Adaptations to Stride Patterns and Head Movements During Walking in Persons With and Without Multiple Sclerosis

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ADAPTATIONS TO STRIDE PATTERNS AND HEAD MOVEMENTS DURING WALKING IN PERSONS WITH AND WITHOUT MULTIPLE SCLEROSIS

A Dissertation Presented

By

JEBB G. REMELIUS

Submitted to the Graduate School of the University of Massachusetts Amherst in partial fulfillment of the requirements for the degree of

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ADAPTATIONS TO STRIDE PATTERNS AND HEAD MOVEMENTS DURING WALKING IN PERSONS WITH AND WITHOUT MULTIPLE SCLEROSIS

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DEDICATION

Many times each day I give thanks for my wonderful wife and friend Tracy, for the time and love she gives, sharing her warm and open hearted approach to life. My grandfather Arvo Niemi saw this in her the moment they met and he told me to keep her close to me for as long as I can. Arvo, or “Itis” as my sister Eila and I called him as kids, was a great source of inspiration to me in his kindness, good nature and “one day at a time” amenability.
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ABSTRACT

ADAPTATIONS TO STRIDE PATTERNS AND HEAD MOVEMENTS DURING WALKING IN PERSONS WITH AND WITHOUT MULTIPLE SCLEROSIS

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Many people with multiple sclerosis (MS) have difficulty with walking, which can decrease their sense of mobility. Gait stability was investigated by studying stride parameters and head movements at preferred and fixed speeds in those with MS. First, walking gait data were recorded at preferred and fixed walking speeds from 19 individuals with MS and 19 controls. Traditional gait parameters were compared, as was swing foot to center of mass (CoM) timing at mid-swing. Second, walking gait data in healthy young adults (n=20) were recorded at preferred speed and while stepping over an obstacle. Study 2 developed novel swing definitions, measures of coordination between the swing foot and body CoM, and head movements as they pertain to field of view orientation during walking. Third, these novel measures were used to study the swing phase of walking in people with MS.

The first investigation revealed that the MS group walked with lengthened dual support times across all speeds, but shortened swing time and altered swing foot timing at fixed speeds in comparison to controls. Those with MS adopted a gait
strategy with increased dual support time, despite forcing changes to swing that may reduce gait stability.

In the second investigation, novel measures of swing showed alterations to phases of swing and in coordination between the swing foot and CoM under different gait tasks. This study also showed that the field of view was closer to the body during obstacle condition steps compared with unobstructed gait.

In the third study, these novel measures showed that at all speeds the MS group shortened early swing and lengthened mid swing while late swing remained unchanged compared with controls. Coordination measures illustrated adaptations in swing foot dynamics that may partially ameliorate altered swing foot timing. The MS group oriented the field of view closer to the body earlier in swing compared with controls. Those with MS have functionally adapted swing to increase time over the stance foot and rely more on visual perception, yet shorter early swing may afford fewer opportunities to plan a step or cope with gait disturbances while walking.
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GLOSSARY

**Anterior-posterior (A/P):** the fore and aft direction of the body in the sagittal plane.

**Walking gait cycle:** the period between consecutive right heel strikes.

**Stance phase:** the period when the right foot is in contact with the ground.

**Swing phase:** the period when the right foot is off the ground.

**Dual Support:** the period when both the right foot and left foot are in contact with the ground.

**Step-landing area:** the location on the ground where the foot is placed at the termination of the swing phase of locomotion.

**Physical dual support boundary:** A/P distance spanning from the heel of the trailing foot to the toe of the leading foot during the bipedal stance phase of walking gait.

**Physical unipedal boundary:** A/P distance spanning from the heel of the stance foot to the toe of the stance foot during the swing phase of walking gait.

**Physical boundary:** The anterior limit of the lead foot (toe) in dual support; this lead foot is the stance foot during the next swing phase of walking gait.

**Anticipated boundary:** A/P distance from the heel of the stance foot to the anterior limit of the upcoming step-landing area; referred to as heel-strike location.

**Center of Mass (CoM):** the point where the whole mass of the body is considered to be concentrated, a function of the mass and position of each of the body segments, modeled as rigid bodies.

**Center of Pressure (CoP):** the point where the pressure distribution under the feet is considered concentrated at a point.

**Optic flow:** the pattern of apparent motion of the environment in the visual field caused by relative motion between the observer and the scene.

**Frankfurt Plane:** a plane referenced to anatomical landmarks of the head that most nearly align with the surface of the earth during quiet stance, passing through the center of the vestibular rings (upper margin of the ear canal or porion) and the inferior limit of the orbit of the eye (orbitale).
Field of view limits in the sagittal plane: from the head neutral Frankfurt Plane (horizontal = 0°), the visual field subtends an angle of 110°, with 45° degrees above horizontal and 65° below horizontal.

Proximal Visual Intersect (PVI): the point on the ground where projection of the lower limit of the visual field meets the transverse plane at a given head orientation.

Exproprioception: a summary term for visual perception that combines proprioception (information regarding the position of the body's segments in space) and exteroception (information regarding the environment outside of the body).

Tau-gap (τ-gap): temporal boundary closure measure based on visual perception of optic flow induced by self-motion towards a boundary or feature of the environment.

Time to Contact (TtC): temporal boundary closure measure based on sensory perception of motion towards a boundary. Measured by computing distance to a boundary and velocity toward that boundary.

Motion-gap (m-g): physical distance to a boundary.

Coupling-gap ratio (K): the rate of closure of two temporal gaps as a ratio of one to the other.

Multiple sclerosis (MS): an immune-mediated neurological disease that causes damage to the myelin sheaths and neurons of the central nervous system.

Expanded Disability Status Score (EDSS): A rating of neurologic impairment developed by Kurtzke (1983)
CHAPTER I

DEVELOPMENT OF THE PROBLEM

Introduction

Walking is a rhythmic pattern of voluntary movements coordinated to transport the body overground. The old riddle "What walks on four legs in the spring, two legs in summer and three legs in the fall?" notwithstanding, the bulk of a human lifespan is spent locomoting bipedally. In order to walk in this manner, humans initiate an alternating pattern of stance and swing, where at least one foot is on the ground and each foot in turn is picked up and swung forward. When both feet are on the ground the perimeter of foot contacts describe a physical dual support boundary (Figure 1.1) and the movement of the body center of mass (CoM) within the dual support boundary is important to maintaining an upright posture during gait (Winter, 1983; Winter, 1987).

However, perhaps most important for the step-to-step persistence of gait is the proper reestablishment of foot contact that ends swing. During swing the body moves over the physical boundaries of the single stance limb of the heel and toe in the sagittal plane. Importantly, by early swing the step-landing area must be prospectively selected and the information necessary for picking a step-landing area is largely extracted from visual information (Patla, 1991; Marigold, 2008). The distance to a selected step-landing area determines step length and describes an anticipated boundary (Figure 1.1). This distance to the anterior limit of the anticipated boundary must be traversed in a timely manner by the swing-foot,
before the arrival of the CoM to keep the body upright and progressing with a predictable regular gait pattern.

![Diagram of physical and anticipated base of support (BoS)](image)

Figure 1.1. Physical and anticipated base of support (BoS). A physical BoS spans foot contact of a step; an anticipated BoS spans foot contact of a step that is forming.

During walking, CoM motion can be described in terms of the approach and crossing of the anterior limits of these two boundary configurations, physical and anticipated. While both feet are on the ground, the CoM is behind the toe of the leading foot (left foot, Figure 1.1) and inside the physical boundary. In swing however, the CoM travels beyond the toe of the stance foot, and the body enters an unstable equilibrium (Whittle, 1991). The unstable equilibrium of walking gait describes moments when the CoM is outside the anterior limit of the physical boundary but moving in a controlled manner. Ending swing quickly or decreasing step length (shortening the distance to the anterior limit of the anticipated boundary) can minimize the duration of this unstable state. Thus, the CoM is “recaptured” and is again held within a new physical boundary. The distance to the anticipated boundary (to the step-landing area) can be extracted post-hoc from the kinematic record of a gait collection in order to study the CoM and swing-foot approaches.
For a person walking, the location of a step-landing area is anticipated from visual clues embedded in the optic flow, as vision is the primary means of extracting a spatial representation of the external environment. Optic flow is the pattern of apparent motion of features in the environment caused by relative motion. Optic flow for sighted individuals is shaped by overall body movement, and compensatory head movements serve to functionally stabilize the visual field and allow for perception of stationary features on the ground (Gibson, 1958; Owen & Lee, 1986; Warren et al., 1986; Patla et al., 1999; Bradshaw & Sparrow, 2001; Berg & Mark, 2005; Reynolds & Day, 2005; Buckley et al., 2008; Marigold & Patla, 2008). Since the interaction with the ground is so important in step selection, the lower visual field is most relevant (Marigold & Patla, 2008). Where the limit of the lower visual field intersects the ground represents the last point in the environment visually perceptible when moving forward. For this reason, the portion of the environment inside the field of view proximal to the body may hold a special importance in the regulation of gait.

How the anticipated boundary is visually selected and how the CoM is recaptured within this new physical boundary is important for understanding the control of gait in healthy walkers, but may be even more important for those with chronic challenges to balance such as in persons with multiple sclerosis (MS). A means to quantify these spatio-temporal approaches to the physical and anticipated boundary approaches is to adopt the temporal boundary closure measures known as tau-gap (τ-gap) (Lee & Kalmus, 1980) and the closely related time-to-contact (TtC) (Slobounov et al., 1997) formulation. The concept of τ-gap and TtC are
attractive because they describe motion as actions yet relate these behaviors to a temporal margin that is directly perceptible. While $\tau$-gap was formulated to describe the visual perception of a boundary approach, TtC describes internally perceptible body features such as CoM motion towards physical boundaries of the feet (Haddad et al., 2006; Remelius et al., 2008). In this way, actions currently occurring can be perceived and actions can be shaped to realize the goal of movement. To better understand how actions may be prospectively guided in walking, $\tau$-gap can be used to describe the visual selection of an anticipated boundary location while TtC can be used to describe the transit of the CoM over the physical boundary and the closure of the CoM and swing-foot on the anticipated boundary. Furthermore, temporal boundary-related measures of stepping during gait may help in understanding changes that precipitate the breakdown of mobility and stability that often comes with chronic challenges to balance.

**Walking gait cycle**

To study locomotion in any population, it is helpful to describe walking gait on a cycle-by-cycle basis. A gait cycle is defined by the duration of a stride and the distance it carries us (Figure 1.2). A stride consists of all events between consecutive heel strikes of the same foot, typically the right foot. In this way, a stride starts with a stance phase (typically about 60% of a stride for walking) where the right foot is on the ground. A stride also contains a swing phase where the right foot is off the ground (typically about 40% of a stride for walking) and this is when the anterior limit of the anticipated boundary is hypothesized to be visually selected (Figure 1.1).
A distinction is made within stance where both feet are on the ground, called dual support, and where the physical boundary is formed. Dual support occurs at the beginning and middle of a gait cycle (Figure 1.2), once for each leg leading the body, and these periods are typically summed (totaling approximately 30% of the cycle). Slower gait speeds generally cause the percentage of stride time spent in dual support to increase, stride time to decrease and stride length to shorten (Larsson et al., 1980). Stride time and stride length are intimately entwined as both are dependent on the anthropometrics of the legs. A person’s legs have an intrinsic natural frequency of swing given their mass and length that loosely sets their preferred cadence (Winter, 1983). However, these stride characteristics are not fixed and can change over time and even change from stride to stride. This is an important feature of gait as the variability of gait patterns may relate to gait stability and flexibility. Gait patterns must be stable such that upright posture is maintained, and flexible in order to cope with unexpected perturbations that may be encountered during overground locomotion.

Figure 1.2. Stride definitions of walking gait (adopted from Uustal & Baerga, 2004).
Measures of gait stability are often based on cycle-to-cycle variability. Variability represents the nonuniformity of stride measures across strides and is typically reported through the standard deviation of such measures as stride time or stride length (Winter, 1983). However, increases in this form of gait variability are not exclusively linked to loss of stability, although it may be a signal that gait has become unstable. The association between variability and stability remains unclear partly because distinct signatures of instability are difficult to ascertain (Herman et al., 2005). Additionally, a moderate amount of variability in gait patterns is natural and healthy, whereas very low variability may also be an indicator of gait instability (Bruijn et al., 2009).

The most common spatial-temporal outcome measures of gait reported are the rate of anterior CoM displacement (velocity), and step or stride length. Additionally, stride width and the foot angle relative to average walking path direction can be reported. Generally, as gait speed decreases, stride length decreases, and stride width increases along with the external rotation of the foot. The reporting of additional variables depends on the scope of the data collected. Gait variables can also include the kinetics of ground reaction force under the foot or pressure distribution data. Gait reports commonly include measurements of swing-foot clearances above the ground and joint angles for the ankles, knees and hips (Saunders et al., 1953), or further up the segmental chain, pelvis, torso and head displacement and orientation. To create stable gait, the actions of these variables must emerge in a coordinated pattern. Segments of the body must move relative to one another and contribute to the efficient transport of the body as a
whole (Van Emmerik et al., 2005). Additionally, body movements must be patterned in order to enhance information pick-up by the sensory systems. For example, the feet must contact the ground on the plantar side of the foot to activate proprioceptive touch receptors, and the head must be appropriately positioned during each step for the pick-up of visual information. How gait is controlled has been investigated for many years, but without question, locomotion is guided to some degree by visual information in sighted individuals (Gibson, 1958; Hollands et al., 2002; Lee & Lishman, 1977; Marigold & Patla, 2008; Owen & Lee, 2004; Patla, 1997; Warren et al., 1986). A means to understand behavior driven by visual perception called τ-gap may help better understand the patterns of gait.

**Tau-gap descriptions of human motion**

Tau (τ) is a temporal measure consisting of a first order time to closure of any motion-gap forming between a moving organism and contact with a boundary (Figure 1.3) (Lee & Kalmus, 1980; Lee et al., 2001; Lee, 2005; Pepping & Grealy, 2007). A motion-gap is based on instantaneous position and approach motion towards a perceived boundary, and the τ-gap describes the distance to that boundary in time. The earliest τ-gap studies by David Lee sought to explain the braking behavior of automobile drivers as they approached obstacles but the most common examples are a hummingbird feeding, and a diving gannet bird folding its wings in the moments before contact with the water surface (Figure 1.3A & B; Lee & Lishman, 1977; Lee, 2009).
Figure 1.3. Motion-gap (m-g) between: A) humming bird and a feeder, B) diving gannet and the water, and C) coupling-gap between the hand (H) and the target (T) towards a goal (G) (Lee, 2009).

Features of the global movement of the body can be described in this way by the $\tau$-gap, but as the body is composed of many segments and these segments are coincidently moving in a coordinated manner, the motions of these elements must be coupled in some way. A coupling-gap can be used to describe how two motion-gaps interact. The coupling-gap can be used to characterize behavior between segments where a closure is scaled to match another target or opposing segment (Figure 1.3C; Equation 1.1). If the rates of closures of two motion-gaps ($x$ and $y$) remain in a constant ratio ($K$) over time ($t$), the gaps are coupled (Lee et al., 2001).

\[ \tau(x,t) = K \tau(y,t) \]  
Equation 1.1

In human posture research, the tau concept has been adapted as time-to-contact (TtC), and the boundary approached is the physical base of support where the anatomical edges of the feet are in contact with the support surface (Figure 1.4) (Slobounov et al., 1998). Postural adaptations described with TtC are somewhat different from tau, as TtC is not driven purely by visual perception of the motion-gap.
but by a direct perception of the movement of the body’s center of pressure (CoP) or CoM towards and away from the physical base of support (Slobounov et al., 1998). TtC was developed as a method to understand postural activity when both feet remain in place and has evolved from descriptions of quiet standing (Haddad et al., 2006) to behavior during more challenged postures (Forth et al., 2006; Van Emmerik et al., 2010), the postural phase of gait initiation (Remelius et al., 2008) and the CoM activity preceding a step due to a postural perturbation (Hasson, Van Emmerik, & Caldwell, 2008). However, TtC holds potential for use in more dynamic tasks, including locomotion where the stability boundary changes.

Figure 1.4. The fixed stability boundary typical of postural time-to-contact studies and the center of pressure trajectory (CoP) during quiet standing.

Approach to a new stability boundary, where the support of the body has been simplified as being centered at the CoP, has been described for the first step of walking. Austad & van der Meer (2007) computed τ-gap closure in seconds of the CoP when it was moving above 10% of its velocity peak in the first step of walking, and showed that closure rates change during early development. As a child becomes a more accomplished walker, closure of the τ-gap of the CoP near heel-strike result
in less forceful, “softer” touch contacts with the boundary, whereas in early development of gait initiation children tend to move the CoP resulting in “harder”, collision-like contacts with the BoS, but not so hard that they fall over. However, τ-gap analysis in the Austad & van der Meer (2007) study was limited to the first step.

