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COORDINATION OF LOWER LIMB SEGMENTS DURING OBSTACLE CLEARANCE IN HEALTHY ADULTS AND PATHOLOGICAL POPULATIONS

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Résumé

Conformément au « planar law » de la coordination intersegmentaire, lorsque les angles d'élévation de la cuisse, de la jambe et du pied sont tracés pour un cycle de marche, ils ont tendance à créer une boucle qui s'oriente sur un plan spatial en 3-dimensions (3D). Le principal objectif de cette thèse a été d'utiliser des techniques d'analyse relatives à cette loi pour mieux comprendre comment le système nerveux central (SNC) coordonne et met en œuvre les ajustements locomoteurs anticipatoires (ALAs) pour le franchissement d'obstacle dans les populations saines et pathologiques. L'étude 1 a examiné les points généraux pour les ALAs normaux et l'étude 2 a relié les amplitudes du mouvement segmentaire au travail mécanique effectué par les muscles fléchisseurs de la hanche et du genou. Les études 3 et 4 ont déterminé comment des déficiences respectives de l'ataxie cérébelleuse autosomique récessive de type 1 (ARCA-1) et d'un accident vasculaire cérébral (AVC) peuvent avoir une incidence sur le contrôle locomoteur. Chez les adultes sains, le déphasage entre les segments adjacents se sont révélés être corrélé à des caractéristiques du plan formé par les angles d'élévation du segment, et ces différences de phase ont changé systématiquement avec l'augmentation de la hauteur de l'obstacle. Il a été proposé que le SNC ajuste un patron locomoteur de base pour les contraintes environnementales par la manipulation des différentes phases de l'angle d'élévation entre les segments adjacents ainsi que par l'amplitude de l'angle d'élévation. L'étude de suivi a déterminé que lorsque les obstacles les plus hauts étaient franchis, les décalages de phase de la cuisse pour la jambe d'attaque, et pour la jambe de l'autre membre ont augmenté. Le travail effectué par les muscles de la hanche et du genou a influencé l'élévation de la cuisse différemment pour les membres d'attaques et suivants et il a été conclu que ces muscles n'ont pas de rôles spécifiques pour l'élévation et la progression du membre inférieur lors de l'ALA. Au lieu de cela, ces puissances musculaires peuvent résulter du contrôle dynamique des angles d'élévation. Lorsque l'on observe la coordination chez les participants avec ARCA-1 et AVC les trajectoires des angles d'élévation du membre inférieur ont continué à s'orienter sur un plan spatial en 3D. Chez les participants avec ARCA-1, une plus grande différence de phase entre les segments de la cuisse et la jambe a suggéré d'être un mécanisme de contrôle lorsque ce groupe augmente volontairement le dégagement des orteils pour enjamber des obstacles. Chez les participants avec un précédent d'AVC, cette différence de phase a été plus grande dans le membre non parétique, ce qui a été interprété comme nécessaire pour élever le membre en compensation pour le mauvais appui du membre parétique. Les résultats de cette thèse suggèrent des mécanismes de contrôle d'élévation segmentaire pour la mise en œuvre d'ALAs et mettent en évidence les mécanismes de compensation volontaire d'un tel contrôle dans des populations pathologiques.

Abstract

According to the planar law of intersegmental coordination, when elevation angles of the thigh, shank, and foot are plotted for a gait cycle, they tend to create a loop which orients on a plane in 3-dimensional space. The main goal of this thesis was to use analysis techniques related to this law to gain a better understanding of how the central nervous system coordinates and implements anticipatory locomotor adjustments (ALAs) for stepping over obstacles in healthy and pathological populations. Study 1 examined general issues of normal ALAs, while study 2 related the timing and amplitude of segment motion with mechanical work by hip and knee flexor muscles. Studies 3 and 4, determined how impairments such as autosomal recessive cerebellar ataxia type-1 (ARCA-1) and a previous stroke respectively affect locomotor control. In healthy adults, phasing differences between adjacent segments were shown to be correlated to characteristics of the plane formed by the segment elevation angles and these phase differences changed systematically with increasing obstacle height. It was proposed that the CNS adjusts a basic locomotor pattern for environmental constraints by manipulating elevation angle phase differences between adjacent segments as well as elevation angle amplitudes. The follow-up study determined that as higher obstacles were cleared, leading limb thigh phase lead and trailing limb shank phase lag increased. The work done by the hip and knee flexor muscles influenced thigh elevation differently in the leading and trailing limbs and it was concluded that these muscles do not have simple specific roles in elevating and progressing the lower limb during locomotion. Instead, these muscle powers may result from elevation angle waveform control dynamics. When observing coordination in the ARCA-1 and stroke participants, plotted segmental elevation angle trajectories continued to covary on a plane. In the ARCA-1 participants, a larger phase difference between the thigh and shank segments was suggested to be a voluntary control mechanism to increase toe clearance over obstacles. In participants with a previous stroke, this phase difference was greater in the non-paretic limb which was interpreted as being necessary to elevate this limb in compensation for poor support by the paretic limb. The results of this thesis suggest mechanisms of segment elevation control to implement ALAs and highlights voluntary compensatory mechanisms in such control in pathological populations.

Preface

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This thesis is comprised of four manuscripts (Chapters 3-6) prepared for scientific publication. The following text provides information on the titles, co-authors, role of the student, and any modifications from the original for each individual manuscript.

Chapter 3 – Study 1

Titled: *"Segmental control for adaptive locomotor adjustments during obstacle clearance in healthy young adults"*

Primary Author: M J MacLellan
Coauthor: B J McFadyen
Affiliation: Centre for Interdisciplinary Research in Rehabilitation and Social Integration, Department of Rehabilitation, Faculty of Medicine, Laval University, Quebec, Canada
Role of the Primary Author: protocol design, data collection, data and statistical analysis, and writing of the manuscript under the supervision of the coauthor.
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Chapter 4 – Study 2

Titled: *"Relationships between segment elevation angles and muscle power during obstacle clearance reveal multiple goals for the mechanical work at hip and knee joints"*

Primary Author: M J MacLellan
Coauthor: B J McFadyen
Affiliation: Centre for Interdisciplinary Research in Rehabilitation and Social Integration, Department of Rehabilitation, Faculty of Medicine, Laval University, Quebec, Canada
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Chapter 5 – Study 3

Titled: *"Increased obstacle clearance distance in people with ARCA-1 results in part from coordination changes between the thigh and shank segments"*

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Chapter 6 – Study 4

Titled: *"Comparison of locomotor control mechanisms for segmental coordination between non-paretic and paretic limbs during obstacle clearance following stroke"*

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With love to Catherine and Veronica

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List of Abbreviations

3D – Three-Dimensional

A1 – 1st ankle muscle power burst

A2 – 2nd ankle muscle power burst

ALA – Anticipatory Locomotor Adjustment

ANOVA – Analysis of Variance

ARCA-1 – Autosomal Recessive Cerebellar Ataxia Type 1

CA – Cerebellar Ataxia Group

CNS – Central Nervous System

COM – Centre of Mass

COP – Centre of Pressure

CPG – Central Pattern Generator

EMG - Electromyography

H – Healthy Adult Group

H1 – 1st hip muscle power burst

H2 – 2nd hip muscle power burst

H3 – 3rd hip muscle power burst

H3D – delayed 3rd hip muscle power burst

ICARS – International Cooperative Ataxia Rating Scale

IRDPQ – l'Institut de réadaptation en déficience physique de Québec

iRED – Infrared Emitting Diode

K1 – 1st knee muscle power burst

K2 – 2nd knee muscle power burst

K3 – 3rd knee muscle power burst

K3D – delayed 3rd knee muscle power burst

K4 – 4th knee muscle power burst

K5 – 5th knee muscle power burst

MLR – Mesencephalic Locomotor Region

MV – Matched Velocity Group

PCA – Principal Component Analysis

Chapter 1: Introduction

1.0: General Introduction

Human locomotion is usually adapted in order to arrive at a desired location. For example, to arrive at a goal, a person may need to walk around an object, step up to a new level, or step over an obstacle. Such tasks can be hazardous if not executed properly, which in turn could lead to falling. In particular, the task of stepping over obstacles can be hazardous due to the risk of tripping if the foot comes into contact with the obstacle. One of the risk factors associated with this task could be poor inter-joint coordination in the lower limb; the joints in the lower limb need to be controlled with the utmost accuracy to avoid falling. This chapter will review how successful obstacle clearance is accomplished in healthy adults and 2 pathological populations (people with cerebellar ataxia and people who have sustained a previous stroke). It will also describe a paradigm for coordination analysis used in human locomotion. Following this review of the literature, 4 studies will be presented that examine lower limb coordination and discuss the underlying central nervous system (CNS) control mechanisms during the task of obstacle clearance. One study will examine the adaptations of kinetic patterns in order to elevate and progress the lower limb during obstacle clearance, with the three others using the planar law of intersegmental coordination to study CNS control in obstacle clearance in healthy adults, adults with cerebellar ataxia, and adults with a previous stroke.

1.0.1: Statement of Problem

The following review of literature will present current knowledge about kinematic and kinetic patterns of human locomotion during level walking and obstacle clearance. Previous studies suggest that specific joints may have roles in coordinating elevation and progression of the lower limb during locomotion (Niang & McFadyen, 2004). Although this idea has been suggested, there has been no study testing this hypothesis specifically. This review will also discuss current methods used in analyzing inter-joint coordination during these tasks in healthy adults, adults with cerebellar ataxia, and adults with a previous stroke. The review of literature will show that coordination analyses previously used for locomotion are descriptive in nature and only provide some insight into the underlying CNS control of movement. Such insight may provide evidence of specific variables which are controlled by the CNS in order to properly coordinate locomotion. The following review of the literature will also introduce a paradigm used in coordination analysis (the planar law of intersegmental coordination) which has been related to CNS control mechanisms and will subsequently be used to analyze inter-joint coordination during obstacle clearance in the present thesis.

1.1: Kinematics and Kinetics during Level Walking

During the human gait cycle, there are 2 main phases, the stance phase (while the foot is in contact with the surface, taking up approximately 60% of the gait cycle) and the swing phase (while the foot is airborne prior to subsequent contact, taking up approximately 40% of the gait cycle). A short amount of time (approximately 20% of the gait cycle) where both feet are in contact with the ground, termed double support, is the most stable part of the gait cycle. These important terms, as well as others described below, will be used throughout this thesis when describing locomotion.

Studies of human locomotion have provided us with information on lower limb joint angles, moments, and powers during a normal gait cycle. The gait cycle begins with heel contact, the instant in which the foot makes contact with the surface. At this time, the knee and ankle begin to flex, accepting the weight of the body following contact. Stance phase then continues with the hip extending in order to progress the upper body in a forward motion. As the body continues to progress over the foot (with the ankle dorsiflexing), the hip continues to extend during a period known as mid-stance. Prior to the end of stance, the hip and knee begin to flex while the ankle starts to plantarflex. Plantarflexion at the ankle leads to a forward progression of the body, which is termed push-off and continues until toe-off, the point at which the toe leaves the surface. At this point, the hip and knee continue to flex and the lower limb moves upwards and forwards (early swing), which is followed by an extension of the knee joint to prepare for heel contact (late swing) thereby ending the gait cycle. This summarizes the joint angular kinematics during a gait cycle.

Although joint kinematics describe motion at each joint, muscle power (an energetic measure) is an important measure because, taken into account with the muscle moment and angular motion of a joint, it provides insight into the mechanical cause of movement (Winter, 2005). Muscle power analyses are informative since they determine how energy is partially exchanged in the system. A positive muscle power represents a generation of energy by the muscles while a negative muscle power represents absorption of energy by the muscles. During the gait cycle, there are a variety of muscle power bursts which are important in the control of gait (and will be referred to throughout this thesis). A well accepted form of identifying these muscle power bursts is related to the joint where the burst occurs and the order that it occurs across the gait cycle (Winter & Sienko, 1988). These bursts are briefly described below.

At the ankle, there are 2 main power bursts. The first power burst (A1) is a negative energy absorption period by the ankle plantarflexors as the shank (and body) rotates forward over the foot. This is followed by A2, a large power generation burst by the plantarflexors as the foot pushes off the surface, generating the energy to move the body forwards. The knee has 4 main power bursts. The initial power burst (K1) is an absorption burst by the knee extensors as the knee flexes when accepting the weight of the body. K2 is a generation burst by the knee extensors as the knee extends during stance to raise the centre of gravity of the body. This is followed by K3, an absorption burst by the knee extensors as the knee flexes during foot push-off and early swing. The final power burst (K4) is an absorption burst by the knee flexors at the end of swing to decelerate the moving shank and foot segments. At the hip, an initial power generation by the hip extensors (H1) occurs when accepting the weight of the body as the hip extends. H2 is an absorption burst by the hip flexors in order to decelerate the thigh throughout stance phase. Finally, H3 is a power generation burst by the hip flexors as the hip flexes prior to toe-off and through swing to progress the limb upwards and forward.

To date, locomotor patterns have predominately been described at each lower limb joint separately. What is interesting (and much less understood) is how the CNS coordinates these joints together in an orderly fashion. Bernstein (1967) described coordination as a system of elements achieving a common goal. In the case of locomotion, lower limb joints are coordinated to support and progress the body in a forward direction. There are many types of coordination that can be examined and this thesis will focus on inter-joint coordination: how the lower limb joints (hip, knee, and ankle in each limb separately) function together during locomotion.

Inter-joint coordination during human locomotion has been analyzed in a variety of ways. One of the more popular methods is through angle-angle graphs which were developed by Greive (1969). In these graphs, joint angles are plotted against each other (for example: knee angle vs. hip angle). During a gait cycle in human locomotion, angle-angle plots form loops which provide a description of the underlying inter-joint coordination. Winstein & Garfinkel (1989) described typical shapes seen in such angle-angle plots and the interpretations of each of these shapes. Horizontal or vertical segments (straight lines) indicate that one joint is changing while the other remains constant. A diagonally oriented line in the positive direction indicates that the two joint angles are being coordinated in-phase at a constant rate, while a negative diagonal line indicates joint motion is out of phase. If both joints reach a maximum and switch directions simultaneously, this represents a “turning-point” where inter-joint movements change direction in a coordinated fashion. Finally, a rounded trajectory suggests the joints have a phase offset and the coordination is decoupled. Along with these patterns, changes in the area confined by the loop and the perimeter can provide information regarding coordination. Hershler & Milner (1980) stated that the area confined by angle-angle plots indicate the total possible joint range of motion (eg. a greater area would indicate a larger range of motion). Hershler & Milner (1980) also stated that the perimeter of angle-angle plots can represent how smooth the inter-joint coordination is (eg. a larger perimeter caused by multiple joint reversals creating a jagged loop indicates jerky or uncoordinated movements by the joints).

In healthy adults, hip-knee angle-angle plots have a distinct turning point, which signifies heel contact and when the extending knee begins to flex and the flexing hip begins to extend simultaneously (Steiner, Capildeo, & Rose, 1982). Following heel contact, the hip extends with the knee joint relatively constant (producing a straight line on the angle-angle plot) which is followed by flexion of the knee that reaches a peak in swing where extension of both joints begins to prepare for subsequent heel contact (Hershler & Milner, 1980). The hip angles change relatively slowly during knee extension/flexion, creating a pyramid shaped loop (Hershler & Milner, 1980). In knee-ankle angle-angle plots, a diagonal line (with negative slope) is shown during stance indicating that motion about the knee and ankle is 180 degrees out of phase (Winstein & Garfinkel, 1989). At pre-swing, a turning point occurs where the knee and ankle angles reach a maximum and turn to flexion and extension simultaneously, followed by a straight diagonal trajectory (negative slope) again suggesting constant relationship between knee flexion and ankle extension (Winstein & Garfinkel, 1989). In early swing, a rounded segment is presented suggesting that knee and ankle motion is phase offset and late swing is characterized by a vertical line indicating the knee is extending when the ankle angle is constant (Winstein & Garfinkel, 1989).

Inter-joint coordination has also been examined in healthy adults using relative phase analysis. Relative phase analysis uses dynamic systems theory to describe the motion of joints or segments in the body. Phase plots consist of plotting a segment's angular position with the segment's angular velocity. Clark & Phillips (1993) concluded that when this method is used to analyse the lower limb segments during locomotion, the behaviour is similar to that of a limit cycle (a cyclic system that loses and gains energy for the cyclic behaviour to continue). In order to analyze the coordination of 2 joints or segments, the phase angle of one joint or segment is subtracted from the phase angle of the second joint or segment at each point in time, yielding a measure of relative phase (Barela, Whitall, Black, & Clarke, 2000). This type of analysis is beneficial since it compresses 4 variables (the position and velocity of 2 joints or segments) into 1 variable (Barela et al., 2000). This 1 variable can then be used to study the coupling of 2 joints or segments during a multi-joint movement and determine if the joints or segments are moving together (in-phase) or in opposite directions (out of phase) (Burgess-Limerick, Abernethy, & Neal, 1993).

Relative phase analysis has previously been used to detail the coordination of the thigh and shank segments during locomotion in healthy adults. Clark & Phillips (1993) were the first to conclude that there is a complex non-linear relationship between these segments during locomotion. Following heel contact, the thigh segment tends to lead the shank segment and the magnitude of this lead varies throughout stance (Clark & Phillips, 1993; Kurz &

Stergiou, 2002). Although there is a greater out of phase relationship between the segments earlier in stance (Kurz & Stergiou, 2002). At toe off, the thigh angle phase is approximately 90° ahead of the shank segment and this lead is maintained throughout the first half of the swing phase (Clark & Phillips, 1993). The relationship between the segments becomes less out of phase when the thigh segment reverses forward swing (Clark & Phillips, 1993). Near the end of swing, the shank segment begins to swing in the posterior direction and movement is in phase with the thigh segment which continues for a short amount of time until the shank segment leads that of the thigh before heel contact (Clark & Phillips, 1993). These observations led Clark & Phillips (1993) to conclude that the intersegmental relationship appears to be one in which the thigh acts as a forcing oscillator that drives ahead, stops, and changes direction with the shank segment following its action.

Other methods of coordination analysis exist, and one termed the planar law of intersegmental coordination will be given in greater detail below. However, to now, one can see that there is a complex interaction of joint movements during human locomotion that are performed in a highly coordinated manner. It is thought that the control of this movement originates in the CNS by a locomotion pattern generator. This is explored next.

1.1.1: Central Nervous System Control of Level Walking

Studies have shown that much of the information needed to generate a locomotor pattern is present in the spinal cord. This idea originated from work by Graham Brown (1911) when observing muscle activations in the cat. In these studies, Brown transected all lumbarsacral dorsal roots as well as performed a complete transection of the cat spinal cord. Following the transection, alternating rhythmic contractions of ankle flexors and extensors were observed in the cat hindlimbs suggesting the isolated spinal cord has the ability to generate rhythmic muscle activation patterns. Brown noted that flexor activation of one limb coincided with extensor activation of the other and hypothesized a “half-centre” model for this muscle activation where one muscle activation centre existed for each limb and this centre divided into 2 parts, one for flexors and one for extensors. Each of these half-centres would be connected with inhibiting pathways whereby the activation of one half-centre would also send strong inhibition to the opposing half-centre.

These results lead to further studies examining locomotor pattern generators in cats (Grillner & Zangger, 1975, 1984). In these studies, mesencephalic cats (cats with a transection located at the level of the midbrain, therefore preventing the cerebral cortex from influencing the locomotor pattern) walked on a treadmill while the weight of their body was supported. In order for locomotion to occur, electrical stimulation was administered to the mesencephalic locomotor region (MLR) which has been shown to produce walking in such cat preparations (Shik, Severin, & Orlovsky, 1966). Locomotion was observed in these cats following ipsilateral (Grillner & Zangger, 1975, 1984) and bilateral (Grillner & Zangger, 1984) transection of the spinal cord dorsal roots in order to remove kinesthetic sensory information from the limbs. Following transection, it was shown that the characteristic flexion and extension muscle activation patterns remained in order for locomotion to continue, although there was greater variability in the timing of the muscle activation patterns. As well, muscle activation patterns were not necessarily simple alternations of flexor and extensor activity as suggested by Brown (1911). For example, Grillner & Zangger (1984) showed that the semitendinosis muscle retains a double burst pattern following dorsal root transection.

From the results of these studies, it was concluded that some muscles in the limbs required for locomotion are programmed by a central pattern generator (CPG) which is not dependent on afferent information originating from the limbs (Grillner & Zangger, 1975, 1984). It was further suggested that this central program does not simply alternate the activation of flexor and extensor muscles, but activate and terminate muscle activations with precise timing (Grillner & Zangger, 1975, 1984). Although these studies show that afferent information is not required to generate a basic locomotor pattern, it is generally agreed that sensory information and supraspinal signals are required for fine tuning the locomotor pattern and adapting the locomotor pattern for the environment (Delcomyn, 1980; Dietz, 1992; Duysens & Van de Crommert, 1998; Grillner & Zangger, 1975, 1984; Shik & Orlovsky, 1976).

Even though a large body of literature exists supporting the theory of a locomotor CPG in cats, there is much less evidence supporting this theory in humans. This is due in part to the populations used to study such phenomenon and the general complexity of the human CNS. Since much of the literature supporting a locomotion CPG comes from cats with transected spinal cords, the human equivalent used is people that have been in accidents in which the spinal cord was partially or completely severed (Bussel et al., 1988; Calancie et al., 1994). In these studies, rhythmic activity of the lower limb flexors and extensors were observed following spinal cord injury. These studies concluded that the lower limb activity in the participants was completely generated by the spinal cord since the activity appeared below the level of spinal cord transection, therefore providing evidence of a locomotor CPG in humans. Since the mechanism of injury is trauma in such populations, the spinal transections studied in humans are not as controlled as in cats. These injuries are generally associated with multiple traumas to the spinal cord and other CNS structures. Although these studies do have limitations, they are one of the better human models we have to study the existence of locomotion CPGs in humans.

A body of literature examining locomotion in infants also supports the idea of a locomotor CPG in humans. Forssberg (1985) observed stepping in infants when they were held over a moving treadmill. Although this locomotor pattern differed from that of an adult (exaggerated flexion of hip and knee, large variability between infants and between steps), there was a large degree of synchrony of lower limb muscle activation (Forssberg, 1985). Forssberg (1985) argued that these results are proof of the existence of an innate locomotor rhythm generator which is genetically coded at birth and subsequently develops into a matured form of locomotion. Further studies have shown that infants are also able to walk on split-belt treadmills in which detailed adjustments of stance and swing time need to be made (Thelen, Ulrich, & Niles, 1987; Yang et al., 2004; Yang, Lamont, & Pang, 2005). Infants are also able to walk on these treadmills when the belts are moving in opposite directions (Yang et al., 2005) and have led to arguments that a CPG exists for each limb (Yang et al., 2004; Yang et al., 2005). These studies have supplemented the research arguing that CPGs are present in the human spinal cord, but there has not been any definitive work on the subject.

Although this brief review provides evidence suggesting the presence of a locomotor CPG, the fact remains that we have no definitive proof this does exist in humans. Even so, many theories of locomotor control are based upon the assumption that human locomotion is controlled by a CPG. These basic locomotion patterns must be adapted in order to continue locomotion when changes occur in the environment.

1.2: Kinematic and Kinetics during Adaptive Locomotion (obstacle clearance)

Adapting locomotion is of utmost importance to daily life; it is very rare that one performs only straight-forward level walking all of the time. Locomotor patterns need to be adapted in order to circumvent objects, step up to a new level, or to step over an obstacle. When a person changes locomotor dynamics due to a foreseeable environmental change or a change in movement goal, this is called an anticipatory locomotor adjustment or ALA (McFadyen & Carnahan, 1997). It is thought that ALAs are an adjustment of the basic locomotor pattern rather than an implementation of an entirely new locomotor pattern (McFadyen, Magnan, & Bouchard, 1993; McFadyen & Winter, 1991; McFadyen, Winter, & Allard, 1994; Taga, 1998). This section will describe ALAs related to stepping over obstacles.

Obstacle clearance is important in the study of adaptive locomotion since it is a task that people accomplish on a daily basis. In order to successfully step over an obstacle, the leading and trailing limbs increase hip and knee flexion (Chou & Draganich, 1997; Patla, Prentice, Robinson, & Neufeld, 1991) which results in a toe clearance of approximately 0.10 m (Patla & Rietdyk, 1993). As the height of an obstacle increases so does flexion of the hip, knee, and ankle (Austin, Garrett, & Bohannon, 1999; Chou & Draganich, 1997; Patla & Rietdyk, 1993). Increasing obstacle depth results in less flexion at the knee and ankle during clearance (Patla & Rietdyk, 1993). Joint angle patterns also change depending on the proximity of the obstacle. During unilateral obstacle clearance, hip flexion decreases when an obstacle is closer in proximity (McFadyen et al., 1993), while a closer obstacle leads to decreases in flexion of the hip, knee, and ankle in the trailing limb (Chou & Draganich, 1998a, 1998b). Evidence has shown that the trailing limb, when stepping over an obstacle, is controlled independently of the leading limb (Patla, Rietdyk, Martin, & Prentice, 1996) and differences do exist between leading and trailing limbs. One main difference is the use of visual information during clearance in each limb; only the leading limb has the ability to use on-line vision to make trajectory corrections (Mohagheghi, Moraes, & Patla, 2004). In general, toe clearance in the leading limb has been shown to increase when stepping over higher obstacles (Austin et al., 1999; Patla & Rietdyk, 1993) while no changes in clearance distance are present in the trailing limb when stepping over obstacles of different sizes (Chou & Draganich, 1997), although clearance has been shown to increase when the foot is placed further from the obstacle in the leading (McFadyen et al., 1993) and trailing (Chou & Draganich, 1998b) limbs.

The noted kinematic changes in the leading and trailing limbs during obstacle clearance are the result of a reorganization of muscle powers during clearance. Specifically, K3 and H3 muscle powers decrease while a new knee flexor power generation (K5) appears in order to flex the knee and hip (McFadyen & Winter, 1991) with the hip being flexed through distal intersegmental forces at the thigh (Patla & Prentice, 1995). This has been confirmed through computational modeling (McFadyen et al., 1994). As obstacle height increases so does K5 power, while no changes are observed in hip power generation (Niang & McFadyen, 2004). The appearance of a K5 generation burst to elevate the limb over the obstacle also occurs in the trailing limb and similar increases in this burst with obstacle height have been shown (Niang & McFadyen, 2004). However, there is a delay of K3 power (which was referred to as K3D by Niang & McFadyen, 2004) and H3 is separated into H3 (which decreases as obstacle height increases) and H3D (delay H3 burst which increases as obstacle height increases) (Niang & McFadyen, 2004). The differences observed in muscle powers were suggested to occur due to the specific trajectory of the trailing limb; initially elevating the limb using the knee (K5) and progression of the limb through swing with the hip H3D power burst after the limb has been elevated (Niang & McFadyen, 2004).

This suggestion of the trailing limb having two separate goals during swing introduces the idea of joint specificity where the knee has a specific role in elevating the foot while the hip has a specific role in progression of the limb through swing. Using forward dynamic modelling, Neptune, Zajac, & Kautz (2004) have suggested that the hip flexors (iliacus and psoas in particular) accelerate the hip into flexion and swing the limb forwards in pre-swing and into swing, therefore supporting this hypothesis. Joint specificity may be a method of how the CNS coordinates lower limb movement when stepping over obstacles. This type of control may be seen in the cat, whereby specific pyramidal tract neurons are activated sequentially in order to first elevate the paw then flex the elbow to swing the limb when the forelimb steps over an obstacle (Drew, Andujar, Lajoie, & Yakovenko, 2008; Lavoie & Drew, 2002). A similar hypothesis has been made where a "shoulder-centreed" pattern is used to control arm movements during horizontal reaching (Galloway & Koshland, 2002). Aside from these studies, not much research has been conducted on this topic. It would be important to identify if limb movement is controlled in this way because it would have important applications for mobility rehabilitation and understanding motor coordination.

Inter-joint coordination has been examined in obstacle clearance using methods related to relative phase analysis. These studies have shown that hip-knee and knee-ankle joint pairs rotate in-phase throughout swing when clearing an obstacle in the leading and trailing limbs (Lu, Yen, & Chen, 2008; Yen, Chen, Liu, Liu, & Lu, 2009). Lu et al. (2008) noted that the only out of phase joint movements were observed in the stance limb knee-ankle joint pair when the contralateral limb was clearing the obstacle (for both leading and trailing limbs). This out of phase movement was accounted for by the modulation of full body movement rotating over the ankle joint in stance thereby maintaining stability (Lu et al., 2008). These authors did not show any significant differences in joint angle relative phase between obstacle heights ranging from 10-30% of each participant's leg length (Lu et al., 2008; Yen et al., 2009). Similar results have been shown using a relative phase analysis on segment elevation angle pairs when healthy adults cleared an obstacle while running (Stergiou, Scholten, Jensen, & Blanke, 2001). Differences in relative phase between leading and trailing limbs were observed when examining the variability of relative phase plots; the variability of relative phase was smaller in the leading limb when compared to the trailing limb and this was attributed to a lack of visual feedback available in the trailing limb (Lu et al., 2008). This type of analysis for the examination of inter-joint coordination during obstacle clearance is questionable since it was unable to discriminate coordination differences between differing obstacle heights even though it has been shown that joint angle amplitudes change drastically during such movements (as discussed above). Due to these facts, perhaps other forms of coordination analysis which are more sensitive to inter-joint coordination changes should be used to examine obstacle clearance.

1.2.1: Central Nervous System Control of Adaptive Locomotion

As previously stated, it is generally agreed that sensory information and supraspinal signals are required for fine tuning the locomotor pattern and adapting the locomotor pattern for the environment. For example, observations have shown that limb sensory information can modulate the output from the CPG therefore adapting the locomotor pattern. When tactile stimulation is applied to the dorsal hindlimb paw during swing phase in spinal cord transected cats, the entire limb enhances flexion such as during obstacle clearance, followed by a continuation of the walking pattern (Forssberg, Grillner, & Rossignol, 1975, 1977). Interestingly, this response tends to change when stimulation occurs during stance phase; large limb extensor muscle activation is seen, which has been referred to as a reflex reversal (Forssberg et al., 1975, 1977). Similar muscle activations have been observed during fictive locomotion in the cat (Andersson, Forssberg, Grillner, & Lindquist, 1978). This reflex reversal is suggested to be due to phasic facilitation and inhibition of neural pathways which is dependent on the current signals being generated by the CPG while activating muscle groups sequentially for continuation of the gait cycle (Forssberg et al., 1975, 1977). Since these cats could not use supraspinal input (due to spinal cord transection), this evidence shows that sensory information is useful for the CPG when encountering unexpected perturbations to locomotion.

Supraspinal signals have been shown to be important in altering muscle activation patterns produced by the CPG and for adapting locomotor patterns to the environment. An example of adapting the locomotor pattern to the environment comes from the work of Drew (1988), who recorded pyramidal tract neuron activity as cats stepped over obstacles of 6 different shapes. The author observed that discharge rates in recorded pyramidal tract neurons tended to increase when the cat stepped over the obstacle. Using a correlation analysis, it was shown that some cells may be related to muscle activity at specific joints. Since the firing of the cells occurred prior to activation of the muscles, it was concluded that these cells have a role in the production of this change in movement required when stepping over the obstacle (Drew, 1988). Similar results were observed by Widajewicz, Kably, & Drew (1994) using a more extensive analysis of pyramidal tract cell discharge during a similar protocol. In this study, differing cells seemed to be related to either leading or trailing hindlimb muscle activation and some cells tended to be related to different phases of obstacle clearance; cells tended to be related to swing of the limb over the obstacle, subsequent stance phase following the step over the obstacle, or the gait cycle prior to gait modification. These results suggested that certain pyramidal tract cells were related to specific tasks such as the modification of limb flexor activity to elevate the limb over the obstacle, positioning of the foot following clearance, and stabilizing the body when other limbs are crossing the obstacle (Widajewicz et al., 1994). A small number of cells were also suggested to be related to appropriately timing the delay between forelimb and hindlimb clearance (Widajewicz et al., 1994). These results provide evidence that supraspinal signals are necessary to alter the centrally generated locomotor pattern for the environment (Drew, 1988; Widajewicz et al., 1994). Widajewicz et al. (1994) suggested the mechanism for this control to be through descending signals acting on interneuronal populations within the lumbar spinal cord that may be related to the locomotor CPG. Widajewicz et al. (1994) argued that this mode of action would allow the motor cortex to provide precise control over the modified electromyography (EMG) activity in the limb when ensuring the integration of this control into the centrally generated pattern.

The previous sections have discussed limb biomechanics and neural control of level walking and obstacle clearance in healthy adults and cats. These patterns tend to change in certain pathological conditions. The following section will detail these changes in two pathological populations; adults with a previous stroke and adults with cerebellar ataxia.

1.3: Pathological Locomotion

Locomotor coordination can be affected by certain pathological conditions. An example of such a pathological condition is cerebellar ataxia: a pathological condition affecting the cerebellum. The cerebellum contributes to movement since it is an area that compares information regarding intention of movement and movement execution (Kandel, Schwartz, & Jessell, 2000). People with cerebellar ataxia display symptoms of ataxic gait, which can be simply described as uncoordinated gait. Another example is a previous stroke. People with a previous stroke tend to have asymmetries in their locomotor patterns due to specific brain areas affected by the disorder. Generally, a locomotor task such as stepping over an obstacle can be hazardous in these populations due to changes in stability and a greater risk of tripping. Many studies examining coordination in these populations have used simple kinematic and kinetic analyses which are limited in their description of coordination. The following sections will detail locomotor coordination changes in people with cerebellar ataxia and stroke.

1.3.1: Cerebellar Ataxia

The cerebellum is located in the posterior section of the brain and heavily influences human movement. One of the roles of the cerebellum is to evaluate differences between intention (perceived goal of a movement) and action (actual movement created), though the precise contribution of the cerebellum on movement is unknown (Kandel et al., 2000). The cerebellum contains the vermis, which is a section in the middle of the cerebellum and 2 lateral sections (called cerebellar hemispheres) (Mariotti, Fancellu, & Di Donato, 2005). Functionally, the cerebellum can also be separated into 3 parts: the archicerebellum, paleocerebellum, and neocerebellum (Mariotti et al., 2005). The archicerebellum consists of the flocculonodular lobe and is connected to the vestibular and visual systems in order to coordinate head and eye movements (Mariotti et al., 2005). The paleocerebellum receives signals from the spinal cord and is involved with controlling muscle tone during postural and locomotor activities (Mariotti et al., 2005). Finally, the neocerebellum connects to the pons and cerebral cortex and is related to the planning, initiation, and regulation of movement (Mariotti et al., 2005). Mariotti et al. (2005) define cerebellar ataxia as a "neurological dysfunction of motor coordination which may affect activities such as gaze, speech, gait, and balance". The causes of cerebellar ataxia include toxic, metabolic, immune, and genetic origins (Mariotti et al., 2005).

A type of cerebellar ataxia that will be focused on in the research presented in this thesis is autosomal recessive cerebellar ataxia type 1 (ARCA-1), or recessive ataxia of Beauce. This is an inherited form of cerebellar ataxia that presents itself at middle age and is a slow progressing disorder that evolves into a moderate level of disability (Dupre et al., 2007;

Gros-Louis et al., 2007). The symptoms of this specific form of cerebellar ataxia include significant dysarthria, dysmetria, brisk lower extremity tendon reflexes, and minor abnormalities in saccade and smooth pursuit eye movements (Dupre et al., 2007; Gros-Louis et al., 2007). An interesting note about this form of ataxia is that it is a pure form of cerebellar ataxia and imaging studies have shown diffuse atrophy of the cerebellum with no cerebral cortex, midbrain, pontine, bulbar, or inferior olive atrophy (Dupre et al., 2007). Along with this, there are no extrapyramidal signs or signs of cognitive loss, retinopathy, cardiomyophathy, sensory abnormalities, or autonomic disturbances (Dupre et al., 2007). This makes such a population advantageous in studying cerebellum disorders since most previous research involves participants with many forms of cerebellar ataxia of differing origins affecting various locations on the cerebellum.

1.3.1.1: Assessing Cerebellar Ataxia Severity using the International Cooperative Ataxia Rating Scale

The International Cooperative Ataxia Rating Scale (ICARS) was developed in 1997 by a committee from the World Federation of Neurology (Trouillas et al., 1997). At the time, a number of different tests were used to quantify cerebellar ataxia such as finger tapping, pegboard tests, and pronunciation tests. The committee created the ICARS in order to have a standard in cerebellar ataxia testing for use in controlled drug studies that may be multi-centre or international collaborations. The main criterion in the development of the ICARS was to translate symptomology into quantifiable scores which would be accurately defined to reduce inter-observer bias. It was also decided that a 100 point scale would be used so a percentage of seriousness can be interpreted. The developed test was divided into four subsections which consisted of items related to posture and stance disturbances, limb movement disturbances, speech disorders, and oculomotor disorders. The test is heavily weighted on the limb movement and posture/stance disturbances. The ICARS test has been shown to have high inter-rater, test-retest, and internal reliability (Schmitz-Hubsch et al., 2006; Storey, Tuck, Hester, Hughes, & Churchyard, 2004), but there is a debate as to the validity. Tison et al. (2002) have shown the ICARS to have strong construct and internal validity in a population with cerebellar focused multiple systems atrophy, while Schmitz-Hubsch et al. (2006) indicated much weaker internal validity in a population with spinocerebellar ataxia. These studies suggest that the validity of the ICARS may be dependent on the cerebellar disorder in question. Even though these suggestions have been made, many studies observing gait in people with cerebellar ataxia have made reference to ICARS testing. For example, Ilg, Golla, Thier, & Giese (2007) have shown that the complete ICARS score is highly correlated to increases in step width in people with cerebellar ataxia. As well, the ICARS posture subscore is highly correlated with changes in step width, swing phase duration, lateral body sway, gait velocity, and step length while the ICARS kinetic subscore is highly correlated to variability in swing phase duration.

1.3.1.2: Cerebellar Ataxia and Locomotion

Previously studied forms of cerebellar ataxia have been associated with decreased gait velocity, increases in gait cycle time (stance and swing), reduced step length, (Earhart & Bastian, 2001; Ilg, Giese, Gizewski, Schoch, & Timmann, 2008; Morton & Bastian, 2004, 2007; Palliyath, Hallett, Thomas, & Lebiedowska, 1998), increased step width, increased lateral sway (Ilg et al., 2007), and a drifting in direction (Hallett & Massaquoi, 1993) during level walking. Palliyath et al. (1998) also noted that heel off and toe off times were delayed in the gait cycle, but related this to the slower gait velocity in the population. When walking on a treadmill, Stolze et al. (2002) concluded that people with cerebellar ataxia walked with a decreased cadence, increased stance time, increased double support time, increased step width, and had larger ankle external rotation. Stolze et al. (2002) suggested that the changes in step width and ankle external rotation were compensatory mechanisms in order to increase stability. One of the main symptoms of ataxic gait is variability. People with cerebellar ataxia show increased variability in step length, stride length, stride length symmetry, cadence, step time, foot clearance during swing, and stance time during level walking (Palliyath et al., 1998) and increased variability in step length, step height, ankle angle when walking on a treadmill (Stolze et al., 2002). Although a direct comparison was not made, Palliyath et al. (1998) suggested that changes in some of these general gait variables were due to the decreased gait velocity in participants with cerebellar ataxia.

During the gait cycle, some changes do occur in the lower limb joint angles in people with cerebellar ataxia. Many studies have suggested that amplitudes of lower limb joint angles do not differ in a cerebellar ataxic population when compared to healthy adults (Earhart & Bastian, 2001; Stolze et al., 2002) although Palliyath et al. (1998) showed that the range of motion at the ankle decreases in participants with cerebellar ataxia and Morton & Bastian (2003) showed that peak ankle plantarflexion, knee flexion, and hip flexion decreased in cerebellar participants with primarily a balance deficiency. Palliyath et al. (1998) also showed that time of ankle plantarflexion and peak knee flexion in swing is delayed in a cerebellar ataxic population. Similar to the previously mentioned gait characteristics, people with cerebellar ataxia show greater variability in joint angle trajectories. Earhart & Bastian (2001) noted increased variability in hip and knee joint angle trajectories when participants with cerebellar ataxia stepped on inclined surfaces. Palliyath et al. (1998) also showed greater variability in ankle range of motion, ankle angle at heel strike, foot clearance during swing, knee range of motion during stance, and time to peak knee flexion in swing in level walking.

The use of joint angle-angle plots in locomotion analysis has been very important in illustrating changes in coordination that occur in people with cerebellar ataxia. One of the main symptoms of cerebellar ataxia is decomposition of movement (or joint decomposition); the breaking down of multi-joint movement to single joint movements. Bastian, Martin, Keating, & Thach (1996) suggested from reaching studies that decomposition of movement is a compensatory mechanism since it is most pronounced during accurate movements and is due to inappropriate control of interaction torques across joints. Imaging studies have suggested that damage to the intermediate zone of the cerebellum is most likely related to decreased inter-joint coordination in human locomotion (Ilg et al., 2008). Morton & Bastian (2003) have shown that hip-knee joint decomposition was increased in cerebellar ataxic participants classified as having a leg-placement deficit (determined from a visually guided stepping task). As well, Palliyath et al. (1998) have shown that hip-knee angle-angle plots in people with cerebellar ataxia are compressed in the hip angle direction, suggesting a reduced hip range of motion. Also, the hip-knee loop is not as round due to late knee flexion compared to hip flexion during locomotion (Palliyath et al., 1998). Knee-ankle plots of cerebellar ataxic participants have shown several changes in direction (jagged shape of loop) due to rapid reversal of angular movement at one or both joints (Earhart & Bastian, 2001) and a flattened loop as a result of reduced range of motion at the ankle and delay of ankle movement in swing. Again, these angle-angle plots show a large amount of variability when compared to a healthy population (Ilg et al., 2007; Ilg et al., 2009; Stolze et al., 2002). In particular, Ilg et al. (2007) have used analysis techniques that show increased temporal variability of all angle-angle pairs and this separated the cerebellar ataxic population from populations with other balance deficits (Parkinson's disease and peripheral vestibular failure). It is interesting to note that Ilg et al. (2009) showed that with a rehabilitation program, participants with cerebellar ataxia were able to decrease this temporal variability and perform more coordinated movements.

1.3.1.3: Cerebellar Ataxia and Obstacle Clearance

Presently, Morton, Dordevic, & Bastian (2004) have conducted the only study which examines obstacle clearance in participants with cerebellar ataxia. In this study, an obstacle was placed to one side of the participant so that only the leading limb cleared the obstacle. It was observed that participants with cerebellar ataxia exhibited hypermetria (overshooting of toe clearance while clearing the obstacle) and this was attributed to an increase in knee flexion during clearance. In order to gain insight into why these participants exhibited hypermetria, Morton et al. (2004) also observed control participants clearing a higher obstacle that resulted in a similar foot trajectory to the ones where participants with cerebellar ataxia would overshoot clearance. It was shown that when control participants stepped over this higher obstacle, moment patterns were similar to when cerebellar ataxic participants overshot clearance. These results indicated that when hypermetria occurred, it was due to a voluntary strategy to increase the margin of safety when stepping over obstacles (Morton et al., 2004).

1.3.1.4: Ataxic gait: due to a balance deficit or a coordination deficit?

Symptoms associated with ataxic gait have been thought to occur for two different reasons: due to balance deficits and due to coordination deficits. Morton & Bastian (2003, also reviewed in Morton & Bastian, 2004, 2007) have argued that balance deficits contribute to ataxic gait more so than coordination deficits. Morton & Bastian (2003) examined 3 tasks (lateral weight-shifting task, visually guided stepping task, and walking) in participants with cerebellar damage. The first two of these tasks were used to classify participants with cerebellar ataxia into one of four groups: no deficits, a balance deficit (from the weight-shifting task), a leg-placement deficit (from stepping task), or both deficits. These groupings were used to determine how locomotion differed between the groups. During the walking task, it was observed that participants with only a leg-placement deficit were significantly different than controls in 1 measure (knee-hip joint decomposition index) while participants with only a balance deficit were different in 5 measures (stride length, stride length variability, peak hip flexion, peak knee flexion, and peak ankle plantar flexion). Since the balance deficit group had a greater number of differences than the leg-placement deficit group, it was concluded that the symptoms associated with cerebellar gait ataxia are closely related to balance deficits.

This idea has been disputed by Ilg et al. (2007) who have suggested that symptoms related to cerebellar ataxia are related to joint coordination deficits. Ilg et al. (2007) observed locomotion in healthy, cerebellar ataxia, Parkinson's disease, and vestibular deficient participants. In this study, an analysis was introduced that allowed separation of spatial and temporal variability in joint movement. The main outcome of this study was that participants with cerebellar ataxia had changes in temporal variability of multi-joint coordination patterns while participants with Parkinson's disease and vestibular deficits did not show these changes in temporal variability. Changes in temporal variability in participants with cerebellar ataxia were also correlated to voluntary limb control (as measured by ICARS kinetic subscale). Since these temporal changes were observed only in participants with cerebellar ataxia and not participants with other balance deficits (Parkinson's disease, vestibular deficit), it was concluded that these changes must occur due to limb coordination deficits rather than balance deficits.

