MacKay-Lyons, 2002) in an energy efficient manner: "*... the human body, if not influenced markedly by internal or external factors, will integrate the motion of the various segments of the body and control the activity of the muscles so that the energy required for each step is minimal*" (Inman, 1966, abstract, p. 1).

After learning how to walk as a child, locomotion becomes automated and requires little thought, such that it is often taken for granted. However, as attested by the complexity of deficits following injuries (such as a cerebral stroke) and the difficulty of its rehabilitation, it is clear that there is a greater complexity to walking. Over decades, research have investigated human gait and succeeded in the definition of its characteristics and the mechanisms involved (Baker, 2007). The use of biomechanical and EMG analysis have contributed to the quantification of these mechanisms as described in the following sections.

2.1.1 Definition

Walking is a dynamic and rhythmic task (Kuo & Donelan, 2010) that can be divided into different gait events (Figure 2-1) which together make up a gait cycle. These events are the initial contact, loading response, mid stance, terminal stance, pre-swing, initial swing, mid-swing and terminal swing as defined, for example, by Perry and colleagues (Perry, 1992; Perry & Burnfield, 2010). During these sub-phases, three main functional tasks are required to provide upright walking and forward progression: 1) shock absorption at weight acceptance during initial contact and loading response, 2) single limb support and equilibrium control during mid stance, energy generation to provide forward acceleration during late stance, and 3) the control of the foot trajectory during swing phases (Winter, McFadyen, & Dickey, 1991;

Winter, 1987). Furthermore, a long-standing and still supported theory suggests that gait kinematics, such as horizontal and vertical displacement of the center of gravity, are regulated with the goal of achieving these subtasks in the most energy efficient manner possible (Kao et al., 2013; Saunders, Inman, & Eberhart, 1953).

When gait characteristics are compared across individuals, the gait cycle must be normalized and converted into a percentage. Initial contact starts at 0% and terminal swing ends at 100% of the cycle. The gait cycle is divided into stance and swing phases. During 'normal' gait, a person spends about 60% of the gait cycle in the stance phase as illustrated in Figure 2-1. At its beginning and its end, both feet touch the floor during the so-called double support phase (20-24% of the gait cycle in total). During the remaining approximately 40%, one leg is in swing (single support of the opposite leg).

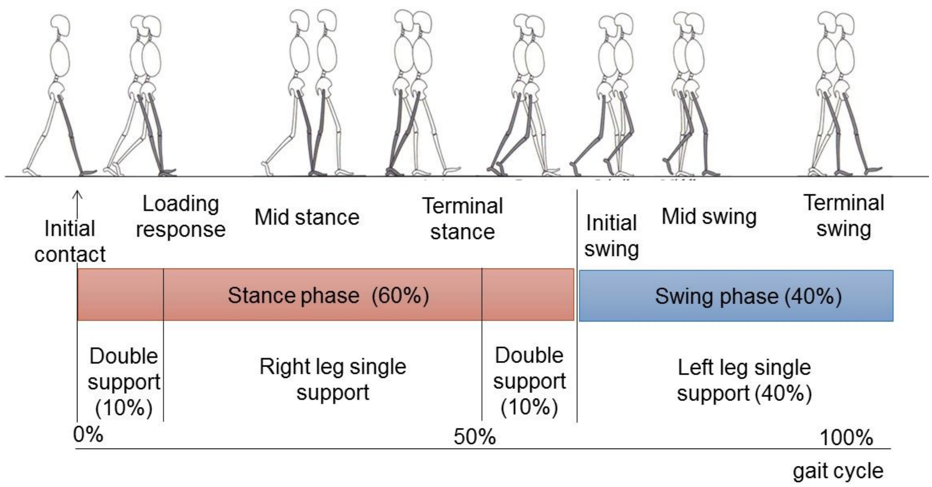

Figure 2-1. Schematic illustration of gait phases in human gait cycle based on the Ranchos Los Amigos terminology: Initial contact and loading response (0 - 10%), mid stance 10 - 30%, terminal stance 30 - 60%, initial swing 60 - 73%, mid 73 - 87% and terminal swing 87 - 100% of the gait cycle (modified from Perry, 1992; SPLER, 2016).

Characteristics of the gait cycle can be quantified by different analysis approaches. For example, the temporal and spatial aspects of the gait pattern can be characterized by its cadence, step length or stance duration (Winter, 2009). In addition, a kinematic analysis can provide a more profound and detailed description of the gait pattern by looking at joint angles, segment displacements, and velocities or accelerations of joint movements during the different phases. Finally, a kinetic analysis can provide a description of the required effort to achieve the task of walking, as it requires the exertion and absorption of force (Nadeau, Betschart, & Bethoux, 2013; Winter, 2009). Each individual's gait cycle contains a basic pattern overlaid with some personal characteristics (Inman, 1966). To a certain degree, individual fluctuations in walking patterns are considered normal and tend to increase with age (Brach et al., 2010). As far as spatiotemporal and functional parameters are concerned, Table 2-1 presents a summary of values for different gait parameters measured during walking over ground from studies including healthy elderly individuals of both genders.

TABLE 2-1. Spatiotemporal and Functional Gait Parameters (Mixed Gender).

Parameter [number of participants studied]	**Range**	**Age (years)**
Step length (cm) [A]	57 - 66	> 60
[n = 241]		
CWS (m/s) [B-D]	1.33 - 1.51	40 - 49
[n = 96 - 941]	1.22 - 1.49	50 - 59
	0.80 - 1.41	60 - 69
Endurance 6MWT (m) [E, F]	607 - 705	48 - 54
[n = 28-35]	563 - 667	55 - 64
	412 - 579	65 - 75
Functional mobility TUG (s) [E]	7 - 9	60 - 69
[n = 281]		

A) (Aboutorabi et al., 2015); B) (Bohannon & Williams Andrews, 2011); C) (Studenski et al., 2011); D) (Salbach et al., 2013); E) (Steffen, Hacker, & Mollinger, 2002); F) (Elazzazi et al., 2012); Abbreviations: cm = centimeter; m = meters; s = second; CWS = comfortable walking speed; 6MWT = 6 Minute Walking Test; TUG = Timed Up and Go test.

The relevant measures for this thesis include step length and walking speed, endurance and functional mobility. Step length symmetry ratios in healthy individuals will be presented separately because of the parameter's particular relevance to this thesis. While it may be assumed that healthy individuals display symmetrical gait patterns, this has been a topic of debate as described in the following section.

2.1.2 Gait Asymmetry in Healthy Individuals

According to Sadeghi et al. (2000) symmetry is defined as "a perfect agreement between the actions of the lower limbs" during walking (p. 35). In more practical terms, this can be defined as the absence of statistical differences when comparing bilaterally measured parameters of gait (Sadeghi et al., 2000). These authors reviewed 25 studies to address the assumption of symmetrical gait in individuals without pathologies. Among the reviewed studies, several differences in step or stride length, foot placement or joint motion can be found. In addition, two studies were reported that found differences between the amplitude of lower limb muscle activity or in the muscular contribution to locomotion (Arsenault, Winter, & Marteniuk, 1986; Ounpuu & Winter, 1989). However, these discrepancies between the gait cycles of each side of the body were considered as a natural and functional consequence of the phenomenon of limb dominance. Furthermore, comparison of gait kinematics of older adults with younger controls revealed that only older adults presented step length asymmetries (Nagano et al., 2013). The asymmetry found in this group of older adults might be explained by the increased asymmetry in lower limb strength found to be associated with ageing (Perry et al., 2007). Therefore, gait asymmetry (spatiotemporal, kinematic or kinetic) can be age-related and not only induced by pathologies. The lack of step length asymmetry found in the younger group was in accordance with the aforementioned findings on muscle activity and joint kinetics (Burnett et al., 2011; Teixeira-Salmela et al., 2008). For example, the analysis of trunk and lower limb muscle activity in 31 healthy young adults during different functional tasks, such as walking, revealed no asymmetry between sides during any task (Burnett et al., 2011). In agreement with this observation, Teixeira-Salmela and colleagues (2008) found no statistical difference in the joint power and mechanical work of lower limbs when comparing between dominant and non-dominant sides.

These functional and natural asymmetries in gait parameters found in older individuals emphasize the use of a range of symmetry rather than talking about the lack of statistical differences between sides. Indeed, the ranges of spatiotemporal symmetry were measured in

81 healthy participants (mean age = 64.2 [SD 22.4] years) during walking over ground for that purpose (Patterson et al., 2010a). With regards to the step length parameter, healthy participants showed a mean value of 1.03 (SD 0.02). The range of "normal" asymmetry was between 0.92 - 1.08 in healthy individuals, where a ratio of "1" indicates perfect symmetry. To perform statistical analyses of these asymmetries, the authors recommended the use of a simple ratio which divides the larger value by the smaller value in order to obtain values always ≥ 1. If the direction of asymmetry (shorter paretic or non-paretic side) is relevant, the use of a ratio with respect to the side was recommended by a recent topical review on gait symmetry (e.g., dominant/non-dominant, paretic/non-paretic) (Lauzière et al., 2014b). Overall, it is clear that to achieve the task of walking and to provide a symmetrical pattern of walking requires complex control of joint biomechanics and muscle activity. The biomechanics and muscle activity underlying the task of walking in healthy individuals will be described in the following section.

2.1.3 Net Joint Moments During Walking

During the stance phase at comfortable walking speed, the ground reaction forces (GRF) quantify the forces exerted on the ground during walking. The vertical component (F_V) represents the vertical acceleration including the weight support, the anteroposterior component ($F_{A\text{-}P}$) is associated with the braking and forward propulsion phases and the third component represents the mediolateral acceleration ($F_{M\text{-}L}$) of the body (John et al., 2012; Kesar et al., 2011; Nadeau et al., 2013). Using kinematic and anthropometric models, these forces can be used to calculate the net joint moments (Winter, 2009).

During walking, the main contribution to support and forward acceleration of the body is within the sagittal plane (Eng & Winter, 1995) and includes the hip, knee and ankle joints (Winter, 1980). At the hip, the sagittal joint moment profile at comfortable walking speed is biphasic and characterized by an extension moment at initial stance (Eng & Winter, 1995). This extension moment subsequently turns into a flexion moment from mid stance to toe off producing a hip flexion to accelerate the leg into swing. An extension moment appears at the end of swing phase contributing to a deceleration of the swinging leg and to prepare for initial contact (Eng & Winter, 1995).

The knee reaches a peak flexion moment in the sagittal plane directly after initial contact. This is associated with “breaking” the forward acceleration of the thighs and the trunk (Kepple, Siegel, & Stanhope, 1997; Zajac, Neptune, & Kautz, 2003). This short flexion moment is followed by an immediate increase in extension moment which contributes to the preservation of an upright position, the transfer of the trunk over the foot, and preventing the knee from collapsing. During mid to late stance, another flexion moment occurs, followed by a short extension moment to accelerate the leg into extension for the swing. During swing, a flexor moment occurs at the end of this phase resulting from the knee flexors which absorb energy to decelerate the swinging leg prior to initial contact (Eng & Winter, 1995).

At the ankle, joint moments occur solely during stance phase. A small extension moment occurs during initial contact which contributes to the controlled lowering of the foot to the ground (Eng & Winter, 1995; Winter, 2009). The subsequent large increase in plantarflexion moment reaches its peak before toe off (terminal stance) which is considered to be the main contributor to forward propulsion of the body (Kepple et al., 1997; Liu et al., 2006; Zajac et al., 2003) and to the ipsilateral and contralateral step length (Allen et al, 2011; Lauzière et al, 2014). The contribution to step length was confirmed by the analysis of plantarflexor net moments and impulses in individuals post-stroke with different step length asymmetries (Allen, Kautz, & Neptune, 2011; Lauzière et al., 2014a). For example, individuals post-stroke and healthy controls both displayed significant relationships between the change in plantarflexion moment and the change of contralateral step length induced by SBT walking (Lauzière et al., 2014a).

The previously described net joint moments are particularly influenced by the muscle activity acting about a joint (Winter, 2009). The muscle activity involved in abled-body gait was studied in the 1940s by the team of Vern Inman in gait analysis of amputees (Sutherland, 2001; Whittle, 1996) and continued over the last decades (Den Otter et al., 2004; Liu et al., 2006; Mann & Hagy, 1980; Shiavi, 1985; Winter, 1987; Zajac et al., 2003). Indeed, a basic, normal pattern of muscle activity in healthy individuals during walking over ground at comfortable speed emerges from this literature. Nonetheless, it has to be taken into account that the pattern of muscle activation during walking can be influenced by several intrinsic factors. These include age, walking velocity (Den Otter et al., 2004; Hortobagyi et al., 2008), postural configuration of the trunk (Forth & Layne, 2008; Grasso, Zago, & Lacquaniti, 2000) and the recording methods (type of electrodes, signal filtering or normalizing techniques, etc.) and analyses used (Chowdhury et al., 2013; De Luca, 1997; Kleissen et al., 1997). These factors explain some inconsistencies in muscle activity patterns observed in healthy individuals presenting similar limb kinematics and walking speed (Clark et al., 2010; Ivanenko, Poppele, & Lacquaniti, 2004; Shiavi, 1985; Winter, 1987).

Muscle activity underlying the task of walking is of particular interest to the current work. Therefore, the tool used to quantify muscle activity in this thesis will be first described and discussed prior to the description of the contribution of lower limb muscle activity during walking in order to facilitate the understanding of the current work.

2.1.4 Electromyography

Surface electromyography (EMG) analysis is considered a reliable and frequently used tool to quantify muscle activity during different functional tasks (Frigo & Crenna, 2009). Furthermore, as described by one of the pioneers in EMG analysis, it is a tool which needs to be handled with care:

> "*Electromyography is a seductive muse because it provides easy access to physiological processes that cause the muscle to generate force, produce movement, and accomplish the countless functions that allow us to interact with the world around us… To its detriment, electromyography is too easy to use and consequently too easy to abuse*." (De Luca, 1997 , p. 3)

2.1.4.1 The Electrophysiology of EMG Signals

Surface EMG is a very common tool used to quantify muscle function during gait (De Luca, 1997; Frigo & Crenna, 2009; Konrad, 2005; Ricamato & Hidler, 2005). In contrast to joint kinetics, EMG signals allow a direct interpretation of muscle function (Basmajian & De Luca, 1985; Frigo & Crenna, 2009). Surface EMG signals are considered to represent anatomical and physiological properties of a muscle (Chowdhury et al., 2013) and a direct analysis of locomotor commands issued to a muscle (Frigo & Crenna, 2009). Each skeletal muscle emanates an electrical signal when it contracts (Basmajian & De Luca, 1985). The quantification of these electrical signals provides a means by which the intensity (amplitude), duration and timing (on- and offset) of muscle activity can be determined. In general, EMG signal treatment consists of a filter procedure which is used to clean the signal from baseline noise and/or amplitude artefacts, signal smoothing to obtain an activation profile of the muscle during a task, and normalization for time and amplitude (De Luca, 1997; De Luca et al., 2010; Konrad, 2005). For details about the actual methods applied in the scope of this doctoral work, we refer to the method sections of the papers, respectively and Appendix I.

2.1.4.2 Interpretation of EMG Signals

In the current thesis we concentrate on the use of EMG for analysis of gait patterns. The use of EMG in the context of gait analysis can provide insight into the appropriate timing of muscle activity and in the identification of deficits of lower limb function. For example, a premature onset of activation was found in the plantarflexor muscle in individuals post-stroke during the stance phase (Knutsson & Richards, 1979; Raja, Neptune, & Kautz, 2012). This abnormal timing was associated with a compensatory strategy to support forward propulsion. Another way to interpret temporal properties about an electrical signal is to look at the frequency content of the signal (Konrad, 2005). An EMG signal contains a specific frequency or power spectrum which illustrates the different frequencies at which a signal has been detected. These frequencies range from 6 - 500 Hz with the most power contained in the 20 - 150 Hz range (Konrad, 2005). The power spectrum can be used as an index of muscle fatigue (De Luca, 1997), since it is considered to shift towards lower frequency bands in cases of muscular fatigue (Basmajian & De Luca, 1985; Moritani, Muro, & Nagata, 1986).

Interpretations of signal amplitude are most frequently done on the peak or mean activation of the muscle. Moreover, activation profiles can be used to analyze muscle synergies during different tasks (Clark et al., 2010; Ivanenko et al., 2004; Safavynia, Torres-Oviedo, & Ting, 2011). To analyse synergies, activation profiles of different muscles that are active in a similar manner during a task are merged to a ‘motor module’ (Clark et al., 2010; Safavynia et al., 2011). This allows insight into the interaction or coordination of different individual muscles during a specific task (Safavynia et al., 2011). This sophisticated method has been used to demonstrate that healthy individuals require up to 2 - 5 different lower limb synergies during walking at comfortable speed (Clark et al., 2010; Ivanenko et al., 2004). Nonetheless, in order to use EMGs in these different analyses, it is important to understand how the EMG signal is treated and processed.

2.1.4.3 Factors Affecting EMG Signals

In terms of signal analysis, the EMG signal is influenced by several factors from its "conception" (i.e., the actual electrical signal produced by the MUAPs) to the rectified and smoothed activation profile used for interpretation. These factors can be separated into extrinsic and intrinsic categories and are considered as having "basic or elemental effect on the signal" (De Luca, 1997, p. 6). In surface EMG, extrinsic factors are mainly associated with the active electrodes, such as their configuration (size, shape), location on the muscle and orientation. For example, a slight deviation in electrode placement between sessions, or its distance to the innervations can alter a signal's amplitude or frequency spectrum (De Luca, 1997). Fortunately, the confounding influence of these factors can be controlled prior to signal recording by using literature-backed standards of electrode positioning and use (Melaku, Kumar, & Bradley, 2001; Merletti, 1999; Rainoldi, Melchiorri, & Caruso, 2004). In contrast to extrinsic factors, intrinsic factors are basically physiological, anatomical and biomechanical characteristics of a muscle and more difficult to control for. They include the number of active motor units, fiber type composition, fiber diameter, blood flow in the muscle, as well as the amount of tissue between the surface of the muscles and electrodes, and condition of the skin. All these intrinsic factors might change impedance of the electrode-skin junction and consequently the signal input. However, skin condition can be controlled with appropriate skin treatment prior the EMG recording (ISEK standards) (Merletti, 1999).

Yet, even when recommendations of appropriate preparation are considered, an EMG signal is always contaminated with unavoidable noise (De Luca, 1997). Thus recommendations for signal filtering should be considered during pre- and offline processing. As far as filtering during pre-processing is concerned, active electrodes are configured to process EMG signals. For further details about our EMG recording system (TeleMyo DTS), the Noraxon Inc. website can be visited. As far as offline processing is concerned, the literature recommends the application of a bandpass filter between 10 - 400Hz (De Luca et al., 2010; Konrad, 2005; Merletti, 1999). However, frequencies up to 20 Hz are considered likely to be contaminated with motion artefacts such that the use of a 20 Hz high-pass frequency filter is also recommended. Overall, most of the signal power can be found within 20 - 200Hz (De Luca et al., 2010).

Furthermore, when intending to compare EMG signals between individuals or across sessions, performing amplitude normalization is highly recommended by the literature (Konrad, 2005; Merletti, 1999) to eliminate bias induced by factors such as electrode placement or skin condition. To normalize for the amplitude, the actual detected EMG signal is expressed relative to a maximal value of EMG activity. The use of a peak value obtained during isometric maximal voluntary contraction (MVC) or a peak or mean value obtained during a reference task (e.g., walking) are valid normalization procedures (Burden, Trew, & Baltzopoulos, 2003). While the isometric MVC procedure has been useful to inform clinicians about the actual extent of muscle activity required during gait, it is not ideal for dynamic EMG analysis in a stroke population considering their deficits in voluntary activation post-stroke. Rather, the normalization of the signal using a peak value from a reference task (i.e., walking) results in the lowest values of inter-individual variability and is considered more appropriate for the normalization of lower limb EMG in individuals post-stroke (Teixeira-Salmela et al., 2012).

2.1.4.4 EMG Activity – Force Relationship

A further aspect to consider when interpreting EMG activity is the relationship between an EMG signal and the actual force produced by the muscle, which depends on several factors (Perry & Bekey, 1981). During isometric contractions, EMG activities of upper and lower limb muscles have a positive, non-linear relationship with the force production (Basmajian & De Luca, 1985). During functional tasks, a negative relationship was found between EMG response and changes in muscle length (Loram, Maganaris, & Lakie, 2005; Lunnen, Yack, & LeVeau, 1981). For example, both an increase in the length of the triceps surae group during standing or an increased length of the biarticular hamstrings during MVC for knee flexion at different hip joint angles led to a reduction in EMG activity of these muscles. However, another study disagreed with these findings, demonstrating that the biarticular hamstring EMG activity remained consistent during MVC for knee flexion at different hip or knee angles (Mohamed, Perry, & Hislop, 2002). However, uniarticular kneeflexors (e.g., short head of the biceps femoris muscle, gracilis muscle) show decreased EMG activity with lengthening (knee extension). The discrepancies between these studies were mainly explained by methodological factors such as differences in force production tasks or electrode types (wire or surface). Nonetheless, such controversy suggests that one must be cautious in interpreting changes in EMG activity during a dynamic task, as the different factors involved in the EMG-force relationship may vary from muscle to muscle and from task to task.

These questions of the EMG-force relationship are of particular importance when investigating evoked EMG during a functional task (Perry & Bekey, 1981). More precisely, during a complex task such as gait, muscles not only contribute to force production in order to generate kinetic energy but also to absorb such energy (Winter, 2009; Zajac et al., 2003). This requires muscles to contract in both a concentric and eccentric manner and at different velocities (Winter & Scott, 1991b; Zajac et al., 2003). Indeed, during constant force production, EMG activity in lower limbs is generally higher during concentric compared to eccentric contractions (Eloranta & Komi, 1980) as confirmed by recent work (Disselhorst-Klug, Schmitz-Rode, & Rau, 2009; Linnamo, Strojnik, & Komi, 2006). However, these studies point out that the EMG activity is strongly dependent on the joint angle and can vary between muscles as observed in the quadriceps components (VL, VM, RF) (Eloranta & Komi, 1980).

In addition, during concentric contraction torque-velocity displays a positive relationship with EMG activity, which is not always the case during eccentric contraction (Perry & Bekey, 1981; Westing, Cresswell, & Thorstensson, 1991). Therefore, during a dynamic task such as walking, information about muscle weakness or strength cannot be directly deduced from EMG activity on its own. Nevertheless, it is a valid, non-invasive tool which provides direct information about muscle function which is frequently impaired in individuals post-stroke (see section 2.3). The following section will describe the contribution of muscle activity to the subtasks involved in human locomotion.

2.1.5 Muscle Activity During Walking

As previously mentioned, muscle activity during walking has been studied over decades. The most frequently studied muscles in gait include: The uniarticular tibialis anterior (TA), soleus (SOL), vastus lateralis (VL) gluteus medius (GLM) and maximus (GMAX), the biarticular gastrocnemius lateralis (GL) and medialis (GM), rectus femoris (RF), biceps femoris (BF), and semitendinosus (ST) (hamstrings: HAMS) (Shiavi, 1985; Winter & Yack, 1987). In particular, TA, GL, VL, BF and ST are considered as excellent muscles for EMG analysis since electrode positioning and determination of innervation zones are straightforward and rapid (Rainoldi et al., 2004). In the following paragraphs, the contributions and activation profiles of these lower limb muscles will be described using both EMG analyses and simulation studies during the different phases of walking. Figure 2-2 illustrates some of the profiles and contributions described.

2.1.5.1 Initial Contact and Early Stance

At heel contact, when weight transfers onto the leg (0-20% of gait cycle), the TA muscles, HAMS, and the glutei group (i.e., GMAX and GLM) reach their peak activity (Liu et al., 2006; Mann & Hagy, 1980; Winter & Yack, 1987). The GMAX is one of the principal muscles contributing to the upright position and forward progression of the body during this phase (Zajac et al., 2003) and during increases in walking speed (Jonkers, Delp, & Patten, 2009; Liu et al., 2006). The quadricep muscles (VL, RF) reach their peak after the TA, HAMS, and GLM with an absorption of the impact force resulting from weight-acceptance (Liu et al., 2006; Zajac et al., 2003) and to prevent the knee from collapsing. The TA, HAMS and GLM all decrease their activity at mid stance (Winter & Yack, 1987).

Recent findings using EMG analysis suggest that the RF is only active during the stance to swing transition and not during early stance in contrast to what has been assumed over decades of research (Nene, Byrne, & Hermens, 2004). In particular, RF EMG activity during the early stance was found with surface electrodes but not with fine wire EMG. Therefore, it has been suggested that the activity measured at early stance in the RF was due to crosstalk with the VL. This has been confirmed by Barr et al. (2010) who demonstrated that RF activity was not present during early stance when measured with fine wire electrodes in 20 healthy individuals across different walking conditions (Barr, Miller, & Chapin, 2010). These findings are perhaps not surprising considering that the muscle bellies from the VL and RF are relatively close. Furthermore, RF muscles have low signal quality when using surface EMG (Rainoldi et al., 2004) as well as a large cycle to cycle variability relative to the TA and GL muscles (Kleissen et al., 1997; Rainoldi et al., 2004; Winter & Yack, 1987).

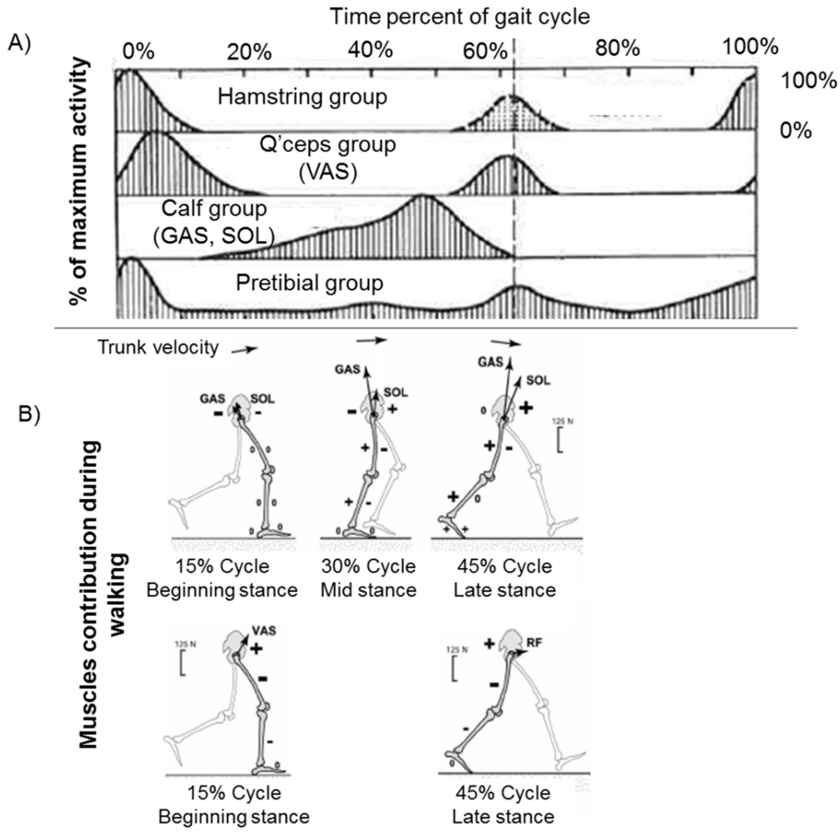

Figure 2-2. Illustration of muscle activity during gait. A) Muscle activity profiles during one gait cycle (0 - 100%). B) Muscles contributing to acceleration (+) or deceleration (-) of a segment. Abbreviations: Q'ceps = Quadriceps; VAS = vastus lateralis; GAS = gastrocnemius lateralis; SOL = Soleus. (A) Modified from Basmajian & De Luca (1985); (B) Modified from Zajac, Neptune & Kautz (2003).

2.1.5.2 Mid Stance to Swing Initiation

At mid stance, the concentric work of the quadriceps muscle group contributes to trunk acceleration and forward progression during stance (Liu et al., 2006; Zajac et al., 2003).

During late stance, the RF muscles support the acceleration into knee and hip extension by contracting eccentrically (Zajac et al., 2003). This in turn assists with the forward acceleration of the trunk. Furthermore, according to Liu et al. (2006), the GLM supports upright position during mid stance and forward acceleration during late stance.

Other important contributors to forward progression are the plantarflexor muscles (GL, GM, SOL) whose peak activation occurs at late stance (about 50% of gait cycle) (Winter & Yack, 1987). The SOL is mainly responsible for body support via energy transfer to the trunk and forward motion of the body (Liu et al., 2006; Zajac, Neptune, & Kautz, 2002). The GL and GM mainly deliver energy to the leg, driving its forward acceleration. Accurate foot clearance is provided by the support of the dorsiflexors (Shiavi, 1985). As such, it has been suggested that the energy transferred to the ipsilateral leg by these muscles influences ipsilateral step length. Along a similar vein, it has been suggested that the contribution of the plantarflexors to forward acceleration influences contralateral step and increased walking speed with the support of hip flexors (e.g., RF) and extensors (e.g., GMAX) (Balasubramanian et al., 2007; Nadeau et al., 1999; Neptune, Sasaki, & Kautz, 2008; Requiao et al., 2005).

2.1.5.3 Mid and Terminal Swing

At swing, the knee and hip flexors increase their activity to control the forward swinging of the leg and prepare it for foot contact (Winter & Yack, 1987; Zajac et al., 2003). On a distal level, the deceleration of the foot is controlled by the TA, which is active particularly from the end of the swing phase until to the subsequent initial contact of the foot with the ground (Shiavi, 1985). It has also been suggested that the TA is active during the mid swing, however this observation is highly variable and sometimes absent (Shiavi, 1985). These muscular contributions to the forward progression of gait are symmetrical between sides in healthy younger individuals (Burnett et al., 2011; Teixeira-Salmela et al., 2008) but appear to deviate from symmetry with age as described in section 2.1.2. Furthermore, lesions of the central nervous system can drastically alter the muscle function and biomechanics involved in the task of walking. As introduced in chapter 1, a cerebral stroke frequently leads to impaired muscle function and gait deficits which are often related to striking limitations in the ability to perform daily life activities. The following section will provide insight into the consequences of a cerebral stroke with particular emphasis on changes in muscle function and gait abilities.

2.2 Cerebral Stroke: Overview

A cerebral stroke is a sudden loss of brain function caused by cell death due to a rupture of vessels in the brain (hemorrhagic) or an interruption of blood flow (ischemic) to the brain cells (Bamford et al., 1990). Like every other organ, the cells in the brain (neurons, glia cells, etc.) need blood to function and stay alive. The consequences of a stroke depend on several factors, including the type of stroke (i.e., hemorrhagic, ischemic), the brain region affected and the size of the area damaged (Heart and Stroke Foundation, 2016). Based on the World Health Organisation (WHO), the most important behavioral risk factors for stroke include tobacco, an unhealthy diet, the use of alcohol, and physical inactivity (World Health Organization, 2016). Stroke is among the leading causes of long-term impairments in North America (Heart and Stroke Foundation, 2016; Truelsen et al., 2006; World Health Organization). Based on recently updated WHO statistics, the number of new cases (incidence) is decreasing in developed countries due to better control of risk factors (World Health Organization, 2011). However, stroke prevalence has been increasing exponentially throughout the years (Truelsen et al., 2006; World Health Organization, 2011). Statistics suggest that this increase is due to two main factors, namely: 1) an increase in the aging population; and 2) an increased number of younger individuals becoming affected (Heart and Stroke Foundation, 2016; Kissela et al., 2012). Worldwide, cerebral strokes have left five million people permanently disabled. In Canada alone, 317,500 people live with the consequences of a stroke. As a consequence, it is not surprising that a considerable proportion of all costs of long-term care and rehabilitative nursing are oriented towards stroke patients. For example, in the Netherlands, stroke is the second most costly disease in the elderly (World Health Organization, 2011). An epidemiological review showed that lifetime costs for individuals post-stroke in the United States can be upwards of $90,000 US, with annual costs ranging from $16,000 to 26,000 US. These estimates are similar to the costs for treating stroke in New Zealand and Europe, where inpatient rehabilitation is considered the most expensive part of the total costs (37%) (Mahler et al., 2008). Taken together, the healthcare burden imposed by stroke highlights the critical importance of investigating more cost- and time-efficient rehabilitation strategies.

2.3 The Functional Consequences and Burden of a Stroke

The functional consequences of a cerebral stroke depend on the location of the stroke and the size of induced damage as aforemetioned (Heart and Stroke Foundation, 2016). Both the location and the size of the lesion have a consequence on observed deficits, with larger lesions producing greater deficits. For example, a stroke at brain stem level, the "oldest' part of the brain, can produce impairments in basic functions such as chewing, swallowing, speaking, breathing or the control of body temperature. In contrast, strokes in the cerebellum cause deficits in feedback and feedforward control of movement. This leads to difficulties in the control of movement coordination and balance, and is accompanied by dizziness, nausea or even vomiting. Nonetheless, most strokes affect the cerebrum, the "youngest part of the brain" (Bamford et al., 1990). Strokes located in the cerebrum can cause sensorimotor deficits such as difficulty or inability to grasp, speak, read, do math, learn, and/or remember new information. Of particular interest to the current thesis, are the regions in the cerebrum known as the motor and sensorimotor cortices responsible for movement control (Heart and Stroke Foundation, 2016). When this region becomes affected due to stroke, it often induces an upper motoneuron (UMN) lesion of the nervous system (Brown, Cannon, & Rowland, 2013). The role of the UMNs is to send motor commands to motoneurons in the spinal cord that control the muscles in order to control movement. Hence, when a stroke affects the motor cortex, it results in deficits such as muscle weakness, spasticity and overactive excitability to tendon reflexes (Adams, Gandevia, & Skuse, 1990) particularly on the contralesional side to the damaged hemisphere.

Of particular importance, muscle weakness is considered one of the cardinal symptoms of motoneuronal lesions (Arene & Hidler, 2009; Brown et al., 2013). As it is explored in the current thesis, the quantification of markers such as muscle weakness in the context of lower limb muscle activity and biomechanics in individuals post-stroke can provide insights into the strategies involved in SBT-induced changes of gait pattern.

2.3.1 Muscle Weakness

Muscle weakness is the reduced ability to generate forces of normal magnitude by a voluntarily-induced muscle contraction (Arene & Hidler, 2009). Most often, impairments in force production are quantified during MVC of a muscle or muscle group. The analysis of individuals with an UMN lesion due to a cerebral stroke showed that the maximum strength of the paretic side (contralesional) was between 24% to 94% of the maximum strength on the non-paretic side (ipsilesional) in ankle, knee and hip flexors and extensors (Adams et al., 1990; Bowden, Taylor, & McNulty, 2014; Chou et al., 2013; Knarr, Zeni, & Higginson, 2012). Furthermore, individuals post-stroke also displayed a slight reduction in maximal strength for lower limb muscles on the non-paretic side when compared to healthy individuals (Adams et al., 1990). The analysis of maximal static strength in 48 individuals post-stroke revealed even lower values of 19.8 - 33.9% on the paretic and 60.1 - 89.5% on the non-paretic side when compared to healthy controls (Andrews & Bohannon, 2000). Such muscle weakness was found to have a negative impact on important daily functions such as walking. Indeed, slow walking is associated with weakness of the plantarflexors (Kim & Eng, 2004; Nadeau et al., 1999), hip and knee power generation and absorption during walking (Kim & Eng, 2004). Furthermore, dorsiflexor strength appears to be the main contributor to temporal symmetry ($R^2 = 0.36$) and smaller plantarflexion joint moment impulses during the stance phase of gait has been linked to larger asymmetry in step length (Allen et al., 2011; Balasubramanian et al., 2007).

While these parameters of force, power generation, and impulse can be used to quantify force production of a muscle or muscle group, they do not inform directly about deficits in muscle function. This limitation requires using other techniques, such as EMG (see section 2.1.4. for technical details), to better assess stroke-induced muscular deficits. In humans, the analysis of muscle activity using EMG is considered as the sole opportunity for a direct analysis of movement commands issued to an individual muscle that is associated with a particular movement (Basmajian & De Luca, 1985; Frigo & Crenna, 2009). In this thesis the analysis of muscle activity in individuals post stroke is of particular interest, considering that the aforementioned stroke-induced lesions in the motor- and sensorimotor cortex affect the peripheral nerves and muscles involved in movement. The following sections provide an overview of the stroke-induced changes in muscle activity.

2.3.2 Altered Muscle Activity

Electromyography analyses can be used to explore how muscle activity is functionally altered due to stroke. During MVC of a muscle group, EMG analyses have revealed that individuals post-stroke present deficits in muscle activation. Over decades researchers have agreed that after a cerebral stroke, muscle activity and force production are weaker particularly on the paretic side (Bourbonnais & Vanden Noven, 1989; Chou et al., 2013; Horstman et al., 2008; Klein et al., 2010). For example, during a constant increase of force production (0% to 100%), the rate of EMG amplitude increase, quantified with the root-mean-square (RMS), of the paretic dorsiflexor is less compared to the non-paretic side (Chou et al., 2013). Furthermore, during MVC, normalized EMG (RMS/max M-wave) is reduced bilaterally on upper and lower limb muscles in individuals post-stroke compared to healthy individuals, particularly on the paretic side (Bowden et al., 2014; Klein et al., 2010).

2.3.2.1 Neuromuscular Control

The decreased activation of muscles on the paretic side during voluntary contraction observed in EMG activity points to reduced voluntary activation of those muscles, i.e. deficits in neuromuscular control (Bowden et al., 2014; Horstman et al., 2008; Miller, Flansbjer, & Lexell, 2009). The level of voluntary activation of a muscle is measured by comparing the torque produced during MVC with and without stimulation of the peripheral nerve or the motor cortex (Beaulieu et al., 2014; Bowden et al., 2014). These tests showed that reduced voluntary activation could be due to several different factors working in isolation or together: a reduction in recruitment of motoneurons, changes in neural firing rate, reduced connectivity in the descending corticospinal tracts, and/or a reduced ability to regulate descending neuronal input to activate skeletal muscles (Beaulieu et al., 2014; Bowden et al., 2014). The reduction in voluntary contraction observed in the lower limb muscles after stroke appears to be moderately to strongly correlated with walking speed (10 Meter Walking Test), functional capacity (Timed Up and Go test, TUG), independency (Functional Ambulation Categories, FAC), and dynamic balance (Berg Balance Scale, BBS) (Horstman et al., 2008; Klein et al., 2010).

Beyond studies on voluntary activation, the analyses of spinal reflex modulations have also been insightful. The analysis of spinal reflex modulations involves peripheral nerve stimulation evoking an H-reflex response and activation of motoneurons (Christie & Kamen, 2006; Wolpert, Pearson, & Ghez, 2013). The H-reflex (Hoffman reflex) is the electrical equivalent to a stretch reflex and detected with EMG. The excitability of a motoneuron is dependent on descending inhibitor and excitatory signals as well as their modulation through spinal interneurons (Knikou, 2008; Wolpert et al., 2013). As such, hypo- or hyperactive stretch reflexes point to lesions of the central nervous system. In particular, such studies have found that there is an abnormal regulation of the descending pathways and impaired processing on a spinal level in individuals post-stroke (Bowden et al., 2014; Finley, Perreault, & Dhaher, 2008; Horstman et al., 2008; Li & Francisco, 2015; Miller et al., 2009). The impaired regulation of descending pathways is thought to arise from diminished activation of the premotor and primary motor cortex cells (Bergfeldt et al., 2015) and reduced inhibitory control of corticospinal and reticulospinal pathways (Li & Francisco, 2015). The latter was considered to contribute particularly to spasticity. On the spinal level, impaired processing and modulation has been explained mainly by a reduced inhibition of afferent information. In particular, individuals with chronic stroke showed a reduced presynaptic inhibition of Ia afferents, with reduced reciprocal inhibition, but increased facilitation in group Ib and group II afferents (Aymard et al., 2000; Dyer et al., 2009; Dyer et al., 2014; Li & Francisco, 2015). Other studies also point to increased afferent input to the alpha-motoneurons (Li & Francisco, 2015) and reduced motoneuron coupling between motor unit activities (Nielsen et al., 2008).

These impairments in muscle activity can have functional consequences on activities critical for daily life such as walking. The following section describes the main characteristics observed in post-stroke gait and the modifications in muscle activity and biomechanics that have been found to be related to post-stroke gait. A particular focus will be set on the parameter of interest in this work, which is step length asymmetry post-stroke.

2.4 Locomotion After Stroke

Cerebral strokes often lead to substantial deficits in walking ability as mentioned in the introduction section of this thesis. The probability of regaining independent walking function three months after a cerebral stroke is about 60% in individuals who were non-ambulatory in the first month of rehabilitation (Preston et al., 2011). As mentioned in the introduction, even among the individuals who regained walking capacity, a substantial number lived with persisting gait deficits (Ada et al., 2009; Kollen et al., 2005; Lord et al., 2004; Patterson et al., 2010b; Patterson et al., 2015). Reduced walking speed (Ada et al., 2009; Lord et al., 2004; Mulroy et al., 2003), endurance (Dunn et al., 2015; Eng et al., 2002) and gait asymmetry (Patterson et al., 2010b; Patterson et al., 2008a) are frequently reported characteristics of post-stroke gait. Walking speed over ground in individuals post-stroke ranges from 0.1 - 1.08 m/s (Ada et al., 2013; Ada et al., 2009; Perry et al., 1995; Taylor et al., 2006) compared to 0.80 - 1.40 m/s for healthy older individuals (60 - 69 years) with a good life expectancy (Bohannon & Williams Andrews, 2011; Salbach et al., 2013; Studenski et al., 2011). Deficits in walking endurance have been quantified using walking distance during a 6-Minute Walking test (6MWT). The distance covered by individuals with chronic stroke ranged from 171 - 318 m (Dunn et al., 2015; Lord et al., 2004). This range is below the distance (600 m) required for full community ambulation including traversing crosswalks or going to shopping malls (Andrews et al., 2010; Salbach et al., 2013).

As discussed earlier (see section 2.3), impairments in gait ability (speed, symmetry, endurance) following stroke are likely due to impairments in the neuromuscular system and muscle function due to neurological damage. This has been supported by the modifications in EMG activity observed during walking in individuals post-stroke which is discussed next.

2.4.1 Muscle Activity in Walking Post-stroke

Several studies have explored the changes observed in muscle activity during walking following cerebral stroke damage. Impaired muscle activity in post-stroke gait has been described in terms of 1) alterations in amplitude of muscle activity (Knutsson & Richards, 1979; Mulroy et al., 2003) and 2) alterations in timing of activity with respect to the gait cycle (Buurke et al., 2008; Den Otter et al., 2007; Knutsson & Richards, 1979). In general, compared with the non-paretic side or healthy controls, there is reduced amplitude of activity in paretic plantarflexors, knee extensors (Harris et al., 2001; Hesse, Konrad, & Uhlenbrock, 1999; Knutsson & Richards, 1979; Lamontagne et al., 2002; Lamontagne, Richards, & Malouin, 1998), and hip extensors (De Quervain et al., 1996; Harris-Love et al., 2004; Knutsson & Richards, 1979). Such reduction in activity is compensated for by prolonged quadriceps activation and with a more flexed knee, hip and trunk position to provide body support during single stance in individuals walking at slow speeds (Mulroy et al., 2003). Furthermore, individuals post-stroke frequently presented deficits in timing of muscle activity. Early onset of activity can be observed in the paretic gastrocnemius muscle during the terminal swing and initial stance phases of gait (Arene & Hidler, 2009; Knutsson & Richards, 1979). In addition, a prolonged duration of activity can be found in the paretic gastrocnemius muscle (GM and GL) during the double support phase, and in the paretic biceps femoris (BF) and rectus femoris (RF) muscles during single support compared with healthy controls (Chow, Yablon, & Stokic, 2012; Den Otter et al., 2007). In contrast, an absence or strong reduction of muscle activity has often been observed in the distal paretic dorsiflexor muscle – the tibialis anterior (TA) – during the late swing or early stance phases (Burridge et al., 2001; Perry et al., 1995).

The stroke-induced changes in activation onset and duration of muscle activity can result in a modulation of the coactivation duration between antagonist muscles. Indeed, longer coactivation duration of proximal limb muscles (BF, RF) was found bilaterally in individuals with chronic stroke (Den Otter et al., 2006; Den Otter et al., 2007; Detrembleur et al., 2003). However, results are less conclusive for the distal lower limb muscles, including the ankle antagonists TA versus GL and GM. According to Den Otter et al. (2006), paretic ankle coactivation duration did not differ from non-paretic duration. In contrast, the 11 individuals post-stroke studied by Chow et al. (2012) presented a shorter duration of antagonist coactivation during stance phase of gait when compared to the non-affected side. However, when compared to healthy controls, duration as well as amplitude of coactivation were generally found to be longer for both the paretic and non-paretic sides. Furthermore, slower walking in individuals early after stroke was associated with longer coactivation durations between ankle antagonists (Detrembleur et al., 2003; Lamontagne, Richards, & Malouin, 2000). It has been suggested that these changes reflect an adaptation to postural instability (Lamontagne et al., 2000).

Some of the modifications in muscle activity observed in individuals post-stroke that lead to different muscular contributions to walking can be due to compensatory strategies. For example, a two-dimensional simulation method provided insight into the contribution of muscle activity to the stance phase in individuals post-stroke when compared to healthy controls walking at slow speed (Higginson et al., 2006). Non-paretic and particularly paretic plantarflexors showed a reduced contribution to body support at late stance in comparison to healthy controls. This was compensated for by the paretic hip flexors and extensors (RF, HAMS, and GMAX) which provided larger contributions to body support compared to the non-paretic side. These findings were in accordance with the analysis of joint kinetics (net joint moments, moment impulse and muscular effort) in post-stroke gait. Hip flexors showed an increased contribution to body support or forward propulsion when walking at faster speeds (Milot, Nadeau, & Gravel, 2007; Nadeau et al., 1999) in individuals post-stroke when compared to healthy controls. Hip extensors were found to be more involved in post-stroke participants with plantarflexor weakness in order to provide symmetry in step length during walking (Allen et al., 2011).

2.4.1.1 Muscle Activity and Neuromuscular Deficits

The modifications in EMG activity observed during gait in individuals post-stroke have been associated with spinal and supraspinal changes in the modulation of neural information controlling muscle activity. To explore such changes in neural modulation, evoked reflex responses can be used as indicated in section 2.3.2.1 above. In particular, individuals post-stroke display hyperexcitable stretch reflexes on the heteronymous spinal pathways between knee and ankle extensors (Dyer et al., 2009; Dyer et al., 2014) and hip and knee muscles (Lewek et al., 2007). For example, femoral nerve stimulation has been shown to facilitate the soleus H-reflex in individuals post-stroke compared to healthy controls. The increased facilitation was associated with an increased coactivation of knee and ankle extensors during the stance phase of gait ($r = -0.62$ to -0.73, $p < 0.05$).

Individuals post-stroke also present a larger and velocity-dependent reflex response of the quadriceps muscle to imposed hip extension when compared to healthy controls (Finley et al., 2008; Lewek et al., 2007). For example, the increased response was in line with a prolonged sustained activity of the quadriceps muscle (Lewek et al., 2007). This prolonged response has a moderate to strong negative correlation (Pearson product moment = -0.70 to -0.79) with swing-phase peak knee flexion and velocity of knee flexion and hip extension (Lewek et al., 2007). Based on these findings, it has been suggested that the prolonged quadriceps activity is associated with "spastic stiffed-leg" gait. However, one limitation of the study by Lewek et al. (2007) is that the reflex response was tested in the supine, unloaded position. During walking we experience the load of body weight, which influences the output of EMG in healthy individuals (Dietz, 2002; Kuo & Donelan, 2010). In addition, the reflex responses of lower limb muscles are strongly dependent on the position of the subject or the task they are performing (Bove et al., 2006; Faist, Dietz, & Pierrot-Deseilligny, 1996).

2.4.1.2 EMG and Gait Parameters Relationships

The relationship between muscle activity and gait parameters post-stroke is currently unclear, with findings being inconsistent in the literature. Some studies have demonstrated robust relationships between these variables. For example, self-induced increases in walking speed over ground results in an overall increase of activity of the four lower limb muscles (GM, TA, ST, and RF) as assessed in a group of individuals with chronic stroke (Lamontagne & Fung, 2004). In contrast other studies have not found clear relationships between EMG and gait. According to Den Otter et al. (2006), improvements in gait capacity (speed, temporal symmetry, independence) were not associated with a 'normalization' of temporal patterning in muscle activity during gait post-stroke. This discrepancy could be explained with the separate contribution of spatial (amplitude) and temporal features to the increase in speed. In healthy individuals the increase or decrease in speed led to an increase in amplitude of muscle activity but presented marginal changes in timing (Den Otter et al., 2004; Hof et al., 2002; Nymark et al., 2005).

Furthermore, individuals post-stroke used a variety of strategies to achieve an increase in speed (Jonkers et al., 2009; Jonsdottir et al., 2009). For example, in the study of Jonsdottir et al. (2009), two thirds of participants depended on paretic hip work as a major resource whereas the remaining participants increased work at the level of the paretic ankle. In addition, a lack of association was found between muscle activity and the gait pattern itself (De Quervain et al., 1996). Early findings showed that the walking pattern (hyper-extension of the knee, or flexed knee) was not related to a particular pattern in EMG activity of the lower limbs (Buurke et al., 2008; De Quervain et al., 1996). As explained in section 2.1.4, the inconsistency in these findings is probably due to the fact that EMG activity does not directly explain changes in power production, i.e. work. Furthermore, given the use of different compensatory strategies to achieve the task of walking in individuals post-stroke, it is reasonable to expect heterogeneity of EMG activity across subjects despite similar modifications of spatiotemporal gait parameters (Allen et al., 2011; Higginson et al., 2006; Jonkers et al., 2009; Jonsdottir et al., 2009). It is also possible that some gait parameters may be more related to a particular pattern of EMG activity than others. Of particular importance to the current thesis, to date it remains unclear if changes in step length and improvements in step length asymmetry post-stroke underly a particular and consistent pattern of EMG activity.

2.4.2 Step Length Asymmetry Post-stroke

The study of post stroke gait asymmetry is currently of great interest in the literature. In contrast to stroke-induced deficits in walking speed or lower limb impairments which tend to improve with time, gait asymmetry tends to worsen (Patterson et al., 2010b). Gait asymmetry is also associated with a loss of bone density (Jorgensen et al., 2000; Marzolini et al., 2014), increased neuromuscular pain (Norvell et al., 2005), decreased balance control during walking (De Bujanda et al., 2003), and higher energy consumption (Awad et al., 2015; Detrembleur et al., 2003; Ellis et al., 2013). Furthermore, step length asymmetry in particular tends to be more resistant to conventional therapy compared to other gait parameters (Patterson et al., 2015). For example, at discharge from inpatient rehabilitation only 14% of the patients improved step length symmetry compared to 30% and 62% of patients improving gait speed and balance, respectively. In addition, while therapeutic tasks such as fast walking over ground (Lamontagne & Fung, 2004), bodyweight supported treadmill walking (Dawes et al., 2008; Hassid et al., 1997), aerobic treadmill training (Silver et al., 2000), swing resistance during walking (Yen et al., 2015), and auditory cueing (Thaut et al., 2007) led to improvements in temporal gait asymmetry, they do not affect spatial gait asymmetry.

According to several investigations, around 33% to 49% of individuals post-stroke present step length asymmetries (Balasubramanian et al., 2007; Finley et al., 2008; Patterson et al., 2010a; Patterson et al., 2010b; Patterson et al., 2008a). Step length asymmetries appear less frequently than temporal asymmetries (60%) which are most obvious during double support duration, swing or stance duration (Patterson et al., 2010a). The majority of individuals with step length asymmetry have a longer step on the paretic side with values reported ranging from 47% - 76% (Balasubramanian et al., 2007; Patterson et al., 2010a). The degree of step length asymmetry, however, varies substantially among individuals (Allen et al., 2011; Patterson et al., 2010a; Patterson et al., 2008a). Furthermore, having an asymmetry in step length does not necessarily indicate an asymmetry in temporal parameters (Lauzière et al., 2014b; Patterson et al., 2010a). For example, the analysis of 161 individuals post-stroke revealed significant but moderate correlations between step length ratio and swing time ratio (r = 0.47) and stance time ratio (r = 0.58) (Patterson et al., 2010a).

There are several deficits, based on correlation analyses that are associated with step length asymmetry. These include isokinetic ankle plantarflexor peak force (r = 0.53) and total work (r = 0.53) (Hsu et al., 2003) as well as ankle plantarflexor spasticity which account for 46% and 53% of the variance in symmetry, respectively (Hsu et al., 2003; Lin et al., 2006). Associations between asymmetry and the Brunnstrom motor recovery stages (r = -0.53, $p < 0.001$) showed that participants with more severe deficits presented higher asymmetries (Balasubramanian et al., 2007). The direction of asymmetry (shorter paretic or non-paretic step) was associated with propulsion of the paretic leg during walking (Balasubramanian et al., 2007). In particular, the step length ratio was negatively correlated with paretic leg propulsion (r = -0.78, $p < 0.001$) indicating that individuals with less paretic propulsive force walked more asymmetrically and with a shorter non-paretic step length compared to the paretic side.

The step length asymmetry observed post-stroke is likely representative of compensatory mechanisms for lower limb deficits. This is supported by studies from Allen et al. (2011), which explored this question using 55 individuals with chronic stroke during walking over ground at comfortable speed. Among these participants, all presented impaired paretic plantarflexor function, quantified with anterior-posterior force impulse (impulse = time integral) and plantarflexor moment impulse. Impulses were reduced when compared to healthy controls. The impaired paretic plantarflexor impulse was compensated for by an increase in bilateral hip flexor or nonparetic plantarflexor and knee extensor moment impulses. The group with the hip flexor strategy presented symmetrical step lengths (n = 17). In contrast, the group using the non-paretic plantarflexors and knee extensors to compensate for deficits presented a longer paretic step length compared to the non-paretic side (n = 28). The participants with shorter paretic step lengths did not present a particular compensation strategy, which may be due to the small number of individuals in that group (n = 9) compared to the number of individuals in the other groups.

More recently, the use of a split-belt treadmill has led to new and promising findings in terms of changing step length and improving its symmetry over the long term (Reisman et al., 2013; Reisman et al., 2007). The following section will present a summary of the current knowledge and findings about the principles and effects of SBT walking on individuals post-stroke and healthy controls.

2.5 The Principle of the SBT Protocol

The use of treadmills to perform walking tasks has provided substantial information about how gait is affected post stroke. While traditional treadmills have been used extensively, more recent explorations of gait asymmetry has led to the increasingly greater use of the split-belt treadmill in therapy and research. The split-belt treadmill has two belts and the speed of each belt can be modified independently. When used with a specific protocol, the SBT can produce step length asymmetry in healthy controls (Reisman et al., 2005). In individuals post-stroke, step length asymmetry is reduced following SBT exposure and tends to remain so after three months post-training (Reisman et al., 2013). These results demonstrate that individuals post-stroke have the capacity to adapt their walking pattern to a perturbed walking condition and store this new gait pattern when the perturbation is removed (Reisman et al., 2007). Furthermore, the repeated exposure led to motor learning with storage of the adapted gait pattern over long-term during walking over ground (Reisman et al., 2013). Cerebellar circuits in the central nervous system are considered mainly responsible for the locomotor adaptation and the storage of new motor output (Morton et al., 2010) which partially explains why individuals with cerebral lesions still have the capacity to adapt their gait pattern. Chapter 2.6.1.6 will provide a more detailed description of the current knowledge on structures involved in locomotor adaptation. The changes observed in step length symmetry during SBT walking resulted from a protocol centered on the principle of error-augmentation, described below. Such error-augmentation protocols have seen extensive use in studies investigating locomotor adaptation (Choi & Bastian, 2007; Dietz et al., 1994a; Jansen et al., 2013; Lauzière et al., 2014a; MacLellan et al., 2014; Malone & Bastian, 2010; Mawase et al., 2013; Ogawa et al., 2014; Prokop et al., 1995; Reisman et al., 2005; Vasudevan & Bastian, 2010).

In our everyday lives, we experience "errors" of movement that need to be corrected to preserve the motor goal. For example, adjusting hand movements to an unfamiliar computer mouse, or adapting leg and trunk movements to walking in ski boots or high heels. In other words, an "error" is induced by an external force, such as changes in the environment or the body (e.g., pain). Instead of constantly correcting our movements to this error, error-feedback enables us to re-calibrate the motor output with each trial of the movement and to adjust it with a feedforward mechanism (Bastian, 2008; Martin et al., 1996). According to Bastian (2008) and Martin (1996) such re-calibration and adjustment of motor output on a "trial-and-error" basis can be defined as motor adaptation when the following criteria are satisfied: 1) the initial action during the original condition A needs to be retained (e.g., walking), but requires changes in one or more parameters (e.g., step length) for the new condition B; 2) the adaptation of these parameters occurs gradually and continuously with repetition of the behaviour; and 3) once adapted to condition B, when subjects are exposed to condition A again, they present after-effects and must gradually de-adapt in a continuous manner to the prior (initial) state of condition A. This suggests that the system still expects condition B, and condition A represents a new "error"-inducing situation. Furthermore, after-effects indicate that the motor output of the adapted gait pattern was stored and retained (Malone et al., 2012; Reisman et al., 2005).

This definition of locomotor adaptation is used, since the literature on SBT-induced adaptation relevant in this work refers to this definition or principle when explaining the theory of their results (Bastian, 2008; Lauzière et al., 2014a; Malone & Bastian, 2013;

Malone, Bastian, & Torres-Oviedo, 2012; Reisman et al., 2005). Nevertheless, for upper extremity, this definition for motor adaptation (tested mainly within target-oriented movements) might not be applicable (Hwang & Shadmehr, 2005; Shadmehr & Mussa-Ivaldi, 1994). Furthermore, the locomotor after-effects depend strongly on the context (Hamzey, Kirk, & Vasudevan, 2016; Malone & Bastian, 2010) as will be discussed in section 2.6.4.

The SBT is an interesting tool to induce error-augmentation during walking as its two belts allow control of the speed of each leg independently. The protocols used in these SBT studies have the following common characteristics (Figure 2-3): 1) participants initially walk with both belts at equal speed (tied-belt configuration; baseline period); 2) this is then followed by a period of adaptation with belts used at unequal speeds (split-belt configuration) and 3) a post-adaptation period where both belts are back to equal speeds (tied-belt configuration) (Choi & Bastian, 2007; Dietz et al., 1994a; Jansen et al., 2013; Lauzière et al., 2014a; MacLellan et al., 2014; Malone & Bastian, 2010; Mawase et al., 2013; Ogawa et al., 2014; Prokop et al., 1995; Reisman et al., 2005; Vasudevan & Bastian, 2010). For the split-belt configuration, the aforementioned studies mostly used pre-defined fixed speeds with a 2:1 or 3:1 ratio meaning that one belt moves twice or three time faster than the other.

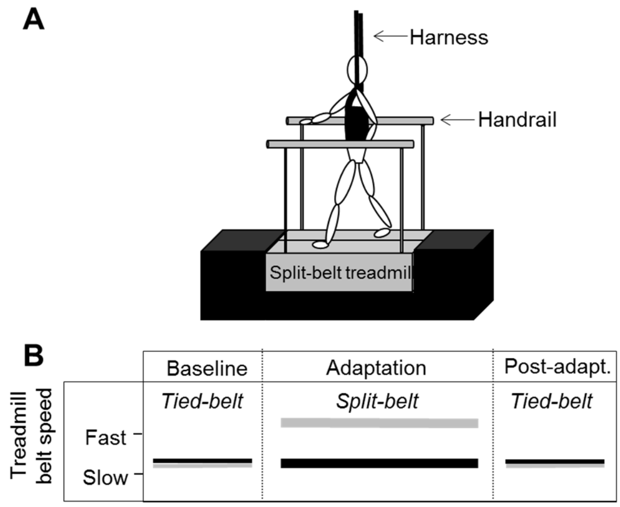

Figure 2-3. A) Schematic representation of a split-belt treadmill (SBT) with handrails and harness. B) Example of a SBT protocol with the tied-belt configuration (Baseline; belts at equal speeds), followed by a split-belt configuration (Adaptation) and finally with the belts at

equal speeds again (tied-belt; Post-adaptation); Modified and adjusted from Lauzière et al. (2014a).