Hof et al. (2008) reported on a modeling study of steady-state gait as it relates to contact of the CoM with the physical base of support (portrayed again by CoP). Hof and colleagues describe a measure called the extrapolated center of mass (XcoM), which represents a projection of the CoM forward in time based on its velocity. In this way, XcoM is very similar to velocity-based measures of TtC. This XcoM is used to describe the temporal margin between the anticipated future location of the CoM and the CoP. The size of the discrepancy between the XcoM and the CoP is hypothesized to relate to gait stability.

Hof’s XcoM is similar to a TtC depiction of the CoM behavior, except that the boundary that the XcoM is approaching is the CoP and not the physical base of support (Hasson et al., 2008). This distinction is important for gait as the CoP can continue to be positioned behind the CoM early in dual-support (after heel contact of the leading foot), since contact precedes weight loading and transit of the CoP under the new stance foot. Describing the τ-gap from the CoM to either an anticipated base of support, or the early portion of dual support, are not aspects of Hof’s model, but may be important to understanding gait stability. Furthermore, the addition of a τ-gap description of how an anticipated step landing is visually selected may bring the study of tau in gait closer to Lee’s original concept of a visually driven phenomenon. In this way the rate of visual closure that drives fishing
gannets to move from being “flyers” to “divers” may also apply in human locomotion where we move from being a “stander” to a “stepper”. Although changing modes during human bipedal motions may be subtler than the gannet’s full speed dives into the ocean, they may have more commonalities than meets the eye.

**Visual control of gait**

During gait, the head and optic centers are moving through space. To understand the visual system’s contribution to the control of movement including gait, David Lee (1980) implores us to investigate two things: the *service of activity* to which we are attending, and the *circumstance* in which the visual system is operating, namely the motion of the head. While the service of activity was described by Gibson (Gibson, 1979) as a primary kind of perception relating to the focus of what we are visually attending to, it may be highly variable (Oliva, 2003) but it is tied to the motion of the head, especially during fixations where vectors of the eye, head and gaze are closely aligned (Straumann et al., 1991). Motion of the head is easily described by motion-capture kinematic data and given that the service of activity during fixations is tied to head movement, a great deal of the information the visual system is using to control movement may be examined (Lee & Kalmus, 1980).

To better understand the circumstance in which vision is operating, kinematic recording of how the head is moving is a good place to start. During the dynamics of gait, head movement is occurring in part to perceive the environment beyond the limits of the visual field (Marigold & Patla, 2008). The field of vision is
constrained by the anatomical features of the eye orbits, but this limitation is
overcome by moving the head. By moving the head, more information on the
environment and body positioning is perceptible. The vertical limits of the visual
field in the pitch plane are approximately 110°, with around 45° above neutral and
65° below neutral (Figure 1.5) (Patla, 1997).

To make between-participant comparisons, the motion of the head can be
referenced to its neutral horizontal plane (0°). The neutral plane of the head
(primarily describing pitch) established at the World Congress of Anthropology in
Frankfurt Germany in 1884 is called the Frankfurt Plane (Figure 1.6 inset). The
Frankfurt Plane is referenced to anatomical landmarks that most nearly align with
the surface of the earth during quiet stance. The Frankfurt Plane is defined as
passing through the center of the vestibular rings (upper margin of the ear canal or
porion) and the inferior limit of the orbit of the eye (orbitale) (The American
heritage medical dictionary, 2007). Visual information below the anatomical limit of
the orbitale ridge (at around -65°) is inaccessible at a given head orientation, thus
there exists an intersection of the lower limit of the visual field and the ground. This
gaze/ground intersection is here introduced as the proximal-visual-intersect (PVI).
The behavior (position and motion) of the PVI along the ground may aid in
understanding the selection of step-landing areas and the closure of temporal gaps
between pauses in the PVI progression and the CoM or swinging foot in normal
overground locomotion. These relationships may be especially important for
individuals with compromised balance. In preliminary work, the PVI slows to nearly
zero velocity in early swing when near each anticipated step-landing area during
gait (Figure 1.7). It is hypothesized that these near-zero velocity pauses in the progression of the PVI along the ground predict a step-landing area or anticipated boundary.

Figure 1.5. Visual field limits during: A) quiet stance and level gaze, and B) stepping with the gaze pitched downward. Head neutral plane (Frankfurt Plane) represented at $0^\circ$ in A and $-30^\circ$ in B (adapted from Trevarthen, 1968).

Figure 1.6. Proximal visual intersect (PVI) with the ground (circled: location of local PVI velocity minima) during early-swing showing pause near step-landing area (lighter). Inset: Frankfurt Plane and a $-65^\circ$ declination describing the inferior limit of the lower visual field.
Figure 1.7. Anterior displacement of the proximal visual intersect (PVI) across the ground (thin line) and anterior PVI first derivative velocity (thick line) from an adult walking at preferred speed over 1.5 gait cycles. Gait events: right toe off (RTO), right heel-strike (RHS), left toe-off (LTO), and left heel-strike (LHS). Note minima in PVI velocity at A-D follows toe-off events.

It is known that visual information is an important aid in predicting a step-landing area, although visual proprioception may continue to guide step placement during swing (Berg & Mark, 2005; Bradshaw & Sparrow, 2001; Hollands et al., 2002; Marigold & Patla, 2008; Patla et al., 1999; Reynolds & Day, 2005; Warren et al., 1986). It is hypothesized that gait parameters are optimized at preferred speed to minimize head movements (Latt et al., 2008), yet motion of the head remains present when walking at preferred speed. Head pitch is presumed to be present to counter vertical head translation induced from a step (Hirasaki et al., 1999; Moore et al., 2001; Mulavara & Bloomberg, 2002), but head movement during gait, even when minimized at preferred speed (Pozzo et al., 1990), may have a functional role in the generation of gait events such as the initiation and termination of the swing phase of gait (Hirasaki et al., 1999). Much research has focused on the visual control...
of locomotion yet the phenomenon is not fully understood. Further investigation is necessary to determine the role of vision in step-landing selection and swing termination in overground walking. Additionally, the behavior of the PVI may serve as an indicator of the reliance of vision in the planning and execution of a step.

Classically, visual function has a two-fold division between visual exteroception and visual proprioception, but Lee (1980) proposed a summary term, exproprioception, that does not differentiate between these activities. Visual exproprioception is a synthesis of all available visual information originating outside the body (exteroception), and visual proprioception relating to available information regarding body positioning. The lower portion of the visual field is important to people walking over irregular terrain, but in the case where internal proprioception of the lower limbs is diminished due to neurological damage, it is presumed that exproprioception from the lower visual field has heightened importance.

**Multiple sclerosis**

MS is a chronic disease where an immune-mediated response damages the myelin sheath that surrounds neurons of the central nervous system (CNS). In MS, CNS damage is typically progressive or mixed with remissive periods between demyelinating episodes. During a remission period, symptomatic levels generally decrease and level off as scar tissue (sclerosis) forms at sites where the body has imperfectly repaired myelin sheaths damaged during an exacerbation (Noseworthy et al., 2000). The onset of MS is generally between 20 and 40 years of age and
women are 2-3 times as likely to contract MS as men at a young age, although the incidence in men and women is nearly equal for those diagnosed with MS later in life. In the United States, there are 400,000 people living with MS and an additional 200 people per week are being diagnosed with MS, yet the cause(s) of MS are unknown (Heesen et al., 2008).

The most common tool used to describe disability due to MS is the Expanded Disability Status Scale (EDSS) score (Kurtzke, 1983), a measure that focuses on walking ability and functional systems of the CNS. Individuals with MS consistently rate walking and vision as the most important factors in maintaining a good quality of life (Heesen et al., 2008). Fortunately, individuals with MS can typically walk with no clinical disability for many years after diagnosis and over the course of the disease, less than one third of people with MS actually lose the ability to walk. Functional gait is prominently featured in the scoring of quality of life measures (Krupp et al., 1989) and common disability assessment tools that heavily weight unassisted and assisted walk distances (Kurtzke, 1983). Studies of physical limitations due to MS typically attempt to correlate disability with mobility and physical activity (Motl & Snook, 2008; Snook & Motl, 2009), and although research on movement dysfunction in MS has begun, relatively little is known compared to aging, stroke or other neurological conditions such as Parkinson’s disease.

Many of the common signs of MS can negatively affect the dynamics of gait. These signs include proprioceptive loss, ataxia (discoordination), vision disturbances, altered sensory system function, and decreased muscle strength, which can all affect balance (DeLisa, 1985; Kelleher et al., 2009; McAlpine, 1985;
Additionally, symptoms of MS can include fatigue and depression (McAlpine, 1985). However, altered balance or vision problems are often the initial symptom that has brought an individual to seek medical advice that leads to a diagnosis of MS (Calabresi, 2004). Even when there is no clinical gait disability apparent, self-reported changes in balance and changes in laboratory based measures of gait patterns can be observed. These changes to gait exhibited by people with MS include: slower walking speed, shorter steps and a longer dual support phase, reduced ankle range of motion, and increased hip flexion (Benedetti et al., 1999; Gutierrez et al., 2005; Kelleher et al., 2009; Martin et al., 2006).

Unfortunately, it remains unclear to what degree changes in gait patterns reported are caused by disability due to MS or simply reduced gait speed. For example, in healthy populations, power law relationships exist between gait speed and stride length, and gait speed and stride time while linear relationships exist between stride length and step time (Larsson et al., 1980), but these relationships have yet to be confirmed in the MS population.

**Statement of the problem**

The purpose of the proposed studies is to better understand the visual regulation of human gait as measured by $\tau$-gap referenced to head motion, the regulation of swing and CoM transit over the feet as measured by TtC, and in particular the changes in gait patterns brought on by alterations in gait speed and disability due to the neurological disease multiple sclerosis. This will be accomplished by characterizing the temporal approach of the body to a step-landing
area with \( \tau \)-gap, and CoM approach to the physical base of support and the anticipated base of support with TtC while walking at preferred speed, and across a range of speeds between fast and slow. Preferred speed is significant as it reflects what may be occurring during daily living, and setting gait speeds allows for direct comparisons in the case that preferred speeds are different between groups. It is important to understand how the CoM is recaptured within the new physical base of support formed at the point of heel contact. The CoM recapture within the physical base of support is dependent on two things: (1) the step-landing area being accurately predicted and (2) the swing-foot reaching the anticipated base of support before the CoM. The control of the swing movement and the selection of where and when to terminate swing to form a new physical base of support is central to keeping the body stable and upright and maintaining the anterior CoM progression of gait.

The CoM approach to the physical and anticipated base of support is described by the TtC variable. During the approach of the CoM to the anticipated base of support, keeping the equilibrium of gait intact depends on the motion of the swing-foot and a coupling ratio (K) will be used to describe how the CoM and the swing-foot are coordinated in an approach to the anticipated base of support. A method termed the proximal-visual-intersect (PVI) is proposed to understand how vision is used to select anticipated base of support locations.
Significance of studies

Head motion allows for flexibility in orienting the visual field during dynamic movements, but describing the outcome of coordinated translations and rotations of the head can be unwieldy. The proximal visual intersect technique for studying visual field orientation simplifies complications of head motion by presenting data in a single transverse plane position. Describing the τ-gap between the PVI and the body illustrates how steps are formed in healthy populations. Similarly, TtC descriptions of motion toward step-landing areas can further illustrate the dynamics of step formation. By relating movement of the body toward the PVI with τ-gap, and relating movement of the CoM and swing the foot toward the step-landing area with TtC, a characterization of the dynamics of gait can be presented that has been lacking in the field of biomechanics and motor control. Furthermore, the behavior of τ-gap, and TtC may reveal strategies of boundary approach that can aid in the understanding of the breakdown of stability and mobility in balance impaired populations.

Two common departures from normal patterns of gait are cautious or reckless gait indicating slow careful walking or bold aggressive walking styles. A clear indicator of gait strategy may be gained by measuring the length of τ-gaps in walking and how TtC margins are coupled between the CoM to the anticipated base of support and between the swing-foot and the anticipated base of support. When the foot arrives well before the CoM, a safer approach to stepping is indicated. More careful step formation may be associated with disability due to MS and greater
caution taken in walking. Alternatively, the closer these two TtC margins are to being closed simultaneously, the more reckless the gait strategy. A reckless gait may indicate less attention being paid to the environment.

How these τ-gaps and TtC coupling ratios change when walking at slow, medium, and fast gait speeds in younger healthy adults allows for the creation of a response profile of CoM temporal gap closure. Next, how these τ-gaps and TtC coupling ratio patterns appear in a matched speed profile in a group of adults and a group of mildly to moderately affected individuals with MS will further illustrate the role of temporal gap closure in the formation of steps during walking in people with challenged balance. Finally, the influences of changes in walking speed and disability in mild-to moderately impaired individuals with MS may be better separated by manipulating gait speed in the MS group and a group of age matched peers.

**Assumptions**

- Kinematic markers accurately represent underlying 3D skeletal motion.
- Rigid body model assumptions hold true.
- The walking path in the lab resembles natural overground walking conditions.
- Individuals with low levels of disability due to multiple sclerosis represent the general population of individuals living with MS at similar disability levels.
- The participants accurately report sensitivity thresholds during sensory system evaluation.
Hypotheses

Study 1: Development of tau-gap control methodology

The purpose of this study is to describe how step-landing areas are visually selected during the movement of the proximal visual intersect by adapting the temporal boundary approach measure known as \( \tau \)-gap to the dynamics of gait. Once an understanding of how the anterior limit of the anticipated base of support is formed by the step-landing area selection, approaches of the body CoM towards the physical and anticipated base of support locations that form during each step will be studied by using a temporal gap-closure technique called time-to-contact. Additionally, during the closure to the anterior limit of the anticipated base of support a coupling ratio is explored between the step-landing area, the CoM and the swing-foot. These characterizations of gait are to be made from the analysis of gait patterns typical to young healthy adults walking at their preferred speed. To understand how the visual selection of step-landing areas may be influenced by features in the environment, participant will be asked to walk at preferred speed and step over an obstacle in their path.

Study 1 will allow development of the \( \tau \)-gap and TtC measures for studying the dynamics of gait:

1.1. At preferred speed determine the \( \tau \)-gap between the velocity minima in anterior progression of the proximal-visual-intersect and CoM arrival at the anticipated base of support.
1.2. At preferred speed determine the rate at which the TtC between the CoM and the physical base of support is closing after mid-swing.

1.3. At preferred speed determine the rate at which the TtC between the CoM and the anticipated base of support is closing.

1.4. At preferred speed determine (1) the magnitude and stability over time of the coupling-gap ratio between the CoM to anticipated base of support gap and the swing-foot to the anticipated base of support, and (2) whether the coupling-gap ratio will favor the swing-foot arriving before the CoM.

1.5. At preferred speed while stepping over an obstacle determine how the \( \tau \)-gap between the velocity minima in anterior progression of the proximal-visual-intersect and CoM arrival at the anticipated base of support are different from unperturbed step-landing area anticipation.

Study 2: Tau-gap control adaptations to changes in walking speed

The purpose of this study is to discern how patterns governing the formation of steps during walking described in Study 1 change when the same group of participants varies gait speed. This is important for understanding how motion patterns change when walking at different speeds, as changes in walking speed frequently occur with alterations in the health of the locomotor system. The influence of gait speed on the selection of step landing area can be related to head pitch and the manner in which the body is carried. Changes in approach of the body CoM towards the physical and anticipated base of support locations across a range of gait speeds will allow for descriptions of how non-impaired people typically alter
gait from their preferred rate of locomotion. Additionally, a profile of coupling ratios across gait speeds between the anticipated base of support, the body CoM and the swing-foot itself may reveal how changes in gait speed affect the ability to form a physical base of support with the swing-foot before the CoM arrives.

Study 2 will examine the following hypotheses:

2.1. The anterior progression of the proximal-visual-intersect will reach a minimum velocity near the anticipated base of support, which will be closer to the body during slow gait and farther from the body during fast gait

2.2. The τ-gap between the CoM and the anticipated base of support will be shorter with increasing speeds and longer with slower speeds.

2.3. The TtC temporal gap between the CoM and the physical base of support will close earlier in the swing phase at faster speeds, and later in the swing phase at slower speeds.