From these two studies alone, it is difficult to determine if symptoms related to cerebellar ataxia are related to balance or coordination deficits. The study by Morton & Bastian (2003) used two tasks that were assumed to be directly related to balance or coordination deficits, but this may not be the case. In the balance task used (and many other balance tasks), all body segment movements must be coordinated to ensure stability, which can tie the two deficits together and a single one cannot be isolated. In addition, the leg placement task used is highly driven by visual input; a sensory system that is affected by cerebellar ataxia. This conclusion leads one to believe that further analysis is needed to truly determine the contributions of balance and coordination deficits to ataxic gait.

1.3.2: Stroke

A previous stroke can be very detrimental to one's health due to the increased risk of falling associated with this neurological disorder. Forster & Young (1995) conducted a study in the United Kingdom which reported fall incidence in 108 stroke patients. They observed that 46% of these patients fell at least once while being treated in hospital and 73% fell in the first 6 months following discharge, with a total of 270 falls being reported. Participants who fell 2 or more times were less socially active at 6 months following stroke and were more depressed compared to the other participants (Forster & Young, 1995). This high incidence of falling shows the importance of understanding specific locomotor changes that occur following stroke.

Olney & Richards (1996) stated that the pathophysiological basis of stroke is damage to motor cells and pathways in the CNS caused by haemorrhage (bleeding of a blood vessel) or occlusion (closure of a blood vessel) affecting the arterial blood supply of the brain. This usually affects one side of the brain and typically occurs with little or no warning. Most ischemic strokes (caused by occlusion) are associated with atherosclerosis and thrombosis, while hemorrhagic strokes are usually associated with hypertension or aneurysms (Kandel et al., 2000). Other causes of stroke include cardiac disease, trauma, infection, neoplasm, blood dyscrasia, vascular malformation, immunological disorders, and exogenous toxins (Kandel et al., 2000). Cruz, Lewek, & Dhaher (2009) have suggested that lower limb movement and locomotion disruptions following a stroke are related to descending neural pathway damage resulting in muscle weakness, exaggerated reflex activity, and impaired coordination. As well, the sensory tracts may be interrupted by a previous stroke (Perry, 1969). The following sections will highlight how a previous stroke affects locomotion and obstacle clearance.

1.3.2.1: Locomotion in people with a previous Stroke:

A major effect of stroke on a person can be decreased locomotor capacity. Friedman (1990) examined locomotor capacity in 197 older adults that were admitted to hospital due to a stroke and observed that 57% of these participants could not walk independently 7 days post-stroke. Of these participants, only 42% were able to gain gait independence between 1-4 months following stroke. Since a previous stroke can have such a debilitating effect on locomotion, rehabilitation is important in this population to regain mobility. The following describes locomotor changes that occur following a stroke.

Generally, participants with a previous stroke have a lower gait velocity when compared to healthy individuals (Brandstater, de Bruin, Gowland, & Clark, 1983; Knutsson & Richards, 1979; Lehmann, Condon, Price, & deLateur, 1987). Brandstater et al. (1983) examined gait speeds in people with differing severity of stroke and noted a velocity range from 0.16 – 0.65 m/s (compared to 1.14 m/s observed in a healthy population). This lower gait velocity is accompanied with a decreased stride length, decreased cadence, increased stride time, increased stance time, and increased double support time when compared to healthy individuals (Brandstater et al., 1983; Burdett, Borello-France, Blatchly, & Potter, 1988; Chen et al., 2003). Other results have shown that durations of double support, push-off, and mid-stance are not significantly different to healthy individuals who walk at a speed matched to people with a previous stroke, suggesting that some of these gait characteristics may be due solely to a decreased walking speed rather than a stroke (Lehmann et al., 1987). The severity of a stroke can have an affect on some temporal locomotion variables. For example, Chen et al. (2003) showed a decreased velocity, cadence, single support, and increased stride time and double support time in lower functioning stroke patients when compared to higher functioning stroke patients. As well, at later stages of motor recovery, some of these measures become closer to a healthy population (Brandstater et al., 1983). One of the hallmarks of gait in people with a previous stroke is asymmetry. Studies have shown when comparing paretic and non-paretic limbs; the paretic limb has a greater stride time and swing time, and a decreased single support time when compared to the non-paretic limb (Brandstater et al., 1983; Chen et al., 2003).

During locomotion, a previous stroke also affects joint angles at the hip, knee and ankle. At the paretic hip, flexion is decreased at heel contact while hip extension is decreased at toe-off when compared to healthy adults (Burdett et al., 1988; Chen et al., 2003). When walking speed is controlled for in a healthy adult population, hip extension remains decreased at toe-off (Lehmann et al., 1987). Hip flexion is also decreased in the paretic limb during the swing phase when compared to healthy adults (Burdett et al., 1988; Chen et al., 2003). As well, hip angles are lower in the paretic limb during stance and swing when compared to the non-paretic limb (Chen et al., 2003). Knutsson & Richards (1979) noted a hip hiking strategy in the paretic limb to assist in foot clearance during swing. Two main kinetic differences are observed at the hip on the paretic side in people with a previous stroke. During stance, the hip extension moment is increased to provide support on the affected side (Olney, Griffin, & McBride, 1998; Olney & Richards, 1996). As well, the H3 power burst is increased on the paretic (Olney & Richards, 1996) and non-paretic sides (Cruz et al., 2009; Olney et al., 1998) in faster walking stroke patients which is suggested to occur to compensate for decreased plantarflexor generation in order to propel the body forwards (Jonkers, Delp, & Patten, 2009; Olney et al., 1998).

For the knee, flexion is increased at heel contact (Burdett et al., 1988) and the knee becomes more extended at toe-off on the paretic side when compared to healthy adults (Burdett et al., 1988; Knutsson & Richards, 1979). In participants with greater stroke severity, the knee becomes hyperextended at this time (Knutsson & Richards, 1979). During swing phase, knee flexion is decreased when compared to healthy adults (Chen et al., 2003) and when healthy adults walk at a speed matched to a stroke population (Lehmann et al., 1987). Knee angles during stance and swing are also lower in the paretic limb when compared to the non-paretic limb (Chen et al., 2003). Examination of knee kinetics showed that the K3 burst increased on the non-paretic side (Olney, Griffin, Monga, & McBride, 1991) and an increased knee extensor support moment in stance in the affected limb (Olney et al., 1998; Olney & Richards, 1996) which has been suggested to occur to compensate for decreased support observed in ankle moments (Olney et al., 1998).

During stance phase, ankle plantar flexion angle at heel contact is increased (Burdett et al., 1988). Ankle plantarflexion is decreased at toe-off when compared to healthy adults (Burdett et al., 1988; Chen et al., 2003) and when healthy adults walked at a speed matched to stroke participants (Lehmann et al., 1987). In swing phase, ankle dorsiflexion is decreased when compared to a healthy population (Chen et al., 2003). As well, ankle angles are decreased during stance and swing in the paretic limb when compared to the non-paretic limb (Chen et al., 2003). Kinetic analysis of the ankle shows that the A2 power burst is decreased on the affected side (Olney et al., 1998; Olney et al., 1991), which may be related to decreased gastrocnemius EMG activity (Knutsson & Richards, 1979) and a lower stretch reflex threshold in the plantarflexors (Knutsson, 1981).

An examination of whole body centre of mass energy and work shows the inefficiency in stroke locomotion. In fact, the total energy and work done by stroke participants during locomotion have been observed to be twice as large as healthy adults (Iida & Yamamuro, 1987). Olney, Monga, & Costigan (1986) observed 3 patterns that accounted for these inefficiencies: gross irregularities in kinetic and potential energy curves with almost no opportunity for energy exchange, some mirroring of kinetic and potential energy patterns with only limited energy exchange occurring, and exaggerated mirroring of head and trunk kinetic/potential energy with potential energy exchange being dominated by swing of the affected limb (which shows a hip hiking pattern).

Angle-angle plots have been used to observe lower limb coordination in participants with a previous stroke. Compared to a healthy population, hip-knee plots tend to show smaller patterns (decreased loop area) as well as horizontally and vertically directed segments in the paretic limb when compared to the non-paretic limb suggesting periods were joint movement did not occur together (Giannini & Perell, 2005). Daly, Sng, Roenigk, Fredrickson, & Dohring (2007) have also shown jagged edges (suggesting multiple reversals between joints) and no consistency between strides for hip-knee plots. Knee-ankle plots in a population with a previous stroke display long periods where knee motion occurs without movement of the ankle in the paretic limb (Giannini & Perell, 2005). As well, Winstein & Garfinkel (1989) noted a large amount of decoupling between knee and ankle joint movements and large distortions of the loop in a population with a previous stroke when compared to a healthy population. These methods have also been used to observe lower limb joint coordination changes in people with a previous stroke during rehabilitation programs and have shown that these angle-angle plots tend to resemble a healthy population with less jagged edges and a larger range of motion (Chin, 1982; Daly et al., 2007; Steiner et al., 1982).

Other methods of analyzing lower limb coordination in a population with a previous stroke have also been used. Winstein & Garfinkel (1989) used phase plane plots to determine coordination differences in the knee and ankle joints. The authors showed that the overall shape of the phase planes were distorted which was attributed to improper control of muscle timing and muscle weakness. Barela et al. (2000) used an analysis observing the relative phase between the thigh and shank segments although this method did not appear to be as useful in determining coordination in a stroke population since the authors noted that relative phase was similar between paretic and non-paretic limbs during most of the stance phase and that mostly timing differences in relative phase occurred when comparing a stroke population to a healthy population.

1.3.2.2: Obstacle Clearance in people with a previous Stroke:

People who have sustained a previous stroke are less likely to successfully clear obstacles during locomotion. This has been shown when walking over ground (Said, Goldie, Patla, Sparrow, & Martin, 1999) and when obstacles are suddenly dropped onto a treadmill (Den Otter, Geurts, de Haart, Mulder, & Duysens, 2005). Both of these studies showed that success rates did not depend on whether the participant was leading with the paretic or non-paretic limb (Den Otter et al., 2005; Said et al., 1999). With a rehabilitation program that includes obstacle clearance training, the capacity for this population to clear obstacles increases (Jaffe, Brown, Pierson-Carey, Buckley, & Lew, 2004). This section will outline how a previous stroke affects kinematic and centre of pressure patterns in the leading and trailing limbs when stepping over an obstacle.

Changes that occur in the leading limb during obstacle clearance in people with a previous stroke depend on whether the paretic or non-paretic limb is leading. When the paretic limb is leading, it is placed closer to the obstacle prior to and following clearance when compared to healthy adults walking at their comfortable velocity, but not when their walking velocity was matched to that of the stroke participants (Said et al., 2005). Said et al. (2008) showed that medial/lateral centre of mass (COM) velocity was increased and the

COM was positioned closer to the non-paretic stance limb at the moment of clearance in the stroke population when compared to healthy adults walking at comfortable velocity. As well, COM velocity in the forward direction was reduced, and the centre of pressure (COP) was located closer to the non-paretic stance heel at clearance (Said et al., 2008). When the non-paretic limb is leading, it is placed closer to the obstacle following clearance and the foot lands flatter on the floor with a decreased horizontal velocity when compared to healthy adults walking at their comfortable velocity, but not when their walking velocity was matched to that of the stroke participants (Said et al., 2005). The COM-COP distance in the supporting paretic limb was greater when compared to healthy adults walking at a slower velocity (which would lead to greater instability) (Said et al., 2008).

Said et al. (2005) also determined that joint angle trajectories and toe clearance distances were similar between participants with a previous stroke and healthy individuals. Since some of these changes were not significantly different to healthy adults walking at a velocity matched to participants with a previous stroke, Said et al. (2005) suggested that these changes occurred partly due to gait speed and not necessarily because of a stroke. Significant differences from healthy individuals walking at a slower velocity were found for COM velocity in the forward direction and COP location. Said et al. (2008) attributed these differences to the inability of the stroke population to generate enough energy at the plantarflexors to accelerate the COM and to using a cautions strategy when stepping over the obstacle. Said, Goldie, Patla, & Sparrow (2001) also observed an increased step time in the leading limb which was attributed to a safety strategy to increase time for swing limb modification, but this study grouped paretic and non-paretic leading limb clearances together and the differences between limbs can not be determined.

In the trailing limb, changes in movement patterns were again dependent on whether the paretic or non-paretic limb was trailing. When the paretic limb was trailing, toe clearance decreased when compared to a healthy population walking at a comfortable velocity and the foot landed flatter and closer to the obstacle following clearance when compared to a healthy population walking at a slower velocity (Said et al., 2005). When the non-paretic foot was trailing, stroke participants landed with a flatter foot and a decreased horizontal velocity when compared to participants walking at comfortable velocity and decreased foot obstacle distance following clearance when compared to healthy individuals walking at a slower velocity (Said et al., 2005). Since no differences in horizontal foot velocity at contact were observed between stroke participants and healthy adults walking at a slower velocity, these results were partially attributed to the slower gait velocity in the stroke population (Said et al., 2005). The distance between the foot and the obstacle was lower when compared to healthy adults walking at a slower velocity, which may actually be detrimental to obstacle clearance and increase the risk of failure (Said et al., 2001). Again, no significant differences were observed in lower limb joint angle trajectories during clearance.

From these results, it is interesting to note that no differences between paretic and non-paretic lower limb joint angle trajectories were observed during obstacle clearance (except for changes in ankle angle at heel contact) although the previous section notes many changes that occur for level walking. One of the reasons for this may be the size of the obstacle used in the studies noted above. Said et al. (Said et al., 2005; Said et al., 2008) used a 0.04 m high obstacle for each of these studies. One of the reasons we may not see many differences in lower limb joint angle trajectories is because an obstacle of this height is not adequate in challenging the CNS for an obstacle clearance task. Changes in lower limb joint angle trajectories may be evident with a larger obstacle that increases the challenge to the CNS.

These results show that some adaptations when a stroke population steps over an obstacle are beneficial to safety (such as stance limb COP location being closer to the heel during lead limb clearance) while others seem detrimental to safety (such as a decreased distance between the foot and the obstacle following clearance). It is most likely due to these detrimental changes that an increased rate of obstacle clearance failure occurs in a stroke population. Further study is needed to determine if a decrease in coordination of the lower limb joints is a factor in obstacle clearance failure.

The previous section has discussed changes in inter-joint coordination that result from specific pathological conditions. The coordination analyses discussed in these sections provides a measure of the coordination of two joints or segments at one time. This information is important, yet human locomotion requires coordination of three joints or segments at one time. The planar law of intersegmental coordination is a unique coordination analysis technique since it takes into account all three lower limb segments. As well, this analysis technique has been suggested to be related to specific CNS control mechanisms and is not purely descriptive in nature much like the previously used coordination analysis techniques. The following section will discuss the planar law of intersegmental coordination and its use in human locomotion.

1.4: A paradigm for examination of inter-joint coordination: The Planar Law of Intersegmental Coordination

Previous research has shown that absolute angles of the lower limb segments (angles of segments with respect to the vertical axis) tend to follow specific planar patterns. This was initially shown by Lacquaniti & Maioli (1994a) when cats stood on tilted support surfaces. In this study, the forelimb and hindlimb joint angles of these cats co-varied on the different support surfaces such that if the scapula, shoulder, and elbow (hip, knee, and ankle in the hindlimb) were plotted in 3-dimensional space, the points representing the absolute angles of the proximal segments and the relative joint angles of the distal segments tended to lie on a plane in this space. When trajectories of these angles were plotted in such a 3-dimensional space during platform tilt perturbations to cats, similar planar patterns were observed (Lacquaniti & Maioli, 1994b). These results led Lacquaniti & Maioli (1994a, 1994b) to conclude that this planar pattern represents a neural constraint on the covariation of joint angles which indicates a simplification of CNS control.

Borghese, Bianchi, & Lacquaniti (1996) applied these techniques to human locomotion and a similar planar pattern was observed. In humans, when segment elevation angles (absolute angles of each segment with respect to the vertical) for one lower limb (thigh, shank, and foot) are plotted in 3-dimensional space for a gait cycle, they tend to form a teardrop shaped loop which orients on a plane in space (Figure 1.1). This has been observed in level walking (Borghese et al., 1996), walking at differing velocities (Bianchi, Angelini, &

Lacquaniti, 1998; Bianchi, Angelini, Orani, & Lacquaniti, 1998), on an incline (Noble & Prentice, 2008), up stairs (Ivanenko, d'Avella, Poppele, & Lacquaniti, 2008), when a single limb steps over a 0.3 m high obstacle (Ivanenko, Cappellini, Dominici, Poppele, & Lacquaniti, 2005), backwards walking (Grasso, Bianchi, & Lacquaniti, 1998), with a bent posture (Grasso, Zago, & Lacquaniti, 2000), with body weight support (Grasso et al., 2004), in a curved trajectory (Courtine & Schieppati, 2004), and on a slippery surface (Cappellini, Ivanenko, Dominici, Poppele, & Lacquaniti, 2010). As well, this pattern has been shown to persist during locomotion in pathological populations such as Parkinson's Disease (Grasso et al., 1999), spastic paraparesis (Dan, Bouillot, Bengoetxea, & Cheron, 2000), spinal cord injury (Grasso et al., 2004), and forefoot rheumatoid arthritis (Laroche et al., 2007). Studies have also documented the development of this plane in infants during level walking (Cheron et al., 2001; Dominici, Ivanenko, & Lacquaniti, 2007; Ivanenko et al., 2008) and during anticipatory locomotor adjustments (Dominici, Ivanenko, Cappellini, Zampagni, & Lacquaniti, 2010). In humans, this planar pattern has been suggested to occur due to a simplification of motor control in the CNS by decreasing the available degrees of freedom in the lower limb (Borghese et al., 1996).

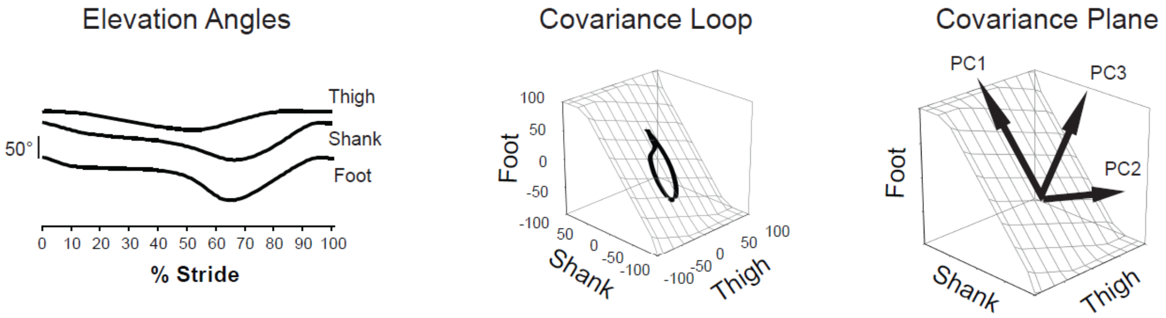

Figure 1.1: Plots illustrating the planar law of intersegmental coordination. When elevation angles (left) of the thigh, shank, and foot are plotted together in 3D space, they tend to form a teardrop shaped loop which orients on a plane in this space (centre). The characteristics of this plane are defined using a principal component analysis such that principal components (PC) 1 and 2 define the plane and PC 3 defines the orientation of this plane (right).

Although this planar pattern persists in these various forms of locomotion, specific characteristics of this pattern tend to differ which include planarity, covariance plane orientation, and covariance loop width. Planarity can be defined as a measure of how close the segment elevation angles tend to orient on a plane, which is usually presented as a percentage. For example, Borghese et al. (1996) showed that planarity during level walking was greater than 99%. Studies have shown that planarity decreases very slightly (although measures of planarity continued to remain over 99%) when walking on an inclined ramp (Noble & Prentice, 2008) and when walking in a curved trajectory (Courtine & Schieppati, 2004). Noble & Prentice (2008) attributed this slight decrease in planarity to online adaptive adjustments in elevation angle waveforms that were required for the completion of the task.

In each of the above mentioned modes of locomotion (as well as for locomotion in pathological populations), the orientation of the covariance plane tends to change from that of level walking. Bianchi, Angelini, Orani et al. (1998) observed that as healthy adults increased gait speed, the orientation of the covariance plane rotated about the longitudinal axis of the covariance loop. It was determined that the angle between the covariance plane and the thigh axis in the 3-dimensional space was the most sensitive to changes in plane orientation and this angle is used to numerically represent the covariance plane orientation (either in degrees or as a direction cosine) (Bianchi, Angelini, Orani et al., 1998). To provide an example of this change in covariance plane orientation, Ivanenko et al. (2005) applied this analysis technique to healthy adults when stepping over a 0.3 m obstacle with a single limb and observed that the orientation of the covariance plane rotated counterclockwise about the longitudinal axis 14 degrees when compared to level walking. The purpose of covariance plane rotation has previously been attributed to net mechanical energy output during the locomotor task (Bianchi, Angelini, Orani et al., 1998; Lacquaniti, Grasso, & Zago, 1999). In particular, Lacquaniti et al. (1999) stated that changes in covariance plane orientation may represent underlying CNS control to economize limb energy expenditure during locomotion.

A planar law of intersegmental coordination characteristic which has not received as much attention as planarity and covariance plane orientation is the width of the covariance loop. If one was to look at the planar covariance illustrations for the various modes of locomotion listed previously, it can be seen that the width of the covariance loop differs greatly. Courtine & Schieppati (2004) showed changes in covariance loop width when young adults walked in a curved trajectory. Results showed an increase in the outer and a decrease for the inner lower limb covariance loop widths when compared to level walking, although an explanation for this change was not provided (Courtine & Schieppati, 2004).

These planar law of intersegmental coordination characteristics are also related to lower limb segment movement. Bianchi, Angelini, Orani et al. (1998) first described this relationship when observing healthy young adults walking at increasing speeds. In this study, a Fourier harmonic analysis was used on each segment elevation angle trajectory to determine its fundamental harmonic. Results showed that the fundamental harmonic phase difference between the shank and foot segments was highly correlated ($r = 0.92$) to the orientation of the covariance plane. This relationship was attributed to a need to bring the foot faster through swing and prepare for heel contact as walking speed increases (Bianchi, Angelini, Orani et al., 1998). Courtine & Schieppati (2004) observed a similar relationship ($r = 0.85$) between the fundamental harmonic phase difference between the thigh-shank segments and the width of the covariance loop. These relationships are thought to be related to the underlying CNS control during locomotion.

Since characteristics from the planar law of intersegmental coordination correlate highly with fundamental harmonic phase differences between adjacent segments in the lower limb, it has been argued that these phase differences may be one of control variables in the CNS for locomotion. Studies by Shen & Poppele (1995) examining feline locomotion showed that elevation angles are highly related to forward and backward motion of the lower limb axis. Using mathematical modelling, Shen & Poppele (1995) also showed that altering phase differences between segments can change overall limb length and they concluded segment phase and amplitude are control variables in the CNS. Similarly, Das & McCollum (1988) provide an illustrative example for human locomotion of how knee angle can be altered in swing by changing the phase difference between thigh and shank. In relation to the CPG, it is proposed that neural oscillators in the CNS control segment elevation angle waveforms with segment phase differences and waveform amplitude being two of the control variables (Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999). How does this relate back to the planar law of intersegmental coordination? Barliya, Omlor, Giese, &

Flash (2009) developed a mathematical model which showed the direct relationship between shank-foot phase difference and covariance plane orientation. As well, this study concluded that eccentricity of the planar covariance loop is mainly determined by amplitudes of segment elevation angle fundamental harmonics and these amplitudes may “stabilize” the covariance plane so rotations about other axes do not occur.

The above findings are very important because together, they provide a theory on how locomotor control is simplified by the CNS. During locomotion, lower limb segment elevation angles tend to show little variability when compared to lower limb joint angles and it was suggested that the CNS takes into account dynamic interactions when issuing motor commands to preserve the kinematic invariance of these angles (Borghese et al., 1996). Segment elevation angles could theoretically be sensed by the CNS by integrating gravitational information from the vestibular system and limb segment position from the proprioceptors (Borghese et al., 1996). Barliya et al. (2009) noted that this strategy would be useful when walking up inclines since the change in gravitational direction can be detected and used as a spatial reference.

Since this method of analyzing coordination seems to apply to many forms of locomotion and takes into account all 3 lower limb segments (instead of 2 in other coordination analyses), it appears to be appropriate for analyzing locomotor coordination. As well, this method has been related to CNS control, which may provide insight as to the underlying mechanisms of locomotor control.

1.4.1: Criticism to Planar Law of Intersegmental Coordination Theory

There are disagreements as to the origin of the Planar Law of Intersegmental Coordination (Hicheur, Terekhov, & Berthoz, 2006). Hicheur et al. (2006) argued that lower limb segment elevation angles orient on a plane due to biomechanical constraints and not neural mechanisms as discussed in the previous section. In this study, locomotion was examined during level walking, fast walking, backward walking, and moderate running. Results from this study showed that there was a steady increase in elevation angle amplitude between the foot and shank segments as locomotor speed increased, but this increase was not observed in the thigh segment. There was also a high correlation between the foot and shank elevation angles at all speeds but the thigh segment did not correlate well with the foot or shank segments. Using linear regression, it was also shown that the angle between the regression line for the shank and foot segments and the plane including all segments was nearly constant at 1.2 degrees in all locomotor conditions. It was argued that these results suggested that thigh segment movement was independent of planar covariation and planar patterns are due to strong mechanical coupling between the foot and shank. This was further argued when a sinusoid wave was replaced for the thigh segment elevation angles planarity still occurred. It should be noted that a cosine wave was used, which has a similar pattern to the thigh segment elevation angle during locomotion. Hicheur et al. (2006) concluded that the Planar Law of Intersegmental Coordination exists solely due to biomechanical coupling between the foot and shank segments and there is no central involvement.

This was greatly disputed by Ivanenko et al. (2008) who believe there is CNS involvement to this planar pattern. Ivanenko et al. (2008) revisited past experimental data and collected new data to argue a neural contribution to the Planar Law of Intersegmental Coordination. In this study, data was presented for treadmill walking at various speeds, overground crouched walking, hopping, uphill walking on a treadmill, stepping up stairs, air-stepping with body-weight support, and unsupported walking in toddlers. Planar covariation was observed in all of these models of locomotion. It was noted that planar covariation was observed in hopping, although there is a low correlation between foot and shank elevation angles during this task ($r = 0.09$). This finding rejected the hypothesis of Hicheur et al. (2006) that planar covariation only occurs due to biomechanical coupling of the foot and shank segments, although it can not be excluded that biomechanical coupling may have some role in planar covariation. During unsupported walking in toddlers, elevation angles increasingly confine to a plane throughout development and it was suggested that independent walking experience is essential for emergence of a mature covariation pattern, therefore arguing for a central component to planar covariation. It was concluded from this study that the planar covariation “strategy” emerges from both biomechanical and central constraints.

From the above review of literature, it can be seen that inter-joint coordination in the lower limb during locomotion is a complex task that is highly regulated by the CNS. Locomotor patterns are usually adapted by the CNS to the environment to reach a specific goal. For example, when stepping over an obstacle, the knee may be involved in elevating the limb while the hip progresses the limb through swing. Locomotor coordination is altered in certain pathological conditions, but this is not well understood. A previous stroke and cerebellar ataxia are examples of such pathological conditions each of which having a differing effect on locomotor control (cerebellar ataxia is symmetrical while a stroke leads to asymmetries in control). Coordination has been analyzed using various techniques such as angle-angle plots and relative phase analyses. Yet, such coordination analyses have been mostly descriptive in nature. Using analysis techniques related to the planar law of intersegmental coordination may provide some insight into the underlying CNS control mechanisms in both healthy and impaired individuals.

1.5: Objectives and Hypotheses:

ALAs are usually needed to arrive at a locomotor goal. During such adjustments, the inter-joint coordination is altered but we do not have a great understanding as to how this is accomplished. Current studies in coordination analysis are mostly descriptive in nature and do not provide any insight into CNS control of this coordination. This thesis will further examine coordination of the lower limb during level gait and obstacle clearance.

As previously stated, lower limb joint coordination during locomotion tends to follow patterns as stated by the planar law of intersegmental coordination, although this law has not been used extensively for the analysis of obstacle clearance. Since this law has been related to locomotor control, such analyses may provide information on how the CNS controls movement during ALAs. As well, joint specificity may be a mechanism with which the CNS coordinates lower limb movement. Although this idea has been theorized, it has not been tested exclusively. This thesis will use an obstacle clearance paradigm in which obstacle height and depth will be adjusted to force changes in limb elevation and progression in order to determine if joint specificity for the hip and knee joints exists during obstacle clearance. To gain insight into the mechanisms of coordination deficit, analyses will be conducted in a healthy young adult population and extended to two populations with CNS control deficits (hemiplegia due to stroke and coordination disorders due to ARCA-1). People with a previous stroke have asymmetric locomotion and this analysis may provide insight into how the CNS control differs between each limb. As well, the population with cerebellar ataxia to be used in this study is unique and documented to have pure diffuse and homogeneous cerebellar atrophy (ARCA-1). The studies presented in this thesis will therefore add to the current literature by discussing specific CNS control mechanisms during obstacle clearance in healthy adults, as well as in 2 pathological populations (stroke and ARCA-1).

1.5.1: Specific Objectives

Study 1:

To better understand how ALAs are controlled by applying the planar law of intersegmental coordination to locomotor patterns of both leading and trailing limbs in a paradigm in which obstacle height and depth are manipulated.

Study 2:

To determine how individual phase shifts in fundamental harmonics of the thigh and shank segments contribute to the increase in phase difference between these segments as higher obstacles are cleared, and then use such information to determine if joint specificity is a mechanism of control in the CNS by relating the amplitudes and timings of these segment elevation angles to work done by the hip and knee joints.

Study 3:

To better understand how a deficit such as ARCA-1 affects leading limb segment coordination during obstacle clearance using the planar law of intersegmental coordination as well as suggest specific locomotor control mechanisms for this population.

Study 4:

To understand differences in segmental control mechanisms between the paretic and non-paretic limbs in participants with a previous stroke when leading during obstacle clearance by using the planar law of intersegmental coordination.

1.5.2: Specific Hypotheses

Study 1:

It is hypothesized that ALAs are controlled by similar lower limb segment elevation angle phase mechanisms as shown in level walking. It is further hypothesized that planarity would be maintained across obstacle clearance conditions, supporting the view of an emergence of an adapted pattern from basic level walking control, but with systematic changes due to step elevation and length requirements that are limb dependent.

Study 2:

It is hypothesized that the phase difference between the thigh and shank segments will highlight different control strategies between each limb when clearing an obstacle. It is further hypothesized that the work done by the hip muscles will be strongly related to progression of the lower limb and the work done by the knee muscles will be strongly related to elevation of the lower limb, suggesting a joint specific mechanism of control by the CNS.

Study 3:

It is hypothesized that increases in toe clearance will be observed in the ARCA-1 participants as previously observed by Morton et al. (2004). It is further hypothesized that these increases in clearance will be related to phasing between adjacent lower limb segments highlighting a preservation of an elevation waveform phasing control mechanism in this population to voluntarily increase the margin of safety during obstacle clearance.

Study 4:

It is expected that planarity will remain high in the paretic and non-paretic limbs during level walking and obstacle clearance, suggesting a similar simplification of CNS control in each limb as seen in healthy adults during level walking. It is further hypothesized that differences in characteristics of the planar pattern of segment elevation angles (specifically covariance loop width and covariance plane orientation) between paretic and non-paretic limbs will occur which will highlight differing coordination strategies for each limb.

Chapter 2: General Methodology

The following sections will discuss supplementary methodology which will not be covered directly in the studies presented in this thesis. This section will begin with a description of the participant populations used, including inclusion and exclusion criteria. Following this, a description of the equipment used and the preparation of this equipment will be provided. Finally, details will be provided regarding the treatment of the raw data and the statistical analyses used.

2.1: Participants and Declaration of Ethics

For each of the studies described in this thesis, different methods were used for recruitment and differing criteria were used for participant inclusion and exclusion. Each study contained in this thesis was approved by the ethics committee at l'Institut de réadaptation en déficience physique de Québec (IRDPQ) and Université Laval. The following sections will describe recruitment methods for each study.

Studies 1 (Chapter 3: Segmental control for adaptive locomotor adjustments during obstacle clearance in healthy young adults) and 2 (Chapter 4: Relationships between segment elevation angles and muscle power during obstacle clearance reveal multiple goals for the mechanical work at hip and knee joints):

These two studies involved the same participant group for analysis. In these studies, ten healthy young adults between the ages of 18 and 55 were recruited from the community using posters placed around the IRDPQ. Participants were included if they had normal or corrected to normal vision and the ability to walk independently. Participants were excluded if they had any previous musculoskeletal or neurological disorders which were verified orally with the participant.

Study 3 (Chapter 5: Increased obstacle clearance distance in people with ARCA-1 results in part from coordination changes between the thigh and shank segments):

Two participants groups (cerebellar ataxia and control) were recruited for this study. Participants with cerebellar ataxia were recruited with the assistance of a local neurologist

(Dr. Nicolas Dupré, Centre hospitalier affilié universitaire de Québec). A total of eight participants were recruited who were above the age of 18, were diagnosed as having recessive ataxia of Beauce, had normal or corrected to normal vision, and the ability to walk independently. Participants were excluded if they had any previous musculoskeletal or neurological disorders aside from cerebellar ataxia. The control participants for this study consisted of eight adult volunteers recruited from the community using posters placed around the IRDPQ and a recruitment announcement sent to the Université Laval community by electronic mail. These participants followed the same inclusion and exclusion criteria as for studies 1 and 2, as well as being matched to cerebellar ataxia participants for the group average of sex and age.

Study 4 (Chapter 6: Comparison of locomotor control mechanisms for segmental coordination between non-paretic and paretic limbs during obstacle clearance following stroke):

In this study, six participants with a previous stroke were recruited for the study with the assistance of the Stroke unit at the IRDPQ. Participants were included in the study if they had a gait speed less than 1.0 m/s and were able to walk independently (without an aid). These participants were included if they had suffered their first stroke and the stroke was located above the midbrain (cerebellar and brainstem strokes were excluded). All participants were above 18 years of age and had normal or corrected to normal vision. Participants were excluded if the had any previous musculoskeletal or neurological disorders aside from a previous stroke.

2.2: Equipment

2.2.1: Custom Made Obstacle

The obstacle that participants stepped over during the study was a custom made apparatus which allowed for the formation of obstacles of various heights and depths (Figure 2.1). The apparatus consisted of a stable wooden base outfitted with a commercial spring loaded vinyl window blind. Above this spring loaded blind were 4 vertical rods. Two of these vertical rods were lined up with the spring loaded blind and the remaining 2 rods were behind the blind, each placed upon a track that would allow horizontal movement of the rod (in which the depth of the obstacle could be manipulated). Two long strips of metal with cylindrical sleeves at each end were connected between the 2 front and 2 rear vertical rods. The cylindrical sleeves allowed vertical movement of the long strips of metal and to adjust the height of the obstacle. After the positions of the long metal strips were set, the vinyl material could then be wrapped around the long strip of metal in the front of the obstacle and connected to the rear strip of metal using Velcro, therefore forming a rectangular shaped obstacle.

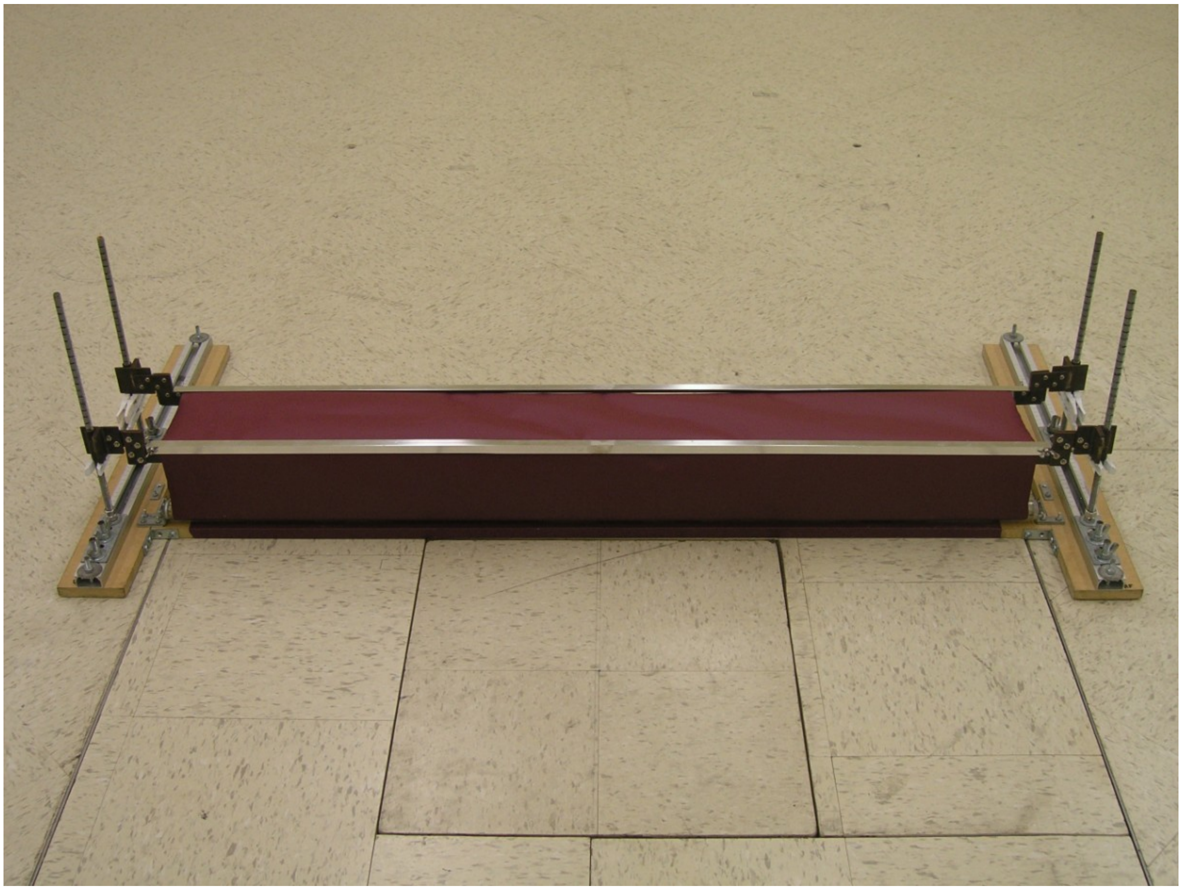

Figure 2.1: Illustration of the custom made obstacle participants stepped over in the studies.

2.2.2: Optotrak 3020 System

For all studies, 3-dimensional (3D) movement of each participant was collected using a 3 position sensor Optotrak 3020 system (Northern Digital Inc., Waterloo, Onatrio, Canada). Each position sensor in the Optotrak 3020 system contained 3 opto-electronic sensors which collected infrared signals emitted by infrared emitting diodes (iREDs) indicating where the marker is located in 3D space. These iREDs were placed on the participant's body segments in order to track 3D segment movement. Each iRED was connected to a strober pack which was then connected by wire to a controller unit. The cameras used in the experimental setup were also connected to this controller unit. One of the main tasks of this controller unit is iRED identification, which is performed by activating each iRED in the setup sequentially at a high frequency so that only 1 iRED is activated at each instance. The Optotrak controller unit was connected to a personal computer, which allowed for the recording of marker locations at each frame of collection. Prior to data collection the Optotrak 3020 system must be calibrated, which is detailed in section 2.3.1.

2.2.3: Advanced Mechanical Technology, Inc. (AMTI) Force Platforms

Three AMTI multi-axis force platforms were used in this study to collect ground reaction forces and moments during locomotion. Each force platform was built into the floor therefore creating a surface which was flush with the floor. Under the floor, these force platforms were bolted to the ground in specific configurations to allow for collection of leading and trailing limb ground reaction forces. Each of these force platforms outputs electronic currents corresponding to forces along 3 orthogonal axes and 1 moment about each of the 3 axes. These forces and moments are measured using electronic strain gauges mounted within the force platform with currents flowing through them. Changes in current flowing through each strain gauge are proportional to the loading on the force platform which allows for the measurement of force and moments of force. These signals are then amplified and sampled at a specified frequency by an analog-to-digital converter and recorded on a personal computer.

2.3: Experimental Procedures

2.3.1: Preparation of Equipment

Prior to data collection, the measurement systems used in these studies need to be prepared. For the Optotrak camera system, there are 2 calibration procedures that must be completed. The first is a dynamic calibration, where a calibration object (an object with iRED markers placed in known locations) is moved throughout the collection volume. The goal of this calibration procedure is to ensure that each Optotrak camera bar identifies each known iRED position correctly. A second calibration, termed an alignment, is performed in order to define the global coordinate axes in the laboratory. To define the global coordinate axes, the calibration object is placed in a single position with specific iREDs pointing in the direction of the desired axes. The desired axes used in these studies were the direction of locomotion (x-axis), the vertical direction with respect to the floor (y-axis), and in the medial-lateral direction of participant movement (z-axis). When these calibrations are complete, the system is ready to collect iRED position data.

AMTI force platforms come pre-calibrated from the manufacturer; therefore there are no direct calibration procedures. Along with the force platform, the manufacturer provides a calibration matrix, in which calculations are made off-line in order to convert the voltage readings from the force platform into forces (in Newtons) and moments (in Newton-meters). Prior to data collection the force platform must adjusted so that there are no voltage differences between the electronic strain gauges, a process which provides a reference voltage when the force platform is unloaded.

2.3.2: Preparation of Participants

When participants arrived to the laboratory, the first task was reading and signing the informed consent ethics forms (Appendix A) to ensure each participant understood their participation in the experiment. Participants were then directed to change into clothes appropriate for data collection (shorts, running shoes, and a t-shirt). When participants were ready, various measurements of segment length and circumference were recorded (Appendix B) which were required to create a 3D biomechanical model (details of the biomechanical model are in section 2.5). In order to track the motion of each body segment, triads of iREDs (which were affixed to plastic plates) were placed on the lateral side of the feet, shanks and thighs, between the posterior superior iliac spines for the pelvis, slightly below the midpoint between the scapulae for the trunk, and on an adjustable strap for the head.

Once the iREDS were placed on the participant and wires from the iREDS were affixed to ensure they did not disrupt movement, a calibration trial was collected to locate each segment and determine the neutral orientation of each segment in 3D space. Following this calibration trial, anatomical landmarks (Appendix C) were digitized using a probe containing 6 iREDs. The end of this probe was placed on each anatomical landmark and the camera system collected data identifying the location of the probe endpoint in 3D space. The collection of these points was used off-line to determine the locations of the anatomical landmarks on the biomechanical model which were subsequently used to determine joint centres of rotation.

2.4: Experimental Protocol

The experimental protocol used in each of the subsequent studies is detailed in the following chapters (Chapters 3-6). In brief, each of these studies examined lower limb joint coordination during obstacle clearance. In studies 1,2, and 4, the obstacle configurations consisted of 3 different obstacle heights (0%, 10%, and 20%, of participant leg length) and 3 obstacle depths (0%, 10%, and 20% of participant step length) making 9 blocks of trials that were presented randomly to the participant. In study 3, the obstacle configurations consisted of 3 different obstacle heights (0%, 5%, and 15%, of leg length) with a consistent depth (0.025 m). Each of these obstacles was presented randomly across trials to the participant.

2.5: Data Analysis

2.5.1: Treatment Raw Data

Steps must be taken following data collection to prepare the raw iRED position data for data analysis. When collecting iRED position data, there may be small section of marker trajectories missing if the Optotrak camera does not receive a signal from the iRED. This may be due to the angle of the iRED (which would not be in view of the cameras) or due to a brief period that the marker is covered. To fill in these data ‘holes’, a cubic spline data interpolation was used which estimated missing marker positions. Once all missing data were accounted for, the data were then filtered to remove any noise in the raw signal. This was done using a 2nd order dual-pass Butterworth filter with a low pass cut-off frequency of 6 Hz. This filtered data was then used to build the biomechanical model described next.

The biomechanical model used in the studies was a 3D link-segment model containing 9 body segments (feet, shanks, thighs, pelvis, trunk, and head). The digitized anatomical landmarks collected in section 2.3.2 were used in this model to define joint axes using custom software which was modelled using equations from KinGait3 software (Milad Ishac, University of Waterloo, Waterloo, Canada). 3D joint angles were calculated using a Z-Y-X Euler rotation sequence and were determined with respect to the proximal segment. Velocities and accelerations of 3D position data were calculated using formulas outlined in Winter (2005).

Raw data collected from the force platforms were filtered using a 2nd order dual-pass Butterworth filter with a low pass cut-off frequency of 50 Hz. These data calculate joint reaction forces and net muscle moments using Newton-Euler inverse dynamics equations (Winter, 2005). Anthropometric data for each body segment was derived using methods suggested by Yeadon & Morlock (1989) and Dempster (1955) from the collected body measurements outlined in section 2.3.2.

All remaining data analysis techniques are explained fully in the subsequent chapters of this thesis.

2.5.2: Dependent Variables

The dependent variables used in the studies presented in this thesis differed depending on the study. The dependent variables that were observed in all studies presented in this thesis include foot placement (distance from the toe to the obstacle prior to clearance and distance from heel to obstacle following clearance), foot trajectory (including clearance over obstacle), and lower limb joint angles. All studies included measures of lower limb segment (thigh, shank, and foot) elevation angles, as well as specific coordination measures including planarity, covariance plane orientation, and covariance loop with (in relation to the planar law of intersegmental coordination), and fundamental harmonic phase differences between the thigh-shank and shank-foot segments which are described in section 2.5.3.1. Study 2 also included measures of lower limb joint kinetics (ankle, knee, and hip net muscle moments and net muscle powers).