2.6 SBT-induced Adaptation and After-effects in Locomotion

2.6.1 Effects of SBT in Abled-body Locomotion

In general, the results described in this section refer to the findings by Reisman et al. (2005), which are representative of the results from other investigations describing SBT-induced effects on spatiotemporal parameters (Choi & Bastian, 2007; Dietz et al., 1994a; Finley et al., 2008; Lauzière et al., 2014a; Malone et al., 2012; Ogawa et al., 2014; Prokop et al., 1995). In healthy subjects, two types of changes in spatiotemporal parameters were observed during walking with split-belt configuration, fast and immediate changes, and slow and adapting changes (Figure 2-4) (Reisman et al., 2005).Throughout this section the expression "fast leg" refers to the side which was walking on the faster belt during adaptation and vice versa for the "slow leg".

2.6.1.1 Fast Changing Spatiotemporal Parameters

Immediately after switching from a tied- (baseline) to split-belt configuration (adaptation), healthy participants present changes in intralimb parameters (Reisman et al., 2005). More precisely, stride length and swing time increase on the fast leg and decrease on the slow leg, albeit the slow leg maintained the same belt speed. Leg stance time becomes shorter on the fast leg compared to the slow leg. During post-adaptation (tied-belt) these intralimb parameters immediately return to baseline and therefore no after-effects can be observed. Additionally, the comparison of intralimb joint angles (knee angle relative to hip or ankle angle) showed slight changes that did not differ significantly from the beginning to the end of adaptation period. As such, these three parameters do not actually adapt during walking with split-belt configuration and can therefore be described as "reactive changing parameters". These reactive and fast changes can be observed across all tested speed ratios (2:1, 3:1, and 4:1).

2.6.1.2 Slowly Changing Spatiotemporal Parameters

Similar to intralimb parameters, interlimb parameters (i.e., involving both legs) are altered immediately when the belts become set at split-belt configuration (Reisman et al., 2005). However, in contrast to intralimb parameters, they do show adaptation during this period (Figure 2-4). More precisely, at the beginning of the split-belt configuration, the interlimb parameters of step lengths and double support duration are initially shorter on the fast side compared to the slow side. This asymmetry is considered to create an "error-feedback" signal.

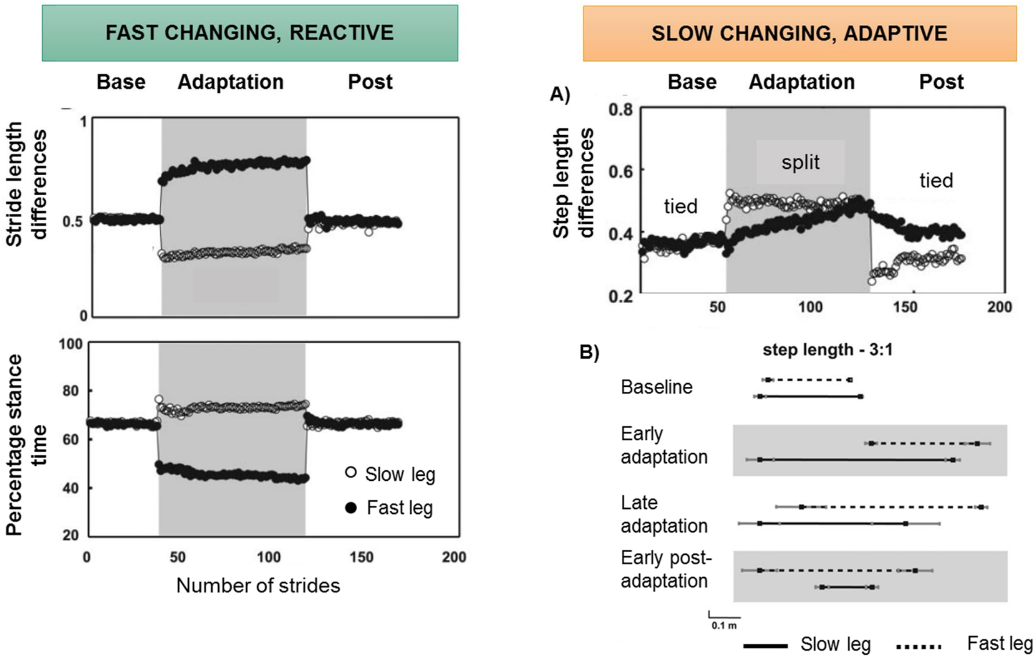

Figure 2-4. Left-hand figures illustrate changes in reactive (stride length and % stance time) parameters. On the right-hand side the adaptive parameter (step length) is shown as symmetry (A) and for each side individually (B). Data is illustrated as differences between limbs. During split-belt configuration, belts had a 3:1-speed ratio. Adapted from Reisman et al. (2005).

The initial asymmetry observed during split-belt configuration gradually (trial-by-trial) returns back towards baseline symmetry regardless of the fact that the belts are still moving at unequal speeds (Reisman et al., 2005). These adjustments are due to changes in both the slow and fast legs. Furthermore, during post-adaptation, when belts are set back to their baseline configuration (tied-belt), these adapted parameters display after-effects. The after-effects can be characterized by a reverse pattern of asymmetry to the pattern observed during the beginning of adaptation. Step length was now larger on the side that walked on the fast belt, when compared to the slow belt. During that period (post-adaptation), the after-effects de-adapted gradually towards baseline symmetry. Finally, the motor asymmetry after effect is accompanied by a perception by participants that the speeds of the split belts were now unequal, even though they were now the same.

2.6.1.3 The Effect of Belt Speeds

The speed of the slower belt and the ratio of belt speeds influenced the magnitude and rate of adaptation. In the Reisman study (2005), the slower belt was kept constant at 0.5 m/s and the faster belt changed between 1.0, 1.5, or 2.0 m/s. Participants showed larger asymmetries and required more strides to adapt at higher belt speed ratios. With regards to observed after-effects, they were larger with a speed ratio of 1:2 compared to a ratio of 2:3 (i.e., the greater the difference between belt speeds, the greater the asymmetry during adaptation and the after-effects in this study). This observation is complemented by the findings of Vasudevan and colleagues (2010), who compared after-effects on abled-body step length at different combinations of belt speeds. Vasudevan et al. (2010) showed that the largest after-effects were found when the speed of the slow belt during adaptation is close to the comfortable speed of the participants (0.7 m/s) compared to faster (1.4 m/s) or slower (0.525 m/s) speeds. To summarize, the largest after-effects appear to be generated when the slow belt speed is closest to the natural speed, and the difference between the slow and fast belt speeds are greatest (ratio 2:1).

2.6.1.4 How Much Time Is Required for Adaptation and After-effects?

The SBT protocols can take anywhere from 6 - 15 minutes to induce adaptation (Lauzière et al., 2014a; Malone & Bastian, 2010; Malone et al., 2012; Reisman et al., 2005; Vasudevan & Bastian, 2010). Adaptation of step symmetry can take up to 200 strides approximately (Malone & Bastian, 2010). Distractions occurring during the dual task can slow down the adaptation (i.e., making it take longer) whereas conscious correction and effort can speed up the adaptation by about 100 strides (Malone & Bastian, 2010). Similar effects can be observed during the after-effects, but with de-adaptation (washout) taking less time in general compared to the adaptation. Specifically, participants distracted during de-adaptation required approximately 100 strides to de-adapt compared to approximately 50 strides in the group that performed conscious correction and the control group. The shortest time required to obtain after-effects was found to be five minutes of adaptation (Vasudevan & Bastian, 2010).

2.6.1.5 How Could The "Error" Be Calculated and Corrected?

Motor control and adaptation have been studied over decades and different theories have been proposed to explain how the brain can generate such behaviour, such as the internal model (e.g.,(Kawato, Furukawa, & Suzuki, 1987) or equilibrium point theory (Feldman, 1986). With regards to the current thesis, we discuss motor adaptation in reference to internal model theories, as the relevant references that form the backbone of this thesis follow this theoretical framework (Malone et al., 2012; Reisman et al., 2005; Reisman et al., 2007). In particular, these studies frequently cited a group of pioneers in the field of motor control research who put forth these theories, using the concept of the internal model to study the mechanisms of adaptation in reaching movements (Shadmehr & Mussa-Ivaldi, 1994; Thoroughman & Shadmehr, 1999). This concept of an internal model has received strong support from the literature on reaching movements (see for example Emken et al., 2007; Hwang & Shadmehr, 2005; Lackner & DiZio, 2000) and locomotion (Bastian, 2008; Malone et al., 2012; Martin et al., 1996). This internal model is considered to represent a sensorimotor map which covers the information about body dynamics associated with a well-learned movement (Bastian, 2008; Hwang & Shadmehr, 2005). However, as indicated previously, the theories on motor adaptation in upper limb movements differ marginally with locomotor adaptation mainly because of the neurological structures involved. A considerable cortical involvement was suggested for upper limb motor adaptation (Shadmehr & Mussa-Ivaldi, 1994; Thoroughman & Shadmehr, 1999). This is not in accordance with locomotor adaptation, as it will be discussed in the following section (2.6.1.6). Consequently, descriptions and definitions about motor adaptation in the context of this work always refer to locomotor adaptation.

The assumption that a movement becomes adapted is based on the after-effects that occur with the removal of the error-inducing situation as shown during the post-adaptation in the SBT protocol (Malone et al., 2012; Reisman et al., 2005). After-effects suggest that the nervous system stores the adapted motor output, hence re-calibrates the initial motor output (Bastian, 2008; Malone et al., 2012; Martin et al., 1996; Reisman et al., 2005). What is the reference used to determine whether motor output deviates from the well-learned pattern?

As shown in the previous section (2.6.1), the interaction between limbs (i.e., interlimb coordination) is of particular relevance for motor adaptation during walking (Dietz et al., 1994a; Malone et al., 2012; Prokop et al., 1995; Reisman et al., 2005). It has been suggested that the timing between left and right heel-strikes (i.e., 'when' the feet land) and their spatial location (i.e., 'where' the feet land) provide the main references for feedback and feedforward control of SBT-induced locomotor adaptation (Malone et al., 2012). In addition, the analysis of spatiotemporal and kinetic parameters in healthy individuals has led to the assumption that the nervous system pays particular attention to inputs from the leg walking on the slow belt to drive adaptation of interlimb parameters during split-belt configuration. Vasudevan et al. (2009) assumed the "sensory cues" are more "salient" from the slow belt (Vasudevan et al, 2009, p. 6), since stance duration is longer on the slow compared to the fast side during adaptation (Reisman et al., 2005; Vasudevan & Bastian, 2010). The following section describes current evidence and assumptions on the neural structures involved in locomotor adaptation and after-effects.

2.6.1.6 Which Structures Are Involved in Locomotor Adaptation?

The "error" induced in the well-learned movement is detected by the central nervous system when the proprioceptive information from the affected muscles and joints deviates from that expected (Hwang & Shadmehr, 2005; Lackner & DiZio, 2000). Signals from muscle spindles are considered as particularly relevant for sensorimotor adaptation (Lackner & DiZio, 2000; MacLellan et al., 2014) since they respond to changes in muscle length and are sensitive to the rate of change (Pearson & Gordon, 2013). Thus, it is not surprising that an important role in representation of position sense is attributed to the afferent feedback from muscle spindles. However, as far as feedforward control is concerned, the system has to anticipate the "new" state or dynamics of the movement (Wolpert et al., 2013). This requires several repetitions when exposed to repetitive perturbation or modification of the movement environment (e.g., split-belt configuration) (Bastian, 2008) which defines the aforementioned "trial-and-error"-based exposure process necessary for motor adaptation.

Both feedforward and feedback control mechanisms can contribute to motor control and adaptation of locomotion. Feedforward control is the generation of a motor command to a respective muscle group or a combination of muscles to achieve a desired movement (Wolpert et al., 2013). Feedback control is less straightforward, requiring a comparison between the desired movement and the perceived movement to drive correction. In feedback control, the information about the differences between the movement that was desired (i.e., voluntary, target-oriented) from the actual movement produced provides an error signal that can influence motor adaptation through feedforward control mechanisms to achieve the desired movement. According to the literature, the feedforward and feedback control of locomotion is considered to be mainly regulated by an interaction of afferent information and cerebellar circuits, as it will be outlined in the following paragraph.

Indeed, the investigation of individuals with cerebellar damage revealed that their feedforward control was disrupted (Choi et al., 2009; Morton & Bastian, 2006). When exposed to SBT walking on belts with different speeds on each leg (split-belt configuration), participants were not able to adapt spatial (step length) nor temporal (double support duration) gait parameters. In contrast, individuals with cerebral lesions showed the capacity to adapt and store spatiotemporal parameters during split-belt configuration similar to healthy controls (Reisman et al., 2005; Reisman et al., 2007). These results suggest a strong contribution of the cerebellar circuits in the prediction of appropriate motor outputs leading to motor adaptation and the eventual production of after-effects. This assumption is supported by neurophysiological studies attributing an anticipatory role to the cerebellum. For example, the results of investigations using magnetoencephalography (MEG) led to the description of the cerebellum as a "detector of change or deviation in a sequence of sensory events" (Ivry, 2000)p. 116). The response of cerebellar circuits has been shown to increase just prior to the actual anticipated stimulus giving experimental evidence that the human cerebellum can sustain a template of temporal information from sensory and motor inputs (Tesche & Karhu, 2000). Furthermore, ventral and dorsal spinocerebellar fibers increased firing during perturbation of walking (Ito, 2000) which suggests that these fibers might send some error-feedback to the cerebellum.

Yet, the cerebral involvement in locomotor adaptation is not completely excluded. As previously mentioned, distraction slowed down the adaptation of step length to unequal belt speeds, but did not prevent it (Malone & Bastian, 2010). Further, conscious correction of step length ("make your steps even") led to a faster correction of the disrupted walking pattern. In the same study, the authors found results which indicate that distinct neural circuits in the cerebellum and spinal networks are responsible for temporal and spatial adaptation. Interestingly, distraction or conscious correction affected the adaptation rate of spatial features associated with step length (center of oscillation) but not of temporal features (interlimb phasing). The center of oscillation describes whether the leg oscillates about a flexed (foot in front of the hip), extended (foot behind the hip) or neutral position to hip (Figure 2-5). The interlimb phasing represents the time between peak angles of the two limbs. The dissociation of spatial and temporal adaptation led to the assumption that spatial features are controlled through the lateral section of the cerebellum, which is strongly connected with the cortical sensorimotor structures via the thalamus (Lisberger & Thach, 2013). The observation that the adaptation of temporal features was not influenced by distraction or conscious correction suggests that the adaptation of temporal features is mainly controlled by automatic control, likely regulated by the previously mentioned spinocerebellar tract and the central parts of the cerebellum (vermis) (Morton & Bastian, 2006).

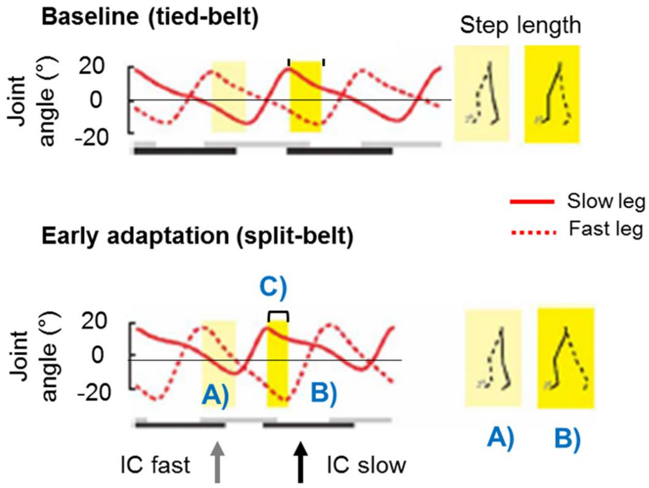

Figure 2-5. Illustration of SBT-induced changes in center of oscillation (A, B) and limb phasing (C) at tied-belt (baseline) and split-belt (early adaptation). IC slow describes the initial contact of the leg walking on the slow belt during adaptation period (continuous lines) and vice versa for the IC fast (dotted lines). A) shows that the slow leg oscillates less in extension compared to baseline in contrast to the fast leg (B) which oscillates more in extension. This leads to a shorter interlimb phasing duration (C). Modified from Choi & Bastian (2007).

Finally, interactions at the spinal cord during adaptation cannot be ignored. Central pattern generators (CPGs) at the spinal level could be involved in the distinct control of spatial versus temporal features. Evidence exists that CPG circuits "shape" the final motor output based on afferent feedback from mechano- and proprioceptors (MacKay-Lyons, 2002). Animal studies led to the assumption that separate circuits of interneurons are responsible for either rhythm control (e.g., cadence) or pattern generation (recruitment of muscle synergies / motor modules) (Dietz, 2003; Forssberg et al., 1980; Lafreniere-Roula & McCrea, 2005; Malone & Bastian, 2010; Rybak et al., 2006). Split-belt studies in animals (e.g., in spinal cats [Forssberg et al., 1980] and spinal turtles [Stein & McCullough, 1998]) showed that these animals with spinal lesion were able to produce distinct rhythmic movement patterns and belt-speed dependent adjustments of walking parameters on each limb, respectively. These authors suggested the presence of neurons that generate rhythmic movements in the spinal cord and for each limb separately. The presence of these rhythm generators was suggested as well to exist in humans. This assumption was based particularly on studies on individuals with complete spinal cord injuries showing rhythmic contractions of lower limb muscles as presented in the review by Duysens and Van de Crommet, (1998). Particularly interesting for the present work are the results of a split-belt treadmill study on infants who were not able to walk independently and thus less likely to be influenced by cortical control (Patrick et al., 2014). In these infants, the leg walking on the faster belt shows different adjustments in spatiotempral parameters than the slower leg also suggesting that in humans, pattern generators exist in the spinal cord to control each limb separately. However, it has to be taken into account that evidence about the mechanisms and role of CPGs in humans is still relatively scarce and interpretations should be handled with care (Dietz, 2003; MacKay-Lyons, 2002; MacLellan et al., 2014). Thus, based on theoretical work on the internal model and the current knowledge of the literature, it can be assumed that the interaction between afferent feedback and the feedforward adjustments regulated by the cerebellum tend to play a major role in locomotor adaptation. As a consequence, the cerebellum in particular appears to be involved in the processing and regulation of joint mechanics and muscle activity to achieve successful adaptation of interlimb parameters.

2.6.2 Biomechanics Underlying SBT-induced Adaptation and After-effects

2.6.2.1 The Role of Muscle Activity in SBT Walking

A small body of literature exists about the contribution and changes in muscle activity involved in SBT-induced adaptation and after-effects. Until now, these findings have been limited to healthy individuals. The SBT protocols were based on either the speed ratio of 2:1 (Finley, Bastian, & Gottschall, 2013; Jansen et al., 2013; MacLellan et al., 2014; Ogawa et al., 2014) or different combinations of speeds (Dietz et al., 1992). Many of these studies have used different approaches to explore muscle activity in SBT walking. Jansen et al. (2013) studied the muscle activity involved in artificially (i.e., SBT) -induced limping and asymmetrical gait (Jansen et al., 2013). Asymmetry was quantified using the acceleration of the center of mass. Ogawa et al. (2014) studied mechanisms underlying SBT-induced adaptation within muscle activity and ground reaction forces and Finley et al. (2013) investigated muscle activity and energy consumption (metabolic power) (Finley et al., 2013; Mawase et al., 2013; Ogawa et al., 2014). The latter study investigated whether lower limb muscle activities (BF, VL, TA, and

GL) increased or decreased in line with energy consumption. MacLellan et al. (2014) used a non-negative matrix factorization to quantify synergies (activity pattern) of lower muscles with the focus on the temporal adaptation of synergies and interlimb coordination. Dietz et al. (1994a) studied changes in amplitude of lower limb activity but during different combinations of belt speed ratios, compared walking with equal belt speeds. These studies focused particularly on the analysis of muscle activity while participants walked with unequal belt speeds (split-belt configuration; period of adaptation). The SBT-induced after-effects on muscle activity were only presented by Ogawa et al. (2014). Although the methods and outcome parameters varied substantially among these studies, all found bilateral changes in lower limb muscle activity which is detailed below. These bilateral changes point to a bilateral organisation of muscle activity during walking with split-belt configuration.

2.6.2.1.1 *Changes in EMG Amplitude During Adaptation*

Considering the amplitude of muscle activity, the analysis of four lower limb muscles (BF; RF; TA; gastrocnemius medialis; GM) revealed that during early adaptation, the proximal muscles (BF and RF) and the ankle dorsiflexor (TA) showed significantly higher EMG activity compared to baseline (Ogawa et al., 2014). On the slow walking side, the RF increased during stance phase and swing phase and the TA increased during stance phase. Interestingly, the TA increased as well on the fast side but during swing phase together with the BF muscle. The antagonist of the TA muscle, the plantarflexor (GM), did not change on either side during the early adaptation period.

From early to late adaptation, both proximal muscles (BF and RF) decreased their EMG activity to the baseline level. Likewise, the TA on the slow side increased activity during early stance followed by a clear phenomenon of adaptation towards baseline. In contrast, the TA on the fast leg retained the increased activity through the adaptation period. Despite a lack of change in the early adaptation, the plantarflexor GM decreased its activity at the end of the adaptation on the slow leg. A general decrease in muscle activity from early to late adaptation was observed as well in the Finley study (2013) associated with a decrease in energy consumption.

The increase of dorsiflexor activity on the slow belt during adaptation is in agreement with the results by Dietz et al. (1994a). Furthermore, Dietz et al. showed that participants presented a premature activation of the plantarflexors during stance phase on the fast belt. The proximal muscles (RF, BF) did not present any significant changes. In summary, there is a general agreement that distal lower limb muscles play a particular role in SBT-induced locomotor adaptation. In addition, the more recent SBT studies (Finley et al., 2013; Ogawa et al., 2014) also revealed a phenomenon of adaptation in proximal and distal muscles. Such adaptation was not described by Dietz et al. (1994a). In this study, authors reported only that an initially chaotic pattern turned into a regular burst-like pattern which was maintained through time. The analysis on muscle synergies however revealed that during SBT walking, healthy individual referred up to four different activity patterns (MacLellan et al., 2014). Among two of these pattern, the contribution was largest from the slow or fast side plantarflexor muscles and the contralateral hamstring group (BF and ST). The extensors of the hip (GM) and knee (VM, VL, and RF) of either slow or fast side defined the other two patterns.

2.6.2.1.2 *Changes in EMG Amplitude During Post-adaptation*

As far as SBT-induced after-effects on muscle activity are concerned, only Ogawa et al. (2014) investigated muscle activity immediately after walking with split-belt configuration. When belts were returned to equal speeds, after-effects were found on both sides and particularly so during the stance phase. The RF increased bilaterally, whereas the plantaflexors showed an increase on the slow side and the dorsiflexors showed an increase on the fast side compared to baseline. The after-effects were not particularly discussed in the Ogawa paper (2014). However, in that study an inverted ("switched") pattern of effects on dorsiflexors was observed, since it was increased on the slow side during adaptation and on the fast side during post-adaptation. Such inverted after-effects were found in spatiotemporal parameters (e.g., step length) (Malone & Bastian, 2013; Reisman et al., 2005).

2.6.2.1.3 *Changes in EMG Timing during SBT Walking*

The study by MacLellan and colleagues (2014) quantified the interlimb coordination during SBT walking in healthy individuals. The interlimb coordination of muscle activity was quantified based on the temporal pattern (activation profiles) of lower limb muscles using cross-correlation analysis on muscle activation pattern. Before and during the SBT walking, two out of four patterns presented a bilateral shift towards earlier activation during walking with split-belt configuration compared to tied-belt walking. In the remaining two synergies, no change in temporal aspects was detected among walking conditions. Moreover, the temporal pattern of these synergies changed in the same manner between sides which resulted in retention of the initial interlimb coordination. This led to the conclusion that motor output (in terms of timing) during SBT walking is regulated based on tight bilateral organization. It has been suggested that such tight organization is regulated by limb-specific CPGs which are mediated by proprioceptive input and supraspinal signals. As aforementioned, the presence of CPGs in humans is still debated in literature (Dietz, 2003; MacKay-Lyons, 2002). However, it has been hypothesized that CPGs are particularly involved in temporal (cadence) control of automatic and rhythmic movements such as walking. Therefore, the temporal and bilateral shifts observed by MacLellan et al. (2014) could be explained by these spinal generators.

2.6.2.1.4 *The Main Players in Feedback and Feedforward Control*

Distal lower limb muscles tend to be particularly involved in split-belt induced locomotor adaptation. Indeed, Ogawa et al. (2014) concluded that the TA and GL play an important role in predictive and reactive feedback strategies underlying SBT-induced adaptation of gait patterns. The increase in slow TA activity during early adaptation was found during early stance phase, in accordance with the increase in braking force which represents the backwards acceleration or “braking” during initial contact. In addition, TA EMG and braking force showed similar patterns of adaptation between early and late adaptation periods. In contrast, slow plantarflexors and the propulsive force (both active during the push-off phase) did not show any pattern of adaptation and only a very small after-effect. This led to the suggestion that the plantar- and dorsiflexors play a role in feedback and -forward control during SBT walking. Since these changes were found on the slow side, it can be concluded that the slower side plays a major role as a “reference” for feedback information to adjust the feedforward command during SBT walking.

Ogawa et al. (2014) went further in their interpretation and concluded that the pattern of adaptation observed in the slow dorsiflexor and braking force at initial contact indicates an increase in ankle stiffening as a consequence of walking with unequal speeds. The hypothesis that the feedforward control comes particulary into play during initial contact was supported by the findings from Mawase et al. (Mawase et al., 2013). This group observed that the vertical GRFs during initial contact presented a stepwise adaptation to split-belt walking as well as after-effects during the period of post-adaptation. While the theory of feedforward control is plausible, it should be handled with care and seems less evident compared to the feedback and feed–forward hypothesis. Investigations on joint stiffness quantified this parameter by means of joint angles and joint torques (Bressel & McNair, 2002; Chung et al., 2004). For example, an augmentation of ankle stiffness can lead to a reduced range of motion in dorsiflexion (Chung et al., 2004) in line with an increase in the joint torque-angle ratio when compared to control values ((Bressel & McNair, 2002; Chung et al., 2004). These two parameters, which were used to define joint stiffness, were not considered in the study by Ogawa et al. (2014).

Furthermore, the assumption that the plantar- and dorsiflexors play a role in feedback and feed–forward control during SBT walking, respectively, seems reasonable when taking into account early findings on neural control mechanisms of the ankle antagonists. It has been considered that the dorsiflexor activity was mainly regulated by central mechanisms (Dietz, 1992; Dietz et al., 1994a), which confirms a potential contribution of this muscle to feedforward mechanisms. As far as plantarflexors are concerned, they were considered as being mainly regulated by proprioceptive feedback mechanisms mediated on a spinal level (Dietz, 1992). Therefore, it is plausible that this muscle group contributes particularly to the transduction of error-feedback information from the distal lower limb to the central nervous system to adjust feedforward control.

Overall, studies of muscle activity and GRFs during SBT walking suggest that each leg plays a different role during SBT walking but with a keen interaction between each in order to adjust the gait pattern (Dietz et al., 1994a; MacLellan et al., 2014; Mawase et al., 2013; Ogawa et al., 2014).

2.6.2.2 Lower Limb Net Joint Moments

The kinetics of movement involved in SBT-induced adaptation and after-effects were studied recently and by a small number of investigations (Lauzière et al., 2014a; 2016; Mawase et al., 2013; Ogawa et al., 2014). Kinetic parameters investigated during SBT-induced adaptation include lower limb net joint moments and the abovementioned GRFs (Ogawa et al., 2014). The analysis of lower limb net joint moments revealed particular implications of the distal lower limb in locomotor adaptation (Lauzière et al., 2014a) which is in agreement with EMG and GRF studies.

Lauzière et al. (2014a) analyzed the after-effects of SBT walking at a 2:1 ratio on lower limb net joint moments. The protocol used was based on the Reisman group (Reisman et al., 2005). However, in the Lauzière study (2014a) belt speeds were not predefined but rather were based on each individual's comfortable speed. Nonetheless the after-effects of step length corresponded to the findings of Reisman et al. (Reisman et al., 2005). Step length was longer on the side that walked on the fast belt during the adaptation period compared to the side on the slow belt. In particular, the most pronounced changes in net joint moments were found at the ankle joint ($p < 0.005$) (Lauziere et al, 2014, Figure 3). On the fast side, the average peak plantarflexion moment was reduced by 29.1% during 20 - 60% of the gait cycle. In contrast, a significant increase of 4.1% of average peak plantarflexion moment was found after slow belt walking during 20 - 55% of the gait cycle. In addition, the average peak hip extension moment during early stance increased about 32.4% on the fast side when compared to baseline. The increase in the peak extension on the fast side combined with the increase in slow peak plantarflexion moments could be due to the counteracting relationship between propulsive and braking force during walking. The increase in slow plantarflexion moment can be considered to lead to an increase in propulsive force, which subsequently has to be balanced out by the braking force during subsequent initial contact of the contralateral leg. The influence of the slow plantarflexor on fast leg movements is supported by the strong correlation found between the peak plantarflexion moment on the slow side with contralateral changes in step length ($r = 0.88$; $p < 0.001$). Consequently, a major role in step length after-effects can be attributed to the plantarflexion moments.

2.6.3 Effects of SBT Walking on Post-stroke Locomotion

While use of SBT in healthy individuals has provided novel insights into the mechanisms of adaptation, SBT has also been used successfully to study post-stroke locomotion. First of all, cerebral lesions do not prevent affected individuals from adapting their gait patterns and storing the adapted pattern during error-augmentation-based SBT walking (Lauziere et al., 2015; Lauzière et al., 2014a; 2016; Malone & Bastian, 2013; Reisman et al., 2007). In accordance with healthy controls, individuals with stroke initially increased their asymmetry in interlimb parameters during early adaptation (split-belt configuration) and adapted as well towards their baseline ratio of symmetry from early to late adaptation (Malone & Bastian, 2013; Reisman et al., 2007). Thus, according to Malone and Bastian (2013), individuals post-stroke do have a "default" gait pattern which is the reference for adaptation.

The after-effects observed in individuals post-stroke also display a similar pattern to that found in healthy controls (Malone & Bastian, 2013; Reisman et al., 2007). The leg that walked on the fast belt during adaptation (split-belt configuration) presented an increase of step length and double support duration during the subsequent tied-belt walking (post-adaptation) (Reisman et al., 2005; Reisman et al., 2007) and sometimes vice versa on the slow leg. In the case of an initial asymmetry in these parameters, SBT walking led to a reduction in asymmetry of interlimb parameters when the side with the shorter step length walked on the fast belt during split-belt configuration (Lauzière et al., 2014a; 2016; Malone & Bastian, 2013; Reisman et al., 2007).

Despite the similarity in parameters and direction of adaptation change (i.e., increase or decrease of asymmetry) between healthy and post-stroke individuals, the rate of adaptation has been shown to be affected by the cerebral stroke. Malone & Bastian (2013) showed that individuals post-stroke required longer exposure to the SBT to adapt and de-adapt spatial and temporal parameters to the same magnitude as healthy controls. This suggests that next to the aforementioned structures involved in locomotor adaptation (cerebellum, proprioceptors, and limb specific CPGs), cerebral networks are involved in adaptation of spatial and temporal parameters in individuals post-stroke. The implication of cerebral mechanisms in adaptation is also supported by observations described earlier that conscious correction or distraction during

SBT walking modifies the rate of adaptation in healthy individuals (Malone & Bastian, 2010). However, because of the cortical remapping and neural rewiring that occurs in response to the cerebral stroke (Dancause et al., 2005; Dawes et al., 2008), it remains unclear whether the cerebral mechanisms involved in adaptation in individuals post-stroke are the same as in healthy controls.

2.6.3.1 Biomechanics Studied in SBT Walking Following Stroke

The impact of SBT-induced perturbation on other locomotor parameters than the spatiotemporal ones described earlier has rarely been studied in individuals post-stroke. Indeed, changes in muscle activity have only been described in healthy individuals (section 2.6.2.1). While this is a keen gap of knowledge in the literature, the after-effects on lower limb net joint moments and effort during SBT walking have been investigated in individuals post-stroke (Lauzière et al., 2014a; 2016), described below.

2.6.3.1.1 *The Effects on Net Joint Moments*

Lauzière et al. (2014a; 2016) explored the effects of strokes on net joint moments during gait. They based their protocol on the Reisman protocol (Reisman et al., 2005; Reisman et al., 2007), but performed it twice for the individuals with stroke. During the adaptation period of the first condition, belt speed was doubled on the non-paretic side (non-paretic-fast condition). This was followed by a second condition with an increase in belt speed on the paretic side (paretic-fast condition). During both conditions, six minutes of walking with split-belt configuration (ratio 2:1) led to similar after-effects on joint mechanics for the participants with chronic stroke (n = 20) when compared to healthy controls (n = 10) (Lauzière et al., 2014a). Slow belt walking led to an increase in peak net plantarflexion moment (20 - 55% of the gait cycle) during post-adaptation compared to baseline. The average increase in the stroke group (16.5%) was larger than in the control group (4.1%). A slight but significant decrease (8.1%) of plantarflexion moment after fast belt walking was also found in individuals post-stroke during 20 - 60% of the gait cycle. It has to be emphasized that the average peak plantarflexion moments were smaller on the paretic than non-paretic side. While this resulted in an increase in plantarflexion asymmetry during paretic fast-belt walking, it led to a reduction in asymmetry after non-paretic fast-belt walking. Furthermore, during both conditions the peak plantarflexion moments were positively correlated with observed after-effects in contralateral step length for individuals post stroke (r = 0.559 and r = 0.535), with slightly smaller Pearson's correlation coefficients than in healthy individuals.

There are two main conclusions that can be taken from these studies looking at net joint moments. First, the largest after-effects of SBT walking were found at the ankle joint in both individuals post-stroke and healthy controls. Second, plantarflexion moments tend to play a major role in contralateral changes of step length during post-adaptation. These observations have led to the hypothesis that paretic fast belt walking might improve step length symmetry in individuals with a shorter paretic step length but at the cost of a reduction in the utilization of the paretic side.

2.6.3.1.2 *The Role of Muscular Effort*

In addition to net joint moments, the group of Lauzière et al. (2016) also investigated the relationship between the muscular effort required during walking and the capacity to adjust step length and ankle joint moments after SBT walking. To do so, the so-called muscular utilization ratio (MUR) was calculated for the plantarflexor group (Milot et al., 2006; Requiao et al., 2005). The MUR represents the ratio of the net joint moment measured during walking and the potential maximal moment produced at the specific joint during MVC. It expresses the muscular effort required during a specific task such as walking. The analysis tested the relationship between the level of effort (MUR) required from paretic plantarflexors during walking at a comfortable speed (baseline) with the capacity to adjust paretic joint moments and non-paretic step length after walking six minutes at split-belt configuration (post-adaptation). As previously described, paretic slow belt walking led to an increase of paretic peak net plantarflexion moment and contralateral (non-paretic) step length (Lauzière et al., 2014a). In accordance with these earlier findings, individuals with an initially shorter non-paretic step length improved their step length symmetry after paretic slow-belt walking (Lauzière et al., 2016). Such improvements in symmetry were in line with an increase in the level of effort (MUR) on the paretic side. Furthermore, it was found that participants who required high levels of effort in the paretic plantarflexors at baseline showed reduced capacity to increase non-paretic step length and paretic plantarflexion moment during post-adaptation.

In contrast, the participants who presented low levels of effort in this muscle group showed a higher capacity to change these parameters. Strong negative correlations were found between the initial MUR on the paretic side and the percentage of change in paretic plantarflexion moment at late stance (r = -0.70; p = 0.001) and non-paretic step length (-0.65; p = 0.003) (Lauzière et al, 2016, Figures 1-2). These results led to the assumption that the participants presenting low levels of effort had better capacities to adjust joint moments and step length because they could mobilize residual strength from the plantarflexors. Thus, the group with high levels of effort might in general be limited by their weakness in those particular muscles.

2.6.4 Transfer of SBT After-effects to Walking Over Ground

The previously described SBT-induced after-effects on abled-body and post-stroke gait were all quantified during walking with tied-belt configuration on the SBT immediately after the split-belt configuration. Walking on a treadmill, however, can be considered as less functionally relevant to everyday life than walking over ground. Therefore, after an injury the final aim is to ultimately improve walking over ground and not just on a treadmill. Indeed, walking over ground and on a treadmill are two different walking contexts. On a treadmill, the ground is moving and the body stays in place whereas during walking over ground the body has to generate and absorb force to provide forward acceleration. Nonetheless, based on the analysis of muscle activity it has been suggested that both walking conditions require similar motor control strategies (Hesse et al., 1999; Kautz et al., 2011). In contrast, studies on kinetic and kinematic gait parameters found that individuals post-stroke tend to increase temporal gait parameters (i.e., cadence and percent stance time) (Brouwer, Parvataneni, & Olney, 2009; Kautz et al., 2011) and decrease GRFs (i.e., peak propulsion and vertical forces) (Brouwer et al., 2009; Harris-Love et al., 2004) during treadmill walking compared to walking over ground. Treadmill walking was also found to require a higher metabolic demand compared to walking over ground for individuals post-stroke (Brouwer et al., 2009).

Furthermore, walking over ground has been considered to require higher "attentional demand" because of an increased "distraction" (Savin, Morton, & Whitall, 2014) and dynamic equilibrium control (Eng & Winter, 1995). Unfortunately, the authors did not precisely indicate the source of "distraction". A possible explanation for higher "attentional demand" is the navigation and avoidance of potential obstacles required during walking over ground that is less important during treadmill walking. This suggests a more complex decoding and regulation of sensory input by the central nervous system during walking over ground compared to the treadmill. This is likely due to the importance of the integration of visual information during walking over ground towards a target (Deshpande & Patla, 2005) as well as in obstacle avoidance (Gerin-Lajoie, Richards, & McFadyen, 2005; 2006; McFadyen et al., 2007; Weerdesteyn et al., 2008). Indeed, the removal of visual inputs during SBT adaptation influenced the transfer of SBT-induced after-effects to the over ground gait. Healthy individuals showed larger transfers of after-effects to the over ground when vision was entirely removed during treadmill walking (Torres-Oviedo & Bastian, 2010). This was explained by the removal of the visual context which limited the creation of a treadmill-specific internal model. Thus the central nervous system had to rely exclusively on proprioceptive and vestibular inputs to generate the internal model, which seem less (but not completely) able to distinguish between the tasks (i.e., treadmill vs. over ground walking) as with visual cue.

Nonetheless, despite the differences in walking demand between treadmill and over ground, SBT-induced after-effects were transferable to over ground in individuals post-stroke and healthy controls (Hamzey et al., 2016; Reisman, Wityk, et al., 2009). In both studies the magnitude of after-effects was slightly smaller during walking over ground. Interestingly, the within-group comparison revealed that the stroke group presented a larger extent of transfer to the ground compared with the controls (Reisman, Wityk, et al., 2009). This supports the aforementioned assumption that the cerebral cortex is involved in task- / condition-specific locomotor adaptation. The analysis from Hamzey et al. (2016) revealed that healthy individuals transferred about 40% of after-effects in step length to the over ground condition. Taking these observations into account, can we expect motor learning with a successful long-term reduction in gait asymmetry during walking over ground in individuals post-stroke? This question was investigated by the Reisman group in a recent paper (Reisman et al., 2013) and will be discussed next.

2.6.5 Repeated Exposure to SBT Walking

The potential for long-term improvement of gait symmetry post-stroke during walking over ground was evaluated with a training study on repeated SBT walking using the error-augmentation strategy (Reisman et al., 2013). Symmetry was significantly reduced after 12 sessions of training that persisted for up to three months. Thirteen participants were trained three days/week for four weeks. The protocol consisted of 30 minutes of treadmill walking at unequal speeds at a 2:1 ratio. Fast belt speed was at 1.0 m/s and the slow belt speed at 0.5 m/s. The side with the shorter step length was trained on the faster belt. The treadmill training was combined with approximately 5 - 10 minutes of training over ground to maintain the improved symmetry. Among the parameters of symmetry measured, only step length symmetry was reduced significantly from baseline to post-training evaluation (p = 0.05). Movement kinematics and kinetics were not assessed. Seven out 12 participants exceeded the predetermined minimal criteria of change (0.020) defining the group of responders. The comparison of different clinical parameters (e.g., Fugl-Meyer score, perceived exertion, percentage of change of perceived exertion, initial walking speed, and asymmetry) revealed that only perceived exertion and its percentage of change from the first to the last training differed between the group of responders and non-responders. Based on these findings, it was concluded that the larger the perception of effort required and the bigger its reduction from the first to the last training session, the larger are changes in step length symmetry.

In conclusion, repeated exposure to SBT walking tends to be an up and coming approach when the goal is longer-term reduction of step length asymmetry. Furthermore, SBT protocols that induce locomotor adaptation appear to confirm the theory that repeated exposure to motor adaptation leads to motor learning, the retention of a new movement pattern over longer term (Bastian, 2008; Malone, Vasudevan, & Bastian, 2011). However, it remains unclear which muscles are contributing to these longer-term changes in symmetry and what the impact of repeated exposure to error-based SBT walking has on joint kinetics. Furthermore, the reinforcement of step length symmetry during additional 5 - 10 minutes applied in the training study described above (Reisman et al., 2013) represents a potential bias, since actual contribution of the error-augmentation-based SBT training to the reduction in asymmetry remains unclear. One of the goals of the current thesis is to further explore these questions.

Summarizing the previous literature review on SBT-induced changes in post-stroke gait, two major questions remain unresolved: 1) Which muscles are involved in the SBT-induced adaptation and immediate after-effects on step length in post-stroke gait? 2) What changes in muscle activity and gait kinetics are in line with the long term improvements of step length asymmetry during walking over ground?

Chapter 3. Objectives and Hypotheses

3.1 Main Hypotheses and Objectives

It is clear from the literature that the use of the SBT can improve the symmetry of interlimb parameters for a short-term period in individuals post-stroke. In addition, it has been suggested that repeated exposure to SBT walking has the potential to reduce step length asymmetry over the longer term (Reisman et al., 2013). Currently, the changes in muscle activity involved in these SBT-induced perturbations have not yet been investigated in the stroke population. Likewise, the biomechanical and EMG modifications underlying long-term improvements of step length symmetry after a training intervention have not been reported. The goal of the current thesis is to address this gap in the literature. The quantification and analysis of the causes of step length changes induced by SBT adaptation within these biomechanical parameters will provide a better understanding of the effects of SBT walking and provide recommendations of clinically relevant protocols. Therefore, the main objectives of this thesis were to investigate short- and long-term changes in lower limb muscle activity in individuals post-stroke undergoing error-augmentation-based SBT protocols to alter their step length. These objectives include the investigation of muscle activity in the scope of a cross-sectional analysis of lower limb muscle activity during and after SBT walking as well as the analysis of lower limb muscle activity and biomechanics during walking over ground after repeated exposure to SBT walking.

In line with the cross-sectional analysis (paper #1), the hypotheses were the following: 1) Based on the findings by Lauzière et al. (2014a) we expect that distal muscles will show the most pronounced activity increase in combination with the after-effects of step length. 2) Also based on Lauzière et al's findings, we hypothesize that the changes in plantarflexor muscle activity would be associated with the changes in SL. For plantarflexor muscles, the changes will be more pronounced after walking on the slower belt compared to walking on the faster belt. 3) During adaptation, the pattern of adaptation will be characterized by an increase or decrease of muscle activity at the beginning of perturbation and a return towards baseline at the end of the six minutes of perturbation.

As a secondary objective, the outcome parameters in the stroke group will be compared to those of healthy controls. For the spatiotemporal parameters, we expect to reproduce data similar to previous studies (Malone & Bastian, 2013; Reisman et al., 2005; Reisman et al., 2007) while for the EMG, we expect differences in changes of EMG activity between groups, considering the altered muscle activity and neuromuscular control secondary to a cerebral stroke (section 2.3). We expect less consistency in muscle groups and a higher number of muscles involved in step length after-effects in the post-stroke group. Furthermore, the frequently observed abnormal muscle timing in individuals post-stroke led to the hypothesis that we will find longer durations of coactivation of the antagonist controlling ankle (tibialis anterior and gastrocnemius lateralis muscles) and knee (vastus lateralis and semitendinosus muscles) after SBT walking when compared to healthy controls.

As far as the training study is concerned, we determined the following three hypotheses:

1) In agreement with previous findings about repeated SBT walking (Reisman et al., 2013), we expect that our protocol will replicate the results of step length asymmetry in the individuals post-stroke. 2) Since our protocol allows adjustment of training speed at each session and contains generally less breaks than Reisman's protocol, we hypothesize that concomitant improvement in walking speed and endurance will be observed. 3) Based on previous studies on joint kinetics during SBT walking (Lauzière et al., 2014a), we expect that plantarflexor muscles and hip extensors will be the main contributors to the reduction in step length asymmetry post-training. More precisely, during stance phase of gait, the utilization of plantarflexors will be augmented on the leg trained on the slow belt, whereas the hip extensor activity will increase on the side of the faster belt.

3.2 Specific Objectives for Scientific Papers in this Thesis

The specific objectives in line with the papers presented in this thesis are as follows:

Paper #1:

1) To analyze changes in EMG amplitude of twelve lower limb muscles during and after SBT walking in individuals post-stroke.

2) To examine the associations between changes in EMG amplitude and changes in step length after six minutes of split-belt treadmill walking (post-adaptation).

<u>Supplementary data paper #1:</u>

S1) To compare SBT-induced effects on EMG amplitude of twelve lower limb muscles in individuals post-stroke with the effects measured in a group of healthy age-matched controls.

S2) To assess the changes in coactivity of ankle and knee antagonists and interlimb coordination of twelve lower limb muscles after six minutes of split-belt treadmill walking in individuals post-stroke and healthy controls.

Paper #2:

1) To test the effects of a new error-augmentation-based protocol on step length asymmetry, walking speed, functional capacity, and endurance during walking over ground in individuals post-stroke.

<u>Supplementary Data Paper #2:</u>

S1) To present preliminary findings on the feasibility of the protocol in terms of practicality and readiness for implementation in a clinical environment.

Paper #3:

1) To quantify the immediate and one-month post-training effects of the previously tested (paper #2) training protocol on lower limb muscle activity and joint kinetics during walking over ground in individuals post-stroke.

Chapter 4. Methods

This section presents the methodological aspects used in this doctoral thesis. First, the cross-sectional (paper #1) and the pilot training study (papers #2 and #3) are described. Paper #1 presents the results of a cross-sectional analysis on lower limb muscle activity during and after SBT walking in individuals post-stroke. A main focus was set on the changes in EMG amplitude in combination with the SBT-induced after-effects on step length. Papers #2 and #3 present the effects of a repeated SBT intervention on step length symmetry, muscle activity and joint kinetics in twelve individuals post-stroke. A detailed description of the treatment and analysis of EMG data is presented in Appendix I.

4.1 Methods: Cross-sectional Study

In the scope of the cross-sectional study, two groups of participants were recruited. The first group consisted of individuals who suffered a cerebral stroke more than six months ago and received intensive functional rehabilitation in a rehabilitation center in greater Montreal (Quebec, Canada). For supplementary analyses, a control group of 10 healthy and age-matched individuals was recruited. The methods and study protocol were approved prior to the studies by the research ethics board of the Centre de recherche interdisciplinaire en réadaptation du Montréal métropolitain (CRIR) institutions.

4.1.1 Participants

The inclusion criteria for the individuals post-stroke were 1) having a first unilateral stroke $\geq$ 6 months prior to the study; 2) being capable of walking independently over short distances at $\geq$ 0.5 m/s without assistive devices or physical assistance; and 3) tolerating three hours of walking tasks with breaks. Exclusion criteria included cerebellar lesions, major pain, heminegligect or - anopsia, signs of major depression quantified with a score $\geq$ 10/15 on the Geriatric Depression Scale (Sheikh & Yesavage, 1968), severe cognitive deficits defined with a score < 25/30 at the Folstein Mini-Mental Exam (Folstein, Folstein, & McHugh, 1975), cardiorespiratory problems and other medical and cognitive conditions that could affect walking capacity and the ability to understand instructions, as verified by medical records. As far as the control group was concerned, inclusion criteria were 1) not having any neurological, degenerative, orthopaedic or cardiac disease; or 2) any major pain affecting their walking capacity; and 3) the ability to tolerate three hours of walking tasks with short breaks. All participants had to sign the consent form which was approved by the ethics committee of the CRIR (see Appendices V and VI).

4.1.2 Recruitment

Individuals post-stroke were recruited from the following three rehabilitation centers: Institut de réadaptation Gingras-Lindsay (IRGLM), Centre de réadaptation Lucie-Bruneau (CRLB) and Hôpital de réadaptation Villa Medica (HRVM). Clinicians and patients were informed about the recruitment by posters summarising the study and the criteria of eligibility. Two other recruitment strategies were used for individuals post-stroke, as well as healthy control participants: 1) Individuals were contacted by phone based on a list of participants who had been subjects in previous studies and who were willing to be contacted for other studies; and 2) by the network of our colleagues (students, engineers and researchers) of the pathokinesiology laboratory at the IRGLM.

A total of 22 individuals post-stroke and 10 healthy controls were recruited. For paper #1, 16 individuals post-stroke were included. Ten healthy individuals were analyzed as a control group. Data from the control group is presented in section 5.1.13.2 as supplementary results following paper #1.

4.1.3 Clinical Evaluation

To evaluate participants' eligibility and to characterize their physical and mental conditions, clinical evaluations were conducted. This section describes the clinical assessments used.

4.1.3.1 Assessments Used for Individuals Post-stroke and Healthy Controls

Both groups were initially questioned about their medical conditions and medication intake. Individuals post-stroke had to provide information about the time since their cerebral stroke and, if possible, about the area of the lesion. Healthy controls were asked about previous medical problems which could affect participation in the project (e.g., cardio-pulmonary problems, visual deficits, etc.). Both groups were asked to rate the presence and severity of pain in the lower limbs during activity and at rest by using the Visual Analog Scale of 10 cm (Hawker, Mian, Kendzerska & French, 2011). Bilateral sense of touch (cutaneous sensibility) was evaluated with the Semmes-Weinstein monofilaments (Collins et al., 2010) and bilateral sense of vibration with a diapason at 128 Hz at the malleoli laterali (Kokmen, Bossemeyer, & Williams, 1977). Movement and position sense was tested at the hips, knees and ankles. To do so, blind-folded participants were instructed to imitate the position of one leg (as set by the evaluator manually) with their other leg. Functional mobility was assessed with the Timed up and Go (TUG) test (Podsiadlo & Richardson, 1991) and walking speed was quantified by the 10 Meter Walking Test (10MWT) (Perera et al., 2006) at comfortable and fast speeds, respectively. To evaluate participants' clinical balance, the Berg Balance Scale (BBS) was used (Berg, Wood-Dauphinee, & Williams, 1995). Furthermore, healthy individuals were asked for there more dominant side which was needed for the adaptation period in the experimental protocol. The dominant side walked on the faster belt. This test was not conducted for individuals post-stroke since they were tested with each paretic and non-paretic side walking on the faster belt.

4.1.3.2 Additional Assessments for the Stroke Group

In addition to the previously described evaluations, individuals post-stroke were evaluated for hemineglect or - anopsia, cognitive deficits (Folstein Mini-Mental Exam) and signs of depression (Geriatric Depression Scale). This was followed by an evaluation of lower limb function. Deficits in lower limb motor function were quantified with the Chedoke McMaster Stroke Assessment (Gowland et al., 1993). To characterize ankle muscle tone and lower limb spasticity, the Modified Ashworth Scale (Bohannon & Smith, 1987) and Composite Spasticity Index (Levin & Hui-Chan, 1993) were used, respectively.

4.1.4 Movement Analysis on the Split-belt Treadmill

Once participants were included in the project, they underwent a second evaluation session which included the three-dimensional (3D) analysis of movement and the recording of muscle activity during walking at different conditions on the SBT.

4.1.4.1 The Split-belt Treadmill (SBT)

The instrumented SBT from Bertec Corporation (USA) used for the experimental setup is built with two belts (Figure 4-1). Each belt has two integrated force plates which allow recording the forces exerted on the belts. The speed of each belt can be regulated independently from 0.01 to 6.67 m/s with an increment range of 0.01 m/s^2 to 0.25 m/s^2. Handrails on left- and right-hand sides, as well and a harness, were used for safety purposes without unloading bodyweight.

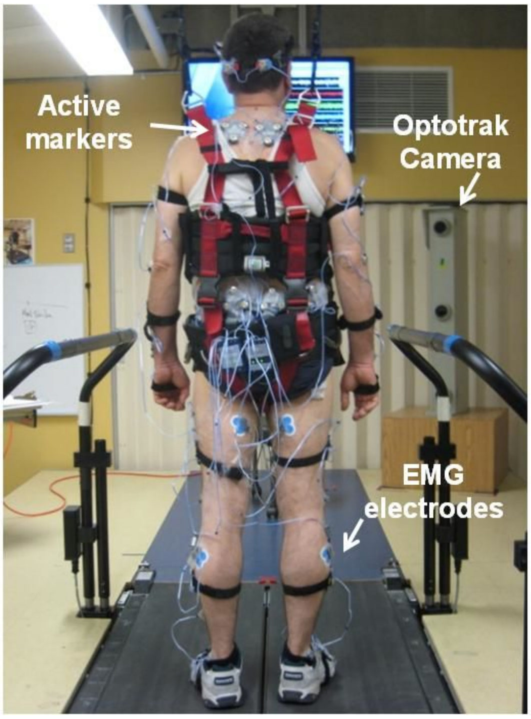

Figure 4-1. Instrumented participant on the split-belt treadmill used for data collection described in the present thesis. The participant is secured with a harness and instrumented with surface electrodes and active markers.

4.1.4.2 Instrumentation for Biomechanical Analysis and Recording of Muscle Activity

Prior to the instrumentation, participants' body-weight, height, as well as length and circumference of lower limb segments were measured. This information was required for offline treatment of kinematic and kinetic data. To obtain biomechanical data, the 3D-motion analysis system Optotrak Certus (Northern Digital Inc., Waterloo, ON, Canada) was used with 75 infrared markers placed across all segments bilaterally (feet, shanks, thighs, pelvis, trunk, head, upper arms, lower arms and hands) (Figure 4-1). For the cross-sectional analysis, only

34 markers (lower limbs and pelvis) were analyzed. The infrared signals from the active markers were captured by four Optotrak cameras at 30 Hz and resampled at 60 Hz to correspond to the data points of the GRFs. These forces were captured from the treadmill integrated force plates at 600Hz. These forces were resampled at 60Hz to correspond to the kinematic data.

Muscle activity was registered bilaterally for the following six lower limb muscles: tibialis anterior (TA), gastrocnemius lateralis (GL), vastus lateralis (VL), rectus femoris (RF), semitendinosus (ST), and gluteus medius (GLM). Self-adhesive surface electrodes (Ag/AgCl; Ambu_BlueSensorM) were placed in a bipolar configuration with 1cm inter-electrode-distance over the muscle belly perpendicular to fiber orientation of the target muscle after skin preparation based on the ISEK EMG standards ((Merletti, 1999). Palpation and series of isometric contractions for specific muscle groups were used for signal validation within signal representation during testing on a screen in front of the evaluator. The wireless EMG signal was captured at 1200Hz, pre-amplified (500x) using a 16-channel surface EMG Direct Transmission System (Noraxon Inc., USA) (CMRR: 115dB at 60Hz; input impedance of 100 MOhms).

4.1.5 Experimental Setup and Protocol on the SBT

4.1.5.1 Familiarization and Determination of Belt Speed

Prior to the actual protocol and signal recording, a period of familiarization had to be conducted. To do so, participants walked 10 minutes on the treadmill with both belts at equal speeds, without instrumentation (markers and electrodes) to familiarize themselves with treadmill walking and to determine comfortable walking speed. Individuals post-stroke walked initially at about 0.4 m/s and healthy individuals at 1.0 m/s. In both groups, speed was progressively augmented by 0.1 m/s until participants provided verbal feedback that they were walking at a comfortable speed. At that point, speed was augmented by 0.05 m/s until the participants perceived that they were no longer walking at a comfortable speed. This was followed by a reduction of belt speed (0.05 m/s) until the participants confirmed they were walking at a comfortable speed once again. Maximal walking speed was tested subsequently with a similar procedure until the participants confirmed having achieved their maximal walking speed. During the determination of comfortable walking speed, participants were asked to walk without holding the handrails since this was one of the conditions during the main protocol.

4.1.5.2 Experimental Protocols

After the familiarization period, participants were instrumented with the infrared markers and the EMG electrodes. For safety purposes participants wore a harness but without bodyweight support. The testing protocol was based on previous literature which analyzed the effects of SBT walking on the gait pattern in individuals post-stroke and healthy groups (Lauzière et al., 2014a; Malone et al., 2012; Reisman et al., 2005; Reisman et al., 2007). It included three periods: 1) baseline period (tied-belt) followed by 2) an adaptation period with unequal belt speeds (split-belt configuration) and finally 3) a post-adaptation period with a tied-belt configuration identical to the baseline (Figure 4-2). The protocol of adaptation was conducted twice with the stroke group and once with the healthy controls. Participants were asked to walk without holding the handrails during the baseline and post-adaptation periods. They were allowed to hold the handrails during the six minutes of the adaptation period and when required for safety purposes during baseline or post-adaptation. Furthermore, foot positioning on the belts and the participant's position on the treadmill were permanently observed by an assessor facing the participant.

4.1.5.2.1 *Baseline Period*

During the baseline period, participants walked three minutes at tied-belt configuration. The speed of this period was identified as the slow speed. For individuals post-stroke, this corresponded to their comfortable speed and for healthy controls, this corresponded to 70% of their comfortable speed. This is due to the fact that comfortable speed was adjusted in healthy controls to match walking speed of the stroke group to facilitate group comparisons of outcome measures. Individuals post-stroke frequently walk at speeds on average 30% lower than the walking speed of age-matched healthy controls (Patterson et al., 2012; von Schroeder et al., 1995). At the end of the three minutes, biomechanical and EMG signals were registered for 30 seconds (s) from the 90th to the 120th second.

4.1.5.2.2 *Adaptation Period*

The adaptation period of split-belt configuration took six minutes with a belt speed ratio set at 2:1. The slower belt speed corresponded to the belt speed during baseline, i.e., it was at a comfortable speed for individuals in the stroke group and at 70% of speed for individuals in the control group as stated earlier. The speed of the faster belt was twice as fast as the slow belt speed. Individuals post-stroke walked with the non-paretic leg on the faster belt (NP-fast condition). The control group walked with the dominant leg on the fast belt. All individuals were allowed to hold handrails during this period (Reisman et al., 2007). Signals were registered during the following three time points for 30 seconds: 1) early adaptation (EA; 0-30s), 2) mid adaptation (MA; 180s-210s; not analyzed) and 3) late adaptation (LA; 330s-360s). In individuals post-stroke, the protocol was also performed with the paretic leg on the faster belt during adaptation (P-fast condition).