2.4. The coupling-gap ratio between the CoM to anticipated base of support gap and the swing-foot to the anticipated base of support gap will favor the swing-foot arriving before the CoM at all speeds. However the coupling ratio will be higher at slow speeds, indicating earlier arrival of the swing-foot relative to the CoM, and lower in fast speeds, indicating that the foot arrives at nearly the same time as the CoM.
Study 3: Spatio-temporal patterns of walking at preferred and set speeds in people with MS

The purpose of this study is to examine gait patterns of individuals with low to moderate levels of disability due to MS in comparison to individuals without MS while walking at preferred, slow, medium and fast speeds. As individuals with MS often walk with a different gait speed than their healthy counterparts, this study allows direct comparisons of gait kinematics without the confounding influence of differences in gait speed.

Study 3 will examine the following hypotheses:

3.1. Preferred walking speed will be similar in individuals with MS in comparison to those without MS as seen in preliminary results but contrary to existing gait reports in MS (Benedetti et al., 1999; Martin et al., 2006).

3.2. An increase in stride length and a decrease in the dual support portion of walking will be observed as gait speed increases in both groups (Larsson et al., 1980). Both groups will have linear responses to gait speed increases in stride length, stride time and dual support. However, the slopes of these regressions will show that MS favor longer dual support periods in comparison to the control group (Benedetti et al., 1999).

Study 4: Tau-gap control of walking at preferred and set speeds in MS

The purpose of this study is to examine how τ-gap closure measures developed in studies one and two on young healthy adults regarding the PVI and CoM approach to physical and anticipated base of support locations describe
walking patterns in older healthy adults and older adults with mild-to-moderate disability due to MS. Examining τ-gap patterns across a range of gait speeds will allow for descriptions of how aging and balance impairment may alter gait and trunk/head orientation without the influence of potentially different preferred walk speed. Patterns of boundary approach may reveal how disability due to MS affects how a physical boundary is formed with the swing-foot before the CoM arrives. Finally, the influence of gait speed on the selection of step landing area can be related to head pitch and the manner in which the body is carried.

Study 4 will examine the following hypotheses:

4.1. The anterior progression of the proximal-visual-intersect will reach a minimum velocity near the anticipated base of support, which will be significantly closer to the body in the MS group compared to the control group even at matched step lengths across the speed profile.

4.2. The τ-gap between the CoM and the anticipated base of support will be shorter across the speed profile in the MS group indicating a more cautious boundary approach than that of the control group.

4.3. The TtC temporal gap between the CoM and the physical base of support will close later in the swing phase across the speed profile in the MS group indicating a more cautious boundary approach than that of the control group.

4.4. The coupling-gap ratio between the CoM to anticipated base of support gap and the swing-foot to the anticipated base of support gap will favor the swing-foot arriving before the CoM at all speeds for both groups. However, the
coupling ratio will be higher at all speeds for the MS group, indicating earlier arrival of the swing-foot relative to the CoM.

4.5. Individuals with MS will exhibit greater trunk flexion and greater head pitch during all speeds in comparison to controls due to a more cautious gait pattern.

Summary

There have been many studies of human locomotion and great insights have been gained on how gait patterns are formed. However, the τ-gap and TtC have had limited use in descriptions of human walking even though they have been shown to be powerful tools in describing a multitude of human and animal behaviors. The TtC measure adapted here can describe a temporal closure with a physical boundary, and τ-gap can describe closures with a visually anticipated boundary, which occurs during each step of walking gait. Therefore, an exploration of how temporal closures are controlled and coupled may explain a locomotor control mechanism that the literature has lacked. The application of this work to the MS population, a group with potentially serious difficulties maintaining balance (Chung et al., 2008) and coordination (Remelius et al., 2008), provides insights into the breakdown of mobility and stability and their increased risk of falls (~52%) (Nilsagard et al., 2009; Peterson, 2007).

Another primary goal of this research is to provide data on speed matched gait patterns of individuals with MS that has been lacking in the literature. When it can be determined what the speed independent changes are that people with MS
experience, biomechanical variables including stride length, dual support time, lower limb joint range of motion and upper body segmental coordination can be related to disability instead of being confounded by gait speed. The manner in which people with MS adapt to changes in gait speed as compared to individuals without MS is important for understanding potential pathways to instability and maintaining mobility. When these functional adaptations are quantified, these findings can be applied to the development of gait training protocols that feature a reliance on vision or coordination to improve stability and reduce the high incidence of falls in those with MS.

The methods developed here may also be used to better understand running gait as well by deriving more meaningful summary measures from upper body dynamics such as the PVI in contrast to simply factoring in the mass and position of the head for anthropometric summary measures adding to more accurate CoM quantification. The PVI can continue to be utilized in psychological studies of head orientation in relation to mood and emotional state, as the distance in front of the body that one is looking may have direct correlation to the mood of the person being studied.
References


CHAPTER II
LITERATURE REVIEW

Introduction

Gait is one of the most common of all human movement patterns but it is also very complex. Despite variability in gait patterns from individual to individual and across a single person’s lifespan (Beauchet et al., 2009; Kang & Dingwell, 2008; Kirtley et al., 1985; Saunders et al., 1953; Winter, 1983; Winter, 1987), generally uniform patterns typify healthy gait. Departures from established gait parameter norms (Larsson et al., 1980) can signify problems with the balance, coordination and stability necessary to maintain the mobility afforded by gait (Benedetti et al., 1999; Crenshaw et al., 2006; Gutierrez et al., 2005; Martin et al., 2006; Remelius et al., 2008). The basis of traditional gait analysis is to report a series of spatio-temporal variables such as gait speed, stride length and stride time, or joint angle patterns of the lower limbs, although research has begun to focus more on upper body kinematics. However, many of these earlier studies only report the results of adaptations to gait and must speculate as to the underlying reasons the altered patterns emerge. Methods that describe temporal closures can be used to understand how the movements of gait are prospectively guided with vision, by a measure known as tau-gap (τ-gap), or with multiple sources of sensory perception in a measure called time-to-contact (TtC). These approach estimates may facilitate descriptions of how gait patterns are controlled and why outcome measures of gait are altered when facing challenges to balance and stability.
This review seeks to first describe the patterns of healthy gait, and how temporal closure may underlie control of those patterns. Once the visual control of stepping during gait is understood in healthy young individuals, the changes in gait patterns that can occur when coping with disability due to multiple sclerosis (MS) will be evaluated. In order to understand changes in gait due to disability, it is necessary to be familiar with what is causing that disability and this review provides a detailed depiction of the nature of MS, disability due to MS and how custom tailored gait interventions for individuals with MS have succeeded in normalizing gait patterns.

**Human walking gait**

The human walking gait serves to transport the body across the ground in an upright fashion at speeds typically ranging between 0.5 and 1.5 m/s. Gait patterns are described in terms of gait speed, the length of a stride, and the duration of the gait cycle. Gender-specific trends have been identified that link gait speed, stride length and stride time (Larsson et al., 1980). As gait speed increases, stride length and stride rate increase (Figure 2.1A, B). Additionally, as stride duration increases (or cadence slows) more time is spent in both stance and swing but stance times increase disproportionately in comparison to swing times (Figure 2.1C). Additionally, as walking speed increases, other features of gait change as well, including narrower stride widths, smaller foot adduction, and smaller lateral sway of the CoM (Danion et al., 2003). To understand stride patterns it is common to normalize events to 100% of the stride duration between subsequent footfalls of the
same foot (Figure 1.2). In this way, temporal patterns can be compared across a range of gait speeds or cadences. Normalized stride duration shows that the composition of a stride changes with gait speed when the percentages of stance and swing are compared (Murray et al., 1984).

Figure 2.1. Correlations in gait patterns of healthy men and women, ±1 SD: A) velocity vs. stride length, B) gait velocity vs. cadence, C) stride duration vs. time spent in stance (ST), swing (SW) and dual support (DS). Note horizontal (velocity) log scale in A&B (adapted from Larsson et al., 1980)

Individuals are often compared while walking at their preferred gait speed, but walking kinematics can appear quite different in a typical gait report simply because of differing preferred gait speeds (Larsson et al., 1980). One means to eliminate potential speed differences is to set gait speed or cadence or both. Gait speed can be set in overground gait studies with verbal instruction or on a motorized treadmill by belt speed. An alternate approach to minimize stride rate differences is to employ a metronome to set stride frequencies (cadences) (Winter, 1983). In this way, comparisons of gait patterns at comparable stepping rates are possible. Cadences of walking typically range from 80-120 steps/min. In this range, people increase gait speed by equally increasing cadence and stride length in a linear fashion (Kirtley et al., 1985; Larsson et al., 1980; Murray et al., 1984). Fixing
cadence is a frequently employed technique to normalize gait, however even with set cadences the gait speed of an individual remains dependent on leg length, as people typically step with a stride length set by their anthropometrics.

Leg length, cadence and, as a result, preferred gait speed vary from person to person making comparisons between individuals challenging. Scaling biomechanical data to an individual’s body mass, height or leg length can reduce between subject variability. Since preferred step frequency and step length are to a large degree set by an individual’s leg length, scaling is common but makes results difficult to interpret as variables become unit-less or individual based. For example, scaled gait speed is often reported in statures, or body heights traveled, per second. Scaling to anthropometrics is an aspect of the science of allometry, which can relate movement to energy costs with a power law (Minetti et al., 1994).

Even when using techniques to scale data, the paradox of gait is that the more that it is studied, the more complex it appears. What makes for stable gait is not fully understood even though gait is a common everyday movement that is second nature to most of us. The inherent complexity of human movement is due in part to the many degrees of freedom in the body (Bernstein, 1967), and the unstable nature of gait as we carry over 2/3 of our body weight above half our height across the ground (Winter, 1983; Winter, 1987). The degrees of freedom problem arises in a biomechanical study of the human body, as there are almost limitless ways to configure the segments of the body with its many joints to accomplish the same goal. Also, part of the degrees of freedom problem stems from the redundancy of the biomechanics observable in human movements. Redundancy describes how the
joints of the body can be moved in the same way with different muscle activations such that it is not directly observable which muscles are activated and to what extent they are contracting when creating joint torques. Furthermore, these patterns of muscle activation are likely to be different during each cycle even if movements are very similar (Bernstein, 1967). While the movement patterns that form functional gait are technically infinite, gait has many commonalities from person to person despite the uniqueness inherent in an individual’s gait patterns. The large amount of variation in how we accomplish overground transport, however, makes characterization of gait challenging.

**Walking speed, stability and gait strategy**

Walk speed is a primary measure of gait performance and at least three factors determine self-selected speed. First, people will walk at a speed that results from preferred step frequency and step length and in such a way that the metabolic cost of transport is near a minimum (Kramer & Sarton-Miller, 2008; Massaad et al., 2007; Saibene, 1990). Second, gait speeds may be modulated by the amount of propulsive force generated (shear in the posterior direction) (Orendurff et al., 2008). Third, gait patterns can be adapted to improve stability and reliability of locomotion, which can be partly accomplished by minimizing unstable states during stepping (Benedetti et al., 1999). Often an outcome measure used to quantify stability during gait is variability of stride length and width, but gaining favor are methods of quantifying the kinematics and coordination of the upper body segments.
(Kavanagh et al., 2005; Latt et al., 2008; Menz et al., 2003; Schrager et al., 2008; Van Emmerik et al., 2005).

Increased gait variability is classically associated with decreased stability, but recent work has revealed that detectible amounts of variability are an indicator of healthy gait (Barak et al., 2006; Haddad et al., 2006; Herman et al., 2005; Jordan et al., 2007; Segal et al., 2008; Van Emmerik & Van Wegen, 2002; Van Emmerik et al., 2005). The function of variability in gait may be to allow for adaptability and research seeking to understand the nature and structure of this variability has shown that gait pattern variability contains complex fractal signatures (Herman et al., 2005; Terrier et al., 2005). Complexity and stability of gait can be assessed by non-linear tools designed to extract parameters such as the Lyapunov exponent or the fractal dimension, and these features may be sensitive indicators of gait stability and adaptability (Herman et al., 2005). However, research has yet to forge strong links between dynamic stability and these signatures of complexity. Some have measured stability during normal gait by Lyapunov exponent, which seems to follow a “U” shaped curve across a range of gait speeds with a minimum near preferred speed (England & Granata, 2007). This result suggests that lower gait speeds do not guarantee a more stable gait, and that gait stability is not truly commensurate with variability (Bruijn et al., 2009). As such, it remains unclear whether increased gait variability increases or decreases overall gait stability. Reports of increased variability at slower gait speeds in healthy populations are an indicator that a slow gait speed may not always be the most stable pattern. As such, slower gait speeds in
patient populations may be an adaptation to other limiting factors such as muscle weakness or fatigue.

If an individual is experiencing problems with balance, loss of coordination (ataxia) or perceptual problems such as sensory loss, more stability during walking may be realized by using a cautious gait strategy (Missaoui & Thoumie, 2009; Thoumie & Mevellec, 2002). Cautious gait typically features a reduced separation of center of pressure (CoP) and center of mass (CoM) in the transverse plane, slower approaches (longer contact times) to stability boundaries, wider stance widths, increased dual support times, shorter steps, carrying the body’s CoM lower to the ground and increased trunk pitch (Giladi et al., 2005; Herman et al., 2005; Remelius et al., 2008; Schrager et al., 2008). The kinematic determinants of the cautious gait strategy are hypothesized to be the result of adaptations to changes in sensory system function (Deshpande et al., 2008; Eils et al., 2004) or difficulty with coordination (Giladi et al., 2005) that has lead to greater variability during locomotion. The cautious gait is commonly seen in the elderly and most people with cautious gait are aware of balance limitations, and self-report their need for caution to avoid falls (Nutt et al., 1993). Shorter steps are thought to improve stability during walking, as unstable unipedal (swing) phases of gait are abbreviated and dual support times can be prolonged (Nutt et al., 1993). Lower limb dynamics similar to the cautious gait can be elicited in healthy adults by simply reducing plantar sensation with ice (Eils et al., 2004), which illustrates the links between proprioception and cautious gait. However, other factors besides perceived instability can influence gait, including emotional state.
Embodiment theories suggest a relationship between bodily expression and the way that emotions are expressed (Michalak et al., 2009). These results indicate that gait patterns associated with sadness and depression are linked to slower gait speeds, reduced vertical head and arm movements but also a more ‘slumped’ posture and greater lateral sway. The spatio-temporal patterns of gait associated with slower speed are also reduced in depressed individuals, including: stride length, double support, and cycle duration (Lemke et al., 2000). Chosen gait strategies and general emotional states greatly broaden the range of possible gait patterns and it remains unclear what influence these changes have on gait variability.

**Upper body segmental and inter-segmental coordination during walking**

In current gait research, more attention is being paid to segmental interactions above the pelvis and to joint actions in planes other than the sagittal (Hodt-Billington et al., 2008). Some research exists on torso movement patterns during gait in older adults (Van Emmerik et al., 2005), stroke populations (Hodt-Billington et al., 2008) and in people with Parkinson’s disease (Van Emmerik et al., 1999). In younger and older healthy individuals, there exists a speed dependent transition between coordination modes in upper body kinematics. This transition is apparent such that at slow gait speeds the upper body tends to move as a single rigid segment, characterized by in-phase movement of pelvis and trunk, whereas faster gait speeds tend to elicit an anti-phase counter rotation between trunk and pelvis movements (Van Emmerik & Wagenaar, 1996).
Trunk movement during cautious gait tends to include a greater trunk incline realized by pitching the shoulders forward and down and the pelvis rearward. The increased trunk pitch is presumed to be an adaptation that reduces the risk of falling by lowering the whole body CoM and minimizing lateral body sway (Nutt et al., 1993). However, the trunk inclination of cautious gait may serve to position the head in such a manner that visual perception is improved, which may be especially important to those with sensory deficits.

Gait induces movements of the head that must be regulated to avoid interference with the functioning of the visual and vestibular sensory systems. Researchers studying acceleration of the head during gait found that gait speed, cadence and step length are selected to minimize accelerations of the head (Latt et al., 2008). However, Pozzo et al. (1990; 1991) found that head accelerations were very similar to accelerations experienced at the hip during locomotion. This group concluded that in order to deal with such a constant jarring, which can potentially blur visual information, gaze is stabilized by co-varying head pitch with vertical displacement of the trunk. When the visual field is stable, essential information can be extracted regarding self-motion from the optic flow field. Vision can be a powerful aid in controlling locomotion (Patla, 1997), especially for those with challenges to balance and stability. As such, these compensatory head movements seem especially important for balance-impaired populations. An individual presumably adopts a cautious gait strategy in an effort to enhance stability and increased forward pitch of the trunk and head may serve to keep head movements controllable. Additionally, increased trunk and head pitch position the optic centers
closer to the ground and may enhance visual perception of the ground and lower limb orientation. It is known that the lower visual field is important during gait in healthy individuals when walking over irregular terrain (Marigold & Patla, 2008), and it is presumed that the lower visual field has a similar or heightened importance for individuals with movement and balance disorders, especially if they are taking shorter steps (Azulay et al., 2002; Frzovic et al., 2000; Grossman & Leigh, 1990; Karst et al., 2005). However, it remains to be determined if increasing trunk and head pitch is improving the pick-up of visual information in individuals who adopt a cautious gait.