2.5.3: Statistics

The statistical analyses used in the presented studies are detailed in each chapter (Chapters 3-6). In brief, studies 1 and 2 use parametric statistical tests (analysis of variance, Tukey post-hoc test, and Pearson correlations) since the populations used in these studies are assumed to have normal variance distributions in the variables studied. Studies 3 and 4 use non-parametric statistical tests (Friedman, Wilcoxon, Mann-Whitney U, and Spearman correlations) since it is assumed that the pathological populations will have inconsistent variability in the variables studied. The statistical tests were performed using SPSS statistical package (16.0 for Windows, SPSS Inc., Chicago, USA). One of the main statistical methods used in studies 2-4 is a principal component analysis (PCA), which is described in the following section.

2.5.3.1: Principal Component Analysis (PCA)

The statistical technique used to quantify variables related to the planar law of intersegmental coordination is a PCA. PCA is a statistical method aimed at mode reduction and detection of invariant and variant properties of coordination (Daffertshofer, Lamoth, Meijer, & Beek, 2004). One principle component is derived for each variable introduced into the analysis. In the case of segment elevation angles (the variables used when observing the planar law of intersegmental coordination), the inputs for the analysis are segment absolute angular changes over time for the foot, shank, and thigh during a single stride cycle (3 variables). Therefore, the analysis will output 3 principle components. In basic terms, when a PCA is applied to segment elevation angles, it is very much like a rotation of the original axis system to a new orientation which describes the variability of the data. The first principal component is aligned along an axis in the direction of maximum variance (Ivanenko et al., 2008). The second principal component will be aligned in a direction of maximum variance perpendicular to the first principal component (Ivanenko et al., 2008). The final (third) principal component will be in a direction of maximum variance orthogonal to the plane created by the first and second principal components (Ivanenko et al., 2008).

The use of PCA for segment elevation angles of the foot, shank, and thigh allow for determination of 3 important measurements: planarity, covariance loop width, and covariance plane orientation. Planarity is quantified by the explained variance of the first and second principal components. In theory, if the data were to orient perfectly on a plane, the explained variance of the first 2 principal components would be 100% (all data orienting on the plane). When PCA is applied to segment elevation angles in level walking in healthy young adults, approximately 99% of the variance is explained by the first 2 components, suggesting high planarity. Covariance loop width is estimated using the percent variance explained by the second principal component; which is always oriented in the direction of the covariance loop width (Courtine & Schieppati, 2004). Covariance plane orientation is quantified using the eigenvector projecting the 3rd principle component onto the thigh axis due to its sensitivity to plane orientation changes (Bianchi, Angelini, Orani et al., 1998). This statistical analysis will be used extensively throughout this thesis.

Chapter 3: Study 1

“Segmental control for adaptive locomotor adjustments during obstacle clearance in healthy young adults”

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3.1: Abstract

Anticipatory locomotor adjustments (ALAs) are used during locomotion to perform tasks, such as obstacle clearance, although not much is known as to how these ALAs are implemented by the central nervous system (CNS). The current study applied the planar law of intersegmental coordination to both leading and trailing limbs in a paradigm in which obstacle height and depth were manipulated to propose how ALAs are controlled. Ten healthy young adults stepped over nine obstacle conditions. Full-body 3D kinematic data were collected and elevation angles of the foot, shank, and thigh in the sagittal plane were calculated. For each limb within each trial, a principal component analysis was applied to limb segment trajectories. As well, a Fourier harmonic series was used to represent segment elevation angle trajectories, and phase differences between adjacent segments were determined. Planarity was consistently high in both limbs for all obstacle conditions, although significant differences between obstacle heights were observed. Increases in covariance loop width and rotation of the covariance plane accompanied changes in planarity. As observed in previous studies, fundamental harmonic phase differences between adjacent segments were highly correlated to plane characteristics and these phase differences changed systematically with increases in obstacle height. From the results, it is proposed that if a given environment requires a change in locomotion, the CNS adjusts a basic locomotor pattern if needed through the manipulation of the phase differences in the fundamental harmonics of the elevation angles between adjacent

segments and elevation angle amplitude (with a constraint being intersegmenal elevation angle planarity).

3.2: Introduction

Human locomotion must be adapted to accomplish daily tasks. For example, walking towards a goal may include steering, stepping to a new level, and the clearance of obstacles. These tasks are preformed using anticipatory locomotor adjustments (ALAs). Using computer modeling, McFadyen et al. (1994) suggested that ALAs used when stepping over an obstacle are a reorganization of a basic locomotor pattern, rather than a separate pattern in itself. Other work purports the same idea providing more detail of visuomotor coupling that might underlie such reorganisation (Taga, 1998). However, control variables are not specifically discussed and not much is still known as to how these locomotor adjustments are coordinated by the central nervous system (CNS).

When stepping over an obstacle, the CNS reorganizes the locomotor pattern to increase hip and knee flexion in the leading and trailing limbs (Chou & Draganich, 1997; McFadyen & Winter, 1991; Patla et al., 1991) which in turn elevates the foot over the obstacle. As the height of an obstacle increases, so does flexion of the hip, knee, and ankle (Austin et al., 1999; Chou & Draganich, 1997; Patla & Rietdyk, 1993). One study has observed obstacle clearance over a low obstruction of increasing depths and this resulted in less flexion at the knee and ankle during clearance (Patla & Rietdyk, 1993). Differences have also been documented between leading and trailing limbs. In particular, the trailing limb has a lower toe clearance (Patla et al., 1996), a delayed hip flexor power burst (Niang & McFadyen, 2004), and ankle dorsiflexion (Chou & Draganich, 1997; McFadyen et al., 1993) when compared with the leading limb to avoid collisions with the obstacle at a closer proximity. Although these studies describe the changes in joint movement when stepping over obstacles, they do not detail the coordination of the movement by the CNS.

A method used previously to quantify lower limb coordination during locomotion is the planar law of intersegmental coordination. According to this law, if lower limb segment (foot, shank, and thigh) elevation angles are plotted for a gait cycle, they create a plane in

3D space. Previous work has shown planarity of elevation angles in many modes of walking including level (Borghese et al., 1996), at differing velocities (Bianchi, Angelini, & Lacquaniti, 1998; Bianchi, Angelini, Orani et al., 1998), on an incline (Noble & Prentice, 2008), up stairs (Ivanenko et al., 2008), backwards (Grasso et al., 1998), with a bent posture (Grasso et al., 2000), with body weight support (Grasso et al., 2004) and on a curved trajectory (Courtine & Schieppati, 2004). It has been concluded that this planar pattern is an indication that the CNS decreases the available degrees of freedom when coordinating locomotion (Bianchi, Angelini, Orani et al., 1998; Borghese et al., 1996; Lacquaniti et al., 1999). Although these forms of locomotion all show planar patterns, characteristics of this plane differ between them. Ivanenko et al. (2005) applied the planar law of intersegmental coordination to a single limb stepping over an obstacle 0.3 m in height during locomotion and a counter-clockwise rotation of this plane about the longitudinal axis and an elongation of the covariance loop width were observed. Other studies have shown high correlations between the width of the covariance loop and the fundamental Fourier harmonic phase difference between thigh and shank elevation angles (Courtine & Schieppati, 2004) as well as between the orientation of the covariance plane and the phase differences of the fundamental harmonics of the shank and foot elevation angles fundamental harmonic phase (Bianchi, Angelini, Orani et al., 1998). These correlations have lead Bianchi, Angelini, Orani et al. (1998) to suggest that neural oscillators in the CNS control segment elevation angles through sinusoidal waveforms with segment phase differences and waveform amplitude being two of the control variables. The idea that segment angular phase and amplitude may be controlled by the CNS is not new. Shen & Poppele (1995) studied cat hindlimb motion during locomotion and observed that timing of segment elevation movement was related to the overall orientation and length of the limb throughout movement. As well, Das & McCollum (1988) provide an illustrative example of how knee angle can be altered by manipulating the phase difference between the thigh and shank segments. In particular, the authors showed how knee flexion patterns during swing can differ by keeping the elevation angle trajectory of the thigh constant and phase shifting the relative shank angle trajectory. These results provide a suggestion on how degrees of freedom are simplified by the CNS.

Obstacle avoidance is an important activity of daily living and a threat to safety if not coordinated properly. However, how it emerges from level walking patterns and how lower limb segments are coordinated differently across different obstacle configurations and between leading and trailing limbs is not known yet. The current study applies the planar law of intersegmental coordination to both leading and trailing limbs in a paradigm in which obstacle height and depth are manipulated to propose how ALAs are controlled. It was expected that planarity would be maintained supporting the view of an emergence of an adapted pattern from basic level walking control, but with systematic changes due to step elevation and length requirements that are limb dependent due to differences in proximity to the obstacle and visual information available.

3.3: Methods

Participants were 10 young adults (6 females/4 males, 27.7 ± 5.7 years, 72.0 ± 16.0 kg in weight, 1.72 ± 0.07 m in height). They provided informed consent prior to participation in the study according to ethics guidelines from the IRDPQ and Laval University.

3.3.1: Protocol

During the study, participants stepped over obstacles that were manipulated for both height and depth. Three obstacle heights [(0, 10 (range 0.082–0.095 m) and 20% (range 0.164–0.19 m) of participant leg length] and three obstacle depths [~4, 10 (range 0.068–0.087 m), and 20% (range 0.136–0.174 m) of participant step length] were used for a total of nine obstacle conditions. These conditions were presented in randomized blocks of 5 trials, totaling in 45 trials. The 0% obstacle height conditions with 10 and 20% depths consisted of participants stepping over a piece of paper of the required depth that was fixed to the floor. The ~4% obstacle depth condition with 10 and 20% heights consisted of a 0.03-m deep obstacle. A condition with no obstacle present was used for the 0% height per ~4% depth condition.

Before each trial, participants were positioned approximately five steps away from the obstacle. The experimenter provided a verbal signal for the participant to start walking and to continue until the experimenter asked the participant to stop (approximately 5 steps following clearance). Participants were not told what side to lead with when stepping over the obstacle. Nine of the participants naturally stepped over the obstacle leading with their right foot, and one participant lead with the left foot. It should be noted that the choice of leading limb did not affect the overall observed patterns. To keep consistency within participants, a trial was repeated if a participant stepped over the obstacle leading with the opposite limb, which occurred minimally throughout the study.

Full-body 3D kinematic data were collected (75 Hz) using a 3-bar Optotrak system (Northern Digital Inc., Waterloo, Canada). Non-collinear triads of IRED markers on plastic plates were fixed to the feet, shanks, thighs, pelvis, trunk, and head segments. Each segment had one triad of markers. A calibration trial was collected and anatomical landmarks (5th metatarsal, medial/lateral malleolus, medial/lateral femoral condyles, left/right iliac crest, and left/right anterior superior iliac spine) were digitized including virtual points for the heels and toes of each foot to determine their trajectories. Kinematic data were filtered using a dual-pass second order Butterworth filter with a cut-off frequency of 6 Hz. Ground reaction forces and moments were collected (1,000 Hz) using AMTI force plates (Advanced Mechanical Technology, Inc., Watertown, MA) under the leading and trailing limbs prior to clearance. Force plate data were filtered using a dual-pass second order Butterworth filter with a cut-off frequency of 50 Hz.

3.3.2: Data analysis

The kinematic data were used to create a nine segment biomechanical model (feet, shanks, thighs, pelvis, trunk, and head). From this model, normalized stride length over the obstacle (horizontal distance between heel contact points, divided by participant leg length) and normalized maximum step height over the obstacle (maximum height of toe, divided by participant leg length) were calculated. As well, horizontal toe–obstacle distance prior to clearance, horizontal heel–obstacle distance following clearance, toe clearance height (vertical distance from virtual toe markers to front of obstacle at crossing), and heel clearance height (vertical distance from virtual heel markers to rear of obstacle crossing) were calculated. The model was also used to calculate sagittal relative joint angles of the ankle, knee, and hip. The maximum flexion angle during the stride over the obstacle was used in statistical analyses. A stride corresponded to heel contact prior to obstacle clearance to heel contact following clearance when the obstacle was present. As well, elevation angles of the foot, shank, and thigh segments were calculated (using the formulas provided in Borghese et al. (1996) for each limb over one stride per trial. The range of each segment elevation angle trajectory was calculated by subtracting the maximum angle from the minimum angle.

Inter-joint coordination in each limb was quantified using the planar law of intersegmental coordination (Bianchi, Angelini, & Lacquaniti, 1998; Bianchi, Angelini, Orani et al., 1998; Borghese et al., 1996; Courtine & Schieppati, 2004; Ivanenko et al., 2005; Ivanenko, Cappellini, Dominici, Poppele, & Lacquaniti, 2007; Ivanenko et al., 2008). The elevation angles for the thigh, shank, and foot were normalized to 100% of stride duration and the mean was subtracted from each segment trajectory. A principal component analysis was then used on the group of three segment elevation angles for each stride to determine planarity, plane orientation, and planar covariance loop width. Planarity was quantified by the percent variance explained by the first two principal components. In theory, the first two principal components would explain 100% of the variance of a plane. The percent variance explained by the second principal component alone was used as an indication of loop width. The orientation of the plane was quantified using the direction cosine between the third principal axis (the axis orthogonal to the covariance plane) of the loop and the positive semi-axis of the thigh segment. This method is sensitive for determining rotations of the covariance plane about the longitudinal axis of the loop (i.e., from the upper point to the bottom depression of the covariance loop). A Fourier series using 10 harmonics was then used to represent the time course of segment elevation trajectories of the foot, shank, and thigh segments. The fundamental harmonic of each segment trajectory was then used to calculate phase difference between adjacent segments (thigh shank and shank foot) by subtracting the distal segment phase from the proximal segment phase (Bianchi, Angelini, & Lacquaniti, 1998; Bianchi, Angelini, Orani et al., 1998; Courtine & Schieppati, 2004).

3.3.3: Statistical analysis

Significant changes in normalized stride length, normalized maximum step height, maximum joint flexion angles, segment elevation angle range, planarity, loop width, plane orientation, and fundamental harmonic phase differences were observed using an obstacle height (0, 10, 20% height) by obstacle depth (~4, 10, 20% depth) two-way analysis of variance (ANOVA) for each limb. To determine differences between leading and trailing limbs; a dependent measures t-test was used which grouped all obstacle conditions together. Significant changes in horizontal toe–obstacle distance prior to clearance, toe clearance height, heel clearance height, and horizontal heel–obstacle distance following clearance were observed using an obstacle height (10, 20% height) by obstacle depth (~4, 10, 20% depth) two-way ANOVA for each limb where all 0% obstacle height conditions were removed. Again, to determine differences between leading and trailing limbs, a dependent measures t test was used. If any ANOVA was found to be significant, a Tukey post hoc test was used to determine differences between obstacle configurations. To determine the relationships between thigh–shank phase difference and loop width and between shank–foot phase difference and plane orientation for each limb, Pearson correlations were used. For all statistical analyses, significance was determined when $p < 0.05$.

3.4: Results

3.4.1: Joint angles and stride characteristics

Table 3.1: Maximum joint flexion angles and elevation angle range for lead and trail limbs when clearing obstacles.

	Lead				Trail			
	Level Walking	10% Height	20% Height	Mean	Level Walking	10% Height	20% Height	Mean
Max Joint Angle (°)								
Dorsiflexion	13.2	16.4	18.1	15.9	14.4	14.0	15.5	14.6
SD	+/- 2.5	+/- 3.5^{a}	+/- 4.5^{a}	+/- 4.1	+/- 2.2	+/- 3.3	+/- 3.2	+/- 3.0
Knee Flexion	65.6	90.6	104.4	86.9	66.1	94.8	109.1	90.0
SD	+/- 5.5	+/- 6.1^{a}	+/- 6.4^{a,b}	+/- 17.2	+/- 3.6	+/- 5.3^{a}	+/- 6.9^{a,b}	+/- 18.8^{c}
Hip Flexion	27.0	46.6	58.0	43.9	23.7	31.1	36.6	30.5
SD	+/- 4.9	+/- 6.5^{a}	+/- 6.8^{a,b}	+/- 14.2	+/- 2.8	+/- 6.6^{a}	+/- 7.3^{a,b}	+/- 7.9^{c}
Angle Range (°)								
Foot	111.8	109.5	110.1	110.4	109.7	129.4	141.0	126.7
SD	+/- 7.3	+/- 6.1	+/- 6.8	+/- 6.9	+/- 6.3	+/- 8.8^{a}	+/- 8.8^{a,b}	+/- 15.2^{c}
Shank	87.1	91.3	95.0	91.1	86.9	100.4	112.7	100.0
SD	+/- 3.4	+/- 2.7^{a}	+/- 3.4^{a,b}	+/- 4.5	+/- 3.0	+/- 4.3^{a}	+/- 5.9^{a,b}	+/- 11.6^{c}
Thigh	48.6	71.7	83.7	68.0	46.3	55.9	60.0	54.1
SD	+/- 4.1	+/- 3.7^{a}	+/- 5.3^{a,b}	+/- 15.3	+/- 4.0	+/- 3.9^{a}	+/- 5.3^{a,b}	+/- 7.3^{c}

[a] – significantly different from level walking
[b] – significantly different from 10% height
[c] – significantly different from lead limb

Maximum relative joint flexion angle generally increased with obstacle height (Figure 3.1; Table 3.1). Maximum relative ankle dorsiflexion angle increased significantly with obstacle height in the leading limb ($F_{(2,81)} = 13.722$, $p < 0.001$) and did not change in the trailing limb ($p > 0.05$). Further analysis showed that the maximum dorsiflexion angle was greater in the obstacle conditions when compared to level walking ($p < 0.001$). Maximum ankle joint dorsiflexion did not differ between leading and trailing limbs ($p > 0.05$). Maximum relative knee flexion angle increased significantly with obstacle height in the leading ($F_{(2,81)} = 306.59$, $p < 0.001$) and trailing ($F_{(2,81)} = 477.66$, $p < 0.001$) limbs. Trailing maximum relative knee angle was greater than the leading ($p < 0.037$). Maximum relative hip flexion angle increased with obstacle height in the leading ($F_{(2,81)} = 192.937$, $p < 0.001$) and trailing ($F_{(2,81)} = 33.701$, $p < 0.001$) limbs, although flexion was greater in the leading when compared with the trailing limb ($p < 0.001$). All segment elevation angle ranges increased significantly with obstacle height (trailing foot range $F_{(2,81)} = 111.24$, $p < 0.001$, leading shank range $F_{(2,81)} = 46.63$, $p < 0.001$, trailing shank range $F_{(2,81)} = 239.00$, $p < 0.001$, leading thigh range $F_{(2,81)} = 515.76$, $p < 0.001$, and trailing thigh range $F_{(2,81)} = 73.22$, $p < 0.001$) except for the leading foot elevation angle range. An obstacle depth effect was also observed in the leading thigh elevation angle range ($F_{(2,81)} = 4.16$, $p < 0.019$) with the 20% obstacle depth condition having the greatest range ($p < 0.014$). Comparisons between limbs showed that trailing segment range was greater than leading in the foot ($p < 0.001$) and shank ($p < 0.001$) segments while thigh elevation range was greater in the leading when compared with the trailing limb ($p < 0.001$).

The obstacles presented to participants caused no change in leading limb normalized stride length ($p > 0.05$, Figure 3.2a) although obstacle height had an effect on trailing limb normalized stride length ($F_{(2,81)} = 5.24$, $p < 0.008$). Further analysis of the trailing limb showed the 20% obstacle height significantly increased normalized stride length when compared with level walking and 10% obstacle height ($p < 0.007$, Figure 3.2b). No differences were observed between leading and trailing limbs ($p > 0.05$, Figure 3.2c).

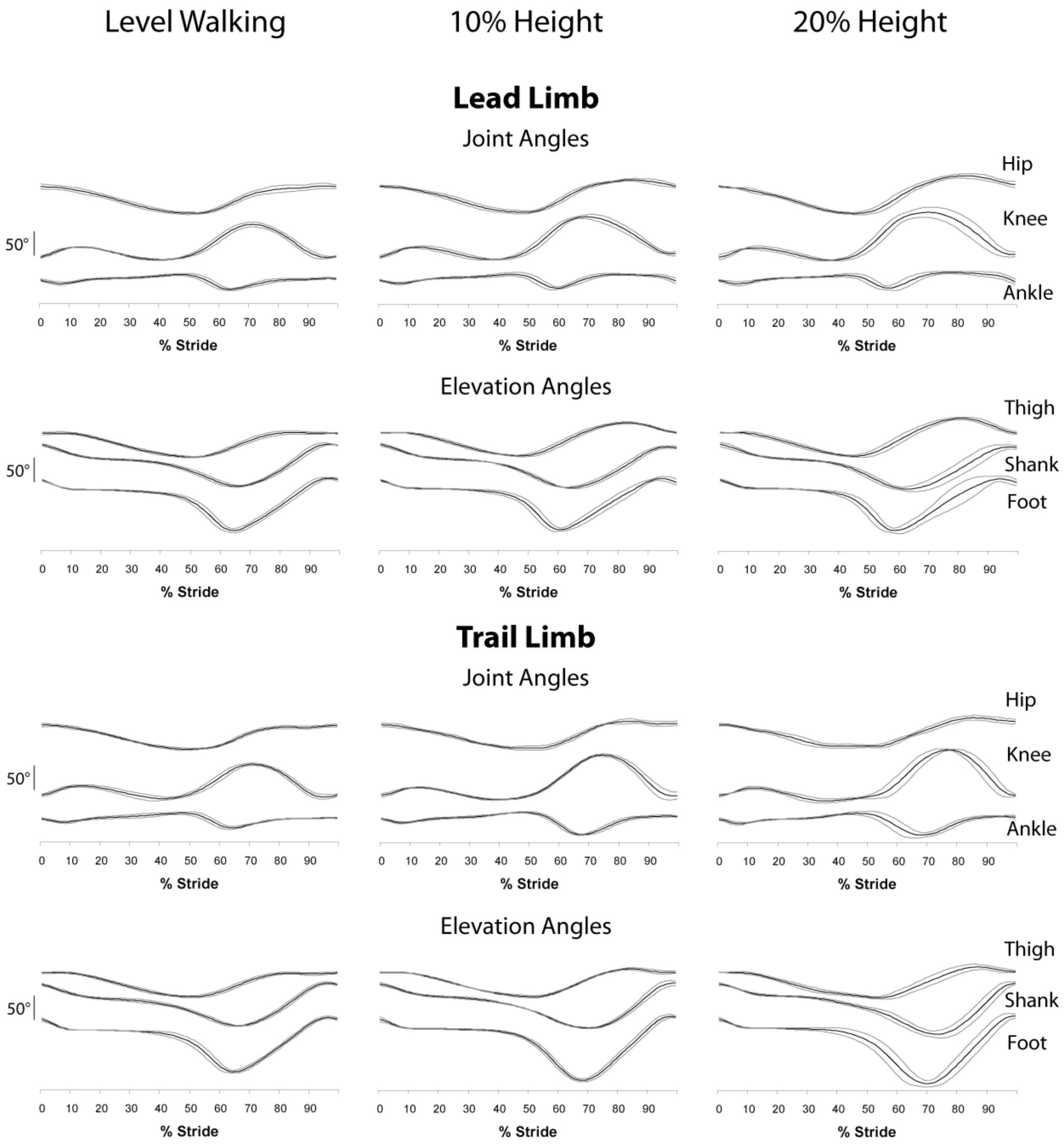

Figure 3.1: Mean (black) +/-2 SD (grey) bilateral joint angle and elevation angle trajectories during level walking and over obstacles of heights of 10 % and 20 % of leg length (~4% of stride length in depth) for a typical participant.

As expected, normalized maximum step height increased significantly with obstacle height in the leading ($F_{(2,81)} = 131.65$, $p < 0.001$, Figure 3.2d) and trailing ($F_{(2,81)} = 138.89$, $p < 0.001$, Figure 3.2e) limbs. Further analysis showed that each increase in obstacle height was accompanied with significant increases in normalized maximum toe height in the leading (p

< 0.001) and trailing ($p < 0.015$) limbs. A comparison of limbs showed that the leading limb had a significantly higher normalized maximum toe height ($p < 0.009$, Figure 3.2f).

3.4.2: Foot proximity and clearance

Horizontal foot placement prior to obstacle clearance and vertical toe–obstacle distance at clearance was not affected by obstacle height or depth in the leading and trailing limbs ($p > 0.05$). However, leading limb clearance was significantly higher than trailing clearance ($p < 0.001$, Figure 3.3a). Heel clearance height in the trailing limb showed a significant obstacle depth effect ($F_{(2,54)} = 8.54$, $p < 0.001$, Figure 3.3b), where heel clearance was significantly higher for the 20% obstacle depth condition when compared with the ~4 and 10% obstacle depths. As well, trailing heel clearance was significantly greater than leading heel clearance ($p < 0.001$, Figure 3.3c) while no differences were observed in the leading limb ($p > 0.05$). Horizontal foot placement after the obstacle was significantly affected by obstacle depth in the leading limb ($F_{(2,54)} = 10.88$, $p < 0.001$, Figure 3.3d) but not the trailing limb ($p > 0.05$). Further analysis showed that heel–obstacle distance was significantly lower for the leading limb in the 20% obstacle depth condition when compared with the ~4 and 10% obstacle depth conditions ($p < 0.042$).

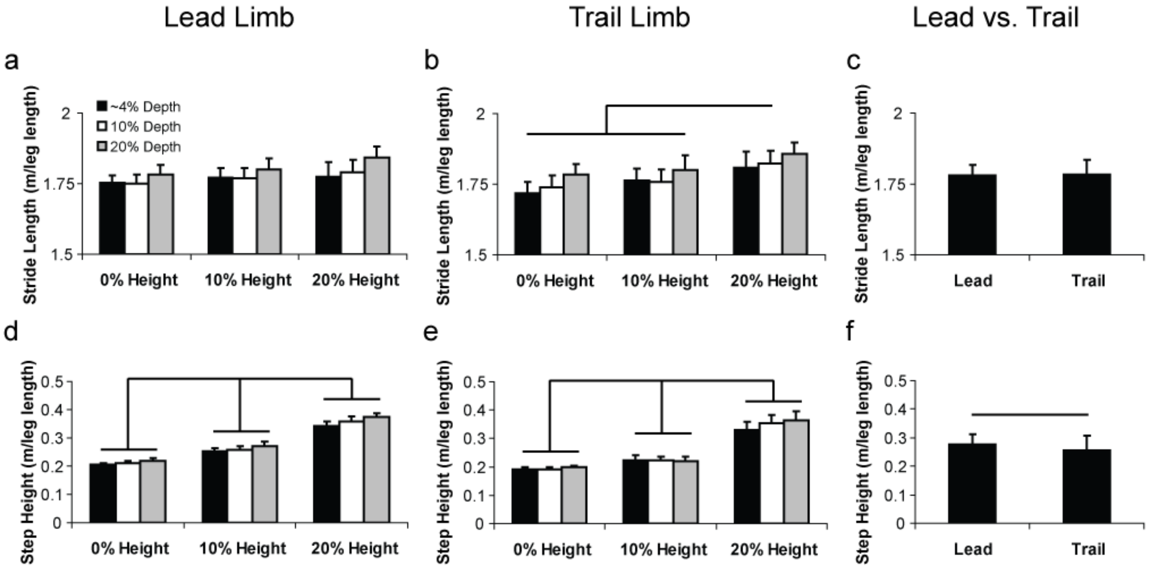

Figure 3.2: Comparisons of normalized stride length (a,b,c) and step height (d,e,f) for level walking and clearance over the different obstacle conditions. Significant differences ($p < 0.05$) are denoted by horizontal bars.

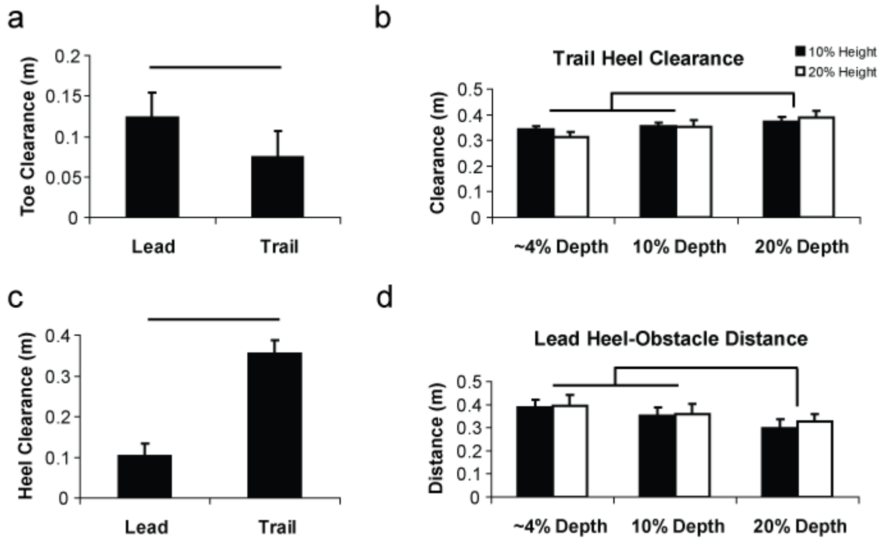

Figure 3.3: Toe clearances across limbs (a), trail heel clearance across obstacle depth conditions (b), heel clearance across limbs (c), and lead heel-obstacle distance following clearance (d). Significant differences ($p < 0.05$) are denoted by horizontal bars.

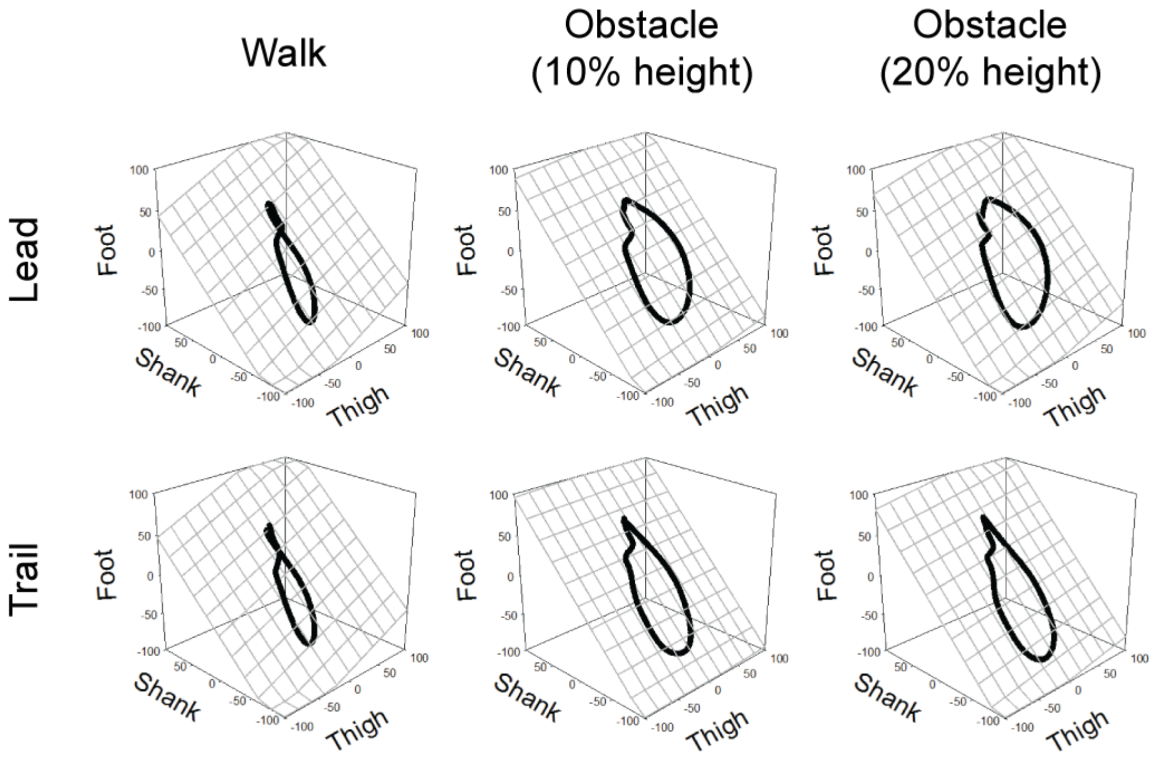

Figure 3.4: Representative covariation plots of elevation angle trajectories for level walking and obstacle conditions with heights of 10 % and 20 % of leg length (~4% of stride length in depth) in the leading and trailing limbs. In the figure, one can see the rotation of the covariance plane and a widening of the covariance loop as obstacle height increases.

3.4.3: Intersegmental coordination

Planarity was consistently high in both limbs for all obstacle conditions (PCs 1 and 2 accounted for 98.99 ± 0.73%; Figure 3.4), although significant height effects were observed in the leading ($F_{(2,81)} = 11.25$, $p < 0.001$, Figure 3.5a) and trailing limbs ($F_{(2,81)} = 19.36$, $p < 0.001$, Figure 3.5b). Further analysis showed that elevation angles were more planar in the level walking condition when compared with the obstacle conditions (leading $p < 0.007$, trailing $p < 0.001$). Planarity was also significantly higher in the trailing when compared with the leading limb ($p < 0.036$, Figure 3.5c).

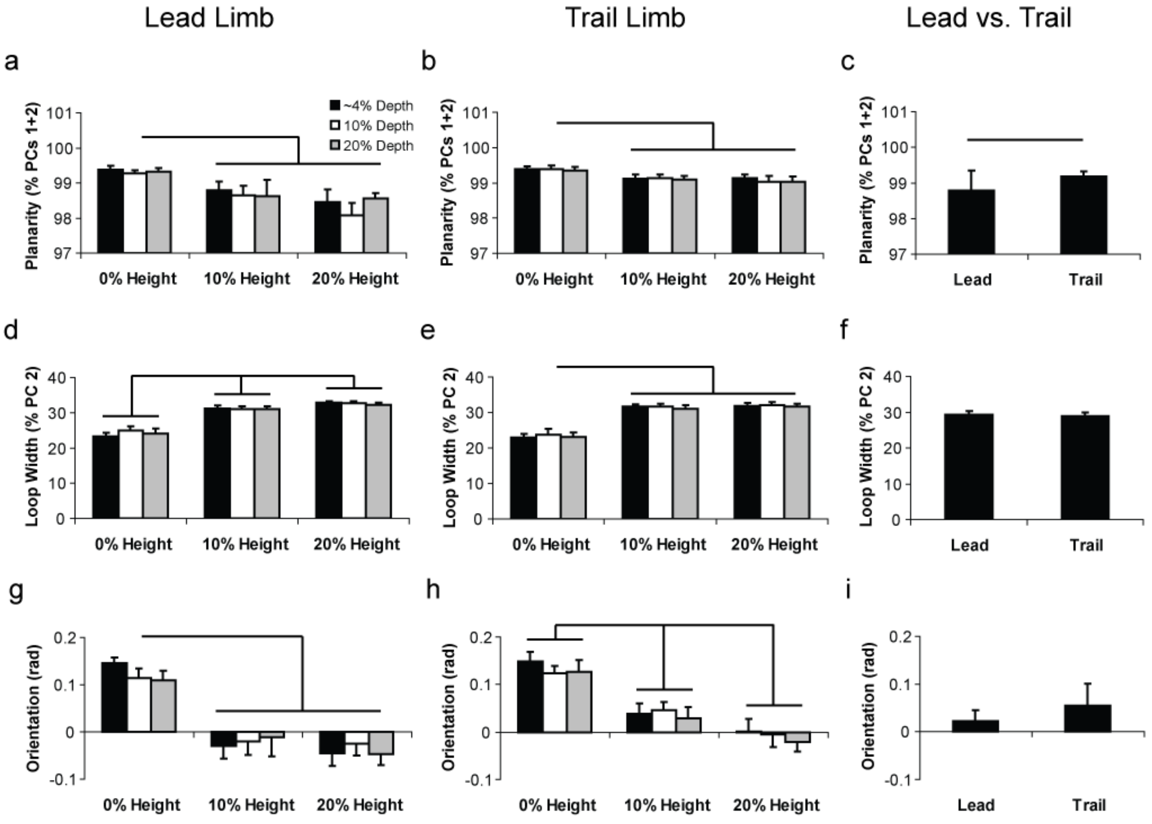

Figure 3.5: Comparisons of planarity (a,b,c), loop width (d,e,f), and plane orientation (g,h,i) across obstacle conditions. Significant differences ($p < 0.05$) are denoted by horizontal bars.

The width of the planar covariance loop increased with obstacle height. For the leading limb, a significant height effect was observed ($F_{(2,81)} = 613.68$, $p < 0.001$) and post hoc analysis showed that the width of the loop increased with each increase in obstacle height ($p < 0.004$, Figure 3.5d). In the trailing limb, a height effect was also shown ($F_{(2,81)} =$

266.33, $p < 0.001$) and further analysis showed that the loop width during obstacle clearance was greater than level walking only ($p < 0.001$, Figure 3.5e). No differences were observed between leading and trailing limbs ($p > 0.05$, Figure 3.5f).

The covariance plane rotated counter-clockwise about its longitudinal axis (orientation becomes more negative) when stepping over obstacles. This was shown by significant height effects in the leading ($F_{(2,81)} = 135.85$, $p < 0.001$, Figure 3.5g) and trailing ($F_{(2,81)} = 56.36$, $p < 0.001$, Figure 3.5h) limbs. In the leading limb, further analysis suggested that the plane rotation was greater in the obstacle conditions when compared with level walking ($p < 0.001$), but no differences existed between obstacle height conditions ($p > 0.05$). In the trailing limb, the plane rotated significantly more counter-clockwise with each increase in obstacle height ($p < 0.004$). Plane orientation did not differ significantly between leading and trailing limbs ($p > 0.05$, Figure 3.5i).

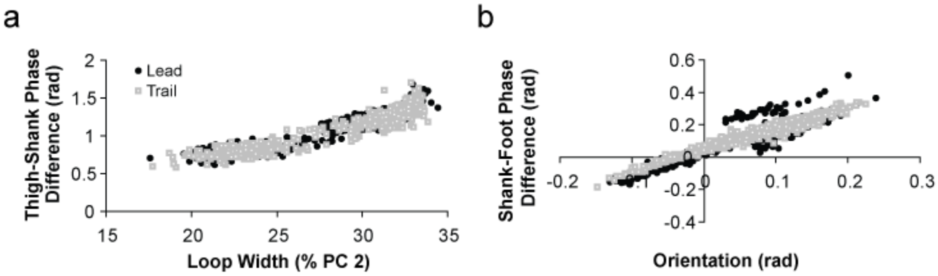

Figure 3.6: Plots of the relationships between thigh-shank phase difference and covariance loop width (a) and shank-foot phase difference and covariance plane orientation (b). Each point on the plot for the lead (black dots) and trail – (grey dots) limbs denotes a single trial.

Now that it has been shown that elevation angles tend to lie on a plane when stepping over obstacles of various configurations and that characteristics of this plane (loop width and plane orientation) change with obstacle height, it will be determined if there are strong relationships between these characteristics and fundamental harmonic phase difference as shown previously for level walking (Bianchi, Angelini, Orani et al., 1998; Courtine & Schieppati, 2004). Pearson correlations showed significant relationships between loop width and thigh–shank fundamental harmonic phase difference for the leading ($r = 0.955$, $p < 0.001$) and trailing ($r = 0.938$, $p < 0.001$) limbs (Figure 3.6a) as well as between plane orientation and shank–foot fundamental harmonic phase difference for the leading ($r = 0.940$, $p < 0.001$) and trailing ($r = 0.979$, $p < 0.001$) limbs (Figure 3.6b).

Because the significant relationship between specific plane characteristics and fundamental harmonic phase difference in adjacent segments has been demonstrated, changes in coordination by observing how these phase differences change when clearing an obstacle will be determined. Analysis showed that the thigh harmonic consistently led that of the shank in all obstacle conditions and this phase difference increased significantly with obstacle height for the leading ($F_{(2,81)} = 308.27$, $p < 0.001$, Figure 3.7a) and trailing ($F_{(2,81)} = 230.43$, $p < 0.001$, Figure 3.7b) limbs. Further analysis suggested that this phase difference increased significantly in each obstacle condition (leading $p < 0.001$, trailing $p < 0.012$). A significant increase was also observed for thigh–shank phase difference in the leading limb when compared with the trailing limb ($p < 0.032$, Figure 3.7c).

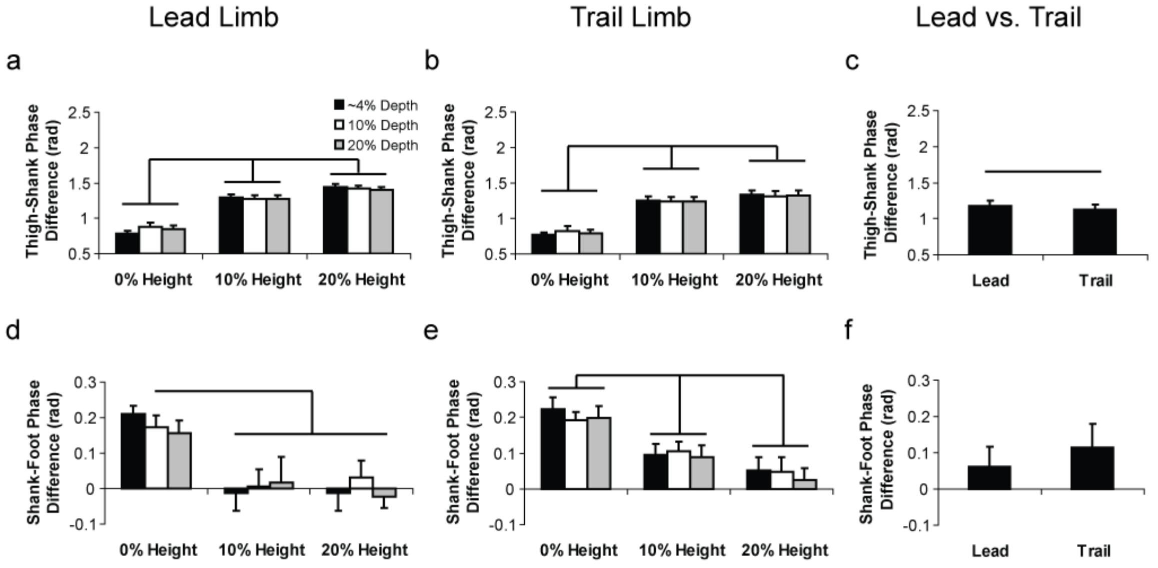

Figure 3.7: Comparisons of thigh-shank phase difference (a,b,c) and shank-foot phase difference (d,e,f), for level walking and clearance over the obstacle. Significant differences ($p < 0.05$) are denoted by horizontal bars.

Phase relationship changes were also observed between the shank and foot during obstacle clearance. In the leading limb, a significant height effect was shown ($F_{(2,81)} = 38.32$, $p < 0.001$, Figure 3.7d) and further analysis showed that shank–foot phase difference decreased when stepping over the obstacle when compared with level walking ($p < 0.001$). The trailing limb showed a significant height effect as well ($F_{(2,81)} = 36.06$, $p < 0.001$, Figure

3.7e) and post hoc analysis showed that the phase difference of the shank with respect to the foot harmonic significantly decreased as obstacle height increased ($p < 0.016$). Shank–foot harmonic difference was also similar between limbs ($p > 0.05$, Figure 3.7f).

3.5: Discussion

As participants stepped over the obstacles presented in this study, changes in lower limb coordination were observed in the leading and trailing limbs. Similar to what has been observed previously for obstacle clearance (Austin et al., 1999; Chou & Draganich, 1997; McFadyen & Winter, 1991), maximum joint flexion angles tended to increase with obstacle height that were likely related to increases in elevation angle range. These changes were accompanied by a slight decrease in segment elevation angle planarity and increases in elevation angle loop width and covariance plane rotation. The changes observed in elevation angle plane characteristics seem to be related to fundamental harmonic phase differences between adjacent segments, which suggest simplified elevation angle waveform control by the CNS to accommodate environmental constraints during locomotion.

3.5.1: Underlying kinematics for the avoidance of obstacles of different configuration

In this study, participants were asked to step over obstacles that differed in both height and depth. Changes in obstacle height led to increases in toe elevation in the leading and trailing limbs while obstacles of approximately 0.15 m in depth did not have a significant effect on stride length. Patla & Rietdyk (1993) used a similar protocol with low obstacles of different depths and found that the deepest obstacle (0.268 m) caused step length changes. In the current study, only the greatest obstacle height increased stride length significantly, but there was still no effect of obstacle depth. This increased stride length for the highest obstacle was most likely due to a higher foot trajectory given that a greater amount of space would be needed to elevate and lower the foot smoothly. However, because obstacle depth was not a factor, these results supplement those of Patla & Rietdyk (1993) in that changes in obstacle depth up to at least 20% of leg length have minimal effects on step kinematics regardless of obstacle height.

Foot placement after obstacle clearance tended to decrease as obstacle depth increased. Given that stride length and foot position prior to obstacle clearance did not change, differences in heel–obstacle distance after clearance are expected for a deeper obstacle. In addition, because heel clearance height at the far end of the obstacle was consistent between obstacle conditions, the CNS may control for this clearance distance instead of foot placement after the obstacle.

Finally, although toe clearance did not differ between obstacle conditions, differences were observed between the leading and trailing limbs. Toe clearance was significantly higher in the leading limb when compared with the trailing limb as seen previously (Patla et al., 1996). This highlights that the leading limb may be controlled separately from the trailing limb and this will be discussed in the following sections.