4.1.5.2.3 *Post-adaptation Period*

During the three minutes of post-adaptation, belt configuration was identical to the one during the baseline period (tied-belt). All participants were asked to release the handrails as fast as possible after having started the post-adaptation period. As soon as the handrails were released, signals were registered during the entire period. The first 30 seconds of recording were used for data analysis (early post-adaptation; EPA). At that moment of the protocol, after-effects were observed similar to those seen in previous studies. As mentioned earlier, individuals post-stroke accomplished the protocol of adaptation twice. Participants took a break of at least 10 minutes until they felt well rested prior to the second condition to reduce likelihood of general and muscular fatigue. During this break, participants were allowed to walk around. This should have provided an additional washout of the after-effects on step length induced by the post-adaptation period during the first (NP-fast) condition. Then the three described periods (baseline, adaptation, post-adaptation) were repeated but with the paretic leg on the fast belt (P-fast condition). The data obtained during the periods of adaptation and post-adaptation were always compared with the corresponding baseline (Figure 4-2; Baseline 1 in NP-fast condition, Baseline 2 in the P-fast condition). To quantify the washout of the after-effects between the NP-fast condition and the P-fast condition, data were also analyzed during the last 30 seconds of NP-fast post-adaptation (150-180s).

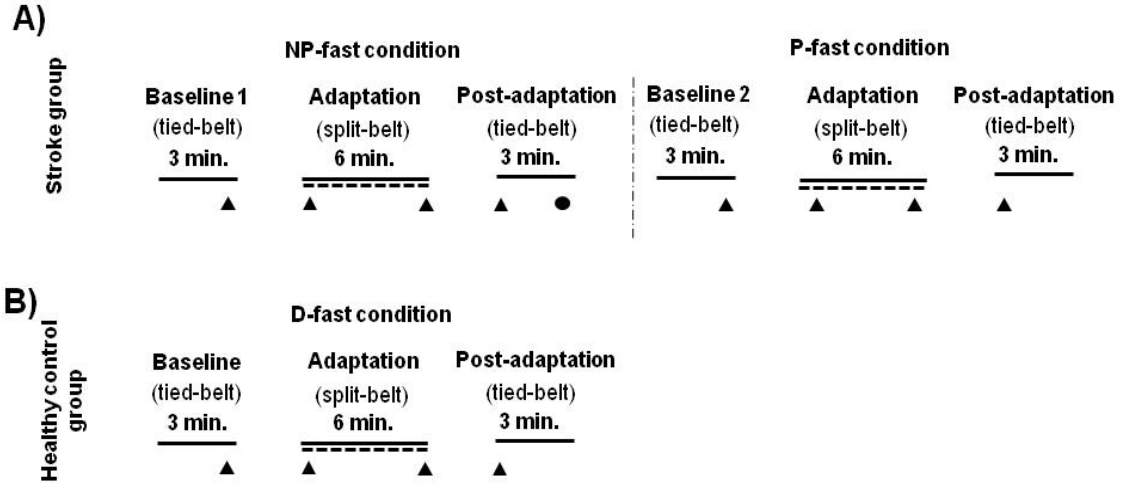

Figure 4-2. Schematic illustration of the experimental protocol; A) Participants in the stroke group walked at two conditions (NP-fast and P-fast condition). B) Healthy individuals conducted the protocol with their dominant side on the fast belt. During the first condition, belt speed was increased on the side of the non-paretic leg (NP-fast condition). During the second condition, the paretic leg walked on the fast belt during Adaptation (P-fast condition). Conditions included baseline, adaptation (split-belt), and post-adaptation. Continuous lines represent the belts running at comfortable speed for individuals post-stroke or at 70% of comfortable speed for the healthy controls. Dotted lines represent the faster belt. The ▲ are time points of signal registration. The ● indicates the registration time at the end of post-adaptation to quantify the washout between after-effects of early post-adaptation during the NP-fast condition and the baseline of the P-fast condition.

4.1.6 Data Analysis

4.1.6.1 Biomechanical Data

In-house software programs developed by the engineers at IRGLM allowed reconstruction of the 3D movements (kinematics) of the segments by using the coordinates obtained from the active infrared markers. Furthermore, in-house programs allowed detection of the force onset and offset produced on the belt by using The Teager-Kaiser Energy Operator (TKEO) (Li, Zhou, & Aruin, 2007; Solnik et al., 2010). The TKEO was calculated by using the variability of the signals. As soon as the variability of the signal exceeded the predefined threshold (average standard deviation of the baseline signal [obtained during standing position]), the signal was considered as being "on". The offset was similarly determined in the opposite manner. The combination of the vertical GRF and the kinematic information allowed calculation of spatiotemporal parameters which were of particular interest considering our cross-sectional study. The analyzed spatiotemporal parameters were 1) step length, 2) cycle duration and 3) double support duration. The marker position of the heel was used to calculate step length. One step was defined as the anterior-posterior distance between the trailing and leading heel markers for two consecutive contacts. Cycle duration was obtained from the time between subsequent contacts of the same foot (Lauzière et al., 2014a; Reisman et al., 2007).

4.1.6.2 Muscle Activity

Offline, EMG signals were first visually inspected and outliers with high content of artefacts or no burst activation were removed (Iglewicz & Banerjee, 2001). In order to minimize the effect of movement artefacts, signals were band-pass filtered using a Butterworth 4th order zero phase filter at 20-400Hz (Knarr et al., 2012; Konrad, 2005; Merletti, 1999). A moving window of 250 ms (300 time points) centered on each signal point was used to calculate the root-mean-square (RMS) values. The resulting RMS curve was expressed as a percentage of the RMS peak value (%max) obtained during walking at the corresponding baseline (Chowdhury et al., 2013; Ricamato & Hidler, 2005; Teixeira-Salmela et al., 2012). To quantify inter-leg coordination of EMG activity, a cross-correlation analysis was used. The cross-correlation analysis (CC) provided a comparison of the shape and timing of muscle activation profiles (Wren et al., 2006). A strong correlation reflected better inter-leg coordination. The signal obtained during a gait cycle was phase shifted over the profile of the same muscle of the contralateral side (Nelson-Wong et al., 2009; Wren et al., 2006). The coefficient (Rxy) corresponding to the highest correlation was then retained. The muscles with large changes in correlation between conditions were further analyzed for signal duration for stance and swing phases where activity was determined when the RMS values exceeded 99.7% deviation from the lowest RMS value of the eight gait cycles (Konrad, 2005). Duration of EMG signal during stance and swing phase of the gait cycle was determined by using individual mean stance duration of the corresponding period. EMG signals were time normalized to 100% of gait cycle. Change values in EMG amplitude and duration were expressed by subtracting the values obtained during baseline from its corresponding post-adaptation value.

4.2 Methods: Pilot Training Study

To achieve the objectives set for the pilot training study, individuals with chronic stroke and residual step length asymmetry were evaluated and trained. This second project was realized in collaboration with the clinicians of the outpatient unit of the Constance-Lethbridge Rehabilitation Center (CRCL). The study was approved prior to the start of the study by the research ethics board of the CRIR.

4.2.1 Participants

Inclusion and exclusion criteria for this second study corresponded to the criteria presented for the cross-sectional analysis (section 4.1.1). Two additional inclusion criteria were added for the training study in comparison to the cross-sectional analysis: 1) participants were required to display a step length asymmetry (step length ratio ≥ 1.08) (Patterson et al., 2010a) caused by a shorter non-paretic or paretic step length, and 2) they had to be able to conduct six minutes of walking without a break and achieve at least 200 m doing so. Once included in the project, participants were informed not to start or follow other training or physical exercise during the study enrolment in order to control factors influencing changes of lower limb symmetry during the training period (2-3 weeks).

4.2.2 Recruitment

4.2.2.1 Recruitment of Participants

Individuals with hemiparesis secondary to a cerebral stroke were recruited from the neurology department of the outpatient units at the CRCL and IRGLM. In addition, eight eligible participants from the cross-sectional study were contacted. Ten participants were considered as sufficient in order to test the feasibility of a protocol within a pilot study (Billingham 2013). This number also corresponded to the number recruited during the time frame set for the collaboration with the clinicians from the outpatient unit (June 2014 - September 2014). Two additional participants had to be recruited between October and November 2014, because of a loss of biomechanical data from two original participants.

4.2.2.2 Recruitment of Physiotherapists

In order to investigate the feasibility of the training and the SBT use, physiotherapists were involved in recruitment and training of participants. All physiotherapists of the neurology department at the CLRC (n = 4) were invited to participate in a one-hour information session. During this session, the relevance, purpose and methods of the study were explained. Three therapists accepted to participate in the information session and the study. The CLRC was in possession of a split-belt treadmill which had only been used for research purposes. Only one of the therapists had worked with a SBT prior to the study.

4.2.3 Clinical Evaluation

The clinical evaluation was conducted at the CLRC or IRGLM depending on convenience for the post-stroke participants. The evaluation was conducted by the four physiotherapists. Clinical assessments were conducted during four time points (Table 4-1): 1) One week prior to the training (pre-evaluation 1), 2) one day prior to the training (pre-evaluation 2), 3) post-evaluation (1-2 days after the last training session) and 4) at the four-week follow-up (follow-up).

TABLE 4-1. Schedule of Clinical and Biomechanical Evaluations.

	Pre-Evaluation 1	Pre-Evaluation 2	Post-Evaluation	Follow-Up
Time point of evaluation	One week prior to training	1-2 days prior to training	1-2 days after the last training session	Four weeks after training
Clinical evaluation	Yes (functional and cognitive assessments; Evaluation of gait ability)	Yes (walking speed)	Yes (evaluation of gait ability)	Yes (evaluation of gait ability)
3D-motion and EMG analysis	No	Yes	Yes	Yes

4.2.3.1 Pre-Evaluation 1

Participants were first informed about study procedures and asked to sign a consent form approved by the research ethics board of the CRIR institutions.

The assessments used to evaluate functional and cognitive deficits were the same used for the cross-sectional analysis (sections 4.1.3.1 and 4.1.3.2). Since the evaluation of gait ability is of particular relevance in this pilot study, the assessment parameters are listed again: Step length symmetry, walking speed (10MWT), endurance (6MWT) and capacity (TUG) were tested during walking over ground. In order to quantify step length and the direction of asymmetry, participants walked over a long paper carpet with light-colored pencils afixed to the backside of the shoe. This led to small coloured points on the paper carpet allowing for measurement of step length.

4.2.3.2 Pre-Evaluation 2

Once included in the project, the gait patterns of participants during walking over ground were assessed. This included a 3D-motion capture and the recording of EMG activity of the lower limbs. Participants were instructed to walk over a 5 m walkway with integrated force plates at comfortable speed. Participants were not informed about the force plates to avoid a modification of their gait pattern. At least four to five valid gait cycles were registered.

For safety purposes, a physiotherapist was always present and walking next to the participants. During the evaluation session, breaks were allowed when required. The entire registration session was filmed by video cameras. After the biomechanical and EMG recordings, markers and EMG electrodes were removed and participants were familiarized with the training protocol. For a total of five minutes, participants walked on the SBT with unequal belt speeds.

4.2.3.3 Post-Evaluation

At maximum 2 days after the last training session, participants were invited to the post-evaluation. During this session, the four clinical gait parameters (step length symmetry [paper carpet]), walking speed (10MWT), capacity (TUG) and endurance (6MWT) were assessed. This was followed by the EMG and 3D-motion recording of gait data over ground at comfortable speed in the laboratory as described in 4.2.3.3.

4.2.3.4 Follow-Up

The follow-up evaluation was conducted four weeks after the last training session. This evaluation included the clinical evaluation of the participants' gait ability, as well as EMG and 3D-motion analysis during walking over ground at comfortable speed as described in 4.2.3.3.

4.2.4 Biomechanical and EMG Analysis

4.2.4.1 Instrumentation for Biomechanical Analysis and Recording of Muscle Activity

Marker and electrode placement for EMG and 3D-motion analysis was identical to the instrumentation procedure of the cross-sectional analysis to obtain kinematic data and muscle activity as described in sections 4.1.4.2 and 4.1.4.3. In contrast to the cross-sectional studies, the signals were recorded while participants walked over ground instead of on the treadmill. Therefore, GRFs were captured by three AMTI (Advanced Mechanical Technology, Inc. USA) force plates integrated into the floor. Furthermore, for EMG analysis, each electrode position was marked on the participant's skin with a pencil during the first evaluation session to reproduce electrode placements each session.

4.2.5 Protocol of the Repeated Split-belt Walking

The SBT training consisted of six sessions evenly distributed over two to three weeks. The protocol was based on previous studies which analyzed the effects of a single session (Lauzière et al., 2014a; Malone & Bastian, 2010; Vasudevan & Bastian, 2010) and repeated exposure (Reisman et al., 2013) to SBT walking.

4.2.5.1 Preparation

Prior to each training session, participant's heart rate (Polar electro Oy) and blood pressure were taken at rest. The age related maximum heart rate was calculated using the Karvonen Formula (ACSM, 2013). The upper limit of the heart rate was set at 80% of the maximal heart rate as suggested by the Guidelines of the American Heart Association (Fletcher et al., 2001). This was followed by the determination of the participant's comfortable walking speed. The walking speed was determined each session with the same method as described in section 4.1.5.1. The adjustment of the participant's comfortable speed at each training session was based on the findings by Vasudevan and colleagues (Vasudevan & Bastian, 2010) that showed the largest magnitude of after-effects was found when the speed of the slower belt was close to comfortable walking speed compared to faster or slower speeds.

4.2.5.2 Training

The training consisted of 20 minutes of walking during a split-belt configuration with belt speeds set at a ratio of 2:1. Participants initially walked on the treadmill with both belts set at equal speeds. After some steps, belt speed on the side with the shorter step length was doubled in a progressive manner until a 2:1 ratio was attained. If the participant did not achieve the 2:1 ratio, an overall reduction in speed of both the slow and fast belts was made in order to enable the participant to walk at the intended 2:1 ratio. Participants were allowed to take breaks if their heart rate exceeded 80% of their maximal heart rate or they reported an exertion of 5 on the Borg rating scale which represents severe exertion (Borg et al, 1982). The training was restarted when the participant's heart rate lowered back to its level at rest. In the case of fatigue, participants were allowed to stop the training before 20 minutes had elapsed. However, participants were required to maintain at least six minutes of locomotion as this represents the shortest duration tested in the literature that led to successful adaptation and after-effects (Lauzière et al., 2014a; Vasudevan & Bastian, 2010).

4.2.5.3 Post-training

After the training, participants walked three minutes with tied-belt configuration (ratio 1:1). This period served as a cool down and to prepare for the next task of walking over ground after the SBT training that consisted of a repeated perturbation over 20 minutes as previously described. No form of verbal reinforcement was applied to promote the use of a symmetrical walking pattern in participants.

4.2.6 Symmetry Ratios

The ratios used to calculate step length symmetry during the cross-sectional analysis differed from the ratio used for the training study for statistical reasons. In the cross-sectional analysis, we tested the effects of two conditions, one with the non-paretic (NP) side walking on the faster belt and the other with the paretic (P) side on the faster belt, on paretic and non-paretic step length. Using the ratio NP/P allowed us to analyze the effect changes in paretic or / and non-paretic side on step length symmetry.

However, as previously described during the training study, all participants were trained with the side displaying a shorter step length on the faster belt. Thus, the "shorter step" referred always to the side trained on the faster belt and vice versa for the "longer step". Consequently, to calculate the step length asymmetry for the training study, the side with the longer step length was divided by the side with the shorter step length. This allowed us to identify the effect of fast or slow belt walking on improvements of symmetry.

4.3 Statistical analysis

The statistical analyses used in the two projects presented in this thesis are not described in this section. A detailed description of the statistical analyses can be found in the three papers presented in section 5.

Chapter 5. Results

The results of this doctoral work are presented in three scientific papers (papers #1 to #3). Among these, two have been submitted and are currently under revision. Paper #3 should ideally be submitted after papers #1 and #2 have been accepted for publication since its content is linked to the first two papers. The results section includes also supplementary data at the end of the corresponding paper. These results will be presented at the end of each section which presents the corresponding paper. The scientific papers of this doctoral work are the following:

#1. **Betschart M**, Lauzière S, Miéville C, McFadyen BJ, Nadeau S. Changes in lower limb muscle activity after walking on a split-belt treadmill in individuals post-stroke. Paper submitted to the Journal of Electromyography and Kinesiology, April 22, 2016.

#2. **Betschart M**, McFadyen BJ, Nadeau S. Repeated split-belt treadmill walking improved gait ability in individuals with chronic stroke: a pilot study. Paper submitted for the second revision to the Journal of Physiotherapy Theory & Practice, March 29, 2016.

#3. **Betschart M**, McFadyen BJ, Nadeau S. Quantification of lower limb muscle activity and joint moments underlying the reduction of step length asymmetry over ground after repeated split-belt treadmill walking. Paper in preparation for submission to the Journal of Rehabilitation Research & Development.

In the following sections, each of the three papers presented is preceded by a short text explaining the contribution of each author and a preamble highlighting the underlying purpose and relevance of the study. Tables and figures are presented prior to the reference list of the respective scientific papers.

5.1 Paper #1: Changes in Lower Limb Muscle Activity After Walking on a Split-belt Treadmill in Individuals Post-stroke.

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Paper submitted to the Journal of Electromyography and Kinesiology, April 22, 2016.

As principle author, I confirm having contributed to the majority of the manuscript and the procedures required to obtain the results presented in this paper. These procedures include the clinical evaluation of participants with stroke and healthy controls, collection, treatment and interpretation of EMG and biomechanical data. Beyond my contribution, Séléna Lauzière and Carole Miéville were particularly involved in the recruitment of participants, protocol development, clinical evaluation of the participants, and data collection. Bradford J. McFadyen provided advice for the treatment of EMG signals, interpretation of results and was involved in revision of the manuscript. Sylvie Nadeau supervised and contributed to each of these procedures starting from the protocol development to the final revision of the manuscript.

5.1.1 Preamble

The present paper aimed to provide novel insight into the mechanisms involved in SBT-induced adaptation and after-effects in individuals who suffered a stroke. As will be described in the paper’s introduction, the capacity of stroke survivors to adjust their gait pattern to an error-augmentation-based SBT protocol was studied by means of the analysis of spatiotemporal parameters, joint moments, and muscular effort (Reisman et al, 2007; Malone et al, 2013; Lauzière et al, 2014, 2015, 2016). Nevertheless, none of these studies analyzed the lower limb muscle activity involved in these adjustments of spatiotemporal parameters or joint moments. As described in the literature review, stroke survivors frequently suffer from different neuromuscular deficits which impede or prevent their ability to walk. Therefore, it was considered as pertinent to evaluate the effects of SBT walking on lower limb muscle activity. This paper will provide first investigation of changes in muscle activity underlying SBT-induced after-effects on step length in individuals with chronic stroke.

5.1.2 Abstract

Background: There is growing evidence that stroke survivors can adapt and improve step length symmetry in the context of split-belt treadmill (SBT) walking. However, less knowledge exists about the strategies involved in such adaptations. This study analyzed lower limb muscle activity in individuals post-stroke related to SBT-induced changes in step length. ***Methods***: Step length and surface EMG activity of six lower limb muscles were evaluated in individuals post-stroke (n = 16) during (adaptation) and after (after-effects) walking at unequal belt speeds. ***Results***: During adaptation, significant increases in EMG activity were mainly found in proximal muscles ($p \leq 0.023$), whereas after-effects were observed particularly in the distal muscles. The plantarflexor EMG increased after walking on the slow belt ($p \leq 0.023$) and the dorsiflexors predominantly after walking on the fast belt ($p \leq 0.017$) for both, non-paretic and paretic-fast conditions. Correlation analysis revealed that after-effects in step length were mainly associated with changes in distal paretic muscle activity ($0.522 \leq r < 0.664$) but not with functional deficits. Based on our results, SBT walking could be relevant for training individuals post-stroke who present shorter paretic step length combined with dorsiflexor weakness or individuals with shorter non-paretic step length and plantarflexor weakness.

Keywords: Stroke, Split-belt treadmill, Muscle activity, After-effects

5.1.3 Introduction

Split-belt treadmill (SBT) walking has shown promising results in short- and long-term reduction of step length (SL) symmetry after stroke (Reisman et al., 2007; Reisman et al., 2013; Lauzière et al., 2014). The SBT was used to disrupt the walking pattern causing the individual to adjust (Bastian, 2008). Current knowledge suggests that such disruption of a well-learned movement pattern (e.g., walking) is interpreted as an "error" by feedback and -forward mechanisms of the neuromuscular system. To correct this error, interlimb parameters such as SL were progressively adapted trial-by-trial towards the initial symmetry (Malone et al., 2014). When the belts were returned to equal speeds, the modified walking pattern was retained during the first steps (after-effect) and gradually de-adapted towards the initial asymmetry. In all study cohorts, the after-effects on SL were consistent in the direction of change. More precisely, individuals post-stroke with and without asymmetry (Reisman et al., 2007; Lauzière et al., 2014; Malone et al., 2014) and healthy controls (Reisman et al., 2005) showed an increase of SL on the side that walked on the fast belt (fast leg) and a reduction for the slow leg. In individuals post-stroke with initial SL asymmetry, this led to a more symmetrical gait (Reisman et al., 2007; Reisman et al., 2013). Such improvements in SL symmetry, that are considered as more resistant to conventional gait rehabilitation when compared to walking speed and dynamic balance (Patterson et al., 2015), still need to be better understood with respect to underlying mechanisms. The analysis of lower limb muscle activity and ground reaction forces in healthy individuals suggested that during split-belt walking, stiffening the ankle was a key feedforward strategy to accommodate the new walking demand (Ogawa et al., 2014). Whether this strategy is also used in individuals post-stroke with a high incidence of impaired muscle function (Hafer-Macko et al., 2008; Arene et al., 2009) is unknown. Indeed, knowledge about the strategies individuals post-stroke use in line with SL adaptation and after-effects is still marginal. A recent cross-sectional analysis of SBT-induced after-effects on joint moments in individuals post-stroke and healthy controls found that the after-effects were most pronounced in increased plantarflexion moments as compared to hip and knee moments when walking on the slower belt (Lauzière et al., 2014). These changes were associated with the changes in contralateral leg SL ($p \leq 0.015$). Despite different sensorimotor deficits and SL asymmetries, after-effects in joint moments were consistent among individuals post-stroke and in accordance with healthy controls. Adding knowledge about the effects of SBT on bilateral muscle activity could also provide evidence for using this approach as a clinical intervention (Reisman et al., 2013). The objective of this study was to assess the changes in muscle activity during and after split-belt walking in individuals post-stroke with and without SL asymmetries in order to gain insight into the contribution of muscle activity for after-effects in SL. We hypothesized that muscle activity will show a consistent pattern in changes such as the one found for SL, particularly for distal muscles as seen in joint moments (Lauzière et al., 2014).

5.1.4 Methods

5.1.4.1 Participants

Sixteen participants with their first cerebral stroke six months or more prior were recruited (Table 5-1). Participants had to be able to walk 10 meters at 0.5 m/s or greater without technical or physical assistance. Exclusion criteria were medical and cognitive conditions that could affect walking capacity verified by medical records. All participants provided signed consent and the protocol was approved by the local ethics committee.

5.1.4.2 Clinical Evaluation

Participants' comfortable and fast over-ground walking speeds and functional mobility were assessed with the 10 Meter Walking (Flansbjer et al., 2005) and the Timed Up and Go (Bohannon, 2006) tests respectively. Lower limb motor recovery was evaluated with the Chedoke-McMaster Stroke Assessment (Gowland et al., 1993). Resistance to passive movement was tested for the ankle muscles using the Ashworth Scale (Pandyan et al., 1999).

5.1.4.3 Experimental Setup

The present protocol was based on previous investigations (Reisman et al., 2007; Lauzière et al., 2014). Participants were exposed to two conditions involving baseline, adaptation and post-adaptation periods. Participants walked with a harness without bodyweight support. Baseline and post-adaptation periods both consisted of three minutes walking at comfortable speed (tied-belts). Participants were asked to not hold the handrails during these conditions. During the adaptation period, participants walked six minutes with belt speed doubled (2:1 ratio of fast:slow) on either the paretic or non-paretic side. For safety, participants were allowed to hold the handrails. During the first condition, the non-paretic leg walked on the fast belt during adaptation (NP-fast condition) and for the second condition, the paretic leg walked on the fast belt during adaptation (P-fast condition). To ensure sufficient washout of the after-effects on SL, participants walked three minutes during the post-adaptation period followed by a break of at least 10 minutes or until they felt well rested prior to the second condition.

5.1.4.4 Data Collection

Muscle activity was recorded bilaterally for the: tibialis anterior (TA), gastrocnemius lateralis (GL), vastus lateralis (VL), rectus femoris (RF), semitendinosus (ST), and gluteus medius (GLM). Self-adhesive surface electrodes (Ag/AgCl; Ambu_BlueSensorM) were placed in a bipolar configuration with 1cm inter-electrode-distance over the muscle belly perpendicular to fiber orientation of each muscle after skin preparation (Merletti, 1999). Wireless EMG signals were captured at 1200Hz, pre-amplified (500x) using a 16-channel surface EMG Direct Transmission System (Noraxon Inc., USA) (CMRR: 115dB at 60Hz; input impedance of 100 MOhms).

Kinematic data were captured at 30 Hz in three dimensions (Optotrak Certus Motion Capture System, NDI) from 34 markers placed on the feet, shanks, thighs and pelvis and resampled to 60 Hz. The forces exerted on the instrumented treadmill (Bertec Corp. USA) were also collected (600 Hz) and resampled to match the kinematic data. Registration of EMG, kinematic and kinetic signals was conducted during 30 seconds (s) at the end of baseline (90 - 120s), early (0 - 30s) and late adaptation (330 - 360s) and during early post-adaptation as soon as the participant released the handrails. For participants using the handrails (90% of participants), the maximum duration of handrail use was less than two gait cycles. During the first condition, data were also collected during the last 30 seconds of post-adaptation (150-180s) to quantify the washout of the after-effects (late post-adaptation).

5.1.4.5 Data Analysis

Mean data used for analyses were obtained from the first ten consecutive gait cycles of each 30-second period recorded. Vertical ground reaction forces were used to detect initial foot contact and toe off using the Teager-Kaiser energy operator (Li et al., 2007). Cycle duration was obtained from the time between subsequent contacts of the same foot (Reisman et al., 2007; Lauzière et al., 2014). SL was defined as the anterior-posterior distance between consecutive trailing and leading heel markers. SL ratio was calculated by dividing the non-paretic SL by the paretic SL with "1" representing perfect symmetry. Offline, a pre-processing of raw signals was conducted by removing visually detected artefacts with a two-stage peak detection algorithm using custom LabView^{TM} programs (O'Keeffe et al., 2001). Signals were then band-pass filtered (Butterworth 4th order zero phase filter at 20 - 400Hz). Root-mean-square (RMS) values were created using a 250 ms moving window and the resulting curve was expressed as a percentage of the RMS peak value (Lamontagne et al., 2000). EMG signals were time normalized to 100% of the gait cycle.

5.1.5 Statistics

Descriptive statistics were computed for all variables. SL and EMG data were tested for normal distribution (Shapiro-Wilk Test; (Razali et al., 2011)). Repeated-measures ANOVA was used to test the effect of period (baseline, early and late adaptation, early post-adaptation), condition and side (non-paretic, paretic) with a significance level at *alpha* = 0.05 which was adjusted by using a Bonferroni correction for post-hoc multiple comparisons (SPSS 21; IBM Corporation). Data from post-adaptation were always compared with the corresponding baseline data (NP-fast or P-fast). The statistical model to test the washout of the SL after-effects included only the periods of the NP-fast condition and the late post-adaptation. Pearson's and Spearman's correlation coefficients were used to evaluate the association of changes in muscle activity from baseline to the early post-adaptation with changes in SL and clinical parameters, respectively.

5.1.6 Results

Participants' demographic information and gait characteristics are presented in Table 5-1. The results of SL and muscle activity expressed as mean and standard deviation refer to the side that walked on the slow or fast belt during adaptation, respectively.

The repeated-measures ANOVA revealed significant interactions among factors (Condition x Period x Side) for SL (p = 0.000) and muscle activity (p = 0.012), except for the dorsiflexors (TA). Consequently, posthoc analyses (factor period) were conducted for paretic and non-paretic muscles and SL during NP-fast and P-fast conditions, independently.

5.1.6.1 Step Length

5.1.6.1.1 *Adaptation*

During early adaptation of the NP-fast condition, participants presented a bilateral increase of SL ($p \leq 0.003$) particularly on the slow side (paretic; +14 cm) (Figure 5-1 A-B; left panel, dotted lines). The ratio became more asymmetrical (1.16 (0.22)) but without approaching significance (p = 0.286) (Figure 5-1 C). After five minutes of walking, both sides differed from baseline SL ($p < 0.001$) as a result from an increased fast (non-paretic) SL and a maintained slow (paretic) SL. Consequently, the ratio returned towards symmetry (0.98 (0.17)). The P-fast adaptation (Figure 5-1; right panel) also induced a bilateral increase of SL ($p < 0.001$) during early adaptation which remained different from baseline after 5 minutes of walking ($p \leq 0.002$). Ratios of the P-fast condition during adaptation did not differ from the baseline ratio (p = 1.000).

5.1.6.1.2 *Post-adaptation*

Both conditions led to a decrease of slow SL ($p \leq 0.001$) after split-belt walking (early post-adaptation) when compared to baseline (Figure 5-1 A). Fast SL remained significantly increased during the NP-fast condition (p = 0.000) and showed a tendency to remain increased during P-fast condition (p = 0.076). During NP-fast walking, inverted after-effects can be observed when comparing the SL ratios of the early adaptation (0.94 (0.15)) with the early post-adaptation (1.16 (0.22)) while in P-fast, the ratio decreased with corresponding values of 1.00 (0.17) and 0.78 (0.20). SL symmetry improved in most of the participants walking with their shorter SL on the fast belt (n = 10/12). The after-effects found during the first condition (NP-fast) were all washed out before the second condition began since values returned to baseline during late post-adaptation for the paretic (p = 0.533) and non-paretic sides (p = 1.000) as well as for the SL ratio (p = 0.268).

5.1.6.2 Muscle Activity

5.1.6.2.1 *Adaptation*

The NP-fast condition lead to an increase in slow (paretic) VL (+24%max) and RF (+26%max) ($p \leq 0.021$) (Figure 5-2). At late adaptation, no statistical differences to baseline activation level were found in any muscle. During the P-fast condition, the slow (non-paretic) VL (+32%max; p = 0.000) and fast RF (+31%max; p = 0.023) increased in activity during early adaptation (Figure 5-3). After five minutes of split-belt walking the slow VL remained increased still with 26%max activity higher than baseline (p = 0.001). Also, the increase observed in early adaptation for the slow RF (+34% max) became significant at late adaptation (+26%max; p = 0.003). During early adaptation of both conditions, fast GL increased on average 26%max (NP-fast) and 50%max (P-fast) but without statistical significance ($p \geq$ 0.072). The muscles not described (TA, ST, and GLM) presented negligible amounts of changes and comparisons with baseline.

5.1.6.2.2 *Post-adaptation*

Both conditions showed increased fast VL and TA in combination with an increased slow GL ($p \leq 0.025$) during early post-adaptation which gradually returned to baseline activity (Figures 5-2 and 5-3). During the P-fast condition, an increase in activity was additionally found in the fast RF (26%max; p = 0.035) and the increase in the slow ST approached significance (16%max; p = 0.056).

5.1.6.3 Correlation of After-effects

Bilateral after-effects on SL found during the NP-fast condition were moderately associated with increased paretic GL activity during early post-adaptation ($p \leq 0.022$; r > 0.60) (Figure 5-4 A). During the P-fast condition, the after-effects on paretic SL was correlated with the increase of the paretic TA (r = 0.570; p = 0.021) (Figure 5-4 B1) and was also associated with SL ratio changes (r = -0.522; p = 0.038). The non-paretic ST was negatively associated with after-effects on non-paretic SL (r = -0.584; p = 0.017) and the SL ratio (r = 0.577; p=0.019). No correlations were found for VL, RF and GLM and no muscle activity correlated with any clinical parameters.

5.1.7 Discussion

The present study analyzed changes in lower limb muscle activity associated with SBT-induced after-effects on SL in individuals post-stroke. The increase or decrease of after-effects on SL and muscle activity was consistent among participants despite different initial asymmetries and functional deficits. In accordance with previous findings, the current protocol led to a longer SL for the side walking on the faster belt and a decrease on the side walking on the slow belt when compared to baseline (Reisman et al., 2007; Lauzière et al., 2014). A related increase in EMG of slow plantarflexors (GL), fast dorsiflexors (TA) and fast muscles controlling the knee and hip (VL, RF) was found and the changes in GL and TA activity were associated with after-effects in SL and SL ratio.

5.1.7.1 After-effects in Muscle Activity

During both conditions fast VL, fast TA, and slow GL showed increased EMG activity immediately after split-belt walking (early post-adaptation) when compared to baseline. The increase in slow plantarflexor EMG during NP-fast condition and the association found with SL after-effects support the findings by (Lauzière et al., 2014). Using the same protocol, an increase in average group ankle plantarflexion joint moment after slow belt walking was found during both conditions. The increase in slow plantarflexion moment was moderately associated with the increase in contralateral (fast) SL during NP-fast (r = 0.559) and P-fast conditions (r = 0.535) (Lauzière et al., 2014). Thus, it was hypothesized that the slow plantarflexors contribute to contralateral changes in SL. However, in the present study, during P-fast walking, the correlation with the slow GL was negligible (r = 0.005). SL changes correlated with the changes in fast (paretic) TA and slow (non-paretic) ST EMG. The distal contribution of the paretic GL and TA to changes in SL is interesting considering that about 60% percent of our participants showed deficits in active plantar- and dorsiflexion (Table 5-1). Thus, despite these deficits, participants showed the capacity to modify distal muscle activity in combination with SL after-effects. However, impaired distal paretic lower limb function might account for the differences of EMG and SL changes found between the two conditions (NP-fast and P-fast). The P-fast condition led to smaller after-effects with less increase in fast SL compared to NP-fast condition. Thus, during the P-fast condition, slow GL might still contribute to an increase of fast SL as seen in the NP-fast condition, but probably not sufficiently to create significant increases from baseline in fast (paretic) SL. This could explain the additional implication of proximal muscles (slow ST and fast RF). Together with the hip extensor gluteus medius, the ST muscle is considered to contribute to forward propulsion in the first half of stance (Higginson et al., 2006; Allen et al., 2014). Based on visual inspection, an accentuated increase in ST activity can be found during 0 - 25% of the gait cycle (Figure 5-6, section 5.1.13.1). Thus, the increase in slow ST could be explained as a compensation for the reduced contribution of slow plantarflexor muscles for support during stance and forward propulsion (Higginson et al., 2006; Allen et al., 2014). Previous analysis of hip joint moments revealed an increase in hip extension moment only after fast belt walking in both – stroke and healthy control – groups control groups (Lauzière et al., 2014). However, explaining changes in slow ST by joint moments needs to be done with caution because of its biarticular and biphasic function and the different external and internal forces influencing hip joint moments.

In terms of fast RF activity, the increase was mainly observed during stance phase (0 - 50% of the gait cycle, Figure 5-6, section 5.1.13.1) when the knee extensors are needed to support and decelerate the body and subsequently provide forward acceleration of the trunk at a late stance (Zajac et al., 2003). Therefore, the paretic RF muscle could compensate for reduced forward propulsion by the non-paretic plantarflexors. However, interpretations considering the RF activity obtained from surface EMG need to be considered with care. It has been emphasized that surface EMG of the RF muscle during early stance results mainly from VL crosstalk (Nene et al., 2004; Barr et al., 2010). In the present study, an attempt was made to avoid crosstalk between muscles and we are confident that RF muscle results are not coming from the VL. Furthermore, it is considered unlikely that the EMG after-effects observed during the P-fast condition were particularly influenced by muscular fatigue or the after-effects produced during the NP-fast condition since participants were given a break including walking around and sitting until they felt well rested. Furthermore, after-effects are considered as real effects since the values during post-adaptation were compared with the corresponding baseline values for the two conditions.

5.1.7.2 Adaptation

A main observation was that during both conditions, the muscles which presented after-effects during early post-adaptation did not increase activity during the adaptation period. During both conditions, increases in slow, proximal muscles (RF & VL) reached significance at early adaptation, but after-effects were found in the opposite muscles (fast VL & RF). The GL muscles showed a similar inverted pattern between early adaptation and early post-adaptation during both conditions. Inverted after-effects were found in SL during NP-fast condition in accordance with previous investigations on locomotor adaptation (Reisman et al., 2005; Reisman et al., 2007; Malone et al., 2014). Secondly, despite similar after-effects in muscle activity during both, NP-fast and P-fast post-adaptation, a phenomenon of adaptation was only found during NP-fast condition. During this condition, SL symmetry adapted towards baseline which was not observed during the P-fast condition. This indicates that the adaptation of muscle activity could be related to the symmetry of inter-limb parameters. More precisely, when SLs were more symmetrical compared to baseline, muscle activity returned towards baseline. Did the system recalibrate muscle activity based on the symmetry of inter-limb parameters and consider the pattern during late adaptation as the new “correct pattern”? Such adaptation of proximal muscles was found in healthy individuals during 10 minutes of split-belt walking (ratio 2:1) (Ogawa et al., 2014). The theory of “recalibration” of muscle activity could be supported with the current explanation of the neural mechanisms of locomotor adaptation. Based on feedback and -forward internal representation of body dynamics, the walking pattern is (re-) calibrated to adjust for the consistent perturbation (Bastian, 2008).

Another strategy in adjusting muscle activity to adapt to split-belt walking suggested by (Ogawa et al., 2014) for healthy individuals is to stiffen the ankle as a feedforward control mechanism. This was based on an increase of fast TA activity during early stance which was in line with the increase and adaptation of braking forces. In the present study, large variability in muscle activity and only significant changes in proximal muscles do not allow conclusions of a stiffening strategy. In general, the large variability of EMG activity during adaptation requires a careful interpretation of the data. The fact that participants held the handrails during adaptation, but not during post-adaptation, probably accounts for the increased variability of EMG during adaptation compared to post-adaptation. In healthy individuals, unloading 15% of body weight leads to significant reduction in TA, GL and VL muscle activity (Fischer et al., 2015). Based on vertical GRFs here, variability was observed with the use of handrails while walking. However, despite large variations in muscle activity during adaptation, a consistent pattern of increased muscle activity was seen in line with after-effects during early post-adaptation.

It still remains unclear how muscle activity would change after repeated exposure to split-belt treadmill walking. Long-term effects could be found in muscles that show after-effects during post-adaptation (tied-belt) since these were in line with changes in SL and SL symmetry. Yet, the protocol used in the investigation of repeated exposure to split-belt walking consisted of 30 minutes of walking during adaptation (Reisman et al., 2013). Thus, such a protocol could lead to an increase of activity in muscles that change during the adaptation period. These assumptions require further research to investigate effects of repeated exposure on split-belt walking on muscle activity in order to improve knowledge about evidence-based therapy approaches to the recovery of gait symmetry and muscle function post-stroke.

5.1.7.3 Limitations

In the present study, the main goal was to provide information about the effects of muscle activity in a general group of individuals with chronic stroke. A subgroup analysis of individuals with a shorter paretic SL or non-paretic SL could be conducted to analyze the changes in muscle activity regarding SL asymmetry. Also, participants held the handrails during the period of adaptation for safety purposes. This probably led to the large variability in muscle activity as mentioned previously. Further analysis including subgroups and biomechanical parameters will be necessary to better understand the changes in muscle activity during the period of adaptation.

5.1.8 Conclusion

Increases in fast SL were in line with changes in slow plantarflexor and fast dorsiflexor activity. Additional changes were found for proximal muscles on the fast side. The frequently impaired paretic plantarflexor increased activity during the NP-fast condition and paretic TA activity increased during the P-fast condition. Further, P-fast walking might require more contribution from proximal muscles to overcome the split-belt induced changes. The findings concerning after-effects were generally consistent among a group of individuals post-stroke with different SL asymmetries and distinct deficits in lower limb function. In contrast, large variances were found during the period of adaptation indicating that individuals post-stroke use different strategies to overcome the SBT-induced repetitive disruption of their gait pattern. Based on our results, split-belt treadmill walking could be relevant for training individuals post-stroke who present shorter paretic step lengths combined with dorsiflexor weakness, or individuals with shorter non-paretic step length and plantarflexor weakness.

5.1.9 Acknowledgments

We would like to acknowledge the participants of our study and thank Philippe Gourdou, Michel Goyette, Youssef El Khamlichi and Daniel Marineau for their technical support.

5.1.10 Declaration of Conflicting Interests

There are no conflicts of interest to disclose.

5.1.11 Funding

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TABLE 5-1. Patients Demographics (N = 16).

	Demographics		**Impairments**		**Gait characteristics**		
Participants	Age (years)	Time post-stroke (months)	Chedoke (Leg/Foot) Range: 0-7	Ashworth Scale 0-4	Treadmill CGS/FCGS (m/s)	10MWT CGS (m/s)	Step length symmetry (NP/P)
P1[f]	53	8	5/3	0	0.90/1.20	1.27	1.02
P2[f]	24	131	5/5	3	0.85/0.95	1.18	0.87
P3_R	55	75	4/4	0	0.85/1.35	1.40	0.93
P4	41	7	3/3	0	0.60/1.00	0.60	1.15
P5[f]	38	150	6/5	1	0.55/0.74	0.60	1.14
P6[f]	63	11	6/4	1	0.60/0.65	0.56	0.94
P7[f]	69	98	3/4	1	0.40/0.50	0.61	0.87
P8_R	60	10	6/1	1	0.45/0.60	0.81	0.64
P9	54	30	6/5	1	0.53/0.80	0.91	0.87
P10	50	183	6/5	1	0.55/0.95	1.09	1.12
P11	46	6	7/7	0	0.80/1.40	1.29	1.09
P12[f]	66	65	3/3	1	0.40/0.55	0.51	0.99
P13	44	58	5/2	3	0.60/0.85	0.65	0.69
P14	39	31	5/4	2	0.60/1.00	1.08	0.93
P15_R	29	18	7/7	0	0.70/1.20	1.20	1.00
P16	66	10	6/5	0	0.55/0.70	0.98	0.80
Mean (SD)	**49.8 (13.4)**	**78.7 (12.5)**			**0.62 (0.16) /0.90 (0.28)**	**0.92 (0.30)**	**0.94 (0.15)**
Median			**5/4**	**1**			

Abbreviations: CGS = comfortable gait speed; FCGS = fast comfortable gait speed; m = meter; s = second; 10MWT = 10 Meter Walk Test; P = paretic, NP = non-paretic; SD = standard deviation; f = female; R = Right hemiparesis.

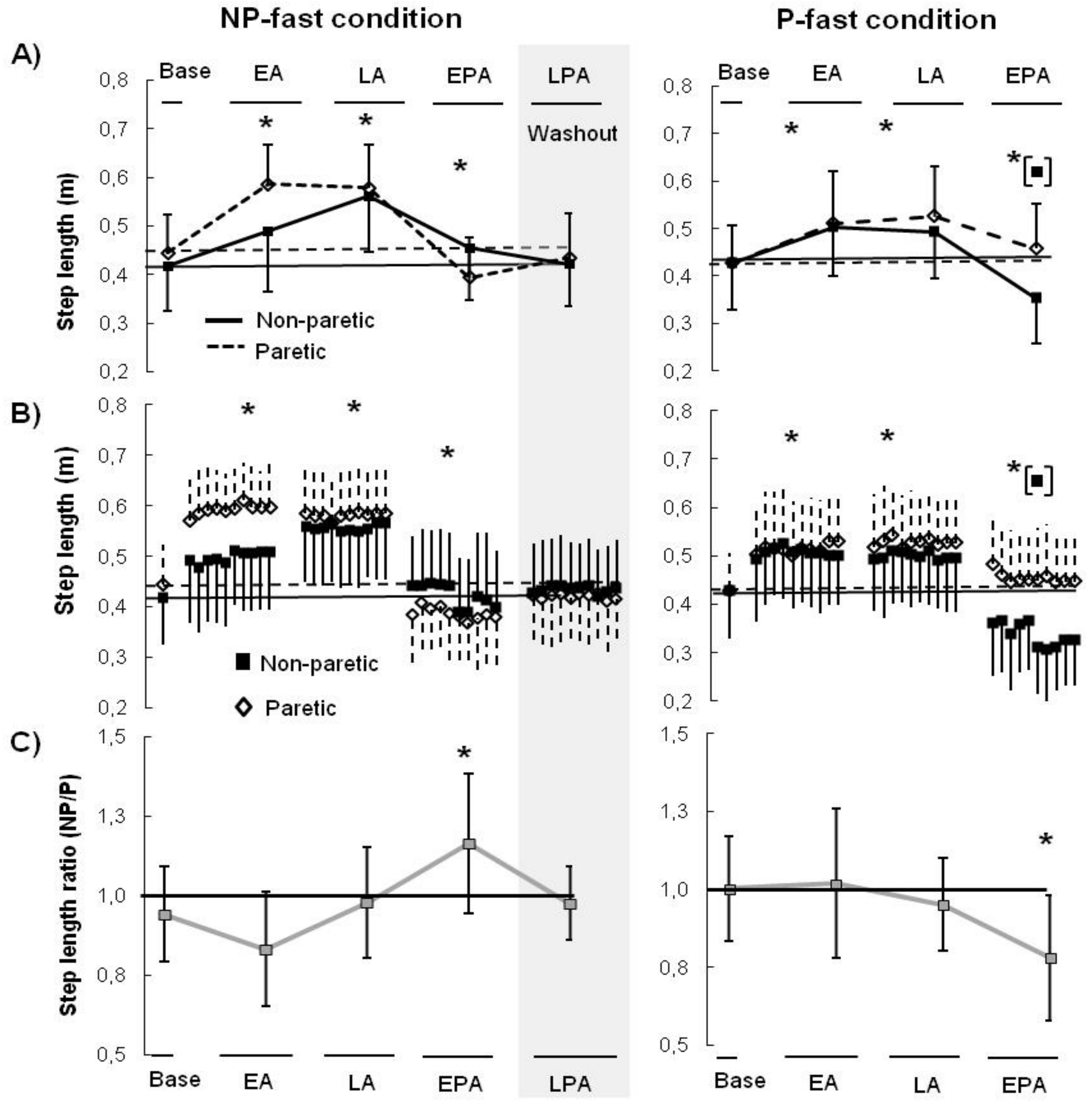

Figure 5-1. Group mean of step length (SL) for the paretic (A dotted line; B ◊) and non-paretic (A solid line; B ■) limbs as well as SL ratio (C, 1.0 = perfect symmetry) during baseline (Base), early adaptation (EA), late adaptation (LA) and early post-adaptation (EPA) of NP-fast and P-fast conditions. Late post-adaptation (LPA) presents the washout of after-effects between conditions. Standard deviations are shown with dotted (paretic side) and continuous lines (non-paretic). *significant difference to baseline after Bonferroni correction; [■] denotes significance only on the non-paretic side.

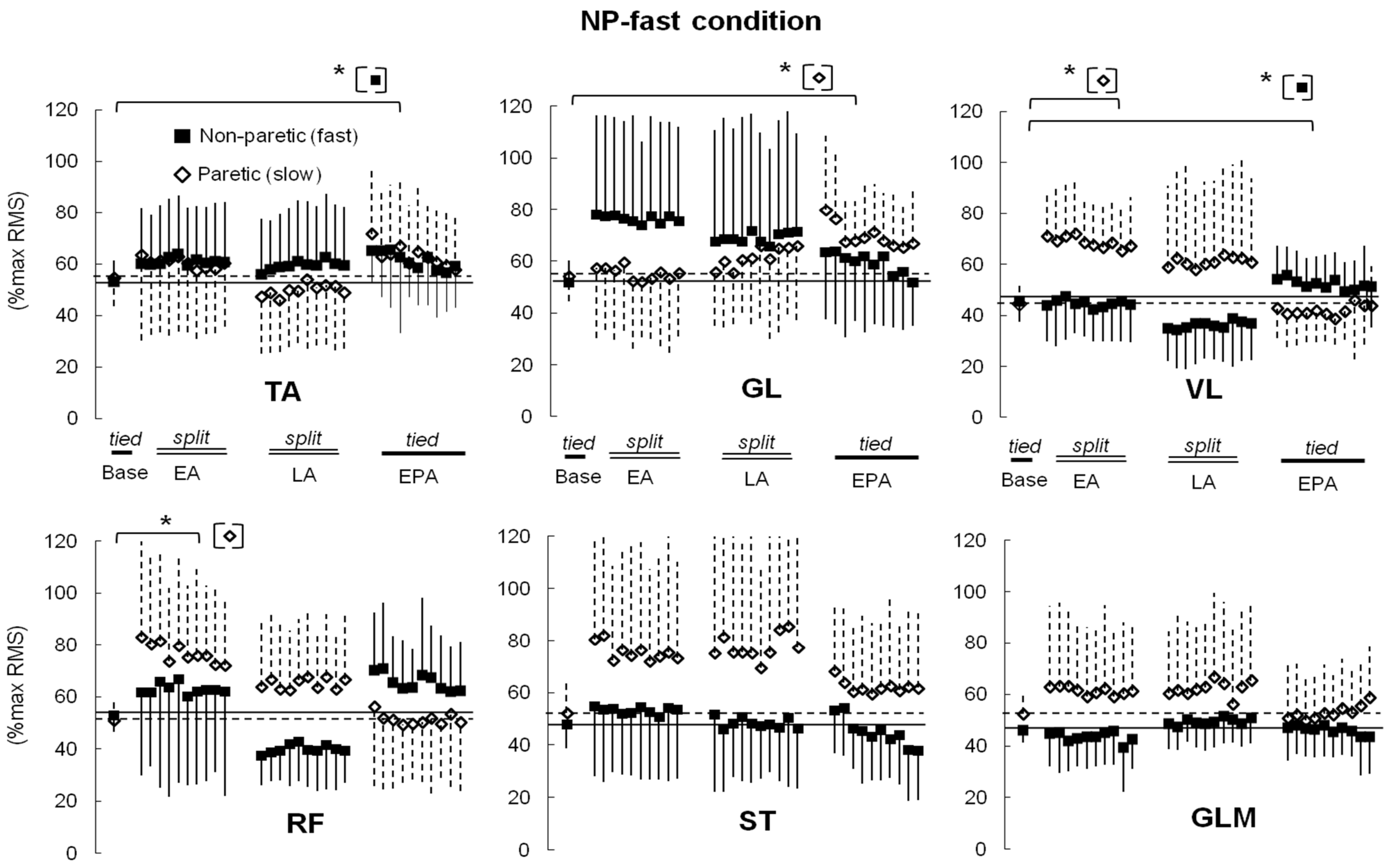

Figure 5-2. Group mean and standard deviations of normalized EMG for 10 consecutive gait cycles of non-paretic (NP) (■) and paretic (P) (◊) muscles during the NP-fast condition for early adaptation (EA), late adaptation (LA) and early post-adaptation (EPA). *Indicates significant changes when compared to baseline. Abbreviations: tibialis anterior (TA), gastrocnemius lateralis (GL), rectus femoris (RF), semitendinosus (ST), vastus lateralis (VL), and the gluteus medius (GLM).

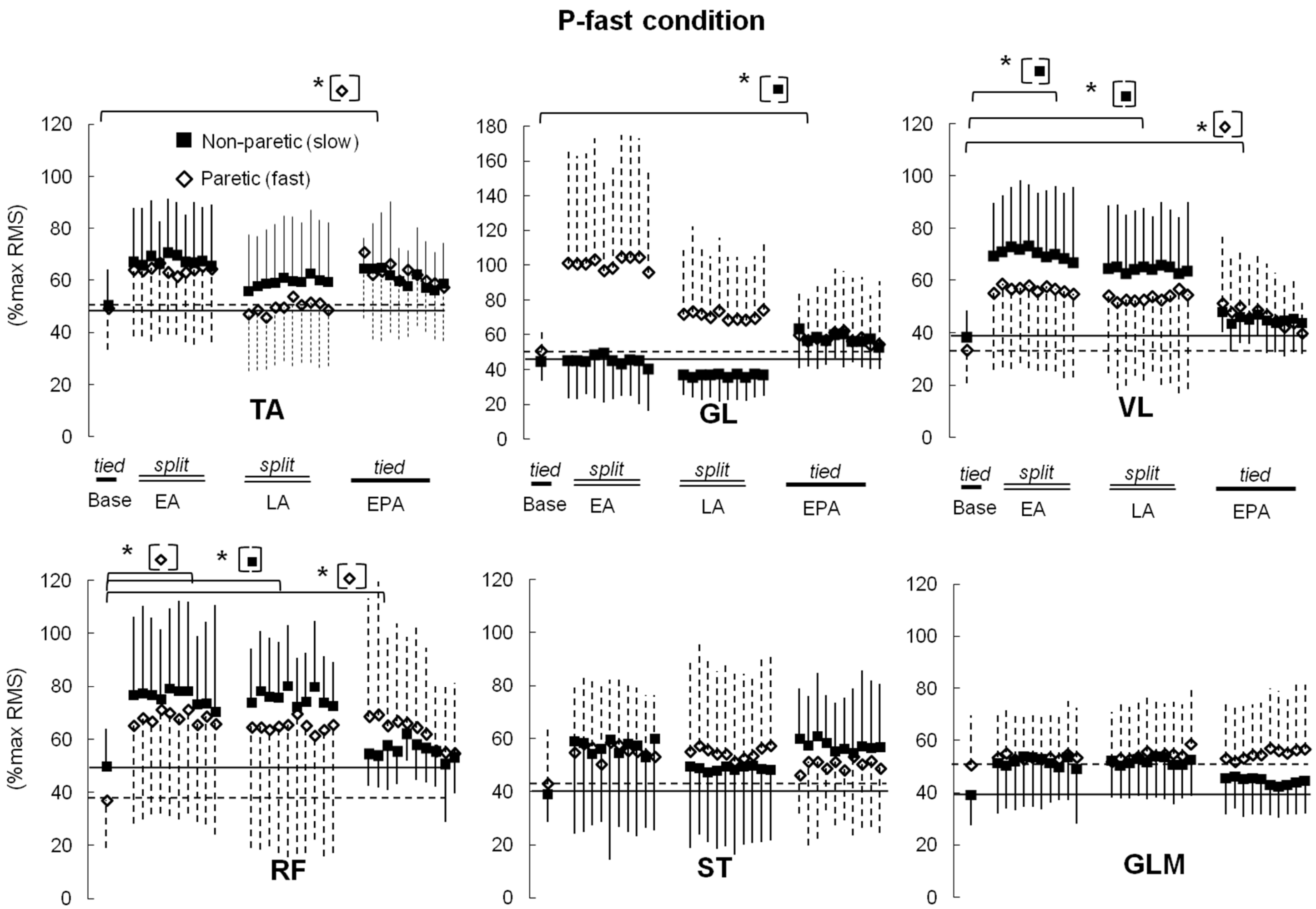

Figure 5-3. Normalized EMG of (NP) (■) and paretic (P) (◊) muscles during P-fast condition. Figure legend corresponds to Figure 5-2.

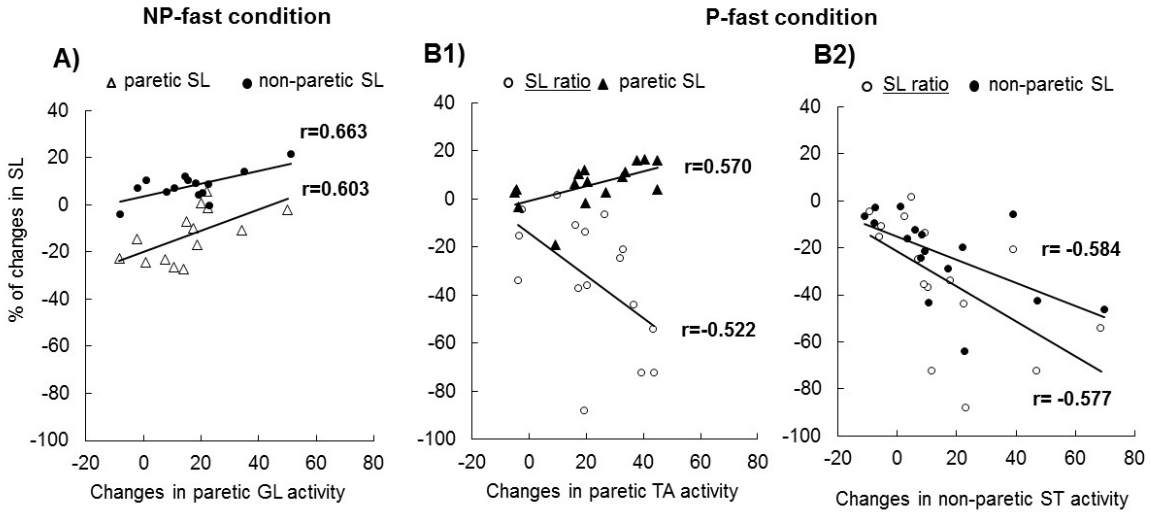

Figure 5-4. Associations between changes (%max) in paretic GL activity and % of change in paretic (Δ) and non-paretic (●) SL during NP-fast condition. For the P-fast condition, associations between the changes in paretic TA (B1) and ST activity (B2) with % of changes in SL (paretic▲; non-paretic●) and SL ratio (○).

5.1.12 References

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5.1.13 Supplementary Results Paper #1:

The supplementary results are organized in three sections. The first section presents the profiles of EMG signals obtained from the 16 indviduals post-stroke analyzed in paper #1 (section 5.1.13.1). The subsequent section (5.1.13.2) presents step length and amplitude of muscle activity results from 10 healthy controls during SBT walking. Finally, the changes in EMG timing are presented for both individuals post-stroke and healthy controls in a third section (section 5.1.13.3).

5.1.13.1 EMG Profiles Stroke Group

The following figures (5-5 and 5-6) were included in the submission to the journal Electromyography and Kinesiology in April 2016 as supplementary figures. The purpose of these figures was to provide the reader with an idea of the timing and shape of the muscle activity during a gait cycle for the muscles whose activity changed significantly in line with step length after-effects.

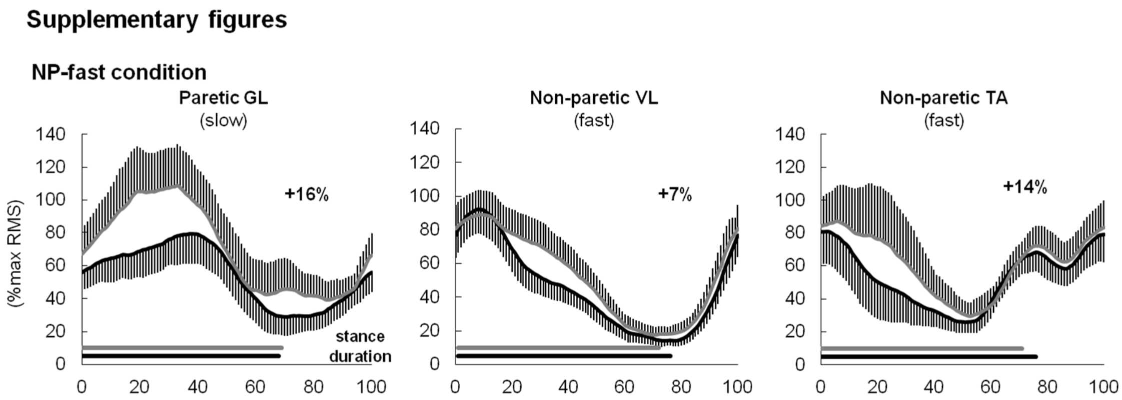

Figure 5-5. Illustration of lower limb muscles with significant changes from baseline to early post-adaptation during the NP-fast condition: paretic gastrocnemius lateralis (GL), non-paretic vastus lateralis (VL) and tibialis anterior (TA). Presented are group mean EMG activation profiles and corresponding standard deviations (black: baseline; grey: post-adaptation) over normalized gait cycle. The horizontal lines represent the duration of stance during baseline (black) and post-adaptation (grey). Percentage of change of the mean RMS is illustrated next to the activation profiles.