**The role of perception in the control of locomotion**

The role of perception in the control of locomotion is possibly best understood using a holistic approach that integrates perception (particularly vision) and the actions of the performer. Since sighted individuals use vision in an integrated fashion during locomotion to oversee other perceptual information (Pick & Saltzman, 1978), study of how vision is used for the control of movement will benefit from an integrated approach as well. Current theory used to describe how we perceive information and control movement has emerged from a fusion of independent ideas from two individuals: Gibson and Bernstein. James J. Gibson (Gibson, 1958) forged a perspective 50 years ago that he termed an ecological approach to perception. While Nikolai Bernstein (Bernstein, 1967) who was self-taught put more emphasis on understanding movement itself for developing theories of control and coordination.
Gibson claimed that optic flow and texture gradients formed the basis for a ‘ground theory’ of space perception, which he claimed were not merely additions to established depth cues such as parallax and relative size. Gibson argued that the task of perceiving the 3-D world is less problematic if (a) the properties of the external world are considered and (b) viewing the world is not reduced to a series of static snapshots. Gibson went on to assert that a central task facing an organism is not just to perceive the 3-D environment but how to get around using vision (Gibson, 1958, pp. 183):

‘How does it react to the solid surfaces of the environment without collision … ? What indicates to the animal that it is moving or not moving with reference to them? What kind of optical stimulation indicates approach to an object? And how does the animal achieve contact without collision? What governs the aiming and steering of locomotion?’

For Gibson, the answers to these questions all lay within the optic flow field generated from movement of the optic array. To Gibson, guidance of locomotion, maintenance of balance and the timing of actions were not processes subsequent or even parallel to seeing, but the essence of perception itself. He went on to reject the idea of ‘stimulus’ for perception and talked instead of the pick-up of available ‘information’ i.e., what is bump-able-into, or what is walk-on-able and claimed that the properties relating to an animal’s needs are in the optic array. Gibson termed these features relating to an animal’s needs ‘goal-objects’, which lead to his first suggestion of ‘affordances for action’ fifty years ago (Gibson, 1958; Gibson, 1966).

Optic flow is the pattern of apparent motion, identified from the motion of objects, surfaces and edges in the environment. Optic flow stimuli can emerge from
relative motion between the eyes and the environment. When the environment is not moving, optic flow must be occurring due to self-motion. One feature of optic flow called expansion seems especially useful in the guidance of locomotion (Figure 2.2) (Gibson, 1966). The rate and the center of expansion in a relatively static environment are the result of the motion in a particular direction. The center of expansion is a special feature of optic flow. The center of expansion is the point that the body is traveling towards. More specifically, the center of expansion is the point that the optic array and head are moving toward. As in Figure 2.2, expansion occurs radially from the center and the further from the center of expansion, the more the expansion appears as parallel flow towards the edges of the field of view.

Figure 2.2. Optic flow expansion is centered at the geometric midpoint of this image.

Nikolai Bernstein is most widely known for his studies that formed the theoretical and experimental foundations of the fields of both biomechanics and motor control. His ideas freed neuroscience from the constraints of the reflex tradition by framing an animal’s behavior as active instead of merely reactive. An
organism’s active interaction with the environment includes overcoming environmental resistance to realizing a goal. The complexity of movement control, as demonstrated by Bernstein, is linked to the fact that the system that controls voluntary movements also is the main means of realizing the goals of the organism. This system requires a high level of reliability and accuracy as the activity of this system induces changes in the organism-environment interactions. Interacting with the environment must be accomplished while controlling the many degrees of freedom of the human body to create patterned coordinated movement (Bernstein, 1967).

Bernstein’s other major contribution to neuroscience is called the equal simplicity principle. Bernstein discussed equal simplicity in terms of the performance of voluntary goal-oriented movements. The principle of equal simplicity says that a given objective can be performed with equal ease or simplicity in many ways. This concept also describes actions that have motor-redundancy, another term pioneered by Bernstein (1967), such as the ability to write with a pencil in either the left or right hand, or between one’s teeth. In perceptual situations, we may perceive things visually with a large range of different head orientations. Finally, in Bernstein’s observations of human movement he was impressed by how humans have such fine end-point control, or equi-finality, over the body and how this end-point control can be extended to include the tool they are grasping. How an individual can overcome the degrees of freedom problem of the trunk and arms to create rhythmic, nearly identical blows of a hammer on an anvil was the focus of some of Bernstein’s earliest studies (Bernstein, 1967).
The ideas of Gibson and Bernstein continue to raise debates in the field of motor control. Bernstein's ideas cast doubt on whether motor outcomes are the result of motor programs chosen to yield specific outcomes, instead focusing on movement control by the coordination of the multiple degrees of freedom (Latash, 1998). Meanwhile, B.J. Rogers (2009) has catalogued the responses to Gibson's ideas as to whether his ideas have “deep implications”, as claimed by Reed (1988) or lead to an “oversimplified approach”, as claimed by Sutherland (1991). Bill Warren (2009) called Gibson's ideas a milepost in the development of an information-based approach to perception and action. From the foundations these two individuals laid down for understanding movement control and perception, David Lee (2005) has framed five principles governing motion:

1. *Movement requires perceptual guidance.* This is because movement is not only the product of muscular forces, but also gravity, friction and other forces that are not entirely predictable. These external forces might alter a planned movement's outcome. Therefore, movements must be continually guided based on the perception of how the movement is unfolding and adjustments can be made such that the desired outcome is realizable. One or more perceptual systems may be employed for these ends. The articular proprioceptive system is always actively sensing movements of the skin, joints and muscles. The vestibular system is actively picking up information on the motion of the head (especially when the head is moving, which is almost always). The visual system can be active in many ways, but one cannot be looking everywhere at once, and the gaze must be directed efficiently.
Vision also seems to play the role of the overseer of perception, keeping the other perceptual systems in tune (Lee & Kalmus, 1980).

2. **Movement requires intrinsic guidance.** Movements must be formed to suit its purpose and therefore must be guided intrinsically. Walking style comes from within the walker while the eyes guide the progress across the ground. The patterns of walking that come from within the individual guide the movement but the perceptual systems must monitor that movement to ensure that it is unfolding as planned. Additionally, intrinsic guidance may be related to emotional features such as mood, which has been shown to have an effect on gait patterns (Lemke et al., 2000; Michalak et al., 2009).

3. **Movements are prospectively guided.** Movements rhythmically flow ahead in time. Because an organism has limited power with which to create movement, power must be managed prospectively to ensure that movements are completed properly. For example, there may be unintended outcomes of a jump if there is insufficient power to cover the distance to the landing.

4. **Movement information embraces the future.** In order for movements to be perceptually guided, the perceived information used to guide the movement must allow adequate extrapolation of the movement into the future. Information must have a temporal structure that extends it beyond the immediate present. In this way upcoming circumstances can be planned for and movements directed accordingly.

5. **Movement guidance is simple, rapid and reliable - and probably follows universal principles.** This is evident from watching any animal in motion. That a small animal with a small nervous system can perform an act of dexterity and
precision that is comparable to our own suggest that movement guidance is controlled by simple principles.

These rules help to form a basis for understanding how a movement is governed. The fusion of these rules creates seamless motion while interacting with the environment. Studying movement with these five guidelines in mind can yield insights into how coordinated movement occurs. As a child playing hopscotch can count out-loud and jump to specific targets at the same time, we look for simple principles that make this possible.

A postulated mechanism for the perceiving and controlling of movement relative to a target or goal is the τ-gap theory put forth by Lee & Lishman (1977). A tau gap is a universal temporal variable that describes a first order time (t) to closure of a physical distance, or motion-gap (x) at the current rate of closure (Lee & Kalmus, 1980):

\[
\tau = \frac{x}{\dot{x}} \text{ where } \dot{x}(t) = \frac{dx}{dt} = v(t) \text{ or } \tau = \frac{\text{distance}}{\text{velocity}} \text{ with units } \frac{m}{m/s} = s \quad \text{Equation 2.1}
\]

The motion-gap describes the closure of the organism toward a goal, in terms of distance and motion towards that goal, such as a hummingbird approaching a flower (Figure 1.3A). The classic example used to illustrate τ-gap is the diving of gannet birds fishing at sea. The bird essentially switches modes from a “flyer” to a “diver” when the wings are reconfigured to ensure a safe entry into the water in order to capture the prey fish (Figure 1.3B). The bird is hypothesized to be using the
expansion in optic flow (Figure 2.2) to gauge the closure with the water surface to
determine the proper moment to initiate the wing folding motion.

The strength of this hypothesis for human locomotion lies in the
simplification of the control of motion such that a “locomotor flow line” in the optic
array that specifies the future course can control locomotion and how stopping for
an obstacle can be simply controlled based on the inherent optic array about the
time-to-contact with that obstacle. In this way, a \( \tau \)-gap is a temporally based
perceptual guidance tool for prospectively controlling movement, as tau
eextrapolates how the motion-gap is changing. The information that an organism
needs to change an ongoing motion-gap closure is then extracted from the rate of
change of a \( \tau \)-gap, defined as tau-dot for braking (b):

\[
\frac{d\tau}{dt} = \dot{\tau}(t) = b
\]

Equation 2.2

Tau-dot is the first derivative or slope of the \( \tau \)-gap variable in time \( t \). For
braking, if a current deceleration is maintained, the organism will stop before or at
the boundary if \( \dot{\tau}(t) \) is held constant at or below 0.5; but if \( \dot{\tau}(t) \) is greater than 0.5 a
collision will ensue (Lee, 2009). Features of the global movement of the body can be
described in this way by the tau gap. However, as the body is composed of many
segments and these segments coincidentally move together in a coordinated manner,
the motions of these elements must be coupled in some way. When behavior begins
to be more complex, a coupling-gap can be used to describe how two motion-gaps
interact. The coupling-gap can be used to characterize more involved behavior
between segments where a closure is scaled to match another target or opposing segment (Figure 1.3C; Equation 1.1). If the rates of closures of two motion-gaps (x and y) remain in a constant ratio (K) over time (t), the gaps are coupled.

Bringing a fork up to and into one’s mouth while eating is a frequently cited example of how a motion-gap (up to the mouth) can be coupled to an angular-gap (into the mouth) within the same movement. A coupling-gap can form between two separate objects as well. In the case of intercepting a moving object such as a ball with the hand, a coupling-gap can explain how the hand can be brought to arrive at a target at the same moment that the ball itself reaches the target (Figure 1.3C). In locomotion, a coincidental closure of the head to a targeted step-landing area, and the closure of the foot toward that location during swing may be coupled. These gaps are essentially time closure estimates that may be perceived directly or ecologically by the organism but can be estimated as contact time based on Newtonian laws of motion. The $\tau$-gap estimate relies on position and instantaneous velocity, and distance to a boundary to determine the temporal gap closure. By specifying a control parameter as a temporal measure, no preference to distance or angle in arbitrary coordinate systems is necessary. This reduction of position and speed information into a temporal measure has been capitalized upon by researchers in the area of postural control in a formulation called postural time-to-contact (Haddad et al., 2006; Hasson et al., 2008; Remelius et al., 2008; Slobounov et al., 1998; Van Wegen et al., 2002).
**Time to Contact while the stability boundary remains constant**

The postural time-to-contact (TtC) measure relates to what is described as a motion-gap and fits under David N. Lee’s umbrella concept of tau (Lee & Lishman, 1977). TtC is a temporal measure that relates movement of the body toward and away from the base of support, or perimeter of the feet, to a temporal-gap. TtC has been used to understand postural strategies of balance (Haddad et al., 2006), the postural phase of gait initiation (Remelius et al., 2008) and the onset of stepping after a postural perturbation (Hasson et al., 2008). TtC is a quantification of approach to a boundary by either the center of pressure (CoP) or the center of mass (CoM) and is reported in seconds. In the Slobounov et al. (2008) approach, TtC is computed as the theoretical time it may take the CoP or CoM to contact the stability boundary, given its instantaneous position relative to the boundary (p), velocity (v), and acceleration (a):

\[
TtC = \frac{-v \pm \sqrt{v^2 - 2a(p_{\text{max}} - p)}}{a}
\]

Equation 2.3

In posture, a short TtC is generally interpreted as a less stable state than a long TtC. As TtC drops, individuals are closer to and/or approaching a boundary with a faster velocity. Depending on the research questions posed, acceleration data can be included in the calculation as it was originally included in the TtC measure (Equation 2.3), and this may make this formulation more sensitive to changes in velocity (Slobounov et al., 1997). The use of TtC in postural studies relates approaches toward a fixed physical boundary to quantities that are probably not
directly perceivable. Arguably, a person does not know exactly where their net CoM or CoP are in space, nor do people perceive the motion of these variables directly. The CoP and CoM are convenient mathematically derived summary measures that are used to compute estimates of the average location of the body and the average point at which ground reaction forces are centered.

The computation of CoP based measures of TtC face an additional challenge of validity as, unlike the CoM, which is a physical measurement of the segmental masses of the body with inertia, the CoP has no mass associated with it and can move instantaneously. Because the CoP can move instantaneously, the meaning of CoP velocity is questionable. CoM measures of TtC generally behave more predictably as inertia limits the changes in velocity possible. Using CoM to compute TtC does show that TtC may be a good estimate of a control parameter that the body uses to regulate upright postures during self-generated and external perturbation where the stability boundary remains unchanged (Hasson et al., 2008).

**Time to Contact during gait initiation**

Gait initiation is a dynamical movement, as the movements of the body develop from small motions typical for quiet upright bipedal stance to the dynamic shifts of the anticipatory postural adjustment. The anticipatory postural adjustment is defined by the posterior and lateral (towards the swing-foot) CoP shift that helps propel the CoM toward the anterior stability boundary limit (the toes of the stance foot). The end of the postural phase of gait initiation occurs at the onset of swing, which coincides with dramatic changes in the stability boundary configuration.
At the start of swing, the CoM has a ballistic trajectory as the body is ‘falling’ in an orbit around the unipedal base of support surrounding the stance foot. Remelius et al. (2008) estimated the CoM TtC as formulated by Slobounov et al. (1997) across the transverse plane during the anticipatory postural adjustment of gait initiation. They showed that people with balance deficits (i.e. multiple sclerosis) employ longer TtC times during voluntary shifts that eventually require changing and crossing the stability boundary (Remelius et al., 2008). This result illustrates how balance-impaired individuals may be using a stability enhancing gait initiation strategy that keeps the CoM over the base of support longer before forming a step. This research used the static stability boundary of the outside edges of both feet as in other postural TtC studies. Thus far, the TtC technique assumes that a fixed stability boundary remains in place. However, TtC holds potential for use in more dynamic movement patterns such as walking where the stability boundary changes.

![Diagram](image.png)

Figure 2.3. Fixed stability boundary typical of postural time-to-contact studies (thick dashed line), as used in postural studies on quiet stance and gait initiation. (adapted from Remelius et al., 2008)
The time-to-contact gap of the initial step of walking has been shown to change during childhood development (Austad & van der Meer, 2007), but similar changes may be present in persons with challenged balance. As a child becomes a more accomplished walker, closure of the τ-gap of the CoP near heel-strike results in less forceful, “softer” touch contacts with the boundary. This is in contrast to early development of gait initiation where children tend to move the CoP resulting in “harder”, collision-like contacts with the base of support, but not so hard that they fall over. For developing children, learning not to overshoot contact leads to smoother more controlled walking patterns. It has yet to be determined if the ability to smoothly close motion-gaps is what is lost due to disability and leads to the disproportionate increases in falls in balance impaired populations. Additionally, as Austad & van der Meer (2007) only computed a motion-gap of the CoP at the step-landing area, the role of vision remains unexplored in stepping selection.

**Time to Contact during locomotion**

Movement of the CoP and how it relates to an “extrapolated” CoM (XcoM) during steady state gait have been reported. In Hof’s method (Hof, 2008), the XcoM is a projection of the CoM based on instantaneous velocity as the body approaches the stability boundary. The XcoM is then compared to the CoP under the feet and the discrepancy between these variables is hypothesized to be related to gait stability (Hof, 2008). The fact that the CoP and CoM interact during human locomotion is well known (Winter, 1983). However, factoring velocity into the description of movement as in the extrapolated CoM theory produces similar results as the time-
to-contact (TtC) measure, which is based on the τ-gap theory. Recent work by Hasson et al. (2008) showed that the XcoM estimate for motion matched velocity based TtC estimates of approach to the stability boundary by the CoM in postural perturbation studies.