3.5.2: Implementation of ALAs may be explained using the planar law of intersegmental coordination

As discussed previously, past work has shown planarity of elevation angles in many modes of locomotion. Ivanenko et al. (2005) studied how the elevation angle plane changed when voluntary tasks were added to locomotion; one of these tasks being obstacle clearance. However, these authors studied walking over a single obstacle (0.30 m in height) by one limb only and observed that segment elevation angles were planar although the orientation of the plane deviated from that of level walking. The present results confirm the conservation of the overall tear-drop shape of the covariance plane loop during clearance of different obstacle configurations in both leading and trailing limbs showing that this pattern is robust when avoiding obstacles of various configurations. Although planarity tended to decrease in the present study as participants stepped over obstacles (and was significantly lower in the leading limb), the first two principal components still explained over 98% of the total variance for the segment elevation angles in all obstacle conditions. Because an extremely large percentage of the variance was explained by the first two components, this suggests that the CNS is able to decrease the available degrees of freedom for stepping over obstacles as suggested for level walking (Borghese et al., 1996; Lacquaniti et al., 1999). This, therefore, provides a mechanism for how lower limb patterns to step over an obstacle emerge from level walking patterns as was previously suggested by modeling (McFadyen et al., 1993; Taga, 1998) rather than being considered as new patterns. However, the slight decrease in planarity may suggest the CNS allows for a small amount of variability when adjusting elevation angle patterns during locomotion. This small amount of variability in turn leads to a slight decrease in planarity.

In the current study, strong relationships were observed between thigh–shank phase difference and loop width as well as between shank–foot phase difference and plane orientation. Although these relationships have been documented previously (shank foot: (Bianchi, Angelini, Orani et al., 1998), thigh shank: (Courtine & Schieppati, 2004)), it was unknown until now if they persisted when elevation angles deviated greatly from level walking patterns. The fact that they do persist may suggest that phase differences between adjacent segments may be related to the control of covariance plane characteristics.

With respect to the more proximal segments, as obstacle height increased, significant increases in the thigh–shank phase difference were observed. These phase differences are most likely related to the relative knee flexion angle increases as participants stepped over the obstacle. When stepping over an obstacle, the CNS may alter the phase relationship between the thigh and shank segments which in part leads to an increase in knee flexion. This idea has been suggested previously in Das & McCollum (1988), but the present study supplements this by providing an experimental example for human walking. The results also showed that the width of the plane loop increased as subjects stepped over increasingly higher obstacles, particularly in the leading limb. A similar analysis was used by Courtine & Schieppati (2004) who concluded that changes in loop width occurred between limbs during curved walking. With visual inspection of the elevation angle loop (see Figure 3.4), it can be seen that the increases in loop width are mainly along the thigh elevation angle axis (with some possible contributions from the shank elevation angle depending on the orientation of the plane). This means the width of the covariance plane loop may be heavily dependent on the thigh segment trajectory. Taken together with the increasing thigh–shank phase difference, it is clear why these two variables correlate.

As for the more distal segments, shank–foot phase difference was lower for obstacle clearance when compared with level walking in the leading limb and this phase difference decreased significantly with obstacle height in the trailing limb. This change most likely occurs due to the increases in ankle dorsiflexion seen in obstacle clearance (Patla & Rietdyk, 1993) and may be controlled by a similar phase mechanism in the CNS as discussed for the more proximal segments. As for the elevation angle plane, it rotated counter-clockwise in both limbs when stepping over an obstacle and was significantly different between obstacle conditions in the trailing limb. Barliya et al. (2009) recently used a mathematical formulation to show a direct relationship between shank/foot phase differences and elevation angle plane orientation. Previous literature has suggested that changes in plane orientation are related to changes in limb mechanical power output, and, in particular, that the CNS may economize limb energy expenditure during locomotion through elevation angle phase changes which in turn affect the orientation of this covariance plane (Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999). Therefore, the counter-clockwise rotations of the covariance plane observed in this study may represent the underlying optimization by the CNS to perform obstacle clearance while minimizing energy expenditure. In other words, the plane orientation changes observed when stepping over obstacles indicates the elevation angles which minimize the energy expenditure to step over the obstacle. Further work relating plane orientation to total work done for obstacle avoidance strides will be required and may even suggest that shank–foot phase difference is directly related to energetic efficiency.

Because the foot and shank movements dictate the orientation of the plane, it is thought that the thigh segment adapts to the shank/foot to maintain planar covariance (Bianchi, Angelini, Orani et al., 1998). This relationship is most likely exhibited through the correlation between thigh–shank phase difference and loop width. Because these relationships are also observed in anticipatory locomotor adjustments, it strengthens the argument that these phase relationships may be one of the variables used by the CNS to coordinate adjusted limb movement during locomotion (Barliya et al., 2009; Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999). In fact, it has been proposed that neural oscillators in the CNS control segment elevation angles through segmental waveforms with phase differences and waveform amplitude being two of the control variables (Bianchi, Angelini, Orani et al., 1998). In the current study, the same control variables appear to hold for stepping over obstacles given the relationships found between covariance loop width and thigh–shank harmonic phase difference and between covariance plane orientation and shank–foot harmonic phase difference.

It has been further suggested that each limb is controlled by a separate neural oscillator (Wannier, Bastiaanse, Colombo, & Dietz, 2001). Although there must be coupling between the oscillators to coordinate inter-limb movements, segment phase difference between limbs was not similar which reiterates the idea that leading limb control is separate from that of the trailing limb (Mohagheghi et al., 2004; Niang & McFadyen, 2004; Patla et al., 1996). Limb differences in thigh–shank phase are most likely due to a delay in trailing limb hip movement to avoid collision during limb elevation for higher obstacles (Niang & McFadyen, 2004). Therefore, this work has shown specific segmental coordination differences related to safety issues supporting previous work on joint kinematics and dynamics.

The results in the current study show that implementation of an ALA by the CNS may be controlled by similar mechanisms as controlling other modes of locomotion. If a given environment requires a change in locomotion, the CNS adjusts the basic pattern by manipulating elevation angle fundamental harmonic phase difference between adjacent segments and elevation angle amplitude (with a constraint being planarity of elevation angles) and a new locomotor pattern emerges which will allow for successful movement through the present environment. The effects of the altered pattern are changes in elevation angle plane orientation, loop width, and the observed relative joint angles.

Throughout the results, we did not see any significant differences between obstacle depth conditions in the coordination variables. This is most likely because the obstacle depth conditions presented here did not require a significant change in stride length. Although a significant change in stride length was not observed, another reason for this could be that when lengthening a stride, the phase relationships between segments are similar to a normal stride; they are just extended over a longer period and intersegmental coordination may not change. Further research will be needed to determine if intersegmental coordination changes when lengthening a stride and if phase relationships between adjacent segments are related to this.

3.5.3: What is the role of vision in the observed intersegmental coordination?

The role of visual information on the planar law of intersegmental coordination has been discussed by Courtine & Schieppati (2004). These authors found no difference in planarity between straight walking and curved walking with and without visual information. It is possible that no differences were observed in these two tasks, because visual information of limb position is used minimally in curved walking. However, vision is used to a greater extent in obstacle clearance, particularly for the leading limb. As stated above, planarity may be a constraint in elevation angle control. Differences observed between leading and trailing limbs, with higher planarity seen in the trailing limb, may also be explained by visual influences. Because online visual information is not available for the trailing limb (Mohagheghi et al., 2004) the CNS may be more cautious and adapt the coordination pattern in a more constrained way and not allow for additional variability when adjusting elevation angles. For the leading limb, where online visual information is available, the CNS may allow for slightly more variability because one is able to receive visual feedback regarding the movement although the movement is still simplified to two degrees of freedom (as shown by the high level of planarity). This shows that variability in planarity may in part be due to two separate factors: increases in obstacle height as well as visual information.

3.6: Conclusions

Using the planar law of intersegmental coordination, the current study showed that systematic changes occur in characteristics of the resulting covariance planar loop (specifically plane orientation and loop width) when stepping over an increasingly higher obstacle. It is proposed that visual input is used to determine if the current locomotor pattern is appropriate for the environment and the CNS adjusts a basic locomotor pattern if needed by manipulating elevation angle fundamental harmonic phase differences between adjacent segments and elevation angle amplitude (with a constraint being planarity of elevation angles). A new locomotor pattern then emerges which is appropriate for the environment. Changes in elevation angle fundamental harmonic phase difference were observed between leading and trailing limbs, which is likely due to altered hip control in the trailing limb to avoid collision with the obstacle and the differences in the amount of visual input available for the clearance of each limb.

3.7: Acknowledgments

This study was funded by a Natural Sciences and Engineering Research Council of Canada grant for Bradford McFadyen and scholarship for Michael MacLellan. The authors thank Mr. Guy St-Vincent for his technical support.

3.8: Bridging Paragraph

Results from Chapter 3 have shown that when healthy adults step over obstacles of various configurations, lower limb segment elevation angles tend to form a planar pattern in the leading and trailing limbs. It was suggested that visual input assesses the appropriateness of the current locomotor pattern for the environment and the CNS makes adjustments to a basic locomotor pattern if a change is required. The conclusions argue that phasing differences between adjacent segment elevation angle waveforms are a control variable by neural oscillators in the CNS. An opposing theory of motor control is that movement kinetics are controlled by the CNS. The idea of joint specificity (Niang & McFadyen, 2004), where muscle power generated at the hip progresses the lower limb forwards and muscle power generated at the knee elevates the limb upward would be an example of this. Chapter 4 will examine if this type of kinetic control exists in the lower limb during obstacle clearance.

Chapter 4: Study 2

“Relationships between segment elevation angles and muscle power during obstacle clearance reveal multiple goals for the mechanical work at hip and knee joints”

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4.1: Abstract

The purpose of this study was to first determine individual segmental contributions to previously reported increases in thigh-shank phase difference when stepping over higher obstacles and, second, to use this information to determine the contribution of mechanical work done by hip and knee muscle flexors to limb progression and elevation during the transition to swing. Ten healthy young adults stepped over obstacles of different heights and depths. Elevation angles of the thigh and shank segments were calculated. A Fourier harmonic series was used to represent segment elevation angle trajectories and to quantify phase shifting across conditions. The positive mechanical work from hip flexor power bursts at toe-off (H3) or delayed into swing (H3D), and the knee flexor power burst at toe-off (K5) were estimated using link segment analyses and related to stride length and maximum toe height as well as the timing of elevation angle turning points. Results showed that as higher obstacles were cleared, leading limb thigh phase lead and trailing limb shank phase lag increased. Correlations between the work done by the H3 muscle power burst and limb progression as well as the K5 muscle power burst and limb elevation showed weak relationships, suggesting the hip and knee joints do not have single specific functions during obstacle clearance. In the leading limb, the onsets of the H3 bursts across conditions synchronized with the minimum peak of the thigh elevation angle despite that fact that these bursts did not increase with obstacle height, suggesting its role in the initiation of thigh forward movement, but not limb progression. This relationship was not shown in the trailing limb, suggesting thigh forward movement initiation occurs due to other mechanisms. A decrease of H3 muscle power and an increase in K5 muscle power with

obstacle height suggested knee flexor influence on thigh elevation amplitude in the trailing limb. It was concluded that these flexor muscle power bursts contribute both to elevating and progressing the lower limb and result from the control dynamics whereby segment elevation angle phase differences may be considered control variables by the CNS using separate bilateral control.

4.2: Introduction

Anticipatory locomotor adjustments (or ALAs) are required in order to walk through environments in which objects obstruct the locomotor path. Such situations include stepping up to a new level or stepping over an obstacle. Although tasks such as these are performed numerous times throughout a person's day, very little is known about how the central nervous system (CNS) controls and coordinates these locomotor adjustments. From a biomechanical perspective, this control has been explored previously predominantly using kinetic and kinematic data.

When human participants step over obstacles, a reorganization of muscle power occurs. In particular, the work done by the knee extensor muscles absorbing energy prior to and following push-off (K3 muscle power burst) and the hip flexor muscles generating energy for hip pull-off (H3 muscle power burst, (Winter, 1987)) decreases and a new energy generation knee flexor muscle power burst (K5) appears in the leading limb (McFadyen & Winter, 1991). This K5 muscle power burst generates the energy needed to flex the knee and hip (McFadyen & Winter, 1991) with hip flexion occurring due to distal intersegmental forces acting at the thigh (Patla & Prentice, 1995), which leads to an elevation of the limb over the obstacle. In the trailing limb, a similar reorganization of muscle powers occurs but the K3 muscle power burst is delayed (termed K3D) and an additional hip flexor muscle burst appears (termed H3D) (Niang & McFadyen, 2004). This change in muscle power reorganization for the trailing limb is thought to occur due to the close proximity of the foot whereby the limb is initially elevated by the K5 muscle power burst and subsequently the H3D muscle power burst progresses the limb through swing therefore avoiding foot contact with the obstacle (Niang & McFadyen, 2004). These observations led Niang & McFadyen (2004) to suggest that the control of this movement is related to specific tasks at each joint whereby the K5 muscle power burst acts to elevate the limb and the H3 (in the leading limb) or H3D (in the trailing limb) power bursts act to progress the limb through swing. This idea was referred to as joint specificity (Niang & McFadyen, 2004). Although the idea of limb progression being due to hip musculature has been suggested through forward dynamic modelling (Neptune et al., 2004), this hypothesis regarding joint specificity has not been tested directly.

Previous studies examining lower limb kinematic data during obstacle clearance have generally focused on joint angle data (Austin et al., 1999; Chou & Draganich, 1997, 1998b; Patla et al., 1991; Patla & Rietdyk, 1993; Patla et al., 1996). A different perspective of lower limb motion can be obtained by using segment absolute angles. In particular, segment elevation angles have been used to describe lower limb movement with theories based on the planar law of intersegmental coordination. According to this law, if the elevation angle trajectories of thigh, shank, and foot segments are plotted in 3-dimensional space for a stride, they create a loop which tends to orient on a plane in this space (Borghese et al., 1996). This phenomenon has been observed in various forms of locomotion (Bianchi, Angelini, & Lacquaniti, 1998; Bianchi, Angelini, Orani et al., 1998; Cappellini et al., 2010; Courtine & Schieppati, 2004; Grasso et al., 1998; Grasso et al., 2000; Ivanenko et al., 2008; Noble & Prentice, 2008) including obstacle clearance (Ivanenko et al., 2005; MacLellan & McFadyen, 2010). Although lower limb segment elevation angle trajectories are planar when stepping over obstacles, the orientation of this plane changes systematically when stepping over obstacles of increasing size (Ivanenko et al., 2005; MacLellan & McFadyen, 2010). As well, the elevation angle range and the phase difference between the thigh and shank segments increases in the leading and trailing limbs which has been related to the observed increases in the width of the loop formed by the elevation angles (MacLellan & McFadyen, 2010).

The significance of the planar law of intersegmental coordination is that it suggests a simplification of control in the CNS (Bianchi, Angelini, Orani et al., 1998; Borghese et al., 1996; Lacquaniti et al., 1999). In addition, it has been suggested that neural oscillators in the CNS control lower limb movement during human locomotion through segment elevation angle waveforms where waveform amplitude and the phase difference between adjacent segment waveforms are control variables (Lacquaniti et al., 1999). Such ideas of waveform control by the CNS have been discussed by Das & McCollum (1988) and Shen & Poppele (1995). MacLellan & McFadyen (2010) suggested that a similar control mechanism is used when implementing ALAs for obstacle clearance, although these theories have not been related to the kinetics of locomotion.

The hip and knee joints have critical roles in successful obstacle clearance. Niang & McFadyen (2004) previously suggested that joint specificity may be a locomotor control mechanism in the lower limb where knee muscle power (K5) elevates the limb over the obstacle and hip muscle power (H3 and H3D in the trailing limb) acts to progress the limb. The current study first determined how individual segmental phase shifts were involved in the increases in thigh-shank phase difference as previously reported (MacLellan and McFadyen, 2010) when stepping over higher obstacles. Then, in order to determine if joint specificity is a mechanism of control in the CNS during obstacle clearance, this information on segmental contributions to phase shifting was used to determine which segments to target in relating the timing and amplitude of segment elevation changes for obstacle avoidance to mechanical work by hip and knee muscle flexors at the transition.

4.3: Methods

Ten healthy young adults (6 female/4male, 27.7 +/- 5.7 years, 72.0 +/- 16.0 kg in weight, 1.72 +/- 0.07 m in height) participated in the study. Prior to data collection, all participants provided informed consent according to ethical guidelines from the Quebec Institute of rehabilitation and physical deficiency (IRDPQ) and Laval University.

4.3.1: Protocol

The experimental protocol used in this study has previously been presented in MacLellan & McFadyen (2010). In brief, participants were asked to step over 9 obstacle conditions in which obstacle height (0, 10, 20% of leg length) and obstacle depth (0, 10, 20% of step length) were manipulated. The 0% height/0% depth condition referred to level walking with no obstacle present. Obstacles obviously always have some depth, thus for these conditions, 0% depth was actually 0.025 m (or approximately 4% of step length), which corresponded to the minimum depth possible for the obstacle used in this study. Each obstacle condition was presented in a block of 5 trials, with a total of 45 trials. Participants were not informed of which limb to lead with when stepping over the obstacle, but were asked always lead with the same limb. This resulted in 9 participants leading with the right limb and 1 with the left limb.

Full body 3D kinematic data we collected at 75 Hz using a 3-bar Optotrak camera system (Northern Digital Inc., Waterloo, Canada). To track body segment movement, triads of non-collinear infrared emitting diodes (IREDS) were affixed to rigid plastic plates which were then subsequently attached to the feet, shanks, thighs, pelvis, trunk, and head segments. After a calibration trial was collected, anatomical landmarks (5th metatarsal, medial/lateral malleolus, medial/lateral femoral condyles, left/right iliac spine, and left/right anterior superior iliac spine) were digitized including virtual points for the heels and toes of each foot in order to determine their trajectories. 3D IRED position data were filtered offline using a dual-pass 2nd order Butterworth filter with a cut-off frequency of 6 Hz.

Ground reaction forces and moments were collected at 1000 Hz from the leading and trailing limbs prior to clearance using 3 AMTI force platforms (Advanced Mechanical Technology, Inc., Watertown, Massachusetts). Kinetic data were filtered offline using a dual-pass 2nd order Butterworth filter with a cut-off frequency of 50 Hz.

4.3.2: Data Analysis

The kinematic data were used to create a 9 linked segment biomechanical model consisting of feet, shanks, thighs, pelvis, trunk, and head. From this model, stride length was calculated as the distance in the direction of progression between digitized heel marker positions at consecutive heel contacts of the same foot and maximum toe height as the peak position of the digitized toe marker during the stride respectively. Each of these values was normalized for participant leg length. Using the digitized points recorded prior to data collection, elevation angles of the thighs and shanks were determined for the stride over the obstacle (heel contact to subsequent heel contact) using the formulas provided by Borghese et al. (1996). From the elevation angle trajectories, the maximum peak of the thigh and the minimum peaks of the thigh and shank were located with respect to the proportion of the stride cycle. These peaks were used to calculate the elevation angle range of the thigh and shank segments. In order to quantify shifts in the elevation angle trajectories according to work related to the planar law of intersegmental coordination (Bianchi, Angelini, Orani et al., 1998), a Fourier series using 10 harmonics was used to represent the trajectories of the thigh and shank segments. From this series, the phase shift of the fundamental harmonic with respect to heel contact was then used to represent the phase shift of the segment trajectory and to subsequently determine phase shift changes between the different obstacle conditions.

Three dimensional Newton-Euler inverse dynamics equations were used to determine reaction forces at joint centres and net muscle moments for the lower limb joints. From these data, muscle power at the hip and knee were determined by the dot product of the net muscle moment and the joint angular velocity at each instant of the gait cycle. The K5 (knee flexor generation), H3 (hip flexor generation), and H3D (hip flexor generation) power bursts were then identified and the mechanical work done by each power burst was estimated using mathematical integration. As well, the onsets and offsets of each of these power bursts were located with respect to the proportion of the stride cycle.

4.3.3: Statistical Analysis

Statistically significant differences between obstacle conditions for the phase shift of the thigh and shank segments, points of maximum peak of the thigh segment trajectory and minimum peaks of the thigh and shank segment trajectories, work done by the H3 power burst, and points of onset and offset of the H3 power burst were determined using an obstacle height (0, 10, 20% height) by obstacle depth (0, 10, 20% depth) 2-way ANOVA for each limb. Since the K5 and H3D power bursts were not observed during level walking or when stepping over obstacles with 0% height, statistically significant changes between obstacle conditions for work done by the K5 and H3D power bursts as well as onsets/offsets of these power bursts were determined using an obstacle height (10, 20% height) by obstacle depth (0, 10, 20% depth) 2-way ANOVA for each limb. In participants where an H3D burst did not occur, these participants were removed from the statistical analysis. To determine differences between leading and trailing limbs for all dependent variables (except H3D which was not observed in the leading limb), a dependent measures t-test was used which grouped all obstacle conditions together. In order to find relationships between kinematic and kinetic data amplitudes, Pearson correlations were applied between work done by the H3 power burst (or the sum of the H3 and H3D power bursts in the trailing limb) with stride length and thigh elevation angle range as well as work done by the K5 power burst with maximum toe height and shank elevation angle range. Relationships between the timing of elevation angle turning points and muscle power burst onset were studied using Pearson correlations between H3 onset time and thigh elevation angle minimum peak as well as K5 onset time and shank elevation angle minimum peak. Each of these correlations were performed for obstacle conditions only since K5 muscle power bursts were not observed in level walking. For all statistical tests, a p-value of 0.05 was used to determine statistical significance.

4.4: Results

4.4.1: Kinematic Characteristics

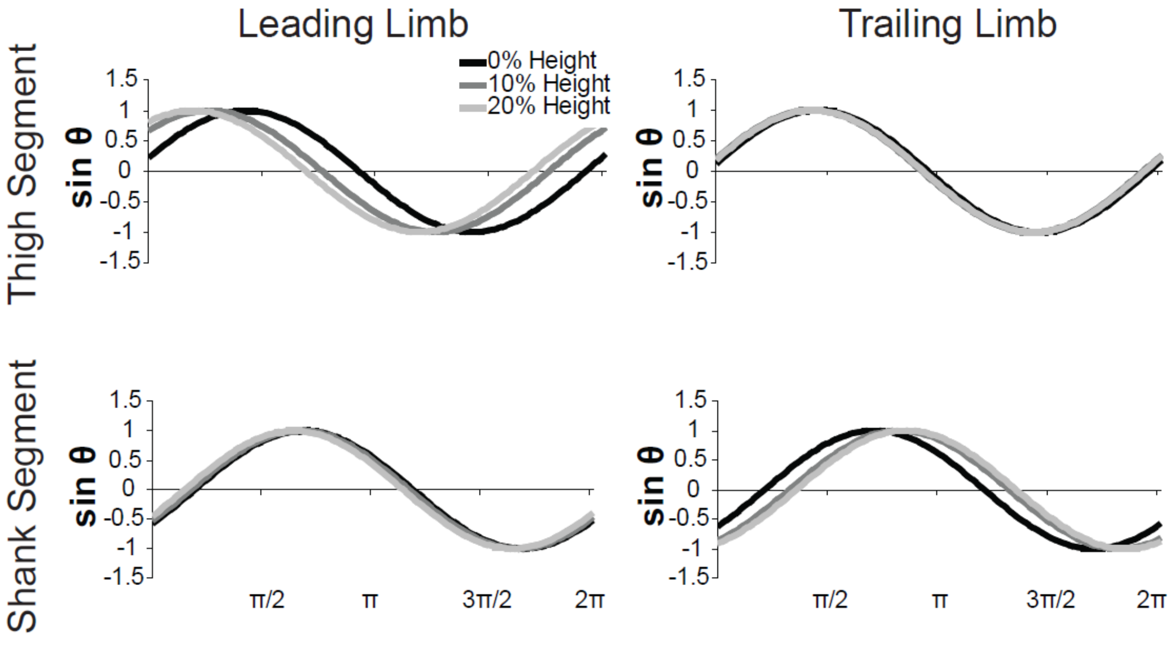

Figure 4.1: Results of average fundamental harmonic phase shifting in the thigh (top) and shank (bottom) segments for the leading (left) and trailing (right) limbs over all participants.

One of the goals of the current study was to determine how the phase shifts of individual segments lead to the increases in thigh-shank phase difference when stepping over higher obstacles. The results showed that the fundamental harmonic phase differences between the thigh and shank segments seemed to be due to shifting by different segments in the leading and trailing limbs. Figure 4.1 illustrates the fundamental harmonic phase shift with respect to heel contact for the thigh (top row) and shank (bottom row) in the leading (left column) and trailing (right column) limbs. On inspection of this figure, it can be seen that greater phase leads are seen in the thigh segment in the leading limb and greater phase lags of the shank segment in the trailing limb.

The phase lead of the thigh fundamental harmonic increased with obstacle height in the leading ($F_{(2,81)} = 382.757$, $p < 0.001$, Figure 4.2a) and was slightly but significantly greater for the obstacle conditions when compared to level walking in the trailing ($F_{(2,81)} = 7.843$, $p < 0.001$, Figure 4.2b) limbs. A dependent t-test showed that the phase lead of the thigh harmonic was greater in the leading when compared to the trailing limb ($p < 0.001$, Figure 4.2c). An obstacle width effect ($F_{(2,81)} = 3.462$, $p < 0.037$) was also shown for the phase shift of the thigh fundamental harmonic in the leading limb where the phase lead was greater in the 20% width obstacle condition when compared to the thin 0% obstacle ($p < 0.030$).

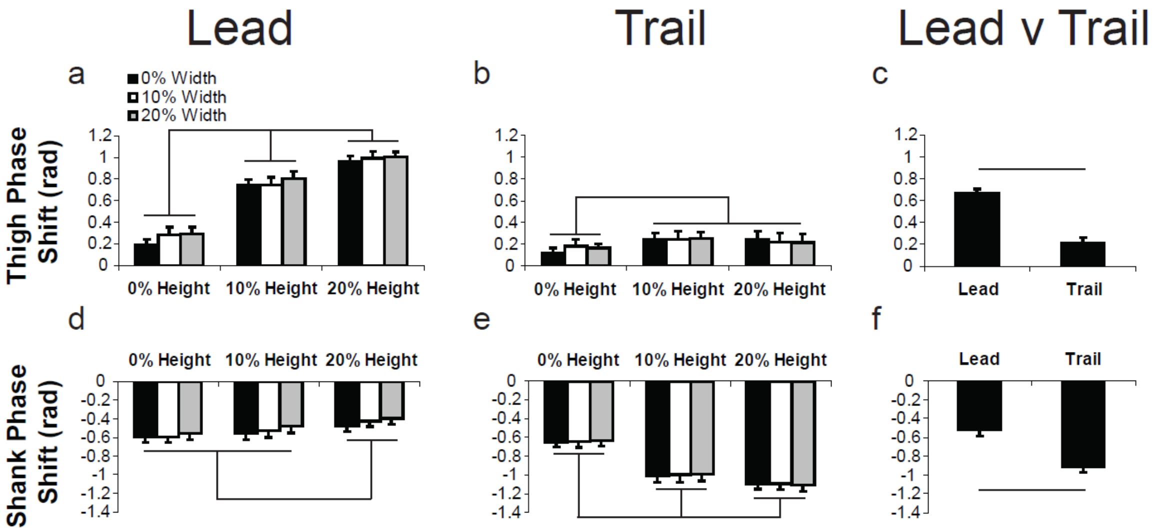

Figure 4.2: Plots of phase shifting in the thigh (top) and shank (bottom) segments. Phase shift values are with respect to heel contact with positive values indicating a leftward shift (phase lead) and negative values indicating a rightward shift (phase lag). Significant differences ($p < 0.05$) are indicated with horizontal bars.

For the phase shift of the shank fundamental harmonic, a significant obstacle height effect was shown in the leading limb ($F_{(2,81)} = 14.574$, $p < 0.001$, Figure 4.2d) as well as for trailing limb ($F_{(2,81)} = 273.28$, $p < 0.001$, Figure 4.2e) although smaller. Further analysis showed that the lag of the phase shift decreased in the 20% obstacle height condition when compared to the 10% obstacle ($p < 0.007$) and level walking ($p < 0.001$) in the leading limb. In the trailing limb, the phase lag increased as obstacle height increased ($p < 0.001$). Comparison between limbs showed that the phase lag was greater in the trailing limb when compared to the leading limb ($p < 0.001$, Figure 4.2f).

Due to these differences in fundamental harmonic phase shifts, the timing of the elevation angle peaks tended to shift as well (Table 4.1). In the leading limb, obstacle height effects were shown for the minimum peak of the thigh ($F_{(2,81)} = 98.99$, $p < 0.001$), shank ($F_{(2,81)} = 28.32$, $p < 0.001$), and maximum peak of the thigh ($F_{(2,81)} = 21.00$, $p < 0.001$) segments and post-hoc analysis showed that these peaks occurred earlier as obstacle height increased (thigh minimum: $p < 0.001$, shank minimum p: < 0.008, thigh maximum: $p < 0.015$). Obstacle height main effects were shown in the trailing limb for minimum peak of the thigh ($F_{(2,81)} = 24.40$, $p < 0.001$), shank ($F_{(2,81)} = 94.93$, $p < 0.001$), and maximum peak of the thigh ($F_{(2,81)} = 5.80$, $p < 0.005$) segments with further tests showing that these peaks occurred later as obstacle height increased (thigh minimum: $p < 0.009$, shank minimum p: < 0.001, thigh maximum: $p < 0.021$). Due to these shifts, each of these peaks occurred significantly earlier in the leading limb ($p < 0.001$).

Table 4.1: Timing of elevation angle peaks for leading and trailing limbs when clearing obstacles. All values in % of gait cycle.

	Lead				Trail			
	Level Walking	10% Height	20% Height	Mean	Level Walking	10% Height	20% Height	Mean
Thigh Min.	50.8	46.7	44.5	47.3	52.5	54.6	56.2	54.4
SD	+/- 2.4	+/- 1.1 ^{a}	+/- 1.5 ^{a,b}	+/- 1.0	+/- 1.6	+/- 2.2 ^{a}	+/- 2.3 ^{a,b}	+/- 1.3 ^{c}
Shank Min.	66.2	63.6	61.8	63.8	67.7	72.7	75.5	72.0
SD	+/- 2.6	+/- 1.7 ^{a}	+/- 2.5 ^{a,b}	+/- 0.8	+/- 1.0	+/- 2.6 ^{a}	+/- 2.5 ^{a,b}	+/- 1.1 ^{c}
Thigh Max.	84.6	81.5	79.1	81.7	86.2	86.5	88.5	87.1
SD	+/- 3.8	+/- 2.4 ^{a}	+/- 3.3 ^{a,b}	+/- 1.5	+/- 3.0	+/- 2.6	+/- 2.6 ^{a,b}	+/- 1.9 ^{c}

[a] – significantly different from level walking
[b] – significantly different from 10% height
[c] – significantly different from lead limb

4.4.2: Kinetic Characteristics

As shown previously, work done by the K5 muscle power tended to increase with obstacle height. This pattern was shown for the leading ($F_{(2,81)}$ = 194.026, $p < 0.001$, Figure 4.3a) and trailing ($F_{(2,81)}$ = 262.55, $p < 0.001$, Figure 4.3b) limbs. The work done by the K5 muscle burst was also significantly larger in the trailing when compared to the leading limb ($p < 0.001$, Figure 4.3c).

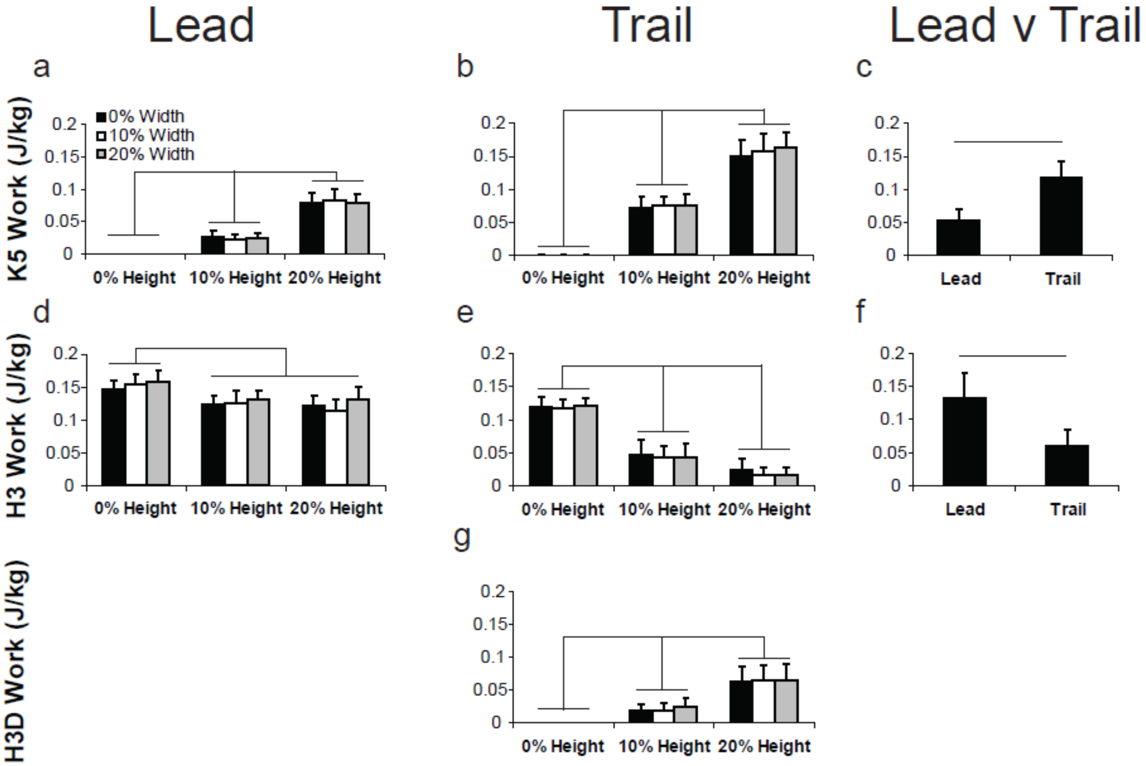

Figure 4.3: Plots of work done for the K5 (top), H3 (middle), and H3D (bottom) muscle power bursts. All magnitudes are normalized to participant body weight. Significant differences ($p < 0.05$) are indicated with horizontal bars.

At the hip, H3 work showed significant obstacle height effects in the leading ($F_{(2,81)}$ = 5.130, $p < 0.008$, Figure 4.3d) and trailing ($F_{(2,80)}$ = 86.685, $p < 0.001$, Figure 4.3e) limbs, much more so for the trailing limb. Further analysis showed that H3 work decreased in the 10% ($p < 0.043$) and 20% ($p < 0.010$) obstacle conditions when compared to level walking in the leading limb and decreased with obstacle height in the trailing limb ($p < 0.009$). Work done by the H3 power burst was also greater in the leading when compared to the trailing limb ($p < 0.001$, Figure 4.3f). As well, the H3 power burst was displayed in all leading limb conditions, and for the trailing limb, in all level walking and 10% obstacle height conditions, and in 9/10 participants for the 20% obstacle height condition. An H3D power burst was observed for the trailing limb in 7/10 participants for the 10% obstacle height condition and in all participants for the 20% obstacle height condition. When the H3D power burst appeared, an obstacle height effect ($F_{(1,45)} = 20.064$, $p < 0.001$, Figure 4.3g) was shown where work done by the H3D burst increased with obstacle height ($p < 0.001$).

Table 4.2: Timing of power bursts for leading and trailing limbs when clearing obstacles. All values in % of gait cycle.

	Lead				Trail			
	Level Walking	10% Height	20% Height	Mean	Level Walking	10% Height	20% Height	Mean
K5 Onset SD	N/A	56.3 +/- 1.8	52.2 ^{b} +/- 2.3	54.2 +/- 1.5	N/A	58.0 +/- 2.7	56.0 ^{b} +/- 2.9	57.0 ^{c} +/- 2.2
K5 Offset SD	N/A	65.8 +/- 4.6	64.9 +/- 3.1	65.4 +/- 3.5	N/A	67.6 +/- 2.5	69.5 ^{b} +/- 2.3	68.6 ^{c} +/- 1.4
H3 Onset SD	52.2 +/- 1.5	47.3 ^{a} +/- 1.4	44.9 ^{a,b} +/- 1.8	48.1 +/- 1.3	53.6 +/- 2.8	53.6 +/- 3.2	52.2 +/- 3.5	53.2 ^{c} +/- 2.6
H3 Offset SD	71.4 +/- 3.4	62.7 ^{a} +/- 2.8	58.3 ^{a,b} +/- 2.4	64.1 +/- 2.5	72.7 +/- 3.2	66.9 ^{a} +/- 6.0	60.1 ^{a,b} +/- 2.3	66.6 ^{c} +/- 2.6
H3D Onset SD	N/A	N/A	N/A	N/A	N/A	70.4 +/- 2.9	69.4 +/- 2.8	69.4 +/- 2.5
H3D Offset SD	N/A	N/A	N/A	N/A	N/A	82.1 +/- 2.9	85.6 ^{b} +/- 2.2	84.3 +/- 1.9

^{a} – significantly different from level walking
^{b} – significantly different from 10% height
^{c} – significantly different from lead limb
N/A – burst did not occur

The onset of the K5 muscle burst (Table 4.2) showed a significant obstacle height effect ($F_{(1,54)} = 58.61$, $p < 0.001$) and post hoc analysis suggested this onset was earlier in the 20% obstacle height when compared to the 10% obstacle height condition ($p < 0.001$). The offset of the K5 muscle burst showed no significant differences with obstacle height ($p > 0.05$). In the trailing limb, the onset of the K5 muscle burst showed an obstacle height effect ($F_{(1,54)} = 7.03$, $p < 0.011$) with the onset being earlier in the 20% obstacle height when compared to the 10% obstacle height condition ($p < 0.011$). The offset of the K5 muscle burst in the trailing limb also showed an obstacle height effect ($F_{(1,54)} = 9.74$, $p < 0.003$) where the offset occurred later in the 20% obstacle height when compared to the

10% obstacle height condition ($p < 0.004$). Comparison between limbs showed that the onset ($p < 0.008$) and offset ($p < 0.008$) of the K5 muscle burst was earlier in the leading limb.

The onset of the H3 power burst (Table 4.2) showed an obstacle height effect in the leading ($F_{(2,81)} = 164.59$, $p < 0.001$) but not the trailing limb ($p > 0.05$). Further analysis of the leading limb showed that H3 power burst onset occurred earlier in the step cycle as obstacle height increased ($p < 0.001$). Obstacle height effects were shown for both the leading ($F_{(2,81)} = 147.72$, $p < 0.001$) and trailing ($F_{(2,80)} = 64.12$, $p < 0.001$) limbs for H3 power burst offset, where the offset occurred earlier as obstacle height increased ($p < 0.001$). Dependent measures t-tests showed that the onset ($p < 0.001$) and offset ($p < 0.007$) of the H3 power burst occurred earlier in the leading limb.

Although obstacle height did not affect H3D burst onset (Table 4.2), an obstacle height effect was observed for H3D offset ($F_{(1,45)} = 21.40$, $p < 0.001$) where offset occurred later in the gait cycle in the 20% obstacle height condition when compared to the 10% obstacle condition ($p < 0.001$).

4.4.3: Kinematic and Kinetic Data Integration

Following the thigh elevation minimum peak during all conditions, an out of phase movement occurs between the thigh and shank (Figure 4.4). At this time the hip and knee are both flexing, leading to an elevation as well as a possible progression of the lower limb. This movement becomes in phase again following the shank minimum peak and continues until the thigh maximum peak. This in phase movement corresponds to forward progression of both segments and the lower limb. The time of the thigh minimum peak (Table 4.1) tends to coincide with the onset of the H3 power burst (Table 4.2) which occurs earlier as obstacle height increases (See Figure 4.4).

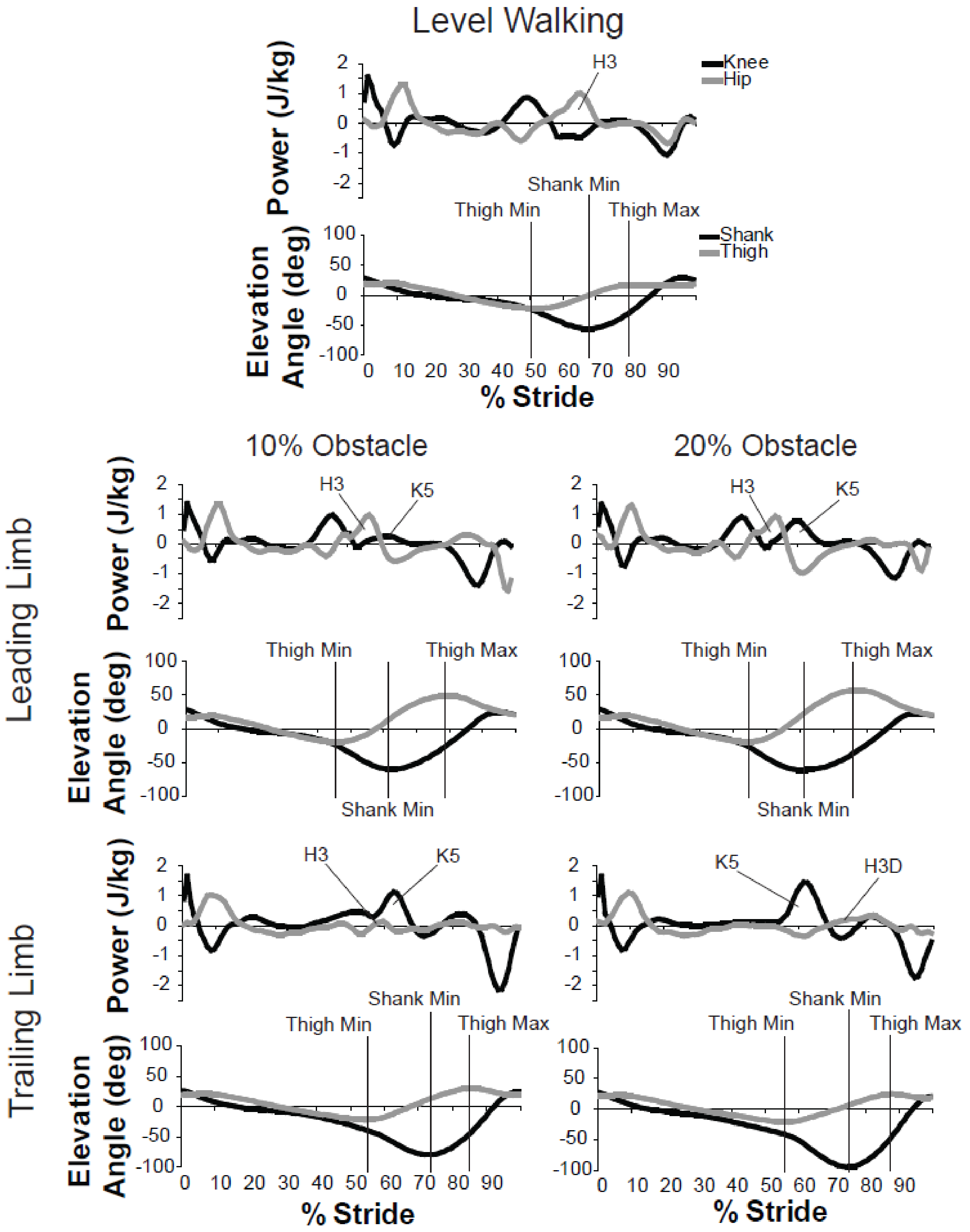

Figure 4.4 Average results from a representative subject relating muscle power bursts (on top) to segment elevation angle movement (below) for level walking (Top middle), and for leading and trailing limbs for the 10% (bottom left) and the 20 % (bottom right) obstacle conditions. Important muscle power bursts (K5, H3, and H3D) are labelled and kinematic events are indicated with vertical bars.

To determine if joint specificity is a control mechanism in the CNS, the work done by the H3 muscle power burst was correlated to step length and thigh elevation range since these variables were chosen to represent progression of the lower limb. As well, the work done by the K5 muscle power burst was correlated to maximum toe height and shank elevation angle range chosen to be representative of lower limb elevation. The plotted relationships of these variables are shown in the top section of Figure 4.5. For the leading limb, Pearson correlations showed significant, but low relationships between H3 work and step length ($r = -0.337$, $p < 0.001$), and H3 work and thigh elevation angle range ($r = 0.204$, $p < 0.001$), although moderate relationships were shown between K5 work and maximum toe height ($r = 0.650$, $p < 0.001$), and K5 work and shank elevation angle range ($r = 0.467$, $p < 0.001$). In the trailing limb, significant correlations were also observed between these variables (H3+H3D work and step length: $r = 0.382$, $p < 0.001$; H3+H3D work and thigh elevation angle range: $r = 0.368$, $p < 0.001$; K5 work and maximum toe height: $r = 0.656$, $p < 0.001$; and K5 work and shank elevation angle range: $r = 0.589$, $p < 0.001$). Although these correlations were significant, the only moderate correlation was observed between work done by the K5 muscle burst and maximum toe height in the leading and trailing limbs.