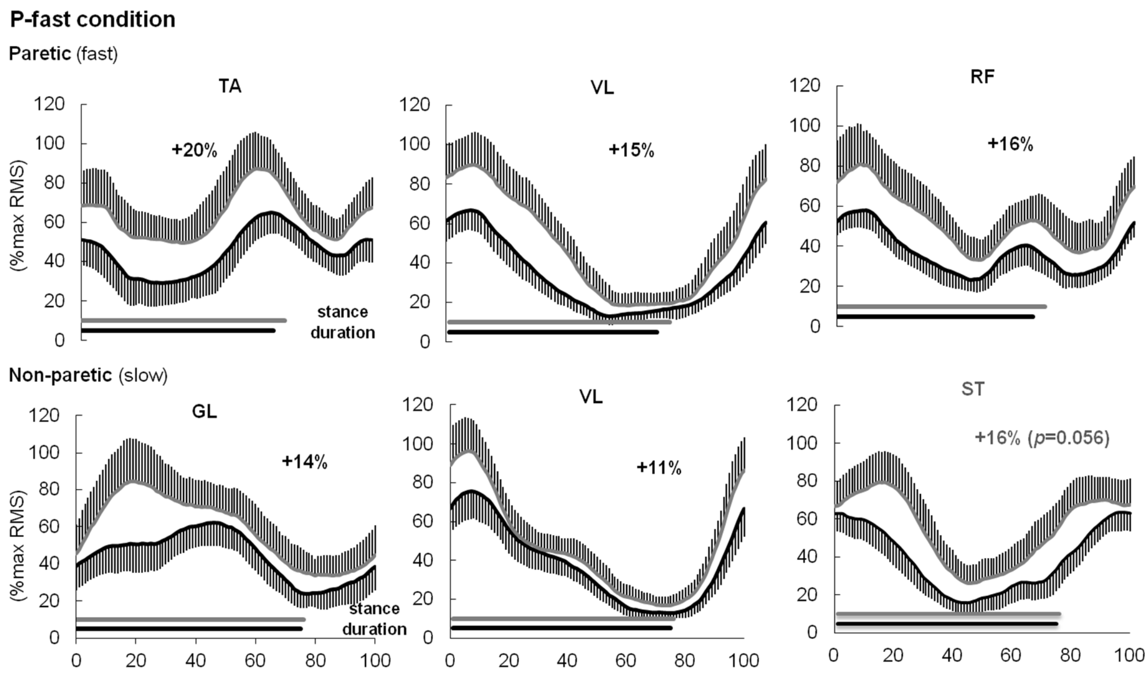

Figure 5-6. Illustration of group average paretic (top panel) and non-paretic (bottom panel) EMG activation profiles with significant changes from baseline to early post-adaptation during the P-fast condition. The muscles are: the paretic tibialis anterior (TA), vastus lateralis (VL), rectus femoris (RF) and non-paretic gastrocnemius lateralis (GL) and VL. The non-paretic semitendinosus (ST, grey) approached significance. Legend corresponds to the legend of Figure 5-5.

5.1.13.2 SBT-induced Changes in Step Length and Muscle Activity in Healthy Controls

This section presents the effects of SBT walking on step length and EMG amplitude of six lower limb muscles in healthy individuals. As presented in the literature review section (section 2.6.1), the effects of SBT walking on the gait pattern of healthy individuals have been studied extensively in cross-sectional studies, but only one study presented changes in EMG amplitude during and after SBT walking in healthy individuals (see section 2.6.2.1). Furthermore, the present protocol differs slightly from the protocols used in those previous studies (i.e., predefined walking speed in previous studies vs. comfortable speed in the present study). Consequently, having a control group walking with the SBT protocol used in our study was considered as essential.

Step length and amplitude muscle activity were evaluated from a healthy control group for comparison with the data measured in our stroke group. Detailed information about the recruitment of healthy controls, their evaluation, and the experimental sessions, as well as data analysis and treatment can be found in section 4.1 (methods of the cross-sectional study). Figure 4-1 illustrates the instrumental setup for EMG and 3D-motion analysis. The protocol for healthy individuals is illustrated on the bottom part of Figure 4-2 (B). All 10 participants successfully accomplished the protocol. Participants presented an average age of 48 years (SD 13.8) and included six females and four males. Average slow and fast walking speeds over ground were at 1.52 m/s (SD 0.20) and 2.44 m/s (SD 0.50), respectively. A comparison of demographic information and gait characteristics between control and stroke group can be found in section 5.1.13.3 (Table 5-2).

5.1.13.2.1 *SBT-Induced Effects on Step Length in Healthy Controls*

The repeated-measures ANOVA revealed a significant interaction among factors (Groups x Period x Side) for step length ($p = 0.000$). The separate group analysis also showed a significant interaction for the side factor (fast and slow step length) ($p = 0.000$). One-factor ANOVAs were performed on each side (fast and slow step length) and revealed a significant effect of Period ($p \leq 0.001$) for step length symmetry and slow and fast step length, respectively.

Adaptation

During comfortable walking at baseline, all participants presented symmetrical step lengths with an average ratio of 1.05 (SD 0.09) (Figure 5-7). At the beginning of the adaptation period slow (non-dominant) step length initially increased (p = 0.032) and the fast (dominant) side tended to decrease (p = 0.076). This led to a significant asymmetry in step length when compared with baseline with a ratio of 0.76 (SD 0.08) (p = 0.001). After five minutes of walking, participants displayed an increase in fast step length (average of +11.2 cm) back to baseline length (p = 0.139). This resulted in an adaptation of symmetry ratio towards the baseline with a ratio of 1.00 (p = 1.000).

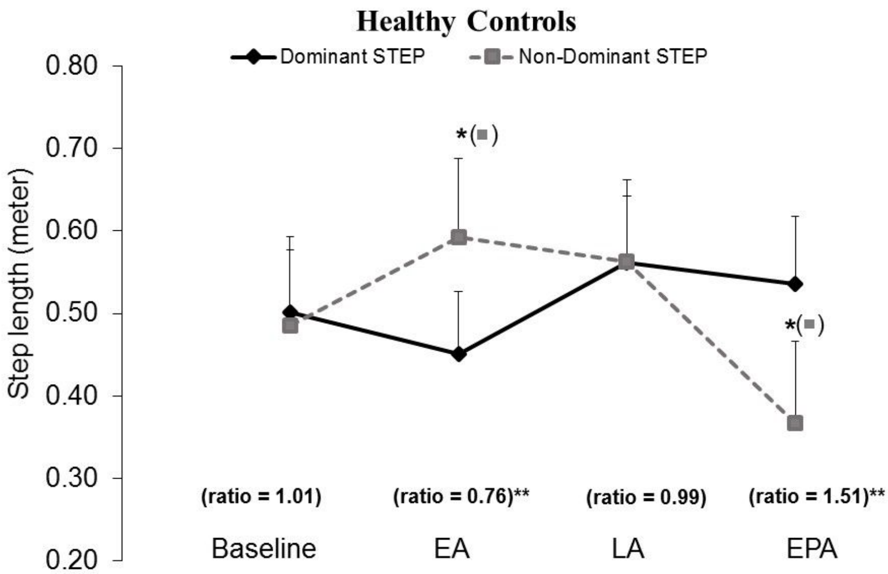

Figure 5-7. Step length of the dominant (black continuous) and non-dominant (grey dotted) side during baseline, early adaptation (EA), late adaptation (LA) and early post-adaptation (EPA). * represent significance for changes in step length and ** for the ratio when compared to baseline. Significance level was set at *alpha* = 0.05 after a post-hoc Bonferroni correction.

Post-adaptation

When the belts were set back to equal speeds after the period of adaptation (ratio 1:1, early post-adaptation), a step length asymmetry was observed with a ratio of 1.51 (SD 0.28) (p = 0.003) when compared to baseline. This ratio was inverted (i.e., in the opposite direction) when compared to the ratio of 0.76 at early adaptation. Slow step length was significantly shorter when compared to baseline (p = 0.007) with an increase in fast step length which did not reach the level of significance (p = 0.175).

5.1.13.2.2 *SBT-Induced Effects on EMG Amplitude in Healthy Controls*

Significant interactions in the repeated-measures ANOVA between the factors Group (i.e., participants with stroke and healthy controls) and Period were found in all muscles (TA, GL, VL, RF, ST, GLM) except the TA muscle during P-fast condition and the ST muscle during NP-fast condition. Sphericity testing revealed that the slow TA, VL, RF, ST and fast ST and GLM required a Huynh-Feldt approach to correct for violation of sphericity.

Adaptation

During the period of adaptation, the ANOVA revealed a significant interaction between Side and Period for the GL and RF muscles ($p \leq 0.002$). A significant effect of Period was found in the TA, GL fast, RF and GLM muscles on the slow side ($p \leq 0.026$). At baseline, group average EMG activity in all muscles was similar on both sides at about 50%max (Figure 5-8). Changing belt speed configuration from equal (ratio 1:1) to unequal (2:1) (early adaptation; EA) led to a bilateral increase in TA, VL and RF muscle ($p \leq 0.044$) with more pronounced changes on the slow side compared to the fast side. For example, the RF presented an increase of 154%max (SD 82%max) on the slow side versus 61%max (SD 35%max) on the fast side. Additionally, an increase of the GLM was observed (p = 0.041) on the slow side with an average increase of 20%max (SD 23%max). On the fast side, an increase in activity was seen for the GL with an average of 68%max (SD 43%max) (p = 0.032) when compared to baseline.

At the end of the adaptation period, the muscles with initial increases displayed a decreased in activity back to baseline ($p \geq 0.130$), except for the RF. Slow and fast RF presented activity higher than at baseline at 67%max (SD 47%max) and 37%max (SD 25%max) ($p \leq 0.036$), respectively.

Post-adaptation

During early post-adaptation, increased activity of the TA muscle (bilaterally) and the RF muscle was observed on the fast side. With regards to the TA muscle, the increase in activity was more prononounced on the fast side with 39%max (SD 39%max) additional activity compared to the 6.5%max (SD 21%max) on the slow side relative to baseline (p = 0.038). The fast RF approached significance (p = 0.065) with an increase of 60%max (SD 58%max) compared to baseline. In opposition to the early adaptation period, the changes in EMG activity of TA and RF muscle were more pronounced on the fast side when compared to the slow side.

Correlations Between Step length and EMG Amplitude During Early Post-Adaptation

The changes in step length symmetry from baseline to early post-adaptation were strongly associated with a concurrent reduction in slow step length (r = -0.895; p = 0.001). Changes in slow step length were also moderately to strongly associated with the changes found in fast TA (r = -0.739; p = 0.036) and slow ST activity (r = -0.896, p = 0.003). The changes in fast (dominant) step length from baseline to early post-adaptation was negatively associated with the changes in slow RF activity (r = -0.708; p = 0.033).

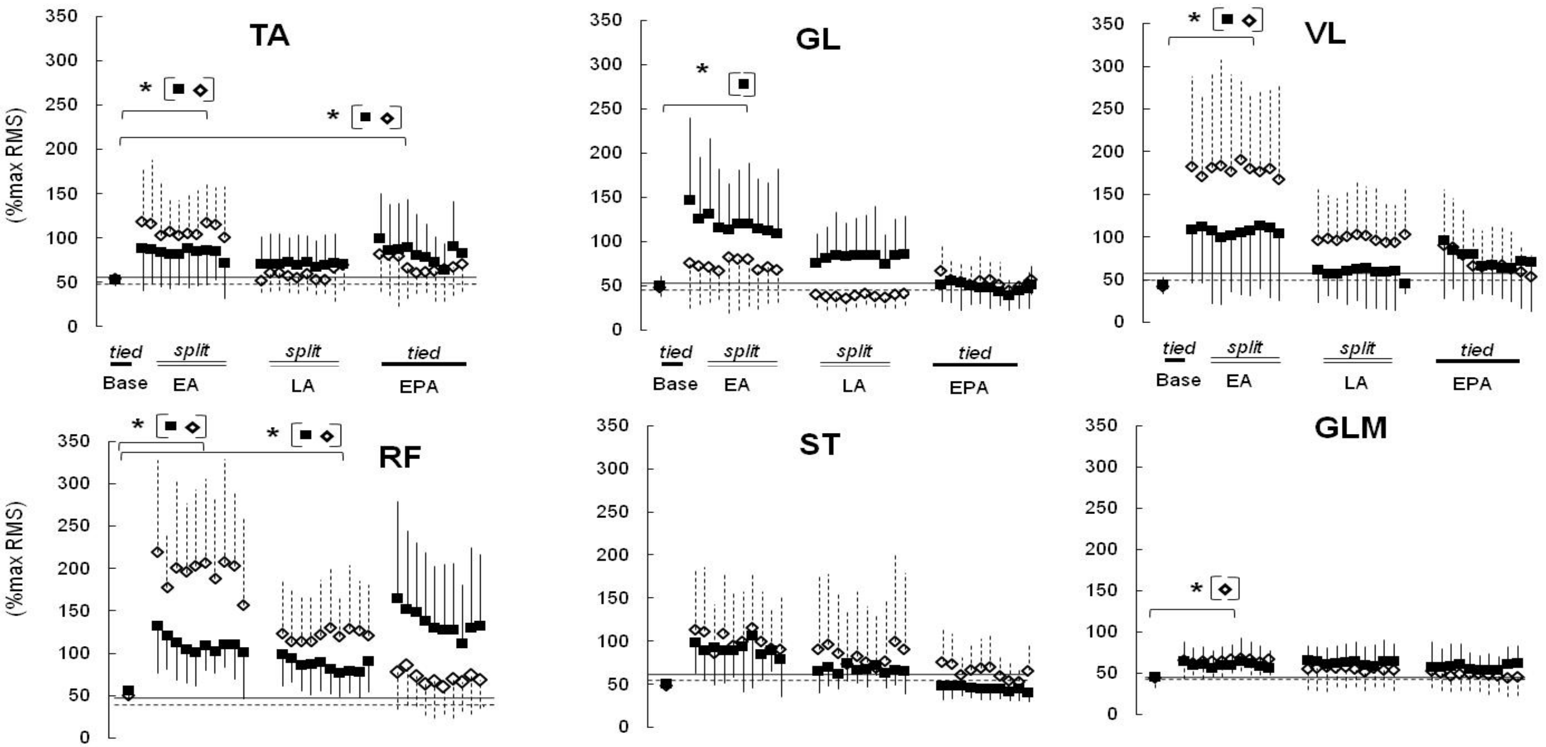

Figure 5-8. Scatter graphs with normalized EMG of six dominant (D) (■) and non-dominant (ND) (◊) muscles during the four periods of walking. For early adaptation (EA), late adaptation (LA) and early post-adaptation (EPA), mean values of EMG are presented for 10 consecutive gait cycles. At baseline (Base), 10 gait cycles are averaged. *Indicates significant changes when compared to baseline for non-dominant [◊] or dominant [■]. Abbreviations: tibialis anterior (TA), gastrocnemius lateralis (GL), rectus femoris (RF), semitendinosus (ST), vastus lateralis (VL), and the gluteus medius (GLM).

5.1.13.3 Effects on Coactivation and Coordination in Individuals Post-stroke and Healthy Controls

Coactivation and interlimb coordination were studied in both healthy and post-stroke individuals to explore the changes in temporal aspects of muscle activity involved in step length after-effects. The method used for the analysis of coactivation and interlimb coordination of muscle activity is described in Appendix I. Coactivation duration and interlimb coordination was compared between baseline and early post-adaptation. Table 5-2 at the end of the section illustrates demographics and functional parameters of the healthy control group combined with the corresponding values of the stroke group.

5.1.13.3.1 *Coactivation*

Muscle coactivation of ankle and knee antagonists was analyzed in 16 individuals post-stroke and nine healthy individuals. The data of one individual in the control group was not valid because of large amplitude artefacts and signal loss observed during offline treatment and had to be omitted from analysis. Shapiro-Wilk testing revealed that the coactivation duration (% of normalized gait cycle) was normally distributed for the antagonists of the knee but not of the ankle. In both groups, coactivation duration of ankle and knee antagonist muscles occurred mainly during the stance phase of gait.

Coactivation of Ankle Antagonists

The comparison of baseline TA-GL coactivation between groups revealed that the non-paretic TA-GL coactivation duration during NP-fast condition was 10 - 12% longer when compared with the dominant side of control subjects (p = 0.000) (Figure 5-9). In contrast, during the baseline of the P-fast condition, no significant differences between groups of non-paretic TA-GL coactivation duration were found (p = 0.095). During both conditions, TA-GL coactivation duration at the paretic ankle did not differ with the coactivation duration of these muscles found in healthy controls (p > 0.186). In general, TA-GL coactivation duration increased after fast belt walking and was unchanged after slow belt walking in individuals post-stroke and healthy controls. In individuals post-stroke, after fast-belt walking coactivation duration of the TA and GL muscle increased on the paretic leg from a mean duration of 7% to

18% (p = 0.028) (Figure 5-9). Healthy individuals increased TA-GL coactivation from 2% to 20% of the gait cycle after fast-belt walking (dominant side; D) (p = 0.011). In both groups, slow-belt walking in healthy controls and individuals post-stroke did not induce significant changes in TA-GL coactivation duration.

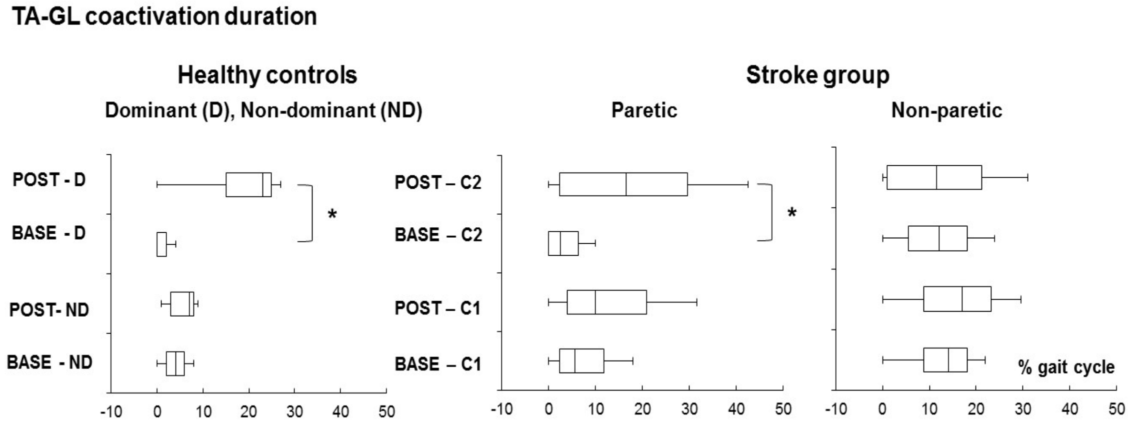

Figure 5-9. Boxplots illustrating coactivation duration between the tibialis anterior (TA) and the gastrocnemius lateralis (GL) for healthy controls (left-hand graph) and individuals post-stroke (middle and right-hand graphs). The duration is represented in % of the gait cycle during baseline (BASE) and post-adaptation (POST). For the stroke group, co-activation is presented for both the NP-fast condition (C1) and P-fast condition (C2). *Denotes significant differences between baseline and post-adaptation period. Significance level was set at *alpha* = 0.05.

Coactivation of Knee Antagonists

Comparison of baseline VL-ST coactivation duration revealed a similar amount of coactivation between groups ($p \geq 0.154$). The group comparison with factors of Period and Side revealed significant interactions between Group x Period (p = 0.049) and Group x Side (p = 0.020). A significant effect of the factor Period was found only in healthy individuals (p = 0.037) with a concomitant reduction in VL-ST coactivation duration after walking on the fast belt (29% to 16%) (Figure 5-10). In individuals post-stroke, SBT walking led to minor and non-significant after-effects in VL-ST coactivation during both the NP-fast and P-fast condition ($p \leq 0.385$).

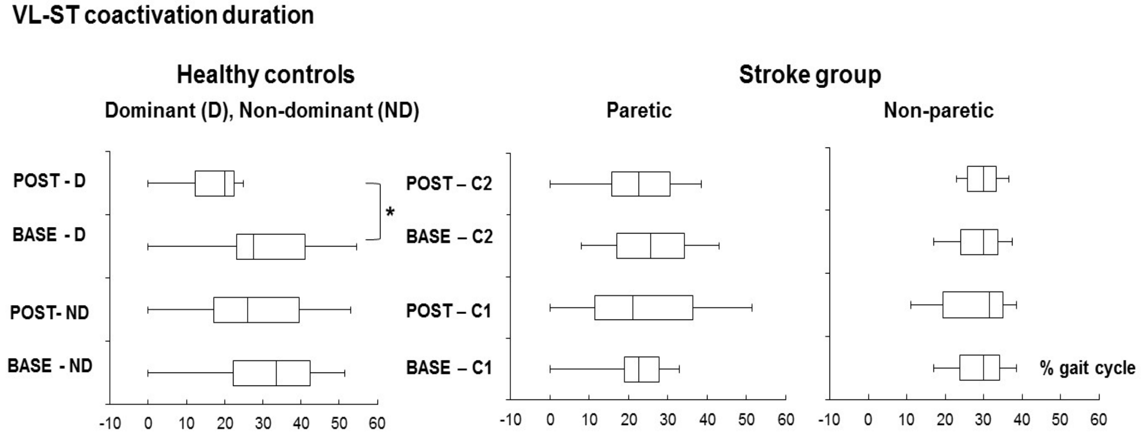

Figure 5-10. Boxplots illustrating coactivation duration between the vastus lateralis (VL) and the semitendinosus (ST) for healthy controls (left-hand graph) and individuals post-stroke (middle and right-hand graphs). The duration is represented in % of the gait cycle during baseline (BASE) and post-adaptation (POST). For the stroke group, co-activation is presented for both the NP-fast condition (C1) and P-fast condition (C2). *Denotes significant differences between baseline and post-adaptation period. Significance level was set at *alpha* = 0.05.

5.1.13.3.2 *Inter-leg Coordination of Muscle Activity*

The cross-correlation coefficients were not normally distributed in the majority of muscles and walking periods in both groups (stroke group n=16; healthy controls n=9). The following results were obtained from non-parametrical statistics. The Rxy can be interpreted as other correlation coefficients with a 1.00 as perfect correlation (Appendix I).

At baseline, healthy controls presented coefficients between Rxy 0.72 and Rxy 0.88 for all muscles except the RF with 0.61 (SD 0.37) (Figure 5-11). After the six minutes of SBT walking, this group showed after-effects that reflected worsened interlimb coordination, passing from strong coefficients in the TA with Rxy of 0.85 (SD 0.27) and the ST with a Rxy of 0.72 (SD 0.31) at baseline, to weak coefficients with a Rxy below 0.46 ($p \leq 0.028$) during early post-adaptation. The VL showed a tendency to reduce the correlation, but only approached significance (p = 0.066). Individuals post-stroke presented similar baseline coefficients to healthy controls in both the NP-fast and P-fast conditions (Rxy 0.62-0.92). Statistical group differences (healthy vs. stroke) in interlimb coordination at baseline were found for the TA and GL muscles ($p \leq 0.041$). Individuals post-stroke did not present significant changes in Rxy coefficients from baseline to early post-adaptation during both NP-fast and P-fast conditions.

TABLE 5-2. Demographics and Functional Parameters of Control (n = 10) and Stroke (n = 16) Participants.

Demographics	**Group**	**Number**		
Gender (m;f)	Control	4;6		
	Stroke	10;6		
	Group	**Mean (SD)**	**Range**	**(*p*-value)[A]**
Age	Control	48 (13.8)	23 - 69	
	Stroke	49.8 (13.4)	24 - 69	0.743
Gait and balance parameters				
Symmetry (D/ND)	Control	1.01 (0.07)	0.88 - 1.09	
	Stroke	0.94 (0.15)	0.64 - 1.15	0.199
CWS 10MWT (m/s)	Control	1.52 (0.20)	1.30-1.75	
	Stroke	0.92 (0.30)	0.51-1.40	0.000
FWS 10MWT (m/s)	Control	2.44 (0.50)	1.9 - 3.28	
	Stroke	1.30 (0.38)	0.72 - 2.00	0.000
CWS TM (m/s)	Control	1.05 (0.17)	0.7 - 1.2	
	Stroke	0.62 (0.16)	0.40 - 0.90	0.000
Balance	Control	56* (0.0)	NA	0.024
	Stroke	55* (3.38)	44-56	

Abbreviations: SD = standard deviation, m = male, f= female, D = dominant, ND = non-dominant, 10MWT = 10 Meter Walking Test, m/s = meters/second, TM = treadmill, CWS = comfortable walking speeds, FWS = fast walking speed, NA = not applicable; *the value represents the group median; [A] Independent group *t*-test.

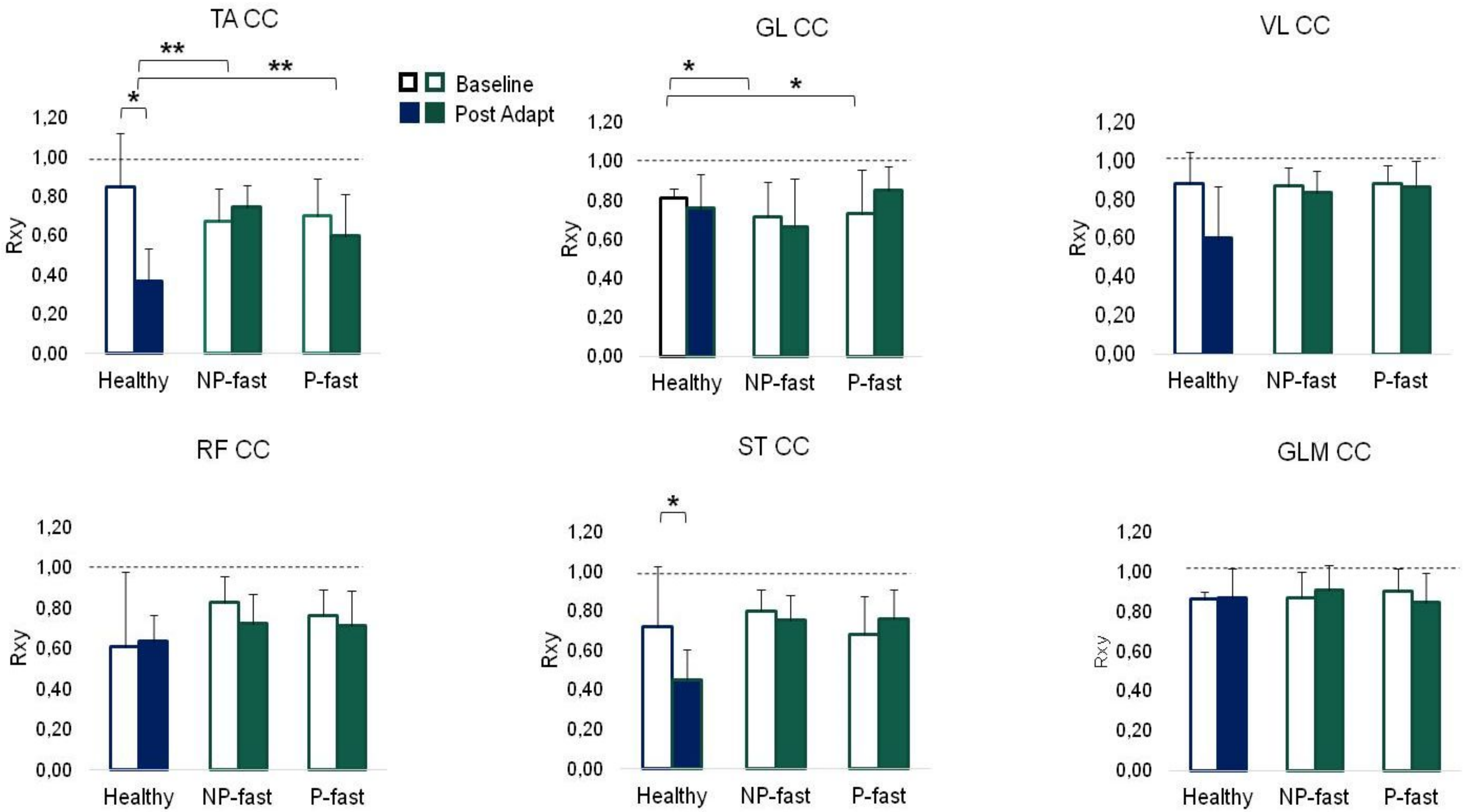

Figure 5-11. Diagrams illustrating coefficient (Rxy) of cross-correlation (CC) for the tibialis anterior (TA), gastrocnemius lateralis (GL), vastus lateralis (VL), rectus femoris (RF), semitendinosus (ST), and gluteus medius (GLM) for individuals post-stroke during both conditions (NP-fast and P-fast) (green) as well as for healthy controls (blue). *indicates significance between groups or period; **indicates significant differences between groups for both, baseline and post-adaptation.

5.2 Paper #2: Repeated Split-belt Treadmill Walking Improved Gait Ability in Individuals with Chronic Stroke: A Pilot Study

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As principle author, I confirm having made substantial contributions to the majority of the procedures required to obtain the results presented in this paper. These procedures included the development of the protocol, recruitment of participants and physiotherapists, instruction of physiotherapists for training purpose, clinical evaluation of participants, data collection in the laboratory, and treatment and interpretation of clinical and biomechanical data. Bradford J. McFadyen participated in the interpretation of results and the revision of the manuscript. Sylvie Nadeau supervised and contributed to each of these procedures starting from the protocol development to the final revision of the manuscript.

5.2.1 Preamble

Paper #2 is one of the two papers presenting the results of the effects of repeated SBT walking on gait ability and biomechanics in individuals post-stroke. As described in the literature review, the positive effects of SBT walking on step length asymmetry in a group of 12 individuals post-stroke were published recently by the Reisman group (Reisman et al, 2013). The analysis of long-term effects to repeated SBT exposure is indeed pertinent. SBT walking – repeated or one-time – led to short-term reductions of step length asymmetry during walking over ground (Reisman et al, 2009, 2013). Furthermore, a tendency of long-term improvements was found in the training study by Reisman et al. (2013). With our training study, we were interested to test a new error-augmentation-based SBT protocol and its short- and long-term effects on gait ability in a group of stroke survivors. The purpose was not only to reproduce and test the findings on step length symmetry by the Reisman group (Reisman et al, 2013) but also to evaluate concomitant effects on other gait parameters that are considered relevant to stroke survivors in order for them to achieve independence in community ambulation. The relevance of this latter aspect will be explained in the introduction section of paper #2. Moreover, we wanted to conduct clinically relevant research and take into account aspects of knowledge exchange between research and practice. Therefore, paper #2 set out to not just the effects of a SBT protocol aimed primarily at improving step length asymmetry and reproducing previous findings, but also to test the efficiency of this protocol in improving other relevant gait parameters as well as its practicality in a clinical environment.

5.2.2 Abstract

This study investigated the effects of repeated split-belt treadmill (SBT) walking on gait ability in individuals post-stroke. Twelve individuals with a first unilateral cerebral stroke (10 males; mean age 53 (SD 8.74); mean time post-stroke 25 months (SD 23.5); nine with left-sided strokes) and initial step length asymmetry (ratio = 1.10-2.05) volunteered for the study. They were trained by physiotherapists from an outpatient rehabilitation center six times over 2-3 weeks using a SBT protocol. After only six sessions of training, all participants reduced their step length asymmetry from an average ratio of 1.39 to 1.17 ($p = 0.002$) and increased walking speed ($p = 0.043$). Improvements in symmetry and speed were retained over one month ($p \leq 0.008$). No effect was observed in participants' endurance, assessed with the 6-Minute Walk Test. These findings suggest that the present SBT protocol has the potential to be an efficient intervention to improve not only SL symmetry but also gait speed, in individuals post-stroke.

Keywords: Stroke, Symmetry, Training, Gait ability

5.2.3 Introduction

Each year, close to 800 000 people in the USA and 50 000 people in Canada experience a cerebral stroke (HSF, 2016; Mozaffarian et al, 2015) leading to a substantial number of stroke survivors living with persistent gait deficits (Lord et al, 2004; Kollen et al, 2005; Patterson et al, 2008; Ada, Dean, and Lindley, 2013). Frequently reported characteristics in gait post-stroke include reduced gait speed (Mulroy et al, 2003; Lord et al, 2004; Ada, Dean, and Lindley, 2013), endurance (Eng et al, 2002; Dunn et al, 2015) and symmetry (Patterson et al, 2008; Patterson et al, 2010; Lauzière et al, 2014) compared to unimpaired individuals. Gait speed in individuals post-stroke has been reported to range from 0.1 - 1.08 m/s (Perry, Garrett, Gronley, and Mulroy, 1995; Taylor, Stretton, Mudge, and Garrett, 2006; Ada, Dean, and Lindley, 2013), while for endurance, distance covered during a six-minute walk ranges from 171 - 318 meters (m) (Dunn et al, 2015). These values are below the suggested speed (0.9 - 1.2 m/s) and distance (600 m) required for full community ambulation including traversing crosswalks or navigating shopping malls (Andrews et al, 2010; Salbach et al, 2013). With regards to gait asymmetry, 33.3% of a group of 54 individuals post-stroke presented inter-limb differences in step length (SL) (Patterson et al, 2008). Gait asymmetry has been associated with a loss of bone density (Jorgensen, Crabtree, Reeve, and Jacobsen, 2000), increased neuromuscular pain (Norvell et al, 2005), more challenge in balance control during walking (De Bujanda, Nadeau, Bourbonnais, and Dickstein, 2003) and higher energy consumption (Detrembleur et al, 2003; Ellis, Howard, and Kram, 2013; Awad et al, 2015). Energy consumption is strongly correlated with the ability to walk in the community (Franceschini et al, 2013). Consequently, reduced gait ability can compromise an individual's independence and social participation (Lord et al, 2004; Franceschini et al, 2013; Salbach et al, 2014).

Studies have shown that treadmill training (with or without body weight support) is very effective in improving gait speed and endurance in individuals with chronic stroke (Macko et al, 1997; Pohl, Mehrholz, Ritschel, and Ruckriem, 2002; Polese et al, 2013; Mehrholz, Pohl, and Elsner, 2014). Fast treadmill walking (Lamontagne and Fung, 2004), body weight supported treadmill walking (Combs, Dugan, Ozimek, and Curtis, 2013) and auditory feedback training (Thaut et al, 2007) led to improvements in temporal, but not spatial asymmetry. The recovery of this parameter tends to be more resistant to conventional therapy compared to other gait parameters (Patterson et al, 2015). At discharge from inpatient rehabilitation, only 14% of patients improved gait symmetry (SL) compared to 30% and 62% of patients improving gait speed and balance respectively.

A new approach involving walking on a split-belt treadmill (SBT) based on the principle of error-driven motor learning (Malone, Vasudevan, and Bastian, 2011) has shown successful reduction in SL asymmetry in individuals post-stroke (Reisman, Wityk, Silver, and Bastian, 2007; Reisman et al, 2013; Lauzière et al, 2014). By repeatedly disrupting a well-learned movement with an external force, the movement pattern is modified. Based on feedback and -forward mechanisms controlled by the neuromuscular system, this deviation from the initial movement pattern is interpreted as an “error” and the pattern is gradually adjusted to the new, predictable demands (adaptation). This error-driven adaptation using a SBT has received more attention over the last decade (Reisman, Block, and Bastian, 2005; Bastian, 2008; Malone, Bastian, and Torres-Oviedo, 2012). Using a SBT, the movement error was induced by walking on two belts with unequal speeds (split-belt configuration). This led to an asymmetry (“error’) in different gait parameters (error-augmentation). Among different study cohorts, individuals without cerebellar lesions showed the capacity to gradually adapt to this initial error towards baseline symmetry (Reisman, Block, and Bastian, 2005; Morton and Bastian, 2006; Reisman et al, 2007). The initial “error” (asymmetry) was then adjusted step-wise with adaptation found in inter-limb parameters (e.g., step length) but not in intra-limb parameters (e.g., stride length). When the belts were reset to equal speeds (tied-belt configuration), participants continued to walk with the new pattern for some steps (after-effects) and corrected progressively towards their initial symmetry. The after-effects in SL were characterized by an increase of SL on the side that walked on the faster belt and a decrease on the side of the slower belt. In healthy individuals, this led to an asymmetry in SL (Reisman, Block, and Bastian, 2005). In individuals post-stroke with initial SL asymmetry, the after-effects were characterized by a more symmetrical SL when the leg with the shorter SL walked on the fast belt during split-belt configuration (Reisman et al, 2007; Malone and Bastian, 2013; Reisman et al, 2013; Lauzière et al, 2014). In terms of the size of after-effects, in healthy individuals, the largest magnitude was found when the slower belt speed was set to the individual's comfortable speed (Vasudevan and Bastian, 2010).

The repeated exposure to such error-augmentation-based SBT walking led to a reduced SL asymmetry during walking over ground with the tendency of a 1-3 months retention (Reisman et al, 2013). In this latter study, participants walked during 30 minutes on a split-belt configuration for twelve training sessions. Belt speed configurations were based on participants' maximal speed during walking over ground. In addition, the SBT training was combined with 10-15 minutes of training over ground with a physiotherapist to reinforce the changes in SL asymmetry (Reisman et al, 2013). Since two interventions of error-augmentation-based SBT and symmetry reinforcement over ground were combined, it remains unclear what the actual contribution of SBT training is to improving SL symmetry. Also, this combination of interventions did not improve gait speed (10MWT). The authors argued that changing gait speed or any other gait parameter was not the scope of their study and therefore belt speeds were not increased between sessions. However, developing interventions leading to an improvement in gait symmetry combined with walking speed were considered important in post-stroke rehabilitation (Awad et al, 2015). Walking faster and more symmetrically led to the largest reduction in energy consumption compared to improving either speed or symmetry alone (Awad et al, 2015).

Thus, the present study investigated the effects of the SBT protocol primarily on SL asymmetry during walking over ground and secondly the co-effects on gait speed, endurance, and functional mobility. Specifically, the present protocol involves a previously used 2:1 ratio of belt speeds (Reisman et al, 2013) adjusted to participants' comfortable speed that has shown solid short-term after-effects in SL and joint moments of individuals post-stroke (Lauzière et al, 2014), concentrating only on the contribution of the SBT protocol to the changes in SL symmetry and optimizing training breaks to increase the probability of improvements in walking endurance. The aim of these modifications was to provide a protocol which leads to clinically relevant and durable reduction in SL asymmetry after a short period of training as well as to investigate its effect on additional gait parameters (speed, endurance, and functional mobility).

5.2.4 Methods

5.2.4.1 Participants

Participants met the following criteria: first unilateral non-cerebellar stroke after sub-acute phase (> 3 months prior to study beginning (Richards, Malouin, and Nadeau, 2015); presence of SL asymmetry (ratio ≥ 1.08) (Patterson et al, 2010) and capacity to walk independently (without aid, including orthoses) over 15 m (Functional Ambulation Category > 4) (Mehrholz et al, 2007) at a speed greater than 0.5 m/s. However, the use of aids for longer distances or outdoors was not a criterion for exclusion. Participants were excluded if they presented major pain preventing them from walking and cognitive deficits which could limit comprehension of training instructions. A minimum walking distance of 200 m over six minutes without a break was also required. Participants were informed to not partake in other gait-related training or physical exercise during the course of the training and assessment period. All participants signed an informed consent prior to the study, which was approved by the local ethics committee.

5.2.4.2 Pre-Evaluation

At pre-evaluation, participants were screened for eligibility for the study by testing functional and cognitive deficits. Further exclusion criteria were the presence of hemineglect (Bells test; Gauthier, Dehaut, and Joanette, 1989) or hemianopsia, signs of major depression (score ≥ 10/15 on the Geriatric Depression Scale (Sheikh and Yesavage, 1968)), and severe cognitive deficits (score < 25/30 on the Folstein Mini Mental Exam, (Folstein, Folstein, and McHugh, 1975)). To account for functional deficits, lower limb impairment, and gait deficits were assessed. The Chedoke-McMaster Assessment was used to evaluate physical impairment of the leg and foot (Gowland et al, 1993). The Composite Spasticity Index was used to assess spasticity of the paretic plantarflexor muscles (Levin and Hui-Chan, 1993).

To characterize gait deficits, SL asymmetry, speed, endurance and functional mobility were assessed. SL was obtained by asking the participant to walk over a long paper-carpet with light-colored pencils affixed to the heel of the shoe (Cerny, 1983). The average of five to six steps was taken for each side to calculate SL and asymmetry. SL symmetry was calculated with two equations defined as follows:

$$\text{NSLD} = \frac{\text{abs (paretic SL - non-paretic SL)}}{\text{paretic SL + non-paretic SL}}$$

Equation 5-1. Normalized SL differences

$$(\text{L/S ratio}) = \frac{\text{Longer SL}}{\text{Shorter SL}}$$

Equation 5-2. Symmetry ratio L/S

The NSLD allowed comparing our results with those of Reisman et al. (2013) (Reisman et al, 2013). The L/S ratio was used in previous studies to determine standards for symmetry thresholds and clinically relevant changes (Patterson et al, 2010; Lewek and Randall, 2011; Reisman et al, 2013). Values of "0" and "1" represent perfect symmetry for NSLD and L/S ratio, respectively.

Comfortable and fast gait speeds were assessed with the 10MWT with an average of two trials (Flansbjer et al, 2005). Gait endurance was tested with the 6-Minute Walk Test (6MWT) based on the standards of the American Thoracic Society (ATS, 2002) and the American Heart Association (Gordon et al, 2004). Functional mobility was assessed with the Timed Up and Go (TUG) test at fast (two trials) and comfortable speeds (two trials) (Podsiadlo and Richardson, 1991).

5.2.4.3 Evaluation of Outcome Parameters

SL asymmetry, gait speed, endurance and functional mobility were also the four outcome parameters. These parameters were assessed on three occasions: prior to the training (pre-evaluation), 1-2 days post training (post-evaluation) and at 4 weeks after discharge from training (follow-up evaluation).

5.2.4.4 Training Protocol

Six sessions of SBT walking were conducted over a period of 2-3 weeks based on the error-augmentation strategy. Participants walked during 20 minutes on a split-belt configuration with a 2:1 ratio of belt speeds with the slower belt set at the participant's comfortable speed. To improve symmetry, the leg with the shorter SL walked on the faster belt (Malone and Bastian, 2013; Reisman et al, 2013; Lauzière et al, 2014). Therefore, participants with a shorter paretic SL had to walk with the paretic leg on the fast belt (paretic-fast group) and vice versa for participants with shorter non-paretic SL (non-paretic-fast group). Comfortable speed was determined on the treadmill at equal belt speeds prior to training at each session. To do so, participants initially walked at the speed of the previous session. This was followed by a progressive increase or decrease of both belt speeds at 0.05 m/s intervals until the participant was walking at a comfortable speed. After the training on the split-belt configuration, participants walked for 3 minutes on the tied-belt configuration (ratio 1:1). This period served to cool down and to avoid that participants walk on the over ground and leave the training situation immediately after being exposed to the repeated perturbation of the split-belt configuration. No verbal reinforcement was conducted during this period and the training did not include any intervention during walking over ground.

Heart rate was measured at rest (HRR) and constantly during the training with a polar heart rate monitor system. Blood pressure was assessed before, during breaks and after the training. Rate of perceived exertion was assessed with a modified Borg Scale (Borg, 1982) at the end of each of the four conditions and prior to a break. Breaks were taken when the heart rate exceeded 80% of age-related maximum or when the exertion was perceived to be higher than severe (> 5) (Gordon et al, 2004; Zamuner et al, 2011). The training continued after the heart rate decreased to the HRR. During the breaks, participants sat on a chair which was placed on the treadmill and no walking was allowed to avoid a de-adaptation of the walking pattern during the 20 minutes. For safety purposes, participants wore a harness without body weight support and were allowed to hold the handrails during the entire time on the SBT.

The training sessions were conducted by four trained physiotherapists in an outpatient facility. To obtain preliminary data about the feasibility of this intervention in a clinical setting, therapists filled out questionnaires after completion of the study. The questionnaires evaluated the therapists' perceptions of the practicality of the protocol and the readiness of its implementation in a clinical setting. A more detailed description of the pilot feasibility testing can be found in the supplementary files.

5.2.5 Statistical Analysis

Descriptive statistics are presented for demographic information, stroke characteristics, the number of breaks and perceived exertion during training (modified Borg Scale). Normal distribution was tested with the Shapiro-Wilk test. The effect of training over time (pre-, post-, follow-up evaluation) on dependent variables was tested with a repeated-measures ANOVA for SL, walking speed, duration of 6MWT and TUG at comfortable speeds, or a Wilcoxon paired signed-rank test for SL asymmetry and the duration of the TUG at fast speeds using SPSS 22.0 (SPSS Inc.). A posthoc pairwise comparison with a Bonferroni adjustment was conducted on parametric data to find the effects on dependent variables between the three evaluations. For non-parametrical statistics, alpha-level was adjusted by the number of comparisons (n = 3) requiring a $p \leq 0.017$. Dependent variables were: SL asymmetry, and comfortable and fast gait speeds, TUG durations and distance covered during the 6MWT. Effect sizes of changes found between evaluations were computed using Hedges's g_{av} (Lakens, 2013), which is a Cohen's *d* value corrected for the sample size using means, SDs and sample sizes and recommended for small samples. These *g* scores can be interpreted the same way as the Cohen's *d* scores: *g* around 0.2 is interpreted as "low", 0.5 "average" and 0.8 as "strong". Spearman's correlation analysis (coefficient = ρ) was conducted to determine associations between changes in SL asymmetry ratios (absolute value) with the other clinical parameters. Furthermore, correlation analysis was conducted to test if the changes in slow and fast SL measured from pre- to post-evaluation were associated with the changes found between pre- and follow-up evaluation.

5.2.6 Results

Twelve participants were included in the study (Figure 5-12). Participants' demographics and stroke characteristics are illustrated in Table 5-3. Asymmetry ratio ranged from 1.10 to 2.05 with four participants having shorter SL on the paretic side (P-fast group). Gait speed over ground ranged from 0.58 to 1.21 m/s (comfortable) and 0.74-1.65 m/s (fast). Different degrees of impairment in foot function ranging from 1-6 for the Chedoke McMaster Score were measured with 10/12 participants having no signs of spasticity (CSI $\leq$ 5). All 12 participants completed the three evaluation sessions and 11/12 participants the six training sessions (Figure 5-12). One participant completed five sessions because of unspecific pain at the dorsum of the paretic foot during sessions four and five. Two participants were able to walk on split-belt condition during the entire 20 minutes without breaks and the others required 1-3 breaks. Nine participants always achieved the belt speed ratio of 2:1 ratio. The lowest speed ratio was 1.6:1 with 0.7 m/s (slow) and 1.12 m/s (fast) (P2, Table 5-3). Comfortable speed on the treadmill increased from 0.05 to 0.30 m/s from the first to the last session with two individuals remaining at the same training speeds. Perceived exertion during split-belt condition ranged from 3-7 (moderate to very severe exertion) during the first session and from 2-5 (slight to severe) during the last session ($p > 0.05$). Total training duration ranged from 40 to 50 minutes including preparation of the participant.

5.2.6.1 Changes in Outcome Parameters

Data were normally distributed except for the SL asymmetry (L/S ratio and NSLD) and the TUG at fast speed. At post-evaluation, 12/12 participants showed improved symmetry (Figure 5-13). NSLD reduced from an average of 0.15 (SD 0.03) to 0.07 (SD 0.04) ($p = 0.003$) and L/S ratio from 1.39 (0.28) to 1.16 (0.16) ($p = 0.002$) post-training (Table 5-4). Eight participants achieved the reduction in asymmetry by a bilateral increase in SL with more pronounced changes on the fast walking side (Figure 5-14). In the non-paretic fast group, one participant decreased SL bilaterally (Figure 5-14 A-B). In the paretic-fast group, two decreased NP SL (slow belt) and showed little change (< 3 cm) for paretic SL (fast belt) (Figure 5-14 C-D).

Comfortable and maximal walking speeds increased after six sessions of training by an average of 0.10 m/s (SD 0.12) and 0.18 m/s (SD 0.18) ($p \leq 0.043$) (Table 5-4). Participants showed a tendency to reduce TUG duration at fast ($p = 0.034$) but not for comfortable speeds ($p = 0.175$) (Table 5-4). The changes in asymmetry and gait speed persisted until the 1-month follow-up ($p \leq 0.004$) and reached significance for the TUG duration at fast speed ($p = 0.008$). There was no training effect on the distance covered during the 6MWT during the post- and follow-up evaluations.

5.2.6.1.1 *Correlation Analysis*

Strong associations were found between the changes of NSLD and L/S ratio from pre- to post-evaluation ($\rho = 0.866$, $p = 0.000$), and from pre- to follow-up evaluation ($\rho = 0.937$, $p = 0.000$). Therefore, only the correlations with the L/S ratio will be presented.

Spearman's correlation revealed that changes in asymmetry from pre- to post-evaluation were associated with initial asymmetry ($\rho = -0.593$, $p = 0.042$) (Figure 5-15 A). Changes in symmetry from pre- to follow-up evaluation were associated with the three measures (Figure 5-15 B-C): Initial symmetry ($\rho = -0.837$, $p = 0.001$) and the initial duration of the TUG at comfortable ($\rho = -0.761$, $p = 0.004$) and fast speeds ($\rho = -0.607$, $p = 0.036$). Changes in SL found from pre- to post-evaluation were strongly associated with changes measured from pre- to follow-up evaluation for both, fast ($\rho = 0.818$, $p = 0.001$) and slow SL ($\rho = 0.811$, $\rho = 0.818$) (Figure 5-15 D).

5.2.7 Discussion

This study aimed to investigate if repeated exposure to new error-augmentation-based SBT training reduces SL asymmetry along with other relevant gait parameters in individuals post-stroke. Six sessions of SBT walking led to improvements in SL asymmetry, gait speed and functional mobility that were maintained over one month in individuals post-stroke.

The effects of error-augmentation-based SBT walking on SL asymmetry in individuals post-stroke (n=12) was previously analyzed by Reisman and colleagues (2013) showing successful outcomes (Reisman et al, 2013). Twelve sessions of combined training including SBT and symmetry reinforcement over ground led to a general reduction in SL asymmetry (*p*

< 0.05). Seven out of 12 participants were considered as "responders" with changes in NSLD ratio > 0.020. Improvements in symmetry approached significance at the one-month follow-up (p = 0.062). Changes in walking speed were not found. However, authors stated that other gait parameters were not targeted for improvement. In the present study, the protocol used by Reisman et al. (2013) was adjusted based on previous findings to achieve durable and clinically relevant improvements in SL asymmetry. Among these were the findings that after-effects were largest when the speed of the slow belt was close to comfortable speed compared to slower or faster speeds (Vasudevan and Bastian, 2010). Therefore, for each session, individuals were trained with the speed of the slow belt corresponding to their comfortable speed on the treadmill at that time. Furthermore, reduced walking speed and endurance were frequently reported in individuals post-stroke, and considered to limit their participation in the community and affect their quality of life (Lord et al, 2004; Salbach et al, 2013). Therefore, a second goal was to investigate the effects of training on walking speed, endurance, and functional mobility in addition to SL asymmetry.

5.2.7.1.1 *Improvements in SL Asymmetry*

All participants in the current study showed a reduction in SL asymmetry after five to six sessions of training. The training induced clinically relevant changes for their L/S ratio (> 0.15) (Lewek and Randall, 2011) in 8/12 participants one day post-training (Figure 5-13 B). Further, when using the second ratio (NSLD), 11/12 participants changed > 0.020 associated with a strong effect size > 1.00 (Table 5-4). Reisman et al. (2013) categorized individuals with changes > 0.020 as "responders" (n = 7/12). Interestingly, the use of only six sessions of a protocol based on participants' comfortable speed appears to be as efficient as 12 sessions with a protocol based on participant's initial fast walking speed over ground (Reisman et al, 2013). In comparison to Reisman et al. (2013), a significant and clinically relevant reduction in asymmetry was achieved without any reinforcement of symmetry after the error-augmentation protocol. This suggests that SL symmetry during walking over ground can be efficiently improved with the error-augmentation associated with the SBT protocol.

Furthermore, the positive and significant correlation found between changes in SL from pre- to post-evaluation, with changes from pre- to follow-up evaluation, (Figure 5-15) strengthens the conclusion that participants maintained their improvements for at least one month.

To reduce the asymmetry, the general strategy was to increase SL bilaterally, being more pronounced on the side trained on the fast belt as found in the study by Reisman et al. (2013). Further, improvements in SL symmetry were associated with initial asymmetry and functional mobility (Figure 5-15 C), suggesting that the training was more relevant for individuals with larger SL asymmetry and more impaired functional mobility. In contrast to Reisman et al. (2013), we did not find significant associations between the changes in symmetry and the level of perceived exertion (Reisman et al, 2013). Future investigations are necessary to better study this aspect.

5.2.7.1.2 *Improvements in Other Gait Parameters*

Improvements were also found in gait speed and functional mobility. Effect sizes of changes in the 10MWT and TUG at comfortable and fast speeds were small to moderate (Table 5-5). However, six participants showed substantial and clinically important increases (> 0.14 m/s; (Perera, Mody, Woodman, and Studenski, 2006)) in their comfortable speed for the 10MWT. Considering that the present protocol primarily aimed to improve SL symmetry, such clinically relevant changes in speed are interesting co-effects. In contrast, the group average reduction in duration to perform the TUG at fast speed post-training was below five seconds and therefore not clinically relevant (Persson et al, 2014).

Yet, how could SBT walking lead to changes in SL symmetry and speed in the present study group, but only in SL symmetry in the Reisman et al's group (2013)? Both studies presented data from 12 participants with similar ranges in NSLD (0.052-0.310 versus 0.046-0.344). Comfortable walking speeds were almost the same in the group analyzed by Reisman and colleagues (2013) with an average speed of 0.68 m/s (responders) and 0.75 m/s (non-responders) as the group average (0.78 m/s) of the present study. Thus, it is assumed that the differences in outcomes between the two studies can be explained mainly by the protocols used.

In the protocol tested by Reisman and colleagues (2013), a participant's fast walking speed over ground was used to define the speed of the faster belt, and belt speed configuration was maintained across sessions. Thus, it would appear that participants did not have to manage speeds faster than they were able to achieve during over ground walking. In the present study, the speed of the slower belt defined the speed configuration. During the first training session 6/12 participants were trained with fast-belt speeds higher than the fast speed they achieved during walking over ground. During the last four training sessions, this number increased to 8/12 participants. Thus, managing higher walking speeds than achievable during walking over ground could explain the improvements in walking speed over ground during post-evaluation in the present group. Studies which compared different training speeds in individuals post-stroke showed that the largest improvements in walking speed were found in the group trained at the fastest speeds (Pohl et al, 2002; Sullivan et al, 2002). This leads to the assumption that fast-belt speeds were more likely contributing to the gait-speed improvements. However, changes in gait speed were not required to improve SL symmetry as seen in the protocol used by the Reisman group (2013). Instead of increasing gait speed, participants decreased cadence from an average of 98 steps/ minute to 89 steps/ minute. Therefore, it is assumed that the essential factor leading to changes in SL symmetry is the unequal belt speeds (ratio 2:1) as found in cross-sectional studies (Bastian, 2008; Malone, Bastian, and Torres-Oviedo, 2012).

In terms of gait endurance, the distance covered during the 6MWT remained unchanged after the six sessions of training. To achieve improvements in endurance post-stroke, walking duration should be at least 20 minutes per day of continuous or accumulated exercise (Gordon et al, 2004). Participants received training 2-3 times a week and took, on average, two breaks during the 20 minutes of SBT walking at a 2:1 ratio. Thus, the duration of the present protocol might not have been sufficient to improve endurance in the present cohort.

5.2.7.1.3 *Future Studies*

The present study indicated that it is not necessary to practice error-augmentation-induced improvements in symmetry additionally during walking over ground to reduce SL symmetry over ground to a relevant degree. It still remains unclear what changes in muscle activity or effort (force production, joint moments) are related to the improvements of gait symmetry. In both the previous (Reisman et al, 2013) and present studies on repeated SBT, the majority of participants presented a shorter SL on the NP-side (n = 8/12). A larger group of individuals with shorter paretic SL would provide knowledge about the strategies used to achieve the reduction in symmetry. Furthermore, the clinically relevant changes after only six training sessions suggest that the use of comfortable speed on the slower belt appears to be more efficient than the use of predefined speeds presented in previous work (Reisman et al, 2013). Finally, the preliminary data presented in the supplementary section of this paper showed that the physiotherapists involved in the study (all experiencing the SBT for the first time) considered the current protocol to be practical for their outpatient setting.

5.2.8 Study Limitations

The current training included only a few training sessions (n = 6) and a double baseline for SL symmetry was not conducted. The latter would have given insight on the actual between-days variance of the participants' SL in comparison with training-induced changes. Nevertheless, 11/12 participants exceeded the criterion of 0.020 used by Reisman et al. (2013) with their changes in NSLD and 8/12 participants showed changes in L/S ratio > 0.15. Thus, the majority of participants showed clinically relevant changes in SL asymmetry.

5.2.9 Conclusion

The use of a comfortable speed-based SBT protocol led to a significant and clinically relevant reduction in SL asymmetry and appears to be appropriate locomotor training to reduce SL symmetry in individuals post-stroke.

In contrast to previous findings, concomitant effects on walking speed were found and improvements were significantly maintained at least 1-month post-training. A larger controlled trial is needed to test if the present protocol is a more advantageous intervention in stroke rehabilitation compared to other locomotor training when improvement in both SL symmetry and gait speed are targeted.

5.2.10 Declaration of Conflict of Interest

There are no conflicts of interest to disclose.

TABLE 5-3. Participants' Demographics and Stroke Characteristics.

Participant	**Age** (years)	**Time post-stroke** (months)	**Paretic side**	**Side of shorter step**	**Initial asymmetry** (ratio L/S)	NSLD	**Initial gait speed (m/s)*** (comf./fast)	**Chedoke Foot/Leg** (/7)	**CSI** (/16)
P1	55	48	L	NP	2.05	0.344	0.74/0.74	3/4	3
P2	53	15	L	NP	1.78	0.280	0.59/0.75	1/6	5
P3	73	9	R	NP	1.45	0.184	0.58/0.85	2/6	4
P4**	58	8	R	P	1.45	0.185	0.92/1.39	5/6	7
P5	49	6	L	NP	1.42	0.172	0.72/1.04	2/6	4
P6	39	40	L	NP	1.31	0.133	1.21/1.65	5/6	5
P7	60	21	L	NP	1.28	0.124	0.74/1.05	3/6	4
P8**,[f]	50	21	L	P	1.27	0.118	0.63/0.76	1/3	7
P9	52	12	R	NP	1.22	0.097	0.62/0.96	3/5	5
P10**	43	88	L	P	1.19	0.086	0.58/0.93	1/4	5
P11	49	19	L	NP	1.15	0.071	1.01/1.64	6/7	5
P12**, [f]	58	14	L	P	1.10	0.046	1.06/1.36	5/7	5
Mean	**53.3**	**25.1**			**1.39**	**0.151**	**0.78/1.09**	**3/6**	**5**
Number			**9L/3R**	**8NP/4P**					
SD	8.7	23.5			0.28	0.05	0.22/0.33	1.8/1.2	1.2

*Initial gait speed was obtained from the 10MWT; m/s = meter/second; comf. = comfortable; CSI = Composed spasticity index: 1 - 5 (normal), 6 - 9 (mild spasticity), 10 - 12 (moderate spasticity), 13 - 16 (severe spasticity) (Levin & Hui-Chan, 1993); **paretic-fast group; [f] = female participant; ratio L/S (see equation B); NSLD = normalized step length difference (see equation A); SD = standard deviation; L = left; R = right; NP = non-paretic; P = paretic.

TABLE 5-4. Gait Parameters at Pre-, Post- and Follow-up Evaluation.