Walking consists of more than the coupling of the CoM and the CoP. The CoP is alternately moving from a position trailing the CoM to one leading it, and the CoP only leads the CoM once the swing-foot has re-contacted the ground and begun accepting body weight. The approach to new stability boundaries (before swing-foot re-contact) during walking is critical to gait persistence and this is coordinated by visual information (Marigold, 2008). By tracking where the gaze is fixated between steps, a study of the approach to that location may allow an analysis such as TtC to become more of a visually driven or optic flow based variable as described by Lee (2009). The gaze during walking often fixates on future stability boundary locations, which will become the next step-landing locations (Hollands et al., 2002). The approach to that boundary may be better understood by describing how motion of the stepping foot and the entire body CoM close the temporal gap between their current position and the visually anticipated future step location.

**Vision as a prospective means to control gait**

Understanding the visual control of locomotion is a large part of gait research (Gibson, 1958). The use of vision during gait is hypothesized to directly influence the dynamics of the lower limbs during gait as well as many other features of gait such as inter-segmental coordination between the torso and head (Hollands et al.,
Vision allows for guidance of locomotion, as vision is a source of exteroception, or perception of the environment beyond the anatomical limits of the body. Vision can also provide perception of the body's own segments in the environment, a function that helps to validate the true location otherwise distilled from our internal sense of proprioception from muscles, joints and skin movement. Lee proposed a summary term for visual information termed exproprioception that does not differentiate between exteroception and visual proprioception (Lee & Kalmus, 1980). Exproprioception is a synthesis of all available visual information and removes the classic two-fold division in the function of vision. This simplification eliminates the need to distinguish between these two types of information used in the continual generation of locomotor patterns of steady state gait, and the monitoring of them. In order for vision to be utilized fully, the head must be oriented properly to pick-up information.

Head orientation determines the limits of the environment and body that are visually perceptible, as the field of vision is constrained by the anatomical features of the eye orbits. Here, the lower limit of the visual field is of primary interest as the features of the stepping surface predominantly influence walking overground. Gauging the role of vision by head movement is simplified somewhat since the vectors of the eye, head and gaze are closely aligned during visual fixations (Straumann et al., 1991), and thus the orientation of the head provides much of the information needed to describe visual system function. In order to make comparisons of head motion between individuals or populations, a common
reference must be selected. A historical measure developed to describe the neutral plane of the head (primarily describing pitch) was established at the World Congress of Anthropology in Frankfurt Germany in 1884 and called the Frankfurt Plane (Figure 1.6 inset). The Frankfurt Plane is referenced to anatomical landmarks that most nearly align with the surface of the earth during quiet stance. The Frankfurt Plane is defined as passing through the center of the vestibular rings (upper margin of the ear canal or porion) and the inferior limit of the orbit of the eye (orbitale) (The American heritage medical dictionary 2007).

Locomotion is possible without exproprioception, as blind individuals navigate and successfully plan each step of walking in the environment without vision. Blind individuals tapping with ‘the long white cane’ are substituting tactile proprioceptive information for visual exteroception with an anti-phase pattern of cane taps where the tap is “clearing” the subsequent step location prior to the swing phase of walking gait (Bickford, 2009). In this way, blind individuals are adapting a gait strategy used by sighted individuals, who use visual exteroception to “clear” step landing location in a bilaterally alternating fashion. However, without visual proprioception the blind must rely entirely on internal sense of body position, thus adding to the challenges they face with mobility.

Visual exproprioception seems to be an inherent part of how visual information is picked up. The research initiated by Gibson and built upon by Lee hinges on direct perception of information, and current research is still uncovering mechanisms that show this to be the case. It seems that information processing occurs at the retina itself, signifying the integrated nature of visual information
processing. New research has determined that there exist multifunctional neural circuits that specialize in the detection of approaching objects, which can theoretically include movements of the body’s own segments (Munch et al., 2009). Munch et al. (2009) have identified approach sensitive ganglion neurons that suppress responses to all non-approaching objects. In the context of control of locomotion, the appearance of the swing-foot as it closes on the upcoming step landing location can be considered an approaching object and very fast acting circuits may be helping to ensure rapid and clear perception of the foot as it enters the peripheral visual field.

How the visual system conveys information, what that information is, and how it forms actions continues to be debated. One exchange that occurred during the late 80’s centered on whether or not the control of step length in treadmill running was principally accomplished by adjusting vertical impulse force. Warren et al. (1986) found evidence that treadmill runners control step length by regulating vertical impulse, which was modulated by adjusting the τ-gap. Warren termed this step length control tau-impulse, but Patla et al. reported a 40% contribution of horizontal impulse in overground running and a strategy that depends on the timing of cuing, which undermined the tau-impulse theory stating that 80% of the variation in stride length was due to vertical impulse control. However, Warren & Yaffe (1989) reported on overground running data that shows only a 20% contribution of horizontal impulse, which may be a consequence of vertical impulse regulation. According to Warren, this difference is due to the differences in dynamics between overground and treadmill running. However, these new data indicate that a global
impulse parameter is regulated that primarily affects vertical impulse, thus
salvaging the tau-impulse theory. Warren worked with Lee on the formulation of
this tau-impulse theory, which is strongly influenced by the Gibsonian philosophy
emphasizing the role of visual flow in the guidance of locomotion. The perspective
added by Patla et al. is not is an example of the multi factorial influences that we use
to control locomotion. Most will agree that optic flow plays a central role in
distinguishing movement cues of the optic array relative to the environment
immediately in the path of the individual who is walking. One way to identify how
the optic centers are moving relative to the environment is to seek the point at
which the head is pointed during locomotion and a means to do this is known as the
head fixation distance.

**Head Fixation Distance (HFD)**

The search for understanding of how head motions are regulated during
dynamic motions has lead to the formulation of a concept known as the head
fixation distance (HFD) that serves to combine measures of head translation and
rotation in a single variable (Pozzo et al., 1990). However, HFD is limited to vertical
translation and pitch of the head, where other features of motion such as surge
(forward and backward translation) are not considered. The HFD is derived from
shift and pitch data on salient points from a locomotor cycle and computed by
triangulation between intersections of projections of the Frankfurt Plane anterior to
the body. The points commonly utilized to triangulate the HFD location are the local
minima and maxima of vertical head translation within one cycle of motion. As such,
on a cycle-to-cycle basis estimates of where a person is fixating during movements such as walking, walking in place, running in place or hopping are discernable (Hirasaki et al., 1999).

Originally formulated by Pozzo and colleagues for overground locomotion, the HFD has found favor in treadmill studies, even though features of optic flow are absent in such paradigms. Regardless of the study, if the motion being examined is cyclical, motion of the head can be portrayed in reference to where the head is pointing. The limitation of this HFD lies in its lack of ability to describe where fixation is occurring during the entirety of a gait cycle and that the outcome measure is a three-dimensional point relative to the head origin that holds little relevance to features in the environment or the rest of the body.

Without environment-relative context, it is difficult to ascertain what features are being fixated upon, unless a specific target is positioned in front of the person as in a treadmill-walking paradigm. The HFD does provide insights into the compensatory mechanisms that are an integral part of stabilizing the visual field such that meaningful information can be extracted. These limitations of the HFD have in part kept this measure from being established as a primary means to quantify the features of visual flow in terms of unconstrained (overground) locomotion. Perhaps projecting the HFD to the transverse plane during locomotion may yield information regarding the distance from the body to fixation points during overground locomotion, but many questions remain regarding the meaning of the results.
Most literature utilizing HFD are primarily concerned with the operation of vestibulo-ocular reflexes including the vestibulo-colic, cervico-colic, cervico-ocular, and visual reflexes ability to produce compensatory head orientation adjustments while experiencing dynamic shifts originating from lower in the kinematic chain. The HFD best describes intermittent positioning of the head, as one point per cycle, and not a continuous variable that may provide insights of behavior within the gait cycle. The ability to modify the amount of head motion during gait may allow for increased reliance on the visual field during gait. Walking is an important ability for maintaining independent mobility and quality of life in a person's daily living. People with MS often have problems with their ability to walk and they may be moving differently to increase the stabilization of the visual field as they may be using visual information differently than non-MS individuals.

**Multiple Sclerosis (MS)**

Multiple sclerosis (MS) is a disease that affects the central nervous system (CNS). MS is an immune-mediated disorder that results in damage of the myelin sheath surrounding CNS neurons. In MS, CNS damage is typically progressive or mixed with remissive periods between demyelinating episodes. During remission in MS, scar tissue (sclerosis) forms at sites where the body has imperfectly repaired the damaged myelin sheaths. The onset of MS is generally between 20 and 40 years of age and affects approximately 2.5 million people worldwide. Women are more than twice as likely to contract MS as men at a young age, although the incidence in men and women is nearly equal for those diagnosed with MS later in life.
Neurologists diagnose approximately two hundred people every day with MS yet the cause(s) remain unknown (Ragonese et al., 2008).

Signs of MS are unpredictable but often include diminished motor drive, weakness, spasticity, and loss of proprioceptive function, poor coordination, fatigue, vision problems, and cognitive impairment (DeLisa, 1985). Signs of MS may be associated with the location and bilateral asymmetry of the sclerotic lesions caused by MS, which can be identified with some success using MRI techniques (Bakshi et al., 2008). A truly accurate way to track MS lesions in vivo remains illusive, and there has been very little success in identifying sites of demyelination before they become sclerotic, which adds to the challenges of understanding what initiates CNS damage.

Many of the symptoms of MS can interfere with mobility and in this way directly affect quality of life. It has been shown that physical activity correlates with neurological impairment and disability in MS (Motl et al., 2008). Over two-thirds of individuals with MS retain the ability to walk during the course of MS; however changes in gait patterns are apparent even with low levels of disability (Martin et al., 2006). Relating the distribution of MS lesions to symptoms of MS is difficult but symptoms frequently include fatigue, problems creating coordinated movement, and difficulty with balance and mobility. These symptoms may partly explain the high incidence of falls (~52%) in people living with MS (Finlayson, 2006; Peterson et al., 2008) and self reported problems with postural stability (McAlpine, 1985). Poor balance and fear of falling, combined with symptoms of fatigue and the uncertainty of how a case of MS will progress, often leads newly diagnosed people
with MS to lower their current level of physical activity and overestimate their future disability (Peterson, 2007). It is important to inform the newly diagnosed that disability progression in MS is slower than previously reported in most cases (Pittock et al., 2004; Tremlett et al., 2006), as decreased physical activity may increase the likelihood of developing other diseases. Diseases such as obesity, cardiovascular disease, osteoporosis, and type-2 diabetes can significantly shorten life expectancy. However, daily activity like a 30-minute walk can curb these tendencies (Hamilton et al., 2007; Pate et al., 1995). Additionally, as life expectancy in people with MS is equal to healthy peers in 90% of cases (Finlayson, 2004; Ragonese et al., 2008), it is important for caregivers to emphasize physical activity that has been shown to help in maintaining mobility and quality of life (Dalgas et al., 2008; Motl et al., 2008; Motl et al., 2009). To help people cope with MS, researchers must learn how to inform people with MS on how they are adapting their mobility to impaired balance control and perceived loss of stability (Lode et al., 2007).

**Characterization of disability due to MS**

Symptoms of MS are widely disseminated and highly variable, which is why a systematic means to characterize neurological involvement is a great aid in associating impairment and mobility. The most common scale used to characterize physical impairments due to MS is the Expanded Disability Status Scale or EDSS, with scores ranging from zero to ten (death due to MS) (Kurtzke, 1983). The EDSS correlates strongly with mobility in the midrange of scores, taking into account maximum walking distance and the use of walking aids.
The EDSS score is also a composite of disability in eight neurological Functional Systems (FS), which are more sensitive to differences in low and high ranges of EDSS (Goodkin et al., 1992). The FS and their approximate prevalence upon MS diagnosis at the time of publication (1983) include Pyramidal (P) 84.9%; Cerebellar (CB) 76.9%; Brain Stem (BS) 73.9%; Sensory (S) 55.2%; Bowel & Bladder (BB) 22.6%; Visual (V) 33.9%; Cerebral-total (Cb) 20.7%; and Other (O) 14.9% (Kurtzke, 1983). Little epidemiologic research is available whether the distribution of prevalence in the FS scores has changed in recent years. Functional systems that may affect mobility include pyramidal (loss of motor drive), cerebellar (lack of coordination), sensory (touch and pain), brain stem (including vestibular), visual, and cerebral (cognition) systems (McAlpine, 1985). Cognitive impairment is a special case because of the potential for impaired judgment or visuo-spatial processing (Hoffmann et al., 2007; J. M. Rogers, 2007). Sensory system (S) impairment due to MS may be another special case affecting mobility, as an awareness of the interaction with the environment is necessary to shape movement patterns.

Increased sensory impairment due to MS increases variability during posture (Horak, 2001; Rougier et al., 2007) and gait (Thoumie & Mevellec, 2002; Thoumie et al., 2005). Severe numbness due to MS can occur in the feet such that individuals cannot feel the floor or know where their feet are. If this sensory loss is causing observable discoordination when a person closes their eyes, it is formally identified as sensory ataxia (Missaoui & Thoumie, 2009). The level of involvement in each functional system and the resulting combination of symptoms due to MS will affect
the ability to perform tasks of daily living, but little research has focused on the FS scores and mobility (Thoumie & Mevellec, 2002). Importantly, cerebellar (Cb) function is critical to coordination, as the cerebellum seems to act as a comparator of sensory and voluntary motor information that, when impaired, causes unstable movement. Questions remain regarding compensatory interactions of the functional systems when affected by MS during dynamic tasks. For instance, increased reliance on visual proprioception may partly compensate for sensory ataxia during gait, but the degree to which this is occurring is unknown.

Efforts to improve quantification of disability in MS have lead to the development of scales that expand on the EDSS and relate more to activities of daily living in people with MS. These tests factor in life satisfaction, as in the MS quality of life inventory (MSQLI) (Fischer et al., 1999), and place greater emphasis on cognitive function compared to the EDSS such as in the MS Functional Composite (MSFC) (Fischer et al., 1999). As the EDSS has a broad scope, especially in the lower score ranges, an important but absent feature of the EDSS is a measure of physical disability in the upper body. The trunk impairment scale (TIS) is a measure of the ability to control upper body movement in MS, which has been adapted from research on stroke patient populations (Verheyden et al., 2006). The TIS can provide detailed information regarding aspects of upper body performance related to mobility including the phasic coordination of trunk and pelvis rotations in the transverse plane (Van Emmerik et al., 2005). It is not yet clear to what degree trunk impairment may influence mobility in people with MS. Additionally, it remains unclear how MS affects torso movement during gait or whether trunk pitch plays a
role due to emotional status or safety concerns or some other factor. We define mobility as the ability to walk, but the salient features describing gait performance remain debatable. In order to understand gait in people with MS, researchers must carefully select features of gait, a process that has just recently begun in MS research but lacks the thoroughness afforded other patient populations.

**Gait patterns of individuals with multiple sclerosis**

Research on movement dysfunction in MS has begun, however relatively little research exists compared to aging, stroke or other neurological conditions such as Parkinson’s disease. Most people with moderate disability due to MS have clinical problems with gait, but the relationship between loss of balance and disability level needs further investigation considering the high incidence of falls (Cattaneo et al., 2007; Nilsagard et al., 2009; Peterson et al., 2008).

Even with minimal impairment due to MS, people have changed their movement strategies to adapt to their functional limitations. The first and most apparent reported gait changes in MS are slower gait speeds and prolonged dual support time, which can appear before clinical disability is recognized (Benedetti et al., 1999; Martin et al., 2006). Gait speed can be influenced by asymmetries in strength or symptoms in lower limbs (Chung et al., 2008; Crenshaw et al., 2006). However, a higher metabolic cost of walking in MS may also be a factor in their slower walking speeds (Olgiati et al., 1988; Tantucci et al., 1996). Alternatively, gait speed may be slower in MS due to muscle weakness or control in the production of propulsive forces (Thoumie & Mevellec, 2002). Higher metabolic costs of walking
due to MS have been related to spasticity and ataxia but not to trunk or lower limb weakness (Olgiati et al., 1988). Fortunately, the cardio respiratory response to walking in the MS population has been shown to be unrelated to fatigue, and as a result, this may not a limitation for walking (Chetta et al., 2004). It remains unclear if stride length in the gait of individuals with MS is set strictly by anthropometrics or by other factors including energy cost or gait stability.