With regards to temporal variables, correlation analysis suggested a strong relationship between H3 power onset and thigh minimum peak in the leading limb ($r = 0.834$, $p < 0.001$), but this relationship was not as strong in the trailing limb ($r = 0.318$, $p < 0.001$) (lower section of Figure 4.5). When stepping over obstacles, the K5 power burst follows the thigh minimum peak (and the H3 power burst) and continues until the peak shank minimum has been reached. A moderate correlation between K5 power onset and shank minimum peak was shown in the leading limb ($r = 0.601$, $p < 0.001$), although this correlation was much weaker in the trailing limb ($r = 0.109$, $p < 0.033$) (lower section of Figure 4.5). In particular for the trailing limb, an H3D muscle power burst begins prior to the instant of peak minimum shank elevation and continues until slightly before the maximum thigh elevation angle is reached.

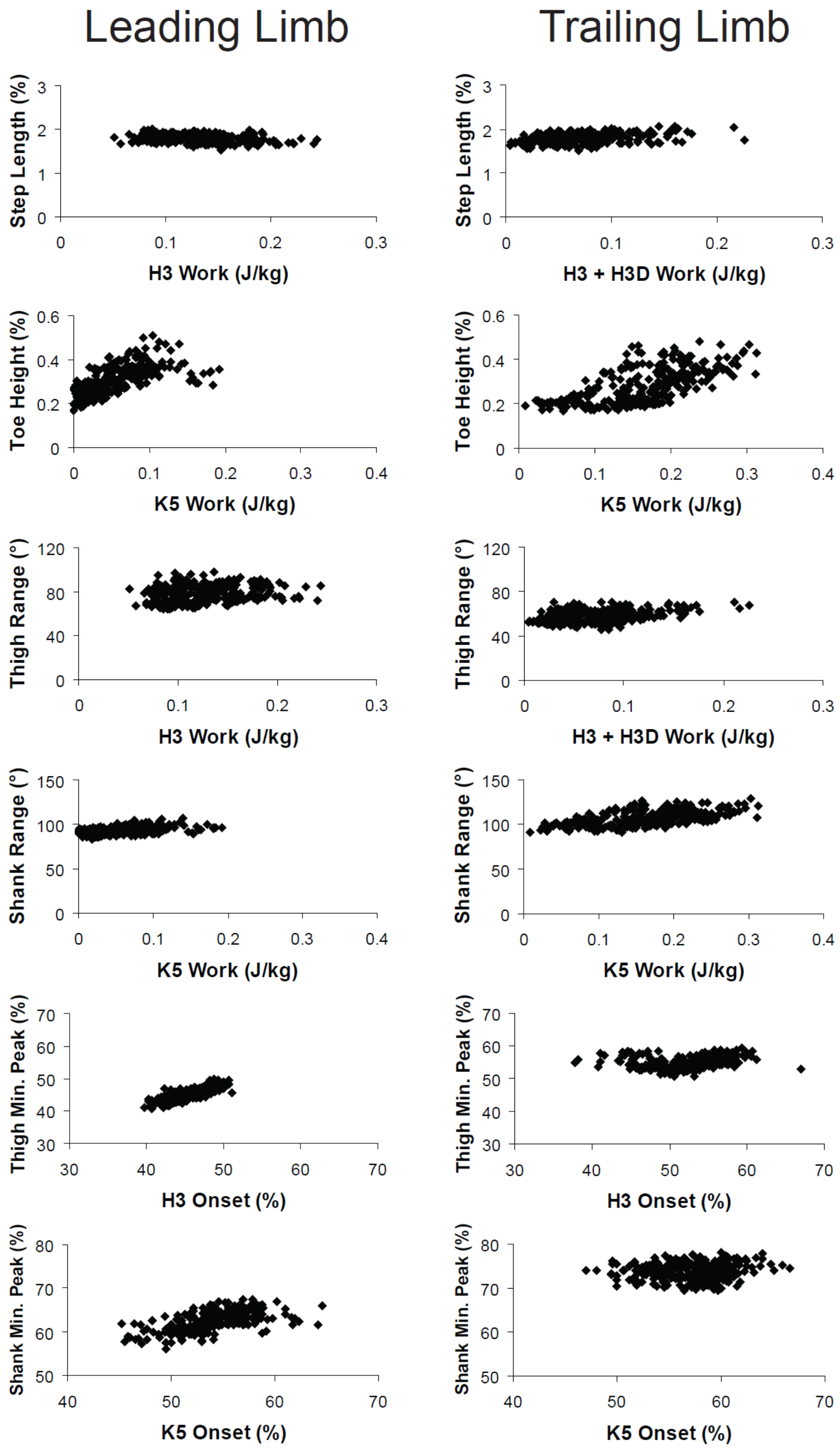

Figure 4.5: Plots of the relationships between kinetic and kinematic data. Spatial parameters are shown in the top eight plots and temporal parameters in the lower four plots.

4.5: Discussion

The current study examined ranges and individual phase shifts of the thigh and shank fundamental harmonics segment elevation angles and related them to the work done by the flexor muscle power bursts at the knee and hip during the transition to swing in order to understand the control underlying lower limb progression and elevation for ALAs. The results showed that in the leading limb, the phase lead of the thigh segment increased, while in the trailing limb, the phase lag of the shank segment increased, with greater obstacle heights. These phase shifts in fundamental harmonics for obstacle clearance resulted in the shifting of elevation angle peaks for the different obstacle heights. As obstacle height increased, the work done by the K5 muscle power burst increased and the H3 muscle power burst decreased while the onsets of the H3 bursts tended to coincide with the minimum peak of the thigh elevation angle. There did not seem to be a strong relationship between the work done by the hip and progression of the lower limb, but a moderate relationship was shown between work done by the knee and elevation of the limb. These results suggest that joint specificity may not be a mechanism of control in the CNS, but the knee joint does play an important role in elevating the limb during obstacle clearance and the hip with initiating forward movement of the thigh. The following text will discuss how these changes are related to CNS control of locomotion.

4.5.1: Thigh-Shank Phase Difference

Bianchi, Angelini, Orani et al. (1998) stated that the phase lead of the shank segment with respect to the foot decreases as walking speed increases. This decrease in phase difference was attributed to the foot having to be swung faster through swing phase to accommodate an earlier heel contact as walking speed increases (Bianchi, Angelini, Orani et al., 1998). More recent work from our laboratory on obstacle avoidance (MacLellan & McFadyen, 2010) has shown that the phase difference of fundamental harmonics between the thigh and shank segment elevation angles increases in both the leading and trailing limbs as higher obstacles are cleared. Building on previous theories raised by Bianchi, Angelini, Orani et al. (1998) and Lacquaniti et al., (1999). MacLellan & McFadyen (2010) suggested that such phase differences between adjacent segment waveforms may be one of the variables controlled by neural oscillators in the CNS during obstacle clearance. The current study focused specifically on individual segmental phase shifting that is related to the phase differences observed previously by MacLellan & McFadyen (2010).

In the leading limb, the thigh increased its phase lead as obstacle height increased and the shank slightly decreased its phase lag, although the changes in thigh phase shift are much greater than those seen in the shank. In the trailing limb, the thigh phase lead was slightly greater in obstacle conditions and the shank increased its phase lag as obstacle height increased, with the changes in shank phase lag being much greater than that of the thigh lead. Therefore, the present study shows that the CNS appears to primarily accomplish the previously reported phase differences between thigh and shank segments by increasing the phase lead of the thigh segment in the leading limb and increasing the phase lag of the shank segment in the trailing limb. These results, therefore, suggest 2 separate CNS strategies for the leading and trailing limbs in order to increase the thigh-shank fundamental harmonic phase difference when clearing obstacles of increasing height. This difference between limbs is most likely related to the need to delay limb swing following the required limb elevation for obstacle clearance in the trailing limb, but not in the leading limb (Niang & McFadyen, 2004). As illustrated in the current study, such a delay appears to be achieved by increasing the phase lag of the shank elevation angle harmonic. These current findings further reinforce the theory that each limb is controlled by a separate neural oscillator (Wannier et al., 2001; Yang et al., 2004).

4.5.2: Integration of Kinematic and Kinetic Results

In order to understand the mechanisms driving human movement, kinematics and kinetics must be integrated together. In particular, muscle powers are of particular importance since they take into account the muscle moment and angular motion of a joint and provide insight into the mechanical cause of movement (Winter, 2005).

At the knee, the K5 muscle power burst increased in the leading and trailing limbs as obstacle height increased. This has been shown in previous studies (McFadyen & Winter, 1991; Niang & McFadyen, 2004). The work done by the K5 muscle power burst was greater in the trailing when compared to the leading limb, which has been suggested with changes in leading limb proximity in McFadyen et al. (1993) and shown in Niang &

McFadyen (2004). As well, the work done by the H3 muscle power burst decreased generally when clearing obstacles in the leading limb and with each increase in obstacle height in the trailing limb, with more work being done by the H3 burst in the leading when compared to the trailing limb. Although these changes in H3 were small, these results differ slightly than those presented by Niang & McFadyen (2004) who showed that work done by the H3 power burst did not significantly differ between level walking and obstacle conditions, but this discrepancy may be due to the non-parametric statistical approach taken by these authors. These differences in work done at the hip and knee muscle groups between obstacle heights can be explained when this information is integrated with the segment phasing results.

When correlating the work done by the knee muscle power burst with maximum toe height (a variables representing limb elevation), moderate relationships were shown in the leading and trailing limbs. This may indicate that the K5 knee muscle power burst probably has a role in elevating the limb, providing quantitative evidence for the suggestions made by McFadyen & Winter (1991) and Niang & McFadyen (2004). Weaker relationships were observed between the work done by the hip muscle power bursts and stride length as well as thigh elevation angle range (variables representing limb progression). Although, the strong correlation between the timings of the H3 muscle power burst onset and thigh elevation angle minimum peak suggests that such hip flexor muscle power may be related to the initiation of thigh movement reversal. When comparing values between Tables 4.1 and 4.2, it can be seen that the thigh segment moves in this positive direction slightly before the H3 muscle power burst initiates. However, given that the onset of the H3 muscle power burst remains synchronized with the change in thigh phasing as obstacle height increases, further supports the H3 muscle power burst as aiding to shift the thigh elevation angle trajectory direction. As well, the K5 muscle power burst occurs around this time to decrease the magnitude of the shank elevation angle peak (leading to greater knee flexion) and also adding energy in order to further flex the hip (McFadyen & Winter, 1991; Patla & Prentice, 1995). The fact that the work done by the H3 muscle power burst was shown to decrease in the obstacle conditions even though the magnitude of the elevation angle range increases with obstacle height (as shown in MacLellan and McFadyen, 2010), further supports the role of H3 muscle power to probably help to reverse thigh rotation and not increase thigh elevation during obstructed walking. The K5 muscle power burst would, for its part, appear to have some direct influence on thigh segment elevation, leading to greater hip flexion as obstacle height increases.

In the trailing limb, the H3 muscle power burst does not seem to be synchronized with the thigh elevation angle minimum peak as seen in the leading limb. As well, the magnitude of H3 work decreases with obstacle height, but this decrease is accompanied with an increase in thigh elevation angle range (MacLellan & McFadyen, 2010). Thus, there must be another source of energy that causes this increase in thigh elevation movement. The work done by the K5 muscle power burst is greater in the trailing when compared to the leading limb. Although shank elevation angle range is increased and thigh elevation angle range is decreased in the trailing limb (as shown in MacLellan & McFadyen, 2010), the greater work done by the K5 muscle power burst in the trailing limb may be a requirement to flex the hip in the presence of much greater decreases in the H3 muscle power burst. Niang & McFadyen (2004) observed the appearance of a delayed H3 power burst (H3D) during obstacle clearance by the trailing limb and concluded that the function of this burst was to progress the limb through swing following trailing foot clearance over the obstacle due to the close proximity of the foot. In the current study, the H3D power burst is initiated during thigh elevation and around the time of K5 offset, as well as after H3 offset. Thus, the function of this delayed power burst may be to provide supplementary energy to continue progression of the thigh to peak maximum elevation following cessation of the K5 power burst. More focused studies will be required to confirm this.

It has been suggested that neural oscillators in the CNS are able to control limb movement during locomotion partially through phase differences in adjacent segment elevation angle waveforms (Das & McCollum, 1988; Lacquaniti et al., 1999; Shen & Poppele, 1995). These ideas have been extended to obstacle clearance in healthy young adults by MacLellan & McFadyen (2010). This current study took a different approach to examine how the underlying muscle kinetics are related to such segment elevation angles to shed light on whether certain joint powers have specific roles related to limb progression and elevation as suggested by Niang & McFadyen (2004). Although relationships were shown between the work done by the K5 muscle power burst and maximum toe height in the leading and trailing limbs, as well as a relationship between H3 onset and thigh movement reversal in the leading limb, this work does not suggest simple joint specificity in CNS control since single functions are not associated with the hip and knee joints. For example, the K5 power burst acts at both the shank and thigh segments (through passive energy transfer as first raised by McFadyen & Winter (1991) and further shown in Patla & Prentice (1995)). Since an increase in elevation of the thigh segment acts to both elevate the lower limb and progress it through swing, it is difficult to assign a single function of the K5 muscle power, although there can be no doubt that it significantly contributes to limb elevation during obstacle clearance. A similar argument can be made for the trailing limb, where the H3 power burst tends to decrease with obstacle height and the K5 muscle power seems to have a greater influence on the thigh segment elevation. Therefore, instead of there being a specific function for each muscle burst, it appears that the work done by the hip and knee muscles coordinate together to perform the obstacle avoidance task successfully.

Since there does not seem to be a single goal for these muscle bursts, this may be further evidence that muscle force is not a control variable in the CNS. Instead, the kinetic patterns seen here allow for the intersegmental phase differences required by the CNS to guide the limb over obstacle with K5 and H3 powers both being contributing to elevation and progression of the limb. Therefore, the phase difference between adjacent segments may be a control variable by the CNS, and the muscle forces, moments, and work done results from the underlying control dynamics. Although these theories are speculative at this time, future work including computer modelling could provide stronger evidence of these control theories.

4.6: Conclusions

The differences in fundamental harmonic phase shifting in the leading and trailing limbs (increase in thigh phase lead in the leading limb, increase in shank phase lag in the trailing limb as obstacle height increases) provides further evidence that the each limb is most likely controlled by separate neural oscillators in the CNS. The work done by the K5 and H3 muscle powers differ in the leading and trailing limbs and appear to provide the energy needed to allow segmental phase changes to occur. Since each of these muscle power bursts function to increase the elevation angle of the thigh segment and the power bursts seem to influence the thigh elevation differently in the leading and trailing limbs, it can be concluded that these muscle powers do not have single roles related to either elevating or progressing the lower limb during locomotion. Instead, these muscle powers result from the control dynamics whereby segment elevation angle phase difference may be one of the variables controlled by the CNS.

4.7: Acknowledgements

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4.8: Bridging Paragraph

Until this point, intersegmental coordination during locomotion has only been examined in healthy individuals. These studies suggest that when stepping over obstacles, an adjustment is made to a basic locomotor pattern which may be facilitated through segment elevation angle waveform shifts in the CNS. It is also suggested that the observed kinetic patterns are the result of this waveform control and not controlled directly. An investigation of locomotor coordination in people with neurological disorders may provide further insight into CNS structures that may be involved in such waveform control and suggest the mechanisms of coordination deficit in such disorders. One such disorder that has been associated with coordination deficit is cerebellar ataxia. The following study will examine locomotor coordination deficits in participants with ARCA-1.

Chapter 5: Study 3

"Increased obstacle clearance distance in people with ARCA-1 results in part from coordination changes between the thigh and shank segments"

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Submitted to "The Cerebellum"

5.1: Abstract

Obstacle clearance can be a hazardous locomotor task if not coordinated with the utmost accuracy. The current study used the planar law of intersegmental coordination to determine how leading limb segment coordination during obstacle clearance is affected by cerebellar ataxia. Eight participants with ARCA-1, caused by mutations in the SYNE-1 gene, and eight healthy adults stepped over obstacles. Healthy adults walked at natural speeds as well as a velocity similar to the participants with cerebellar ataxia, resulting in 3 groups (healthy [H], matched velocity [MV], and cerebellar ataxia [CA]). Elevation angles of the foot, shank, and thigh in the sagittal plane were calculated. A principal component analysis was applied to limb segment trajectories and a Fourier harmonic series was further used to determine temporal phase differences between adjacent segments. Although obstacle clearance was greater in the CA group, the planar nature of the three dimensional covariance plot of segment elevation angles, the covariance loop width, and orientation did not differ between the CA, H and MV groups, suggesting that the planar patterns between elevation angles may not be heavily influenced by the cerebellum. Further analysis led to the observation of a non-linear relationship between covariance loop width and thigh-shank fundamental harmonic phase difference and a decrease in covariance loop width was observed when the fundamental harmonic phase difference between the thigh and shank

segments is greater than 90 degrees. This study supports previous work that a greater safety margin is used in people with cerebellar ataxia when stepping over obstacles, but provides a mechanism of CNS control to facilitate this increase in toe clearance.

5.2: Introduction

The cerebellum has a large influence on movement and dysfunction of this structure can have serious effects on motor coordination. Cerebellar ataxia is one such disorder that can affect the cerebellum. Mariotti et al. (2005) defined cerebellar ataxia as a neurological dysfunction of motor coordination which can affect gaze, speech, gait, and balance. Since successful obstacle clearance during gait is dependent on careful coordination of the lower limb segments, this task may be hazardous to such a population. Morton et al. (2004) studied movements at individual joints during obstacle clearance in such a population. The current study will apply multi-segment coordination analyses to healthy adults as well as adults with cerebellar ataxia in order to determine how cerebellar ataxia affects the coordination of obstacle clearance.

Some studies have shown that lower limb joint angles during locomotion do not differ in people with cerebellar ataxia as compared to healthy adults (Earhart & Bastian, 2001; Stolze et al., 2002). Other studies, however, have shown decreases in ankle range of motion (Palliyath et al., 1998) as well as decreased peak ankle plantarflexion, knee flexion, and hip flexion (Morton & Bastian, 2003) during level locomotion in participants with cerebellar ataxia. Palliyath et al. (1998) have also shown increased variability in ankle range of motion, ankle angle at heel strike, foot clearance during swing, knee range of motion during stance, and time to peak knee flexion in swing in this population. When people with cerebellar ataxia walked on an inclined surface, Earhart & Bastian (2001) have shown greater variability of the knee and hip joint angle trajectories when compared to healthy adults. Since obstacle clearance is accomplished by increasing the flexion angles at the hip and knee (McFadyen & Winter, 1991; Patla et al., 1991), the above mentioned variability and decreases in peak joint angles during locomotion would be detrimental to the clearance of obstacles in persons with cerebellar ataxia.

Obstacle clearance in people with cerebellar ataxia has been examined previously by Morton et al. (2004). In their study, an obstacle was placed to one side of the participants and what appears to be the equivalent of the leading limb (limb further from the obstacle first to clear) cleared the obstacle. When participants with cerebellar ataxia stepped over the obstacle, increases in peak knee flexion were observed which lead to hypermetria (overshooting of toe clearance while clearing the obstacle). Further analysis by this group concluded that the observed hypermetria was a voluntary compensation by the participants with cerebellar ataxia in order to increase the margin of safety when stepping over the obstacle (Morton et al., 2004). Although these authors provided a thorough analysis of single joint kinematic and kinetic data, the coordination between lower limb segments was not provided, although this topic has been suggested to be of importance by Morton & Bastian (2007). As well, only a single obstacle height was used (0.05 m high x 0.10 m deep), which is much smaller than the obstacles that one may encounter on a day to day basis.

Previously, coordination of gait has been examined in people with cerebellar ataxia using angle-angle plots. Such plots have identified a decomposition of multi-joint movement into single joint movements and several changes in joint motion due to multiple reversals between adjacent joints (Earhart & Bastian, 2001; Morton & Bastian, 2003). Such angle-angle plots have shown greater variability in joint coordination when compared to healthy adults (Ilg et al., 2007; Stolze et al., 2002). In particular, Ilg et al. (2007) have shown increased temporal variability of all angle-angle pairs during locomotion which separated a group of adults with cerebellar ataxia from groups of adults with other motor coordination deficits.

Most coordination analyses noted above have involved examining movement at two joints at a time. An analysis technique, termed the planar law of intersegmental coordination, allows for the examination of movement coordination at the thigh, shank, and foot segments simultaneously. This law states that if segment elevation angles for a lower limb are plotted in 3-dimensional space for a gait cycle, they tend to form a teardrop shaped loop which orients on a plane in that space. This has been observed in level walking (Bianchi, Angelini, & Lacquaniti, 1998; Bianchi, Angelini, Orani et al., 1998; Borghese et al., 1996) and various other modes of locomotion (Cappellini et al., 2010; Courtine & Schieppati, 2004; Grasso et al., 1998; Grasso et al., 2004; Grasso et al., 2000; Ivanenko et al., 2008; Noble & Prentice, 2008). As well, this pattern has been shown to persist during locomotion in adults with motor control impairments such as Parkinson's Disease (Grasso et al., 1999), spastic paraparesis (Dan et al., 2000), spinal cord injury (Grasso et al., 2004), and forefoot rheumatoid arthritis (Laroche et al., 2007). The significance of this planar pattern is that it suggests the central nervous system (CNS) decreases the available degrees of freedom, therefore simplifying locomotor control (Bianchi, Angelini, Orani et al., 1998; Borghese et al., 1996; Lacquaniti et al., 1999). As well, characteristics of this plane have been shown to be highly correlated to phasing differences between adjacent lower limb segments suggesting that this phase difference may be a variable of control by neural oscillators in the CNS (Bianchi, Angelini, Orani et al., 1998; Courtine & Schieppati, 2004; Lacquaniti et al., 1999). This law has also been used to examine obstacle clearance in healthy adults (Ivanenko et al., 2005; MacLellan & McFadyen, 2010) and led MacLellan & McFadyen (2010) to conclude that obstacle clearance is controlled by similar segment phasing mechanisms as seen in level walking. Applying this analysis technique to participants with cerebellar ataxia may highlight specific coordination dysfunction and inform about changes in locomotor control in this population.

Many previous studies examining locomotion in people with cerebellar ataxia (Earhart & Bastian, 2001; Ilg et al., 2007; Morton & Bastian, 2003; Morton et al., 2004; Palliyath et al., 1998) have recruited participants with various forms of this disorder from numerous origins (due to tumour removal, stroke, olivopontocerebellar atrophy, pancerebellar atrophy, spinocerebellar ataxia, or cerebellar cortical atrophy). Dupre et al. (2007) have identified a patient population with a pure cerebellar ataxia that has diffuse atrophy throughout the cerebellum with no additional abnormalities in the central or peripheral nervous systems. This form of ataxia has been termed autosomal recessive cerebellar ataxia type 1 (ARCA-1) (or recessive ataxia of Beauce), which is caused by mutations in the SYNE-1 gene. This is an inherited cerebellar ataxia with an onset at middle age progressing slowly and leading to a moderate level of disability (Dupre et al., 2007; Gros-Louis et al., 2007). In order to better understand the cerebellar contributions to locomotor coordination, the cerebellar ataxia participants recruited in this study were all identified as having ARCA-1 with confirmed mutations in the SYNE-1 gene.

Obstacle clearance can be a hazardous task if not coordinated with the utmost accuracy. One of the hallmarks in people with cerebellar ataxia is a lack of coordination during movement. The current study will determine how a deficit such as cerebellar ataxia affects leading limb segment coordination during clearance over obstacles varying in height as well as suggest specific locomotor control mechanisms for this population. This will be examined in a population with ARCA-1 as well as in a healthy population with segment coordination being measured using techniques related to the planar law of intersegmental coordination.

5.3: Methods

Two groups of participants volunteered for this study. The first group consisted of eight participants (4 females and 4 males) with ARCA-1 (aged 43.3 +/- 12.8 years, 1.70 +/- 0.09 m in height, and 80.9 +/- 17.4 kg in mass). The second group of participants were eight healthy adults (4 females and 4 males) selected to match the average age of the ARCA-1 group (aged 42.8 +/- 11.1 years, 1.70 +/- 0.07 m in height, and 68.6 +/- 14.1 kg in mass). All participants provided informed consent in accordance with ethics guidelines from the Quebec Rehabilitation Institute and Laval University.

5.3.1: Protocol

Participants were asked to step over obstacles for which obstacle height (0, 10, 20% of leg length) and obstacle depth (0, 10, 20% of step length) were adjusted, for a total of 9 obstacle conditions. Since obstacles always have some depth, the 0% depth condition was actually 0.025 m, which corresponded to the minimum depth possible for the obstacle used in this study. In the current study, only 3 obstacle conditions will be presented where the leading limb stepped over obstacles with heights of 0, 10, and 20% and a depth of 0.025 m. The 0% height obstacle condition consisted of level walking with no obstacle present. Each obstacle condition was presented in a block of 3 trials, totalling 27 trials presented randomly. For the group of healthy adults, the set of 27 trials was performed twice; once at comfortable velocity and a second time at a slower velocity approximating the average walking velocity of the cerebellar ataxia participants. Participants began each trial 3-4 steps ahead of the obstacle and were not asked to lead with a specific limb. This resulted in 4 of the 8 participants with cerebellar ataxia and all 8 healthy adults leading with the right limb. Participants with cerebellar ataxia were spotted by a physiotherapist for the experimental trials.

Full body 3D kinematic data were collected at 75 Hz using a 3 bar Optotrak 3020 camera system (Northern Digital Inc., Waterloo, Canada). Segment movement was recorded using infrared emitting diodes (IREDS) attached in non-collinear triads to plastic plates which were then fixed to the feet, shanks, thighs, pelvis, trunk, and head segments. A calibration trial was then collected and anatomical landmarks (5th metatarsal, medial/lateral malleolus, medial/lateral femoral condyles, left/right iliac spine, and left/right anterior superior iliac spine) were digitized to estimate centre of mass (COM) positions as well as virtual points for the heels and toes of each foot in order to determine their trajectories. All 3D kinematic data was filtered offline using a 2nd order dual-pass Butterworth filter with a cut-off frequency of 6 Hz.

5.3.2: Data Analysis

Kinematic data were used to create a 9 linked segment biomechanical model which included the feet, shanks, thighs, pelvis, trunk, and head. From this model, clearance distance (vertical distance from the top of the obstacle to the toe at clearance) and full body COM velocity were calculated. Lower limb segment elevation angles (absolute angle with the vertical) of the feet, shanks, and thighs were calculated for the stride over the obstacle (heel contact to subsequent heel contact) from the collected digitized anatomical points as described in Borghese et al. (1996). From each segment angular trajectory, the elevation angle range over the stride was determined by subtracting the maximum from the minimum segment angle.

Inter-joint coordination was assessed using the planar law of intersegmental coordination (Bianchi, Angelini, & Lacquaniti, 1998; Bianchi, Angelini, Orani et al., 1998; Borghese et al., 1996; Courtine & Schieppati, 2004; Ivanenko et al., 2005, 2007; Ivanenko et al., 2008).

Thigh, shank, and foot elevation angle trajectories for the leading limb stride (heel contact to heel contact; from before to after clearance in obstacle conditions) were normalized to 100% of stride time with the mean being subtracted from each trajectory. A three dimensional covariance plot consisted of these three normalized segment elevation angles forming a loop. A principal component analysis (PCA) was then applied to the group of 3 segment elevation angle trajectories to determine covariance loop planarity, width, and orientation. In order to quantify planarity, the percent variances of the first 2 principal components (PCs) were summed together. If the segment elevation angles were to orient perfectly on a plane, the percent variance explained by the first 2 PCs would be 100%. Covariance loop width was determined using the percent variance explained by the 2nd PC, since the axis of this PC is oriented in the direction of the minor axis of the loop formed by the elevation angles. Covariance plane orientation was quantified using the direction cosine between the 3rd principal axis and the positive semi-axis of the thigh segment. Finally, phase differences between adjacent segments (thigh-shank and shank-foot), was quantified by applying a Fourier series using 10 harmonics to represent the time course of each segment elevation angle trajectory by the fundamental harmonic. The phase difference between segments was calculated by subtracting phasing of the proximal segment fundamental harmonic from that of the distal segment fundamental harmonic.

5.3.3: Statistical Analyses

In the current study, statistical analyses were performed within and between each group. Within each group, significant differences between obstacle height conditions (0, 10, 20%) for COM velocity, elevation angle ranges, planarity, covariance loop width, covariance plane orientation, and fundamental harmonic phase differences between segments were determined using a Friedman test. If the Friedman test was found to be significant, a Wilcoxon test was used to determine significant differences between conditions. A Wilcoxon test was also used to determine significant differences between clearance distance in the 10 and 20% obstacle height conditions. Between group differences (healthy [H], matched velocity [MV], cerebellar ataxia [CA]) for all variables was determined using a Kruskal-Wallis test. If the Kruskal-Wallis test was found to be significant, a Mann-Whitney U test was used to determine significant differences between pairs of groups. Statistically significant differences were determined at a p-value of 0.05 or less.

5.4: Results

As CA participants stepped over obstacles, visual observation of the loops formed by the segment elevation angles suggest that segment intra-participant kinematic patterns become increasingly variable (see data of a representative participant in Figure 5.1; top half). This in turn affects the intra-participant variability in width of the covariance loop and orientation of the covariance plane, although intra-participant variability is not a focus in the current study. As well, shapes of the covariance loops for the CA group differ from those of H and MV groups (Figure 5.1; bottom half). The following will detail the differences between obstacle height conditions for each group first, and then make comparisons between groups to determine how coordination during obstacle clearance is affected by cerebellar ataxia.

5.4.1: Healthy Adult Group

The healthy adult group showed no significant differences between obstacle height conditions for clearance distance ($p = 0.320$, Table 5.1). Average full body COM velocity did show a main effect for obstacle height ($p = 0.010$) and further tests showed that this velocity was greater during level walking when compared to obstacle clearance ($p = 0.027$, Table 5.1). Examination of individual segment movement suggested segment elevation angle range (Table 5.1) showed main effects for the shank ($p < 0.001$) and thigh ($p < 0.001$) segments and further analysis showed that elevation angle range increased with obstacle height ($p = 0.004$ for each segment). No significant differences were observed for foot elevation angle range for the different obstacle conditions ($p = 0.355$).

As seen previously (MacLellan & McFadyen, 2010), planarity decreased slightly when healthy adults stepped over obstacles (Figure 5.2). A main effect for obstacle height was shown ($p = 0.001$) and Wilcoxon tests showed that planarity decreased when stepping over the obstacle ($p = 0.004$). As well, the width of the covariance loop increased ($p < 0.001$) and this increase was significant ($p = 0.004$ for each obstacle height condition, Figure 5.2). The orientation of the covariance plane showed a main effect ($p < 0.001$) and further analysis showed that the plane was oriented more counterclockwise when stepping over obstacles as compared to level walking ($p = 0.004$, Figure 5.2).

Table 5.1: Comparisons of kinematic variables between obstacle heights and participant groups.

	Healthy			Matched Velocity			Cerebellar Ataxia		
	Level	10%	20%	Level	10%	20%	Level	10%	20%
Clearance (m)									
Mean	-	0.107	0.104	-	0.100	0.089	-	0.161[C,D]	0.184[b,C,D]
CI	-	0.083 - 0.130	0.079 – 0.128	-	0.065 – 0.135	0.072 – 0.106	-	0.133 – 0.190	0.166 – 0.202
COM Velocity (m/s)									
Mean	1.35	1.28[a]	1.25[a]	0.77[C]	0.75[C]	0.74[C]	0.94[c,D]	0.73[a,C]	0.72[a,C]
CI	1.22 – 1.49	1.17 – 1.38	1.12 – 1.38	0.67 – 0.86	0.70 – 0.81	0.66 – 0.82	0.66 – 1.22	0.40 – 1.06	0.45 – 1.00
Foot Range (°)									
Mean	100.7	104.0	103.8	87.4[C]	91.3[C]	92.3[C]	58.1[C,D]	57.2[C,D]	60.6[C,D]
CI	94.5 – 106.9	100.3 – 107.6	99.3 – 108.3	81.3 – 93.5	84.0 – 98.6	85.9 – 98.8	32.9 – 83.4	32.7 – 81.7	36.7 – 84.4
Shank Range (°)									
Mean	84.2	88.5[a]	91.1[a,b]	74.5[C]	79.5[a,C]	81.0[a,b,C]	65.1[C]	85.3[a]	97.8[a,b,D]
CI	81.2 – 87.2	85.6 – 91.4	87.7 – 94.5	70.5 – 78.5	74.3 – 84.8	76.4 – 85.6	54.7 – 75.6	73.7 – 96.8	86.5 – 109.2
Thigh Range (°)									
Mean	44.4	68.7[a]	80.0[a,b]	36.5[C]	65.5[a]	77.1[a,b]	40.7[D]	67.4[a]	79.2[a,b]
CI	41.2 – 47.6	66.3 – 71.1	77.4 – 82.7	34.2 – 38.8	61.0 – 69.9	72.0 – 82.2	35.4 – 46.0	62.2 – 72.7	69.4 – 88.9

[a] – significantly different from level walking
[b] – significantly different from 10% obstacle
[C] – significantly different from healthy
[D] – significantly different from matched velocity

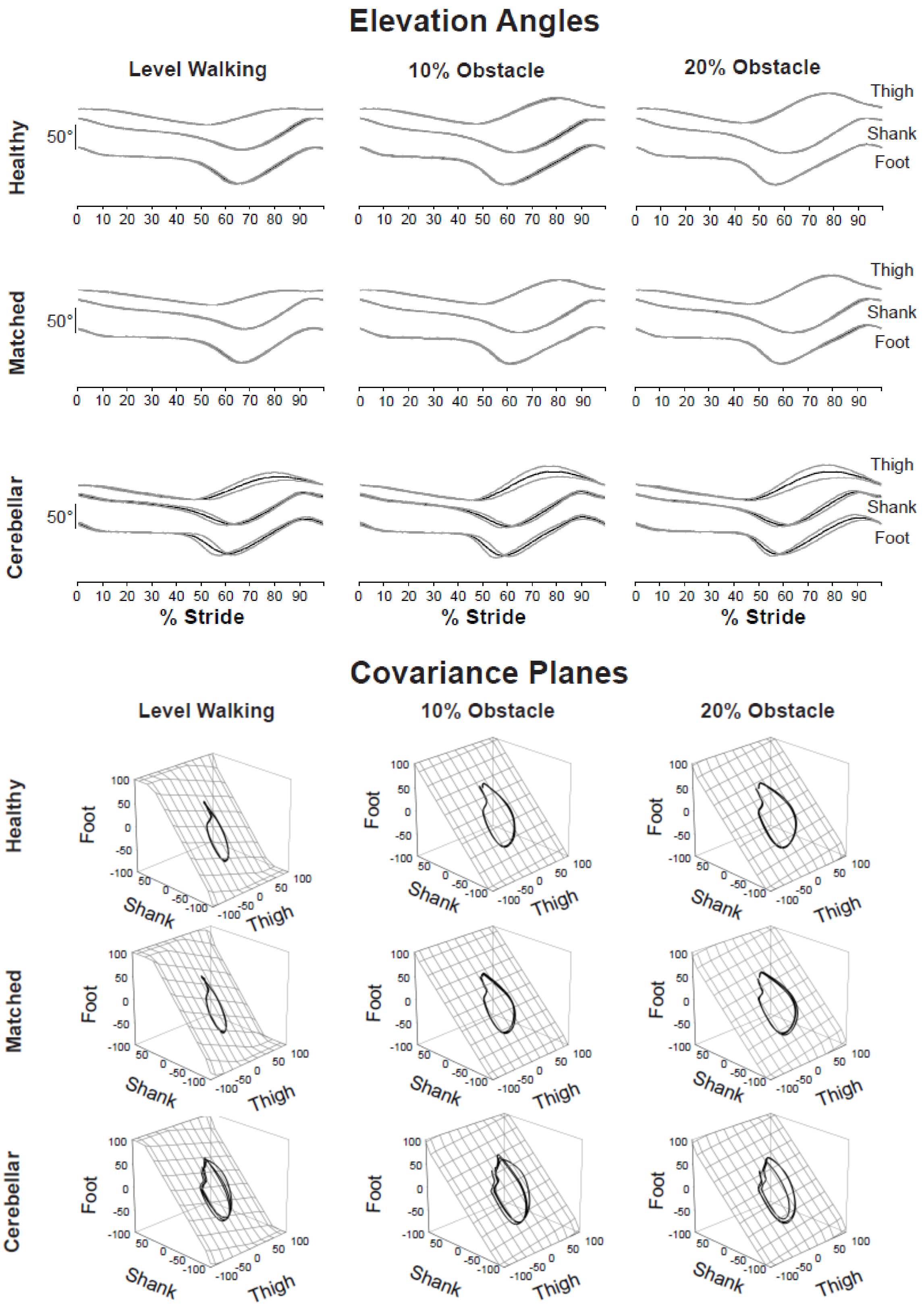

Figure 5.1: Segment elevation angles (top half) and segment covariance patterns (bottom half) for representative participants in each group.

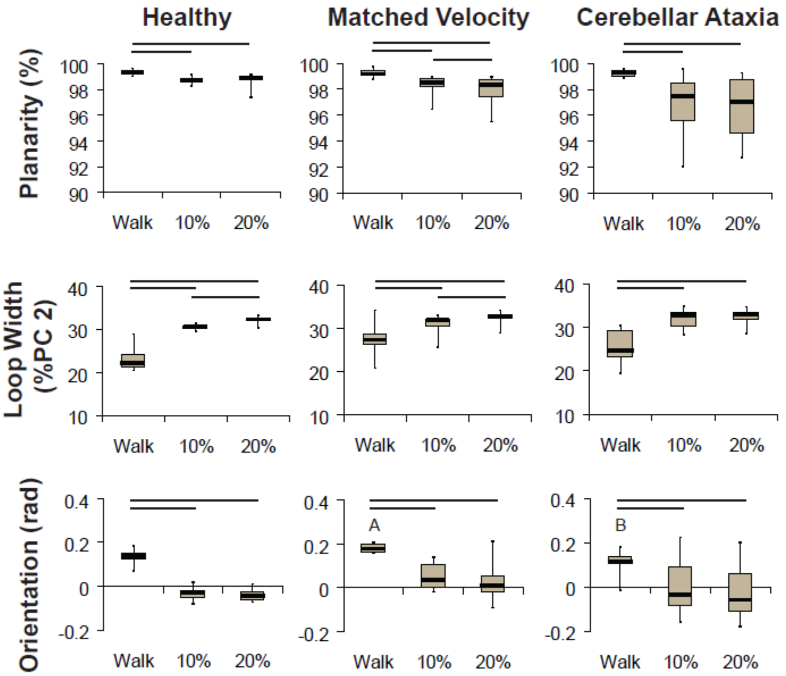

Figure 5.2: Box plots of planarity (top), covariance loop width (middle), and covariance plane orientation (bottom) for participant groups. Significant differences between obstacle conditions are marked with horizontal bars. Significant differences between groups are marked with an 'A' (significantly different from healthy group) and 'B' (significantly different from matched velocity group).

Fundamental harmonic phase differences between segments followed similar patterns to what has been documented previously (MacLellan & McFadyen, 2010) during obstacle clearance. The fundamental harmonic phase difference between the thigh and shank segments showed a main effect for obstacle height ($p < 0.001$) and Wilcoxon tests showed that this phase difference increased with obstacle height ($p = 0.004$, Figure 5.3). A main effect was also observed in the fundamental harmonic phase difference between the shank and foot segments ($p = 0.001$) and further analysis showed that this phase difference was greater in level walking when compared to obstacle clearance ($p = 0.004$, Figure 5.3). Furthermore, the estimated relationship between the thigh-shank fundamental harmonic phase difference and covariance loop width, as well as shank-foot fundamental harmonic phase difference and covariance plane orientation tended to be linear (Figure 5.4).

5.4.2: Matched Velocity Group

When healthy adults walked at a velocity that was matched to the cerebellar ataxia group, the different obstacle heights did not affect clearance distance ($p = 0.191$) or full body COM velocity ($p = 0.067$), although segment range of motion differed significantly between obstacle conditions (Table 5.1). Shank elevation angle range showed a main effect for obstacle height ($p < 0.002$) and this range increased with obstacle height ($p = 0.027$). Similarly, thigh elevation angle range showed a main effect ($p < 0.001$) and also increased with obstacle height ($p = 0.004$). No significant differences between obstacle heights were shown for foot elevation angle range ($p = 0.079$).

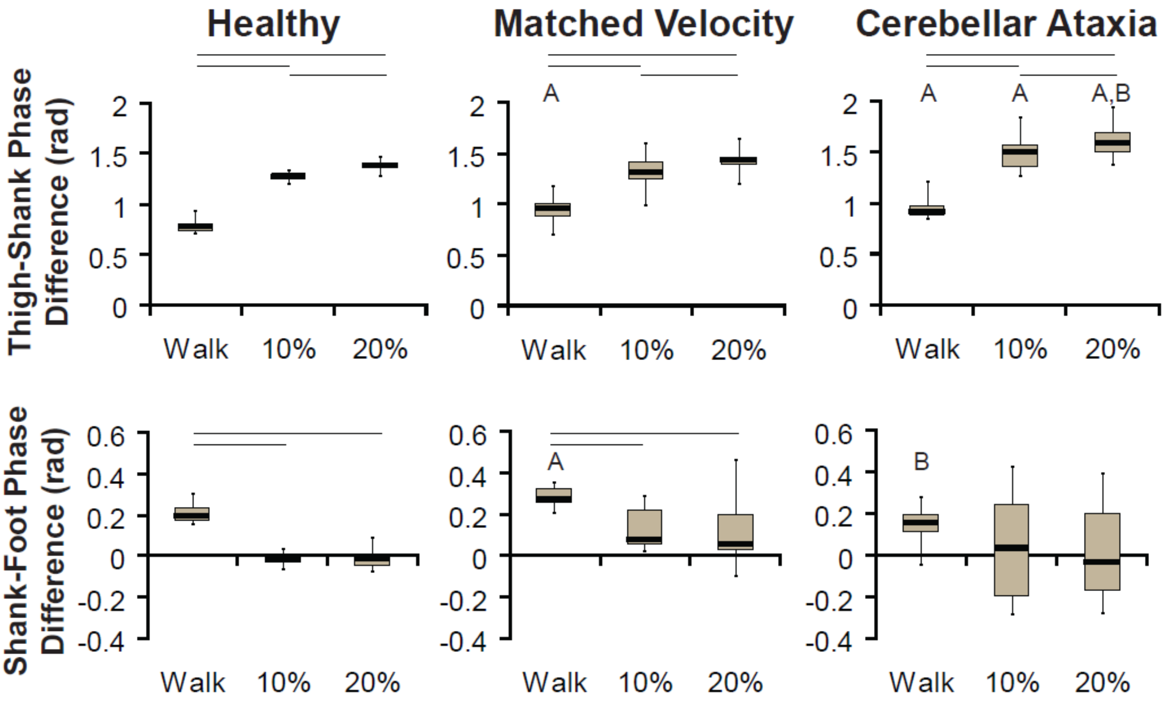

Figure 5.3: Box plots of thigh-shank fundamental harmonic phase difference (top half) and shank-foot fundamental harmonic phase difference (bottom half) for participant groups. Significant differences between obstacle conditions are marked with horizontal bars. Significant differences between groups are marked with an 'A' (significantly different from healthy group) and 'B' (significantly different from matched velocity group).

As seen for comfortable walking velocity, planarity (main effect: $p < 0.001$) tended to decrease during obstacle clearance when compared to level walking ($p = 0.020$, Figure 5.2). The width of the covariance loop showed a main effect for obstacle height ($p = 0.005$) and Wilcoxon tests showed that the width of the covariance loop increased with increasing obstacle height ($p = 0.039$, Figure 5.2). Also, a main effect was shown for covariance plane orientation (p = 0.005) and further tests showed that the covariance plane rotated more counterclockwise as obstacle height increased (p = 0.008, Figure 5.2).

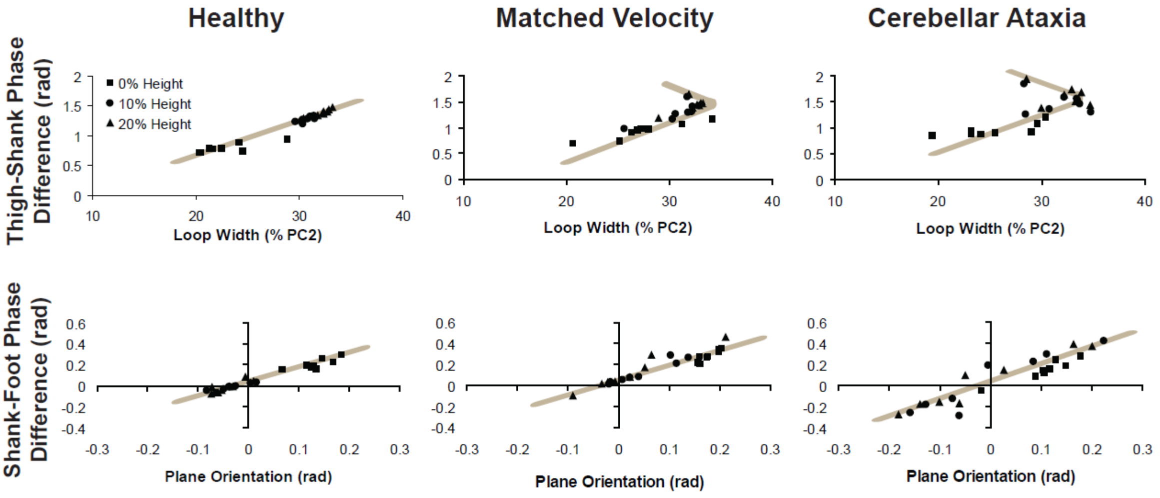

Figure 5.4: Plots containing the relationships between the thigh-shank fundamental harmonic and covariance loop width (top half) and shank-foot fundamental harmonic phase difference and covariance plane orientation. Thick grey lines indicate the estimated linear tendency between variables.