Gait Characteristics (n=12)	**Baseline** Mean (SD)/ Range	**Post-training**	**Follow-up**	***p*-value (Hedges g_{av})**	
				Post-Base	FU-Base
Symmetry (L/S)*	1.39 (0.28)	1.16 (0.16)	1.16 (0.17)	**0.002 (1.01)**	**0.002 (1.00)**
	1.10-2.05	1.03-1.56	1.01-1.63		
Symmetry (NSLD)*	0.15 (0.05)	0.07 (0.04)	0.07 (0.03)	**0.003 (1.71)**	**0.003 (1.93)**
(abs[NP-P]/NP+P)	0.05-0.34	0.01-0.22	0.00-0.24		
10MWT, CWS	0.78 (0.22)	0.89 (0.18)	0.96 (0.28)	**0.043 (0.53)**	**0.004 (0.70)**
(m/s)	0.58-1.21	0.50-1.40	0.67-1.46		
10MWT, FCWS	1.09 (0.33)	1.27 (0.31)	1.29 (0.40)	**0.017 (0.54)**	**0.001 (0.53)**
(m/s)	0.74-1.65	0.65-1.74	0.7-1.96		
TUG, CWS (s)	14.44 (5.13)	12.51 (4.61)	12.06 (4.63)	0.175 (0.38)	0.105 (0.47)
	8.50-24.87	6.9-21.21	6.33-21.73		
TUG, FWS (s)*	11.13(3.9)	10.13 (3.84)	10.13 (4.30)	0.034 (0.25)	**0.008 (0.24)**
	5.00-17.33	6.99-19.77	5.54-20.3		
6MWT (m)	299 (93.4)	325 (110.7)	311 (104.6)	0.652 (0.25)	0.315 (0.12)
	207-505	207-584	203-500		

Abbreviations: L/S ratio (see equation B); abs = absolute value; NP = non-paretic; P = paretic; 10MWT = 10 Meter Walking Test; CWS = comfortable walking speed; FWS = fast walking speed; 6MWT = 6-Minute Walking Test; m = meters; s = second; *Data was not normally distributed and tested with the Wilcoxon signed rank test, significance level is adjusted for comparison p = 0.017; For normally distributed data significance level is at *alpha* $\leq$ 0.05 after post-hoc Bonferroni adjustment.

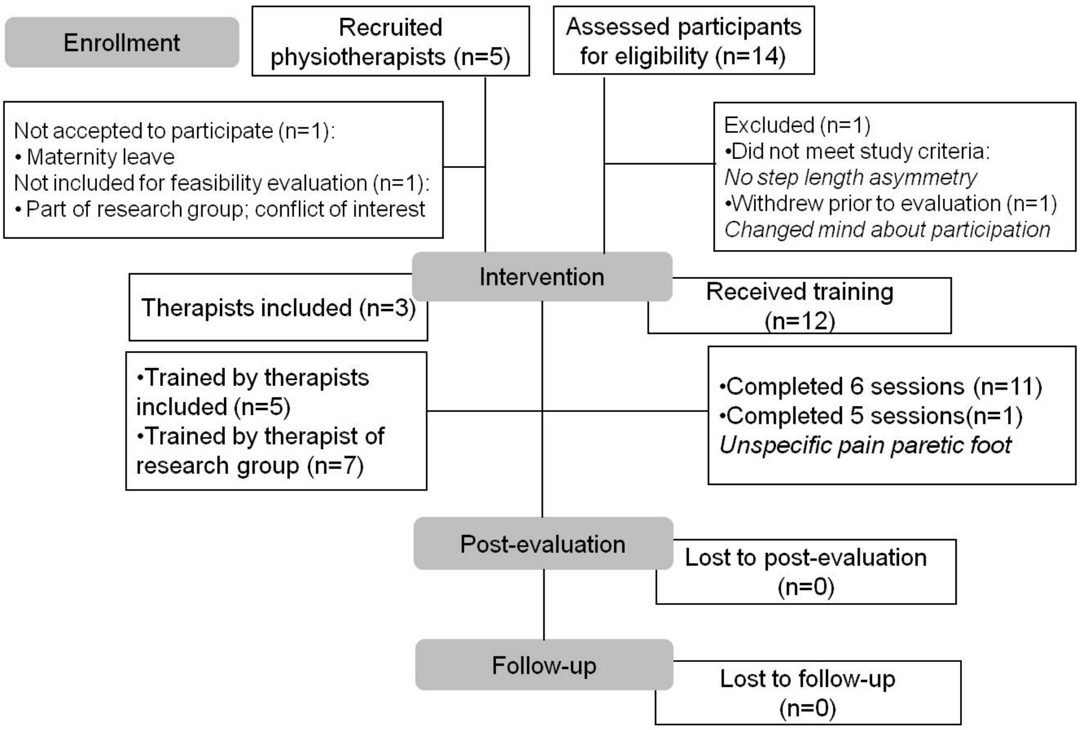

Figure 5-12. Flow-diagram presenting recruitment, training and evaluation processes.

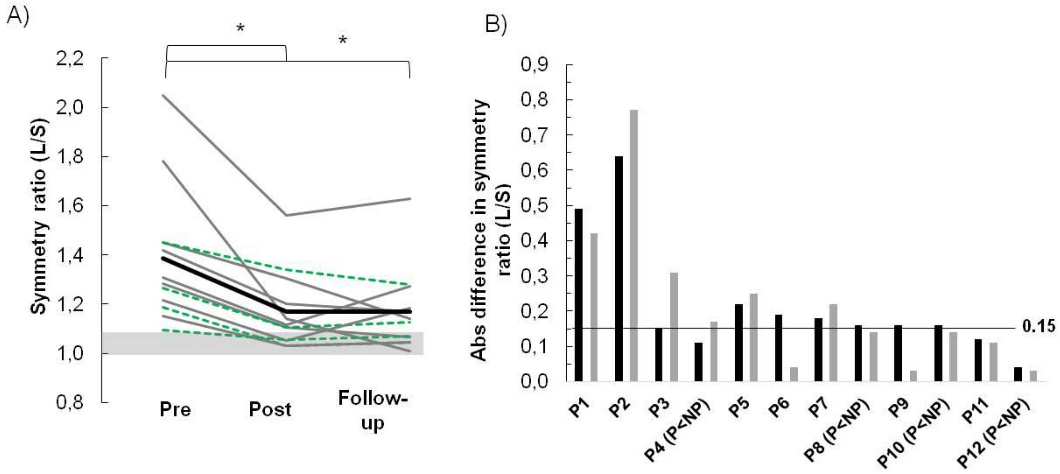

Figure 5-13. (A) Individual (grey and green) and mean (black) step length symmetry ratios (longer/shorter) during pre- post- and follow-up evaluation. Dotted green lines represent the values for individuals with initial shorter step on the paretic side (paretic-fast group). The grey balk on the level of ratio one represents the range of ratio considered as symmetrical (1.00-1.08) according to Patterson et al (2010). (B) Individual absolute changes in symmetry ratio from pre- to post-evaluation (black) and pre- to follow-up evaluation (grey) ordered by the size of the initial asymmetry (P1 to P12). P1 showed the largest initial asymmetry and P12 the smallest. P<NP = paretic fast group. The black horizontal line represents the threshold for clinically relevant change (0.15). *denotes significant differences between conditions between conditions ($p \leq 0.017$).

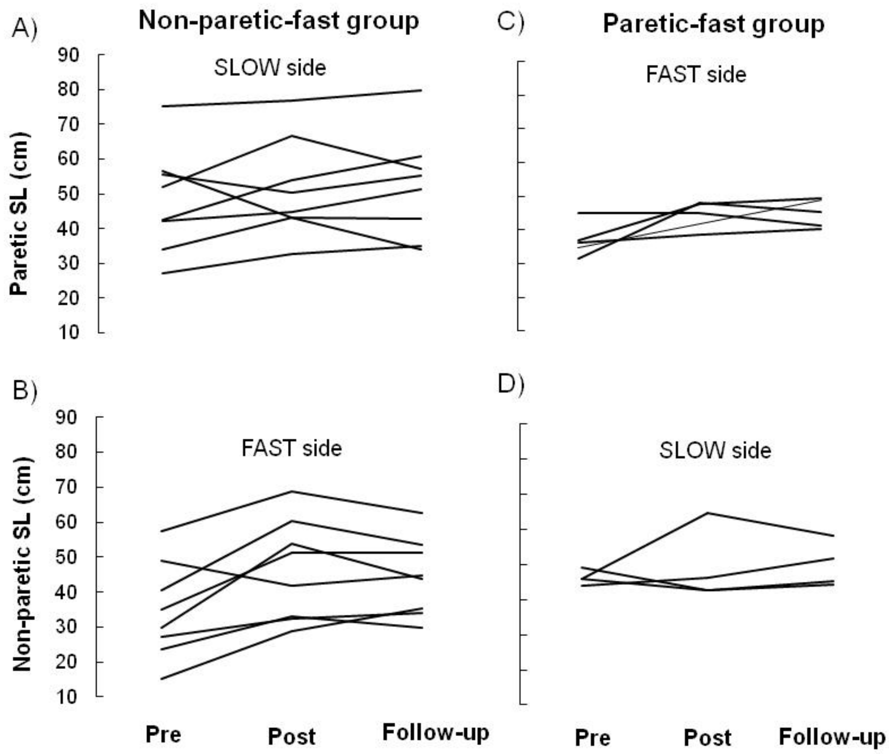

Figure 5-14. Individual step length (SL) during pre, post- and follow-up evaluation. A) paretic and B) non-paretic step lengths of participants trained in the non-paretic-fast group; C) paretic and D) non-paretic step lengths for participants trained in the paretic-fast group. The description FAST side refers to the side trained on the faster belt and vice versa for SLOW side.

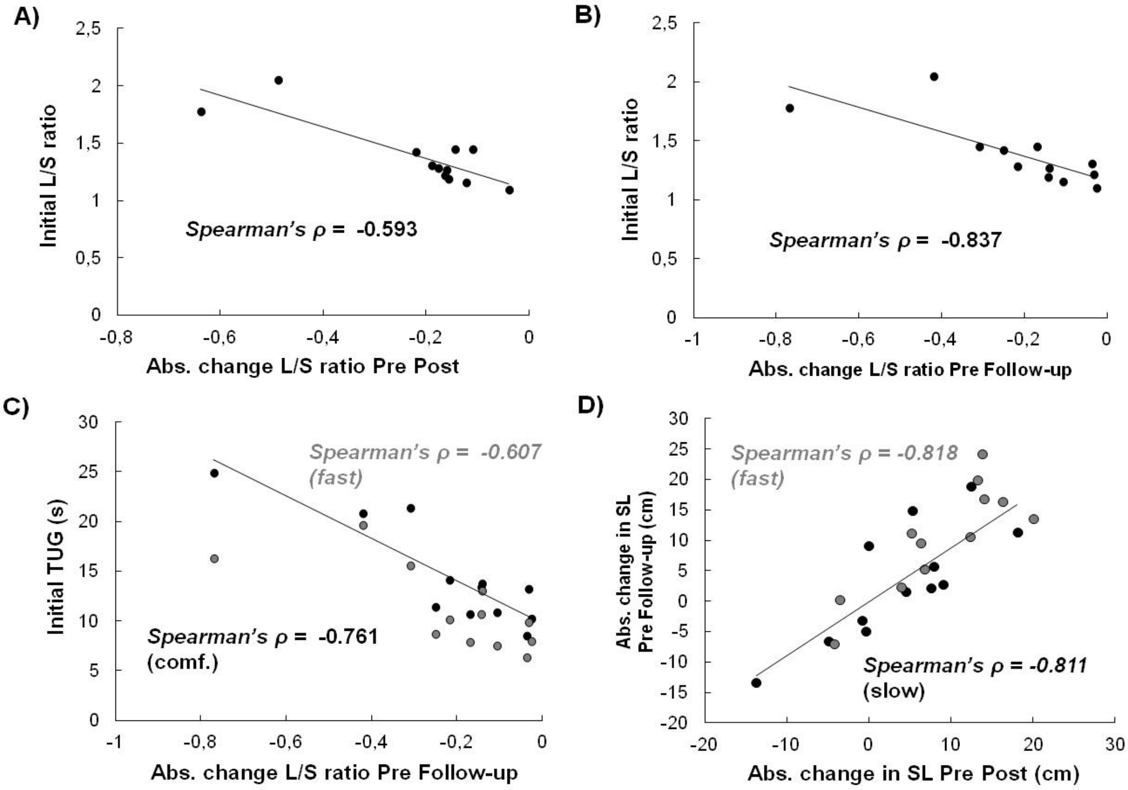

Figure 5-15. Associations between initial L/S ratio and A) absolute change in L/S ratio from pre- to post-evaluation and B) pre- to follow-up evaluation. C) Associations with initial TUG at comfortable (comf.) (black) and fast (grey) speeds with the absolute change in L/S ratio from pre- to follow-up evaluation, respectively. D) Associations between absolute (abs.) change slow (black) and fast (grey) step length (SL) from pre- to post-evaluation with abs. changes from pre- to follow-up evaluation, respectively. Abbreviations: abs = absolute, pre = pre-evaluation, post = post-evaluation, comf. = comfortable. L/S = longer/shorter, s = seconds; *denotes significant differences between conditions ($p \leq 0.05$).

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5.2.12 Supplementary Results Paper #2: Feasibility Study

This section will present the preliminary findings of the feasibility study conducted with the three participating physiotherapists from the rehabilitation center Constance-Lethbridge in Montreal, Canada. The following sections were submitted as supplementary files with the main manuscript for revision to the journal of Physiotherapy Theory & Practice March 29, 2016.

5.2.12.1 Introduction

A training protocol becomes fruitful only when it is applicable and useful in practice. In addition, Reisman et al. (2013) did not report information regarding the feasibility of their training protocol. SBT is a relatively new approach and there is still limited knowledge about its feasibility in practice. Thus, this paper aims to present some preliminary findings on the feasibility of the current protocol in a clinical setting. Physiotherapy has been considered one of the key disciplines of interdisciplinary stroke rehabilitation (Veerbeek et al, 2014). However, based on health services research, a consistent finding is the gap between evidence and practice (Maher et al, 2004). To introduce evidence into routine daily clinical practice is very challenging (Scurlock-Evans, Upton, and Upton, 2014) whereas one of the strategies suggested was the direct involvement of relevant practitioners (Grol and Grimshaw, 2003). The relevant practitioners in this study are physiotherapists. According to literature, this is an occupational group with an interest in evidence-based medicine (Salbach et al, 2007). Seventy-eight percent of 207 physiotherapists practicing stroke rehabilitation considered research findings as useful for clinical practice (Salbach et al, 2007).

The main goal of this study was to investigate the effects of this progressive approach to SBT training on gait ability in individuals post-stroke with different initial SL asymmetries. This supplementary file provides some preliminary findings related to the feasibility of this promising intervention in a clinical setting. We hypothesized that the therapists involved would find the protocol feasible for their clinical practice.

5.2.12.2 Methods

Feasibility

Two areas of feasibility testing were addressed: practicality and implementation (Figure 5-16). An intervention is considered practical when it can be delivered even when personal resources, time and commitment are limited. The implementation assesses the likelihood and extent to which an intervention can be implemented as planned (Bowen et al, 2009). For this purpose, physiotherapists of the neuro-rehabilitation department in an outpatient facility were recruited. The team consisted of 4 therapists working full-time. The physiotherapists were asked to answer the following questionnaires after completion of the study: Organisational readiness for implementing change (ORIC) (Shea et al, 2014); perceived use of SBT and the training protocol (adapted from Davis FD, 1993) and the PERFECT Tool questionnaire (Section 3) (Menon et al, 2010). The ORIC is a 12-item questionnaire to measure commitment and efficacy related to implementing a change. The "perceived use of SBT and the training protocol" was adapted from the model by Davis et al. (1993). In our study, "electronic mail" used in the original questionnaire, was replaced by "SBT and training protocol". Section 3 of the PERFECT Tool was used to measure the perceived changes in clinical practice over the last year, including barriers and facilitators to change. The evaluation of the questionnaires is not based on a qualitative analysis model or concept. The responses from questionnaires were examined and used for discussion purposes in an informal interview with three therapists. One physiotherapist from the research team attended the training sessions. This therapist did not participate in the evaluation of feasibility because of the conflict of interest (Figure 5-12, main manuscript).

Practicality
Number of completed trainings
Number of adverse events
Duration of training achieved at the last session
Sufficient personal resources for the number of therapists required
Time proposed for training sufficient
Implementation
Organisation readiness for implementation:
Organisational readiness for implementing change (ORIC) (Shea et al., 2014)
Usefulness:
Perceived use of a split-belt treadmill (SBT) and the training protocol (adapted from Davis, 1993)
Facilitators and barriers of change in clinical practice:
PERFECT Tool questionnaire (Section 3) (Menon et al., 2010)
Implementation
Organisation readiness for implementation:
Organisational readiness for implementing change (ORIC) (Shea et al., 2014)

Figure 5-16. Areas of feasibility.

5.2.12.3 Statistical Analysis

Since the sample size of therapists was small (n = 3), outcomes of the questionnaires were presented descriptively.

5.2.12.4 Results

5.2.12.4.1 *Practicality*

Participants

All 12 participants completed the three evaluation sessions and 11/12 participants the six training sessions (Figure 5-12). One participant completed five sessions because of unspecific pain at the dorsum of the paretic foot during sessions four and five. Two participants were able to walk during split-belt conditions for the entire 20 minutes without breaks and the others required 1-3 breaks. Nine participants always achieved the belt speed ratio of 2:1. The lowest speed ratio was at 1.6:1 with 0.7 m/s (slow) and 1.12 m/s (fast) (P2).

Comfortable speed on the treadmill increased from 0.05 to 0.30 m/s from the first to the last session with two individuals remaining at the same training speeds (p = 0.005). Perceived exertion during split-belt conditions ranged from 3 - 7 (moderate to very severe exertion) during the first session and from 2 - 5 (slight to severe) during the last session (p > 0.05). Total training duration ranged from 40 to 50 minutes including preparation of the participant.

Personnel Resources

Three out of four physiotherapists in the department agreed to participate in the study while one was on pregnancy leave. The physiotherapists from the department had practical experience in neuro-rehabilitation of 2.5, 3.0 and 12.5 years. After one training session on how to use the SBT, all three physiotherapists were able to apply the protocol on the SBT. A second therapist assisted at the first training session for each participant. For the following sessions, one physiotherapist was sufficient for the training except in one case (P2; Table 5-4), because of the participant's relatively low functional level (speed and endurance). Thus, another therapist assisted for safety purposes. Five participants were trained by the three physiotherapists and seven by the physiotherapist from the research team.

5.2.12.4.2 *Implementation*

The evaluation of the three questionnaires and an informal interview showed that the 3 physiotherapists considered the SBT training protocol as feasible and useful in their clinical environment. For example, the "Perceived ease of use questionnaire" revealed that the physiotherapists strongly agreed that they would use the SBT on a regular basis and agreed that it was easy to learn how to apply the protocol, remember how to perform the protocol and use the treadmill. The PERFECT TOOL questionnaire revealed that the physiotherapists considered being involved with the researchers, and exchanging knowledge among therapists and researchers as major contributors helping to bring change to their practice. For all therapists, "lack of time" and "busy schedule" were the main barriers to implementation. On an organizational level, the physiotherapists generally agreed that there is a readiness to implement change (ORIC, 12 items) with a median score of 4 (Scale 0 - 5; 0 = disagree, 5 = strongly agree).

5.2.12.5 Discussion

The present findings suggest that this protocol is applicable in a rehabilitation setting. The three physiotherapists involved considered the current protocol as practical in their outpatient setting. All of the therapists had their first experience with the SBT within this study. The perceived ease of use questionnaire revealed that the three therapists considered it easy to learn and apply the SBT protocol. The therapists reported that the treadmill is easy to control since it does not differ much from a standard “one-belt” treadmill. The protocol is easy to learn and apply because there is only one factor to control related to determining the 2:1 belt speed ratio based on a participant’s comfortable speed. Based on the PERFECT TOOL questionnaire, it was concluded that the direct involvement of the physiotherapists and the research coordinators of the rehabilitation site early in the process has a positive effect on the implementation of this new intervention as a training tool. In addition, therapists considered the clinical administration ready for and supportive of implementing this new intervention. The outpatient center has ongoing plans to use the protocol in individuals post-stroke. According to the findings, the three therapists feel that the organization will successfully support people as they adjust to this new approach to training their asymmetrical patients.

5.2.12.6 Study Limitations

The current training included six sessions of SBT walking. The number of sessions was based on a compromise to assure the availability of physiotherapists. Six sessions are fewer than the usual number of gait interventions (Tilson, Settle, and Sullivan, 2008; Veerbeek et al, 2014). Nevertheless, the intervention resulted in clinically relevant changes in gait parameters. The sample of therapists from the same outpatient rehabilitation site evaluated the feasibility based on five training sessions which are, on average, only 1.5 participants per therapist. However, this limited amount of training conducted was sufficient for therapists to conclude that the SBT protocol is feasible in their outpatient rehabilitation practice. Furthermore, this study did not investigate the feasibility from the participants’ points of view. Only personal feedback was given by the participants, generally expressing great appreciation for the training.

5.2.12.7 Future Studies

This pilot study gave the first insight into the effects and feasibility of a progressive, speed-dependent approach of repeated SBT walking. The questionnaires used in the scope of this study could be valid tools for the evaluation of the perception of usefulness and readiness for implementation. Future studies should use a larger sample size of individuals post-stroke in order to test whether the small to moderate effect size and the lack of correlation among changes in SL asymmetry and other parameters was due to the small sample size or the SBT protocol.

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5.3 Paper #3: A Pilot Study on the Quantification of Lower Limb Muscle Activity and Joint Moments Underlying the Reduction of Step Length Asymmetry Over Ground in Individuals Post-stroke After Repeated Split-belt Treadmill Walking.

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Paper ready to be submitted.

As principle author, I confirm that I contributed to the majority of the procedures required for the completion of this paper. The results presented in the current paper are obtained from the same group of participants using the same protocol as presented paper #2, (see section 4.2). For this paper I was centrally involved in the data collection and treatment of data obtained from 3D-motion and EMG analysis used for the results in the present paper. Bradford J. McFadyen provided direction in appropriate treatment of EMG signals and the revision of the manuscript. Sylvie Nadeau supervised and contributed to each of these procedures starting from the protocol development to the final revision of the manuscript.

5.3.1 Preamble

Findings from our previous cross-sectional SBT study and others (Lauzière et al, 2014, 2016) suggest a particular contribution of the ankle plantarflexors to step length changes. Furthermore, paretic fast-belt walking has been found to have adverse effects (i.e. a reduction of amplitude) on already impaired paretic peak plantarflexor moments (Lauzière et al, 2014).

In the current paper, we set out to evaluate whether similar conclusions as presented within the aforementioned cross-sectional study can be reached when testing the effects of repeated exposure to SBT on step length asymmetry in individuals post-stroke. Similar to the majority of the cross-sectional studies, Reisman et al. quantified the effects of SBT training solely on using spatiotemporal parameters (Reisman et al, 2013). In contrast, this paper provides first exploration of the effects of repeated SBT walking on lower limb biomechanics. In combination with the findings presented in the paper #2, we hope to provide some basic recommendations and inputs for future research. The paper contains references to our previous papers (i.e., paper #1 and #2). Given that these papers are not yet published in the scope of this doctoral work we refer to these papers as Betschart, McFadyen, and Nadeau (under revision) (paper #2) and Betschart et al. (under revision) (paper #1). As aforementioned, once papers #1 and #2 are published, the present paper will be submitted.

5.3.2 Abstract

The main goal of this study was to investigate changes in muscle activity and joint moments related to step length symmetry improvements in individuals post-stroke following repeated split-belt treadmill (SBT) walking. Twelve individuals with a first unilateral cerebral stroke presenting initial step length (SL) asymmetry (ratio = 1.10 - 2.05) (10 males; mean age 52 (SD 9.3 years); mean time post stroke 23 (SD 24.7 months); nine left-sided stroke) were included. Participants were trained during six sessions of split-belt treadmill walking using an error-augmentation protocol. The training resulted in a reduction in SL asymmetry during walking over ground retained over one month post-training ($p = 0.002$). These improvements were in line with bilateral increases in SL and joint moments. However, changes were particularly pronounced on the side trained on the fast belt and reached significance when comparing pre-training to follow-up data ($p \leq 0.037$). No changes were found in average muscle activity ($p > 0.170$). The findings indicate that a reduction in SL asymmetry is achieved by a variety of biomechanical and muscle activity changes in a group of individuals post-stroke. The side trained on the fast belt and the plantarflexors tended to be strong contributors to symmetry improvements.

Keywords: Stroke, Step length Symmetry, Training, Muscle activity, Joint biomechanics, Split-belt treadmill

5.3.3 Introduction

Stroke survivors have to live with different persistent motor and cognitive impairments (HSF, 2016) which impede their ability to perform activities of daily living, such as self-care or walking (Harris and Eng, 2004; Lord et al, 2004). The probability of regaining independent walking after three to six months post-stroke was found to be about 60% (Lord et al, 2004; Preston et al, 2011). Among the individuals who recover independent walking, a substantial number still present a gait that is often characterized as slow and asymmetrical (Lord et al, 2004; Kollen et al, 2005; Patterson et al, 2008; Ada, Dean, and Lindley, 2013). Step length (SL) asymmetry is considered particularly resistant to conventional gait therapy (Patterson et al, 2015). Only 14% of patients reduced their initial SL asymmetry compared to the 30% and 62% who improved walking speed and balance, respectively. Treadmill interventions (fast walking, bodyweight support) or auditory feedback led to improvements in temporal (e.g. stance duration), but not spatial asymmetries (e.g., step length) (Hassid et al, 1997; Schauer and Mauritz, 2003; Lamontagne and Fung, 2004; Thaut et al, 2007; Combs, Dugan, Ozimek, and Curtis, 2013).

Yet, there is a growing body of evidence about the successful reduction in SL asymmetry in individuals post-stroke after split-belt treadmill (SBT) walking. Repeated exposure to SBT walking led to clinically and statistically relevant reductions in SL asymmetry during walking over ground that was maintained over one month (Betschart, McFadyen, and Nadeau [under revision]; Reisman et al, 2013). These training protocols induced an adaptation of the walking pattern by applying the principle of error-augmentation which was investigated extensively in previous literature within cross-sectional studies (Reisman, Block, and Bastian, 2005; Reisman, Wityk, Silver, and Bastian, 2007; Bastian, 2008; Malone, Bastian, and Torres-Oviedo, 2012; Lauzière et al, 2014a). In accordance with healthy controls, individuals post-stroke showed the capacity to adapt stride-by-stride their SL and double support duration to adjust the gait pattern to the unequal belt speeds (split-belt configuration) (Reisman, Block, and Bastian, 2005; Reisman et al, 2007; Malone and Bastian, 2013). When both belts were set at equal speeds again (tied-belt configuration), immediate after-effects were found. The after-effects were characterized by a larger step on the side which previously walked on the faster belt and often a decrease in SL on the slow belt side, when compared to their initial SL.

In individuals post-stroke with initial SL asymmetry, this led to a more symmetrical pattern provided that the side with the shorter step walked on the faster belt during split-belt configuration (Betschart et al, [under revision]; Reisman et al, 2007; Malone and Bastian, 2013; Lauzière et al, 2014a). These after-effects in SL were moderately to strongly associated with an increase in plantarflexion moment (Lauzière et al, 2014a) and plantarflexor muscle activity (Betschart et al, [under revision]) on the side that walked on the slower belt. Fast belt walking, however, led to an increase in activity of dorsiflexor, hip flexor and knee extensor muscles (Betschart et al, [under revision]) and a decreased plantarflexion moment particularly during the mid and late stance phase of gait (Lauzière et al, 2014a). These changes were found regardless of which side was on the faster belt (non-paretic or paretic leg). Thus, paretic fast-belt walking resulted in a negative effect on paretic plantarflexion moment at push-off. This indicates that individuals with a shorter paretic SL can improve SL symmetry, but there is a risk to reducing the use of the often impaired paretic plantarflexors negatively impacting walking speed (Nadeau et al, 1999; Milot et al, 2006) and step length symmetry (Allen et al, 2011).

Impaired function of paretic lower limb muscles, including weakness and deficits in coordination and activation, is a major consequence secondary to a cerebral stroke (Hafer-Macko, Ryan, Ivey, and Macko, 2008; Arene and Hidler, 2009; Andrews et al, 2010; Sulzer et al, 2010). Indeed, impaired plantar- and dorsiflexor muscles were considered as factors contributing to gait asymmetry (Hsu, Tang, and Jan, 2003; Lin, Yang, Cheng, and Wang, 2006). The present protocol was tested previously and successfully improved SL asymmetry and walking speed in the present group of participants (n = 12) even after one month (Betschart, McFadyen, and Nadeau, [under revision]). Considering the aforementioned in SBT-induced short-term changes in EMG and joint moments, we need to better understand the potential of the present protocol to successfully improve impaired lower limb muscle function.

For this purpose, a pilot investigation was conducted on muscle activity and biomechanical variables during walking over ground in individuals post-stroke trained with a SBT protocol.

Based on previous findings we hypothesized that: 1) plantarflexor muscles and joint moments during late stance phase trained on the slow belt would be the main contributors to the increase or decrease in SL when walking over ground (Betschart et al, [under revision]; Lauzière et al, 2014a); 2) paretic plantar- and dorsiflexors would present increased muscle activity after slow and fast belt walking, respectively (Betschart et al, [under revision]); 3) participants trained with the paretic side on the fast belt would show an increased contribution of the proximal muscles compared to the participants trained with the non-paretic side on the fast belt (Betschart et al, [under revision]).

5.3.4 Methods

5.3.4.1 Participants

Twelve individuals with chronic stroke (nine with left hemiparesis; two women) participated in the study. Inclusion criteria were: first unilateral non-cerebellar stroke ≥ six months prior to study; presence of SL asymmetry (SL ratio ≥ 1.08) (Patterson et al, 2010); and capacity to walk without technical assistance (cane, orthoses) over 15 meters (Functional Ambulation Category > 4) (Mehrholz et al, 2007) at a speed greater than 0.5 m/s. However, the use of aids for longer distances or outdoors was not a criterion for exclusion. Participants had to achieve at least 200 meters during six minutes walking over ground as a minimum requirement of endurance to sustain training bouts of at least five minutes. Deficits that could limit the comprehension of training instructions, such as major pain, hemineglect or hemianopsia, communication deficits (aphasia) or cognitive deficits (Mini Mental State Examination score < 25/30) led to an exclusion from the study. During the course of the study, participants were informed to not partake in other gait related training or physical exercises. All participants signed an informed consent prior to the study, which was approved by the local ethics committee.

5.3.4.2 Evaluation

Participants underwent four evaluation sessions. The first evaluation consisted of a clinical assessment to test the participant's eligibility. Included participants were then assessed on the following three occasions: 1-2 days prior to the training (pre-evaluation), 1-2 days after the last training (post-evaluation) and four weeks after post-evaluation (follow-up evaluation). During these three evaluations, the participant's gait pattern was quantified with clinical and biomechanical measures. No evaluation was conducted during the training sessions.

5.3.4.3 Clinical Evaluation

A questionnaire served to obtain information about stroke characteristics, aforementioned deficits and further medical conditions. A detailed description of the assessments was previously published (Betschart, McFadyen, and Nadeau, [under revision]). Gait ability was assessed with gait speed, SL asymmetry, endurance and functional mobility. Comfortable and fast gait speeds were assessed with the 10 Minute Walk Test (10MWT) with an average of two trials (Flansbjer et al, 2005). Gait endurance was tested with the 6-Minute Walk Test (6MWT) based on the standards of the American Thoracic Society (ATS, 2002) and the American Heart Association (Gordon et al, 2004). Functional mobility was assessed with the Timed Up and Go Test (TUG) at fast (two trials) and comfortable speeds (two trials) (Podsiadlo and Richardson, 1991). SL was obtained by asking the participant to walk over a 5-6 m paper-carpet with light-color pencils affixed to the heel of the shoe (Cerny, 1983). The average of 5-6 steps was taken for each side to calculate SL. The clinical parameters of the participants in the present paper were evaluated within the scope of the previous publication to study the effects of repeated SBT walking on clinical gait parameters (Betschart, McFadyen, and Nadeau, [under revision]). The present paper discusses the effects of SBT walking on muscle activity and joint net moments in this group of participants.

5.3.4.4 Biomechanical Evaluation

5.3.4.4.1 *Data Collection*

The biomechanical and electromyography (EMG) data were collected while participants walked over a 5-7 m walkway at a comfortable speed. Participants were instrumented with 34 active infrared markers placed on the lower limbs and pelvis to capture 3-dimensional kinematic data during walking (Optotrak Motion Capture Certus System). To quantify muscle activity, a wireless TeleMyo DTS system (Noraxon Inc. USA) was used to capture EMG signals from five lower limb muscles assessed bilaterally: tibialis anterior (TA), gastrocnemius lateralis (GL), vastus lateralis (VL), rectus femoris (RF) and semitendinosus (ST). Self-adhesive surface electrodes (Ag/AgCl) were placed over the belly of the targeted muscles. Skin preparation and electrode placement were based on the ISEK standards (Merletti, 1999). Each electrode position was marked on the participant's skin with a pencil during the first evaluation session to reproduce electrode placements each session. Kinematic and EMG data were registered at 60Hz and 1200Hz respectively during walking over ground at comfortable speed. Ground reaction forces (GRFs) were recorded by three floor-integrated force plates at a sampling frequency of 600Hz. For each walking trial, 3-4 gait cycles were registered. For off-line treatment, gait cycles were only valid and used when participants placed their leading foot on the force plate during initial contact. In order to obtain the most natural gait pattern, the location of the force plates was not revealed to the participants. Walking trials were repeated until at least five valid gait cycles were obtained for each leg. To avoid fatigue, participants were allowed to take breaks between the trials.

5.3.4.4.2 *Data Analysis*

Kinematic and force data were filtered with Butterworth, 4th order, zero-lag, filters at cut-off frequencies of 6Hz and 10Hz, respectively. To match the kinematic data with the force data, signals were resampled at 60Hz for offline treatment. Net joint moments on the sagittal plane at the ankle, knee and hip joints were estimated using an inverse dynamic approach (Winter, 2009; Chowdhury and Kumar, 2013). Specific periods of the gait cycle were identified (e.g., push-off phase) and the peak net joint moments were identified (e.g., peak plantarflexion moment during push-off) and further used in the statistical analyses. Cycle duration, SL and walking speed were determined with kinematic data from the heel marker and the vertical GRF. Cycle duration was obtained from the time between subsequent contacts of the same foot. The duration was normalized to 100%. SL was defined as the anterior-posterior distance between the trailing and leading heel markers for two consecutive contacts (Reisman et al, 2007). EMG signals were band-pass filtered using a Butterworth, 4th order, zero-lag filter at 20 - 400Hz. For each trial, RMS values were calculated over 300 signal points and time normalized with the gait cycle (100%). For amplitude normalization, the RMS value was expressed as a percentage of the average peak RMS value obtained during 3-4 gait cycles from the first trial. Thus, the RMS values were normalized with the peak RMS obtained on the same evaluation day (Blanchette, Moffet, Roy, and Bouyer, 2012; Cronin et al, 2015).

SL asymmetry was calculated by a simple ratio using the side with the longer SL divided by the side with the shorter SL (Betschart, McFadyen, and Nadeau, [under revision]; Patterson et al, 2010; Lewek and Randall, 2011).

$$(\mathrm{L/S\ ratio}) = \frac{\text{Longer SL}}{\text{Shorter SL}} = \frac{\text{Slow}}{\text{Fast}}$$

Equation 5-3. Symmetry ratio

A value of “1” represents perfect symmetry. Since we were interested in investigating the effect of fast-belt and slow-belt walking, the description of ‘slow SL’ and ‘fast SL’ was used. Thus, the ‘shorter SL’ in the equation referred always to the ‘fast’ side because the side with the shorter step walked on the faster belt for training. For symmetry ratios of net joint moments and muscle activity, the following index was used.

$$\text{Symmetry index} = \frac{\text{Fast side}}{\text{Slow side} + \text{Fast side}}$$

Equation 5-4. Symmetry index

Perfect symmetry is represented by 0.50. The use of an index prevented from biasing the symmetry ratios because kinetic or EMG values were not always smaller on the side with the shorter SL (fast side) as compared to the side with the longer SL (slow side). Dividing a larger value by a smaller one (e.g., 4/2 = 2) results in a value with a different margin to the symmetry value (= 1) compared to the value obtained from the inverted ratio (e.g., 2/4 = 0.5) (Lauzière, Betschart, Aissaoui, and Nadeau, 2014b).

5.3.4.5 Training Protocol and Sessions

The present protocol was tested in a previous study for its effects on different gait parameters during walking over ground at one day post-training and at a one-month follow-up (Betschart, McFadyen, and Nadeau, [under revision]). The biomechanical variables and muscle activity were obtained from the 12 participants analyzed in the scope of this prior investigation. A detailed description of the protocol can be found in the aforementioned paper.

In general, the training consisted of six sessions (2-3 sessions/week) of error-augmentation-based SBT (Bertec Inc.) walking and was conducted by trained physiotherapists. Participants were trained during 20 minutes of walking on a split-belt configuration with a belt speed ratio of 2:1. Therefore, the slower belt was set at comfortable speed (split-belt configuration). The leg with the shorter SL walked on the faster belt (= fast leg) (Reisman et al, 2007; Malone and Bastian, 2013; Lauzière et al, 2014a). Therefore, participants with a shorter paretic SL had to walk with the paretic leg on the fast belt (paretic-fast group) and vice versa for participants with shorter non-paretic SL (non-paretic-fast group). Comfortable belt speed was tested and adjusted prior to each training session during walking at the comfortable speed used for the previous training. Speeds of both belts were subsequently increased or decreased at 0.05m/s intervals until the participant was perceived to walk at a comfortable speed. The participants were motivated to achieve 20 minutes of split-belt walking and were allowed to take breaks. After the 20 minutes of split-belt walking, participants walked again at tied-belt configuration without any verbal or visual feedback with respect to their gait pattern. This served only as a cooling down period, and for safety purposes to prevent participants from going from constant perturbation directly to walking over ground. During the entire training on the SBT, participants wore a safety harness without body-weight support and were allowed to hold the handrails.

5.3.4.6 Statistical Analysis

Descriptive statistics are presented for demographic information and stroke characteristics. Normal distribution of dependent variables (SL symmetry, walking speed, %max RMS and net joint moments) was tested with the Shapiro-Wilk approach to define whether parametric or non-parametric statistics should be used. The effect of training over time (pre-, post-, follow-up evaluations) on dependent variables (within-subjects) was tested with repeated-measure ANOVA or a Wilcoxon paired signed-rank test using SPSS 22.0 (SPSS Inc.). Effect sizes were computed for effect of time using Hedges's g_{av} (Lakens, 2013), which is a Cohen's d value corrected for the sample size using means, SDs and sample sizes and recommended for small samples. These g scores can be interpreted the same way as the Cohen's d scores: g around 0.2 is interpreted as "low", 0.5 "average" and 0.8 as "strong". For step length the repeated-measure ANOVAs included the additional factor 'condition' next to the factor 'time' for the two conditions of data collection for step length (laboratory and carpet walking) to assess whether the conditions resulted in comparable data. The level of significance was accepted with $p \leq 0.05$ after post-hoc analysis with Bonferroni correction.

5.3.5 Results

Among the twelve participants, only one participant stopped after five sessions because of non-specific pain at the dorsum of the paretic foot, but this pain did not occur during the last evaluation session. However, two participants had difficulties during pre-evaluation to step with only one foot on a force plate because of very short step lengths. Also, EMG data from one participant was lost during post-evaluation sessions due to signal cut-outs. While attempting to collect more trials, the evaluation had to be stopped because of fatigue. Thus, results are presented for ten participants whose characteristics are presented in Table 5-5. Four participants were trained with the paretic side on the fast belt and vice versa for the remaining six participants. The results will be presented referring to 'slow' and 'fast' side, corresponding to the belt on which the leg was trained during split-belt configuration. The slow side always corresponds to the side with the initially longer step length.

5.3.5.1 Step Length Symmetry

Six sessions of SBT training resulted in an improvement in symmetry during walking over ground during clinical and biomechanical evaluations with a strong effect of period (p = 0.002) (Figure 5-17). The symmetry ratios obtained from the two measures corresponded to each other showing no effect of walking condition (p = 0.470) and no interactions were found between period and walking condition (p = 0.221). The following section will only present changes in SL obtained during 3D-motion analysis in the laboratory.

Fast and slow SL presented a significant interaction with the factor of time (p=0.008). Both sides increased SL, but a more pronounced increase was observed on the fast side (Figure 5-18). Fast SL showed an effect of time (p = 0.004) with a significant increase between pre- and the one-month follow-up evaluation (p = 0.038). Slow SL approached significance for effect on time (p = 0.059).

5.3.5.2 Walking Speed

Participants' walking speeds during the data collection showed significant effects of time (p = 0.001) with an average increase of 0.11 m/s from pre- to post-evaluation (p = 0.074) and reaching significance at the one-month follow-up (+0.16 m/s; p = 0.009). Six participants increased speed from pre- to follow-up evaluation by 0.10 - 0.20 m/s, a change considered within the range for a clinically important increase in comfortable walking speed (Perera et al, 2006, Bohannon et al, 2014).

Effect sizes for changes in SL symmetry and speed were in general large particularly between pre- and follow-up evaluation. Effect sizes for all parameters are reported in Table 5-6.

5.3.5.3 Net Joint Moments

A significant increase in peak net joint moments was found only on the limb which was trained on the faster belt and particularly between the pre- and one-month follow-up evaluations (Figure 5-19). Improvements were found in fast plantarflexion moment, knee flexion moment and hip extension moments during the stance phase of gait. No changes were observed for the net moments in dorsiflexion, knee flexion and extension during early stance and hip flexion during late stance.

Plantarflexion moments increased bilaterally after training without interaction effects (p = 0.138). Changes were more pronounced and significant only for the fast side. Post-hoc analysis revealed that the fast plantarflexion moment approached significance from pre- to post-evaluation (p = 0.082) and increased significantly by 12% from pre- to follow-up evaluation (p = 0.016). The increases were observed at the end of the stance phase, from 40-60% of the gait cycle. This corresponds to the initiation of foot push-off.

Fast peak knee flexion moments, observed during 20 - 60% of the gait cycle, was on average 23% (p = 0.002) higher during post-evaluation and showed the tendency to remain increased after one month when compared to pre-evaluation (+25%; p = 0.080). Peak hip extension moments during early stance (0 - 20% of the gait cycle) increased bilaterally from pre- to follow-up evaluation. An increase was found on the fast side (+28%; p = 0.037) and on the slow side approaching significance (+21%; p = 0.084). The Hedges's g_{av} scores for changes in net moments were generally low except for fast plantarflexion (0.60) and fast hip extension moments (0.58) between pre- and follow-up evaluation.

The indices of symmetry in net joint moments did not reveal any significant changes after six sessions of training and at one-month follow-up (p > 0.19) (Table 5-7).

5.3.5.4 Muscle Activity

Group means muscle activity (%max RMS) did not change over time for all muscles analyzed for both slow and fast sides (p > 0.17), nor did the indices of symmetry between sides (p > 0.40). For all muscles tested, effect size remained small with Hedges's g_{av} scores ranging from 0.15 - 0.34. Except the GL muscle showed a Hedges's g_{av} of 0.55 (pre- to post-evaluation) which is considered as moderate (Figure 5-20).

5.3.6 Discussion

The present study investigated changes in lower limb muscle activity and net joint moments in individuals post-stroke during walking over ground after six sessions of SBT training. Biomechanical and EMG data from 10 individuals with chronic stroke were analyzed at one-day and one-month post-training. Clinical and biomechanical results showed that after six sessions of SBT training, all ten participants improved SL symmetry and walking speed. Only one participant had to stop after five training sessions. No other adverse events were found during the training. Lower limb net joint moments increased particularly on the fast side with significance reached at the one-month follow-up evaluation when compared to pre-evaluation. The SBT training did not induce significant changes in group average lower limb muscle activity.

5.3.6.1.1 *Changes in SL and SL Asymmetry*

Repeated exposure to SBT walking reduced SL asymmetry at one-day post-training and was maintained over one month with estimated strong effect sizes (> 1.00). The SL ratios obtained during 3D-motion analysis corresponded to the values obtained during clinical evaluation (paper carpet). For more detailed information about the clinical relevance of individual changes we refer to the paper by (Betschart, McFadyen, and Nadeau, [under revision]).

In general, these changes in SL asymmetry were achieved by a bilateral increase in SL, particularly on the side that was trained on the fast belt (Figure 5-18 A-B), which is in accordance with previous findings (Reisman et al, 2013). The accentuated increase in fast SL confirms the previous suggestion that the side with the shorter SL should be trained on the faster belt if a reduction in SL asymmetry post-stroke is aimed (Reisman et al, 2007; Malone and Bastian, 2013; Lauzière et al, 2014a).

5.3.6.1.2 *Changes in Net Joint Moments*

In line with the changes in SL, the increase of hip extension, knee flexion and plantarflexion moments were found particularly on the side trained on the fast belt. An increase in fast hip extension moment during early stance of gait was found as well as a direct after-effect of six minutes of SBT walking in individuals post-stroke (n = 20) and healthy controls (n = 10) (Lauzière et al, 2014a). The 28% increase in hip extension moment during early stance observed in the present group exceed slightly exceeds 17.1% of change found by Lauzière et al. (Lauzière et al, 2014a). Further, Lauzière et al (2014) reported SBT-induced changes in the plantarflexion moments with strong correlations to contralateral step length changes (Lauzière et al, 2014). While Lauzière et al (2014) reported immediate after-effects of plantarflexion moments on the slow side (mean increase of 16%), repeated exposure to SBT walking led to an increase in plantarflexion moments particularly on the fast side (mean increase of 12%) with an effect size of 0.60 in our participants.

The moderate effect size and the comparable percentages of change with the findings by Lauzière et al (2014) indicate that participants in the current study presented reasonable effects in plantarflexion and hip extension moments. The slow plantarflexion and hip extension moment did not increase significantly (p = 0.131) but showed a moderate effect size of 0.71 and 0.78, respectively. The bilateral, but more pronounced increase on the fast side in plantarflexion and hip extension moments correspond to the bilateral increase observed in step length in the present (Figure 5-18) and in previous studies on repeated SBT walking (Betschart, McFadyen, and Nadeau, [under revision]; Reisman et al, 2013). Changes in plantarflexion moments were found during the late stance phase (Figure 5-19). During this phase, the plantarflexors play a major role in the forward propulsion of the body and swing initiation in healthy (Zajac, Neptune, and Kautz, 2002) and individuals post-stroke (Allen, Kautz, and Neptune, 2014) influencing contralateral (Lauzière et al, 2014; Allen et al, 2011) and ipsilateral step length (Neptune et al, 2003), respectively. Therefore, the plantarflexors are likely the relevant contributors to the increase in SL in the individuals post-stroke analyzed in the current study, as hypothesized. The increase in hip extension moments was found during the early stance phase most likely due to particularly the increase in walking speed. At initial contact, hip extensors counteract the impact of heel contact to provide the support and progression of the body at beginning of stance (Zajac, Neptune, and Kautz, 2003, Siegel et al, 2004). This contribution to support and forward progression increases along with a corresponding augmentation in walking speed (Goldberg, Kautz, and Neptune, 2008).

Indeed, walking speed must be considered when interpreting changes in joint moments. Both plantarflexor and hip extension moments during stance were found to be strongly associated with walking speed in individuals post-stroke (Nadeau, Gravel, Arsenault, and Bourbonnais, 1999; Allen, Kautz, and Neptune, 2014) and healthy controls (Goldberg, Kautz, and Neptune, 2008) in line with their contribution to the forward propulsion of the body (Kepple, Siegel, and Stanhope, 1997; Zajac, Neptune, and Kautz, 2003; Winter, 2009). The after-effects in the study by Lauzière et al. (Lauzière et al, 2014a) were obtained during walking at controlled, comfortable speed imposed by the treadmill. Comfortable speed was determined prior to the split-belt configuration ('adaptation') and used again after 'adaptation'. In the present study, the aim was to investigate the effects of repeated SBT walking during walking over ground, thus participants' "natural" and functional walking environment. Therefore, participants were asked to walk at their comfortable speed in order to obtain their most natural gait pattern. Comfortable speed was higher at one-day (+ 0.11 m/s) and one-month post-training (+ 0.16 m/s) when compared to pre-evaluation.

However, the increase in hip extension and plantarflexion moments was more pronounced on the fast side. The 'asymmetrical' increase in these joint moments suggests a potential contribution to the changes in SL symmetry and not only an association with the increase in walking speed. The contribution to changes in SL symmetry however appears not to be associated with an improvement in joint moment symmetry as the indices in joint moments remained unchanged (Table 5-7). This could be explained by the analysis of fast and slow moments instead of distinguishing by paretic and non-paretic sides. Since two training groups were tested (NP- and P-fast), mean slow moments consisted of paretic and non-paretic joint moments. For example, peak plantarflexion moments were smaller on the paretic side compared to the non-paretic for all participants as illustrated by the representative peak net plantarflexion moments (Figure 5-21). Thus, in the NP-fast group, the NP net joint moment increased which led to an increase in asymmetry of plantarflexion moments. This is in contrast to the improvements in the symmetry of plantarflexion moments in the P-fast group. In the present study, the aim was to identify the effects of slow- and fast-belt walking. However, future studies could emphasize the analysis of a larger sample in each training group to test whether the two training groups present different biomechanical and muscular changes to reduce SL symmetry. Furthermore, six sessions of SBT walking did not lead to significant changes in joint biomechanics and muscle activity at post-evaluation. The changes reached significance at follow-up evaluation. Additional training between one-day and one-month post-training can be excluded as a factor since participants reported that they did not undertake any additional gait training. Yet, an increase in practicing walking without the purpose of training could be a potential factor. Participants reported as informal feedback that they had more "confidence in walking", "walked more", felt that they could walk faster and use "more consciously" their more-affected (paretic) side. Therefore, a larger number of training sessions should be investigated in future studies using biomechanical gait analysis.

5.3.6.1.3 *Changes in Muscle Activity*

Six sessions of SBT training did not induce consistent changes in lower limb muscle activity in the present group of individuals with chronic stroke. The small sample size – based on the study's pilot nature – requires the consideration of a type II error (Sullivan and Feinn, 2012) by concluding that no effect was found when there actually was one. Therefore, effect size calculation and CIs were reported (Table 5-6 and 5-7). The lack of effect is supported by the small effect sizes (range 0.15 - 0.32) found generally for the changes in muscle activity from pre- to post- and pre- to follow-up. Based on this statistical information, one can most likely assume that the SBT training did not induce reasonable and consistent effects on muscle activity in the present group of individuals post-stroke. However, because of the pilot nature of this study with the small sample size, conclusions about the generalization of these findings should be made with care.

A cross-sectional analysis of SBT-induced immediate after-effects on EMG activity (%max RMS) revealed consistent changes in individuals post-stroke ($n = 16$) with different initial asymmetries and lower limb deficits (Betschart et al, [under revision]). As for Lauzière et al. (2014a), Betschart et al. [under revision] analyzed EMG activity during two conditions of SBT walking. Meaning, during the first condition, the non-paretic side walked on the fast belt and vice versa for the second condition. When belts were set at equal speeds again, group mean GL activity on the side that was walking on the slower belt was higher compared to baseline and the TA and VL activity which were higher after fast-belt walking. This was found during both conditions. Additional increases in RF and ST were found when the paretic leg walked on the fast belt. The changes in GL, TA, and ST activity correlated with the changes in SL ($0.522 \leq r < 0.664$).

It is important to consider that the signals in the Betschart et al. [under revision] study were obtained immediately after split-belt configuration and not one day later as in the present study. Furthermore, participants were still walking on the SBT at a comfortable speed on both belts whereas in the present study EMG was analyzed during walking over ground. Previous studies found similar EMG activation of lower limb muscles during (unsupported) treadmill walking when compared to overground gait (Hesse, Konrad, and Uhlenbrock, 1999; Kautz, Bowden, Clark, and Neptune, 2011). Based on such findings it has been assumed that both walking conditions require similar motor control strategies for individuals post-stroke (Kautz et al, 2011). This suggests that the significant and consistent increases in muscle activity found immediately after SBT walking (Betschart et al, [under revision]) are probably due to the SBT-induced after-effects on SL and the controlled speed of the treadmill. Yet, considering the modification of SL symmetry, the contribution of muscle activity during walking over ground could be different compared to treadmill walking. During walking over ground, the limb and trunk motions must be controlled by the individual who also must navigate in a more open space, in contrast to walking on a treadmill with belt-induced speeds. This leads to the assumption that during treadmill (or SBT) walking the task is more constrained compared to walking over ground. Therefore, it seems reasonable to consider that walking over ground leaves more potential for the use of compensatory strategies and eventually requires more attention for individuals post-stroke secondary to deficits in the integration of sensory information (Lamontagne, Fung, McFadyen, and Faubert, 2007; Lamontagne et al, 2010) as assumed by Savin et al. (Savin, Morton, and Whitall, 2014).

The influence of a treadmill belt motion on an individual's gait pattern was brought in line with the finding that stance time symmetry was higher during walking on a treadmill compared to over ground at equal speeds (Hesse et al, 1999; Harris-Love, et al, 2001). These authors hypothesized that the treadmill belt "forces" a timely swing of one limb via the belt-induced posterior movement of the contralateral limb (Harris-Love, et al, 2001). Such an influence of belt movement on the gait pattern of individuals post-stroke might be possible. Yet, its association with muscle activity remains questionable. For example, a general increase of lower limb muscle activity was observed during both, treadmill-induced augmentation of walking speed (Hesse et al, 2001) and self-induced augmentation during walking over ground (Lamontagne and Fung, 2004). In contrast, the analysis of power production and work (J/kg) revealed substantial variability in strategies to the increase of walking speed (Jonkers, Delp, and Patten, 2009; Jonsdottir et al, 2009). Indeed, in the present study, the analysis of individual data revealed a substantial variation among participants in terms of direction and combination of muscles changing EMG activity. For example, some individuals increased GL and ST activity, others more VL and TA activity, on the fast side whereas some showed more pronounced changes on the slow side. However, from a more clinical point of view, such heterogeneity of effect on muscle activity post-stroke after training intervention is not that surprising. Changes in EMG pattern (signal duration and amplitude) were not necessarily associated with changes in gait pattern (Buurke et al, 2008) or gait ability (Den Otter, Geurts, Mulder, and Duysens, 2006) over time. In healthy individuals, four to five different synergies – based on the activation profile – were found in healthy participants (Ivanenko, Poppele, and Lacquaniti, 2004; Clark et al, 2010). Therefore, even in healthy individuals, where SL is considered symmetrical (Patterson et al, 2010), different synergies of muscle groups were found to achieve walking tasks. This indicates that during a specific task, the contribution of muscle activity varies among individuals, depending on the different control strategies required for the task. This could be the case in a population with sensorimotor deficits secondary to a stroke as our group. Nevertheless, regarding the statistics on joint moments presented in this paper, the plantarflexors showed most consistent changes indicating a relevant role for SL changes during walking over ground. Thus, as previously suggested, a larger sample size could test this assumption.

5.3.7 Study Limitations

In the present study, walking speed during biomechanical evaluation was not controlled (standardized) to correspond to the speed at pre-evaluation. However, this would probably have altered participants' natural gait patterns (Liu et al, 2014) and biased the conclusion about effects of SBT training on outcome parameters during walking at participants' individual comfortable speeds over ground.

The method used for EMG amplitude normalization could be a potential contributor to the lack of consistent changes in muscle activity. EMG activity was normalized using a peak reference value from the same day (Methods section). It is likely that when the actual EMG signal is higher, it this is expressed in both, mean and maximal values. This would reduce the chance of detecting the actual increase of activity compared to the other two evaluation sessions. Despite this limit in the interpretation, the normalization of EMG activity is highly recommended when comparing between individuals and different session (Winter, 1991; Konrad, 2005).

In addition, this pilot study revealed that a sample size of ten participants resulted in moderate to small effect sizes for EMG data. The heterogeneity of the training-induced EMG changes and the moderate to small effect size do not allow recommendations considering subgroup analysis. Nevertheless, we suggest a distinct analysis of each training group (P-fast and NP-fast) with larger sample size, respectively. This will enable studying the effects on paretic and non-paretic muscle activity and joint mechanics.

5.3.8 Conclusion

Six sessions of an error-augmentation-based SBT training improved SL symmetry in ten participants. The biomechanical changes in combination with such improvements in SL symmetry were found, particularly on the side trained on the faster belt. The analysis of muscle activity was not conclusive and showed large variability in changes among the ten participants. Thus, the present pilot data suggests that SL symmetry is achieved by a variety of strategies in biomechanical and muscular changes in our group of stroke survivors. However, these findings indicate that the leg trained on the faster belt and the plantarflexors tended to be the main contributor to a reduction in SL asymmetry. Studies with larger samples and a higher number of training sessions are required to test the present assumptions. Based on the variability of EMG data, no particular subgroup analysis can be suggested

5.3.9 Conflict of Interest

There are no conflicts of interest to disclose.

5.3.10 Funding

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Center for their important contribution, the summer student for her help during data collection and analysis, and the engineers from the Pathokinesiology Laboratory for their technical support.

TABLE 5-5. Participants' Demographics and Stroke Characteristics.

Participant	**Age** (years)	**Time post-stroke** (months)	**Paretic side**	**Side of shorter step**	**Initial asymmetry** (ratio L/S)	**Initial gait speed*** (comf./fast)	**Chedoke Foot/Leg** (/7)	**CSI** (/16)
P1	53	15	L	NP	1.78	0.59/0.75	1/6	5
P2**	58	8	R	P	1.45	0.92/1.39	5/6	7
P3	73	9	R	NP	1.45	0.58/0.85	2/6	4
P4	49	6	L	NP	1.42	0.72/1.04	2/6	4
P5	39	40	L	NP	1.31	1.21/1.65	5/6	5
P6**,[f]	50	21	L	P	1.27	0.63/0.76	1/3	7
P7	52	12	R	NP	1.22	0.62/0.96	3/5	5
P8**	43	88	L	P	1.19	0.58/0.93	1/4	5
P9	49	19	L	NP	1.15	1.01/1.64	6/7	5
P10**, [f]	58	14	L	P	1.10	1.06/1.36	5/7	5
Mean / median	**52.4**	**23.2**	**7L/3R**	**6NP/4P**	**1.33**	**0.79/1.13**	**2.5/6**	**5**
SD	9.3	24.7			0.20	0.24/0.35	2.0 /1.3	1.0

*Initial gait speed was obtained from the 10MWT; m/s = meter/second; comf. = comfortable; CSI= Composed spasticity index: 1 - 5 (normal), 6 - 9 (mild spasticity), 10 - 12 (moderate spasticity), 13-16 (severe spasticity) (Levin & Hui-Chan, 1993); **paretic-fast group; [f]= female participant; ratio L/S (see equation); SD = standard deviation; L= left; R= right; NP= non-paretic; P= paretic.

TABLE 5-6. Kinetic and Kinematic Parameters at Pre-, Post- and Follow-up Evaluation (n=10).