A likely factor in selecting a preferred gait speed is stability but it is unclear how to measure stability directly. Gait abnormalities in people with MS may be due in part to impaired stretch reflexes and increased ankle stiffness that have been observed in MS patients with spasticity (Sinkjaer et al., 1996). During gait, people with MS reduce ankle range of motion through extended activation of lower-leg muscles. This strategy of stiffening may be part of a cautious gait strategy to improve stability, which forces an increase in hip range of motion during gait, although it is unclear how trunk-pelvis coordination interacts with this strategy.

Gait in MS in some ways looks like accelerated aging in that gait patterns approximate those of a healthy older adults (slower gait speeds, shortened strides, longer dual support time) (Barak et al., 2006; Kang & Dingwell, 2008; Shkuratova et al., 2004). It remains unclear if older adults’ gait patterns of upper body coordination (Van Emmerik et al., 2005), cautious gait (Herman et al., 2005), or overall gait variability (Heiderscheit, 2000) also appear in people with MS because of the disease or walking speed. In addition, coordination of the trunk in the lateral direction is especially challenging for older adults with balance impairments during
posture (Maki et al., 1994) and gait (Van Emmerik et al., 2005) but it remains unclear how this manifests in people with MS during gait.

Changes in coordination due to MS may explain slower walking speeds and longer contact times with the stability boundary approach during gait initiation (Remelius et al., 2008). These coordination and balance differences during the initiation of gait may relate to strength and sensory system function as well as to the production of appropriately scaled movements. Gait speed is correlated with hamstring strength in people with MS but this correlation becomes stronger when somatosensory loss is factored into the regression (Mevellec et al., 2003; Thoumie & Mevellec, 2002; Thoumie et al., 2005). Somatosensory loss for individuals with MS can range from a subtle reduction in touch sensitivity to complete numbness. The sensitivity losses experienced by those with an affected sensory functional system commonly occur in a proximal to distal pattern (Missaoui & Thoumie, 2009; Rougier et al., 2007; Thoumie & Mevellec, 2002). Thus, the feet are highly susceptible to these neurological changes due to MS.

Research on sensory dysfunction may directly apply to the locomotion patterns of people with MS, especially considering that this is a population commonly affected by somatosensory loss. Difficulty with movement can be compensated for with higher reliance on visual proprioception as in other neurological disorders such as Parkinson’s disease (Azulay et al., 2002). A strategy seen in persons experiencing gait disturbances is characterized by increased downward pitch of the head. This may allow for a more proximal pick-up of visual cues regarding terrain (exteroception) and limb configuration ( proprioception).
This may also be especially important when compensating for distorted, delayed or absent sensory information. However, it remains to be determined if reported changes in gait patterns in MS are related to gait speed or disability due to MS and how much of these gait changes are related to improving exproprioception. Changes in gait patterns may be due to the neurological symptoms of MS, such as loss of peripheral tactile sensitivity (Kelleher et al., 2009), however these changes may also be adaptations to changes in balance.

Generalization of the effects of MS on walking ability remains difficult because symptoms vary greatly between individuals and the degree to which functional systems interact dynamically in the creation of coordinated movement remains unknown. However, understanding cases of MS in terms of successful gait strategies may lead to better and more meaningful treatments and functional outcomes. It is unclear how strongly outcomes of interventions depend on correlations with FS, MSFC and TIS scores, gait performance, tailored gait training or physiotherapy. The goal of training interventions must not be to solely increase gait speeds to normal speeds, but to improve mobility and reduce instability while walking.

**Gait intervention and individuals with MS**

Gait performance in MS can vary from day to day due to fatigue or motivation and there may be little correlation between the duration of the illness and the severity of the symptoms (Albrecht et al., 2001). Additionally, there may be a lack of correlation between objective performance based measures and subjective self-
report measures of functional performance (Goverover et al., 2005). However, strength-training interventions have shown a benefit to people with MS regardless of their limitations (Gehlsen et al., 1986). In addition, people with MS often have problems with foot drop during the swing phase of gait due to changes in neural activation of the muscles below the knee that lift the foot. Typically foot drop due to MS is treated with a foot orthosis and although it is unknown what neurological systems are responsible for this condition (Paul et al., 2008), signs of impairment in pyramidal or cerebellar function may be indicative.

Researchers are continually applying new interventions aimed to alleviate the symptoms of MS, which reflects an effort to improve stability and mobility, compensate for functional impairment, and reduce or prevent injury from falls. One frequently administered intervention is aerobic exercise, which has been shown to improve neurological function and mobility (Kileff & Ashburn, 2005), gait speed and postural stability (Cantalloube et al., 2006; Romberg et al., 2004; Romberg et al., 2005). Similarly, resistance training has been suggested as an intervention, and has been shown to improve gait kinematics in people with MS (Gutierrez et al., 2005; White & Dressendorfer, 2004; White et al., 2004). Home based resistance training was shown to be an effective means to improve strength (DeBolt & McCubbin, 2004) that may lead to improvements in balance, power and mobility. A physiotherapy approach improved disability measures and impairment on timed walk gait assessment, Rivermead mobility index and Berg balance test (Lord et al., 1998). However, quality of life may not necessarily improve in commensurate amounts with decreases in functional impairment (Romberg et al., 2005). It has been shown
that MS itself is not limiting a person’s adaptability to exercise, and an increased exercise training level is correlated with faster gait speeds (Tantucci et al., 1996). Research must continue to probe the root causes for limiting factors of physical activity in MS, which seem unrelated to the capacity to exercise, but that may be part of FS score outcomes. Separating FS scores from changes in mobility due to MS remains difficult because people with MS tend to overestimate their current physical disability level. However, with training, the confidence to be active improves, and people who can move better show less of a tendency to overestimate their future disease state (Janssens et al., 2003). If people with MS realize that physical decline due to the disease can be very gradual, not unlike typical aging, exercise adoption may improve. A recent meta-analysis of 22 articles examined the overall effect of exercise training interventions on 600 individuals with MS, and showed that exercise interventions are associated with small improvements in walking mobility (Snook & Motl, 2009).

Other forms of interventions besides formal exercise prescriptions have also been employed in people with MS, including auditory feedback, virtual reality and robot assisted gait training (Beer et al., 2008), cognitive training, electrical stimulation and vibro-tactile stimulation (Schuhfried et al., 2005). Sensory stimulation of the feet (vibration, cold and hot foot baths, foot dexterity exercises) as a rehabilitation modality has also shown to reduce postural sway due to sensory ataxia (Missaoui & Thoumie, 2009). Auditory feedback was shown to increase gait speed and reduce gait variability (Baram & Miller, 2007). Additionally, musical motor feedback interventions developed for people recovering from stroke may
benefit people with MS in a similar way (Schauer & Mauritz, 2003). Virtual reality-based balance training was shown to increase locomotion speed, endurance and balance (Baram & Miller, 2006; Fulk, 2005). It is unclear if any training effect from these interventions is retained upon leaving the laboratory or how long a training effect may last.

Whole body vibration had strong effects for one week on postural control and mobility in people with MS (Schuhfried et al., 2005). Vibration stimuli may also be beneficial to people with MS in smaller doses as shown in people with Parkinson’s disease and in diabetic patients, where plantar surface stimulation synchronized with stepping made gait less variable (Khaodhiar et al., 2003; Novak & Novak, 2006). It has been established that plantar vibration in healthy adults can influence quiet standing (Bensmaia et al., 2005; Kavounoudias et al., 1998) but the response to vibration during the gait cycle (Forssberg & Hirschfeld, 1988) is largely unexplored in MS. Functional electrical stimulation reduced foot drop due to MS and increased gait speed, although the reasons are unclear (Paul et al., 2008).

For those with more severe disability due to MS, body weight supported treadmill interventions have improved mobility and gait parameters (Giesser et al., 2007). A promising therapy focuses on whole body stability during gait by training voluntary head stabilization to reduce head and trunk oscillations during gait (Cattaneo et al., 2005), which may improve gait stability. Trunk impairment has not been measured as part of gaze stabilization therapy.

These interventions mainly seek to bring gait speed up to normal levels by cueing stepping or enhancing sensory feedback. Careful evaluation of gait variability
in relation to stability may yield insights on how to improve walking gait such that mobility is safeguarded. Additionally, cognitive rehabilitation is an under-explored therapy modality for people with MS and may have beneficial impact on judgment or visuo-spatial processing, but it remains to be determined how cognitive training improves gait stability (O'Brien et al., 2008). Training must remain proportional to impairment in any intervention design and must not simply attempt to bring gait parameters or other measures to “normal” levels. Slower gait speed may have a functional aspect or be part of an attempt to mitigate for increased metabolic cost of walking, or strength loss, or to compensate for poor balance control and stability.

**Summary**

Many questions exist regarding how gait patterns are regulated. These questions may hold heightened importance for those coping with balance impairments due to neurological disorders such as MS. In this chapter, considerable attention has been paid to defining a walking gait cycle, the role of perception in the control of locomotion, how the work of James Gibson, Nikolai Bernstein and David Lee have shaped the understanding of movement and perception, and the neurological disease multiple sclerosis. The gait cycle is well defined and has had extensive study relating to the spatio-temporal patterns of the lower limbs, and the influence of gait speed on the coordination of upper body segments. However, there are questions remaining relating to the manner in which gait is regulated including how step-landing areas are selected and what underlying rules may be driving step length and duration.
Individual and coupled temporal boundary-closures (measured by tau) have been used to describe coordination in the dynamics of motion. The tau theory Lee developed was formulated to describe the visual perception of boundary closure. However, tau theory has proven to be an important basis for other boundary approach theories. Time to contact theory has been applied to postural activity, and the perception of these temporal approaches is hypothesized to derive from proprioceptive and visual sensory information. In TTC studies of postural control, the boundary is fixed, and TTC has had limited application to the dynamics of gait where the boundaries change. In the description of the dynamics of gait, tau theory may be useful in characterizing the approach to an anticipated boundary that is inside the field of view, and TTC may be useful in the characterization of coordinated movements toward proprioceptively perceived boundaries.

Together with the knowledge gained from research on the motion of the visual field during gait, the salient features of the environment for the guidance of a step is accessible. These features of the optic centers may help gain a better understanding of the underlying mechanisms utilized in the formation of the stepping action that is a hallmark of human walking gait.

By forging a better understanding of healthy patterns of gait in young adults, tools become available for comprehending the breakdown of stability and mobility that disease can bring. In order to study the importance of visual information to those with chronic challenges to balance, the body of literature on step regulation and anticipation must be expanded. However, a major step in understanding gait in people with MS is to first understand the adaptations made to accommodate the
dynamics of walking more slowly. Since relationships have been established between stride length and speed in healthy adults (Larsson et al., 1980), similar regressions formed for people with ranges of impairment due to MS may help to describe the relationships inherent in gait patterns in the population of people with MS. Functional gait is best described as the ability to move overground without falling, and as falls in MS are quite common the gait patterns adopted must serve to keep falls to a minimum (Cattaneo et al., 2007; Nilsagard et al., 2009; Peterson et al., 2008). Following the establishment of the patterns of speed adaptation to gait, the potential role of increased visual exproprioception during gait in the MS population can be further investigated. Increased reliance on visual exproprioception may be important in those with MS considering the high incidence of lower limb proprioceptive loss (Kelleher et al., 2009; Missaoui & Thoumie, 2009; Rougier et al., 2007).
References


CHAPTER III

METHODOLOGY

General Introduction

The overarching goal will be to better understand gait by describing with the visual tau-gap measure how head movements aid in the visual selection of a step-landing area and with the postural time-to-contact measure how the body approaches the anterior limits of the physical and anticipated bases of support (physical base of support and anticipated base of support) during each step. Following the adaptation of these techniques to the dynamics of gait at preferred speed, the response of these temporal closure measures to slower and faster gait speeds will be developed in a group of young healthy adults. How temporal gaps are closed and the influences of speed on gap closure may be especially important for populations experiencing chronic challenges to balance. For those with balance disorders, keeping mobility and stability intact presents special challenges and the management of the spatial patterns of gait and temporal gap closure during gait may reveal strategies that enable people with diseases such as MS to retain the ability to walk.

To achieve these goals, four studies will be carried out on data from two gait collections. The data sets are (A) young healthy adults, and (B) adults with mild-to-moderate impairment due to MS and age-matched controls. The first study will be based on preferred speed walking trials of data set A and form the basis for the methodology applied in studies two and four. Study two will focus on a speed manipulation within data set A, intended to induce longer stride lengths and shorter
stride times at faster speeds and shorter step lengths and longer stride times in slower speeds. In study three, stride parameters and lower body gait kinematic patterns extracted from data set B at preferred speed and a set of slow, medium and fast walking speeds will allow direct comparisons of stride characteristics at matched gait speeds that the literature has lacked. In study four, gait trials at preferred speed and across a verbally induced set of gait speeds shall be collected on individuals from data set B.

**Experimental conditions and collection equipment**

Conditions in data set A and B will consist of walking trials at preferred speed, and slow (0.6 m/s), medium (1.0 m/s), and fast (1.4 m/s) speeds. Six-meter trap times of the trial shall be recorded to measure average gait velocity. Successful trials will be within 0.05 m/s of the target speed as determined from the six-meter time trap surrounding the collection volume. The collection volume for all gait trials in data set A and B will be calibrated and recorded with an eight camera Oqus system at 240 Hz (Qualisys Inc, Sweden). In data set A, an Oqus camera will be set to record video of the center four meters of the collection volume in the sagittal plane at 60 Hz to validate head-movements in post-processing and a force plate under the boardwalk that is embedded in the laboratory floor will record throughout the collection (AMTi, Newton MA). In data set B, collections will include foot contact events collected via footfalls over a gait mat (2m X 0.5m) synchronized to the cameras at 60 Hz (RSscan Inc, Belgium). All participants will walk on a 65 cm wide boardwalk for all collections. The total length of the boardwalk will be
approximately 11.75 meters, placed such that calibrated kinematic collection
volume (approximately 4 meters) will be in the middle of the boardwalk, which will
exclude transitional effects of gait initiation and termination. For data set A, the
boardwalk will be flat black carpet. In data set B, the boardwalk will be a flat black
painted plywood surround put in place to elevate the walking surface to the same
height as the sensing elements of the RSscan mat.

Participants in data set A will perform all conditions wearing tight fitting
shorts and T-shirt, and wear lab supplied running type shoes. Participants in data
set B will wear the same type of clothing and shoes, but will repeat all gait speeds in
unshod conditions. Shod foot recordings will be performed because most studies
reporting walking gait patterns of individuals with (and without) MS have been
performed in the shod condition. Bare foot conditions will be performed in data set
B because the sensitivity of the foot pressure system allows events within a footfall
to be sub-divided with detail. Differences in how the feet make contact with the
ground during stance may contain clues pertaining to how gait strategies are
adapted when dealing with chronic challenges to balance. The outcome of these
comparisons will help to understand the alterations to gait parameters commonly
experienced by individuals with a neurological disease such as MS, where gait is
affected but gait disability may be not clinically apparent (Benedetti et al., 1999).

**Kinematic model**

Kinematic data in sets A and B will share identical tracking and calibration
retro-reflective passive marker sets. A thirteen segment biomechanical model will
be built from 56 markers. Segments shall be defined as rigid bodies, with
established anthropometrics (Clauser, McConville, & Young, 1969): head, trunk, pelvis, and bilateral segments: upper arm, lower arm, thigh, lower leg and foot (Figure 3.1).

![Figure 3.1. Marker placement and 13-segment geometry. Light markers: tracking; dark markers: segment definition landmarks.]

The head will be tracked via five markers attached to a rigid hardhat suspension frame placed over the crown of the skull; one marker 4 cm above each superciliary arch, two posterior markers and one above the crown. The Frankfurt Plane will be referenced to the head markers with the visual 3D pointer, by events identifying the bilateral external auditory meatus and inferior orbital arches (Figure 1.6 inset; see also Appendix B) (Johnson, 1950). Once the Frankfurt Plane is identified with the C-Motion pointer, the limits of the lower visual field shall be established and referenced to the Frankfurt Plane.
A methodology is proposed to validate assertions that the lower visual field subtends -65° from neutral (Figure 1.5; 1.6). By a repeated fore and aft pitching of the head during a neutral postural stance pointer events shall be created at the moment where markers placed on the ground in front of the participant shall be just visible. Reference markers placed at the toe (25 mm from the CoM) and 50, 75, 100, and 125 mm anterior of the medial tibial malleoli shall be used as visual targets. Pointer events shall be recorded upon verbal report that the markers approached the limit of their lower field of vision. Thus, an angular measurement will be computed between the horizontal, as measured by the Frankfurt Plane, and the lower limit of the visual field at the point of intersection with the ground (see Appendix C for details).