Examination of fundamental harmonic phase differences showed the difference between the thigh and shank segments had a main effect for obstacle height ($p < 0.001$) and further tests showed that this phase difference increased with obstacle height (p = 0.004, Figure 5.3). A significant main effect was also observed between the shank and foot fundamental harmonic phase difference (p = 0.002) and Wilcoxon tests showed that this phase difference was greater in level walking when compared to obstacle clearance (p = 0.020, Figure 5.3). Examination of the estimated relationship between thigh-shank fundamental harmonic phase difference and covariance loop width showed a slightly different pattern than that observed in the H group (Figure 5.4). This pattern is no longer linear as seen when the pattern is estimated at comfortable velocity, but tends to reach a maximum in covariance loop width as thigh-shank fundamental harmonic phase difference increased. The estimated relationship between shank-foot fundamental harmonic phase difference and covariance plane orientation was estimated to follow a similar linear pattern to that seen in the H group (Figure 5.4).

5.4.3: Cerebellar Ataxia Group

When participants with cerebellar ataxia stepped over the obstacle, clearance distance was significantly greater in the 20% condition when compared to the 10% obstacle height condition ($p = 0.012$, Table 5.1). As well, average full body COM velocity when stepping over the obstacle showed a main effect ($p < 0.001$) and Wilcoxon tests showed that COM velocity was greater for level walking when compared to obstacle clearance ($p = 0.004$, Table 5.1). Although foot elevation angle range did not change for differing obstacle height conditions ($p = 0.236$), shank ($p < 0.001$) and thigh ($p < 0.001$) elevation angle ranges showed main effects for obstacle height (Table 5.1). Wilcoxon tests showed that each of these elevation angles increased with obstacle height ($p = 0.004$).

In terms of intersegmental coordination analysis on the leading limb, the CA group generally displayed a planar pattern of lower limb elevation angles, but a main effect was shown ($p = 0.001$). Wilcoxon tests showed that planarity was greater in level walking when compared to obstacle clearance ($p = 0.004$, Figure 5.2). As well, the width of the covariance loop ($p = 0.001$) and the orientation of the covariance plane ($p = 0.047$) changed with obstacle height (Figure 5.2). Further testing showed that covariance loop width was smaller in level walking when compared to obstacle clearance ($p = 0.004$) and that the covariance plane was rotated in a counterclockwise direction in the obstacle clearance conditions when compared to level walking ($p = 0.020$).

Fundamental harmonic phase difference between the thigh and the shank segments showed a main effect ($p < 0.001$) and Wilcoxon tests showed that this phase difference increased with each increase in obstacle height ($p = 0.004$, Figure 5.3). No significant differences were shown for the fundamental harmonic phase difference between the shank and foot segments ($p = 0.531$, Figure 5.3). As seen in the matched velocity group, the estimated relationship between the thigh-shank fundamental harmonic phase difference and the width of the covariance loop takes on a curved shape (Figure 5.4). This curved shape is much more present than in the MV group. Again, the estimated relationship between the shank-foot fundamental harmonic phase difference and orientation of the covariance plane create a linear pattern.

5.4.4: Comparisons between Participant Groups

Group effects were observed for clearance distance in the 10% ($p = 0.006$) and 20% ($p < 0.001$) obstacle height conditions (Table 5.1). Mann-Whitney U tests showed clearance distance in the CA group was greater than the H (10% - $p = 0.002$, 20% - $p < 0.001$) and MV (10% - $p = 0.005$, 20% $p < 0.001$) groups. No significant differences were observed between the H and MV groups for the 10% ($p = 0.442$) and 20% ($p = 0.156$) obstacle height conditions.

As expected, COM velocity showed main group effects for level walking and each obstacle condition ($p < 0.001$, Table 5.1). During level walking, Mann-Whitney U tests showed that COM velocity was greater in the H group when compared to the CA ($p = 0.004$) and MV ($p < 0.001$) groups. In each of the obstacle conditions, COM velocity was greater in the H group when compared to the MV ($p < 0.001$) and CA ($p < 0.001$) groups.

Foot elevation angle range showed group effects for level walking ($p < 0.001$), the 10% ($p < 0.001$) and 20% obstacle height ($p < 0.001$) conditions (Table 5.1). During level walking, Mann-Whitney U tests showed that foot elevation angle range was greater in the H group when compared to the MV ($p < 0.001$) and CA ($p < 0.001$) groups as well as for the MV as compared to the CA group ($p = 0.032$). For the 10% and 20% obstacle height conditions, foot elevation angle range was again greater in the H group when compared to the MV ($p < 0.001$) and CA ($p < 0.001$) groups as well as greater in the MV when compared to the CA group (10%: $p = 0.034$, 20%: $p = 0.016$). During level walking, shank elevation angle range showed a main effect ($p < 0.001$) where the observed range was greater in the H group when compared to the MV ($p < 0.001$) and CA ($p = 0.001$) groups (Table 5.1). For the 10% obstacle height condition, a main group effect was shown ($p = 0.030$) and Mann-Whitney U tests showed that shank elevation angle range in the H group was greater than the MV group ($p = 0.002$). A main group effect was also shown for the 20% obstacle height condition ($p = 0.001$) where shank elevation angle range was significantly lower in the MV group when compared to the H ($p = 0.001$) and CA ($p = 0.001$) groups. Thigh elevation angle range (Table 5.1) showed a main group effect for level walking ($p = 0.005$) and Mann-Whitney U tests showed that this elevation angle range was lower in the MV group when compared to the H ($p < 0.001$) and CA ($p = 0.043$) groups. No significant group effects were observed for the 10% ($p = 0.442$) and 20% ($p = 0.538$) obstacle height conditions for thigh elevation. Thus, group differences between the CA and the other groups were mostly for foot elevation ranges and level walking.

Planarity (Figure 5.2) was similar between groups for level walking ($p = 0.738$), the 10% ($p = 0.152$) and the 20% obstacle height condition ($p = 0.083$). Covariance loop width (Figure 5.2) did not differ between groups during level walking ($p = 0.061$), the 10% ($p = 0.238$) or the 20% (0.606) obstacle conditions. A group effect was shown during level walking ($p = 0.001$) where the covariance plane was rotated more clockwise in the MV group when compared to the H ($p = 0.001$) and CA ($p = 0.006$) groups. No significant differences between groups were shown for covariance plane orientation in the 10% ($p = 0.051$) and 20% ($p = 0.242$) obstacle height conditions.

The fundamental harmonic phase difference between the thigh and shank segments showed group effects for level walking ($p = 0.030$), the 10% ($p = 0.035$) and 20% ($p = 0.008$) obstacle height conditions. During level walking, the phase difference between the thigh and shank fundamental harmonics was lower in the H group when compared to the MV ($p = 0.036$) and CA (0.003) groups. For the 10% obstacle height condition, the fundamental harmonic phase difference between the thigh and shank segments was greater in the CA group when compared to the H group ($p = 0.003$). As well, this phase difference was greater in the CA group when compared to the H ($p = 0.002$) and MV ($p = 0.034$) groups for the 20% obstacle height condition. The fundamental harmonic phase difference between the shank and foot segments showed significant group effects for level walking ($p = 0.002$, Figure 5.3) where the phase difference was greater in the MV group when compared to the H ($p = 0.007$) and CA ($p = 0.002$) groups. No significant differences between groups for the shank-foot fundamental harmonic phase difference were observed for the 10% ($p = 0.067$) and 20% ($p = 0.301$) obstacle height conditions.

5.5: Discussion

In the current study, healthy participants and participants with ARCA-1 were asked to step over obstacles. This study expanded upon the results by Morton et al. (2004) by using obstacles differing in size, recruiting a population with pure form of ataxia, and by using an analysis to study intersegmental coordination as suggested by Morton & Bastian (2007). In general, the results of the study showed that the participants with cerebellar ataxia had a greater clearance distance over the obstacles than the healthy population as seen in Morton et al. (2004). The analysis of coordination showed that lower limb segment elevation angles when plotted together tended, in the CA group, to orient on a plane. However, changes in the relationship between covariance plane characteristics and fundamental harmonic phase differences between adjacent segments tended to deviate from normal patterns. The following will discuss coordination differences between the groups and suggest why these pattern deviations were observed.

5.5.1: Kinematics of level walking and obstacle clearance

Morton et al. (2004) have previously shown that hypermetria (excessive foot elevation) occurs when people with cerebellar ataxia step over a 0.05 m high x 0.10 m deep obstacle with a single limb. It was concluded that the observed hypermetria is a voluntary strategy in people with cerebellar ataxia to increase the margin of safety when stepping over obstacles since the patterns of movement were similar to that of healthy adults stepping over a higher obstacle. The current study differed from Morton et al. (2004) in that two higher obstacle heights were used, the healthy adults walked at a comfortable velocity as well as the a matched velocity, and the pathological group in the present study had purer form of cerebellar ataxia and walked slower on average. Results from this study showed that clearance distance remained relatively constant over the different obstacle conditions in the H and MV groups. In the CA group, an increased clearance distance was observed for each obstacle height and this clearance distance was greater in the higher obstacle. Therefore, the results from this study expand on those from Morton et al. (2004) and show that a similar increase in clearance distance is observed in slower walking cerebellar ataxia participants and that hypermetria may be increased for differing obstacle heights since there was an increase in clearance distance with the greater obstacle height. As well, since clearances were similar between the H and MV groups, this further suggests the larger clearance in the CA group was not a result of gait velocity. This would support the view that participants with cerebellar ataxia provide a greater safety margin when stepping over higher obstacles.

Leading limb elevation angle ranges of the thigh and shank segments became larger for all groups as obstacle height increased while no differences were observed in the foot elevation angle range across conditions within each group. MacLellan & McFadyen (2010) observed similar patterns in healthy young adults during obstacle clearance and attributed these changes in elevation angle range to the required increase of hip and knee joint angles to successfully step over obstacles. What is interesting to note is that thigh and shank elevation angle range did not differ in the obstacle conditions between the H and CA groups although some differences were found between these groups and the MV group. For the shank elevation angle range, the single difference between the CA and H groups was seen during level walking, where the shank elevation angle range was significantly higher in the H group. The fact that there were few differences in elevation angle range between groups may be related to findings that lower limb joint angle magnitudes in participants with cerebellar ataxia have been shown to be similar to that of healthy individuals in level walking (Stolze et al., 2002) and when walking on inclines (Earhart & Bastian, 2001).

Most of the significant differences between groups were observed in the foot segment elevation angle range, where this range was greater in the H and MV groups when compared to the CA group. These differences in foot elevation angle range may be associated with decreases in ankle range of motion (Palliyath et al., 1998) and peak ankle plantarflexion (Morton & Bastian, 2003) previously observed in people with other forms of cerebellar ataxia. Since the foot elevation angle range was lower in the CA when compared to the MV group, it can be concluded that these changes in foot elevation range are not due to walking speed but from control differences in the CA group.

How is it that people with cerebellar ataxia increase their clearance distance when 2 of the 3 elevation angles remain similar between groups for obstacle avoidance? Although Morton et al. (2004) have found greater knee flexion in a cerebellar ataxia population (which was attributed to the observed hypermetria), the elevation angle ranges observed here do not differ from a healthy adult group. To understand how the greater elevation of the lower limb is achieved, one must look at the intersegmental coordination between the limbs.

5.5.2: Intersegmental coordination during level walking and obstacle clearance

Previous studies observing locomotion in pathological populations such as Parkinson's disease (Grasso et al., 1999), spastic paraparesis (Dan et al., 2000), spinal cord injury (Grasso et al., 2004), and forefoot rheumatoid arthritis (Laroche et al., 2007) have shown that planarity of segment elevation angles remains high (above 97%) during level walking. In the current study, similar observations of high planarity during level walking can be seen in the CA group. Since these magnitudes of planarity remain high in this population, this may provide further evidence that the observed planar pattern may arise from neural oscillators located within the spinal cord for level walking (as suggested by Lacquaniti et al. (1999) with little influence from higher CNS structures. Since a population with a pure form of cerebellar atrophy with no accompanying disorders was involved in this study, this may provide stronger evidence that the cerebellum has little influence on the planar patterns formed by the segment elevation angles during level walking. This also appears to support the conclusions by Morton et al. (2004) that during locomotor tasks, there is less involvement from the cerebellum and more from the brainstem and spinal cord, with the cerebellum being important in balance control rather than direct intersegmental coordination.

Although planarity did not differ significantly between groups, a large amount of inter-group variability can be seen by the illustrated ranges of planarity in Figure 5.2 when participants with cerebellar ataxia stepped over obstacles. This increase in range is predominantly driven by 2 of the 8 participants with cerebellar ataxia. Previous work has suggested that the nucleus interpositus located in the cerebellum may function as one of the structures used to fine tune the swing phase of the gait cycle (Armstrong, 1988; Orlovsky, 1972). The current study utilized participants with diffuse atrophy of the cerebellum which would also have an affect on the nucleus interpositus. The lower values of planarity observed in these specific participants when stepping over the obstacle may be related to more severe atrophy and suggest the influence of this structure on producing the planar pattern during adaptive locomotion, although this is speculative at this time due to the small number of participants recruited in this study. Therefore, further work using a large number of participants with differing severities of cerebellar atrophy should be done to determine if this structure has a function in producing the planar pattern of segment elevation angles and therefore affect the coordination of locomotion during this task.

In each of the groups used in this study, the orientation of the covariance plane rotated counterclockwise as participants stepped over higher obstacles. Previous research has suggested that rotations of this covariance plane are related to economy of mechanical power output of the lower limb (Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999). MacLellan & McFadyen (2010) extended this idea to obstacle clearance and proposed that the observed changes in covariance plane orientation is representative of the segment elevation angles that minimize energy expenditure when stepping over obstacles. As well, a linear relationship was shown between shank-foot fundamental harmonic phase difference and covariance plane orientation in each group studied, which is not surprising since Barliya et al. (2009) have shown the direct mathematical link between these variables.

One major difference that has not been observed in previous studies of either healthy or pathological groups using the planar law of intersegmental coordination is a more complex relationship between the thigh-shank fundamental harmonic phase difference and the width of the covariance loop. Figure 5.4 shows that a non-linear relationship (see estimated trends in Figure 5.4) exists between these variables in the CA and, to a lesser extent, the MV groups. In the current study, covariance loop width increased with each increase in obstacle height in the H and MV groups and was greater for obstacle clearance conditions together when compared to level walking in the CA group. As well, no significant differences in covariance loop width were shown between groups for any condition. It was decided that further analysis was needed to determine why this difference between groups occurred.

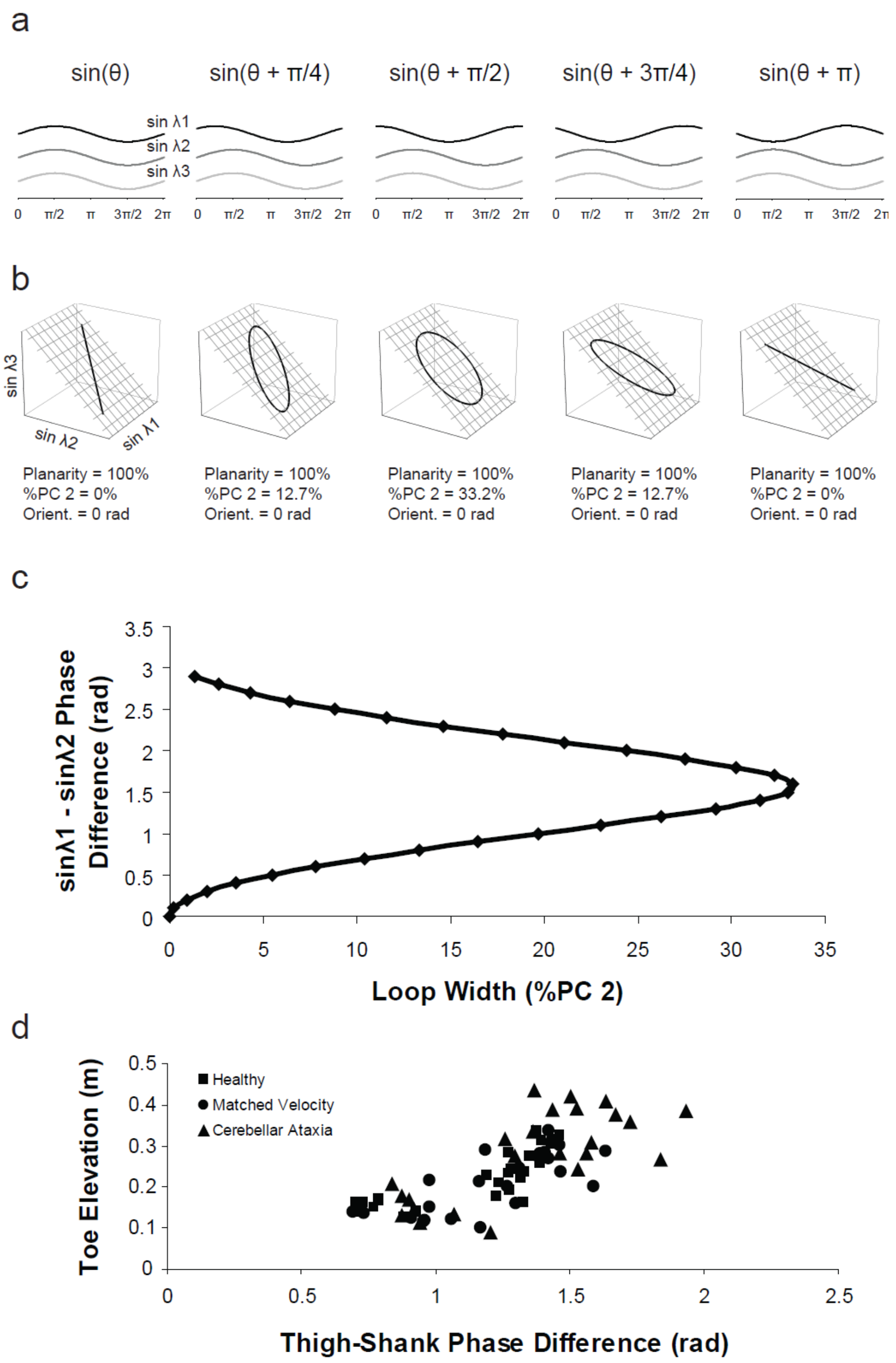

Figure 5.5: Results of a simple model used to determine the non-linear relationship between thigh-shank fundamental harmonic phase difference and covariance loop width illustrating the shifting of the thigh (λ1) waveform (a) and the corresponding covariance patterns formed by each group of waveforms (b). The relationship between a range of phase shifts (0 – 3 radians) and covariance plane loop width (c). Finally, a plot showing the relationship between thigh-shank fundamental harmonic phase difference and toe elevation across conditions (d).

In order to explain why a new pattern was observed between the thigh-shank fundamental harmonic phase difference and covariance loop width, a simple model was developed here using 3 sinusoidal waves, λ1, λ2, and λ3, representing segmental elevation angle fundamental harmonics for the thigh, shank and foot segments respectively (Figure 5.5). The phase difference between λ2 and λ3 (shank and foot segments) was held constant at 0 radians while λ1 (thigh segment harmonic) was submitted to four different phase shifts in time (π/4, π/2, 3π/4, and π radians; the latter representing complete opposite phase or out of phase; see Figure 5.5a). With no phase difference between λ2 and λ3 (shank and foot), the orientation of the covariance plane (see Figure 5.5b) did not alter as proposed by Barliya et al. (2009). The width of the covariance loop, however, did as thigh-shank phase difference changed from 0 and π radians (thigh waveform being in or out of phase respectively) showing different directions on the same plane orientation (see plots in Figure 5.5b). In both of the extreme conditions (0 and π radians), the percent variance explained by the 2nd PC was zero indicating the covariance loop had no width. For all other shifts in λ1, a loop was formed with the greatest width for a π/2 radian (or a 90 degree) shift.

When this behaviour is plotted as shown in Figure 5.5c with λ1 shifted from 0 to 3 radians, the non-linear pattern observed in the present study for MV and CA groups (see Figure 5.4) was found. That is, when the phase difference between λ1 and λ2 is greater than π/2 radians, the width of the resulting covariance loop decreases again. Previous studies examining this relationship (Courtine & Schieppati, 2004; MacLellan & McFadyen, 2010) have included locomotor movements for which this fundamental harmonic phase difference does not surpass π/2 radians and therefore has not been observed before. The present study allowed for a greater range of thigh-shank fundamental harmonic phase difference leading to the observed patterns. Barliya et al. (2009) suggested that elevation angle amplitude determines covariance loop width, but the present results also illustrate how phase differences between the thigh and shank segments relate to the width of the covariance loop.

Now that we have shown that the observed non-linear relationships can be simply determined by a greater phase difference between thigh and shank segments, the next step would be to determine why such phase differences are greater in the CA and to a lesser extent the MV groups. Since obstacle clearance is dependent on elevation of the foot, the relationship between thigh-shank fundamental harmonic phase difference and the maximum height of the toe trajectory was analyzed for the three groups as shown in Figure 5.5d. Pearson correlations were used to quantify the relationships between these 2 variables and r-values of 0.796, 0.740, and 0.734 (respective r-squared values of 0.634, 0.548, and 0.539) were obtained for the H, MV and CA groups respectively. Therefore, over half of the variance of the changing maximum toe elevation is explained by the increasing fundamental harmonic phase difference between the thigh and shank segments. This may also explain the significant differences observed between groups for the thigh-shank fundamental harmonic phase difference. The remaining variance of the changing maximum toe height may be explained by the phase difference between the shank and foot segments as well as the extension of the supporting limb. Future work using modelling or regression analysis using these variables may indicate how the limb is elevated in this population.

Therefore, participants with cerebellar ataxia may control obstacle clearance through similar CNS mechanisms as seen in healthy adults (MacLellan & McFadyen, 2010); whereby segment elevation angles are controlled by neural oscillators in the CNS with the phase difference between adjacent segment elevation angle waveforms being one of the variables controlled by the CNS. Morton et al. (2004) concluded that the increase in limb elevation may be a voluntary strategy in which the margin of safety is increased when stepping over obstacles. This conclusion was based upon observing knee angle acceleration profiles and intersegmental forces between the shank and thigh in a population with cerebellar ataxia. The current study expanded on the study by Morton et al. (2004) by using a technique to examine intersegmental coordination which has been used to understand locomotor control mechanisms. We support these conclusions while also showing possible CNS control mechanisms to facilitate the increased clearance. These results may aid rehabilitation professionals by providing an indicator of what causes hypermetria in participants with cerebellar ataxia and provide a measure of segment coordination leading to a the adaptation of locomotor patterns for obstacle clearance.

5.6: Conclusions

Participants with cerebellar ataxia were able to successfully clear obstacles during locomotion in this study. Although elevation angle ranges for the thigh and shank segments were similar in the CA group when compared to the H group, a larger clearance distance was observed in the CA group. Observations of planarity, covariance loop width, and covariance plane orientation in segmental elevation angles did not reveal how this larger clearance distance was obtained and suggested that walking speed and cerebellar dysfunction did not have a large influence on such planar patterns. Examination of the relationship between the thigh-shank fundamental harmonic phase difference and covariance loop width led to the conclusion that a simple mathematical relationship exists between these variables and what was important was an increased thigh-shank phase difference. A relationship between thigh-shank fundamental harmonic phase difference and maximum toe elevation was observed and it was concluded that the CA group increased this phase difference in order to achieve a greater toe clearance. This increase in toe clearance is therefore most likely a voluntary strategy to increase the margin of safety when stepping over obstacles as discussed previously by Morton et al. (2004), and thigh-shank coordination appears to be the CNS control mechanism for this increase.

5.7: Acknowledgements

This study was funded by a Natural Sciences and Engineering Research Council of Canada grant for Bradford McFadyen and scholarship for Michael MacLellan. The authors thank Mr. Guy St-Vincent for his technical support, Dr. Pierre Provencher for his assistance in recruitment, and Ms. Andréanne Blanchette for ensuring the safety of the participants during the course of this study.

5.8: Bridging Paragraph

As stated, ARCA-1 is a pure form of cerebellar ataxia that causes diffuse atrophy throughout the cerebellum. The previous study shows that planar patterns of segment elevation angles persist in participants with this disorder, supporting previous evidence that there is less involvement from the cerebellum and more from the brainstem and spinal cord during locomotion, with the cerebellum being important in balance control rather than intersegmental coordination (Morton et al. 2004). Since this neurological disorder produces diffuse atrophy throughout the cerebellum, there is symmetry of its effects on the lower limbs. This is not the case in neurological dysfunction caused by stroke. In people with stroke, there is generally an asymmetry on either side of the body due to the specific neural pathways damaged. This leads to asymmetrical locomotion patterns between lower limbs which may be due to the stroke directly or from compensation by the healthy side. This pathology is an example of an interesting case where coordination and control of locomotion may differ significantly between each lower limb. The following study will examine such locomotor control asymmetry in participants with a previous stroke.

Chapter 6: Study 4

“Comparison of locomotor control mechanisms for segmental coordination between non-paretic and paretic limbs during obstacle clearance following stroke”

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6.1: Abstract

The objective of this study was to use the planar law of intersegmental coordination to understand differences in segmental control mechanisms between the paretic and non-paretic limbs during obstacle clearance in persons with chronic stroke. Six persons with chronic stroke stepped over obstacles of two different heights (5 and 15% of leg length) leading with their paretic and non-paretic limbs. Kinematic data were collected and segment elevation angles (absolute segment angular position with respect to vertical) were calculated for the thigh, shank, and foot segments. Established mathematical techniques related to the planar law of intersegmental coordination (quantifying covariance and temporal phase relationships among elevation angles) were then applied to differentiate the control underlying the coordination of these segment elevation angle trajectories between limbs. Results showed that segment covariance in elevation angles followed the planar law of intersegmental coordination during level walking (i.e., 3 elevation angles forming a plane and the variance explained by 2 principle components) for both paretic and non-paretic limbs. During obstacle clearance, however, relationships between covariance plane characteristics and segment phase differences for adjacent elevation angles differed in the non-paretic limb, likely due to a need for greater limb elevation for obstacle clearance during paretic limb support. It was concluded that segment elevation angles in the paretic and non-paretic limbs followed a planar pattern suggesting that the basic control of the paretic limb is not affected by a stroke. Also, temporal differences between the thigh and

shank segments in the non-paretic limb highlight compensation due to insufficient support by the paretic limb. These results highlight the preservation of basic control mechanisms in the paretic limb during obstacle clearance, yet also reveal fundamental motor control changes that most likely result from decreased dynamic stability when the body is being supported solely by the non-paretic limb. Future research in this area may help to pinpoint specific control deficits in this population and lead to improved rehabilitation methods to target deficits in intersegmental coordination during walking.

6.2: Introduction

Human locomotor patterns must be flexible in order to successfully navigate through different environments. While healthy individuals meet these demands with ease, people with disturbed motor control may have problems implementing anticipatory locomotor adjustments (ALAs) necessary for successful navigation through these environments. For example, Said et al. (1999) have shown that people with a previous stroke have more difficulty clearing obstacles, thus increasing the risk of falls. Knowledge of lower limb coordination mechanisms during obstacle clearance in persons with chronic stroke will increase our understanding on locomotor adaptations and help target rehabilitation accordingly.

It has been known for some time that characteristics of the disturbed locomotor control are differentially expressed among persons post-stroke resulting in a variety of movement patterns and walking disabilities. Characteristics of disturbed motor control or impairments include a lack of muscle activation (paresis), hyperactive stretch reflexes (spastcity), excessive coactivation of antagonists and excessive resistance of non-neural components (Dietz, Quintern, & Berger, 1981; Knutsson & Richards, 1979; Lamontagne, Malouin, & Richards, 2000, 2001; Lamontagne, Malouin, Richards, & Dumas, 2002; Lamontagne, Richards, & Malouin, 2000; Richards & Knutsson, 1974; Shiavi, Bugle, & Limbird, 1987). These impairments often result in decreased hip and knee flexion and ankle dorsiflexion of the paretic limb during the swing phase of locomotion when compared to healthy adults (Chen et al., 2003; Knutsson & Richards, 1979; Lamontagne et al., 2001) or to the non-paretic limb (Chen et al., 2003). These decreases in maximum flexion are detrimental to obstacle clearance which requires greater flexion at each of these joints (McFadyen & Winter, 1991).

Said et al. (1999) reported an increased rate of collisions when participants with a previous stroke step over an obstacle. They also increase their step time (Said et al., 2001) and the difference in horizontal distance between their centre of mass and centre of pressure (Said et al., 2008) when compared to healthy individuals. According to Said et al. (2005), however, maximum flexion of the hip, knee and ankle of persons post-stroke when stepping over an obstacle does not differ from healthy individuals. This finding is likely related to the low obstacle height of 0.04m, which is diminished in comparison to obstacle encountered in daily life. Despite this low obstacle height, however, collisions still occurred. Thus, there is a need to better understand the deficits in motor control that lead to collisions during obstacle clearance. Furthermore, previous studies have focussed mainly on examining the body's kinematics during obstacle clearance in persons post stroke (Said, Galea, & Lythgo, 2009; Said et al., 2005; Said et al., 2008; Said et al., 2001; Said et al., 1999), in comparison to healthy adults. To our knowledge, the performance of the paretic and non-paretic limbs has not been compared when walking or stepping over obstacles. Such a comparison is important to enable the identification of interlimb differences in coordination that could provide further insight into the effects of a stroke on locomotor control mechanisms.

In past studies, the planar law of intersegmental coordination (Borghese et al., 1996) has been used to quantify coordination during different forms of locomotion. According to this law, when the segment elevation angles for one lower limb (thigh, shank, and foot) are plotted in 3-dimensional (3D) space for a gait cycle (heel contact to subsequent heel contact), they tend to form a teardrop-shaped loop which fits within a plane in that 3D space. This has been observed in different walking conditions such as level walking (Borghese et al., 1996), walking at differing velocities (Bianchi, Angelini, & Lacquaniti, 1998; Bianchi, Angelini, Orani et al., 1998), on an incline (Noble & Prentice, 2008), up stairs (Ivanenko et al., 2008), backwards (Grasso et al., 1998), with a bent posture (Grasso et al., 2000), with body weight support (Grasso et al., 2004) and on a curved trajectory

(Courtine & Schieppati, 2004). As well, this pattern has been shown to persist during level locomotion in persons with disturbed motor control due to Parkinson’s Disease (Grasso et al., 1999), spastic paraparesis (Dan et al., 2000), spinal cord injury (Grasso et al., 2004), and forefoot rheumatoid arthritis (Laroche et al., 2007). The fact that these elevation angles do form a loop which orients on a plane has been argued to show a reduction in the available degrees of freedom by the central nervous system (CNS) when coordinating locomotion (Bianchi, Angelini, Orani et al., 1998; Borghese et al., 1996; Lacquaniti et al., 1999). It is unknown if such coordination is maintained for hemiparetic gait following stroke.

Other studies have further advanced the planar law of intersegmental coordination to show strong relationships between specific covariance plane characteristics (in particular the width of the loop and the orientation of the plane with respect to the thigh axis) and temporal phase differences between the fundamental waveforms underlying segment elevation angle trajectories (shank-foot phase differences and plane orientation: (Bianchi, Angelini, Orani et al., 1998), thigh-shank phase differences and covariance loop width: (Courtine & Schieppati, 2004)). These findings have led to the argument that segment phase differences may be directly controlled by the CNS during level locomotion (Barliya et al., 2009; Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999) and obstacle clearance (MacLellan & McFadyen, 2010). Therefore, using data analysis techniques related to the planar law of intersegmental coordination may help differentiate underlying locomotor control problems resulting from stroke.

The planar law of intersegmental coordination appears to also hold for stepping over obstacles (Ivanenko et al., 2005; MacLellan & McFadyen, 2010), however, MacLellan & McFadyen (2010) showed that specific covariance loop shape and orientation characteristics changed when stepping over obstacles with the leading limb. Such changes were highly correlated with fundamental phase differences between adjacent lower limb segments, suggesting that anticipatory locomotor adjustments are implemented as an adjustment to the basic level walking pattern by the CNS using segment phase differences as a control variable (MacLellan & McFadyen, 2010). In more recent unpublished work

(MacLellan, Dupré, and McFadyen, submitted), a non-linear relationship between segment phase differences and segment elevation angle covariance loop characteristics has been shown in people with cerebellar ataxia when stepping over obstacles. Such change in the linear relationship was attributed to the need to increase the phase difference between the thigh and shank segments, which led to an increase in clearance distance over the obstacle.

Given that changes in the coordination of the lower limb segments have been revealed by the planar law of coordination in healthy adults and adults with cerebellar ataxia as they step over obstacles, the application of such coordination analysis techniques to the lower limb movements of persons post stroke for level walking and stepping over obstacles could also be exploited to reveal coordination changes between paretic and non-paretic lower extremities. Such a study has the potential to inform about targets for gait assessment and retraining in these persons. The purpose of this study is thus to apply the same analysis techniques to better understand differences in intersegmental coordination between the paretic and non-paretic lower extremities of persons post stroke as they walk over obstacles. It is expected that planarity of segment elevation angles will remain high in the paretic and non-paretic limbs during level walking and obstacle clearance for both limbs, suggesting a similar simplification of CNS control in each limb as seen in healthy adults. It is further hypothesized that differences in specific characteristics of this planar pattern of segment elevation angles (specifically covariance loop width and covariance plane orientation) between paretic and non-paretic limbs will highlight different coordination strategies for each limb.

6.3: Methods

Six adults with a previous stroke (3 female/ 3 male, age: 56.2 +/- 15.4 years, height: 1.66 +/- 0.05 m, weight: 63.1 +/- 9.0 kg, time since stroke: 22.5 +/- 23.8 months, walking speed 0.96 +/- 0.24 m/s) participated in the study. One participant wore an ankle-foot orthosis (AFO) throughout the experiment. All participants provided informed consent prior to participation in the study according to ethics guidelines from the Quebec Rehabilitation Institute (IRDPQ) and Université Laval.

To examine segment coordination during obstacle clearance, participants initiated walking approximately 3-4 steps from an obstacle and stepped over it leading with the paretic or the non-paretic sides on separate trials. The different heights of the obstacles used in the study were normalized to either 5 or 15% of each participant's leg length (greater trochanter to floor) with a constant depth of 0.05 m. Control trials with no obstacle present were used to examine level walking. In all, the protocol consisted of 6 experimental conditions (paretic level, paretic lead 5%, paretic lead 15%, non-paretic level, non-paretic lead 5%, non-paretic lead 15%) presented in a randomized order. The number of trials performed for each condition varied (2-6 trials) for each participant and was dependent on walking capacity and endurance.

Full body 3D kinematic data were collected at 60 Hz by a 3-bar Optotrak system (Northern Digital Inc., Waterloo, Canada). Non-collinear triads of IRED markers on plastic plates were fixed to the feet, shanks, thighs, pelvis, trunk, and head segments. A calibration trial was collected and anatomical landmarks (5th metatarsal, medial/lateral malleoli, medial/lateral femoral condyles, left/right iliac crests, and left/right anterior superior iliac spines) as well as the heels and toes of each foot were digitized to determine their virtual trajectories off-line. Kinematic data were filtered using a dual-pass 2nd order Butterworth filter with a cut-off frequency of 6 Hz.

The kinematic data were used to create a 9 segment biomechanical model (feet, shanks, thighs, pelvis, trunk, and head). This biomechanical model was then used to calculate toe clearance (vertical distance from the upper front edge of the obstacle to the toe) and limb elevation (vertical distance between the hip and ankle joints, therefore removing the effects of the foot segment) of the leading limb when stepping over the obstacle. As well, the model was used to calculate leading limb peak flexion of the hip, knee, and ankle joints for the stride over the obstacle (leading limb stride was defined from heel contact prior to clearance to heel contact following clearance for the first limb to step over the obstacle). Elevation angles of the foot, shank, and thigh segments were also calculated (absolute angles of each segment with respect to the vertical, as calculated in (Borghese et al., 1996)) during the stride of the leading limb over the obstacle. The range of each segment's elevation angle trajectory was calculated by subtracting the maximum elevation angle from the minimum elevation angle.

The planar law of intersegmental coordination (Bianchi, Angelini, & Lacquaniti, 1998; Bianchi, Angelini, Orani et al., 1998; Borghese et al., 1996; Courtine & Schieppati, 2004; Ivanenko et al., 2005, 2007; Ivanenko et al., 2008) was used to quantify coordination of the leading limb when stepping over the obstacle. According to this method, thigh, shank, and foot elevation angles were normalized to 100% of stride duration and the mean over each stride was subtracted from each segment trajectory to normalize the amplitudes of the elevation angles. Principal component analysis (PCA) was then applied to each group of elevation angles to determine planarity (percent variance explained by the first 2 principal components), covariance plane orientation (direction cosine between the 3rd principal axis of the loop and the positive axis of the thigh segment), and the planar covariance loop width (percent variance explained by the 2nd principal component). Theoretically the first 2 principal components should explain 100% of the variance of a plane. The method used to determine orientation of the covariance plane has been found to be sensitive in determining rotations of the covariance plane about the longitudinal axis of the covariance loop (Bianchi, Angelini, Orani et al., 1998). A Fourier series using 10 harmonics was then used to represent the time course of the thigh, shank, and foot segment elevation trajectories. Phase differences between the fundamental harmonic patterns of adjacent elevation angles were found by subtracting the distal segment phase from the proximal segment (thigh-shank and shank-foot) phase (Bianchi, Angelini, Orani et al., 1998; Courtine & Schieppati, 2004). This analysis provided an indication of changes in timing between adjacent segments.

In the current study, statistically significant main effects for joint angle peaks, elevation angle ranges, elevation angle planarity, covariance loop width, covariance plane orientation, and adjacent segment phase differences between obstacle heights (level walking, 5%, 15%) were determined separately for the paretic and non-paretic leading limbs using the Friedman test. If significant main effects were found, the Wilcoxon sign-rank test was used to determine significant differences between each obstacle condition (level walking/5%, level walking/15%, 5%/15%). For toe clearance and limb elevation, the Wilcoxon test was used to determine significant differences between obstacle heights (5%, 15%). To determine statistically significant differences between paretic and non-paretic limbs, a Mann-Whitney U test was used for each obstacle condition. Statistical significance was accepted at a level of $p \leq 0.05$.

6.4: Results

6.4.1: Paretic Limb

When the paretic leading limb stepped over the obstacle, toe clearance was similar ($p = 0.219$) for the 5% (0.115 +/- 0.011 m) and 15% (0.116 +/- 0.028 m) obstacle height conditions. However, limb elevation was greater in the 15% (0.571+/-0.061 m) when compared to the 5% (0.643 +/- 0.051 m) obstacle height condition ($p = 0.016$) as seen by a smaller distance between the hip and ankle joints. As well, main effects were observed for peak knee ($p < 0.001$) and hip ($p < 0.001$) flexion angles for obstacle height (Table 6.1, Figure 6.1). Wilcoxon tests showed that peak flexion in both of these joints increased with each increase in obstacle height ($p = 0.016$). A main effect was also shown for the range in thigh elevation angles ($p < 0.001$) and Wilcoxon tests showed significant increases with obstacle height ($p = 0.016$, Table 6.1). No significant changes were observed in foot ($p = 0.956$) or shank ($p = 0.430$) elevation angle ranges in the paretic limb.

Table 6.1: Maximum joint flexion angles and elevation angle range for level walking and obstacle clearance.

	Paretic			Non-Paretic		
	Level Walking	5% Height	15% Height	Level Walking	5% Height	15% Height
Max Joint Angle (°)						
Dorsiflexion ± SD	1.3 ± 1.1^{c}	1.6 ± 2.0^{c}	3.2 ± 2.6^{c}	12.3 ± 3.3	17.6 ± 2.9	18.6 ± 2.9
Knee Flexion ± SD	34.2 ± 2.7^{c}	54.2 ± 4.8^{a,c}	67.1 ± 4.3^{a,b,c}	66.6 ± 1.2	90.4 ± 5.4^{a}	102.3 ± 4.9^{a,b}
Hip Flexion ± SD	16.8 ± 3.0	44.5 ± 5.0^{a}	56.2 ± 5.1^{a,b}	26.4 ± 2.4	56.7 ± 5.2^{a}	63.5 ± 5.3^{a,b}
Angle Range (°)						
Foot ± SD	63.0 ± 3.5^{c}	60.6 ± 5.5	54.0 ± 7.9	82.4 ± 4.9	60.0 ± 9.6^{a}	60.5 ± 9.7^{a}
Shank ± SD	52.6 ± 1.8	50.1 ± 3.0^{c}	46.7 ± 5.4^{c}	66.2 ± 2.2	61.3 ± 6.7	64.9 ± 6.9
Thigh ± SD	38.6 ± 2.5	58.8 ± 3.2^{a}	67.0 ± 5.5^{a,b}	42.1 ± 1.1	64.3 ± 5.9^{a}	72.0 ± 6.6^{a,b}

[a] – significantly different from level walking
[b] – significantly different from 5% height
[c] – significantly different from non-paretic limb

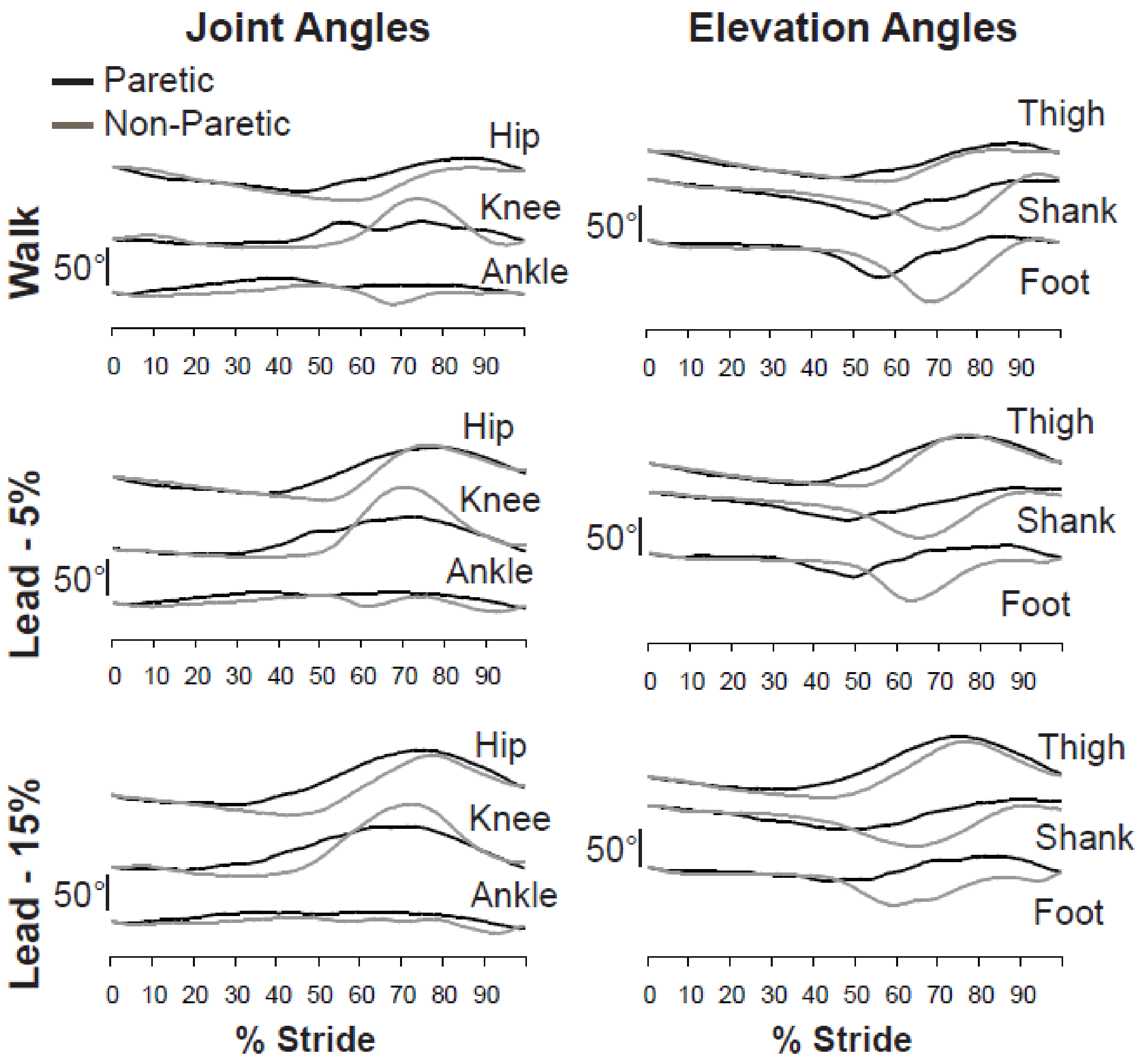

Figure 6.1: Joint and elevation angles of the paretic (black) and non-paretic (grey) limbs during level walking and when leading during stepping over an obstacle. These angles are represented over 100% of a stride (heel contact to subsequent heel contact).