	Pre-evaluation	Post-evaluation	Follow-up evaluation	*Post-hoc analysis* Post-Base	FU-Base
	Mean (SD) (CI 95%)			*p*-value (effect size) [95% CI for mean differences]	
Spatiotemporal parameters					
Symmetry					
(L/S)*	1.28 (0.16)	1.14 (0.09)	1.11 (0.08)	**0.002 (1.12)**	**0.013 (1.35)**
	(1.08-1.78)	(1.02-1.34)	(1.04-1.28)	[0.06, 0.22]	[0.04, 0.30]
Step length					
fast	46.2 (9.2)	50.3 (9.0)	53.7 (10.0)	0.103 (0.44)	**0.038 (1.42)**
(cm) [a]	(36.3-66.0)	(38.2-69.3)	(42.7-76.0)	[-9.12, 0.77]	[-14.69, -0.44]
Step length					
slow	53.5 (12.7)	58.4 (11.0)	58.8 (12.6)	0.157 (0.41)	0.227 (0.40)
(cm)	(40.9-82.2)	(45.4-81.9)	(44.6-87.2)	[-11.40, 1.60]	[-13.60, 3.00]
Gait speed					
comf.	0.71 (0.18)	0.82 (0.18)	0.87 (0.19)	0.074 (0.59)	**0.009 (0.80)**
(m/s)	(0.43-1.08)	(0.64-1.12)	(0.38-1.23)	[-0.24, 0.01]	[-0.28, -0.05]
Kinetic parameters: joint moments (Nm/kg)					
Plantarflexion (20-60%)					
Fast	1.22 (0.26)	1.31 (0.26)	1.37 (0.23)	0.082 (0.32)	**0.016 (0.60)**
	(0.86-1.79)	(0.98-1.84)	(1.05-1.79)	[-0.19, 0.01]	[-0.28, -0.03]
Slow	1.06 (0.17)	1.07 (0.18)	1.16 (0.10)	1.000 (0.04)	0.131 (0.71)
	(0.76-1.24)	(0.76-1.25)	(1.01-1.26)	[-0.09, 0.07]	[-0.23, 0.03]
Knee flexion (20-60%) [b]					
Fast	-0.30 (0.21)	-0.39 (0.21)	-0.40(0.27)	**0.002 (0.41)**	0.080 (0.39)
	(-0.82-0.03)	(-0.87-0.17)	(-1.05-[-0.15])	[0.04;0.14]	[-0.01;0.21]
Slow	-0.16 (0.15)	-0.15 (0.16)	-0.24 (0.20)	1.000 (0.06)	0.571 (0.45)
	(-0.37-0.06)	(-0.40-0.06)	(-0.53-0.05)	[-0.09, 0.07]	[-0.09, 0.26]
Knee extension (0-50%) [c]					
Fast	0.26 (0.21)	0.28 (0.21)	0.33 (0.29)	1.000 (<0.11)	1.000 (<0.27)
Slow	0.39 (0.21)	0.41 (0.24)	0.40 (0.27)	[-0.12, 0.08]	[-0.20, 0.12]
Hip extension (0-20%) [a]					
Fast	0.47 (0.29)	0.60 (0.29)	0.65 (0.31)	0.249 (0.42)	**0.037 (0.58)**
	(0.18-0.77)	(0.14-1.12)	(0.13-0.94)	(-0.33, 0.07)	[-0.36, -0.01)
Slow	0.41 (0.22)	0.38 (0.17)	0.58 (0.19)	1.000 (0.13)	0.084 (0.78)
	(0.09-0.74)	(0.15-0.63)	(0.35-0.94)	[-0.14, 0.19]	[-0.36, 0.02]
Hip Flexion (20-80%) [a,c]					
Fast	-0.53 (0.09)	-0.52 (0.06)	-0.59 (0.10)	1.000 (0.54)	1.000 (0.11)
Slow	-0.48 (0.06)	-0.57 (0.05)	-0.46 (0.06)	[-0.25, 0.24]	[-0.13, 0.25]

*Including both data obtained from the 3D-motion analysis and data from the clinical evaluation; **obtained from 3D-motion analysis; (%) = percentage of gait cycle; [a] = interaction between sides; [b] = side effect; [c] = no main effect; therefore, post-hoc *p*-value, effect sizes and 95% CI are reported considering the total means (both sides). Abbreviations: L/S ratio (equation symmetry ratio); SD = standard deviation; comf. = comfortable walking speed; m = meters; s = second; cm = centimeter; Fast and slow = refers to the side trained on the faster and slower belt, respectively. Significance level was set at $p \leq 0.05$ and adjusted with a Bonferroni correction and is denoted by bold-written *p*-values and effect sizes. Effect size scores: 0.20 = small; 0.50 = medium; 0.80= large (Thalheimer et al., 2002).

TABLE 5-7. Indices of Peak Net Moments at Pre-, Post- and Follow-up Evaluation (n=10).

	Pre-evaluation	Post-evaluation	Follow-up evaluation	*Post-hoc analysis* Post-Base	FU-Base
		Mean (SD) (range)		*p*-value (mean difference) [95% CI for mean differences]	
Indices: fast / (fast+slow); perfect symmetry = 0.50					
Plantaflexion	0.52 (0.08) (0.42-0.63)	0.54 (0.07) (0.45-0.63)	0.53 (0.05) (0.47-0.59)	0.080 (-0.02) [-0.04, 0.00]	1.000 (-0.01) [0.05, 0.03]
Knee flexion	0.75 (0.38) (0.30-1.55)	0.79 (0.26) (0.52-1.20)	0.71 (0.34) (0.27-1.44)	1.000 (-0.04) [-0.23, 0.15]	1.000 (0.04) [-0.19, 0.27]
Hip extension	0.55 (0.17) (0.27-0.74)	0.59 (0.20) (0.28-0.81)	0.50 (0.19) (0.18-0.67)	0.694 (-0.04) [-0.14, 0.06]	0.541 (0.04) [-0.05, 0.13]

Illustrated are indices of joint moments which presented significant changes from pre- to post-evaluation and pre- to follow-up evaluation. SD = standard deviation; CI = confidence interval. Significance level was set at $p \leq 0.05$ and adjusted with a Bonferroni correction; denoted are *p*-values, mean differences with its 95% CI. The ranges illustrate that some participants presented higher values on the slow side and other higher values on the fast side, which explains the symmetry of the group mean values. The 95% CI of the differences range from negative to positive which explains the lack of significant changes.

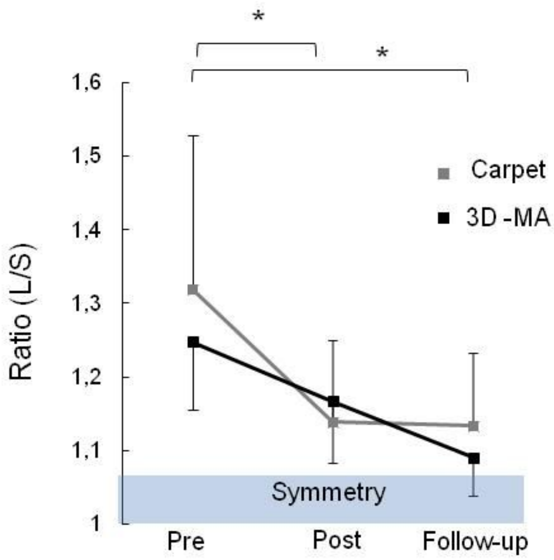

Figure 5-17. Illustration of step length ratio (longer/shorter) obtained by 3D-motion analysis (3D-MA) in the laboratory (black) and by walking over a paper carpet (grey) during pre-, post- and follow-up evaluation. The grey bar between 1 and 1.08 on the y-axis represents the range at which the ratio is considered symmetrical (Patterson et al, 2010). *denotes significant differences between evaluations ($p \leq 0.05$) for both, carpet and laboratory data. The comparison of the ratios obtained during the 3D-motion analysis and walking over a paper carpet did not show any significant differences ($p = 0.470$).

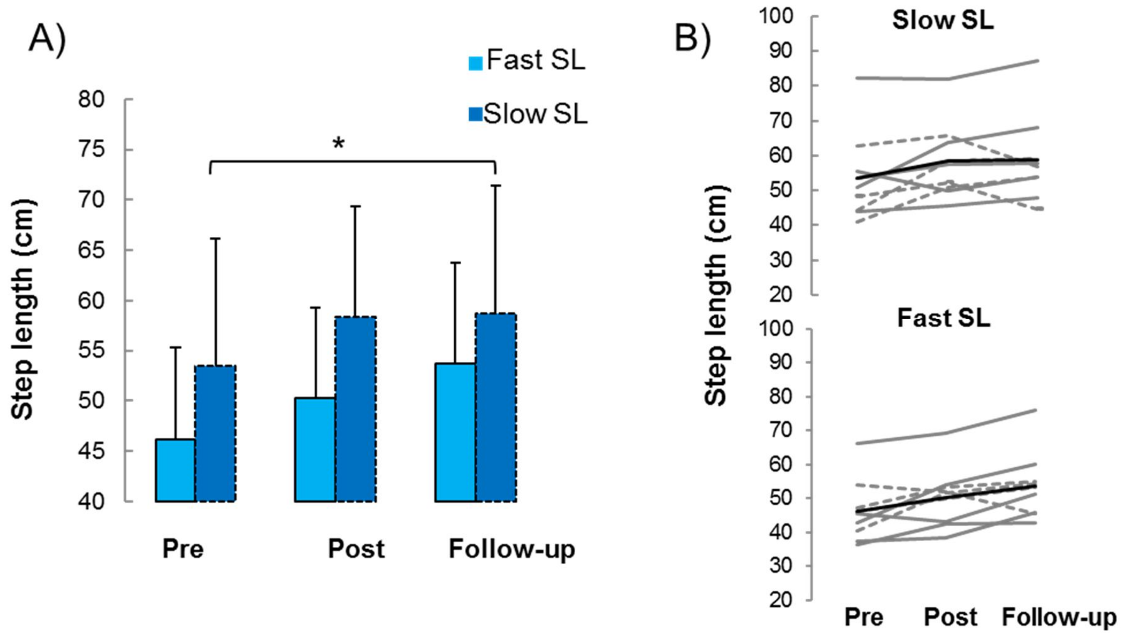

Figure 5-18. (A) Fast (pale blue) and slow (dark blue) step lengths (SL) during pre-, post and follow-up evaluation. B) Individual slow and fast step lengths. Group mean is illustrated in black. Dotted lines represent step lengths for participants trained in the P-fast group. *indicates significant changes from post-hoc analysis ($p \leq 0.05$).

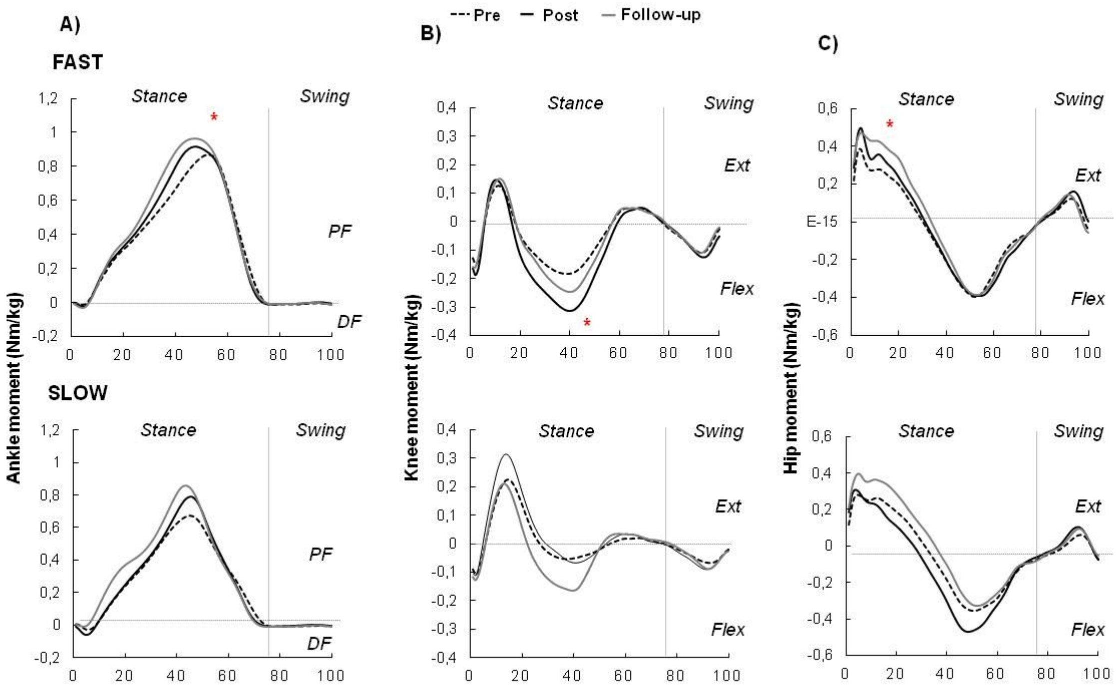

Figure 5-19. Fast (top) and slow (bottom) net joint moments for ankle (A), knee (B) and hip joints (C) normalized for cycle duration (0-100%; x-axes). Illustrated are group net moments during pre- (dotted line), post- (black) and follow-up (grey) evaluations. * indicates statistical significance after post-hoc analysis with Bonferroni correction ($p \leq 0.05$). Abbreviations: PF = Plantarflexion, DF = Dorsiflexion, Flex = Flexion, Ext = Extension.

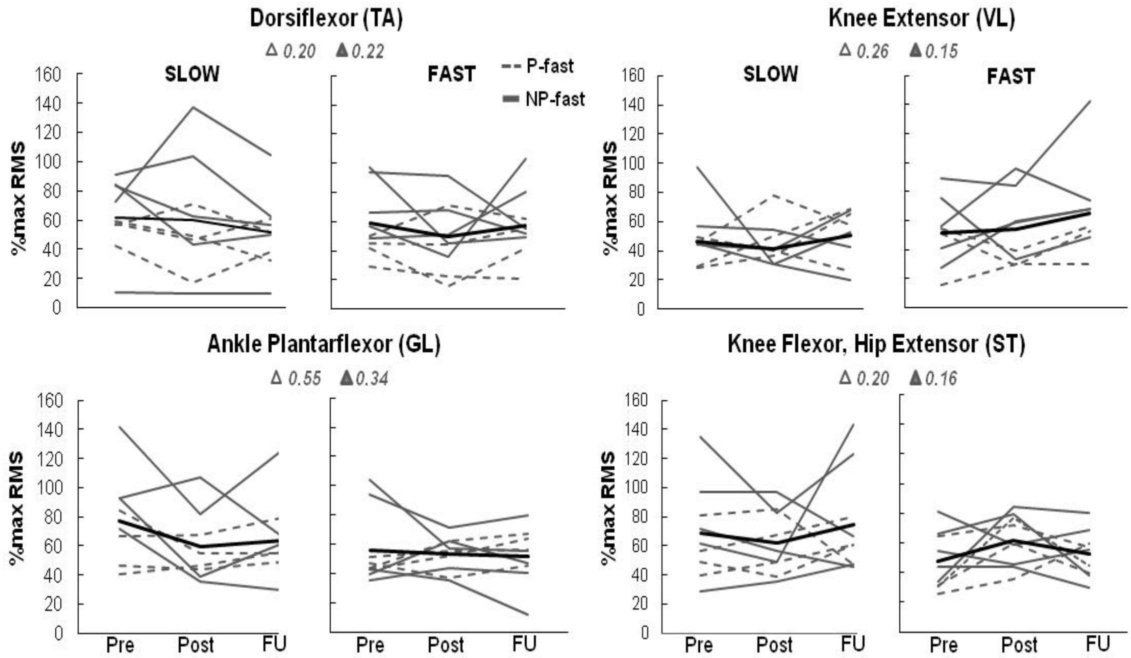

Figure 5-20. Individual EMG activity (%max RMS) during pre-, post- and follow-up (FU) evaluations. EMG values are presented in correspondence to the slow and fast sides with group means (black) and individual means (grey) for paretic-fast trained (dotted) and non-paretic fast trained (continuous) limbs. Effect sizes (Hegdes g_{av}) are reported for the changes pre- to post- [Δ] and pre- to follow-up [▲] evaluation considering mean values from both sides. Abbreviations: TA = Tibialis anterior, GL = Gastrocnemius lateralis, VL = Vastus Lateralis, ST = semitendinosus. *The rectus femoris muscles are not illustrated since data from 6 participants were missing for certain evaluation sessions.

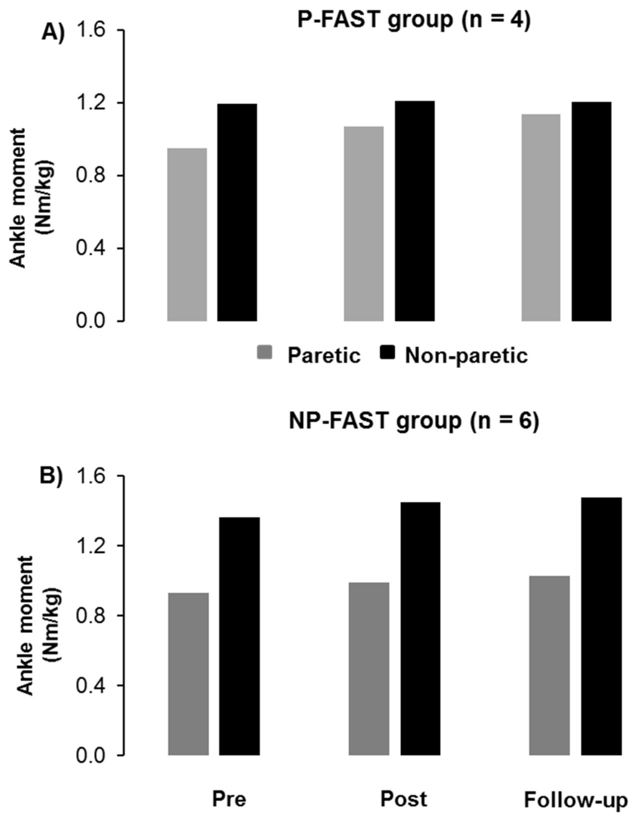

Figure 5-21. Group mean peak plantarflexion moments of the participants trained in the P-fast group. (A) and NP-fast group (B). Peak moments are illustrated for the paretic (grey) and non-paretic (black) side.

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Chapter 6. General Discussion

The aim of the present thesis was to investigate lower limb muscle activity and joint kinetics in combination with SBT-induced modification of step length and step length symmetry in individuals post-stroke. The immediate after-effects of SBT walking on these parameters were tested during a single exposure to an error-augmentation-based SBT protocol within a cross-sectional analysis. In a study project, the effects of repeated exposure to such a protocol on gait ability and biomechanics were assessed in twelve individuals post-stroke forming the pilot group.

The interest behind these studies was driven in part by the novel findings by Reisman et al. (2005; 2007). This group showed that individuals post-stroke have the capacity not only to adapt, but also to store the adapted step length after walking on a SBT with unequal belt speeds. Prior to these studies, groups of researchers used more conventional approaches to improve step length symmetry in individuals post-stroke, such as different types of treadmill walking (Combs et al., 2013; Hassid et al., 1997) and with auditory feedback (Cha et al., 2014; Schauer & Mauritz, 2003; Thaut et al., 2007). These interventions were helpful in modifying temporal aspects of the gait pattern post-stroke, but without any reported changes of step length symmetry. While SBT protocols showed promising results in short- and long-term changes of step length asymmetry post-stroke, it was relevant to assess the lower limb muscle activity to better understand the improvements in step length asymmetry.

This doctoral work provides the first exploration of SBT-induced changes in lower limb muscle activity in a group of individuals with sensorimotor and gait deficits secondary to a cerebral stroke. Furthermore, this work presents first findings on the feasibility of a SBT training protocol in a clinical setting. The following sections discuss the main findings of the three papers along with supplementary results. The thesis hypotheses, study limitations, relevance and suggestions for future research will also be discussed.

6.1 Principal Findings

The first paper (paper #1) demonstrated that six minutes of SBT walking induced quite consistent after-effects in group mean muscle activity among 16 individuals post-stroke with different initial step length asymmetries and lower limb deficits. Both the NP-fast and P-fast walking conditions led to a short-term increase in muscle activity of the slow plantarflexor (GL), fast dorsiflexor (TA), and fast knee extensor muscles (VL) during early post-adaptation. With regards to step length, previous findings on SBT-induced after-effects were reproduced (Malone & Bastian, 2013; Reisman et al., 2007). Changes in EMG activity with after-effects were observed in combination with a decrease in slow step length and an increase in fast step length when compared to baseline. The fast TA and slow GL presented moderate to strong correlations with the changes in step length. This observation supports the hypotheses stating that distal lower limb muscles will present the most pronounced changes during the post-adaptation period and will be associated with changes in step length after-effects. However, as far as the period of adaptation is concerned, only some muscles, which had changed their EMG activity during the early period of adaptation, subsequently had de-adapted their muscle activity during the late adaptation period. As such, the expected pattern of adaptation in EMG activity can only be partially confirmed.

Paper #2 demonstrated that six sessions of SBT walking were sufficient to significantly reduce step length asymmetry during walking over ground in a group of 12 participants. Concomitant effects of the training were found on walking speed and functional capacity whereas walking endurance did not change from pre- to post-training. Thus, the hypothesis that the new error-augmentation-based SBT training would lead to improvements in all four gait parameters (step length symmetry, walking speed, endurance and functional capacity) was partially confirmed. Paper #3 demonstrated that a reduction in step length asymmetry was combined with changes in net joint moments, particularly on the fast side. A less consistent relationship was found between changes in step length asymmetry and changes in muscle activity which did not differ significantly from pre-training to one-day and one-month post-training. However, the significant increase in fast net plantarflexion along with the large effect size estimation for the changes in fast plantarflexor EMG activity support the hypothesis that the plantarflexors play an important major role in adjustments of step length symmetry. However, the hypothesis that the use of the plantarflexor on the side trained on the slow belt would increase was not supported by the present findings.

6.1.1 Participants

Two groups of individuals with chronic stroke were included in this doctoral work. For the cross-sectional analysis (paper #1), a group of 16 individuals post-stroke was assessed and another group of 12 participants was trained with the SBT protocol (papers #2 and #3). Two participants analyzed for the cross-sectional analysis also participated in the training study for papers #2 and #3. These were the only individuals already familiar with SBT walking. No particular differences in outcome measures obtained from these two participants were found when compared with the 10 other participants of the training study. Overall, clinical data showed that both groups displayed similar characteristics in terms of age, time post-stroke (chronic) and gait performance and comparable ranges in sensorimotor deficits to those evaluated in previous gait studies (Dyer et al., 2014; Lamontagne & Fung, 2004; Lamontagne et al., 2002; Milot et al., 2007; Olney, Griffin, & McBride, 1994) as well as investigations on SBT-induced locomotor adaptation (Malone & Bastian, 2013; Reisman et al., 2013; Reisman et al., 2007).

Our participants walked without technical or personal aid for a short distance (15 m) indoors and their comfortable speed over ground ranged from 0.51 - 1.40 m/s similar to the range of the walking speeds reported in previous SBT studies (Lauzière et al., 2014a; Malone & Bastian, 2013; Reisman et al., 2013; Reisman et al., 2007). Lower limb sensorimotor deficits varied among participants as shown by individual data in Table 5-1 and Table 5-3 with most of the participants displaying moderate to light deficits similar to those reported in previous EMG and biomechanical gait studies (Den Otter et al., 2007; Dyer et al., 2014; Lamontagne et al., 2002; Milot et al., 2007) and locomotor adaptation (Malone & Bastian, 2013; Reisman et al., 2007). The presence of ankle spasticity was very limited in both study groups. In each study, only two participants displayed augmented resistance to passive stretch in plantarflexion (Ashworth scale). Signs of ankle clonus were observed only in three participants of the cross-sectional study and classified as mild (not presented in the results). The small proportion of participants with moderate signs or absence of spasticity is similar in proportion to what has been reported in previous gait studies (Dyer et al., 2011; Hsu et al., 2003; Milot et al., 2007).

With regards to average step length asymmetry, the two study groups showed slight differences, while the training group displayed higher asymmetry values. This is not surprising considering the differences in inclusion criteria concerning step length symmetry. In the training study, participants had to display step length asymmetry whereas no such criterion was specified in the cross-sectional study. As far as the participants in the cross-sectional study are concerned, initial step length asymmetry ratios ranged from 0.64 - 1.15 with a mean of 0.94 (SD 0.15) which is close to symmetry according to Patterson et al. (2010a). The step length asymmetries of the participants from the training study are presented in Table 5-3 of paper #2. With symmetry ratios ranging from 1.10 - 2.05, all participants were asymmetrical based on the definition of Patterson et al. (2010a). To enable a comparison of the symmetry ranges of our participants with the participants tested in the previous SBT training study (Reisman et al., 2013), the normalized step length difference (NSLD) was used. The NSLD in the current training study group ranged from 0.046 - 0.344 which corresponds to the average initial NSLD calculated in the participants from the Reisman group (NSLD: 0.052 - 0.310) (Reisman et al., 2013).

Given that 16 of our participants were included in the 20 participants analyzed by Lauzière et al. (2015; 2014a; 2016) who investigated the effects of SBT walking on net joint moments and muscular effort, gait characteristics and sensorimotor deficits were similar across their studies and those included in this thesis. As a consequence, the results of our cross-sectional study add to those reported previously by Lauzière and colleagues. Overall, these similarities in deficits and gait parameters found in our study participants when compared to previous literature emphasize the comparability of the results of our studies with the investigations cited in the following discussion.

6.2 SBT-induced Changes in Outcome Parameters

This section provides an extended discussion of the results obtained from the cross-sectional and training study that goes beyond what was discussed within each respective paper. Overall, the work presented in the current thesis provided a first investigation of the changes in lower limb muscle activity involved in SBT-induced step length after-effects as well as in reduced step length asymmetry after repeated exposure to such SBT protocol. Our results generally reproduced previous changes in step length post-stroke obtained with similar SBT protocols (Malone & Bastian, 2013; Reisman et al., 2013; Reisman et al., 2007). However, the protocol investigated in the present training study appears to be advantageous compared to the one used in a previous study (Reisman et al., 2013) since it resulted not only a reduction of step length asymmetry but also in improvements of walking speed as discussed in paper #2. The observation of after-effects indicates that a cerebral lesion did not prevent our participants post-stroke from adapting their gait pattern to a new walking condition (i.e., split-belt configuration) and that these changes were stored when exposed to the initial walking condition (i.e., tied-belt configuration). Long-term changes on step length symmetry observed in our training study suggest that the repeated exposure to locomotor adaptation led to motor learning. These findings are not surprising according to the internal model, the concept used in the scope of this thesis, which suggests that cerebellar circuits in the central nervous system are mainly responsible for regulating locomotor adaptation and the storage of the adapted locomotor output as decribed in the literature review section.

Nevertheless, the two walking conditions (NP-fast and P-fast) in our cross-sectional study induced different effects on step length and muscle activity in our post-stroke participants. This latter aspect will be discussed in the following sections.

6.2.1 SBT-induced Effects on Step Length in Individuals Post-stroke

Overall, the cross-sectional study demonstrated that for both NP-fast and P-fast conditions, six minutes of walking with the split-belt configuration resulted in consistent after-effects in group mean step length during early post-adaptation. These after-effects were characterized by an increase in fast step length and a decrease in slow step length. However, differences in the amplitude of after-effects were found between the two walking conditions (NP-fast and P-fast). Such differences were not reported in previous papers assessing both walking conditions (Malone & Bastian, 2013; Reisman et al., 2007). These previous studies were particularly interested in the spatiotemporal changes underlying the SBT-induced disruption of the gait pattern. These authors sought to explain the spatiotemporal changes involved in locomotor adaptation and after-effects by exploring changes in symmetry of inter- and intralimb parameters. For example, in both the NP-fast and P-fast walking conditions, the switch from tied- to split-belt configuration led to an initial asymmetry (error) of gait parameters which was adapted towards symmetry in interlimb, but not intralimb parameters (Reisman et al., 2005; Reisman et al., 2007) (section 2.6). In the context of our cross-sectional analysis we aimed to investigate which lower limb muscles could be involved in the decrease of left and the increase of right step length, respectively, in order to obtain a more profound understanding of the strategies involved in producing these after-effects.

The quantification of the step length measure in our study demonstrated that during the early post-adaptation of the NP-fast condition, step lengths from both sides (paretic and non-paretic) differed (i.e., increased or decreased) significantly from their baseline values. In contrast, after the paretic leg walked on the faster belt (P-fast condition), only the changes in non-paretic step length reached significance (Figure 5-1 A-B). Given that the P-fast condition was always conducted following the NP-fast condition, one can assume that the smaller amplitude in after-effects observed during the P-fast condition could be the consequence of a general fatigue or a carry-over from the effects of the NP-fast condition. However, as discussed in paper #1 in conjunction with EMG results, participants rested between the conditions until they felt recovered such that fatigue as a causal explanation can be rejected.

Interestingly, healthy controls also showed significant after-effects only on the slow side (Figure 5-7). Given that our stroke group presented bilateral after-effects during the NP-fast condition, it is assumed that during the P-fast condition the limited amplitude of change in paretic step length was due to stroke-induced neuromuscular deficits limiting the task performance of the paretic side. However, this interpretation must be handled with care since lower limb impairments (e.g., touch sensitivity, CMSA scores, Ashworth score) were not correlated with step length after-effects as observed in the present study (not presented in paper #1) as well as previous investigations (Reisman et al., 2007). On the other hand, the interpretation of neuromuscular deficit effects can be supported by previous investigations that showed participants with more severe sensorimotor deficits on the paretic side showed larger impairments in spatiotemporal gait parameters (Patterson et al., 2008a) and lower performance in short-term motor learning (Dancause, Ptito, & Levin, 2002). The analysis of 36 chronic stroke survivors revealed that those with lower scores on the CMSA (foot and leg) on the paretic side, displayed more severe impairments in spatiotemporal parameters (Patterson et al., 2008a). This indicates larger deficits in the control of paretic limb movements, supported by Dancause et al, (2002) on the capacity of short-term motor learning in individuals post-stroke. Ten individuals with chronic stroke displayed a limited capacity to correct for force-induced movement errors during elbow extension and flexion when compared to healthy controls. The lower the scores on the CMSA (upper limb) of the paretic limb, the greater the inconsistencies of the strategies used to correct the movement and the larger the movement deviations when compared to healthy controls.

Further, stroke-induced changes in the morphology of the skeletal muscle could be a possible explanation for the observed after-effects in paretic step length in our cross-sectional analysis which were reduced in amplitude particularly after fast belt walking (P-fast condition) compared to slow belt walking (NP-fast condition). De Deyne et al. (2004) found that the number of fast-twitch fibers was significantly elevated in the paretic quadriceps muscle (46 - 88% of all fibers) when compared to the non-paretic side (32 - 72%) (De Deyne et al., 2004). Interestingly, an increased number of fast-twitch fibers showed a strong negative correlation with walking speed (r = -0.78). Based on these findings, De Deyne et al. concluded that a cerebral stroke can induce biological changes in muscle phenotypes with potentially negative functional consequences.

Above all, changes in muscle activity found in our cross-sectional analysis support the interpretation that the paretic side has more difficulty in adjusting its step length in the P-fast condition compared to slow belt walking or healthy controls. Our results showed a more pronounced contribution of muscle activity during the P-fast condition when compared to the NP-fast condition (Figures 5-2 and 5-3) or healthy controls (Figure 5-8). Next to the slow (non-paretic) GL, fast (paretic) TA and fast VL that showed after-effects during both conditions, additional increases of EMG activity were found in the fast RF and the slow ST muscles during the P-fast condition that was correlated with slow step length changes. As elaborated in the discussion section of paper #1, these additional changes were hypothesized to compensate for the insufficient contribution of non-paretic plantarflexors such that an increase in paretic step length could occur. Step length after-effects in healthy individuals were combined with a significant increase in activity of slow and fast TA muscles and the fast RF, albeit only approaching significance for the RF muscle ($p = 0.065$) (section 5.1.13.2). Paper #1 discussed the additional bilateral contribution of proximal muscles during P-fast walking by means of biomechanical factors. Nevertheless, deficits in neuromuscular control should be taken into account as contributing factors to the augmented bilateral contribution of proximal muscle considering that such deficits were frequently reported as a consequence secondary to a stroke. Their possible implication will be discussed in the following section.

6.2.2 Lower limb Impairments and After-Effects on EMG Activity

Considering that adaptation of the gait pattern during SBT walking involves changes in the symmetry of inter-limb parameters (Malone et al., 2012; Reisman, Bastian, & Morton, 2010; Reisman et al., 2005; Reisman et al., 2007), it is reasonable to assume that this adaptation involves circuits that control interlimb motor and sensory information. For example, crossed-spinal reflex pathways are considered as relevant contributors to motor control for dynamic stability between limb segments and propulsive acceleration of the body during unperturbed gait in healthy individuals (Gervasio et al., 2015; Hanna-Boutros et al., 2014; Stubbs & Mrachacz-Kersting, 2009). However, evidence about the influence of stroke-affected crossed-spinal pathways on muscle activity post-stroke is very small (Stubbs et al., 2012; Zehr, 2005). Both studies showed that crossed-spinal reflexes persist post-stroke, but were less inhibited in individuals with chronic stroke compared to healthy controls. According to Stubbs et al. (Stubbs & Mrachacz-Kersting, 2009), this impaired reflex regulation between limbs could limit the ability to adjust the gait pattern quickly after mechanical disturbance of the ipsilateral limb. However, these pathways were studied particularly in distal lower limb muscles (Gervasio et al., 2015; Hanna-Boutros et al., 2014; Stubbs & Mrachacz-Kersting, 2009; Stubbs et al., 2012; Zehr, 2005). Indeed, findings on crossed-spinal pathways linking proximal muscles have been very recently investigated by Stevenson et al (2015) who found that knee extension in sitting position induces an inhibitory reflex response of the contralateral BF muscle in healthy individuals. Considering such limited knowledge on proximal crossed-spinal pathways, the interpretation of their potential influence on the additional increase in proximal muscles in the P-fast condition must be handled with care.

Another structure involved in the bilateral regulation of muscle activity during walking and locomotor adaptation is the CPGs mentioned in section 2.6.1.6. Based on current, albeit limited, knowledge in human locomotion, these functional networks are considered to be involved in rhythm generation and the regulation of the pattern of motor output (muscle synergies) (Duysens & Van de Crommert, 1998; Rybak et al., 2006). Furthermore, the CPGs are thought to be influenced by supraspinal centers (Duysens & Van de Crommert, 1998; MacKay-Lyons, 2002) which points to the possibility that these circuits can be altered secondary to a stroke with consequences on the generation of motor output during locomotor adaptation.

Finally, considering that limbs can also be independently controlled (Choi & Bastian, 2007; Nielsen et al., 2008), these circuits cannot be ignored. Compared to the previous examples on crossed-spinal pathways, a larger number of studies exist about impairments of spinal and supraspinal circuits post-stroke involved in the independent control of limbs. Among these impairments are the reduced ability to regulate descending neuronal input to activate skeletal muscles (Beaulieu et al., 2014; Bowden et al., 2014), reduced corticospinal drive to the affected limb (Nielsen et al., 2008), increased afferent input to the alpha-motoneurons (Li & Francisco, 2015), as well as modified modulation of presynaptic inhibition of Ia afferents (Aymard et al., 2000; Dyer et al., 2009; Dyer et al., 2014; Lewek et al., 2007; Li & Francisco, 2015). As such, the damage to these circuits can lead to deficits in muscle function during voluntary activation or functional tasks such as walking, as discussed in sections 2.3.2.1 and 2.4.1 respectively. Although in the present work we did not measure reflex responses, such changes have been frequently reported secondary to a cerebral stroke (section 2.3.2) supporting a role of reduced inhibited reflex responses in producing the observed differences in EMG activity found between NP-fast and P-fast conditions.

While differences in activity of a larger numbers of muscles were associated with step length after-effects during the P-fast condition, the fast TA and VL and the slow GL increased during both conditions. Considering that lower limb deficits could influence the control of muscle activity as discussed above, it is particularly relevant to question the functional relevance of these changes in the fast TA and VL, and slow GL observed in our stroke group. This aspect was not completely addressed in paper #1 and will be discussed in the following section.

6.2.3 The Functional Relevance of SBT-induced Effects on Muscle Activity

The changes in the fast TA and VL, and slow GL observed in the current cross-sectional study were considered as generally positive, functional, and not in line with adverse effects. This conclusion is based on the comparison of SBT-induced effects on EMG activity in the 16 post-stroke participants with those observed in the control group (n = 10) and with previous findings on kinetics during SBT walking. In addition to the larger contribution in muscle activity discussed previously, similarities in changes of EMG activity between individuals post-stroke and healthy controls were also found. These similarities were considered as indicators of functional and “normal” changes. EMG activity during the adaptation period will not be compared between groups due to post-stroke participants’ use of the handrails for stability as explained in more detail in paper #1.

The comparison of SBT-induced changes in muscle activity of individuals post-stroke relative to healthy controls revealed that SBT walking led to three common after-effects in EMG activity between these two groups (Figure 6-1). First, six minutes of fast-belt walking led to an increase in the TA and the RF muscles during early post-adaptation. Second, during that period, all muscles with after-effects de-adapted towards baseline values after approximately 10-15 steps as apparent in Figures 5-2 and 5-3. Third, the TA (fast) and ST (slow) muscles showed moderate to strong correlations with the step length after-effects in our group of healthy controls and in post-stroke participants during the P-fast condition.

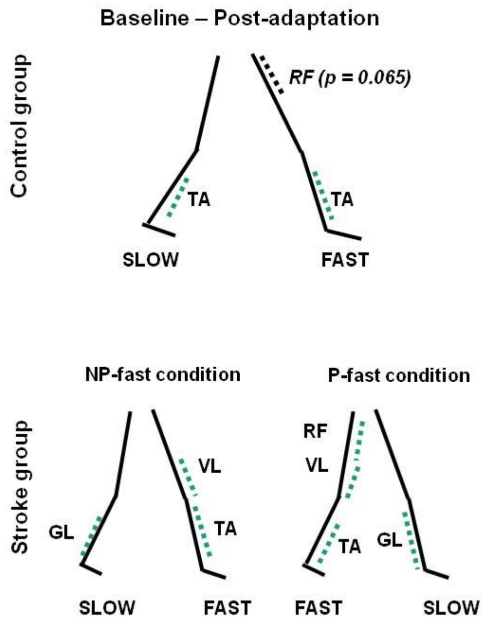

Figure 6-1. Muscles with a significant increase (green dotted lines) or near significance (black dotted lines) from baseline to post-adaptation for healthy controls (upper part) and individuals post-stroke (bottom part). For the stroke group, results from both conditions are presented (NP-fast and P-fast). Abbreviations: NP = Non-paretic, P= Paretic, D = Dominant, ND = Non-dominant, TA = tibialis anterior, GL = gastrocnemius lateralis, VL = vastus lateralis, RF = rectus femoris.

6.2.3.1 Mechanisms of Locomotor Adaptation

The aforementioned common effects between individuals post-stroke and healthy controls were associated with the fast TA muscle which appears to play a particular role in locomotor adaptation in both groups. This role can be explained within the feedforward control in line with the theory of the internal model. According to a recent investigation on locomotor adaptation by Ogawa et al. (2014), the TA muscle is an important contributor to feedforward control during SBT walking. This conclusion was based on the consistent relationship between the pattern of adaptation found in the TA muscle and the braking force during early stance phase of gait. In fact, the TA muscle increased activity to ensure stability and prepare for the increased impact during initial contact (i.e., braking force) provoked by the new walking condition (split-belt configuration). Thus, the changes in TA EMG were considered as a predictive adjustment for the upcoming event of initial contact and therefore can be described as a feedforward control mechanism.

The feedforward hypothesis can be supported by previous studies on motor control that attributed a role in feedforward control to the TA muscle given the strong supraspinal influence on the activity of this muscle (Capaday et al., 1999; Christensen et al., 2001; Dietz et al., 1992; Palmer et al., 2016). For example, the analysis of motor-evoked potentials (MEPs) and reflex responses suggests that increases in TA activity during the stance phase of walking when external perturbations are applied is explained by its role in securing stability around the ankle (Christensen et al., 2001). Enhanced excitability of the corticospinal pathways of the TA muscle was found during stance phase in healthy individuals when compared to voluntary plantarflexion (Capaday et al., 1999; Christensen et al., 2001) or compared to the swing phase (Christensen et al., 2001). In our first study, the EMG activity of fast dorsiflexors in the stroke group increased particularly during stance as apparent in the EMG profiles in the supplementary figures of paper #1 (Figures 5-5 and 5-6). This could be considered as functionally relevant and "normal" due to its presence in healthy individuals.

6.2.3.1.1 *Proximal Muscles in Feedforward Control*

In addition, the quadriceps group (RF: both groups; VL: individuals post-stroke) on the fast leg increased in line with the TA during the stance phase of gait as illustrated for the stroke group in Figures 5-5 and 5-6. It has been established that the interaction between the quadriceps group and the ankle dorsiflexors plays an important role during walking (Lamy et al., 2008; Marchand-Pauvert & Nielsen, 2002). Quadriceps activity can influence the activity of the dorsiflexors and vice versa, mediated through recurrent collaterals and group Ia motoneuronal pathways, respectively. This interaction takes place particularly during heel contact (Lamy et al., 2008) and early stance (Marchand-Pauvert & Nielsen, 2002) when both muscles become coactivated. Of interest to this discussion are the findings by Achache et al. (2010), who showed that in 14 individuals post-stroke, the excitation (late latency response) from paretic ankle dorsiflexors to the quadriceps group activation has been shown to be enhanced compared to healthy controls. These authors concluded that the enhanced excitation might be explained by the dorsiflexors providing assistance in knee stabilization (extension) and leg support at early stance when "balance is particularly challenged". Moreover, healthy individuals presented higher braking forces during post-adaptation on the fast leg compared to baseline (Miéville et al., under revision; Ogawa et al., 2014). Therefore, the increase in VL and RF in our stroke and control group could be explained as an anticipatory mechanism induced by feedforward control to counteract the increased impact force during early stance. This role in knee stabilization during initial stance was also outlined in the discussion of paper

#1 to explain the observed changes in fast VL and RF that were in line with step length after-effects.

6.2.3.1.2 *The Role of Ankle Stiffening*

Thus, the theory of internal models provides a good explanation for how these changes in TA and quadriceps EMG can occur within the context of feedback and feedforward control. However, it still remains unclear what exact mechanisms underlie such feedforward control in individuals post-stroke. Ogawa et al. (2014) suggested that in healthy individuals the TA feedforward control and stabilization of the ankle is in line with an overall stiffening of the ankle joint. While in the present cross-sectional analysis the anterior and posterior GRFs were not analyzed, the supplementary results on coactivation can provide an alternative approach to explore the possibility of a presence of ankle stiffening during post-adaptation in our participants.

An increase in coactivation can be considered as a factor leading to an increase in joint stiffness (Hortobagyi et al., 2008; Simmons & Richardson, 1988). Indeed, the cross-sectional analysis revealed that both the post-stroke participants and healthy controls increased the duration of ankle antagonist coactivation (TA-GL), particularly after fast-belt walking (Figure 5-9). However, as illustrated in Figure 5-9, the coactivation durations varied substantially among participants in comparison to baseline. Nevertheless, using coactivation as an indicator for stiffness, the present SBT protocol might be less appropriate for individuals who already have high coactivation duration or in this case, ankle stiffness. The issue of ankle stiffening is further complicated by the fact that different approaches have been used in literature to define joint stiffness such as joint angles and torques (Bressel & McNair, 2002; Chung et al., 2004; Rydahl & Brouwer, 2004) in addition to the aforementioned coactivation (Hortobagyi et al., 2008; Simmons & Richardson, 1988). Nevertheless, whatever method is used to quantify joint stiffness, it remains unclear whether a stiffening of a joint in line with motor adaptation can be considered as negative or positive and functional.

It is also possible that the initially observed increased stiffness (or increase in coactivation) could simply represent the normal reaction to a new situation as suggested by the results of studies exploring the consequences of external perturbations of goal-oriented movements of the upper extremity (De Serres & Milner, 1991; Gribble et al., 2003; Milner, 2002; Milner & Cloutier, 1998). For example, following repeated exposure to the perturbations of arm trajectories, the initial muscle stiffness adapts (in this case becomes reduced) providing more efficient and less energy-consuming movement control. This interpretation appears to be applicable to the context of SBT walking since reduction in energy consumption was found in healthy individuals during the adaptation period of SBT walking (Finley et al., 2013). Energy consumption was quantified as metabolic power (Watt/kg), which increased initially when the belts were switched from tied- to split-belt configuration and reduced with time, in line with the adaptation of step length towards symmetry. Therefore, considering the after-effects and de-adaptation found in the fast dorsiflexor (TA) during the early stance phase of gait, this muscle appears to play a particularly relevant role in the feedforward adjustments needed during locomotor adaptation (i.e., after-effects) in individuals post-stroke similar to healthy controls. Nevertheless, future research including lower limb biomechanics and controlled trials are necessary to provide a greater characterization of the mechanisms (e.g., joint stiffening) underlying this feedforward control during locomotor adaptation post-stroke.

6.2.3.1.3 *Feedback Control and the Role of Plantarflexors*

Complementary to feedforward control, feedback control in SBT walking is thought to involve the plantarflexor muscle group during the stance phase of gait (Ogawa et al., 2014). This is in agreement with previous findings suggesting that the control of plantarflexors during stance is mainly regulated at the spinal level based on proprioceptive feedback (Dietz, 1992; Dietz et al., 1994a) and is less influenced by supraspinal centers as shown by studies using MEP (Capaday et al., 1999; Ung et al., 2005). Proprioceptive feedback is a major contributor of information necessary for the centrally regulated feedforward control observed during adaptation (Hwang & Shadmehr, 2005; Lackner & DiZio, 2000; MacLellan et al., 2014). Next to the plantarflexor's role as a provider of feedback, this muscle group is an important contributor to support and forward acceleration of the body during the stance phase of gait (Kepple et al., 1997; Sutherland, Cooper, & Daniel, 1980; Winter, 1980; Zajac et al., 2003) as described in section 2.1.3. In the present cross-sectional study, the slow plantarflexors increased their activity significantly in our group of individuals post-stroke during post-adaptation, which was most pronounced during stance phase of gait (Figures 5-5 and 5-6). This increase during stance is likely to be linked to its functional contribution to forward propulsion. These results lead to the conclusion that the increase in activity of the plantarflexor muscles (GL) after slow-belt walking is functionally relevant and a positive change, as discussed further in the following section.

6.2.3.2 Forward Propulsion and Step Length Changes

The interpretation that the increase in plantarflexor activity after slow-belt walking played an essential role in forward propulsion and contralateral step length changes was driven by the timing of increase (Figures 5-5 and 5-6) as well as the moderate to strong correlations of the increase found with step length after-effects (Figure 5-4). During early post-adaptation, the slow plantarflexors showed an increase in EMG activity during mid and late stance (20% to 60%) of the gait cycle in both conditions (NP-fast and P-fast). This was supported by the increase in peak plantarflexion net moment between 20% and 55% of gait cycle observed in individuals post-stroke, as well as after slow-belt walking (Lauzière et al., 2014a). In both investigations, changes in the slow plantarflexor muscles were associated with increases in contralateral step length ($r > 0.60$). This aspect was discussed as well in paper #1.

Interestingly, such contribution from the plantarflexor muscles to contralateral step length after-effects was not observed in our control group. In this group, neither slow- nor fast-belt walking led to changes in plantarflexor EMG activity (Figure 5-8). The lack of changes in plantarflexor EMG corresponds as well to the diminished after-effects observed on fast step length in this group, mentioned above. In agreement with this interpretation are the relatively small increases in peak net plantarflexion moment found in healthy individuals (4.1%) compared to the post-stroke group (16.5%) by Lauzière et al. (2014a). Such small changes in peak net plantarflexor moment might not be readily visible at the level of plantarflexor EMG activity.

A possible explanation for this discrepancy in changes of plantarflexor activity between individuals post-stroke and healthy controls can be found in the differences in belt speeds used during the protocols. This suggestion is based on the positive correlations found between walking speed and the plantarflexor EMG activity (Den Otter et al., 2004) as well as the net moment during stance phase of gait (Goldberg & Stanhope, 2013). In both our cross-sectional analysis and the study by Lauzière et al. (2014a), healthy participants walked at slow-belt speeds which were at 70% of their comfortable speed (range = 0.56 - 0.84 m/s). In the post-stroke group, however, slow-belt speed was set at their comfortable walking speed on the treadmill. Yet, in the Ogawa study (2014), healthy individuals walked as well at very slow speeds (i.e., slow belt = 0.5 m/s), but showed an increase in the plantarflexor EMG during early post-adaptation. Therefore, it is likely that for the healthy individuals in our control group, step length after-effects were mainly regulated by a decrease in slow step length. Consequently, they required less contribution of the slow plantarflexors to increase fast step length compared to our individuals post-stroke. This is in line with the hypothesis that the slow side is used as a reference to drive the adjustment of the lower limb to SBT-induced disruption of gait (Ogawa et al., 2014; Vasudevan & Bastian, 2010). This leads us back to the earlier stated conclusion concerning step length (section 6.2.1), saying that individuals post-stroke are more likely to refer to bilateral feedback during locomotor control than their healthy controls. However, this interpretation appears to be related only to after-effects in EMG amplitude, but not to the temporal features of muscle activity, as discussed in the following section.

6.2.4 EMG Timing and Interlimb Coordination during SBT Walking

Changes in temporal features of EMG activity due to the SBT protocol were modest in comparison to changes in EMG amplitude and differed between the study groups. In individuals post-stroke, interlimb coordination did not change from baseline to post-adaptation, while healthy individuals displayed a reduction in the coordination for the TA and ST muscles with a correlation coefficient (Rxy) smaller than 0.46 ($p \leq 0.028$) (Figure 5-11). With regards to coactivation, its duration changed only in the ankle antagonists (TA-GL) after fast-belt walking in healthy individuals and only after paretic fast-belt walking in the stroke group. The marginal changes in temporal EMG features are in accordance with those described previously in the literature. In the context of SBT walking, the analysis of muscle synergies in healthy individuals revealed that the majority of EMG patterns of four different synergies were not altered in a significant manner between tied- and split-belt conditions (MacLellan et al., 2014). MacLellan et al. (2014) concluded that the little changes in temporal aspects of EMG profiles observed and the similarity between sides during SBT walking reflect a tight link between limbs.

In terms of group differences in interlimb coordination, the absence of change found in temporal EMG post-stroke can be explained by the reduced flexibility of the EMG activation pattern found during walking (Clark et al., 2010), pedaling (Ambrosini et al., 2016) and bilateral isometric force production in individuals with chronic stroke when compared to healthy controls (Kautz & Patten, 2005). These studies evaluated muscle synergies based on temporal patterning (activation profiles) of the respective muscles, similar to how the calculation of interlimb coordination is performed. The analysis of lower limb muscle synergies during walking revealed that individuals post-stroke (n = 55) present a smaller number of synergies (2-4 modules) when compared to healthy controls (4 modules) (Clark et al., 2010). Such a reduction in activation modules was particularly pronounced on the paretic side and interpreted as a compromised capacity to vary motor output, and thus muscle synergies.

Furthermore, temporal aspects in EMG were found to be less affected by modifications in spatiotemporal gait parameters in individuals post-stroke (De Quervain et al., 1996; Den Otter et al., 2006). For example, Den Otter et al. (2006) found that the on- and offset of lower limb muscle activity did not display normalization despite improvements in gait speed and a reduction in swing phase asymmetry. Lower limb muscles continued to show prolonged activation of proximal and distal lower limb muscles on the paretic side along with premature activation of the paretic gastrocnemius medialis.

To summarize, a role in feedforward control was attributed to the TA muscle in both individuals post-stroke and healthy controls. Furthermore, in individuals post-stroke the after-effects in step length were mainly related to the plantarflexor muscles. These findings are in agreement with previous findings on SBT effects in lower limb peak net moments (Lauzière et al., 2014a) in individuals post-stroke and muscle activity (Ogawa et al., 2014) in healthy controls. Furthermore, the results confirm our study hypotheses stated in chapter 3 that suggested a particular contribution of distal lower limb muscles to step length after-effects and an increase in slow plantarflexor activity. In the following section, the effects of repeated exposure to the protocol used in our training study will be discussed taking into account the previously discussed immediate after-effects of SBT walking.

6.3 Effects of Reapeated Exposure to SBT Walking

Repeated exposure to the error-augmentation-based protocol resulted in improvements in step length symmetry and walking speed. These findings confirm previous results suggesting that repeated error-augmentation-based SBT walking is an appropriate intervention to improve step length symmetry post-stroke (Reisman et al., 2013). In line with these interpretations, our results support the theory that repeated exposure to motor adaptation results in motor learning that can be viewed as the retention of an adapted pattern over a longer period (Bastian, 2008; Martin et al., 1996; Reisman et al., 2010). The following sections discuss the effect of the SBT training beyond what was covered in papers #2 and #3 with consideration of the results from the cross-sectional studies.

6.3.1 Step Length Asymmetry

The majority of our participants achieved a reduction in step length asymmetry through a bilateral increase in step length which was more pronounced on the fast side. However, according to cross-sectional studies (paper #1 and (Malone & Bastian, 2013; Malone et al., 2012; Reisman et al., 2005; Reisman et al., 2007), step length after-effects were characterized by a reduction of step length after walking on the slower belt and an increase of step length after fast-belt walking. This distinct behaviour in step length changes observed in the cross-sectional and training studies might be the result of walking speed, which was not controlled for in the training study. This was likely due to the fact that in the training study, participants walked over ground instead on the treadmill. Our participants increased their walking speed post-training, which could have led to bilaterally longer step lengths. However, Reisman et al. (2013) reported a similar strategy in step length changes in their group of participants minus the changes in walking speed post-training. Indeed, such bilateral increase was observed during split-belt configuration (adaptation) in both NP-fast and P-fast conditions in the cross-sectional analysis as illustrated in Figure 5-1 A. Together, these results suggest that the split-belt configuration itself, which was characterized by a speed ratio of 2:1 in both studies, led to these common changes in step length and not the period of post-adaptation (tied-belt configuration) during which we observed the after-effects. Therefore, our results suggest that repeated exposure to these novel locomotor environments provoking adaptation led to learning, in agreement with the theories on motor adaptation and learning (Malone et al., 2011; Martin et al., 1996; Reisman et al., 2010). As such, it can be assumed that the direct demand of feedback and feedforward control required to adjust to switching belt speeds was no longer necessary to the same extent since participants now used new learned patterns (altered step length ratio) within the context of walking over ground. This could explain the lack of pronounced changes in the TA muscle, since its feedforward control was not required in the same manner during walking over ground one day post-training compared to the post-adaptation period in the cross-sectional analysis (see section 6.2.3). Furthermore, as discussed in the following section, the maintained changes in step length post-training are likely associated with the increase in fast and slow plantarflexion net moments during late stance.

6.3.2 Joint Kinetics and Muscle Activity

The effect size estimations used in paper #3 suggest that the plantarflexor muscle group appears to be the main contributor to step length changes induced by six sessions of SBT training. These findings confirm the research hypothesis presented in chapter 3 and are in accordance with the conclusion of paper #1 as well as the results from previous investigations on the relationship between plantarflexion moments and step length post-stroke (Allen et al., 2011; Lauzière et al., 2014a). Nonetheless, our results suggest that the contribution of the plantarflexor muscle to step length was slightly different after repeated exposure to the SBT protocol when compared to the post-adaptation period directly after walking with the split-belt configuration. Immediate after-effects in peak net plantarflexion moments (Lauzière et al., 2014a) and EMG activity (paper #1) increased after slow belt walking. In parallel, step length increased after fast-belt walking. Since these two effects were strongly correlated, it can be concluded that the plantarflexor muscles are a main contributor to changes in contralateral step length. This role of the plantarflexors was less evident in the context of the training study, where repeated exposure to this protocol did lead to a bilateral increase in step length and peak net plantarflexion moments but with significant changes only found on the side trained on the faster belt. The peak net plantarflexion moment on the slow side showed only the tendency to increase post-training, but with a strong effect size of changes as estimated by the Hedges's g_{av} score with a size of 0.71. Therefore, it is likely that the slow plantarflexor muscles contributed during late stance of gait to an increase in fast step length. Yet, as aforementioned significant increase in peak net plantarflexion moment was found on the fast side (Table 5-6 and Figure 5-19, section 5.3). This suggests that the increase in fast peak net plantarflexion moments contributed to the increase in the ipsilateral (fast) step length. This can be explained by the gastrocnemius group of this muscle which is considered to deliver power to the leg for swing initiation (Neptune, Kautz, & Zajac, 2001; Zajac et al., 2003). An augmented propulsion force during late stance, which induced an increase of energy delivery to the leg before swing initiation, could contribute to an increase in ipsilateral step length.

6.3.2.1 Heterogeneity in EMG During Walking Over Ground

In order to better assess the heterogeneity in EMG activity, for each participant and muscle, a single SD of mean activity during pre-evaluation was calculated. In the case where the difference in muscle activity between pre- and post- or pre- and follow-up evaluation exceeded this SD, the difference was considered as a real change. The comparison of individual SDs and differences revealed that for each muscle and side, an average of 50% of participants showed real changes in activity between the compared evaluations. Furthermore, the direction of changes varied substantially among participants as illustrated in Figure 5-20 of paper #3. For example, three participants increased the TA %max RMS of EMG on the side trained on the slow belt. Two participants did not present changes in this muscle and five participants presented a decrease in TA %max RMS. Furthermore, among participants' changes in muscle activity post-training, different combinations of muscles were found as discussed in paper #3. For example, among the aforementioned three participants with an increase in slow TA %max RMS, two presented an increase in the slow VL %max RMS and one showed a combined increase of slow TA %max RMS and fast ST %max RMS. From a clinical perspective, our participants presented a large variety of muscle contributions underlying improvements in step symmetry and walking speed.

A more detailed appraisal of the individual patterns of EMG changes revealed no specific pattern of changes in individuals with similar lower limb deficits or gait patterns.

More precisely, the categorization of participants based on gait speed, the severity of asymmetry, spasticity or the CMSA score did not reveal any conclusive similarities in EMG changes. Furthermore, classifying the present participants' gait pattern based on the categories of De Quervain et al. (1996), we found 4/10 displayed hyperextension of the knee (extension thrust), one had a stiff knee, and 4/10 had a more flexed knee (buckling) during walking. Based on visual inspection, these three gait patterns were not linked to a particular pattern or combination of changes in EMG activity. As mentioned in the discussion section of paper #3 such heterogeneity is not surprising from a clinical point of view. Briefly repeated, this argument as outlined in paper #3's discussion was based on the conclusion of previous studies that individuals post-stroke use a variety of strategies to achieve a specific task (Jonkers et al., 2009; Jonsdottir et al., 2009) which becomes particularly evident during walking over ground and that changes in gait pattern or ability were not associated with a consistent pattern of EMG activity (Buurke et al., 2008; Den Otter et al., 2006).

6.3.2.2 Symmetry in Biomechanical Parameters and Step Length

While asymmetry in step length was reduced post-training, symmetry indices of muscle activity and net joint moments did not change (Table 5-7). Two possible explanations were found for the lack of changes in symmetry of kinetics and muscle activity in the context of this training study.

First, the belt configurations were based on the participants' initial step length asymmetry and not on the asymmetry in peak net plantarflexion moments. Participants were categorized into two training groups with 4/10 trained with the paretic side on the faster belt (P-fast group) and 6/10 with the non-paretic side (NP-fast group) within the scope of paper #3. However, as expected, all ten participants presented lower peak net plantarflexion moments on the paretic side as illustrated in Figure 5-21. Our results suggest that the paretic side should be trained on the faster belt to improve symmetry in this parameter, since peak net plantarflexion moment increased particularly on the side trained on the fast belt. The four participants trained in the P-fast group improved symmetry in plantarflexion moments while the six participants in the NP-fast group remained asymmetrical despite improvements in step length symmetry. However, with regards to statistical analysis, both groups were included, and the test statistics were based on “slow” and “fast” and not “paretic” and “non-paretic” side in order to interpret the effect of the belt speeds on the outcome parameters.