The trunk will be modeled as a rigid segment spanning the distance from the bilateral iliac crest of the pelvis to the bilateral acromion processes and tracked via four markers on a snug fitting backpack.

The pelvis will be constructed from pointer landmarks generated at the femoral greater trochanters and the pelvis iliac crests. Pelvis markers used to reference the landmarks shall also be used as tracking markers: the sacrum, bilateral posterior superior iliac spines and bilateral anterior superior iliac spines.

A rigid cluster of four markers will be attached on the lateral aspect and near the middle of each thigh and lower limb segment with Velcro. Pointer landmarks at the femoral greater trochanter, medial and lateral knee and tibial tuberosity and fibular tuberosity shall be used to construct the lower limb segments.
Each foot will be tracked via five markers; a rigid triad will be affixed to the heel and two locations on the forefoot, one marker on the first and one marker on the fifth metatarsal joint. Additional markers shall be placed on the feet at the second metatarsal joint, the medial, volar and lateral arch as represented by markers at the base of the first, second and fifth metatarsal; the medial apex of the navicular bone, the lateral apex of the peronial bone, the medial apex of the sustentaculum tali, thus replicating the Leardini foot model marker set (Leardini at al., 1999; Leardini et al., 2007).

Bilateral arm segments shall be created from a radius and ulna marker, a lateral elbow marker, and an acromion marker. An additional landmark will be created on the medial elbow that will be referenced from the lateral elbow to the acromion, and ulna.

**Data Reduction**

All 3D kinematics will be exported from Qualisys Track Manager into Visual 3D and analyzed on a cycle-by-cycle basis. The gait cycle is defined as beginning and ending with a right foot heel-strike where the right foot will initially be in front of the rear left foot (Figure 3.2A). In Visual 3D, all marker trajectories will be filtered with a zero-lag recursive second order Butterworth filter with a low pass cutoff frequency of 8Hz in accordance with estimates found in the gait literature. For the purposes of this proposal, only anterior boundary approaches are considered since the anterior direction is primary for progression during walking.
Figure 3.2. Center of Mass trajectory during walking gait (thin line S-curve) and example CoM position (bulls-eye) during dual support stance (A & C) and swing (B) phases of gait. Vertical lines: thick) physical base of support, and thin) anticipated base of support.

Gait events

For both data sets A and B, heel-strike and toe-off events will be extracted from the force sensing structures that are part of the collection equipment.

Following this, a kinematic pattern recognition algorithm in Visual 3D will be used to identify bilateral heel-strike and toe-off events not occurring on the force sensing structures. Kinematic pattern matching has been shown to be highly correlated with footfall events captured by force sensing devices (Stanhope et al., 1990) and as such only a single right and left footfall at each gait speed are necessary to run the pattern matching algorithm. Contact events occurring on the gait mat coincide with
kinematic pattern matching within four microseconds (one frame) of kinematic data. Kinematic patterns of the foot marker data (49 frames center on an event) shall be used to create pattern-matching templates.

For data set A (healthy young adults) force triggered footfall events will be extracted from the analog data collected and exported with the kinematic record into Visual 3D. For data set B (individuals with MS and their age-matched controls without MS) the time-synchronized recordings from the RSscan gait mat identified footfall events in Visual 3D.

**Visual selection of anterior limit of the anticipated boundary**

In order to study the use of the lower visual field during gait it is necessary to know the size of the visual field and have a repeatable measure of head orientation. The visual field is approximately 110° in the vertical plane, with approximately 45° of the field above the neutral plane (0° in Figure 1.5A) and 65° below (Patla, 1997). Measuring head pitch during gait relative to its neutral plane will enable the 110° vertical span of the visual field (specifically the -65° of the lower visual field) to be mapped to the cranium. An anatomical based measure of the neutral plane known as the *Frankfurt Plane* will provide a repeatable identification of the 0° plane of the visual field (Figure 1.6 inset). The Frankfurt Plane (*The American heritage medical dictionary 2007*) is a craniometric plane determined by the upper margin of the auditory meatus and the inferior borders of the orbits and describes the most parallel plane of the head relative to the ground (Figure 1.5A: 0°).
Vision is limited in the vertical (and horizontal) plane due to the anatomical bounds of the visual system, namely the orbits of the eyes in the cranium. We cannot see everywhere at once and if the head is kept level, there is a period before our swinging foot re-contacts the ground, where the step-landing area is no longer visible as it is below the lower limit of the field of view. At the point when we no longer see the step-landing area, stepping becomes purely anticipatory. This anticipatory period of swing co-occurs with the moment when the CoM is no longer over the physical base of support. If the head is pitched forward, the intersection between the lower visual field and the ground is closer to the body and anticipatory margins do not have to be as long in duration. By mapping the intersect of the lower visual field, anchored to the neutral plane of the head, and the ground, temporal margins measured as τ-gaps between the body and this last point seen can be estimated. In this way, a simple transverse plane measurement of head/visual field movement is specified.

The movement of this proximal visual intersect (PVI) has specific patterns of displacement that are clearly identifiable and are hypothesized to relate directly to the perception of τ-gap closure. The PVI has a distinct first derivative velocity profile in the anterior direction during gait (Figure 1.7). The minima in velocity occur twice per gait cycle just after toe-off of each swing-foot and the location of the PVI at these velocity minima is the anticipated base of support and very near the location that eventually becomes the next physical base of support or step-landing area (Figure 3.3).
Figure 3.3. Anterior displacement of the CoM, PVI, and right and left toes over one gait cycle (RHS to RHS) from a healthy adult walking at preferred speed. Note: PVI location at B and C (minimum velocity points in Figure 1.7) appears located near anticipated boundary (dashed lines for emphasis).

The body experiences complex patterns of cyclic motions as it moves through space during gait. To completely specify the motion of the body in space, six independent parameters are required. Relative motion can be uniquely expressed by translation and rotation in the three Cartesian planes: anterior-posterior (A/P), medio-lateral (M/L) and vertical (Z). These directions are also known as surge (A/P), sway (M/L), and heave (Z) (Maritime dictionary 2009). Angular orientations of the body are typically described by its attitude in reference to specified planes: frontal (M/L & Z), sagittal (A/P & Z), and transverse (A/P & M/L). These rotations are also known as roll (frontal), pitch (sagittal), and yaw (transverse). Each of these six quantities can be measured for each segment of the body. It is this inherent complexity of human motion that drives the need for summary measures that allow
the interpretation of gait dynamics. The PVI provides an outcome variable that
describes the three translations and three rotations of the head segment as by
motion of a 2-dimensional point.

When the PVI changes velocity (Figure 1.7), the center of expansion in the
visual field changes location. It is hypothesized that when PVI velocity is high, the
optic flow is mostly parallel (Figure 3.4A), and that when the PVI velocity is low, the
location of the PVI at that moment describes a temporary centering of the optic flow
field at the step-landing area (Figure 3.4B). In this way, when the center of
expansion is near the lower limit of the visual field, τ-gap describes the temporal
closure between the head and the step-landing area (anterior limit of the
anticipated base of support). It is further hypothesized that this temporary center of
expansion provides a person the information needed to predict a step-landing area.
The minima in PVI velocity occur soon after toe-off, suggesting a time dependent
relationship between the pause of the PVI and the formation of a step in progress.

Figure 3.4. A) parallel flow moving past the PVI at high velocity, B) centered optic
expansion when the PVI velocity is nearly zero.
Time to contact between the CoM, and physical and anticipated boundaries

Increased time spent in dual support increases the time spent by the CoM over the physical base of support. The temporal margin to the anterior limit of the physical base of support during dual support phase measured with TtC will allow a greater understanding into how long a person chooses to spend in dual support. In the dual support phase, the TtC to the anterior limit of the physical base of support (right foot toe, Figure 3.2A) by the CoM will be computed until the CoM makes contact with that anterior limit (TtC=0). The CoM contact with the anterior limit of the physical base of support may occur after toe-off of the left foot.

As soon as the TtC to the physical base of support equals zero, the body enters the unstable equilibrium of walking gait and placing the swing-foot near an anticipated step-landing location recaptures the CoM. As the PVI describes where this location is on the transverse plane, and τ-gap may describe the temporal margin between the body and that anticipated base of support location, the swing-foot must be placed near the area that has been deemed appropriate for foot placement in a timely manner. Since the PVI has moved beyond the step-landing area by the time the swing-foot arrives, the location is no longer visible and TtC may provide a more appropriate description of motion. During the swing-phase of the left foot the anticipated base of support is formed between the heel of the stance-foot (right) and the anticipated step-landing area of the left foot (Figure 3.2B). The exact location and timing of swing termination will be extracted from the kinematic record by going forward in the time record data to locate where and when the foot lands on the ground. The step-landing area signifies the preceding τ-gap computation. First, a
TtC computation will be performed on the motion-gap between the CoM and the anticipated base of support anterior boundary. Next, a second TtC computation will be performed between the swing-foot toe-marker and the anticipated base of support anterior boundary. Finally, the coupling ratio (K, Equation 1.1) of the movement will be estimated for the slope of the regression line between the two temporal gaps.

**Data set A: Studies 1 and 2**

Data set A will form the basis for addressing the research questions and hypotheses in studies one and two. All participants volunteering for this collection will be included in both study one and two.

**Participants**

Participants will consist of 18 young healthy males and females between 21 and 40 years of age. Sample size estimate was performed to ensure a statistical power of 0.80 such that detection of differences greater than .05 seconds in the temporal measures with 95% confidence is possible (Eng, 2003). Participants must have a body mass index (BMI) in the normal range (18-25) and not wear glasses for vision correction (contact lenses are acceptable). Approval for the participation of human subjects in this investigation will be obtained from the University of Massachusetts School of Public Health and Health Sciences Human Institutional Review Committee.
Walking Protocol

Each participant will be requested to walk at their preferred pace six times across the laboratory collection area with verbal instructions to walk at a comfortable speed. The same instruction to walk at a comfortable speed will be given for both (1) unobstructed walking trials and (2) obstacle clearing (stepping over) trials. Six-meter trap times of the trial shall be recorded to measure average gait velocity. Speed manipulation trials will follow preferred walking speed trials in random order. Velocity trap times for the six-meter transit shall be set to yield the following average gait speeds: slow (0.6 m/s), medium (1.0 m/s), and fast (1.4 m/s). Successful trials will be within 0.05 m/s of the target speed. Participants will perform each speed condition until four successful times are recorded.

Study 1

Introduction

The purpose of this study will be to examine motion of the head relative to the ground during walking at preferred speed that may guide the selection of anticipated step-landing areas, and how the approach to that location may be guided by perception of a temporal τ-gap. Second, to examine approaches of the body CoM towards the physical and anticipated base of support locations that form during each step using a temporal gap-closure technique called time-to-contact (TtC). Third, during the closure to the anticipated base of support a coupling ratio is explored between the step-landing area, the CoM and the swing-foot. These
characterizations of gait are to be made from the analysis of gait patterns typical to young healthy adults walking at their preferred speed.

**Statistical Analysis**

Study 1 will allow development of the $\tau$-gap and TtC measures for studying the dynamics of gait. The statistical analysis will consist of (1) descriptive statistics of tau-gap, TtC and $K$, and (2) t-tests assessing differences these measures as well as step to step variability of these measures between unobstructed and obstacle conditions. The hypothesis relating to the obstacle trials will be tested with a within participant repeated measures analysis of variance (measures repeated under unconstrained and obstacle conditions) to determine the presence or absence of significant differences in $\tau$-gap, TtC and TtC coupling behavior across speeds. $P$-values, $F$-values, degrees of freedom, 95% confidence intervals and effect sizes will be reported.

**Study 2**

**Introduction**

The purpose of this study will be to discern how rules governing the formation of steps during walking described in study one, regarding temporal gap-closure of the visual selection of step-landing areas and body CoM motion toward physical and anticipated base of support locations, change when the same group of participants from study one varies gait speed. Changes in patterns of visual step-landing selection and approach of the body CoM towards the physical and
anticipated base of support locations at different gait speeds will allow for
descriptions of how non-impaired people typically alter gait from their preferred
rate of locomotion. Secondly, a profile of coupling ratios across gait speeds between
the anticipated base of support, the body CoM and the swing-foot itself may reveal
how changes in gait speed affect the ability to form a physical boundary with the
swing-foot before the CoM arrives. Finally, the influence of walking gait speed on the
visual selection of step landing area can be related to head pitch and the manner in
which the body is carried.

**Statistical Analysis**

Study two consists of a speed manipulation portion of data set A and as each
individual performs all speed conditions, performed in a random order. All
hypotheses in this study will be tested with a within participant repeated measures
analysis of variance (measures repeated at each walking speed) to determine the
presence or absence of significant differences in $\tau$-gap, TtC and TtC coupling
behavior across speeds. P-values, F-values, degrees of freedom, 95% confidence
intervals and effect sizes will be reported.

**Data set B: Studies 3 and 4**

Data set B will form the basis for addressing the research questions and
hypotheses in studies three and four. All participants volunteering for this collection
will be included in both study three and four.
Participants

Participants will consist of 20 individuals with MS at mild-to-moderate levels of disability (0 < EDSS < 4) and 20 age and gender matched individuals without MS. Sample size estimate was performed to ensure a statistical power of 0.80 such that detection of differences greater than .05 seconds in the temporal measures with 95% confidence is possible (Eng, 2003). Approval for the participation of human subjects in this investigation will be obtained from the University of Massachusetts School of Public Health and Health Sciences Human Subjects Review Committee.

Participants with MS shall have been relapse-free for at least six months before collection, and no participants will have a significant history of lower limb injury. Participants must have a body mass index (BMI) in the normal range (18-25), will have greater than 20/200 vision, and be free from vertigo or other acute balance disorders. Study collection hours will be scheduled for morning sessions to alleviate undue levels of symptomatic fatigue in the MS group.

Functional Assessment

A battery of questionnaires and tests of motor function, sensory system function and cognition shall be administered by the study representatives and will be utilized to characterize the group of participants in data set B.

- The first visit to the laboratory for all participants will begin with reading and signing the informed consent.
- Von Frey fiber filament testing and the Biothesiometer shall be used to assess peripheral sensory sensitivity to touch and vibration. A 128 Hz tuning fork will also be struck and placed in
contact with the foot near the medial side of the first metatarsal and time until cessation of sensed vibration shall be recorded, twice for each foot.

- Motor drive will be assessed by ten-second toe-taps, performed twice with each foot.
- The expanded disability status score (EDSS) will be computed from the self-reported EDSS questionnaire and neurologist administered report.
- With instructions to walk as quickly and safely as possible every participant performed the twenty-five foot timed walk test twice. Estimated steady state walking performance will be collected at normal and "brisk" speeds over the center 20 meters of a 30 m hallway.
- A nine-hole peg test will be administered and timed, in order to assess upper limb coordination, repeated two times with each hand, starting with the dominant hand and alternating to the non-dominant.
- A paced serial addition test will be administered once to each participant as a measure of cognitive function.

**Walking Protocol**

Each participant will begin the study by walking at their preferred walking speed, followed in random order by slow (0.6 m/s), medium (1.0 m/s) and fast (1.4
m/s) walking speed trials. Participants will be verbally instructed to walk at a comfortable speed such as one used outdoors on a pleasant day. Speed manipulation trials will follow preferred walking speed trials in random order. Velocity trap times for the six-meter transit shall be set to yield average gait speeds: slow (0.6 m/s), medium (1.0 m/s), and fast (1.4 m/s) speeds. Successful trials will be within 0.05 m/s of the target speed. Participants will perform each speed condition until four successful times are recorded.

**Study 3**

**Introduction**

The purpose of this study is to examine the spatio-temporal walking patterns of individuals with MS across a range of gait speeds. As individuals with MS often walk with a different gait speed than their healthy counterparts, this study allows direct comparisons of gait kinematics without the confounding influence of differences in gait speed.

**Statistical Analysis**

Study three compares the preferred speed condition and the results of the speed manipulation between the groups of participants of data set B (mild-to-moderate disability due to MS and age-matched non-MS controls). Two separate analyses will be performed: (1) t-tests to assess differences in preferred walking speed, stride time, stride length, dual support time between the MS and non-MS groups, and (2) a within participant repeated measures analysis of variance
ANOVA) to determine main group effects and interactions in these same parameters with changes in walking speed.