Sample trajectories from a single participant of the covariance of segment elevation angles during level walking and obstacle clearance can be seen in Figure 6.2. This figure highlights the shape of the planar covariance loop, which deviates in the paretic limb from the teardrop-shaped loop seen during walking in the non-paretic limb and in healthy adults (Bianchi, Angelini, & Lacquaniti, 1998; Bianchi, Angelini, Orani et al., 1998; Borghese et al., 1996; Courtine & Schieppati, 2004; Ivanenko et al., 2005, 2007; Ivanenko et al., 2008; MacLellan & McFadyen, 2010).

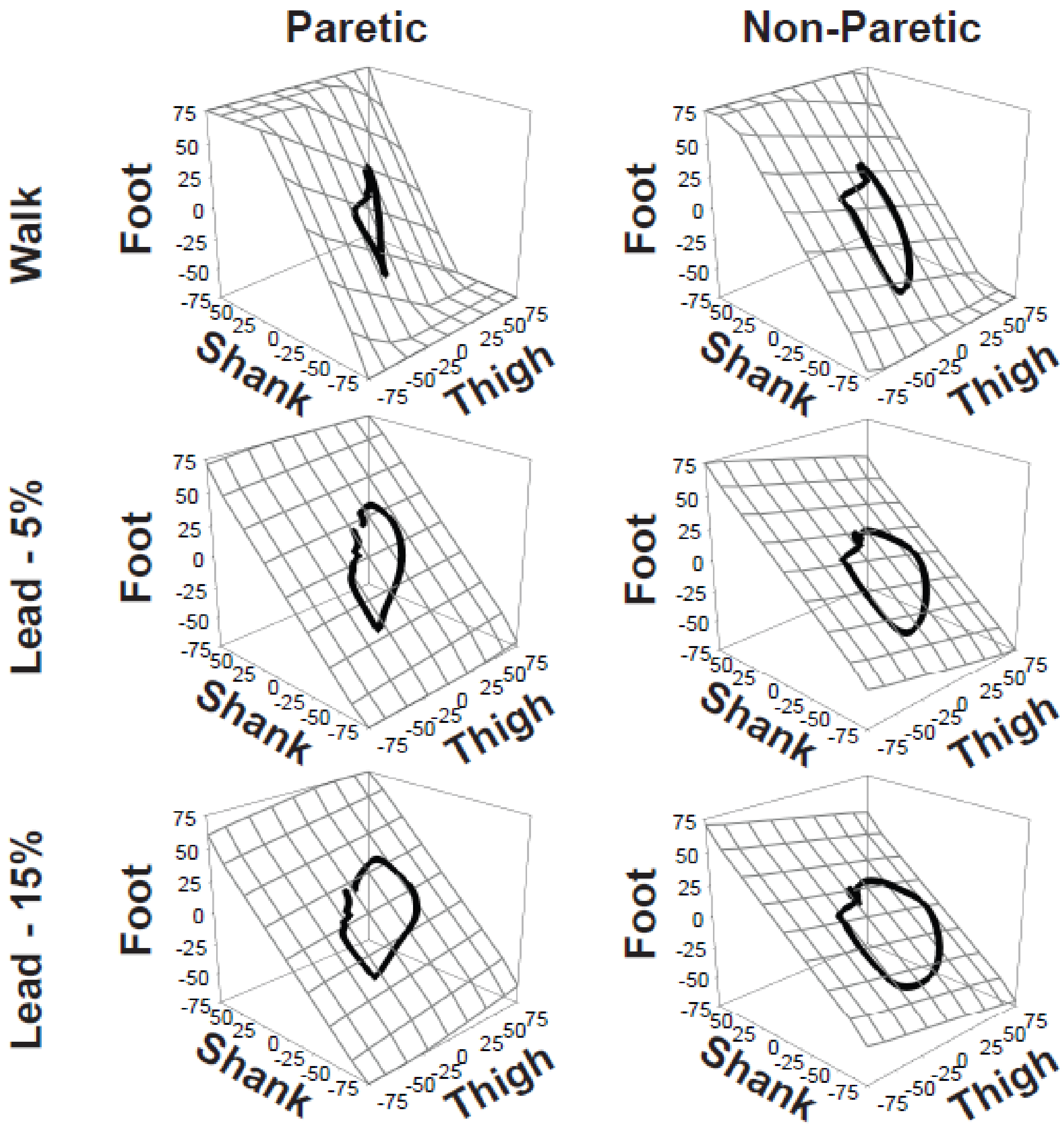

Figure 6.2: Representative covariance loops for the paretic and non-paretic limbs during level walking and when leading during stepping over an obstacle.

Although shapes of these loops differed between participants, the overall covariance loop pattern was preserved. In addition, planarity remained high for the paretic leading limb (97.98 +/- 0.54% of the total variance was explained by the first 2 principal components) and this did not change for differing obstacle heights (p = 0.072, Figure 6.3 top row). The width of the covariance loop in the paretic leading limb showed a main effect for obstacle height (p = 0.012) and post hoc analysis showed that loop width increased significantly for each obstacle height (p = 0.047, Figure 6.3 middle row). Similarly, the orientation of the covariance plane showed a main effect ($p < 0.001$). It was observed that this plane rotated more counterclockwise as obstacle height increased ($p = 0.016$, Figure 6.3 bottom row).

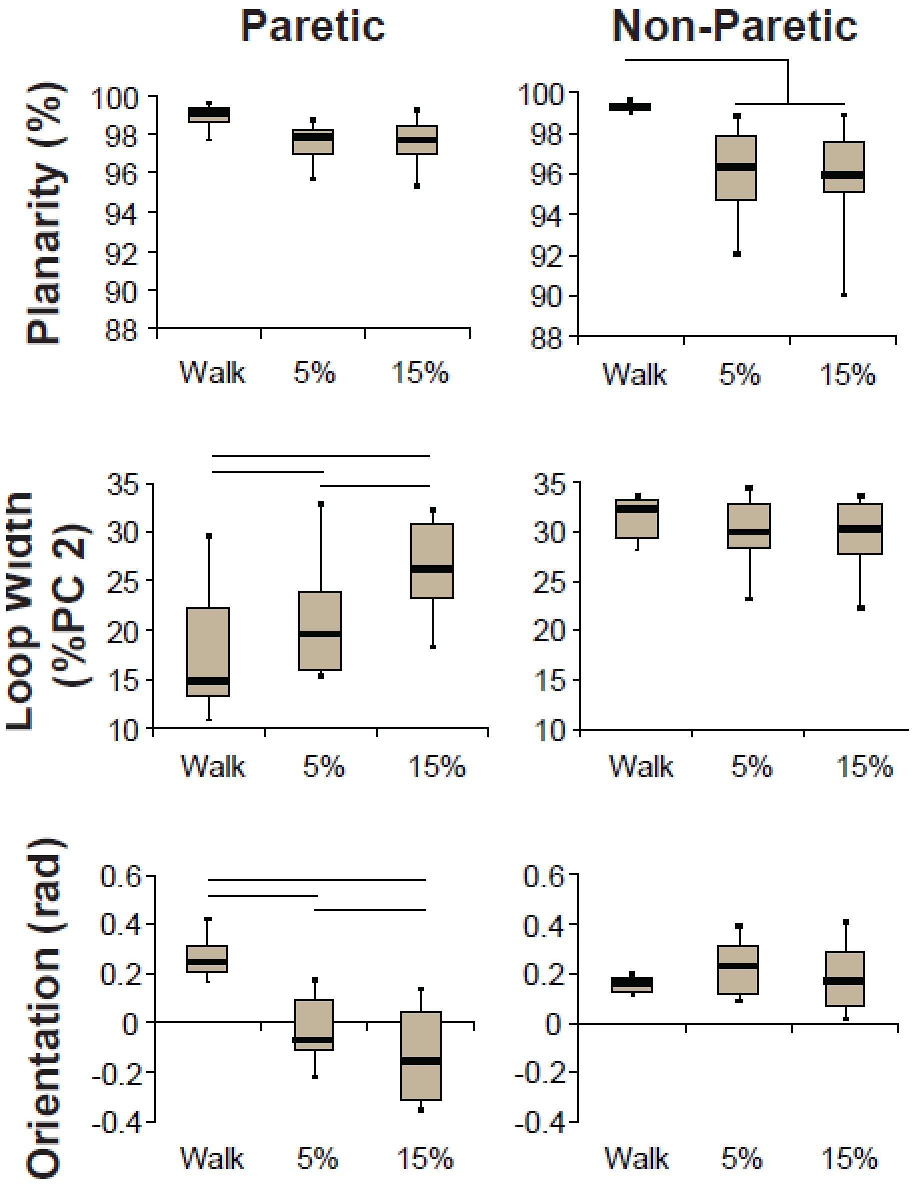

Figure 6.3: Box plots illustrating changes in planarity (top), planar covariance loop width (middle), and plane orientation (bottom) for the paretic and non-paretic limbs. Significant differences ($p \leq 0.05$) between conditions are indicated with horizontal bars.

The phase difference between the thigh and shank fundamental harmonics increased in the paretic leading limb ($p < 0.001$) and Wilcoxon tests showed the harmonic difference increased with obstacle height ($p = 0.016$, Figure 6.4 top row). The phase difference between the shank and foot fundamental harmonics decreased significantly in the paretic limb ($p < 0.001$). The Wilcoxon test showed that this phase difference decreased significantly with each increase in obstacle height ($p = 0.016$, Figure 6.4 bottom row).

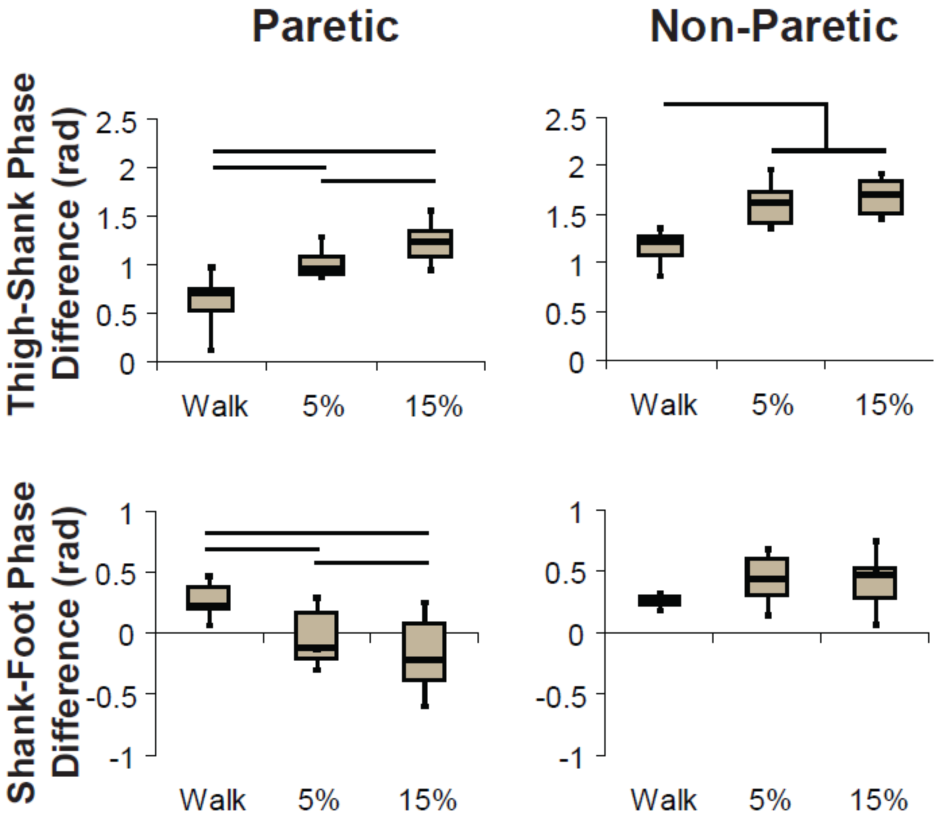

Figure 6.4: Box plots illustrating changes in thigh-shank (top) and shank-foot (bottom) fundamental harmonic phase differences for the paretic (left) and non-paretic (right) limbs. Significant differences ($p \leq 0.05$) between conditions are indicated with horizontal bars.

6.4.2: Non-paretic Limb

When the non-paretic limb leads in stepping over the obstacle, toe clearances were comparable for the 5% (0.130 +/- 0.027 m) and 15% (0.139 +/- 0.024 m) obstacle height conditions ($p = 0.500$). Although clearance distance did not differ, limb elevation was significantly greater in 15% (0.502 +/- 0.037 m) when compared to the 5% (0.579 +/- 0.031 m) obstacle condition ($p = 0.016$). As with the paretic limb, main effects were observed for peak knee ($p < 0.001$) and hip ($p < 0.001$) flexion and post hoc analysis showed that these peak joint angles increased with obstacle height ($p = 0.016$, Figure 6.1 and Table 6.1). The range in foot elevation angles increased for the non-paretic limb ($p = 0.008$) and Wilcoxon tests showed that this range was greater in level walking when compared to the obstacle conditions (p = 0.016, Table 6.1). Similarly, the thigh elevation angle range of the non-paretic limb showed main effects (p < 0.001) and post hoc tests revealed increases with each obstacle height (p = 0.016, Table 6.1). No significant differences were observed in the non-paretic shank elevation angle range (p = 0.184).

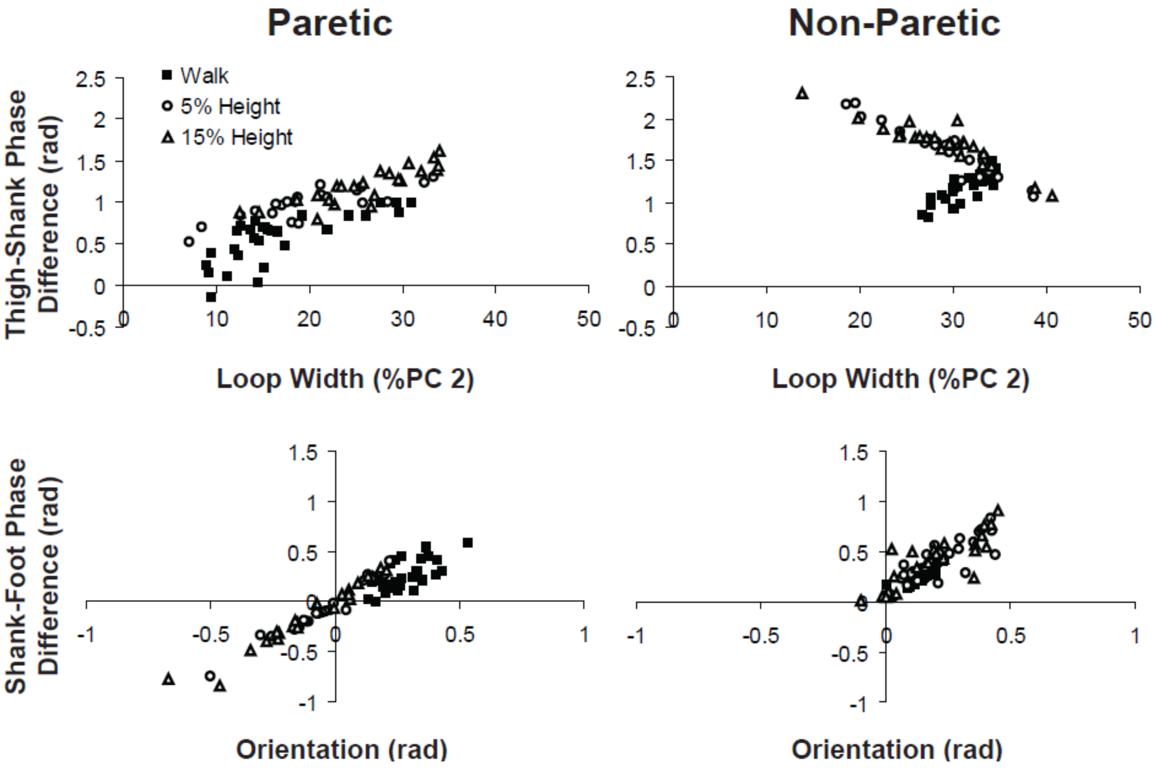

Figure 6.5: Plots illustrating the relationship between thigh-shank fundamental harmonic phase difference and covariance plane loop width (top) and shank-foot fundamental harmonic phase difference and covariance plane orientation (bottom) during obstacle clearance in the paretic (left) and non-paretic (right) limbs.

With respect to the planar law of intersegment coordination, Figure 6.3 shows that planarity decreased significantly on the non-paretic side (p = 0.008) and was significantly lower when stepping over obstacles (p = 0.016, Figure 6.3 top row), although Figure 6.2 shows that the shape of the planar covariance loop is similar to that observed for level walking in healthy adults (Bianchi, Angelini, & Lacquaniti, 1998; Bianchi, Angelini, Orani et al., 1998; Borghese et al., 1996; Courtine & Schieppati, 2004; Ivanenko et al., 2005, 2007; Ivanenko et al., 2008; MacLellan & McFadyen, 2010). As the height of the obstacle increased, no significant changes were observed in covariance loop width (p = 0.956, Figure 6.3 middle row) or the orientation of the covariance plane (p = 0.570, Figure 6.3 bottom row).

The phase difference between the thigh and shank fundamental harmonics showed a main effect in the non-paretic side (p = 0.006). Wilcoxon tests showed that the fundamental harmonic phase difference was greater in the obstacle conditions when compared to level walking (p = 0.016, Figure 6.4 top row). No significant differences were observed between obstacle conditions for shank-foot fundamental harmonic phase difference in the non-paretic limb (p = 0.184, Figure 6.4 bottom row).

The relationship between the thigh-shank fundamental harmonic phase difference and the covariance loop width tended to follow a non-linear pattern in the non-paretic limb, where a maximum covariance loop width was reached and followed by a decrease in covariance loop width as the harmonic phase difference increases (Figure 6.5, top right). Although a different pattern was observed between these variables, a linear relationship was once again observed between covariance plane orientation and the fundamental harmonic phase difference between the shank and foot segments as seen in the paretic limb (Figure 6.5, top left).

6.4.3: Comparison between Paretic and Non-paretic Limbs

Obstacle foot clearance did not differ significantly between paretic (0.116 +/- 0.019 m) and non-paretic (0.135 +/- 0.025 m) leading limbs (5%: p = 0.294, 15%: p = 0.155). However, limb elevation was significantly greater in the non-paretic limb (5%: 0.579 +/- 0.031 m, 15%: 0.502 +/- 0.037 m) when compared to the paretic limb (5%: 0.643 +/- 0.051 m, 15%: 0.571+/-0.061 m) in the 5% (p = 0.021) and 15% (p = 0.021) obstacle height conditions.

Peak ankle dorsiflexion was significantly greater in the non-paretic limb (Figure 6.1, Table 6.1) during level walking (p = 0.013) and at obstacle heights of 5% (p = 0.004) and 15% (p = 0.013). As well, knee flexion was significantly greater in the non-paretic limb at each obstacle height (p < 0.001). No differences were observed between limbs for peak hip flexion (level: p = 0.090, 5%: p = 0.066, 15%: p = 0.090). Foot elevation angle range was significantly greater on the non-paretic side during level walking (p = 0.008) as well as for shank elevation angle range for the 5% (p = 0.047) and 15% (p = 0.002) obstacle height conditions (Figure 6.1, Table 6.1). No significant differences were observed between limbs for thigh elevation angle range (level: p = 0.120, 5%: p = 0.242, 15%: p = 0.155).

When examining characteristics of coordination, planarity was consistently high and did not differ between paretic and non-paretic leading limbs (level: p = 0.294, 5%: p = 0.197, 15%: p = 0.120). Loop width was observed to be greater on the non-paretic side when compared to the paretic side during level walking (p = 0.004) and in the 5% obstacle condition (p = 0.021, Figure 6.3). Moreover, the orientation of the plane was rotated more counterclockwise for the non-paretic limb during level walking and for the paretic side in the 5% (p = 0.021) and 15% (p = 0.021) obstacle height conditions.

The phase lead of the thigh fundamental harmonic with respect to the shank was greater in the non-paretic limb when compared to the paretic limb for level walking (p = 0.002), as well as the 5% (p = 0.001) and 15% (p = 0.004) obstacle height conditions (Figure 6.4). As well, the phase lead of the shank fundamental harmonic with respect to the foot was lower on the paretic side in the leading limb for the 5% (p = 0.013) and 15% (p = 0.008) obstacle height conditions.

6.5: Discussion

This study examined intersegmental coordination changes when persons post-stroke stepped over obstacles while leading with their paretic and non-paretic limbs. As seen in previous studies, successful obstacle clearance was achieved by increasing flexion of the hip and knee joints. This was associated with increases in different segment elevation angles (thigh segment in the paretic limb, thigh and foot segments in the non-paretic limb) which tended to maintain planar patterns in both limbs when plotted for all obstacle heights presented. Changes in characteristics of this planar pattern were found to be related to phase differences in the fundamental harmonic patterns between adjacent segments in the paretic and non-paretic limbs. In particular, the relationship between the thigh-shank harmonic phase difference and covariance plane loop width in the non-paretic limb followed a non-linear pattern not normally seen. These observations are discussed below.

6.5.1: Kinematic comparisons for leading limbs

When the leading limb steps over an obstacle, generally a toe clearance of about 0.10 m is observed (Patla & Rietdyk, 1993) in healthy participants. In the current study, a similar toe clearance was found and although a difference between the limbs of approximately 0.02 m was noted, it was not significant. Conflicting results have been previously reported by Said et al. (2001) and Said et al. (2005) for persons with stroke, with one study suggesting a trend for increased toe clearance (Said et al., 2001), while the other study showed that clearance in the paretic and non-paretic leading limbs was not significantly different than that seen in healthy persons. Said et al. (2005) attributed this difference to changes in posture during clearance, but there were also differences (which were not discussed in the study) between the ages of participants and time post-stroke. Moreover, one must consider that differences may also be the result of the large variability in disturbed motor control.

In healthy adults, successful obstacle clearance is achieved by increasing flexion at the hip and knee joints (Austin et al., 1999; Chou & Draganich, 1997; McFadyen & Winter, 1991). A similar trend was observed in the present study where peak hip and knee flexion increased as obstacle height increased in the paretic and non-paretic limbs. Previous work by Said et al. (2005) has also shown that joint angles when stepping over obstacles in persons with a previous stroke were similar to that of healthy adults, but comparisons between limbs were not addressed. In the current study, comparison between the paretic and non-paretic limbs showed that peak ankle dorsiflexion and knee flexion were significantly greater in the non-paretic limb during level walking and at each obstacle height. These changes in joint angles can be directly related to changes observed in segment elevation angles.

When observing changes between obstacle conditions, the current study showed similar changes in the thigh elevation angle range in both paretic and non-paretic limbs as obstacle height increased. Fewer changes were observed for foot elevation angle range (increases seen in the non-paretic limb) and none were observed for shank elevation angle range. Given the similar timing between relative joint angles and absolute segment angles (see Figure 6.1), the observed changes in peak hip and knee flexion with increases in obstacle height are therefore most likely related to the increases in thigh segment elevation angle range. Similar increases in thigh segment elevation angle range were observed in young healthy adults when stepping over an obstacle (MacLellan & McFadyen, 2010) and may highlight the importance of increasing hip and knee flexion to successfully step over the obstacle with both paretic and non-paretic leading limbs. The differences in peak knee flexion and ankle dorsiflexion angles between paretic and non-paretic limbs are most likely related to decreases in the shank segment elevation angle range which may be related to deficient multi-segment coordination in the paretic limb.

6.5.2: Differences in coordination between leading limbs

Recent research has used the planar law of intersegmental coordination to examine obstacle clearance (Ivanenko et al., 2005; MacLellan & McFadyen, 2010) and these studies have shown that planar patterns of segment elevation angles that occur during level walking are preserved when stepping over an obstacle. Moreover, planar patterns have been shown during level walking in persons with specific pathological conditions (Dan et al., 2000; Grasso et al., 2004; Grasso et al., 1999; Laroche et al., 2007). This is the first study to analyze locomotion in people with chronic stroke using the planar law of intersegmental coordination and focus on adaptations for obstacle avoidance. In this study segment elevation angles remained highly planar during level walking and in the leading limb on both sides for obstacle clearance. Although planarity decreased slightly as obstacle height increased when the non-paretic limb led (as reported in healthy young adults by (MacLellan & McFadyen, 2010), no significant differences were observed between paretic and non-paretic limbs. It has been suggested that a planar pattern of elevation angles indicates a reduction of lower limb degrees of freedom in the CNS control of level walking (Borghese et al., 1996; Lacquaniti et al., 1999) as well as during obstacle clearance (MacLellan & McFadyen, 2010). These results therefore suggest the preservation of fundamental locomotor control mechanisms related to the adjustment of a basic motor control pattern in both limbs despite a previous stroke.

In the paretic limb, the results show that the width of the planar covariance loop increased and the covariance plane rotated counterclockwise about the longitudinal axis of the covariance loop when higher obstacles were presented. In healthy young adults and adults with cerebellar ataxia, these increases of the planar covariance loop width have been shown to be directly related to an increased fundamental harmonic phase difference between the thigh and shank segments (MacLellan, Dupre, and McFadyen; submitted). Since similar increases in thigh-shank phase difference were observed in the paretic limb, this highlights the need for the CNS to increase this phase difference to successfully step over the obstacle in this limb (leading to the previously mentioned increase in hip and knee flexion). As well, similar rotations in covariance plane orientation have been observed when healthy adults step over increasingly higher obstacles (MacLellan & McFadyen, 2010). This change in covariance plane orientation has been related to lower limb mechanical power output economy (Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999) and may be related to the increased need for mechanical energy in order to elevate the limb over the obstacle in the paretic limb.

Although changes in covariance plane characteristics in the paretic leading limb follow patterns seen in young healthy adults, interestingly, these patterns were not conserved in the non-paretic limb. Both planar covariance loop width and plane orientation did not change significantly as obstacle height increased. As previously mentioned, changes in covariance loop width are related to the fundamental harmonic phase difference between the thigh and shank segments. In the non-paretic limb, an increase in this phase difference occurred during obstacle clearance when compared to level walking, but this was not reflected in planar covariance loop width. As well, the orientation of the covariance plane did not rotate in a counterclockwise direction as seen in the paretic limb. This lack of change in covariance plane orientation has also been observed between different modes of locomotion in infants (Dominici et al., 2010) and this result was attributed to the hypothesis proposed by Bernstein (1967) that freezing or reducing the degrees of freedom is a strategy in early motor skill acquisition to simplify the coordination between multiple body segments. This may also be true for the non-paretic leading limb when stepping over an obstacle, and further results suggest that this may be due to a compensatory mechanism as explained below.

In previous studies (Bianchi, Angelini, Orani et al., 1998; Courtine & Schieppati, 2004; MacLellan & McFadyen, 2010), high correlations have been reported between planar covariance loop width and thigh-shank fundamental harmonic phase difference as well as covariance plane orientation and shank-foot fundamental harmonic phase difference. It has been suggested that the CNS controls lower limb segment motions during locomotion by encoding elevation angle waveforms, with phase difference being one of the control variables (Barliya et al., 2009; Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999). This has also been suggested when implementing anticipatory locomotor adjustments (MacLellan & McFadyen, 2010). In particular for young healthy adults, the thigh fundamental harmonic pattern leads that of the shank and this lead increases as obstacle height increases, while the shank fundamental harmonic leads that of the foot and this lead decreases when stepping over obstacles (MacLellan & McFadyen, 2010).

When the paretic limb leads, a linear relationship was observed between planar covariance loop width and thigh-shank fundamental harmonic phase difference as well as between covariance plane orientation and shank-foot fundamental harmonic phase difference. A similar linear relationship was observed between covariance plane orientation and shank-foot fundamental harmonic phase difference in the non-paretic limb. Although these relationships appear to be linear, the spread of the data points appears to be greater than that seen in the non-paretic limb. This may suggest that during locomotion, the paretic limb is controlled by similar mechanisms in the CNS as suggested previously in healthy adults (Barliya et al., 2009; Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999), but problems may occur in coordinating the required kinematic patterns which leads to the greater spread of the data.

In the non-paretic limb, a non-linear relationship was observed between planar covariance loop width and thigh-shank fundamental harmonic phase difference, which was similar to that seen in the leading limb of participants with cerebellar ataxia when stepping over an obstacle (MacLellan, Dupré, and McFadyen; submitted). MacLellan, Dupré, and McFadyen (submitted) attributed this pattern to a simple mathematical relationship between these variables, but still noted larger fundamental harmonic phase differences between the thigh and shank segments that were suggested to lead to a greater clearance distance in the cerebellar ataxia group. In the current study, similar increases in the fundamental harmonic phase differences between the thigh and shank segments in the non-paretic limb were observed in all conditions. A major difference in this study is that this increase in phase difference was not accompanied with an increase in clearance distance. The reason is related to changes in limb elevation. The lower limb was elevated a greater distance in the non-paretic when compared to the paretic limb. Since limb elevation increases while clearance does not differ between the two limbs, this suggests that while the non-paretic limb is stepping over the obstacle, the paretic stance limb may not support the weight of the body as well requiring the non-paretic limb to be elevated higher to avoid contact with the obstacle. Said et al. (2008) concluded that stability is threatened during obstacle clearance in people with a previous stroke due to the changes observed in the difference in distance between the centre of mass and centre of pressure. According to the present results, this threat to stability most likely occurs due to disturbed multi-segment coordination in the paretic limb and instability is probably heightened when the paretic limb is supporting the body while the non-paretic limb is in swing phase over the obstacle. Therefore, the non-paretic limb must compensate for this disturbed control and elevate the limb a greater amount to step over the obstacle successfully.

As stated in the methods, 1 participant wore an AFO throughout the study. Generally, the data from this participant was similar to the other participants except for covariance loop width and thigh-shank harmonic phase difference when stepping over obstacles in the non-paretic limb. This participant displayed the lowest and highest values for these variables respectively. Again, this could highlight a compensatory mechanism whereby the decreased support of the ankle joint in the paretic limb required the greatest amount of limb elevation which was obtained by the greatest phasing difference between the thigh and shank segments. Since this study had only the 1 participant with an AFO, this is speculation and further work would be required to confirm this.

According to the above discussion, this study highlights the mechanisms of compensation that occur in the non-paretic limb due to insufficient support in the paretic limb when stepping over obstacles. Future work in this area is needed to better understand the clinical implications of these findings. For example, it would also be of interest to examine the effects of training aimed at improving control and stability of both limbs.

6.6: Conclusions

Obstacle clearance can be a hazardous task for people post stroke due to motor control deficits. The present study highlighted coordination differences between lower limb segments when persons post stroke walk over obstacles of increasing heights with either the paretic or the non-paretic limb leading. The basic planar patterns of the covariance plots of lower limb segment elevation angles tended to be similar to those observed in young healthy adults for both limbs, even when clearing relatively high obstacles. This shows that fundamental locomotor patterns are preserved for the persons with chronic stroke studied. However, while some changes were observed in the orientation and width of the covariance loop in the paretic limb, these changes were not shown in the non-paretic limb which was interpreted as evidence of a decrease in degrees of freedom in the non-paretic limb to simplify CNS control. Relationships between these plane characteristics and segment phase differences in the fundamental harmonics of elevation angles highlighted compensation mechanisms in the non-paretic limb as it steps over the obstacle that were believed to be due to insufficient support in the paretic limb. Future research in this area may help to pinpoint specific deficits and lead to improved rehabilitation methods to target deficits in intersegmental coordination during walking.

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Chapter 7: General Conclusions

7.1: Advancement of Knowledge in Locomotor Control Mechanisms

The general goal of this thesis was to use an obstacle clearance paradigm and apply the planar law of intersegmental coordination to suggest how the CNS implements ALAs in healthy individuals and determine deficits of ALA implementation in participants with specific pathologies. In general, this thesis supports the theory that phase differences between adjacent lower limb segment elevation angle waveforms may be one of the control variables for locomotion used by the CNS. Although this has been hypothesized previously for level walking (Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999), the current work supported a similar theory for the implementation of ALAs in healthy adults and in specific pathological populations. The following discussion will revisit the objectives and hypotheses set in the beginning of this thesis and discuss possible human locomotor control mechanisms proposed in this thesis.

In study 1 (Chapter 3) of this thesis, the main objective was to apply the planar law of intersegmental coordination to locomotor patterns of both the leading and trailing limbs in an obstacle clearance paradigm to better understand how ALAs are controlled by the CNS. In this study, covariance plots of lower limb segment elevation angles were found to display planar patterns for all obstacle conditions in both the leading and trailing limbs. Although planarity of covariance loops was maintained for all obstacle conditions, systematic increases were observed in covariance loop width and covariance plane orientation which were found to be highly correlated to phasing differences between adjacent lower limb segments. The consistent planar patterns during level walking and obstacle clearance suggested that a similar locomotor pattern is used during these tasks. This finding is significant since it provides further evidence that the CNS adjusts a basic locomotor pattern when an ALA is used to achieve a locomotor goal, instead of implementing an entirely new locomotor pattern all together. Since the characteristics of this plane are highly related to segment phasing, it was concluded that the CNS adjusts this basic locomotion pattern partially through the control of segment elevation angle waveform phase differences, with a constraint on the system being planarity of these waveforms. In fact, results from Barliya et al. (2009) and Chapter 5 of this thesis show how phasing between adjacent segments directly determines such plane characteristics. Previous work by Bianchi et al. (1998) and Lacquaniti et al. (1999) has suggested that the orientation of the covariance plane may be directly related to mechanical energy output during the gait cycle. When one takes a closer look at the results from this thesis, this conclusion may be challenged. For example, results from Chapter 5 suggested that the orientation of the covariance plane did not differ between healty adults and adults with ARCA-1. By extension, this would suggest that mechanical energy output would be similar between these groups even though a larger clearance distance was shown in the ARCA-1 group when compared to the healthy group, which is most likely not so. Instead, the orientation of the plane is most likely related to the phase difference between the shank and foot segments as suggested throughout this thesis and mathematically shown by Barliya et al. (2009).

These results support the hypotheses stated that ALAs are controlled by lower limb segment elevation angle phase mechanisms as shown in level walking (Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999) and that planarity is maintained across obstacle clearance conditions, supporting the view of an emergence of an adapted pattern from basic level walking control. These findings are highly significant because they suggest CNS control mechanisms for the implementation of ALAs.

The following study (Chapter 4) examined how individual phase shifts in fundamental harmonics of the thigh and shank segments contribute to the increase in phase difference between these segments as higher obstacles are cleared, and then use such information to determine if joint specificity is a mechanism of control in the CNS by relating the amplitudes and timings of these segment elevation angles to work done by the hip and knee joints. The examination of thigh and shank elevation angle peaks led to the conclusion that the phase difference between these segments was caused by an increased phase lead in the thigh segment in the leading limb and an increased phase lag of the shank segment in the trailing limb, supporting the hypothesis stated in Chapter 1. This builds on the results from Chapter 3 and shows two distinct mechanisms for increasing the phase difference in the leading and trailing limbs when stepping over obstacles. These separate mechanisms were attributed to differences in the initiation of swing between the limbs. Moderate relationships were shown between the work done by the K5 muscle power burst and maximum toe height, suggesting the knee contributes to limb elevation during obstacle clearance. The mechanical work done by the hip did not appear to be related to progression of the lower limb; a weak correlation was shown between work done by the H3 power burst and stride length as well as between work done by the H3 power burst and thigh elevation angle range. Although, a strong relationship was shown between the onset of the H3 muscle power burst and the reversal of thigh segment movement in the leading limb. These results suggested that the H3 power burst initiates thigh progression in the leading limb, but it does not provide the energy needed to increase the elevation range of the thigh segment. In the trailing limb, the H3 power burst was not synchronized with thigh elevation angle minimum peak. What does increase in each limb is the work done by the K5 muscle power burst, which is most likely responsible for the increase in thigh elevation angle range through intersegmental forces as proposed by McFadyen & Winter (1991) and Patla & Prentice (1995). In the trailing limb, the H3D power burst most likely assists the knee in forward progression of the thigh segment.

The results from this study suggests that the idea of joint specificity is not a control mechanism in the CNS since single functions are not associated with the hip and knee joints, therefore rejecting the hypothesis stated in Chapter 1. Instead, this thesis offers a different view as to how work done at the hip and knee joints is related to elevation angle control by the CNS and may subsequently change the way rehabilitation professionals focus their programs for patient populations. Also, these results support the conclusions of Chapter 3 and suggest that segment elevation angle phase difference may be one of the variables controlled by the CNS, with joint kinetics not being controlled directly but result from the control dynamics. This suggestion conflicts with previous work which suggests kinetics (forces and muscle activations) are directly controlled by the CNS rather than kinematics (originally shown by Evarts, 1968). The control of forces and muscle activations directly by the CNS would be a very complicated task. For example, inertia and the effects of gravity are constantly changing in a task like locomotion and these factors need to be accounted for in some way during such control. It is most likely due to these factors that we see no single goals associated with muscle power bursts at the hip and knee joints. As well, this may explain the differences in muscle power burst patterns between the leading and trailing limbs. Instead, this work suggests a segment elevation angle waveform mode of control which is simplified when compared to kinetic control.

One may now wonder how movement kinematics and kinetics are integrated into such a control mechanism. A main argument made in this thesis is that a basic locomotor pattern exists and ALAs are performed by adapting this basic pattern. It is suggested here that the basic locomotor pattern consists of segment elevation angle waveforms which when plotted together form a teardrop shaped loop in 3D space. This basic locomotor pattern is most likely stored within neural oscillators based within the spinal cord. When an ALA is required, higher CNS structures play a role in adapting the basic locomotor pattern. Widajewicz et al. (1994) stated that descending signals from the motor cortex may provide precise control over muscle activity and integrate this into the centrally generated pattern when adapting locomotion to the environment. Using this hypothesis, it can be suggested that such descending signals provide the muscle activity needed to adapt the phase and magnitude of segment elevation angles when the environment requires a change in locomotor pattern. One possibility of how this is accomplished is through muscle activation building blocks or motor primitives. Examination of the wiping reflex in the frog (Tresch et al. 1999) has shown that limb muscle activity can be broken down into a smaller number of activation components. A similar breaking down of muscle activation patterns has been shown in human locomotion (Ivanenko et al. 2004). Therefore, descending signals from higher CNS structures may act on spinal neural oscillators and dictate the required segmental phase shifting that is needed to successfully walk through an environment. The neural oscillators then send combinations of muscle activation patterns (or motor primitives) to generate the muscle forces which ultimately lead to phase changes in segment elevation angle waveforms. Although this is highly speculative and additional research is required, this provides a new perspective to control mechanisms during locomotion and when adapting locomotor patterns.

In the first two chapters of this thesis, locomotor control strategies were developed in healthy adults. Chapters 5 and 6 aimed at using these locomotor control theories to determine the mechanisms of coordination dysfunction for two pathological populations. Chapter 5 examined how a deficit such as ARCA-1 affected coordination of the leading limb when stepping over obstacles. The previously mentioned coordination analysis techniques were applied to the data and were subsequently used to suggest locomotor control mechanisms in this population. Leading limb segment elevation angles were found to display planar patterns for level walking and obstacle clearance. Since the presence of cerebellar ataxia had very little effect on the resulting segment elevation angle planarity, these results appear to support the conclusions by Morton et al. (2004) that during locomotor tasks, there is less involvement from the cerebellum. This may be due greater influences from the brainstem and spinal cord during locomotion with the cerebellum being important in balance control rather than segment coordination as suggested by Morton et al. (2004). This argument of brainstem control is further supported by work from Drew (1988) and Widajewicz et al. (1994) which highlighted the role of pyramidal tract neurons during ALAs in cats. Since the participants with ARCA-1 in this study had no brain structures other than the cerebellum affected by the disorder, this could suggest that the cerebellum does not have an effect on this control or that brain plasticity compensates for dysfunctional cerebellar areas. Although this argument was made, this study suggests some speculation that the nucleus interpositus located in the cerebellum may influence the observed planar patterns during locomotor adjustments since Orlovsky (1972) and Armstrong (1988) suggested this structure may function to fine tune the swing phase of the gait cycle. As seen in healthy individuals, systematic changes were shown in covariance loop width and covariance plane orientation and these plane characteristics were shown to be highly related to phasing between adjacent lower limb segments as shown in Chapter 3, therefore supporting the stated hypothesis. As well, the study in Chapter 5 supplements the results in Chapter 3 showing a non-linear relationship between thigh-shank fundamental harmonic phase difference and the width of the covariance loop, which is associated with a greater foot clearance over the obstacle.

The final study of this thesis examined differences in segmental control between the paretic and non-paretic leading limbs in a group of participants with a previous stroke during obstacle clearance using the planar law of intersegmental coordination. Planarity was maintained in the leading paretic and non-paretic limbs, supporting the hypothesis that a simplification of CNS control occurs as seen in healthy adults and adults with ARCA-1. Moreover, differences in segment elevation angle covariance plane characteristics were observed between the paretic and non paretic limbs post stroke, highlighting differing coordination strategies for each limb. Such differences between limbs and adaptations of the non-paretic limb have previously been reviewed in Olney & Richards (1996). Further analysis showed that a similar non-linear relationship between thigh-shank fundamental harmonic phase difference and covariance loop width occurred in the non-paretic limb as seen in participants with ARCA-1 in Chapter 5. Although this pattern indicated an increased phase difference between the thigh and shank segments on the non-paretic side, this change was not accompanied with an increase in foot clearance as seen in Chapter 5. Instead, this change in phase difference between the thigh and shank was related to greater flexion of the non-paretic limb during paretic limb support. This highlights a CNS control mechanism in the non-paretic limb to compensate for insufficient support in the paretic limb. Rehabilitation professions may be able to implement this information into exercise programs and provide patients with and alternate strategy to successfully clear obstacles if training in the paretic limb is unsuccessful.

The final two studies in this thesis do have limitations related to sample size and render them exploratory at this time. Each of these studies involves only a small number of participants (8 with ARCA-1 and 6 with a previous stroke). As well, each of these studies has missing data due to individual function and endurance issues seen in many pathological populations. Although this limits the statistical power in each of these studies, the preliminary results still suggest that people with these pathological conditions control locomotion through similar elevation angle mechanisms as seen in healthy adults. The participants in each of these studies had relatively mild forms of impairment due to the fact that the studies required individuals to walk and step over obstacles independently, which limits what results can be attributed to the severity of impairment. It is recommended that future studies use larger populations of participants with varying severity of impairments to determine how such factors contribute to locomotor control and the locomotor compensations discussed in this thesis.

Together, the results of this thesis support the theory that phasing between lower limb segment waveforms could be one of the variables controlled by neural oscillators in the CNS during human locomotion (Bianchi, Angelini, Orani et al., 1998; Lacquaniti et al., 1999). More importantly, this thesis suggests that similar mechanisms could be used by two specific pathological populations. If waveform phasing is a control variable during locomotion in the CNS, the current work suggests that the resulting kinetic patterns are not directly controlled by the CNS. Instead, they provide the energy to modulate segment elevation angle waveforms. As well, this thesis supports the theory that such waveform control is responsible for increases in safety margin when people with cerebellar ataxia step over obstacles. In addition, this thesis highlights how the planar law of intersegmental coordination can be used to identify segment phasing compensations in the non-paretic limb in participants with a previous stroke. This work therefore suggests a mechanism of control when implementing ALAs during human locomotion, which has not been examined previously.

7.2: Future Directions

The work described in this thesis can be developed further to determine fundamental control mechanisms during human locomotion and the results may have implications in rehabilitation such as in the development of prosthetic devices and for indicating measures of coordination for rehabilitation assessment.

A tool to help determine if segment waveform phasing is a control variable in the CNS would be computer modelling of human locomotion. For example, a biomechanical model could be developed in which the only control variables are phase differences between adjacent segment waveforms and waveform amplitude, with a constraint on the model being planarity of the resulting elevation angle waveforms. If such a model could successfully walk, this may provide further evidence that such inputs are control variables in the CNS. This model could also be tested on environments in which ALAs need to be implemented, such as obstacle avoidance, stepping up/down to a new level, or even recovery to an external perturbation to suggest if this type of control is utilized in these situations. Although this would not provide definitive proof of waveform control in the CNS, this would strengthen the argument that such control exists.

The motor control theories discussed in this thesis could also be used in the development of robotic prosthetic devices or lower limb exoskeletons to be used in rehabilitation. Robotic lower limb prosthetic devices which include ankle (below knee amputation) or knee and ankle (above knee amputation) joints can be developed using the idea of segment waveform control. Such prosthetic devices could mimic kinematic patterns of intact adults and motors placed at the knee and ankle joints could assist in generating the energy needed to produce the kinematic waveforms (as suggested in Chapter 4 regarding the resulting kinetic patterns during locomotion and obstacle clearance). As well, this idea could be extended to the creation of robotic exoskeletons to be used in rehabilitation to assist patients with mobility dysfunction in walking. The use of a robotic exoskeleton could help patients re-learn kinematic patterns used in healthy adults and maximize rehabilitation potential.

Finally, with more work on the topic, segment phasing could possibly be a measure used to determine coordination dysfunction in rehabilitation. The studies in this thesis showed that phase differences between the thigh and shank segments revealed compensations in people with cerebellar ataxia (to voluntarily increase foot clearance when stepping over obstacles) and a previous stroke (due to decreased support of the paretic limb when the non-paretic limb was in swing). It is possible that such information could be used by rehabilitation professionals to assist in the correction of human locomotion. Although sophisticated imaging techniques and data analysis would be needed for such a rehabilitation measure, the future may provide the technological advances for this idea to come to life.