There is an additional explanation that could be provided, which might have become more evident in a larger study sample. This explanation is based on the limitations in motor performance and on the underlying compensatory strategies used. A recent publication by Lauzière et al. (2016) showed that the capacity to adjust step length and paretic plantarflexion moments during post-adaptation was influenced by the level of muscular effort in the paretic plantarflexors (section 2.6.3.1.2). In that study, data from participants with shorter non-paretic step lengths were analyzed. Participants walked for six minutes, with the non-paretic side on the faster belt, which led to an increase in non-paretic step length and thus improved step length asymmetry as after-effects during post-adaptation. The capacity to increase non-paretic step length and paretic plantarflexion moment during post-adaptation was limited in the subgroup which displayed high levels of effort in the paretic plantarflexors during the baseline period when compared to the participants with a low-level of effort in this muscle. According to these findings, improvements in the symmetry of plantarflexor moments in line with the improvements in step lengths symmetry appear to be less likely in individuals who already exert high levels of effort in the paretic plantarflexor muscle. Furthermore, the analysis of muscular effort (muscular utilization ratio; MUR) of lower limb muscles during level walking revealed that some individuals post-stroke with plantarflexor weakness achieve higher walking speeds by compensating through the use of hip flexors during the push-off phase of gait (Nadeau et al., 1999). Therefore, an MUR-based analysis on the effects on effort after repeated exposure to SBT walking could reveal insight into possible compensatory strategies used in individuals post-stroke to achieve improvements in step length symmetry. Moreover, these findings about MUR subgroups support the assumption stated in paper #3, that our group of participants used different strategies and muscle contributions to improve step length asymmetry during walking over ground.

6.3.3 Feasibility

The results presented in this thesis on gait ability and biomechanics suggest that repeated exposure to SBT walking has the potential to form a basis for therapeutic interventions aimed at gait recovery after stroke. In addition, from a clinician's viewpoint, this intervention appears promising and applicable in a rehabilitation setting as described in detail in section 5.2.12. By means of three valid questionnaires on the practicality of the protocol and SBT use (Perceived Usefulness questionnaire; [Davis, 1993]), readiness for implementation on an individual (PERFECT Tool; [Menon et al., 2010]) and organizational level (ORIC; [Shea et al., 2014]), it has been shown that the three participating therapists in our training study considered the protocol and the use of the SBT as practical and ready for implementation. These positive findings, while very encouraging, emphasize the need for further qualitative research in line with controlled trials to strengthen the present conclusion and to support the implementation of this intervention in clinical settings.

6.4 Study Limits

While much insight can be gained from the findings of the cross-sectional analysis and the training protocol, there are three particular study limitations that can be highlighted in these two projects. The limits are related to the study protocols, the EMG analysis and the generalization of the findings, as discussed below.

6.4.1 Limitations of the Study Protocols

6.4.1.1 Cross-sectional Analysis

The SBT protocol contained two walking conditions with each consisting of three periods (baseline, adaptation, and post-adaptation). These two walking conditions were not randomized between participants. All participants started with the condition where all walked with their non-paretic side on the faster belt during the adaptation period (NP-fast condition). This was followed by a second condition which was identical except that the paretic side was placed on the fast belt during the period of adaption (P-fast condition). The decision to present these conditions in the same order across participants was based on the experimental experience with healthy controls, where the protocol – including the preparation of the participants – was generally long, which could be fatiguing for individuals who have suffered a stroke. Consequently, the walking conditions were set in a consistent, non-randomized order to ensure that data would be obtained from at least one common condition (NP-fast).

Furthermore, to limit the extent to which the effects observed during the second condition (P-fast) were not biased by the after-effects obtained from the first condition (NP-fast), participants had to take a break between conditions until they felt well rested. As discussed in paper #1, the after-effects during NP-fast condition were sufficiently washed out (Section 5.1). Moreover, the values obtained from the adaptation and post-adaptation periods were always compared with their corresponding baseline. Therefore, we are convinced that the effects observed during the P-fast condition are valid and not biased by the NP-fast condition. However, following the completion of this analysis it is clear that, in fact, the two conditions can be tolerated by post-stroke participants such that for future studies, a randomized design is recommended.

While handrail-holding during the period of adaptation (split-belt configuration) limited the comparison between healthy controls and post-stroke participants, and the interpretation of muscle activity and joint kinetics during this period, its use is still highly recommended in this study population for security purposes. Furthermore, previous studies using the SBT protocol have also reported that participants held on to handrails during the protocol (Malone & Bastian, 2013; Reisman et al., 2007). Controlled trials testing the effects of SBT walking with and without handrail holding will be required to investigate the influence of holding handrails on gait parameters during SBT walking.

6.4.1.2 The Training Study

Another possible study limitation was the lack of a double baseline testing on step length symmetry. A double baseline testing would have given information about participants' actual variability of step length symmetry. This baseline variability could have served as a benchmark for actual change. However, during post-training, all participants exceeded the baseline variability of asymmetry (NLSD = 0.020) presented in the previous investigation on split-belt training (Reisman et al, 2013). Moreover, 8/12 participants exceeded the reference value for minimal clinical relevant change which is a change in L/S ratio over 0.15 (Lewek & Randall, 2011) as described in paper #2.

A further limitation can be found in the number of training sessions. The combined analysis of values from both measure types (3D-motion analysis and clinical evaluation) resulted in significant and clinically relevant changes post-training. Nonetheless, the six sessions should not be taken as a benchmark for the number of training sessions. Among the twelve participants who reduced their step length asymmetry after six sessions of training only four fell within the range of symmetry (L/S ratio $\leq$ 1.08) (Figure 5-13 A, paper #2). Consequently, it can be concluded that the remaining eight participants were likely to require more session to achieve an L/S ratio below 1.08. Therefore, in order to achieve symmetry, it is highly recommended to quantify step length asymmetry before each training session, in order to determine the number of sessions still required to achieve symmetry and to avoid over-correction.

6.4.2 Limitations of the Parameters Tested

6.4.2.1 Cross-sectional Analysis

As described in section 2.1.4., the purpose of the cross-sectional study was to quantify muscle activity during walking, and EMG is a common and reliable tool to do so. However, as exemplified in section 2.1.4.4, EMG analysis is bounded in the interpretation of force requirements or kinetic energy generation and absorption during a dynamic task. Therefore, in the context of the present cross-sectional analysis, it might have been interesting to relate changes in EMG with those of joint kinematics and kinetics. However, considering that the effects of an error-augmentation-based SBT protocol on muscle activity had not been investigated in individuals post-stroke prior to our investigation, we decided to concentrate on EMG analysis and refer to the work of Lauzière et al. (2014a; 2016) for comparison with kinetic data. Since 16/20 participants from the studies of Lauzière and colleagues were common to our cross-sectional analysis, the results between these studies are comparable. Nonetheless, the analysis of joint angles in line with power production can reveal additional information about force production of a muscle during specific gait events (i.e., initial contact) in the context of SBT walking. Therefore, the use of a combined analysis of EMG and kinetic data is suggested for future studies. In our study, the statistical analysis did not consider the comparison of EMG activity relative to specific sub-phases of gait. Instead, visual inspection of mean activation profiles of EMG illustrated in Figures 5-5 and 5-6 were related to a specific phase of gait. Such categorization of EMG activity into sub-phases could have improved the interpretation of the contribution of a muscle activity during specific gait events.

6.4.2.2 Training Study

Another possible limitation could be due to the choice of the technique of EMG amplitude normalization. To interpret the changes in muscle activity induced by the SBT training, the EMG signal was normalized using the peak value obtained during the corresponding session (e.g., post-evaluation; see section 4.2.4 for further details). This was necessary due to the fact that the electrodes had to be repositioned for each session. Despite the measures of electrode placement taken during the first evaluation session (pre-evaluation), it is still possible that the electrode was not placed exactly at the same point over the muscle belly from session to session. Furthermore, skin conditions can change between sessions. These factors can vary the amplitude of the EMG signal (De Luca, 1997; De Luca et al., 2012) as described in chapter 2.1.4.3. As discussed in the same section (2.1.4.3), the normalization to a reference peak value results in smaller variability than the normalization to a mean value (Burden et al., 2003) and was considered as most reliable in individuals post-stroke when compared to other normalization techniques (Teixeira-Salmela et al., 2012). In addition, Blanchette et al. (2012), showed that the normalization method we used (reference value from the respective day) was sensitive enough to detect significant differences in activity of lower limb muscles in healthy individuals after four days of locomotor-perturbation training on a treadmill. Considering all these points, the variability of changes in muscle activity found in our post-stroke participants is more likely due to an actual variability of strategies in EMG changes than to the method of amplitude normalization.

Nonetheless, the chosen normalization method requires careful scrutiny when comparing EMG data with kinetic values such as net joint moments. Since the muscle activity was normalized with a reference peak value from the respective evaluation date, conclusions about an actual increase or decrease of EMG activity when compared to the other evaluation days should be handled with care. Conversely, net joint moments were normalized to the participant's bodyweight and therefore showed actual change in net moments between evaluation sessions. Nevertheless, this issue is more relevant to future studies since in the results presented in paper #3 there was a lack of significant changes in EMG activity making a comparison less relevant overall.

6.4.3 Generalization

Considering the stroke characteristics of our entire study cohort (n = 26), the participants seem representative of a relatively large population of stroke survivors, strengthening the generalization of our findings. Our participants were chronic stroke survivors with ages ranging from 29 years to 73 years, displayed mild signs of stroke (Wolf, Baum, & Conner, 2009), were independently ambulant with speeds slightly slower than healthy controls and they suffered from a large range of lower limb gait deficits as described in section 6.1. According to Wolf et al. (2009) the severities of cerebral strokes are increasingly “mild”. For example, the percentage of individuals with mild stroke ranges from 67% (out of 304 individuals) (Duncan et al., 1997) to 49% in a cohort of 7,740 (Wolf et al., 2009). Individuals who suffered a mild stroke presented physical impairments, but were frequently able to walk (Duncan et al., 1997). Indeed, Lord et al. (2004) showed that about 60% of 115 participants regained the capacity to walk independently in the community. Lower limb and gait deficits are still frequently present in independent-ambulant individuals post-stroke (Ada et al., 2009; Dunn et al., 2015; Lauzière et al., 2014b; Mulroy et al., 2003; Patterson et al., 2008a; Perry et al., 1995). According to a large retrospective epidemiological study on US stroke population (years: 1999-2005), the incidence of cerebral strokes in younger individuals (< 55 years) is increasing (Kissela et al., 2012). Indeed 16/26 (61%) of our participants were younger than 55 years with an average of 49.8 (13.4%) in our cross-sectional analysis and 53.3 (SD 8.7) in the training study (Tables 5-1 and 5-3).

The generalizability of our findings was further demonstrated by the reproduction of previous findings on short and longer term changes in step length and step length asymmetry (Lauzière et al., 2014a; Malone & Bastian, 2013; Reisman et al., 2005; Reisman et al., 2013). As far as immediate after-effects on muscle activity are concerned, the results of the cross-sectional study are considered strong enough to be generalizable. In particular, both walking conditions led to an increase in slow GL, fast TA, and fast VL in a group with a large variety of lower limb deficits. However, the results of the training study indicate a more limited generalization when considering the EMG and net joint moment results. The variability and the generally small to moderate effect size in EMG data and net joint moments emphasize the need for a larger sample size before a more robust generalization of the results can be attained. In addition, while the present training protocol resulted in improvements of walking speed, participants in the Reisman study (2013) did not increase speed post-training. This is most likely due to the differences in how the belt speed on the slower side was determined as discussed in paper #2. Furthermore, each study analyzed a small number of 12 participants and did not use control groups. Therefore, the interpretation about the superiority of our protocol relative to the protocol of Reisman et al. (2013) regarding the improvements of walking speed shoyld be considered with some caution.

6.5 For Future Research

The previously discussed limitations and the results of the pilot training study lead to several recommendations and suggestions for future studies. Some of these recommendations are relevant to the two study types used in the current thesis (cross-sectional and training) while others are specific to the investigation of effects of repeated exposure to SBT walking (training).

6.5.1 Recommendations Concerning Parameters Analyzed

The analysis of muscle synergies in the context of SBT walking in individuals post-stroke would help further insight into the functioning of different motor modules during locomotor adaptation. This approach has been used in the past, for example by MacLellan et al. (2014) and Clark et al. (2010), to study the contributions of lower limb muscles during walking. These authors defined synergy as motor modules which are based on combinations of muscles that coactivate during a specific task and the amount each muscle contributes to this task (Clark et al., 2010; MacLellan et al., 2014; Safavynia et al., 2011). According to Clark et al. (2010), individuals post-stroke showed less variety in motor modules when compared to healthy controls, indicating a reduced flexibility in muscle coordination. MacLellan et al. (2014) were able to demonstrate that healthy individuals presented four different patterns of muscle activity during SBT walking, that is two patterns related more to knee controlling muscles (VM, VL, and RF) and two others to plantarflexor (GL, GM, and SOL) and contralateral hamstring (BF) muscles either on the slow or fast side. As mentioned in section 6.2.4, the temporal pattern of these modules changed only marginally during different walking conditions on a SBT. Therefore, analysis of synergies in individuals post-stroke and the context of SBT walking could reveal information about the influence of the motor modules and their variety on the capacity of locomotor adaptation post-stroke. This could provide insight into compensatory strategies used by individuals post-stroke. Furthermore, in the context of a training study it could reveal information about whether coordination flexibility improves in combination with improvements in gait ability.

Another interesting aspect of coordination that could be investigated in the context of locomotor adaptation concerns the link between arm and leg movements during walking. Upper and lower limbs present a rhythmic movement pattern during walking (Zehr & Duysens, 2004). This coordination and connection between upper and lower limbs has been found to be preserved in individuals post-stroke (Klarner et al., 2014; Stephenson, De Serres, & Lamontagne, 2010; Stephenson, Lamontagne, & De Serres, 2009). However, in contrast to healthy controls, in individuals post-stroke, the quality of this coordination, as quantified with a cross-correlation coefficient, varied with walking speed (Bovonsunthonchai et al., 2012). It has been observed that during walking at comfortable speed, the coordination between non-paretic upper and paretic lower limb was better compared to the coordination of the paretic upper and non-paretic lower limbs. The opposite was observed when participants walked with fast speeds. The inversion was attributed mainly to the paretic upper limb presenting better adjustments to the non-paretic lower limb at faster speeds. As far as SBT walking is concerned, it appears that the movements of both upper limbs are particularly influenced by the leg walking on the faster belt in healthy individuals (MacLellan et al., 2013). This effect of speed on coordination underlines the relevance of analysing upper and lower limb coordination during and after walking with unequal belts speeds and its influence on step length symmetry in individuals post-stroke.

Furthermore, the analysis of hip, pelvis, and trunk movements is recommended for future analyses with SBT protocols to gain insight into possible strategies in the frontal plane of motion underlying improvements in step length asymmetry. This suggestion is based on the presence of compensatory strategies, such as paretic hip hiking and circumduction, described in post-stroke gait (Stanhope et al., 2014). Therefore, it is reasonable to assume that potential strategies in frontal plane biomechanics and muscle activity also exist during SBT walking and could affect the SBT training-induced reduction in step length asymmetry during walking over ground.

6.5.2 Recommendations for Training Studies

6.5.2.1 Parameters Analyzed

As described in the literature review, within the MUR, it has been possible to find some explanations for the limited capacity of individuals post-stroke to adjust their step length in the short term after SBT walking (Lauzière et al., 2016). The initial level of effort used in paretic plantarflexor muscles by participants in that study influenced the capacity to increase non-paretic step length during the post-adaptation period. Based on these results Lauzière et al. (2016) concluded that some individuals are limited by their plantarflexor weakness in improvements of step length symmetry whereas others show better capacities to adjust step length due to the residual strength left in this muscle. Therefore, the analysis of the plantarflexor MUR in the scope of a training study would allow exploring the aformentioned hypotheses on different subgroups based on compensatory strategies and the different training requirements needed to achieve optimal improvements in step length symmetry post-stroke in these different subgroups.

In the context of a subgroup analysis, an additional recommendation would be to consider conducting statistical analysis that separates data based on the paretic and non-paretic side instead of the slow and fast side. As mentioned, in the analyses of this thesis, the global aim was to test the effect of belt speed (i.e., slow, fast) on individuals with different directions of asymmetries in order to give recommendations for the future use of the protocol in a clinical setting. Indeed, we demonstrated that for the majority of participants, the increased step length was pronounced on the leg trained on the fast side compared to the leg trained on the slow side. Nevertheless, using a larger group of participants and performing subgroup analyses based on initial asymmetry (shorter paretic step or non-paretic step) would enable the evaluation of effects of training on biomechanics of the paretic and non-paretic legs. Using such analysis, the role of initial step length asymmetry and functional deficits in SBT-induced effects might be more evident. For example, the analysis of individuals post-stroke with different initial step length asymmetries showed that the group with shorter non-paretic step length weakness of paretic plantarflexors compensated with the non-paretic plantarflexors and knee extensors (Allen et al., 2011). However, in the group with shorter paretic step length (n = 4), no consistent results on such compensations were found. Therefore, the authors hypothesized that the direction of step length asymmetry observed in their post-stroke participants was associated with the compensatory mechanisms.

6.5.2.2 Study Protocol

Three principal recommendations can be offered considering future analyses investigating the long-term effects on gait parameters of a SBT training protocol. The three recommendations are the following: first, the use of a double baseline testing of the step length asymmetry prior to the beginning of the training; second to adjust the speed of the slower belt for each training session and third to use a larger sample size. The first two recommendations were mentioned previously in the sections study limitations (6.4.1.2) and generalization (6.4.3) and will, therefore, not be commented on further. Considering the third recommendation, a sample size calculation was conducted. Within our training study, we aimed to provide information for the study designs and methods of larger controlled trials. For that purpose, a sample of at least 20 participants post-stroke is suggested in order to have a 0.85 probability to state that the effect of time (pre- post- and follow-up) is not influenced by a small sample size. This suggestion is based on a sample size calculation using the variances and average differences in EMG activity of the cross-sectional analysis and is appropriate for a 1-factor analysis of variance without control group (Chow, 2011; Kadam & Bhalerao, 2010).

6.6 Clinical Implications

The previous sections discussed the findings of the studies presented in this thesis and provided recommendations for future research on the short- and long-term effects of SBT walking on step length, lower limb muscle activity and biomechanics in individuals who suffered a cerebral stroke. In the following sections, the more clinically relevant findings will be discussed in relation to therapeutic interventions. In addition, some key messages for clinicians will be provided.

6.6.1 The Relevance of Assessing and Training Step Length Asymmetry

The present work becomes critical when adjustments in step length and step length symmetry are a goal in gait recovery post-stroke. According to the literature, gait speed and endurance are relevant factors for the capacity to walk in the community (Andrews et al., 2010; Salbach et al., 2013). Nevertheless, step length asymmetry is a frequently observed deficit post-stroke, albeit less than temporal asymmetries. Step length asymmetry was found in 33% - 43% of ambulant stroke survivors (Patterson et al., 2010a; 2008a) and in 12/16 participants (75%) of our cross-sectional analysis. In addition, 79.3% of stroke survivors with gait deficits (n = 58) reported that it is very important to regain a “normal appearance” of their gait pattern (Bohannon et al., 1991). However, step length symmetry is not only an aspect of appearance considering its associations with ankle plantarflexor spasticity (Lin et al., 2006) and weakness in forward propulsion during gait (Allen et al., 2011). Therefore, we concluded that more effort should be put forward towards the quantification and treatment of step length asymmetry during gait recovery for ambulant stroke survivors. The use of a paper carpet and pencils affixed to the heel of the shoe is a practical and reliable method to quantify step lengths. This method provided step length symmetry ratios which were in accordance with the ratios measured during the 3D-motion analysis (paper #3), providing a quick and simple manner to assess symmetry without needing costly equipment. As far as the treatment of step length asymmetry is concerned, the results of this doctoral work support previous findings of successful short- and long-term improvements when using the error-augmentation-based principle on a SBT. Recommendations for the use of this protocol will be provided next.

6.6.1.1 Recommendations for the Use of the SBT Protocol

Taking into account the results from paper #1 and #2, it can be agreed that improvements in step length symmetry are most likely achieved when the side with the shorter step length is trained on the faster running belt. Based on our findings and those of previous studies, we suggest the use of a 2:1 ratio between belt speeds. In addition, we recommend that the slower belt should be set at the patient’s comfortable speed. As far as repeated exposure to the SBT protocol is concerned, we suggest an adjustment of the comfortable speed for each training session, particularly when concomitant effects on patient's walking speed is one of the goals of the gait training. However, this last recommendation is still lacking in evidence and requires further research using controlled trials comparing the adjustment of confortable speed to a protocol without adjustments of comfortable speed. Regardless, we strongly recommend a systematic assessment of step length asymmetry before each training session in order to estimate the number of sessions required to attain sufficient reduction in step length asymmetry and avoid the inversion of the initial asymmetry. Nonetheless, the question remains whether a reduction in step length asymmetry leads to a reduction in the aforementioned lower limb deficits (e.g., muscle weakness). Within this doctoral work, the results demonstrated that the dorsiflexors (TA) can be considered as relevant for feedforward control during locomotor adaptation and plantarflexors are most likely strong contributors to the step length after-effects. Therefore, improvement of distal lower limb function (e.g., strength or range of motion) is likely to serve as a facilitator for improvements in step length asymmetry but requires controlled trials to investigate the effects of SBT walking combined with or without, for example, strength training to confirm this hypothesis. The current doctoral work showed that not only is it a safe approach, but it is also simple to implement with the belt speeds only needing to be doubled on the side with the shorter step length and when repeated in the context of training, it can successfully reduce step length asymmetry and increase walking speed most likely by improving distal lower limb muscle function.

Chapter 7. Conclusion

This doctoral work investigated lower limb muscle activity and the biomechanics underlying immediate and long-term effects of SBT walking on step length in individuals who had suffered a cerebral stroke. At first, the effect of a single-session with SBT walking was tested on 16 stroke survivors and 10 healthy controls (paper #1 and supplementary data). Previous findings were reproduced showing that six minutes of SBT walking led to after-effects characterized by a reduced step length after slow-belt walking and an increase after fast-belt walking in both groups. Individuals post-stroke displayed consistent changes in muscle activity underlying these after-effects. Changes in step length and symmetry were associated particularly with those in distal lower limb muscles, which is in accordance with previous studies and the research hypothesis. Based on these findings, it was concluded that repeated exposure to this protocol will lead to a reduction in step length asymmetry and improvements of distal lower limb muscle activity.

Indeed, repeated exposure to the error-augmentation-based protocol led to a successful reduction in step length asymmetry during walking over ground (paper #2). The effects persisted to 1-month post-training. Based on the concomitant long-term improvements in gait ability, the present protocol is likely to be advantageous compared to previous SBT protocols when walking speed and capacity are additional goals to step symmetry in gait recovery. As far as the role of distal lower limb muscles is concerned, even though there was a lack of statistical significance, the effect size estimations of changes in muscle activity and joint moments supported the hypothesis that the plantarflexors play a major role in adjustments of step length after repeated exposure to SBT walking. However, in contrast to the findings from cross-sectional studies (paper #1; Lauzière et al, 2014a), plantarflexors increased joint moments bilaterally and particularly on the fast side. Thus, repeated exposure to fast belt walking did not lead to negative effects on the paretic plantarflexor muscles as hypothesized from previous work (Lauzière et al., 2014a).

A large variability was nonetheless found in muscle activity underlying improvements in step length symmetry post-training during walking over ground. The large heterogeneity of muscle activity was not expected considering the consistent changes reported in the cross-sectional analysis. The difference in walking conditions (treadmill and over ground) is potentially responsible for this discrepancy with over-ground walking enabling a larger degree of freedom in muscle synergies or compensatory strategies to achieve the same task. Overall, the nature of our training study as a pilot approach emphasizes the need for future research using larger sample sizes, control-trials and larger numbers of training sessions to confirm the present findings.

Finally, the SBT protocols were well tolerated by the participants and did not induce severe adverse events. Three physiotherapists considered the protocol and the use of the SBT as an easy-to-learn tool that is ready for implementation in a clinical setting. Thus, we strongly recommend that the studies presented in this thesis be followed up by future research on the long-term effects of repeated SBT walking considering the potential of this intervention to improve gait symmetry and speed in individuals post-stroke.

Chapter 8. References

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Appendix I

Acquisition and Treatment of Electromyography Signals in the Scope of the Thesis "Locomotor After-effects and Rehabilitation of Gait in Individuals with Hemiparesis".

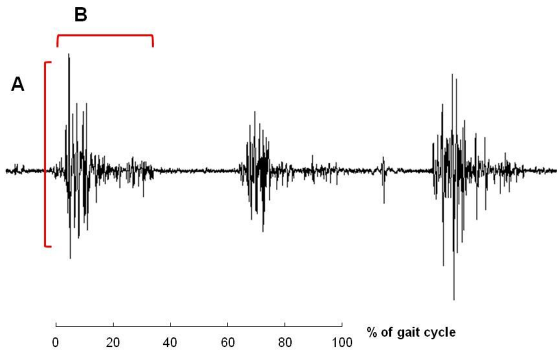
Preamble

This appendix will provide a detailed description of the acquisition and treatment of EMG signal in line with the two studies contained in this thesis. To repeat, during the first study EMG signals were obtained from 16 individuals post-stroke and 10 healthy controls during different walking conditions on a SBT for cross-sectional analysis (Figure 4-2). A detailed description of the protocol can be found in the main manuscript section 4.1.5. In the second study EMG signal were acquired during three different evaluation days (baseline, post- and follow-up evaluation) from 12 individuals post-stroke. Participants' muscle activity was registered during walking over ground at comfortable speed.

Amplitude and timing analyses were performed on the EMG data. Modifications in the frequency features were not considered within the framework of this thesis. Several EMG parameters can be quantified, including the amount of electrical activity a muscle emanates during the contraction (amplitude), as the duration of the muscle's activity and its onset (Figure A1). Furthermore, the activation profile (shape) can be compared among muscles or between sides and be used as an indicator for coordination among muscles (Figure A2). Thus, EMG analyses used in this thesis were performed in order to obtain information about:

A. EMG signal amplitude (intensity of a signal),
B. Duration of active signal (temporal) and
C. Signal profiles (shape, coordination).

The following explanations of pre- and offline EMG treatment apply for the EMG acquisition and treatment performed in both studies.



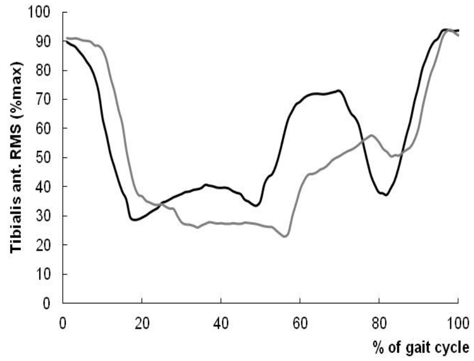


Figure A1. Example of the electrical activity of a lower limb muscle registered during walking showing the amplitude of activity (A) and the burst duration (B) of a signal normalized to the gait cycle (0 - 100%).

Figure A2. Illustration of an activation profile of the paretic (black) and (non-paretic) right dorsiflexor muscle (tibialis anterior) from an individual with stroke during one gait cycle. The profile was obtained by the average mean RMS from 10 gait cycles of comfortable walking.

EMG Detection and Pre-processing

Muscle activity was quantified bilaterally for the following six lower limb muscles: the uniarticular muscles tibialis anterior (TA), vastus lateralis (VL) and gluteus medius (GM) and the biarticular muscles gastrocnemius lateralis (GL), semitendinosus (ST) and rectus femoris (RF). Signal detection and pre-processing was conducted according to the standards proposed by the International Society of Electrophysiology and Kinesiology (ISEK) (Merletti, 1999). More precisely, self-adhesive surface electrodes (Ag/AgCl, Ambu Blue Sensor M) were placed over the muscle belly of the target muscle after skin-preparation with a standardized inter-electrode distance of 2.0 cm (DelSys Incorporated, 1996). For both projects the wireless system TeleMyo DTS (Noraxon Inc., USA) was used. Using the 16-channel surface EMG Direct Transmission System signals were acquired at a sampling frequency of 1200 Hz, pre-amplified by 500 with a Common-Mode Rejection Ratio (CMRR) of 115 dB and an input impedance of 100 MOhms.

Prior to the registration, palpation and a series of isometric contractions of the target muscles were used for signal validation. During signal registration, the raw signal was displayed on a screen. This allowed the evaluators to verify signal abnormalities, such as signal loss or cut-offs. Any trials with abnormalities were noted for offline treatment and when possible repeated.

Offline Treatment

Visual Inspection

Prior to the amplitude processing, the raw signal was visually inspected in order to identify amplitude artefacts and signal cut-offs. The signal was therefore inspected on two levels: 1) the level of amplitude during a sequence of registration and 2) the frequency components of the signal (frequency spectrum). EMG activity during a gait cycle with large sequences of signal cut-offs were excluded from analysis. As far as the amplitude is concerned, spike-shaped burst excursion were observed. This phenomenon is considered as an amplitude artefact (O'Keeffe et al, 2001) and illustrated in Figure A3.

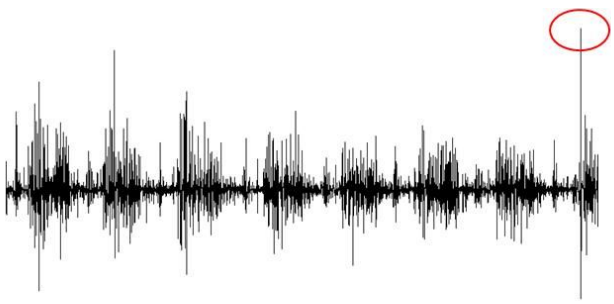


Figure A3. Illustration of several signal bursts whereas one presents a spike-shaped burst excursion encircled with red.

Regarding the frequency spectra of signals, gaps of signal during certain firing frequencies were observed (Figure A4). Comparing data among participants revealed that the signal loss occurred at different frequencies, varied among individuals and was not constantly found in the same muscle. These gaps occurred in a systematic and periodic manner which allowed developing a correcting algorithm. A more detailed description of the mechanisms of artefact and shift correction will be provided in the following section.

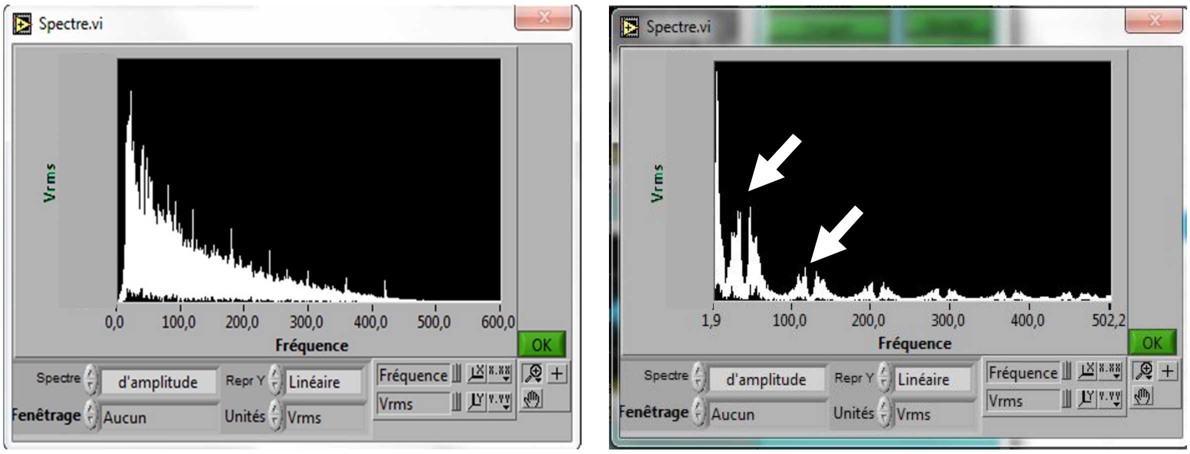


Figure A4. Screen captures of frequency spectres without (left-hand) and with signal gaps (right-hand). The left-hand image represents a usual and correct frequency spectrum of a lower limb muscle during walking. The arrows on the right-hand image indicate the signal gaps. For both images, x-axes represent the firing frequency and the y-axes amplitude of the electrical activity. Thus in the left-hand image, most of the energy is found in the frequency range 10 - 250Hz.

<u>Treatment for Amplitude Artefacts</u>

After visual inspection the affected signals were treated first for artefacts and subsequently for shifts. To account for amplitude artefacts the signal was corrected by means of a software-based two-stage peak detection algorithm described and recommended by O'Keeffe et al. (2001). The O'Keeffe-Algorithm allows the removal of movement artefacts without cutting "real" signal. With a shifting window (250 ms) the largest spike excursion is searched throughout the entire target signal (Figure A5 A). The algorithm detects the highest spike during the sequence the window was shifted. The highest spike will be considered as 100%. To remove and replace this artefact, a cut-off threshold has to be determined. Therefore, the O'Keeffe group suggests a high threshold of 0.5 (O'Keeffe et al, 2001). Using 0.5 as a limit leads to a cut out of all the spikes exceeding 50% of the highest spike, previously determined. The onset and end of the artefact are determined based on a so-called low threshold (Figure A5 B) which is close to zero (or baseline). The section defined as the artefact by the O'Keeffe-Algorithm gets replaced by the general mean value. Image C in Figure A5 illustrates the difference between an original sequence including the artefact (blue) and a correction sequence (green). The general mean RMS of eight cycles before filtering was 10.59 (0.60) microvolt (µV) and 11.12 (0.68) µV after filtering. An estimation of the amount (%) of difference between filtered an unfiltered data revealed that the discrepancies ranged between 10 - 30% for the stroke group in the cross-sectional study and slightly smaller discrepancy in the group of healthy controls.

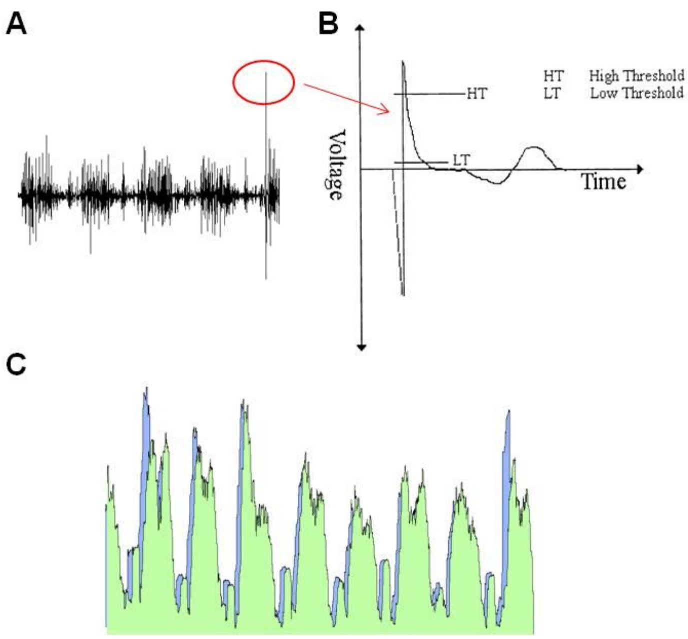


Figure A5. A) raw EMG signal from one subject with amplitude artefacts with peak excursion circled in red. B) Illustration of a single spike. The exponential decay of the spike is characteristic for an artefact. The O'Keeffe algorithm detects and corrects the artefact based on a high (HT) and low threshold (LT) C) Superimposed activation profiles obtained after rectification of the signal presented in A) activation profiles with RMS values without (blue) and with (green) application of the artefact correction (O'Keeffe et al, 2001). Home-based images from an individual post-stroke.

Treatment for Signal Shifts

In case of frequency gaps, the affected signal sequence was corrected by a custom-made LabView based algorithm (LabView 6.1, National instruments Inc.). The careful observation of the raw signal amplitude of the "affected" muscles revealed a pattern of abrupt up and down shifts of the signal to another amplitude level (white dotted line, Figure A6). The constant periodicity of these shifts allowed the development of a detection algorithm to correct them. These signal shifts were identified by the algorithm (red spikes, Figure A6) and corrected by subtracting the amplitude of each shift (green line, Figure A6).

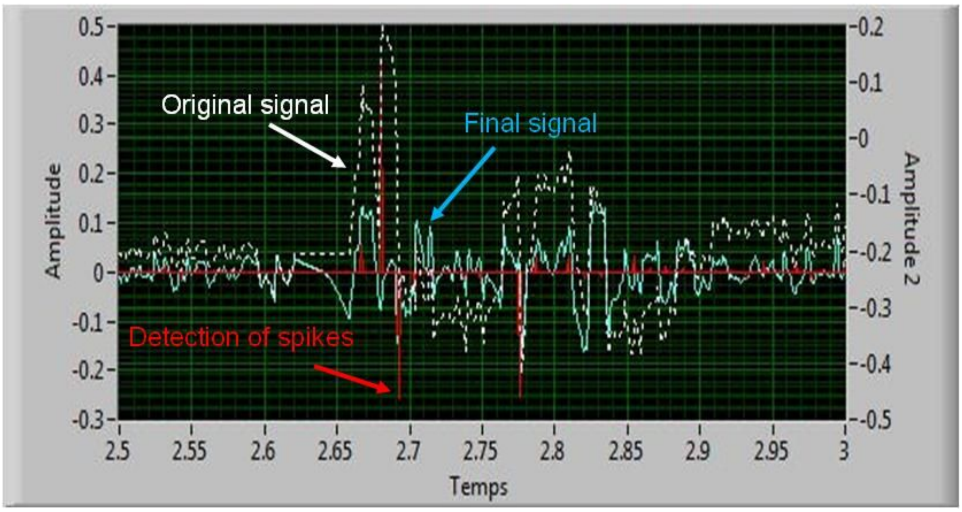


Figure A6. Illustration of strongly zoomed raw signal with signal shifts (white dotted line). The spikes (red) represent the time point of signal shifts. The blue line represents the final, corrected signal.

Signal Filtering

After these corrections, signals were filtered with a band-pass of 20 - 400 Hz during offline processing (De Luca et al, 2010; Knarr, Zeni, & Higginson, 2012; Konrad, 2005). Due to the use of a wireless acquisition system, a signal delay of 440 ms had to be corrected. The 240 ms of delay were predetermined by the Noraxon wireless system and the 200ms were added manually in the analog output receiver in order to avoid signal loss.

Signal Smoothing

After the signal correction and filtering process the signals were transformed into centered running-window root mean square (RMS) values in order to allow an estimation of the signal amplitude. According to previous studies, a 20 ms time window should be used for smoothing signals during fast movements (jumping) and 500 ms for slow, static movements (Konrad, 2005). Therefore, a time window of 250 ms tends to be appropriate for walking tasks at comfortable and fast-comfortable speeds and is recommended by the literature (Konrad, 2005; Merletti, 2001).

Signal Normalization

Time normalization

The RMS values were time normalized to their corresponding gait cycle (0-100%). This allowed interpretation of features of the signal behavior (time point of peak activity or burst duration) during the gait cycle and inter-subject comparison. The time normalization of the smoothed signal resulted in a curve representing the EMG activity profile during one gait cycle (Figure A2).

Amplitude normalization

Amplitude normalization is an essential step in signal analysis for valid inter-subject comparison since amplitude in muscle activity varies substantially among individuals since electrode position and skin condition influencing the signal output (De Luca et al, 2012; Konrad, 2005, Soderberg & Knutson, 2000). A mean dynamic method (Burden, Trew, & Balzopoulos, 2003; Teixeira-Salmela et al, 2012) was applied in order to normalize amplitude. Therefore, the mean RMS EMG was expressed as a percentage of the peak RMS obtained during walking at comfortable speed and calculated. For the cross-sectional analysis the average of the peaks from 10 gait cycles obtained during the corresponding baselines was taken for normalization. For each condition the mean RMS values were normalized with the average of peak values from the baseline period for the corresponding condition. For the second project (training study) the mean RMS values were normalized with the average of peak values observed during five gait cycles from the first walking trial over ground. This mean percentage was the value used for statistical analysis to test the hypothesis of changes of amplitude between periods and conditions. Prior to the signal registration in the training study, participants walked during 2-3 times instrumented on the over ground at comfortable speed to familiarize with the experimental condition.

Analysis of Timing and Interlimb Coordination

Burst duration and timing

The analysis of the timing in EMG signal included the calculation of burst duration and determination of activation onset and offset. To do so, it is recommended by the literature to use a Gaussian distribution of the data of interest (Van Albada & Robinson, 2007). Therefore, the mean RMS from a time normalized signal (one gait cycle) was tested for its normal distribution. Non-normally distributed data is transformed with a statistical approach of logarithm based transformation. Subsequently, from this now normally distributed data a general mean with three standard deviations (3 SD = 99.7%) was calculated. When the signal exceeded three SDs it was considered as active (Mickelborough et al., 2004; Duncan et al., 1990). Figure A7 illustrates the normally distributed general mean and upper and lower limits of the 3SDs (1.085 and 0.926). The green areas in Figure A7 B represent the periods were the signal was considered active. Home-based software (Labview, 2009) was used to calculate the time point during the normalized gait cycle (0 - 100%) at which the signal was considered to be on and off. The activation duration of these time-normalized signals was calculated for stance and swing phase of gait, respectively. The activation durations of the TA, GL, VL and ST during stance and swing were used to calculate coactivation duration between the TA and GL as well as between the VL and ST muscle. For the calculation of coactivation duration time-normalized duration of activity of one muscle (e.g., GL) was subtracted from the one antagonistic muscle (e.g., TA).

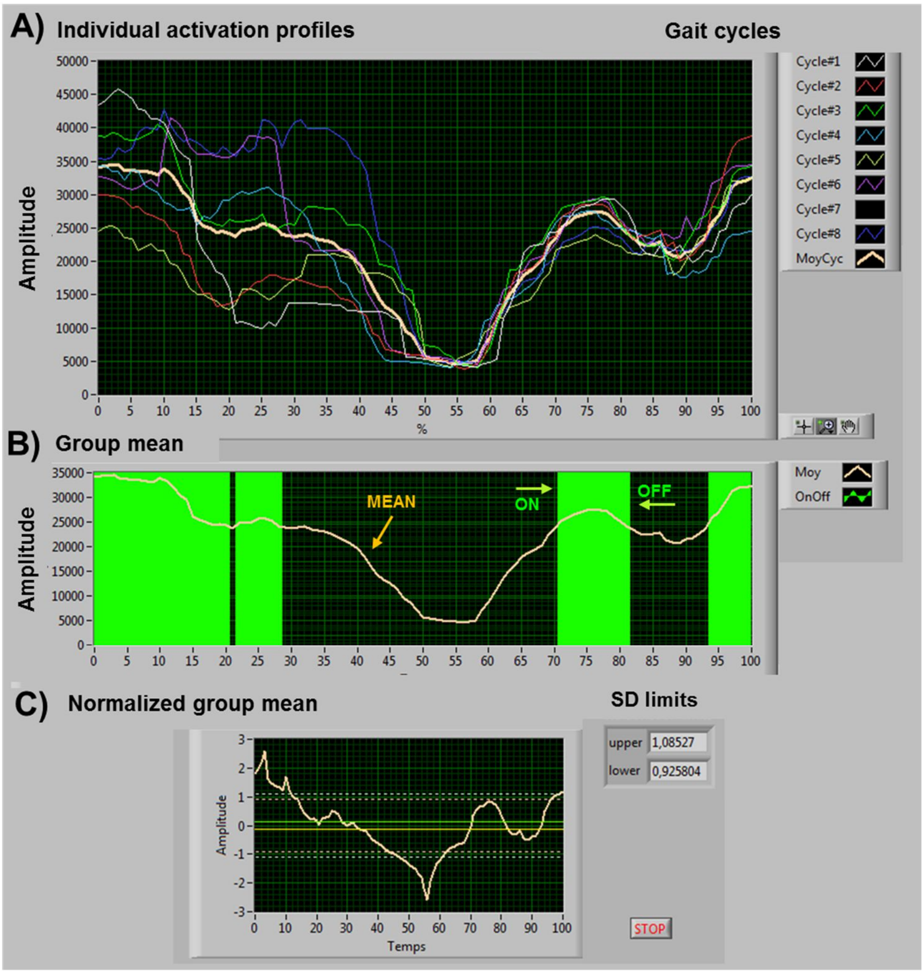
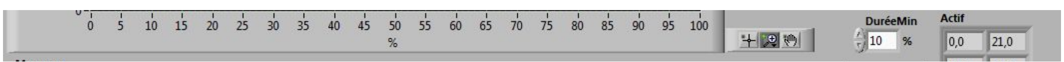


Figure A7. A) Depiction of activation profiles from the dorsiflexors during comfortable walking on the treadmill. Illustrated are individual activation profiles (RMS values) from eight gait cycles and the mean of one participant. B) Shown is the group mean activation profile (pale line). The green areas represent the time where the signal was considered as active. C) presents the normalized mean and the upper and lower limits of standard deviation (SD). Values of the x-axes represent the normalized gait cycle (0 - 100%).

Interlimb Coordination

In order to investigate the coordination between limbs a cross-correlation analysis was performed. The coordination of muscle activity between limbs can be observed by quantifying how similar the shape and timing of muscle activity between left and right sides are during the



different walking conditions. A strong correlation between activity profiles was considered to represent strong inter-limb coordination. Cross-correlation analysis has been considered as a valid method to obtain information about both, the timing and shape of an EMG signal (Nelson-Wong et al, 2009; Wren et al, 2006). The activity profile of each muscle evaluated on the paretic side has been cross-correlated with its non-paretic counterpart (paretic TA with the non-paretic TA). This has been done for each individual and condition separately, such that one profile has been time shifted over the other profile. At the time point of the largest correlation between the curves, the correlation factor was calculated. The paretic and non-paretic activation profile illustrated in Figure A8 presents a correlation coefficient of Rxy = 0.85. Furthermore, a delay of signal onset has been expressed between the two muscles at the time point of the highest correlation between the profiles. The obtained correlation factors (Rxy) represented the interlimb coordination of each muscle group and were used for statistical analysis to compare changes in interlimb coordination for each muscle group among conditions. Therefore, the cross-correlation analysis itself was not considered as a statistical test.

Limitations of the CC Analysis:

As discussed in the study limitations, the cross-correlation analysis did not allow determining the time points at which delay of signals were obtained. Thus in a biphasic muscle, such as the tibialis anterior which is active usually at early stance and swing, it remains unclear whether the delay between sides was defined at the peak activation during stance or during swing.

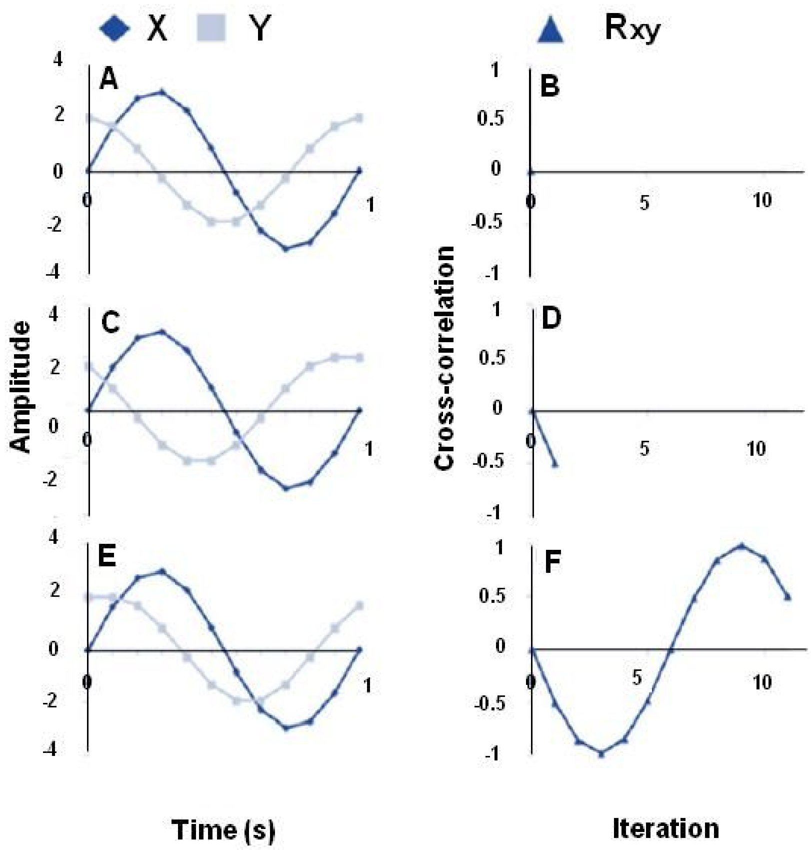


Figure A8. Cross-correlation of two signals (X and Y); Baseline correlation between the two activation profiles (A) contribute to a single point to the cross-correlation function (B). Profile Y gets shifted backwards by one data point (C) which leads to the next correlation value (Rxy) in the function (D). Y is subsequently step-wise shifted backwards until the number of iterations is equivalent to the number of data points in the signal (E) and leads to the final cross-correlation curve (F). Modified from Nelson-Wong et al. (2009).

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Appendix II

Questionnaires Feasibility Study (n = 3)

Questionnaire #1

Perceived Ease of Use of a split-belt treadmill (SBT) and the training protocol

Adapted from Davis FD (1993)

Perceived Ease of Use of a split-belt treadmill (SBT) and the training protocol

1- I find the Split-belt treadmill (SBT) was cumbersome to use.

Strongly agree		Neutral			Strongly disagree	
1	2	3	4	5	6	7

2- Learning to operate the SBT was easy for me.

Strongly agree		Neutral			Strongly disagree	
1	2	3	4	5	6	7

3- Learning to apply the protocol was easy for me (i.e., deciding which belt needs to be increased).

Strongly agree		Neutral			Strongly disagree	
1	2	3	4	5	6	7

4- I found it easy to get the SBT to do what I want it to do.

Strongly agree		Neutral			Strongly disagree	
1	2	3	4	5	6	7

5- It is easy for me to remember how to perform the task (training) using the SBT training protocol.

Strongly agree		Neutral			Strongly disagree	
1	2	3	4	5	6	7

6- It found it a lot of effort to become skillful at using the SBT.

Strongly agree		Neutral			Strongly disagree	
1	2	3	4	5	6	7

7- I found it a lot of effort to become skillful at performing the task (training) using the SBT.

Strongly agree		Neutral			Strongly disagree	
1	2	3	4	5	6	7

8- Assuming the SBT would be available to me, I predict that I will use it on a regular basis in the future.

Strongly agree		Neutral			Strongly disagree	
1	2	3	4	5	6	7

Questionnaire #2

PERFECT Tool Questionnaire (Section 3)

Professional Evaluation & Reflection on Change Tool

Adapted from PERFECT Tool Questionnaire; McGill University (Menon et al., 2010).

PERFECT (Section 3)

DEMOGRAPHICS:

1. In which age group would you belong to?

   21-25 □ 26-30 □ 31-35 □ 36-40 □ 41-50 □ 51-60 □ Over 60 □

Gender: □ M □ F

Do you work full-time or part-time? □ Full-time □ Part-time

How many years have you been working as a physiotherapist? _______

How many years have you been working in neuro-rehabilitation? _______

Type of work setting: Rehabilitation Outpatient

Section 3: Treatment practices

3a) Think of your clinical practice over the **past six months**, please describe any changes you have made with respect to your treatment practices?

I. ____

II. ____

3b) Now, think of your clinical practice over the past year, please describe any changes you have made with respect to your treatment practices that you have not already told us about?

I. ____

II. ____

III. ____

*if no changes were mentioned in 3a or 3b, skip to question 3f

3c) Now I want you to think of (refer to each change listed in 3a or 3b). What were the reason(s) for this change in treatment practice?

**If you are not sure about the question, please see the given examples at the end of the questionnaire*

Past six months

I. ____

II. ____

III. ____

Past year

I. ______________________________

II. ______________________________

III. ______________________________

3d) Now I want you to think of (refer to each change listed in 3a or 3b). What, if anything, helped bring about this change in treatment practice?

**If you are not sure about the question, please see the given examples at the end of the questionnaire*

Past six months

I. ______________________________

II. ______________________________

III. ______________________________

Past year

I. ______________________________

II. ______________________________

III. ______________________________

3e) Now I want you to think of (refer to each change listed in 3a or 3b). What, if anything, made it difficult to bring about this change in treatment practice?

**If you are not sure about the question, please see the given examples at the end of the questionnaire*

I. ____________________

II. ____________________

III. ____________________

Past year

I. ____________________

II. ____________________

III. ____________________

3f) Now, think about your treatment practices over the past year. Given an ideal world is there anything you would have changed?

I. ____________________

II. ____________________

III. ____________________

*If no desired changes were mentioned in 3f, skip to question 4a

3g) What, if anything, would have made it difficult to bring about this change? (refer to each change listed in 3f).

**If you are not sure about the question, please see the given examples at the end of the questionnaire*

I. ____________________

II. ____________________

III. ____________________

Examples for responses :

3c: Examples of reasons for change may include having attended a continuing education course, acquired new knowledge from a professional journal, attended a conference, heard suggestions from colleagues, etc.

3d: Some examples of things that may help bring about change are self-motivation, departmental funding, support from supervisor, etc.

3e: Some examples of things that may make it difficult to bring about change are lack of departmental funding, busy schedule, lack of knowledge, etc.

3g: Some examples of things that may make it difficult to bring about change are lack of departmental funding, busy schedule, lack of knowledge, etc.

Questionnaire #3

Organizational Readiness for Implementing Change (ORIC)

Adapted from Shea et al. (2014)

Organizational Readiness for Implementing Change (ORIC)

Fill out as a team

Note: The term change is representing the use of a split-belt treadmill protocol in gait rehabilitation for individuals post-stroke.

1	2	3	4	5
Disagree	Somewhat Disagree	Neither Agree nor Disagree	Somewhat Agree	Agree

1. People who work here feel confident that the organization can get people invested in implementing this change.	1	2	3	4	5
2. People who work here are committed to implementing this change.	1	2	3	4	5
3. People who work here feel confident that they can keep track of progress in implementing this change.	1	2	3	4	5
4. People who work here will do whatever it takes to implement this change.	1	2	3	4	5
5. People who work here feel confident that the organization can support people as they adjust to this change.	1	2	3	4	5
6. People who work here want to implement this change.	1	2	3	4	5
7. People who work here feel confident that they can keep the momentum going in implementing this change.	1	2	3	4	5
8. People who work here feel confident that they can handle the challenges that might arise in implementing this change.	1	2	3	4	5
9. People who work here are determined to implement this	1	2	3	4	

change.	5			
10. People who work here feel confident that they can coordinate tasks so that implementation goes smoothly.	1 5	2	3	4
11. People who work here are motivated to implement this change.	1 5	2	3	4
12. People who work here feel confident that they can manage the politics of implementing this change.	1 5	2	3	4

Appendix III

Ethics Certification for the Cross-sectional Study

Comité d'éthique de la recherche
des établissements du CRIR

Certificat d'éthique

Par la présente, le comité d'éthique de la recherche des établissements du CRIR (CÉR) atteste qu'il a évalué, lors de sa réunion du 14 juin 2011, le projet de recherche **CRIR-616-0411** intitulé:

« Comparaison de la marche asymétrique et symétrique chez les personnes hémiparétiques chroniques ».

Présenté par: **Sylvie Nadeau, Ph.D.**
Séléna Lauzière, M.Sc., pht
Carole Miéville, M.Sc

Le présent projet répond aux exigences éthiques de notre CÉR. Le Comité autorise donc sa mise en œuvre sur la foi des documents suivants :

- **Lettre d'introduction datée du 30 mai 2011 ;**
- **Formulaire A daté du 26 avril 2011 ;**
- **Formulaire d'évaluation de l'Institut de réadaptation Gingras-Lindsay de Montréal, daté du 5 mai 2011, mentionnant que le projet est acceptable sur le plan de la convenance institutionnelle ;**
- **Grille d'évaluation scientifique du projet de recherche datée du 16 mai 2011 ;**
- **Budget ;**
- **Protocole de recherche intitulé « Comparaison de la marche asymétrique et symétrique chez les personnes hémiparétiques chroniques » ;**
- **Formulaire de consentement destiné aux participants hémiparétiques (versions anglaise et française du 17 juin 2011) ;**
- **Formulaire de consentement destiné aux participants en santé (version française du 17 juin 2011) ;**
- **Lettre d'invitation à participer à une étude pour le recrutement de personnes hémiparétiques suite à un AVC ;**
- **Cahier d'évaluation clinique.**

Ce projet se déroulera dans le site du CRIR suivant : **Institut de réadaptation Gingras-Lindsay de Montréal**

Ce certificat est valable pour un an. En acceptant le présent certificat d'éthique, le chercheur s'engage à :

1. Informer, dès que possible, le CÉR de tout changement qui pourrait être apporté à la présente recherche ou aux documents qui en découlent (Formulaire M) ;
2. Notifier, dès que possible, le CÉR de tout incident ou accident lié à la procédure du projet ;

3. Notifier, dès que possible, le CÉR de tout nouveau renseignement susceptible d'affecter l'intégrité ou l'éthicité du projet de recherche, ou encore, d'influer sur la décision d'un sujet de recherche quant à sa participation au projet ;
4. Notifier, dès que possible, le CÉR de toute suspension ou annulation d'autorisation relative au projet qu'aura formulée un organisme de subvention ou de réglementation ;
5. Notifier, dès que possible, le CÉR de tout problème constaté par un tiers au cours d'une activité de surveillance ou de vérification, interne ou externe, qui est susceptible de remettre en question l'intégrité ou l'éthicité du projet ainsi que la décision du CÉR ;
6. Notifier, dès que possible, le CÉR de l'interruption prématurée, temporaire ou définitive du projet. Cette modification doit être accompagnée d'un rapport faisant état des motifs à la base de cette interruption et des répercussions sur celles-ci sur les sujets de recherche ;
7. Fournir annuellement au CÉR un rapport d'étape l'informant de l'avancement des travaux de recherche (formulaire R) ;
8. Demander le renouvellement annuel de son certificat d'éthique ;
9. Tenir et conserver, selon la procédure prévue dans la *Politique portant sur la conservation d'une liste des sujets de recherche*, incluse dans le cadre réglementaire des établissements du CRIR, une liste des personnes qui ont accepté de prendre part à la présente étude ;
10. Envoyer au CÉR une copie de son rapport de fin de projet / publication ;
11. En vertu de l'article 19.2 de la *Loi sur les services de santé et les services sociaux*, obtenir l'autorisation du Directeur des services professionnels de l'établissement sollicité avant d'aller consulter les dossiers des usagers de cet établissement, le cas échéant.

[redacted]

Président du CER

Date d'émission
17 juin 2011

Appendix IV

Ethics Certification for the Training Study

Comité d'éthique de la recherche
des établissements du CRIR

Certificat d'éthique

Par la présente, le comité d'éthique de la recherche des établissements du CRIR (CÉR) atteste qu'il a évalué, lors de sa réunion du 10 juin 2014, le projet de recherche **CRIR-951-0314** intitulé:

« Effects of Repeated Split-belt Treadmill walking on Gait Symmetry and Muscle Activity in Chronic Stroke Survivors: a Pilot Study ».

Présenté par: **Martina Betschart, candidate au Ph.D.**
Sylvie Nadeau, Ph.D., pht
Bradford J. McFadyen, Ph.D.
Mylène Lajeunesse-Langdeau, étudiante en physiothérapie

Le présent projet répond aux exigences éthiques de notre CÉR. Le Comité autorise donc sa mise en œuvre sur la foi des documents suivants :

- **Lettre d'introduction datée du 24 mars 2014 ;**
- **Formulaire A;**
- **Formulaire d'évaluation du Centre de réadaptation Constance-Lethbridge, mentionnant que le projet est acceptable sur le plan de la convenance institutionnelle ;**
- **Formulaire d'évaluation de l'Institut de réadaptation Gingras-Lindsay de Montréal mentionnant que le projet est acceptable sur le plan de la convenance institutionnelle ;**
- **Grille d'évaluation scientifique du projet de recherche datée du 13 mai 2014 ;**
- **Protocole de recherche ;**
- **Formulaire de consentement (versions anglaise et française du 16 jauin 2014) ;**
- **Questionnaire « Work-sheet Clinical Evaluation »;**
- **Questionnaire « Six-Minute Walt Test »;**
- **Cahier d'évaluation clinique.**

Ce projet se déroulera dans le site du CRIR suivant :

- **Centre de réadaptation Constance Lethbridge**
- **Institut de réadaptation Gingras-Lindsay de Montréal**

Ce certificat est valable pour un an. En acceptant le présent certificat d'éthique, le chercheur s'engage à :

1. Informer, dès que possible, le CÉR de tout changement qui pourrait être apporté à la présente recherche ou aux documents qui en découlent (Formulaire M) ;

2. Notifier, dès que possible, le CÉR de tout incident ou accident lié à la procédure du projet ;
3. Notifier, dès que possible, le CÉR de tout nouveau renseignement susceptible d'affecter l'intégrité ou l'éthicité du projet de recherche, ou encore, d'influer sur la décision d'un sujet de recherche quant à sa participation au projet ;
4. Notifier, dès que possible, le CÉR de toute suspension ou annulation d'autorisation relative au projet qu'aura formulée un organisme de subvention ou de réglementation ;
5. Notifier, dès que possible, le CÉR de tout problème constaté par un tiers au cours d'une activité de surveillance ou de vérification, interne ou externe, qui est susceptible de remettre en question l'intégrité ou l'éthicité du projet ainsi que la décision du CÉR ;
6. Notifier, dès que possible, le CÉR de l'interruption prématurée, temporaire ou définitive du projet. Cette modification doit être accompagnée d'un rapport faisant état des motifs à la base de cette interruption et des répercussions sur celles-ci sur les sujets de recherche ;
7. Fournir annuellement au CÉR un rapport d'étape l'informant de l'avancement des travaux de recherche (formulaire R) ;
8. Demander le renouvellement annuel de son certificat d'éthique ;
9. Tenir et conserver, selon la procédure prévue dans la *Politique portant sur la conservation d'une liste des sujets de recherche*, incluse dans le cadre réglementaire des établissements du CRIR, une liste des personnes qui ont accepté de prendre part à la présente étude ;
10. Envoyer au CÉR une copie de son rapport de fin de projet / publication ;
11. En vertu de l'article 19.2 de la *Loi sur les services de santé et les services sociaux*, obtenir l'autorisation du Directeur des services professionnels des établissements sollicités avant d'aller consulter les dossiers des usagers de cet établissement, le cas échéant.

[redacted]
Président du CÉR

Date d'émission
16 juin 2014

Appendix V

Consent Form for Participants with Stroke (Cross-sectional Study)

(Version in French)

INSTITUT DE RÉADAPTATION
Gingras-Lindsay-de-Montréal

CRIR
Centre de recherche interdisciplinaire en réadaptation du Montréal Métropolitain

Formule de consentement pour votre participation à un projet de recherche

TITRE DU PROJET

Comparaison de la marche asymétrique et symétrique chez les personnes hémiparétiques chroniques.

RESPONSABLE

Sylvie Nadeau, pht, Ph.D	Chercheure, Centre de recherche interdisciplinaire en réadaptation (CRIR), Institut de réadaptation Gingras-Lindsay de Montréal (IRGLM), Laboratoire de pathokinésiologie et d'analyse des activités fonctionnelles. Professeure titulaire à l'Université de Montréal, École de réadaptation. Chercheure responsable du projet.

CO-CHERCHEURS

Cyril Duclos, Ph. D.	Chercheur, CRIR, IRGLM
Séléna Lauzière, pht, M. Sc.	Candidate au doctorat, CRIR, IRGLM
Carole Miéville, M. Sc.	Candidate au doctorat, CRIR, IRGLM
Rachid Aissaoui, Ing, Ph. D.	Chercheur associé au CRIR, site IRGLM et École de technologie supérieure

PRÉAMBULE

Nous vous demandons de participer à un projet de recherche qui implique différentes évaluations se déroulant au laboratoire de pathokinésiologie au 4^{e} étage de l'IRGLM. Ces évaluations visent à étudier l'effet de la marche symétrique sur la stabilité, le coût énergétique et les niveaux d'effort musculaire produits aux membres inférieurs.

Avant d'accepter de participer à ce projet de recherche, veuillez prendre le temps de comprendre et de considérer attentivement les renseignements suivants.

Ce formulaire de consentement vous explique le but de cette étude, les procédures, les avantages, les risques et inconvénients, de même que les personnes avec qui communiquer au besoin.

Le présent formulaire de consentement peut contenir des mots que vous ne comprenez pas. Nous vous invitons à poser toutes les questions que vous jugerez utiles au chercheur et aux autres membres du personnel affecté au projet de recherche et à leur demander de vous expliquer tout mot ou renseignement qui n'est pas clair.

DESCRIPTION DU PROJET ET DE SES OBJECTIFS

Des problèmes locomoteurs sont fréquemment rencontrés chez les personnes ayant une hémiparésie suite à un accident vasculaire cérébral (AVC). Le plus souvent, leur performance est caractérisée par une diminution de la vitesse de marche et par une asymétrie des mouvements entre les deux jambes. Cependant, sur demande, ces personnes peuvent habituellement effectuer la tâche à une vitesse plus élevée et de façon plus symétrique que ce qu'elles font de façon naturelle. La question qui nous intéresse ici est de comprendre pourquoi les personnes hémiparétiques utilisent une stratégie asymétrique alors qu'elles ont les capacités de marcher plus symétriquement. Les résultats de nos travaux antérieurs suggèrent que la perception de l'effort produit afin de réussir la tâche pourrait expliquer la stratégie de mouvements choisie. L'objectif du présent projet est de déterminer les effets réels et perçus d'une marche symétrique sur la stabilité posturale, le coût énergétique et les niveaux d'effort musculaire afin de déterminer si ces facteurs sont explicatifs de la performance motrice mesurée en laboratoire et en clinique chez les individus hémiparétiques. Un objectif secondaire est d'évaluer l'effet de la marche prolongée sur la symétrie du patron de marche, la stabilité, l'effort global et les niveaux d'effort musculaire.

Pour répondre à ces objectifs, 20 participants avec une hémiparésie chronique consécutive à un AVC unilatéral seront recrutés dans deux établissements de réadaptation: l'Institut de réadaptation Gingras-Lindsay de Montréal (IRGLM) et l'Hôpital de réadaptation Villa Medica.

NATURE ET DURÉE DE LA PARTICIPATION

Cette étude comporte deux séances d'évaluation qui auront lieu dans un intervalle d'une à deux semaines. Toutes les évaluations seront réalisées au laboratoire de pathokinésiologie et d'analyse de tâches fonctionnelles du site IRGLM.

Lors de la **première séance**, qui durera environ trois (3) heures, un(e) physiothérapeute évaluera votre santé, votre condition physique ainsi que votre habileté à réaliser diverses activités fonctionnelles via des questionnaires et différents tests standardisés. Ces tests évalueront vos mouvements au niveau des jambes, votre sensibilité, votre équilibre ainsi que votre capacité à réaliser quelques épreuves fonctionnelles. De plus, votre capacité à réduire l'asymétrie de votre patron de marche de façon volontaire sera évaluée par une simple méthode de calcul utilisant l'empreinte du pas sur le sol. Il est possible que suite aux résultats de l'évaluation clinique, nous constations que vous ne répondez pas totalement au type de participants que nous recherchons pour cette étude. S'il en est ainsi, votre participation s'arrêtera après cette première séance et on vous remettra une indemnité compensatoire couvrant vos frais de transport et de stationnement pour cette visite.

Si vous répondez au type de participants recherchés pour l'étude, vous serez invité(e) à réaliser différents types d'effort avec vos jambes. Ces tests serviront à évaluer votre force musculaire avec un appareil appelé dynamomètre. Il s'agit d'un appareil qui permet de mesurer précisément la force maximale lors de poussées avec différentes parties de vos jambes contre l'appareil. Pour cette évaluation de la force, vous serez assis ou couché et des courroies vous stabiliseront et empêcheront les mouvements de certaines parties de votre corps (voir photo 1). Au total, vous aurez à réaliser environ 76 contractions d'une durée d'environ 5 secondes chacune avec différents muscles de vos jambes avec des repos fréquents.

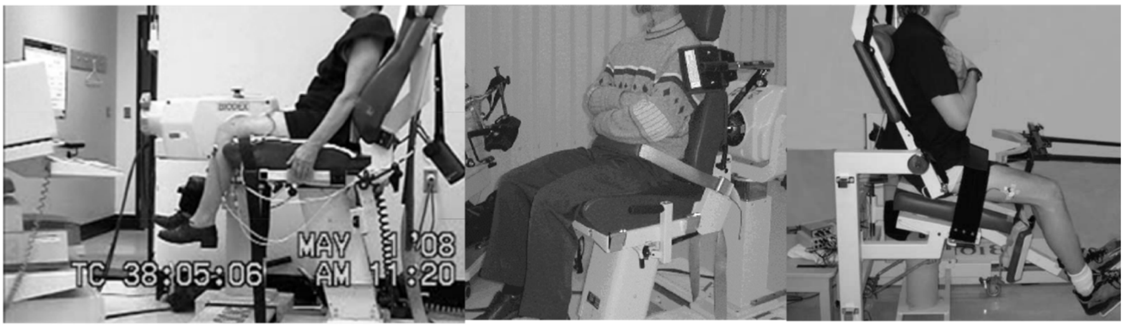

Photo 1. Dynamomètre Biodex et courroies de stabilisation

Finalement, une séance de familiarisation sur le tapis roulant à double courroie sera effectuée avec le port d'un masque nommé Cosmed qui sert à mesurer votre consommation d'oxygène (voir photo 2). Cette familiarisation vous permettra d'expérimenter les différentes conditions de marche utilisées lors de la 2^{e} visite sur le tapis roulant à double courroie (voir photo 3). Comme son nom l'indique, ce tapis roulant possède deux courroies distinctes qui peuvent se déplacer à des vitesses différentes. Ainsi, il permet de faire varier la vitesse de déplacement d'une jambe différemment par rapport à l'autre. Le tapis roulant possède des barres d'appui des deux côtés et également une barre d'appui à l'avant. Ainsi, malgré qu'il vous soit demandé de marcher sur le tapis roulant sans prendre appui avec vos mains, vous pourrez vous stabiliser sur ses barres en cas de déséquilibre. De plus, vous serez encadré en tout temps de deux personnes qui assureront votre sécurité.

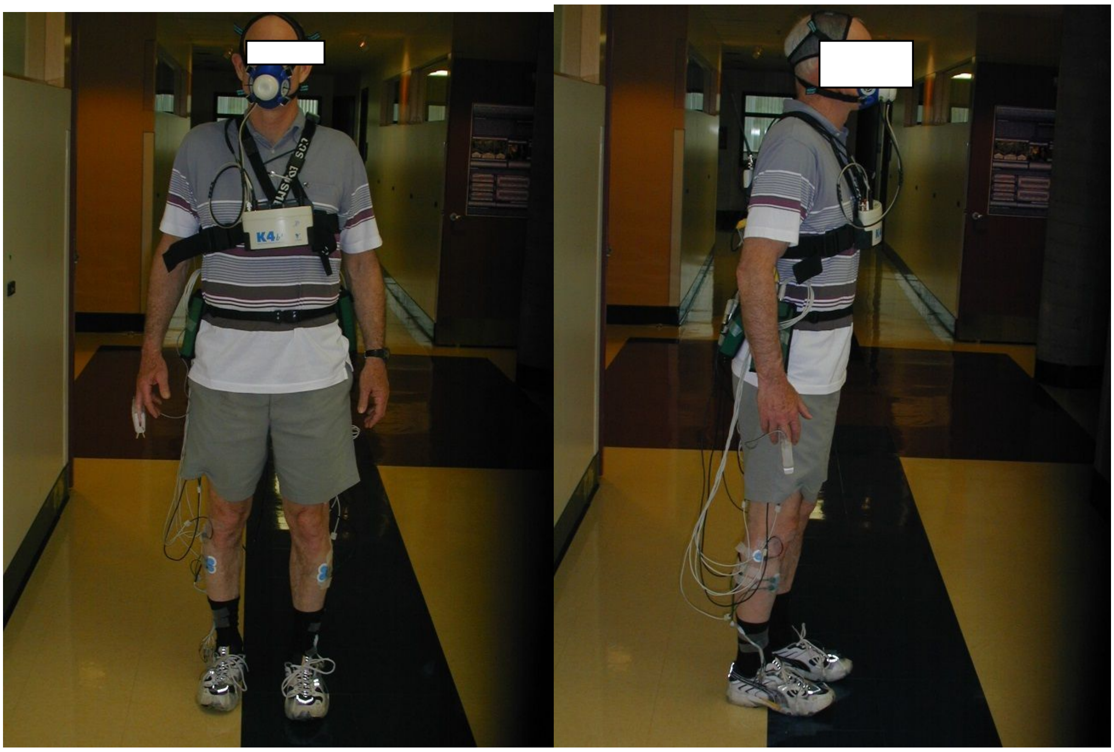

Photo 2. Système d'acquisition des paramètres cardiorespiratoires (COSMED)

Photo 3. Tapis roulant à double courroie

Lors de la **deuxième séance**, qui durera également trois (3) heures, vous aurez à effectuer plusieurs conditions de marche différentes sur le tapis roulant à double courroie. Ainsi, il y aura des conditions où les courroies se déplaceront à la même vitesse et une condition où elles se déplaceront à des vitesses différentes. On vous demandera également de tenter de marcher de façon plus symétrique. Pour ce faire, vous recevrez la consigne de "marchez avec une longueur de pas la plus symétrique possible" et un physiothérapeute donnera des consignes spécifiques telles que "placez le pied gauche/droit plus loin" pour essayer d'obtenir la symétrie la meilleure possible. De plus, il vous sera demandé de marcher lors d'une période prolongée afin d'évaluer les changements dans votre performance motrice lorsque vos muscles sont fatigués. Lors de ces conditions, l'activité de vos muscles sera enregistrée avec des électrodes que nous collerons sur les muscles de vos jambes. Nous mesurerons simultanément les forces que vous produisez sous les pieds à l'aide de plates-formes de forces qui sont situées sous le tapis roulant. Des marqueurs seront collés sur différentes parties de votre corps (pieds, jambes, cuisses, bassin, tronc), pour permettre l'enregistrement de vos mouvements à l'aide d'un système de caméras infrarouges. Tous les essais seront aussi enregistrés à l'aide de deux caméras vidéo afin de nous fournir une image de la manière dont vous exécutez les tâches. Lors de certaines de ces tâches, nous vous demanderons de cocher l'effort, la stabilité et le niveau d'effort musculaire que vous percevez lorsque vous exécutez les diverses tâches locomotrices. Des périodes de repos (2 périodes de repos de 20 minutes) vous seront accordés entre les différentes conditions. Des repos additionnels s'ajouteront au besoin, selon votre endurance physique.

AVANTAGES PERSONNELS POUVANT DÉCOULER DE VOTRE PARTICIPATION

En tant que participant, vous ne retirerez aucun avantage de votre implication au projet de recherche. Par ailleurs, votre participation aura contribué à l'avancement de la recherche dans le domaine de la réadaptation des personnes avec un AVC.

RISQUES POUVANT DÉCOULER DE VOTRE PARTICIPATION

Il est entendu que votre participation à ce projet ne vous fait courir, sur le plan médical, aucun risque que ce soit. Toutefois, dans quelques cas, une irritation cutanée pourrait survenir à l'endroit où ont été collées les électrodes. Si tel est le cas, une lotion calmante sera appliquée. Si l'irritation cutanée persiste plus de 24 heures, vous devrez aviser un des responsables du projet et consulter un médecin. De plus, le risque de pertes d'équilibre lors de la marche ne peut être complètement éliminé. Cependant, lors des moments les plus instables (lorsqu'il y a changement des vitesses des courroies), vous aurez l'autorisation de vous tenir sur les barres d'appui puisqu'aucun n'enregistrement n'est effectué durant cette période. De plus, deux personnes seront à vos côtés afin d'assurer votre sécurité. Le tapis roulant, étant composé de barres d'appui des deux côtés et en avant de vous, vous permettra de vous stabiliser à tout moment lors des différentes conditions de marche au cas où vous auriez une période de déséquilibre.

Il est également entendu que votre participation à cette étude ne nuira d'aucune manière à tout traitement médical ou de réadaptation auquel vous êtes soumis ou pourriez éventuellement être soumis à l'Institut de réadaptation Gingras-Lindsay de Montréal ou à l'Hôpital de réadaptation Villa Medica.

INCONVÉNIENTS PERSONNELS POUVANT DÉCOULER DE VOTRE PARTICIPATION

Il se peut que les efforts demandés lors de l'évaluation en laboratoire provoquent tout au plus une certaine fatigue mais celle-ci ne sera que temporaire. Par ailleurs, les déplacements occasionnés pour la séance d'évaluation peuvent constituer un inconvénient pour certaines personnes.

La pose d'électrodes pour enregistrer l'activité musculaire peut nécessiter le rasage des poils sur les surfaces de la peau où elles seront placées. A ce titre, les règles d’hygiène les plus strictes (rasoirs et collerettes à usage unique, nettoyage de la peau avec de l’alcool) seront mises en place.

ACCÈS À VOTRE DOSSIER MÉDICAL

Vous acceptez que les personnes responsables de ce projet aient accès à votre dossier médical de l’Institut de réadaptation Gingras-Lindsay de Montréal. Nous prélèverons à votre dossier certaines informations sur votre état de santé, sur les tests et mesures réalisés par les cliniciens en lien avec les évaluations décrites plus haut.

AUTORISATION D’UTILISER LES RÉSULTATS

Vous acceptez que l'information recueillie puisse être utilisée pour des fins de communication scientifique, professionnelle et d’enseignement. Il est entendu que l'anonymat sera respecté à votre égard.

CONFIDENTIALITÉ

Il est entendu que les observations effectuées en ce qui vous concerne, dans le cadre du projet de recherche décrit ci-dessus, demeureront strictement confidentielles. À cet effet, tous les renseignements personnels recueillis à votre sujet au cours de l’étude seront codifiés et conservés sous clé dans une filière du laboratoire de pathokinésiologie et d’analyse d’activités fonctionnelles de l’IRGLM par le responsable de l’étude pour une période de 5 ans suivant la fin du projet. Seuls les membres de l’équipe de recherche y auront accès. Après cette période de 5 ans, ces renseignements seront détruits. Cependant, à des fins de contrôle du projet de recherche, votre dossier pourrait être consulté par une personne mandatée par le CÉR des établissements du CRIR, qui adhère à une politique de stricte confidentialité.

INFORMATIONS CONCERNANT LE PROJET

Pour votre satisfaction, nous nous appliquerons à répondre à toutes les questions que vous poserez à propos du projet de recherche auquel vous acceptez de participer. Pour toutes informations ou questions, vous pourrez communiquer avec Sylvie Nadeau, Ph. D. en sciences biomédicales (réadaptation) responsable du projet, au numéro de téléphone 514-340-2111 au poste 2179.

Si vous avez des questions sur vos droits et recours ou sur votre participation à ce projet de recherche, vous pouvez communiquer avec Me Anik Nolet, coordonnatrice à l'éthique de la recherche des établissements du CRIR au (514) 527-4527 poste 2643 ou par courriel à l'adresse : anolet.crir@ssss.gouv.qc.ca

PARTICIPATION VOLONTAIRE ET RETRAIT DE VOTRE PARTICIPATION

Il est entendu que votre participation au projet de recherche décrit ci-dessus est tout à fait libre et volontaire. Il est également entendu que vous pourrez, à tout moment, mettre un terme à votre participation sans aucun préjudice et sans que cela n'affecte les services de santé auxquels vous aurez droit à l'Institut de Réadaptation Gingras-Lindsay de Montréal ou à l'Hôpital de réadaptation Villa Medica. En cas de retrait de votre part, les documents audiovisuels et écrits vous concernant seront détruits.

CLAUSE DE RESPONSABILITÉ :

Il est entendu qu'en acceptant de participer à cette étude, vous ne renoncez à aucun de vos droits ni ne libérez les chercheurs et les institutions impliquées de leurs obligations légales et professionnelles.

INDEMNITÉ COMPENSATOIRE

Une somme de 50$ vous sera remise à chacune des visites (1^{e} et 2^{e}) afin de compenser pour les dépenses encourues par votre participation à ce projet de recherche.

CONSENTEMENT

Je déclare avoir lu et compris le présent projet, la nature et l'ampleur de ma participation, ainsi que les risques auxquels je m'expose tels que présentés dans le présent formulaire. J'ai eu l'occasion de poser toutes les questions concernant les différents aspects de l'étude et de recevoir des réponses à ma satisfaction.

Je, soussigné(e), accepte volontairement de participer à cette étude. Je peux me retirer en tout temps sans préjudice d'aucune sorte. Je certifie qu'on m'a laissé le temps voulu pour prendre ma décision et je sais qu'une copie de ce formulaire figurera dans mon dossier médical.

J'accepte d'être contacté (e) dans le futur par le même chercheur principal pour d'autres études dans un domaine de recherche connexe :

- ☐ non
- ☐ oui (pour une durée d'un an) *
- ☐ oui (pour une durée de deux ans) *
- ☐ oui (pour une durée de cinq ans) *

* *Notez que si vous cochez l'une de ces trois cases, vos coordonnées personnelles seront conservées par le chercheur principal pour la période à laquelle vous avez consenti.*

J'accepte que les données recueillies au cours de cette étude soient utilisées pour d'autres publications scientifiques demeurant en lien (même domaine de recherche) avec le présent projet.

oui ☐ non ☐

Une copie signée de ce formulaire d'information et de consentement doit m'être remise.

________________________________ ______________________________

Nom du sujet Signature de l'intéressé (e)

Fait à ___________________, le ______________________, 20______.

ENGAGEMENT DU CHERCHEUR

Je, soussigné(e), ____, certifie

(a) avoir expliqué au signataire les termes du présent formulaire;

(b) avoir répondu aux questions qu'il m'a posées à cet égard;

(c) lui avoir clairement indiqué qu'il reste, à tout moment, libre de mettre un terme à sa participation au projet de recherche décrit ci-dessus;

et (d) que je lui remettrai une copie signée et datée du présent formulaire.

____

Signature du responsable du projet
ou de son représentant

Fait à ___________________, le _______________ 20__.

Appendix VI

Consent Form for Healthy Participants (Cross-sectional Study)

(Version in French)

INSTITUT DE RÉADAPTATION
Gingras-Lindsay-de-Montréal

CRIR
Centre de recherche interdisciplinaire en réadaptation du Montréal Métropolitain

Formule de consentement pour votre participation à un projet de recherche

TITRE DU PROJET

Comparaison de la marche asymétrique et symétrique chez les personnes hémiparétiques chroniques

RESPONSABLE

Sylvie Nadeau, pht, Ph.D	Chercheure, Centre de recherche interdisciplinaire en réadaptation (CRIR), Institut de réadaptation Gingras-Lindsay de Montréal (IRGLM), Laboratoire de pathokinésiologie et d'analyse des activités fonctionnelles. Professeure titulaire à l'Université de Montréal, École de réadaptation. Chercheure responsable du projet.

CO-CHERCHEURS :

Cyril Duclos, Ph. D.	Chercheur, CRIR, IRGLM
Séléna Lauzière, pht, M. Sc.	Candidate au doctorat, CRIR, IRGLM
Carole Miéville, M. Sc.	Candidate au doctorat, CRIR, IRGLM
Rachid Aissaoui, Ing, Ph. D.	Chercheur associé au CRIR, site IRGLM et École de technologie supérieure

PRÉAMBULE

Nous vous demandons de participer à un projet de recherche qui implique différentes évaluations se déroulant au laboratoire de pathokinésiologie au 4e étage de l'IRGLM. Ces évaluations visent à étudier l'effet de la marche symétrique sur la stabilité, le coût énergétique et les niveaux d'effort musculaire produits aux membres inférieurs.

Avant d'accepter de participer à ce projet de recherche, veuillez prendre le temps de comprendre et de considérer attentivement les renseignements suivants.

Ce formulaire de consentement vous explique le but de cette étude, les procédures, les avantages, les risques et inconvénients, de même que les personnes avec qui communiquer au besoin.

Le présent formulaire de consentement peut contenir des mots que vous ne comprenez pas. Nous vous invitons à poser toutes les questions que vous jugerez utiles au chercheur et aux autres membres du personnel affecté au projet de recherche et à leur demander de vous expliquer tout mot ou renseignement qui n'est pas clair.

DESCRIPTION DU PROJET ET DE SES OBJECTIFS

Des problèmes locomoteurs sont fréquemment rencontrés chez les personnes ayant une hémiparésie suite à un accident vasculaire cérébral (AVC). Le plus souvent, leur performance est caractérisée par une diminution de la vitesse de marche et par une asymétrie des mouvements entre les deux jambes. Cependant, sur demande, ces personnes peuvent habituellement effectuer la tâche à une vitesse plus élevée et de façon plus symétrique que ce qu'elles font de façon naturelle. La question qui nous intéresse ici est de comprendre pourquoi les personnes hémiparétiques utilisent une stratégie asymétrique alors qu'elles ont les capacités de marcher plus symétriquement. Les résultats de nos travaux antérieurs suggèrent que la perception de l'effort produit afin de réussir la tâche pourrait expliquer la stratégie de mouvements choisie. L'objectif du présent projet est de déterminer les effets réels et perçus d'une marche symétrique sur la stabilité posturale, le coût énergétique et les niveaux d'effort musculaire afin de déterminer si ces facteurs sont explicatifs de la performance motrice mesurée en laboratoire et en clinique chez les individus hémiparétiques. Un objectif secondaire est d'évaluer l'effet de la marche prolongée sur la symétrie du patron de marche, la stabilité, l'effort global et les niveaux d'effort musculaire.

Pour répondre à ces objectifs, 20 participants avec une hémiparésie chronique consécutive à un AVC unilatéral seront recrutés dans deux établissements de réadaptation: l'Institut de réadaptation Gingras-Lindsay de Montréal (IRGLM) et l'Hôpital de réadaptation Villa Medica. De plus, 10 participants sains, dont vous faites partie, seront recrutés afin d'avoir des données comparatives.

NATURE ET DURÉE DE LA PARTICIPATION

Cette étude comporte deux séances d'évaluation pour les personnes hémiparétiques mais seulement une pour les personnes saines (durée d'environ 4 heures). Cette séance se déroulera au laboratoire de pathokinésiologie et d'analyse de tâches fonctionnelles du site IRGLM.

Au début de la séance, un(e) physiothérapeute évaluera votre santé, votre condition physique ainsi que votre habileté à réaliser quelques activités fonctionnelles via des tests standardisés.

Vous serez invité(e) à réaliser différents types d'effort avec vos jambes. Ces tests serviront à évaluer votre force musculaire avec un appareil appelé dynamomètre. Il s'agit d'un appareil qui permet de mesurer précisément la force maximale lors de poussées avec différentes parties de vos jambes contre l'appareil. Pour cette évaluation de la force, vous serez assis ou couché et des courroies vous stabiliseront et empêcheront les mouvements de certaines parties de votre corps (voir photo 1). Au total, vous aurez à réaliser environ 76 contractions d'une durée d'environ 5 secondes chacune avec différents muscles de vos jambes avec des repos fréquents.

Photo 1. Dynamomètre Biodex et courroies de stabilisation

Finalement, une séance de familiarisation sur le tapis roulant à double courroie sera effectuée avec le port d'un masque nommé Cosmed qui sert à mesurer votre consommation d'oxygène (voir photo 2). Cette familiarisation vous permettra d'expérimenter les différentes conditions de marche utilisées lors de la visite sur le tapis roulant à double courroie (voir photo 3). Comme son nom l'indique, ce tapis roulant possède deux courroies distinctes qui peuvent se déplacer à des vitesses différentes. Ainsi, il permet de faire varier la vitesse de déplacement d'une jambe différemment par rapport à l'autre. Le tapis roulant possède des barres d'appui des deux côtés et également une barre d'appui à l'avant. Ainsi, malgré qu'il vous soit demandé de marcher sur le tapis roulant sans prendre appui avec vos mains, vous pourrez vous stabiliser sur ses barres en cas de déséquilibre. De plus, vous serez encadré en tout temps de deux personnes qui assureront votre sécurité.

Photo 2. Système d’acquisition des paramètres cardiorespiratoires (COSMED)

Photo 3. Tapis roulant à double courroie

Par la suite, vous aurez à effectuer plusieurs conditions de marche différentes sur le tapis roulant à double courroie. Ainsi, il y aura des conditions où les courroies se déplaceront à la même vitesse et une condition où elles se déplaceront à des vitesses différentes. Dans cette condition, votre marche deviendra asymétrique avec une jambe se déplaçant plus vite que l'autre. De plus, il vous sera demandé de marcher lors d'une période prolongée afin d'évaluer les changements dans votre performance motrice lorsque vos muscles sont fatigués. Lors de ces conditions, l'activité de vos muscles sera enregistrée avec des électrodes que nous collerons sur les muscles de vos jambes. Nous mesurerons simultanément les forces que vous produisez sous les pieds à l'aide de plates-formes de forces qui sont situées sous le tapis roulant. Des marqueurs seront collés sur différentes parties de votre corps (pieds, jambes, cuisses, bassin, tronc), pour permettre l'enregistrement de vos mouvements à l'aide d'un système de caméras infrarouges. Tous les essais seront aussi enregistrés à l'aide de deux caméras vidéo afin de nous fournir une image de la manière dont vous exécutez les tâches. Lors de certaines de ces tâches, nous vous demanderons de cocher l'effort, la stabilité et le niveau d'effort musculaire que vous percevez lorsque vous exécutez les diverses tâches locomotrices. Des périodes de repos (2 périodes de repos de 20 minutes) vous seront accordés entre les différentes conditions si nécessaire. Des repos additionnels s'ajouteront au besoin, selon votre endurance physique.

AVANTAGES PERSONNELS POUVANT DÉCOULER DE VOTRE PARTICIPATION

En tant que participant, vous ne retirerez aucun avantage de votre implication au projet de recherche. Par ailleurs, votre participation aura contribué à l'avancement de la recherche dans le domaine de la réadaptation des personnes avec un AVC.

RISQUES POUVANT DÉCOULER DE VOTRE PARTICIPATION

Il est entendu que votre participation à ce projet ne vous fait courir, sur le plan médical, aucun risque que ce soit. Toutefois, dans quelques cas, une irritation cutanée pourrait survenir à l'endroit où ont été collées les électrodes. Si tel est le cas, une lotion calmante sera appliquée. Si l'irritation cutanée persiste plus de 24 heures, vous devrez aviser un des responsables du projet et consulter un médecin. De plus, le risque de pertes d'équilibre lors de la marche ne peut être complètement éliminé. Cependant, lors des moments les plus instables (lorsqu'il y a changement des vitesses des courroies), vous aurez l'autorisation de vous tenir sur les barres d'appui puisqu'aucun n'enregistrement n'est effectué durant cette période. De plus, deux personnes seront à vos côtés afin d'assurer votre sécurité. Le tapis roulant, étant composé de barres d'appui des deux côtés et en avant de vous, vous permettra de vous stabiliser à tout moment lors des différentes conditions de marche au cas où vous auriez un déséquilibre.

INCONVÉNIENTS PERSONNELS POUVANT DÉCOULER DE VOTRE PARTICIPATION

Il se peut que les efforts demandés lors de l'évaluation en laboratoire provoquent tout au plus une certaine fatigue mais celle-ci ne sera que temporaire. Par ailleurs, les déplacements occasionnés pour la séance d'évaluation peuvent constituer un inconvénient pour certaines personnes.

La pose d'électrodes pour enregistrer l'activité musculaire peut nécessiter le rasage des poils sur les surfaces de la peau où elles seront placées. A ce titre, les règles d'hygiène les plus strictes (rasoirs et collerettes à usage unique, nettoyage de la peau avec de l'alcool) seront mises en place.

AUTORISATION D'UTILISER LES RÉSULTATS

Vous acceptez que l'information recueillie puisse être utilisée pour des fins de communication scientifique, professionnelle et d'enseignement. Il est entendu que l'anonymat sera respecté à votre égard.

CONFIDENTIALITÉ

Il est entendu que les observations effectuées en ce qui vous concerne, dans le cadre du projet de recherche décrit ci-dessus, demeureront strictement confidentielles. À cet effet, tous les renseignements personnels recueillis à votre sujet au cours de l'étude seront codifiés et conservés sous clé dans une filière du laboratoire de pathokinésiologie et d'analyse d'activités fonctionnelles de l'IRGLM par le responsable de l'étude pour une période de 5 ans suivant la fin du projet. Seuls les membres de l'équipe de recherche y auront accès. Après cette période de 5 ans, ces renseignements seront détruits. Cependant, à des fins de contrôle du projet de recherche, votre dossier pourrait être consulté par une personne mandatée par le CÉR des établissements du CRIR, qui adhère à une politique de stricte confidentialité.

INFORMATIONS CONCERNANT LE PROJET

Pour votre satisfaction, nous nous appliquerons à répondre à toutes les questions que vous poserez à propos du projet de recherche auquel vous acceptez de participer. Pour toutes informations ou questions, vous pourrez communiquer avec Sylvie Nadeau, Ph. D. en sciences biomédicales (réadaptation) responsable du projet, au numéro de téléphone 514-340-2111 au poste 2179.

Si vous avez des questions sur vos droits et recours ou sur votre participation à ce projet de recherche, vous pouvez communiquer avec Me Anik Nolet, coordonnatrice à l'éthique de la recherche des établissements du CRIR au (514) 527-4527 poste 2643 ou par courriel à l'adresse: anolet.crir@ssss.gouv.qc.ca

PARTICIPATION VOLONTAIRE ET RETRAIT DE VOTRE PARTICIPATION

Il est entendu que votre participation au projet de recherche décrit ci-dessus est tout à fait libre et volontaire. Il est également entendu que vous pourrez, à tout moment, mettre un terme à votre participation sans aucun préjudice. En cas de retrait de votre part, les documents audiovisuels et écrits vous concernant seront détruits.

CLAUSE DE RESPONSABILITÉ

Il est entendu qu'en acceptant de participer à cette étude, vous ne renoncez à aucun de vos droits ni ne libérez les chercheurs et les institutions impliquées de leurs obligations légales et professionnelles.

INDEMNITÉ COMPENSATOIRE

Une somme de 50$ vous sera remise suite à votre visite afin de compenser pour les dépenses encourues par votre participation à ce projet de recherche.

CONSENTEMENT

Je déclare avoir lu et compris le présent projet, la nature et l'ampleur de ma participation, ainsi que les risques auxquels je m'expose tels que présentés dans le présent formulaire. J'ai eu l'occasion de poser toutes les questions concernant les différents aspects de l'étude et de recevoir des réponses à ma satisfaction.

Je, soussigné(e), accepte volontairement de participer à cette étude. Je peux me retirer en tout temps sans préjudice d'aucune sorte. Je certifie qu'on m'a laissé le temps voulu pour prendre ma décision et je sais qu'une copie de ce formulaire figurera dans mon dossier médical.

J'accepte d'être contacté (e) dans le futur par le même chercheur principal pour d'autres études dans un domaine de recherche connexe :

☐ non

☐ oui (pour une durée d'un an) *

☐ oui (pour une durée de deux ans) *

☐ oui (pour une durée de cinq ans) *

* *Notez que si vous cochez l'une de ces trois cases, vos coordonnées personnelles seront conservées par le chercheur principal pour la période à laquelle vous avez consenti.*

J'accepte que les données recueillies au cours de cette étude soient utilisées pour d'autres publications scientifiques demeurant en lien (même domaine de recherche) avec le présent projet.

oui ☐ non ☐

Une copie signée de ce formulaire d'information et de consentement doit m'être remise.

____________________________________ ____________________________

Nom du sujet Signature de l'intéressé (e)

Fait à __________________, le ___________________, 20______.

ENGAGEMENT DU CHERCHEUR

Je, soussigné(e), ____, certifie

(a) avoir expliqué au signataire les termes du présent formulaire ;

(b) avoir répondu aux questions qu'il m'a posées à cet égard ;

(c) lui avoir clairement indiqué qu'il reste, à tout moment, libre de mettre un terme à sa participation au projet de recherche décrit ci-dessus ;

et (d) que je lui remettrai une copie signée et datée du présent formulaire.

____

Signature du responsable du projet / ou de son représentant

Fait à __________________, le ______________ 20__.

Appendix VII

Consent Form for Participants with Stroke (Training Study)

(Version in French)

INSTITUT DE RÉADAPTATION Gingras-Lindsay-de-Montréal

CRIR Centre de recherche interdisciplinaire en réadaptation du Montréal Métropolitain

constance-lethbridge CENTRE DE RÉADAPTATION • REHABILITATION CENTRE 75 ans • years

Formulaire de consentement pour votre participation à un projet de recherche

TITRE DU PROJET

Effets de la marche répétée sur un tapis à double courroies sur la symétrie de la marche et l'activité musculaire chez les personnes hémiparétiques chroniques : une étude pilote.

RESPONSABLE DU PROJET

Sylvie Nadeau, Ph. D., pht. — Chercheure, Centre de recherche interdisciplinaire en réadaptation (CRIR), Institut de réadaptation Gingras-Lindsay de Montréal (IRGLM), Laboratoire pathokinésiologie; professeure titulaire à Université de Montréal, École de réadaptation

CO-CHERCHEURS

Bradford J. McFadyen, Ph. D. — Chercheur au département de réadaptation, faculté de médecine, professeur à l'Université de Laval, CIRRIS

Martina Betschart, Pt, M. Sc. — Étudiante au doctorat à l'Université de Montréal, Sciences de la réadaptation, CRIR, IRGLM

Mylène Lajeunesse-Langdeau — Étudiante en physiothérapie (B. Sc.) à l'Université de Montréal (1[ière] année)

PRÉAMBULE

Nous vous invitons à participer à un projet de recherche qui implique l'analyse des effets d'un entraînement sur un tapis à double courroies sur la symétrie de votre marche et sur votre activité musculaire.

Avant d'accepter de participer à ce projet de recherche, veuillez prendre le temps de comprendre et de considérer attentivement les renseignements qui suivent. Ce formulaire de consentement vous explique le but de cette étude, les procédures, les avantages, les risques et inconvénients. Vous y trouverez aussi les coordonnées des personnes avec qui communiquer en cas de besoin.

Le présent formulaire de consentement peut contenir des mots que vous ne comprenez pas. Les chercheurs et les membres du personnel affectés à cette étude sont à votre disposition pour répondre à toutes vos questions et demandes d'éclaircissement. N'hésitez pas à les solliciter.

DESCRIPTION DU PROJET ET DE SES OBJECTIVES

Des problèmes à la marche sont fréquemment rencontrés chez les personnes ayant une hémiparésie suite à un accident vasculaire cérébral (AVC). Le plus souvent, leur marche est caractérisée par une asymétrie des pas entre les deux côtés. Une telle asymétrie peut avoir pour conséquences des dépenses d'énergie élevées, des exigences plus grandes en matière de contrôle d'équilibre et des faiblesses musculaires. De plus, l'asymétrie du schéma de marche semble être résistante aux interventions conventionnelles de physiothérapie. Une étude récente a démontré des effets positifs liés à la marche répétée sur un tapis à double courroies. Cependant, les effets positifs précis d'un tel protocole sur les paramètres de marche affectés à la suite d'un AVC (i.e. : activité musculaire) restent inconnus. Les premiers résultats d'une étude effectuée au laboratoire de pathokinésiologie (IRGLM) ont démontré que certains muscles augmentent leur activité à court terme après une séance unique de marche sur un tapis à double courroies.

L'objectif principal de ce projet pilote est de tester un protocole d'entraînement visant à améliorer la symétrie de la marche et l'activité musculaire. Plus spécifiquement, le projet cherche à évaluer :

a) la faisabilité de ce protocole dans un environnement clinique;
b) les effets directs et à long-terme sur le schéma de marche et l'activité musculaire chez les individus ayant subi un AVC.

Image 1 : Illustration d'oxymètre du pouls

Pour répondre à ces objectifs, 10 participants atteints d'une hémiparésie chronique consécutive à un AVC unilatéral seront recrutés dans deux établissements de réadaptation ; le centre de réadaptation Constance-Lethbridge (CRCL) et l'Institut de réadaptation Gingras-Lindsay de Montréal (IRGLM). La participation à ce projet devrait durer environ 6 semaines.

NATURE ET DURÉE DE LA PARTICIPATION

Si vous acceptez de participer à ce projet, vous serez soumis aux étapes suivantes.

1) ***Séance 1 (1 heure)*** *à l'IRGLM*

Un(e) physiothérapeute vous demandera de compléter des questionnaires sur votre état de santé. Il ou elle testera ensuite votre capacité de marche (vitesse, symétrie et endurance) en utilisant des tests standardisés. Afin de mesurer votre vitesse de marche et la symétrie de vos pas, vous devrez marcher au sol environ 2 fois 10 mètres. Pour mesurer votre endurance, le ou la physiothérapeute vous demandera de marcher au sol pendant 6 minutes, le plus rapidement possible mais en toute sécurité. Lors de ce dernier test, vous serez équipé d'un oxymètre du pouls. Ce petit appareil fixé à votre doigt par une pince (*Image 1*) donne des indications du taux d'oxygène dans le sang.

Important : Cette séance fournira des informations sur votre éligibilité au projet proposé. Par conséquent, votre participation pourrait se terminer après cette première séance.

2) ***Séance 2 (2 heures)*** *à l'IRGLM, (1 semaine après la séance 1)*

Si vous répondez aux critères d'inclusion de ce projet pilote, nous devons tout d'abord nous assurer que votre schéma de marche n'évolue pas avec le temps et que les changements observés durant cette étude seront uniquement liés à l'entraînement sur le tapis roulant. Pour ce faire, nous allons à nouveau tester votre marche. Lors de cette séance, vous marcherez au sol quelques fois sur une distance de 5 mètres. Des électrodes, mesurant votre activité musculaire lors de la marche, seront fixées sur vos jambes (*Image* 2). De plus, des capteurs, collés à différents endroits de votre corps (pieds, jambes, bassin), enregistreront vos mouvements à l'aide d'un système infrarouge *(Image 3).* Nous mesurerons simultanément les forces que vous exercez à chaque pas à l'aide de plates-formes de force intégrées dans le sol. Tous les essais seront enregistrés à l'aide de caméras vidéo afin de nous fournir une image de la façon dont vous marchez.

Image 2 : Illustration du système EMG

Après ces tests, vous pourrez vous familiariser avec le protocole d'entraînement grâce à une courte période de marche sur le tapis roulant (environ dix minutes).

Pendant toute la séance, il vous sera possible de prendre des périodes de repos aussi souvent que vous en ressentez le besoin.

L’ENTRAÎNEMENT :

3) ***Séances 3 à 8*** *(1 heure par séance, 6 séances à 3 jours par semaine ; idéalement, un jour de pause entre les séances)*

! L’entraînement se déroulera sur un tapis roulant à double courroies, sans l’utilisation d’électrodes ni de marqueurs. Afin d’éviter les risques de chutes, vous porterez un harnais *(Image 4)* et vous aurez la possibilité de vous tenir aux barres d’appui au besoin.

- Les séances débuteront par une période d’échauffement (5-10min). Pendant cette période, votre vitesse de marche confortable sera déterminée et vous pourrez vous familiariser avec la marche sur le tapis.
- Après cette période d’échauffement, vous marcherez 3 minutes à une vitesse de marche confortable.
- Puis, la vitesse d’une courroie sera doublée et vous marcherez à vitesse asymétrique pendant 4 x 5 minutes (20 minutes). Entre les séquences de 5 minutes de marche, une période repos en position assise vous sera allouée. Pendant ce repos, nous mesurerons votre tension artérielle. Des repos additionnels s’ajouteront au besoin.
- Après ces 20 minutes, les courroies seront remises à la même vitesse et vous marcherez 3 minutes supplémentaires à votre vitesse de marche confortable.

Pendant toutes les séances d’entraînement, un(e) physiothérapeute, ou un(e) assistant(e) de recherche, restera près de vous pour garantir votre sécurité.

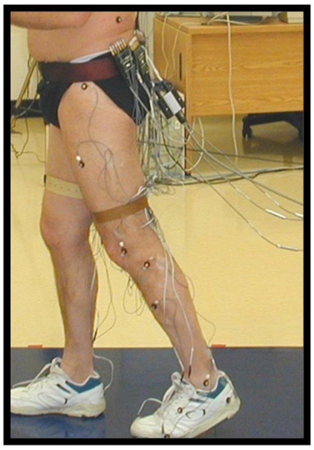

Image 3 : Illustration des marqueurs infrarouges pendant la marche sur le sol

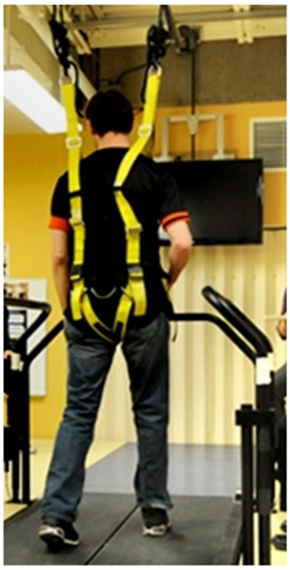

Image 4: Illustration d'équipment

La personne présente vous demandera régulièrement votre perception de l'effort pendant l'entraînement. Votre fréquence cardiaque sera constamment mesurée à l'aide d'un système d'enregistrement (Polar, USA) fixé autour de votre thorax. Cette mesure permettra de nous assurer que vous marchez à un niveau d'intensité sous la fréquence cardiaque maximale préconisée par les standards de l'*American Heart Association.*

4) ***Séance 9 (1-2 heures)*** *à l'IRGLM*

La séance 9 vise à évaluer les effets potentiels de l'entraînement sur votre schéma de marche. Vous serez invité à une nouvelle séance d'évaluation au laboratoire à IRGLM un à deux jour(s) après votre dernière séance d'entraînement, Vous marcherez au sol environ 5 fois 5 mètres comme lors de la séance 2. Le même équipement sera utilisé que lors de cette séance. Vous pourrez aussi prendre des pauses au besoin.

5) ***Séance 10* (1-2 heures)** à l'IRGLM

Pour analyser les effets à plus long terme de l'entraînement, vous serez invité à prendre part à une dernière séance d'évaluation, similaire aux séances 2 et 9. Cette séance devrait avoir lieu un mois après votre dernière séance d'entraînement.

Important : Un des objectifs de cette étude étant l'évaluation des effets de l'entraînement de la marche sur le tapis à double courroies, tout entraînement additionnel pourrait fausser les résultats, Ainsi, nous vous demandons de ne pas commencer d'entraînement supplémentaire ou d'interrompre vos entraînements actuels pendant la durée de votre participation à l'étude. Vous pourrez reprendre vos activités habituelles (entraînement et autre traitement) tel que vous le souhaitez dès la fin de l'étude c'est-à-dire suite à la séance 10.

AVANTAGES POUVANT DÉCOULER DE VOTRE PARTICIPATION

Nous ne pouvons pas garantir des effets positifs ou des avantages en participant à cette étude pilote. Toutefois, vous aurez contribué à l'avancement de la science et à la collaboration entre la recherche et la clinique.

RISQUES POUVANT DÉCOULER DE VOTRE PARTICIPATION.

Nous aimerions vous informer des risques potentiels qui peuvent découler de votre participation.

- L'entraînement pourrait provoquer un effort physique menant à une fréquence cardiaque et une pression artérielle plus élevée que pendant la marche naturelle. Ceci est attendu. Toutefois, puisque les réponses cardiovasculaires sont contrôlées pendant l'entraînement et que votre dossier médical a été consulté avant le début de votre participation, aucun changement préoccupant persistant au-delà de l'entraînement n'est attendu.

- Dans quelques cas, une irritation cutanée pourrait survenir à l'endroit où ont été collées les électrodes. Si tel est le cas, une lotion calmante sera appliquée. Si l'irritation cutanée persiste plus de 24 heures, vous devrez aviser un des responsables du projet et consulter un médecin.
- De plus, le risque de pertes d'équilibre lors de la marche ne peut être complètement éliminé. Cependant, lors des moments plus instables (lorsqu'il y a changement des vitesses des courroies), vous aurez l'autorisation de vous tenir sur les barres d'appui des deux côtés. Pendant toute la séance d'entraînement vous porterez un harnais. Pour l'entraînement et l'évaluation du schéma de marche au sol, une personne sera à vos côtés afin d'assurer votre sécurité.

INCONVÉNIENTS PERSONNELS POUVANT DÉCOULER DE VOTRE PARTICIPATION

- Il se peut que les efforts demandés lors de l'évaluation en laboratoire provoquent tout au plus une certaine fatigue mais celle-ci ne sera que temporaire.
- L'entraînement exigera un certain niveau d'efforts musculaires, ainsi, vous pourriez ressentir des courbatures deux ou trois jours suivant une séance d'entraînement. Cette réaction est toute à fait normale et ne représente pas un risque potentiel pour vous.
- La pose d'électrodes pour enregistrer l'activité musculaire peut nécessiter le rasage des poils sur les surfaces de la peau où elles seront placées. À ce titre, les règles d'hygiène les plus strictes (rasoirs et collerettes à usage unique, nettoyage de la peau avec de l'alcool) seront mises en place.
- Par ailleurs, les déplacements occasionnés pour les séances d'évaluation et d'entraînement peuvent constituer un inconvénient pour certaines personnes.

ACCÈS À VOTRE DOSSIER MÉDICAL

Vous acceptez, que les personnes responsables de ce projet aient accès à votre dossier médical du centre de réadaptation Constance-Lethbridge (CRCL) ou de l'Institut de réadaptation Gingras-Lindsay de Montréal (IRGLM). Nous relèverons, dans votre dossier, certaines informations sur votre état de santé, sur les tests et mesures réalisés par les cliniciens en lien avec les évaluations décrites plus haut (Section III, page 4).

AUTORISATION D'UTILISER LES RÉSULTATS

Vous acceptez que l'information recueillie puisse être utilisée pour des fins de communication scientifique, professionnelle et d'enseignement. Il est entendu que votre anonymat sera respecté.

CONFIDENTIALITÉ

Il est entendu que les observations effectuées en ce qui vous concerne, dans le cadre du projet de recherche décrit ci-dessus, demeureront strictement confidentielles. À cet effet, tous les renseignements personnels recueillis à votre sujet au cours de l'étude seront codifiés et conservés sous clé dans une filière du laboratoire de pathokinésiologie et d'analyse d'activités fonctionnelles de l'IRGLM par le responsable de l'étude et ce pour une période de 5 ans, suivant la fin du projet. Seuls les membres de l'équipe de recherche y auront accès. Ces renseignements seront détruits à l'échéance de ce délai de conservation. Cependant, à des fins de contrôle du projet de recherche, votre dossier pourrait être consulté par une personne mandatée par le CÉR des établissements du Centre de recherche interdisciplinaire en réadaptation (CRIR) ou par l'Unité de l'éthique du ministère de la Santé et des Services Sociaux du Québec, qui adhèrent à une politique de stricte confidentialité.

INFORMATION CONCERNANT LE PROJET

Nous nous engageons à répondre à toutes les questions que vous vous poserez à propos du projet de recherche auquel vous acceptez de participer. Pour toutes informations ou questions, vous pourrez vous adresser à Sylvie Nadeau, Ph. D. responsable du projet, au numéro de téléphone 514-340-2111 ext. 2179.

Si vous avez des questions sur vos droits et recours ou sur votre participation à ce projet de recherche, vous pouvez vous adressez à Me Anik Nolet, coordonnatrice à l'éthique de la recherche des établissements du CRIR, au 514-527-4527 ext. 2643, ou par courriel à l'adresse : anolet.crir@ssss.gouv.qc.ca. Vous pouvez aussi contacter le commissaire local aux plaintes et à la qualité des services de votre établissement (IRGLM : 514-345-5225' CRCL : 514 487-1891 poste 515)

PARTICIPATION VOLONTAIRE ET RETRAIT DE VOTRE PARTICIPATION

Il est entendu que votre participation au projet de recherche décrit ci-dessus est tout à fait libre et volontaire. Il est également entendu que vous pourrez, à tout moment, mettre un terme à votre participation sans aucun préjudice et sans que cela n'affecte les services de santé auxquels vous aurez droit à l'un des deux centres (CRCL, IRGLM). En cas de retrait de votre part, les documents audiovisuels et écrits vous concernant seront détruits, si vous en faites la demande.

CLAUSE DE RESPONSABILITÉ :

Il est entendu qu'en acceptant de participer à cette étude, vous ne renoncez à aucun de vos droits ni ne libérez les chercheurs et les institutions impliquées de leurs obligations légales et professionnelles.

INDEMNITÉ COMPENSATOIRE :

Une somme de 100 $ vous sera remise au prorata de votre participation afin de compenser les dépenses encourues par votre participation à ce projet de recherche.

CONSENTEMENT

Je déclare avoir lu et compris le présent projet, la nature et l'ampleur de ma participation, ainsi que les risques auxquels je m'expose tels que présentés dans le présent formulaire. J'ai eu l'occasion de poser toutes les questions concernant les différents aspects de l'étude et de recevoir les réponses qui m'ont satisfait.

Je, soussigné(e), accepte volontairement de participer à cette étude. Je peux me retirer en tout temps sans préjudice d'aucune sorte. Je certifie qu'on m'a laissé le temps voulu pour prendre ma décision.

J'accepte de ne participer à aucun nouvel entraînement ou traitement physique additionnel pendant ma période de participation. Je comprends que je pourrai reprendre mes activités habituelles suite à la dernière évaluation du présent projet.

J'accepte d'être contacté (e) dans le futur par le même chercheur principal pour d'autres études dans un domaine de recherche connexe :

☐ non
☐ oui (pour une durée d'un an) *
☐ oui (pour une durée de deux ans) *
☐ oui (pour une durée de trois ans) *

* *Notez que si vous cochez l'une de ces trois cases, vos coordonnées personnelles seront conservées par le chercheur principal pour la période à laquelle vous avez consenti.*

J'accepte que les données recueillies au cours de cette étude soient utilisées pour d'autres publications scientifiques demeurant en lien (même domaine de recherche) avec le présent projet.

oui ☐ non ☐

Une copie signée de ce formulaire d’information et de consentement doit m’être remise.

____ _________________________________

Nom du sujet Signature de l’intéressé (e)

Fait à _____________________, le ______________________, 20______.

ENGAGEMENT DU CHERCHEUR

Je, soussigné(e), ____, certifie

(a) avoir expliqué au signataire les termes du présent formulaire ;

(b) avoir répondu aux questions qu'il m'a posées à cet égard ;

(c) lui avoir clairement indiqué qu'il reste, à tout moment, libre de mettre un terme à sa participation au projet de recherche décrit ci-dessus ; et (d) que je lui remettrai une copie signée et datée du présent formulaire.

____

Signature du responsable du projet
ou de son représentant

Fait à _____________________, le ________________ 20__.

Appendix VIII

Abstracts Published (International Conferences)

Abstract #1

2014 ISPGR World Congress

Vancouver, June 29 to July 3

Published in the collection of abstracts of the World Congress

Title: **Changes in paretic and non-paretic lower limb muscle activities during gait on a split-belt treadmill in chronic stroke survivors**

Betschart M., Lauzière S, Miéville C, McFadyen BJ, and Nadeau S

Background and aim: Recent studies have shown that chronic stroke survivors are able to store an adapted gait pattern induced by walking on a split-belt treadmill with asymmetric belt speeds. When the belt speeds were returned to identical speeds, after-effects were observed showing a reduced asymmetry of step length and double stance duration. The aim of this study was to better understand how the motor system induces such adaptations towards symmetry by characterizing lower limb muscle activity during post-adaptation. **Methods**: Surface electromyography (EMG) was used to analyze bilateral muscle activity of six lower limb muscles for 15 chronic stroke survivors (mean age and time post-stroke: 49±13yrs and 84±93 months). The participants were asked to walk on an instrumented split-belt treadmill (Bertec FIT) under different conditions: three minutes with identical belt speeds (baseline); six minutes with the belt speed doubled on the non-paretic (NP) side (adaptation) and three minutes with identical belt speeds (post-adaptation). The protocol was then performed on the paretic (P) side. Ground reaction forces provided the spatiotemporal asymmetry (step length [SL] and double stance duration [DSD]) expressed using an NP/P ratio. EMG was band-pass filtered (20-400Hz) and RMS values were calculated and normalized for amplitude (peak RMS value obtained during baseline) and time (corresponding gait cycle). Repeated measures ANOVAs were used to compare differences of spatiotemporal and EMG values among the three conditions and sides. Pearson correlation was applied to identify associations between changes in muscle activity and spatiotemporal parameters. **Results**: SL and DSD increased after fast belt walking and reduced after slow belt walking when compared to baseline ($p \leq .001$). These changes corresponded to a reduced asymmetry of the SL (7/8) and DSD (12/12) during post-adaptation. Mean RMS values on paretic dorsiflexors increased by 21% ($p=.004$) during the post-adaptation period after walking on the fast belt. For the condition where the non-paretic leg was walking on the fast belt, the paretic plantarflexors EMG activity increased during post-adaptation (18%, $p=.012$) as did the non-paretic hip flexors (20%, $p=.027$) when compared to baseline. Only increased paretic dorsiflexor EMG activity was correlated with changes in SL symmetry ratio (r=+.653, p=.008). **Conclusions:** As shown by others, the present findings demonstrated that chronic stroke survivors have the capacity to adapt their locomotor pattern. The less asymmetrical gait pattern was related to increased muscle activity for distal paretic lower limb muscles and proximal non-paretic lower limb muscles. Analysis on temporal information of muscle activity in combination with biomechanical parameters (joint angles, joint moments) will provide better understanding of the motor adaptation in gait post-stroke created by the use of split-belt treadmill and its use for rehabilitation training.

Abstract #2

36th international symposium of the GRSNC (2014)

Montreal, May 12 to May 13

Published in the collection of abstracts of the 36th international symposium of the GRSNC

Title: Timing and coordination of distal lower limb muscles after split-belt treadmill walking in chronic stroke survivors

Betschart M., Lauzière S, Miéville C, McFadyen BJ, and Nadeau S

Background: Studies have shown that chronic stroke survivors changed step length and double stance duration towards symmetry after walking on a split-belt treadmill with asymmetric belt speeds. The aim of this study was to better understand how the motor system induces such adaptations in gait parameters by studying changes of coordination of lower limb muscle during post-adaptation. **Methods**: Surface electromyography (EMG) was used to analyze bilateral muscle activity of six lower limb muscles for 16 stroke survivors (mean age: 49±13yrs). The participants were asked to walk on an instrumented split-belt treadmill under different conditions: three minutes with identical belt speeds (baseline); six minutes with the belt speed doubled on the non-paretic (NP) side (adaptation) and three minutes with identical belt speeds (post-adaptation). The protocol was then performed on the paretic (P) side. Ground reaction forces and kinematics provided the spatiotemporal asymmetry (step length [SL] and double stance duration [DSD]). EMG was band-pass filtered (20-400Hz) and RMS values of EMG data across the gait cycle were normalized to baseline activity. Cross-correlation analysis was used to evaluate interlimb coordination (timing and activation pattern) during the three conditions. Associations between changes in EMG and spatiotemporal parameters were tested using Pearson's correlation. **Results**: SL and DSD increased on the fast leg and reduced on the slow leg during post-adaptation when compared to baseline ($p \leq .001$). Fast leg dorsiflexor activity increased during post-adaptation ($p=.001$) and led to premature activation on the paretic side during late stance ($p=.045$). Both changes were moderately correlated with changes in DSD ($r=.61$ and $r=.50$, respectively, $p<.05$). The plantarflexors on the slow leg presented no changes in coordination but increased activity after slow belt walking before push-off ($p=0.001$) with a strong positive correlation with changes in P ($r=.65$) and NP ($r=.74$) SL ($p \leq .008$). **Conclusion:** Dorsiflexors tend to be associated with temporal changes in gait whereas the plantarflexors were more related to spatial aspects. These changes were observed during stance to swing transition. The combination with other biomechanical parameters (joint angles, joint moments) for analysis will provide better understanding of the motor adaptation in gait post-stroke created by the use of split-belt treadmill and its use for rehabilitation training.