In conclusion, this thesis explored mechanisms of locomotor control during level walking and the clearance of obstacles. The results of these studies support the possibility that during locomotion and when adapting locomotor patterns, lower limb segment postures may be controlled in the CNS with the observed muscle kinetic patterns resulting from this kinematic control. Furthermore, the results suggest that this control persists in participants with ARCA-1 and with a previous stroke, and voluntary compensations are highlighted in these populations. Future work on this subject may help in the development of robotic prosthetic devices and measures of coordination used in rehabilitation.

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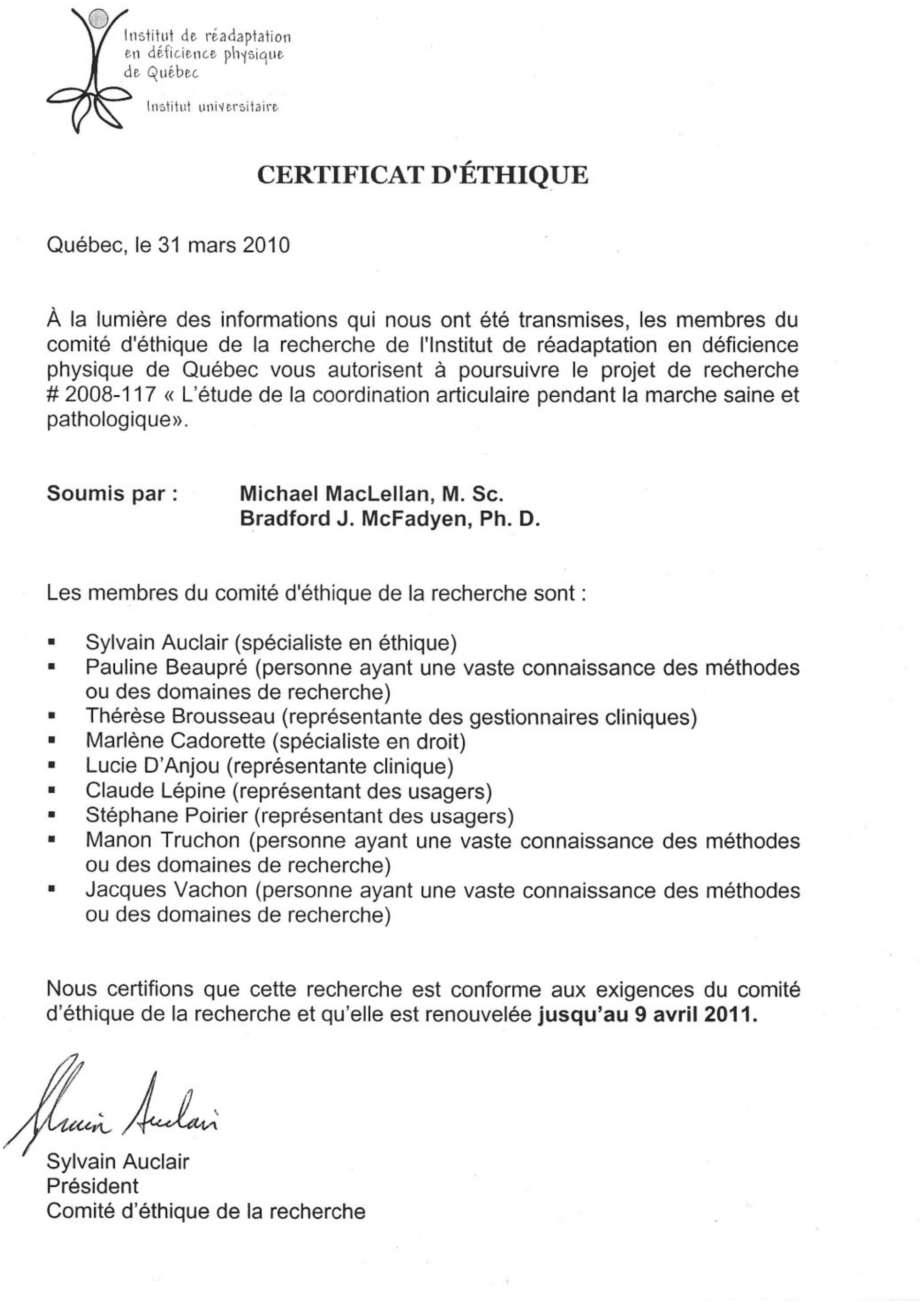
Appendix A

Certificates of Ethics and Consent Forms

Studies 1, 2 and 3



Institut de réadaptation
en déficience physique
de Québec

Institut universitaire

CERTIFICAT D'ÉTHIQUE

Québec, le 31 mars 2010

À la lumière des informations qui nous ont été transmises, les membres du comité d'éthique de la recherche de l'Institut de réadaptation en déficience physique de Québec vous autorisent à poursuivre le projet de recherche # 2008-117 « L'étude de la coordination articulaire pendant la marche saine et pathologique».

Soumis par : **Michael MacLellan, M. Sc.**
Bradford J. McFadyen, Ph. D.

Les membres du comité d'éthique de la recherche sont :

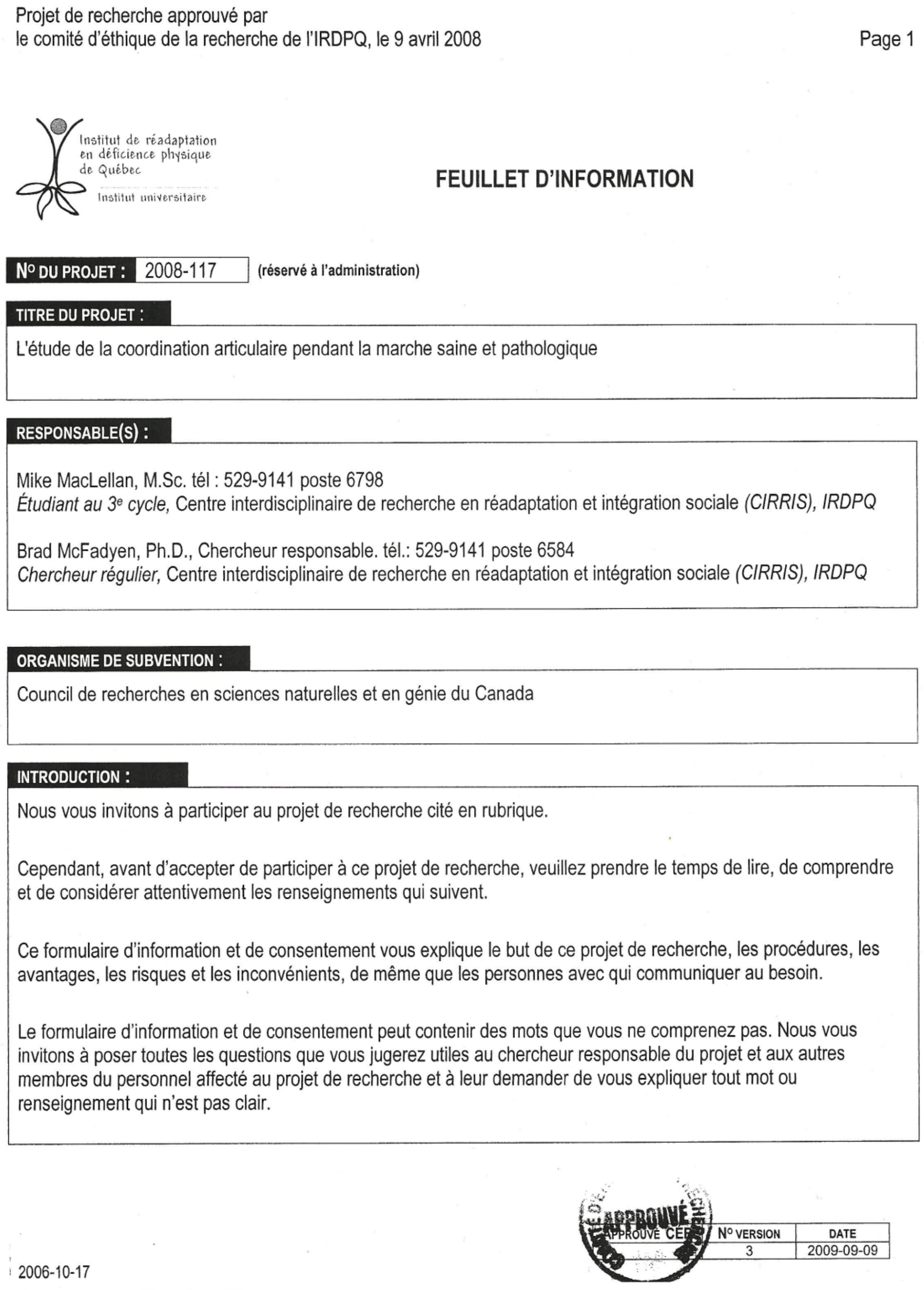
- Sylvain Auclair (spécialiste en éthique)
- Pauline Beaupré (personne ayant une vaste connaissance des méthodes ou des domaines de recherche)
- Thérèse Brousseau (représentante des gestionnaires cliniques)
- Marlène Cadorette (spécialiste en droit)
- Lucie D'Anjou (représentante clinique)
- Claude Lépine (représentant des usagers)
- Stéphane Poirier (représentant des usagers)
- Manon Truchon (personne ayant une vaste connaissance des méthodes ou des domaines de recherche)
- Jacques Vachon (personne ayant une vaste connaissance des méthodes ou des domaines de recherche)

Nous certifions que cette recherche est conforme aux exigences du comité d'éthique de la recherche et qu'elle est renouvelée **jusqu'au 9 avril 2011.**

Sylvain Auclair

Sylvain Auclair
Président
Comité d'éthique de la recherche

Studies 1, 2 and 3



Projet de recherche approuvé par
le comité d'éthique de la recherche de l'IRDPQ, le 9 avril 2008

Page 1

Institut de réadaptation
en déficience physique
de Québec
Institut universitaire

FEUILLET D'INFORMATION

N° DU PROJET : 2008-117 **(réservé à l'administration)**

TITRE DU PROJET :

L'étude de la coordination articulaire pendant la marche saine et pathologique

RESPONSABLE(S) :

Mike MacLellan, M.Sc. tél : 529-9141 poste 6798
Étudiant au 3e cycle, Centre interdisciplinaire de recherche en réadaptation et intégration sociale *(CIRRIS), IRDPQ*

Brad McFadyen, Ph.D., Chercheur responsable. tél.: 529-9141 poste 6584
Chercheur régulier, Centre interdisciplinaire de recherche en réadaptation et intégration sociale *(CIRRIS), IRDPQ*

ORGANISME DE SUBVENTION :

Council de recherches en sciences naturelles et en génie du Canada

INTRODUCTION :

Nous vous invitons à participer au projet de recherche cité en rubrique.

Cependant, avant d'accepter de participer à ce projet de recherche, veuillez prendre le temps de lire, de comprendre et de considérer attentivement les renseignements qui suivent.

Ce formulaire d'information et de consentement vous explique le but de ce projet de recherche, les procédures, les avantages, les risques et les inconvénients, de même que les personnes avec qui communiquer au besoin.

Le formulaire d'information et 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 responsable du projet 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.

APPROUVÉ
APPROUVÉ CÉR

N° VERSION	DATE
3	2009-09-09

2006-10-17

NATURE ET OBJECTIFS DE L'ÉTUDE :

Le but de cette étude est de mieux comprendre les stratégies de coordination de la hanche et genou pendant la marche sur terrain plat ainsi que pendant l'enjambement d'obstacles.

DÉROULEMENT DE L'ÉTUDE :

Si vous acceptez, vous participerez à une session d'expérimentation en laboratoire d'environ 3 heures. Dès que vous arriverez et que vous aurez donné votre consentement, il vous sera demandé de mettre les vêtements appropriés pour l'étude que vous aurez apportés (i.e., shorts, espadrilles, chemise à manches courtes). Nous prendrons votre poids, votre taille et les dimensions de différents segments corporels et nous vérifierons aussi votre vision à l'aide d'une charte. Des électrodes à lumières infrarouge seront collées sur votre tête, tronc, bassin, ainsi que sur vos cuisses, jambes et pieds pour enregistrer vos mouvements par un système de caméras infrarouges, ainsi que par une caméra vidéo.

La tâche demandée sera de marcher avec et sans obstacle à enjamber. La marche entravée comprendra des obstacles de trois hauteurs et trois longueurs différentes ajustées par rapport à la longueur de votre jambe (0 %, 10 %, et 20%) pour un total de neuf conditions. Vous devrez marcher à votre vitesse naturelle sur un parcours comprenant un obstacle ou non. Les périodes de repos seront données après chaque 5 à 6 essais ou quand vous les demanderez.

Si vous acceptez, nous vous demanderons à la fin de cette collecte de données de marcher pour environ 5 essais supplémentaires. Pendant ces essais supplémentaires, vous apportez un microphone pour enregistrer la voix. La tâche demandée sera de marcher à votre vitesse naturelle sur le même parcours sans obstacles et de nous dires les couleurs d'encre des mots qui seront présentées sur 5 écrans (mêmes mots sur chaque écran pour faciliter le regard pendant la marche) que nous ajouterons au plancher. Le but des essais supplémentaires est pour fournir de données préliminaires portant sur la division d'attention pendant la marche dans différentes populations.

Si vous êtes d'accord, les données seront envoyées au Dr Nicolas Dupré, neurologue, afin de transmettre les informations portant sur votre capacité de marche.

RISQUES POTENTIELS ET AVANTAGES POSSIBLES :

Vous pourriez trébucher pendant la marche comprenant un obstacle physique. Cependant quelqu'un restera auprès de vous lors de l'exécution de ces tâches.

Une légère fatigue pourrait survenir lors de l'expérimentation, mais vous pourrez vous reposer entre les essais, ou encore n'importe quand, à votre demande. Votre sécurité sera assurée par un responsable du projet, qui restera au laboratoire pendant toute la durée de l'expérience.

Il se peut que vous ne retiriez aucun bénéfice personnel de votre participation à ce projet de recherche. Toutefois, les résultats obtenus contribueront à l'avancement des connaissances dans ce domaine.

Si vous deviez subir quelque préjudice que ce soit à la suite de toute procédure reliée à ce projet de recherche, vous recevrez tous les soins requis par votre état de santé, sans frais de votre part.

APPROUVÉ CÉR	Nº VERSION	DATE
	3	2009-09-09

DROITS DU PARTICIPANT :

Votre participation est volontaire et vous avez le droit de vous retirer du projet à tout moment sans justification de votre part et sans que cela ne vous porte préjudice. Si vous décidez de ne plus participer à l'étude, vous n'aurez qu'à aviser les responsables du projet par téléphone.

QUESTIONS AU SUJET DE L'ÉTUDE :

Pour toute question concernant cette étude ou encore pour un changement inhabituel de votre condition (blessures, effets secondaires, etc.) vous pouvez contacter :

- Mike MacLellan, M.Sc., étudiant au doctorat
  (418) 529-9141 poste 6798
- Brad McFadyen, Ph.D., chercheur responsable du projet.
  (418) 529-9141 poste 6584
- Guy St-Vincent, M.Sc., coordonnateur du laboratoire.
  (418) 529-9141 poste 6116

Pour toute question d'ordre éthique, veuillez contacter :

- Johanne Trahan
  (418) 529-9141 poste 6036

COMPENSATION :

Si vous avez des frais encourus (ex. : déplacement, stationnement), une compensation (15$) vous sera accordée.

CONFIDENTIALITÉ ET UTILISATION DES RÉSULTATS :

La confidentialité des informations recueillies sera assurée et respectée. Votre nom sera remplacé par un système de codes numériques et seul ce système de code sera utilisé pour les présentations scientifiques. Si vous acceptez en signant un formulaire de consentement à cet effet, vos données et les informations seront conservées dans une banque de donéées permanente. La banque de donnée sera disponible, à la discrétion du Dr Brad McFadyen, pour des fins d'analyses futures. Tous les dossiers contenant des renseignements personnels et des images des participants seront conservés sous clé. Toutes les images vidéo enregistrées au laboratoire seront conservées par le responsable scientifique et serviront uniquement à la vérification du comportement moteur. Ils ne feront l'objet d'aucune référence extérieure au projet de recherche sans votre autorisation écrite, et le cas échéant, de votre tuteur légal. Les noms et les images vidéo seront effacées après une priode de 5 ans suivant la collecte de données.

APPROUVÉ CER	N° VERSION	DATE
	3	2009-09-09

Institut de réadaptation
en déficience physique
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Institut universitaire

FORMULAIRE DE CONSENTEMENT

Nº de projet : **2008-117**

Titre du projet : L'étude de la coordination articulaire pendant la marche saine et pathologique

Responsable(s) du projet : Mike MacLellan, MSc, Brad McFadyen, PhD

1) Le(la) responsable m'a informé(e) de la nature et des buts de ce projet de recherche ainsi que de son déroulement;
2) Le(la) responsable m'a informé(e) des risques et inconvénients associés à ma participation;
3) Ma participation à cette étude est volontaire et je peux me retirer en tout temps sans préjudice;
4) Les données de cette étude seront traitées en toute confidentialité et elles ne seront utilisées qu'aux fins scientifiques et par les partenaires identifiés au formulaire d'information;
5) J'ai pu poser toutes les questions voulues concernant ce projet et j'ai obtenu des réponses satisfaisantes;
6) Ma décision de participer à cette étude ne libère ni les chercheurs, ni l'établissement hôte de leurs obligations envers moi;
7) Je sais qu'aucune rémunération n'est rattachée à ma participation;
8) Le(la) responsable m'a remis un exemplaire du feuillet d'information et du formulaire de consentement;
9) J'ai lu le présent formulaire et je consens volontairement à participer à cette étude;
10) Je désire recevoir une copie des résultats de l'étude ☐ oui ☐ non
11) J'autorise les données à être envoyées au Dr Nicolas Dupré, neurologue ☐ oui ☐ non

Nom et prénom du sujet	**Date de naissance**	**Numéro de téléphone**
Signature du sujet *	Date	
Nom du chercheur	Date	Signature
Assentiment de la personne mineure (si possible)	Date	Signature

- Dans le cas de personnes mineures ou inaptes, remplacer la signature du sujet par celle du parent ou du tuteur

APPROUVÉ CER	Nº VERSION	DATE
	3	2009-09-09

J'accepte que les données collectées soient conservées dans une banque de données permanente. Je sais que les données seront dénominalisées, c'est-à-dire qu'elles seront identifiées par un code numérique. De plus, les images vidéo et mon nom devront être effacés après une période de cinq ans suivant la fin de l'étude. ☐ Oui ☐ Non

Nom du sujet Date Signature

Nom du chercheur Date Signature

Nom du témoin Date Signature

APPROUVÉ

[illegible]	N° VERSION	DATE
	3	2009-09-09

J'accepte que mon nom ainsi que mes coordonnées soient conservés dans la possibilité d'être contacté ultérieurement afin de participer à une autre étude du Dr McFadyen.

☐ **Oui** ☐ **Non**

_________________________ ______________ _________________________

Nom du sujet Date Signature

_________________________ ______________ _________________________

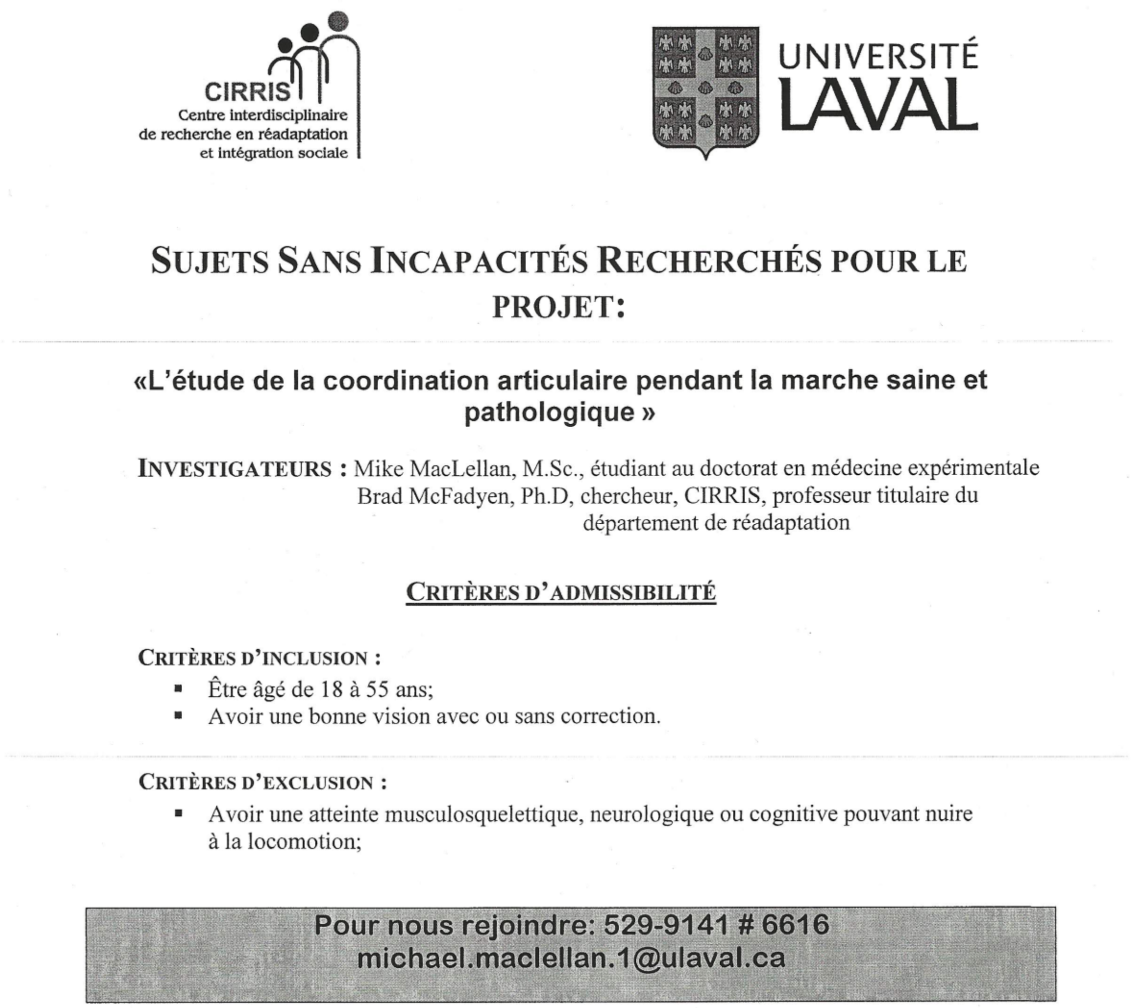
Nom du chercheur Date Signature

_________________________ ______________ _________________________

Nom du témoin Date Signature

APPROUVÉ CÉR	Nº VERSION	DATE
	3	2009-09-09

Studies 1, 2, and 3



CIRRIS
Centre interdisciplinaire
de recherche en réadaptation
et intégration sociale

UNIVERSITÉ
LAVAL

SUJETS SANS INCAPACITÉS RECHERCHÉS POUR LE PROJET:

«L'étude de la coordination articulaire pendant la marche saine et pathologique »

INVESTIGATEURS : Mike MacLellan, M.Sc., étudiant au doctorat en médecine expérimentale
Brad McFadyen, Ph.D, chercheur, CIRRIS, professeur titulaire du département de réadaptation

CRITÈRES D'ADMISSIBILITÉ

CRITÈRES D'INCLUSION :

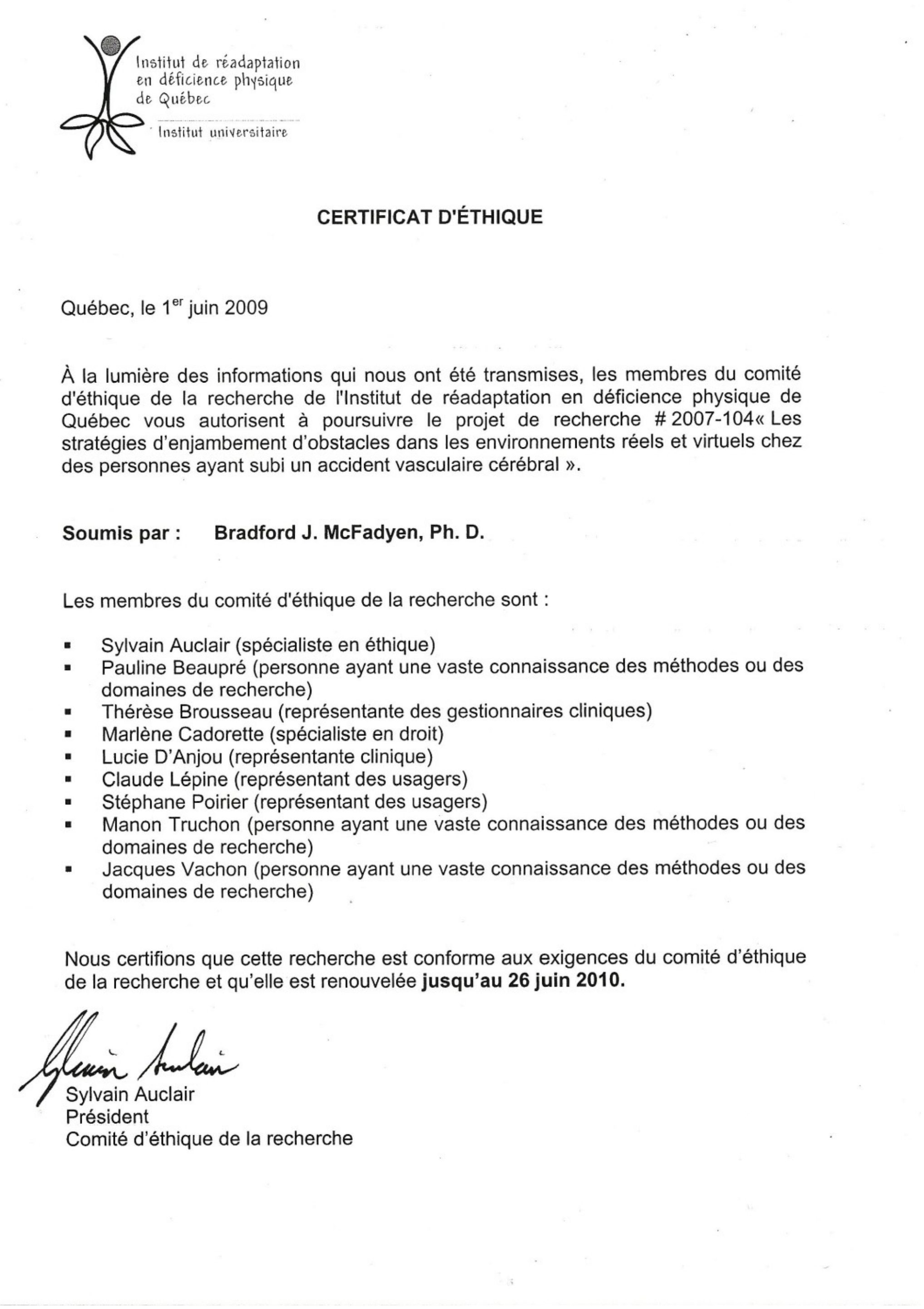
- Être âgé de 18 à 55 ans;
- Avoir une bonne vision avec ou sans correction.

CRITÈRES D'EXCLUSION :

- Avoir une atteinte musculosquelettique, neurologique ou cognitive pouvant nuire à la locomotion;

Pour nous rejoindre: 529-9141 # 6616
michael.maclellan.1@ulaval.ca

Study 4



Institut de réadaptation
en déficience physique
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CERTIFICAT D'ÉTHIQUE

Québec, le 1er juin 2009

À la lumière des informations qui nous ont été transmises, les membres du comité d'éthique de la recherche de l'Institut de réadaptation en déficience physique de Québec vous autorisent à poursuivre le projet de recherche # 2007-104« Les stratégies d'enjambement d'obstacles dans les environnements réels et virtuels chez des personnes ayant subi un accident vasculaire cérébral ».

Soumis par : Bradford J. McFadyen, Ph. D.

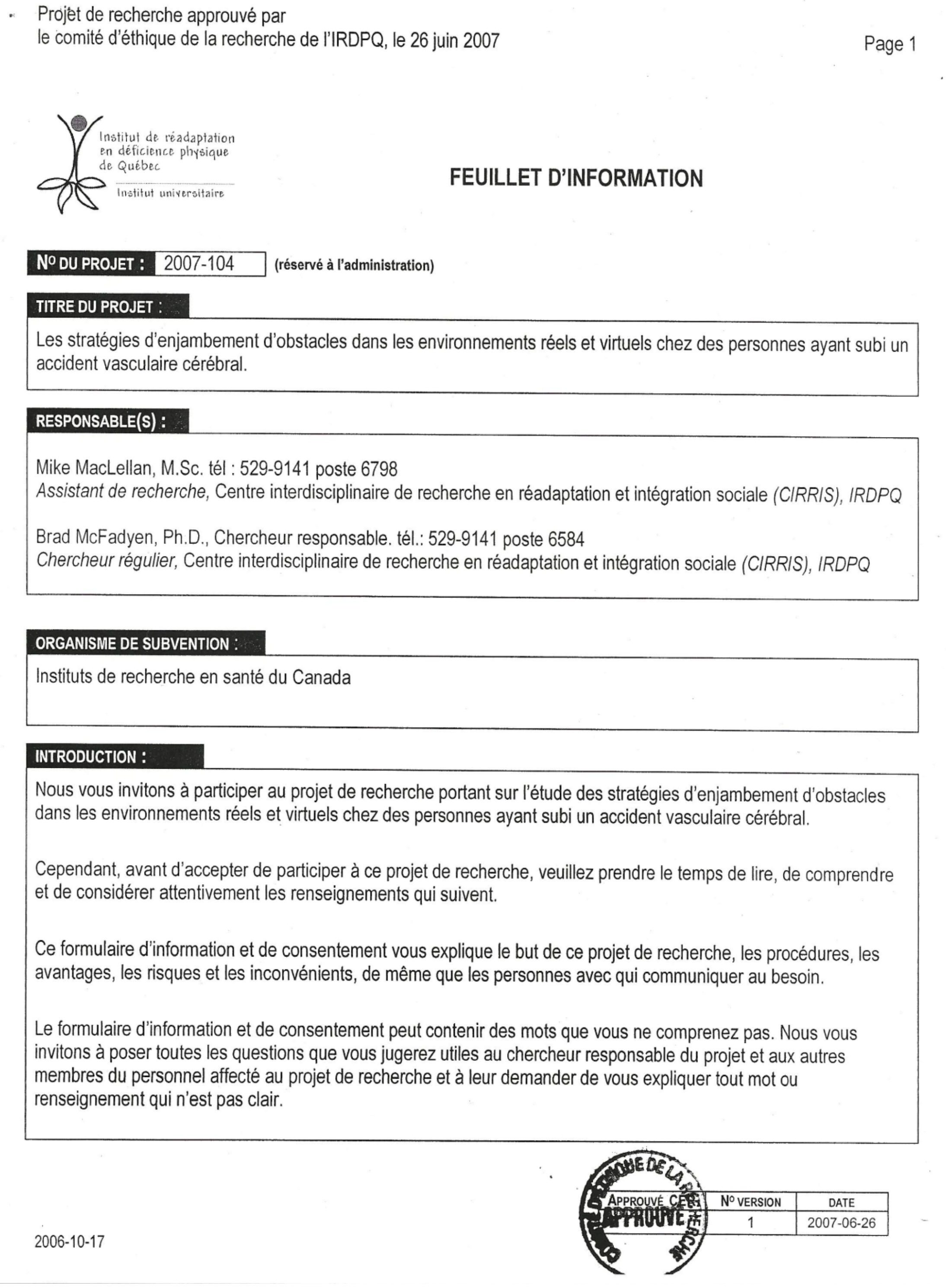
Les membres du comité d'éthique de la recherche sont :

- Sylvain Auclair (spécialiste en éthique)
- Pauline Beaupré (personne ayant une vaste connaissance des méthodes ou des domaines de recherche)
- Thérèse Brousseau (représentante des gestionnaires cliniques)
- Marlène Cadorette (spécialiste en droit)
- Lucie D'Anjou (représentante clinique)
- Claude Lépine (représentant des usagers)
- Stéphane Poirier (représentant des usagers)
- Manon Truchon (personne ayant une vaste connaissance des méthodes ou des domaines de recherche)
- Jacques Vachon (personne ayant une vaste connaissance des méthodes ou des domaines de recherche)

Nous certifions que cette recherche est conforme aux exigences du comité d'éthique de la recherche et qu'elle est renouvelée **jusqu'au 26 juin 2010.**

Sylvain Auclair
Président
Comité d'éthique de la recherche

Study 4



Projet de recherche approuvé par
le comité d'éthique de la recherche de l'IRDPQ, le 26 juin 2007

Page 1

Institut de réadaptation
en déficience physique
de Québec
Institut universitaire

FEUILLET D'INFORMATION

Nº DU PROJET : 2007-104 **(réservé à l'administration)**

TITRE DU PROJET :

Les stratégies d'enjambement d'obstacles dans les environnements réels et virtuels chez des personnes ayant subi un accident vasculaire cérébral.

RESPONSABLE(S) :

Mike MacLellan, M.Sc. tél : 529-9141 poste 6798
Assistant de recherche, Centre interdisciplinaire de recherche en réadaptation et intégration sociale *(CIRRIS), IRDPQ*

Brad McFadyen, Ph.D., Chercheur responsable. tél.: 529-9141 poste 6584
Chercheur régulier, Centre interdisciplinaire de recherche en réadaptation et intégration sociale *(CIRRIS), IRDPQ*

ORGANISME DE SUBVENTION :

Instituts de recherche en santé du Canada

INTRODUCTION :

Nous vous invitons à participer au projet de recherche portant sur l'étude des stratégies d'enjambement d'obstacles dans les environnements réels et virtuels chez des personnes ayant subi un accident vasculaire cérébral.

Cependant, avant d'accepter de participer à ce projet de recherche, veuillez prendre le temps de lire, de comprendre et de considérer attentivement les renseignements qui suivent.

Ce formulaire d'information et de consentement vous explique le but de ce projet de recherche, les procédures, les avantages, les risques et les inconvénients, de même que les personnes avec qui communiquer au besoin.

Le formulaire d'information et 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 responsable du projet 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.

APPROUVÉ CÉR
APPROUVÉ

Nº VERSION	DATE
1	2007-06-26

2006-10-17

NATURE ET OBJECTIFS DE L'ÉTUDE :

Le premier but de cette étude est de mieux comprendre les stratégies mécaniques pour franchir des obstacles physiques chez les personnes ayant subi un accident vasculaire cérébral. Le deuxième but est de comparer les comportements de marche avec obstacles dans les environnements réels et virtuels chez les personnes ayant subi un accident vasculaire cérébral et ce en comparaison avec les personnes en bonne santé.

DÉROULEMENT DE L'ÉTUDE :

Si vous acceptez, vous participerez à deux sessions d'expérimentation en laboratoire d'environ 2 heures chacune. La tâche demandée sera de marcher avec et sans franchissement d'obstacle. La marche avec franchissement d'obstacle comprendra des obstacles physiques de deux hauteurs différentes ajustées par rapport à la longueur de votre jambe (5 % et 15 %) pour un total de trois conditions (sans obstacle, avec un obstacle de 5 % et avec un obstacle de 15 %). Lors de la première visite, vous marcherez avec et sans des obstacles physiques. Lors de la deuxième visite, vous marcherez en portant un casque de réalité virtuelle dans lequel les mêmes conditions (avec et sans obstacles) seront présentées virtuellement. De plus, vous marcherez avec des lunettes cachant la vision périphérique semblable au casque de réalité virtuelle. Pour les deux visites, vous devrez toujours marcher à votre vitesse naturelle pour toutes les conditions pour environ 18 essais au total. Pour les sujets en bonne santé, les conditions seront répétées à une vitesse lente aussi, pour 18 essais supplémentaires. Un harnais sera toujours porté pendant l'étude pour éviter les chutes. De plus, des marqueurs seront collés sur diverses parties de votre corps pour enregistrer vos mouvements par un système de caméras spéciales, ainsi que par une caméra vidéo. Lors de la première visite, nous prendrons votre poids, votre taille et les dimensions de différents segments corporels. Nous vérifierons aussi votre vision à l'aide d'une charte. Enfin, un questionnaire sera administré pendant la première visite pour déterminer le niveau de menace que vous percevez par rapport à des obstacles de différentes hauteurs.

RISQUES POTENTIELS ET AVANTAGES POSSIBLES :

Vous pourriez trébucher pendant la marche comprenant un obstacle physique. Cependant vous porterez un harnais pour éviter une chute. De plus, quelqu'un restera auprès de vous lors de l'exécution de ces tâches.

Une légère fatigue pourrait survenir lors de l'expérimentation. De plus, il est possible que de légères nausées puissent survenir dû à l'utilisation du casque de réalité virtuelle. Cependent, vous pourrez vous reposer entre les essais, ou encore n'importe quand, à votre demande. Vous pourrez enlever le casque de réalité virtuelle aussi au besoin. Votre sécurité sera assurée par un responsable du projet, qui restera au laboratoire pendant toute la durée de l'expérience.

Il se peut que vous ne retiriez aucun bénéfice personnel de votre participation à ce projet de recherche. Toutefois, les résultats obtenus contribueront à l'avancement des connaissances dans ce domaine.

Si vous deviez subir quelque préjudice que ce soit à la suite de toute procédure reliée à ce projet de recherche, vous recevrez tous les soins requis par votre état de santé, sans frais de votre part.

APPROUVÉ CER	N° VERSION	DATE
	1	2007-06-26

DROITS DU PARTICIPANT :

Votre participation est volontaire et vous avez le droit de vous retirer du projet à tout moment sans justification de votre part et sans que cela ne vous porte préjudice. Si vous décidez de ne plus participer à l'étude, vous n'aurez qu'à aviser les responsables du projet par téléphone.

QUESTIONS AU SUJET DE L'ÉTUDE :

Pour toute question concernant cette étude vous pouvez contacter :

- Brad McFadyen, Ph.D., chercheur responsable du projet.
  (418) 529-9141 poste 6584
- Guy St-Vincent, M.Sc., coordonnateur du laboratoire.
  (418) 529-9141 poste 6116

Pour toute question d'ordre éthique, veuillez contacter :

- Johanne Trahan
  (418) 529-9141 poste 6036

COMPENSATION :

Si vous avez des frais encourus (ex. : déplacement, stationnement), une compensation forfaitaire (15$) vous sera accordée.

CONFIDENTIALITÉ ET UTILISATION DES RÉSULTATS :

La confidentialité des informations recueillies sera assurée et respectée. Votre nom sera remplacé par un système de codes numériques et seul ce système de code sera utilisé pour les présentations scientifiques. Si vous acceptez en signant un formulaire de consentement à cet effet, vos données et les informations seront conservées dans une banque de donéees permanente. La banque de donnée sera disponible, à la discrétion du Dr Brad McFadyen, pour des fins d'analyses futures. Tous les dossiers contenant des renseignements personnels et des images des participants seront conservés sous clé. Toutes les images vidéo enregistrées au laboratoire seront conservées par le responsable scientifique et serviront uniquement à la vérification du comportement moteur. Ils ne feront l'objet d'aucune référence extérieure au projet de recherche sans votre autorisation écrite, et le cas échéant, de votre tuteur légal. Les noms et les images vidéo seront effacées après une priode de 5 ans suivant la collecte de données.

APPROUVÉ CÉR	Nº VERSION	DATE
APPROUVÉ	1	2007-06-26

Institut de réadaptation
en déficience physique
de Québec
Institut universitaire

FORMULAIRE DE CONSENTEMENT

Nº de projet : **2007-104**

Titre du projet : Les stratégies d'enjambement d'obstacles dans les environnements réels et virtuels chez des personnes ayant subi un accident vasculaire cérébral.

Responsable(s) du projet : Mike MacLellan, MSc, Brad McFadyen, PhD

1) Le(la) responsable m'a informé(e) de la nature et des buts de ce projet de recherche ainsi que de son déroulement;
2) Le(la) responsable m'a informé(e) des risques et inconvénients associés à ma participation;
3) Ma participation à cette étude est volontaire et je peux me retirer en tout temps sans préjudice;
4) Les données de cette étude seront traitées en toute confidentialité et elles ne seront utilisées qu'aux fins scientifiques et par les partenaires identifiés au formulaire d'information;
5) J'ai pu poser toutes les questions voulues concernant ce projet et j'ai obtenu des réponses satisfaisantes;
6) Ma décision de participer à cette étude ne libère ni les chercheurs, ni l'établissement hôte de leurs obligations envers moi;
7) Je sais qu'aucune rémunération n'est rattachée à ma participation;
8) Le(la) responsable m'a remis un exemplaire du feuillet d'information et du formulaire de consentement;
9) J'ai lu le présent formulaire et je consens volontairement à participer à cette étude;
10) Je désire recevoir une copie des résultats de l'étude ☐ oui ☐ non

Nom et prénom du sujet	**Date de naissance**	**Numéro de téléphone**
Signature du sujet	Date	
Nom du chercheur	Date	Signature

COMITÉ D'ÉTHIQUE DE LA RECHERCHE
APPROUVÉ

Nº VERSION	DATE
1	2007-06-26

J'accepte que les données collectées soient conservées dans une banque de données permanente. Je sais que les données seront dénominalisées, c'est-à-dire qu'elles seront identifiées par un code numérique. De plus, les images vidéo et mon nom devront être effacés après une période de cinq ans suivant la fin de l'étude. ☐ Oui ☐ Non

Nom du sujet ______ Date ______ Signature ______

Nom du chercheur ______ Date ______ Signature ______

Nom du témoin ______ Date ______ Signature ______

APPROUVÉ

Approuvé CÉR	Nº VERSION	DATE
	1	2007-06-26

J'accepte que mon nom ainsi que mes coordonnées soient conservés dans la possibilité d'être contacté ultérieurement afin de participer à une autre étude du Dr McFadyen.

☐ **Oui** ☐ **Non**

Nom du sujet	Date	Signature
Nom du chercheur	Date	Signature
Nom du témoin	Date	Signature

APPROUVÉ

[illegible]	N° VERSION	DATE
	1	2007-06-26

Appendix B

Anthropometric Measurements

I. R. D. P. Q.
Laboratoire d'analyse de la motricité

Date : ____________ **No. du sujet :** ________________.
Nom : ______________________________________.
Dx : ______________________________________.
Evaluateurs : _________________________________.

Mesures Anthropométriques

masse : ____________ **Taille :** ____________.

1- PIED (talon - bout des orteils)

	DROIT	GAUCHE
longueur :	______________	________________.
p1 (malléoles) :	______________	________________.
p2 (arche) :	______________	________________.
p3 (métatarse) :	______________	________________.

2- JAMBE (centre artic. genou - mall. Ext.)

	DROIT	GAUCHE
longueur :	______________	________________.
p1 (plateau tibial) :	______________	________________.
p2 (triceps sural) :	______________	________________.
p3 (jambe dist.) :	______________	________________.

3- CUISSE (Gtroch - centre artic. genou)

	DROIT	GAUCHE
longueur :	______________	________________.
p1 (pli fessier) :	______________	________________.
p2 (mi-cuisse) :	______________	________________.
p3 (condyles) :	______________	________________.

4- TRONC (Acromion - Gtroch)

longueur : ______________.
p1 (mamelons) : ______________ **Larg1 (mamelons) :** ________________.
p2 (nombril) : ______________ **Larg2 (nombril) :** ________________.
p3 (hanches) : ______________ **Larg3 (hanches) :** ________________.

Commentaires : ____.
____.
____.
____.
____.

Appendix C

Marker Positions and Virtual Landmarks

PROJET: SUJETS CBL MARCHE AVEC OBSTACLE

MODÈLE AVEC 27 MARQUEURS

SEGMENTS: **marqueurs réels**

	GAUCHE		DROIT
PIED (triangles)	1	Arrière	16
	2	Milieu	17
	3	Avant	18
JAMBE (triangles)	4	Inférieur	13
	5	Milieu	14
	6	Supérieur	15
CUISSE	7	Condyle fém. latéral sup	10
	8	Milieu	11
	9	Proximal	12
BASSIN (triangle)	19	Pointe gauche	
arriere	20	Milieu	20
		Pointe droite	21
TRONC (triangle)	22	Pointe gauche	
arriere	23	Milieu	23
		Pointe droite	24
TETE (triangle)*	25	Pointe gauche	
arriere	26	Milieu	26
		Pointe droite	27

*Tête: Utilisation du support de casque

pts sondés

	GAUCHE		DROIT
PIED	28 (1)	Bout du pied	35 (8)
	29 (2)	Talon	36 (9)
	30 (3)	Métatarse	37 (10)
JAMBE	31 (4)	Malléole latérale	38 (11)
	32 (5)	Malléole médiale	39 (12)
CUISSE	33 (6)	Condyle fém. latéral	40 (13)
	34 (7)	Condyle fém. médial	41 (14)
BASSIN	42 (15)	Crête iliaque	44 (17)
	43 (16)	Épine iliaque antéro-sup	45 (18)
TRONC	46 (19)	Point glénohuméral	
TÊTE	47 (20)	Oreille	48 (21)