**Study 4**

**Introduction**

The purpose of this study is to discern how results on temporal gap closures in studies one and two on young healthy adults compare with the same speed profiles of temporal gap closure in adults with mild-to-moderate disability due to MS. Differences in pattern of approach of the body CoM towards the physical base of support and anticipated base of support locations forming during each step with TtC across a range of gait speeds will allow for descriptions of how balance impaired people alter gait and arrive at their current preferred rate of locomotion. Secondly, a comparison of coupling ratio profiles across gait speeds between the anticipated base of support, the body CoM and the swing-foot itself may reveal how MS may affect the strategy of forming a physical boundary with the swing-foot before the CoM arrives. Finally, the influence of gait speed on the visual selection of step landing area can be related to head pitch using τ-gap.

**Walking Protocol**

Each participant will begin the study by walking at their preferred walking speed, followed in random order by slow (0.6 m/s), medium (1.0 m/s) and fast (1.4 m/s) walking speeds. Participants will walk on the same boardwalk described in study one.
**Statistical Analysis**

Study four compares the preferred speed condition and the results of the speed manipulation between the groups of participants of data set B (mild-to-moderate disability due to MS and age-matched non-MS controls). Two separate analyses will be performed: (1) t-tests to assess differences in tau-gap, and TtC between CoM and the anticipated base of support, TtC between swing foot and the anticipated base of support and their coupling constant K, trunk lean and head pitch between the MS and non-MS groups, and (2) a within participant repeated measures analysis of variance (ANOVA) to determine main group effects and interactions in these same parameters with increases in walking speed.

**Modifications to study design**

Chapters 1 through 3 include the dissertation proposal, as submitted to the Graduate School in February 2010. After the analysis of Study 1 results and the development of Study 2 conclusions, it was decided that Study 3 would no longer be conducted as part of the dissertation. The reasons for this decision were first, the response of novel measures developed in Study 2 to fixed walking speeds was observable in the control group in proposed Study 4, and second there was no a priori difference expected in the response to fixing walking speeds between the young healthy controls of Study 2 and the healthy adults (control group) in Study 4. The healthy adult control group of Studies 1 and 4 were older than the young healthy group of Studies 2 and proposed Study 3, but showed no other between group differences. The response to fixing walking speeds at slow (0.6 m/s), medium
(1.0 m/s) and fast (1.4 m/s) walking speeds by the control group of healthy adults of Study 4 is included and discussed in Chapter 6. A complementary assessment of the Proximal Visual Intersect behavior was included as Appendix B-D.


CHAPTER IV

GAIT IMPAIRMENTS IN PERSONS WITH MULTIPLE SCLEROSIS ACROSS PREFERRED AND FIXED WALKING SPEEDS

Abstract

Individuals with multiple sclerosis (MS) have been observed to walk with a slower preferred speed and a longer dual support phase of walking. The purpose of this study was to investigate (1) whether previously observed changes in gait parameters in individuals with multiple sclerosis (MS) are the result of slower preferred walking speeds or reflect adaptations independent of gait speed; and (2) the changes in spatio-temporal features of the unstable swing phase of gait in people with MS. This was accomplished by conducting a cross-sectional study to assess changes in gait parameters during preferred, slow (0.6 m/s), medium (1.0 m/s) and fast (1.4 m/s) walking speeds performed in a gait laboratory with instrumented walkway and motion capture system. The MS group had mild-to-moderate impairment (n=19, 16 females) with a median Expanded Disability Status Scale score (EDSS) of 3.75 (range 2.5 to 6), and the control group was gender and age matched (n=19). Parameters of gait investigated included gait speed, stride length, and stride width, cadence, dual support time, swing time, and timing of swing foot and body/head center of mass during swing phase. Individuals with MS walked at slower preferred speeds with longer dual support times compared with controls. In fixed-speed conditions, dual support times were longer and swing times were shorter in MS compared with controls. Stride width was wider for all speed conditions in the MS group. In the fixed speed conditions, the MS group positioned
their head and body centers of mass closer to the anterior base of support boundary when entering the unstable equilibrium of the swing phase. Longer dual support time is part of a gait strategy in MS that is apparent even when controlling for the confounding effect of slower preferred speed. However, a gait strategy featuring longer dual support times may have limitations if potentially destabilizing swing dynamics exist, which especially occur at walking speeds other than preferred for people with MS.

**Introduction**

In multiple sclerosis (MS) central nervous system signals are disrupted after the myelin sheaths surrounding neurons are damaged by an autoimmune response (Noseworthy et al., 2000). MS can affect sensory and motor pathways leading to diminished perception, vestibular sense and control of muscles that can disturb balance and coordination (Frzovic et al., 2000; Noseworthy et al., 2000; Remelius et al., 2008; Van Emmerik et al., 2010). This balance disruption can lead to impaired walking ability and reduced mobility, which is ranked as one of the most important factors for maintaining a good quality of life by individuals with MS (Heesen et al., 2008). The ability to walk is compromised in those with MS, as over 50% of the MS population experiences falls during activities of daily living (Finlayson et al., 2006). These falls occur despite alterations to gait that may be adopted as part of a protective strategy to increase stability during walking (Benedetti et al., 1999). A protective gait strategy is generally non-specific, yet is characterized by slower gait speeds and wider strides during walking gait (Thompson & Nutt, 2007).
Even with mild disability due to MS, individuals adapt gait such that they walk with slower preferred speeds, shorter stride lengths, longer dual support times and altered lower limb kinematics (Benedetti et al., 1999; Gianfrancesco et al., 2011; Kelleher et al., 2010; Martin et al., 2006; Sacco et al., 2010). Walking speed directly influences parameters of gait, including dual support time, stride length, and cadence (Kirtley et al., 1985; Winter, 1983). As such, it remains unclear whether the changes in gait observed in previous studies of walking by individuals with MS are due to slower walking speed or reflect MS-related functional impairments. What is needed is an assessment of the changes in gait parameters in people with MS under controlled speed conditions at speeds other than preferred, to determine whether altered gait patterns reflect adaptations other than walking speed.

When more time is spent with both feet on the ground in dual support, swing foot dynamics intended to hasten the re-establishment of the dual support posture of gait may alter other gait dynamics such that the benefit gained from a longer dual support strategy may be offset, leading to more frequent falls. The unstable equilibrium of swing is defined as the portion of the gait cycle when the whole body center of mass ($\text{CoM}_{\text{body}}$) is positioned in front of the base of support formed by the stance foot in the anterior-posterior direction (Winter, 1983). The anterior boundary of the base of support is the toe of the lead stance foot. Walking has been described as a series of controlled forward falls that occur when the body enters an unstable equilibrium during the swing phase (Winter, 1983). During swing the $\text{CoM}_{\text{body}}$ transits beyond the anterior boundary of the base of support (formed by the toes of the stance foot; ‘$B_p$’ in Figure 4.1) and the body enters the unstable
equilibrium, which ends when the $\text{CoM}_{\text{body}}$ is recaptured by a new base of support formed at swing foot heel strike. During this phase an individual can optimize recovery from gait disturbances by ensuring that the swing foot is among the first segments of the body to move beyond the anterior boundary, thereby keeping the swing foot beneath the body, albeit in the air. Therefore, it is critical to understand the relative timing and sequence in which salient segments cross beyond the boundary of the base of support during the controlled fall phase of gait.

Figure 4.1. A) MS, and B) control example postures at the swing foot crossing of the $B_p$ while walking at fast speed.
The purpose of this study was to determine whether reported changes in gait parameters in people with MS compared to those without MS are the result of slower preferred gait speed or are the result of MS-related adaptations to gait. Secondly, this study reported on intersegmental dynamics related to the formation of the swing phase of gait. Gait data were collected during preferred and fixed-speed walking trials in a group of individuals with mild-to-moderate functional limitations due to MS and a group of age- and gender-matched non-MS controls. Gait parameters that were assessed included spatial (stride length and width) and temporal (cadence, dual support time, swing time, and intersegmental timings for anterior boundary \( B_p \) crossings by the \( \text{CoM}_{\text{body}} \) and \( \text{CoM}_{\text{head}} \) relative to the swing foot) variables. We hypothesized that if earlier reported changes in gait parameters (dual support and swing times) are only observed at preferred speed, then speed is considered a confounding factor in earlier studies. In contrast, if differences in gait parameters are found in the fixed-speed conditions, then the observed differences in stride parameters may be attributed to MS related adaptations that are not due to differences in walking speed.

**Methods**

**Participant Characteristics**

Nineteen volunteers (16 females) with mild-to-moderate impairment due to MS participated in this study (Table 4.1). The MS group evaluated their disability using the self-administered Expanded Disability Status Scale (EDSS) (Bowen et al., 2001) (median=3.75; range 2.5 to 6). None of the participants with MS reported
exacerbations of symptoms in the previous six months, visual acuity less than 20/200 or were non-ambulatory. The distribution of the MS subtypes was: primary progressive [2], relapsing remitting [15], and secondary-progressive [2].

Participants walked unassisted during data collection. The time elapsed since diagnosis ranged from 2 to 26 years (mean=10.3; SD=7.8), and participants were taking the following medications: immunomodulators [15 participants], antidepressants [8], anti-spasticity [6], anti-fatigue [5], anti-convulsive/sedative [2], and anti-vertigo [1].

Table 4.1. Group characteristics, vibration threshold, pressure sensitivity (grams), visual analog fatigue score, fatigue severity scale, foot taps in 10 s, and normal and brisk functional assessment gait recording speed, statures per second, cadence and stride length for MS and control groups. Reported: Mean ± Standard Deviation, F value (degrees of freedom), p value, and 95% Confidence Interval (CI).

<table>
<thead>
<tr>
<th>Participant Characteristics and Functional Assessment</th>
<th>MS</th>
<th>Controls</th>
<th>F(1,36)</th>
<th>p</th>
<th>95% CI</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>51.3 (10.5)</td>
<td>51.8 (11.5)</td>
<td>0.01</td>
<td>0.930</td>
<td>-7.6</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>167.6 (6.7)</td>
<td>164.8 (6.7)</td>
<td>1.70</td>
<td>0.201</td>
<td>-7.5</td>
</tr>
<tr>
<td>Body mass (kg)</td>
<td>75.65 (16.6)</td>
<td>67.39 (10.0)</td>
<td>4.96</td>
<td>0.032</td>
<td>-41.13</td>
</tr>
<tr>
<td>Vibration threshold (volts)</td>
<td>17.7 (13.1)</td>
<td>8.9 (6.4)</td>
<td>6.79</td>
<td>0.013</td>
<td>-15.4</td>
</tr>
<tr>
<td>Pressure threshold (gr)</td>
<td>4.64 (10.23)</td>
<td>0.65 (.70)</td>
<td>2.88</td>
<td>0.098</td>
<td>-8.76</td>
</tr>
<tr>
<td>Fatigue score (VAFS cm)</td>
<td>3.88 (2.39)</td>
<td>2.33 (1.88)</td>
<td>6.71</td>
<td>0.014</td>
<td>-3.18</td>
</tr>
<tr>
<td>Fatigue Severity Scale</td>
<td>5.6 (1.2)</td>
<td>2.7 (1.1)</td>
<td>3.2 &lt;0.001</td>
<td>1.9</td>
<td>3.9</td>
</tr>
<tr>
<td>Foot taps (in 10 s)</td>
<td>34.3 (9.1)</td>
<td>48.4 (9.9)</td>
<td>17.70 &lt;0.001</td>
<td>6.6</td>
<td>18.8</td>
</tr>
<tr>
<td>Normal speed (m/s)</td>
<td>1.33 (.19)</td>
<td>1.46 (.21)</td>
<td>3.79</td>
<td>0.059</td>
<td>-0.01</td>
</tr>
<tr>
<td>Normal speed (statures/s)</td>
<td>0.799 (.123)</td>
<td>0.890 (.134)</td>
<td>4.89</td>
<td>0.033</td>
<td>0.008</td>
</tr>
<tr>
<td>Normal cadence (steps/min)</td>
<td>111.7 10.4</td>
<td>118.5 11.6</td>
<td>0.01</td>
<td>0.065</td>
<td>-14.0</td>
</tr>
<tr>
<td>Normal stride length (cm)</td>
<td>1440 (18)</td>
<td>1480 (10)</td>
<td>0.63</td>
<td>0.431</td>
<td>-29</td>
</tr>
<tr>
<td>Brisk speed (m/s)</td>
<td>1.63 (.24)</td>
<td>1.91 (.27)</td>
<td>11.51</td>
<td>0.002</td>
<td>0.11</td>
</tr>
<tr>
<td>Brisk speed (statures/s)</td>
<td>0.973 (.150)</td>
<td>1.157 (.150)</td>
<td>13.56</td>
<td>0.001</td>
<td>0.083</td>
</tr>
<tr>
<td>Brisk cadence (steps/min)</td>
<td>122.8 (14.2)</td>
<td>137.7 (12.7)</td>
<td>1.38</td>
<td>0.002</td>
<td>-23.7</td>
</tr>
<tr>
<td>Brisk stride length (cm)</td>
<td>1593 (26)</td>
<td>1661 (13)</td>
<td>0.94</td>
<td>0.339</td>
<td>-35</td>
</tr>
</tbody>
</table>
Control participants were age- and gender-matched to those in the MS group (Table 4.1). All participants were otherwise healthy (no orthopedic or other neurological problems besides MS) and sedentary to recreationally active (≤30 min structured exercise a day, three days per week). All participants gave informed consent, as approved by the Institutional Review Board from the University of Massachusetts, Amherst, before testing.

**Protocol**

Participants were tested on two separate visits. In the first visit, participants underwent a functional assessment. Peripheral sensitivity to vibration was assessed with a Biothesiometer (Bio-Medical Instrument Co., OH) as the voltage setting at which vibration is first perceived; and to light touch pressure with a single filament test protocol (North Coast Medical Inc., CA) as force in grams of the smallest filament perceived. Vibration and touch pressure threshold values were reported as an average from bilateral tests on the ball, arch and heel of each foot (Table 4.1). Motor drive was measured as the average number (two repetitions) of foot-taps each participant was able to perform in 10 s with each foot (Kent-Braun et al., 1998) (Table 4.1). Each participant was asked to walk 20 m along a hallway, twice at a “normal” unhurried pace and twice at a faster, “brisk” pace (not as fast as possible). Average speed, cadence and stride length were assessed (Table 4.1). Acute symptomatic fatigue was assessed with a visual analog fatigue scale (VAFS) at the start of the first visit, by marking the current degree of fatigue on a 10-cm scale...
(Table 4.1). The Fatigue Severity Scale measured chronic fatigue (FSS; Krupp et al., 1989).

During the second visit, kinematic and pressure mat recordings were collected for 4 preferred walking speed trials. These were followed in random order by trials at fixed-speeds: slow (0.6 m/s), medium (1.0 m/s), and fast (1.4 m/s). Participants were provided verbal feedback about their speed relative to the target speed. Individuals were requested to walk at each speed until 4 recordings were obtained that fell within ±5% of the target speeds. Total number of walking trials, including those that did not match the targeted speed, was on average 23 (±3) for the MS and 25 (±3) for the control group. Participants wore lab-supplied running shoes.

**Experimental Setup for Gait Collections**

Qualisys track manager software was used to synchronize eight Oqus cameras (Qualisys AB, Sweden) sampled at 240 Hz with a 0.5 x 2.0 m pressure mat (RSscan, Belgium) sampled at 60 Hz. The pressure mat was located in the center of a 0.7 x 14 m wooden walkway and served to identify bilateral heel-strike and toe-off events. A 6 m optical trap provided average gait speed estimates on each gait trial.

Segment kinematics were tracked with fifty 10-mm retro-reflective markers. A 13-segment 3D biomechanical model, based on established anthropometric data (Clauser et al., 1969), was constructed from the kinematic data in order to estimate the position of the CoMbody, the CoMhead and the distal end of each foot during gait collections (Figure 4.1).
**Gait Data Analysis**

Visual 3D (C-Motion, Inc. MD) was used to build the model, filter marker data (8 Hz low-pass bidirectional zero lag) and post-process gait parameters: gait speed, height normalized gait speed (statures/s) (Winter, 1983), stride length, stride width, cadence, dual support time, swing time (total: left and right leg combined), and percent of stride cycle spent in dual support and swing phases.

The intersegmental timing observations were computed for the swing foot toe, CoM$_{body}$, and the center of mass of the head (CoM$_{head}$). These features were selected because the CoM$_{body}$ describes the posture of the entire body relative to the ground (Winter, 1983; Winter, 1995), while keeping the CoM$_{head}$ stable is essential for normal gait (Latt et al., 2008; Peters et al., 2006; Pozzo et al., 1990). The time difference between swing foot crossing of the Bp in comparison to the CoM$_{head}$ and CoM$_{body}$ was used to characterize entry into the unstable equilibrium state during each step. A positive time difference indicates the reference variable (i.e., CoM$_{head}$ or CoM$_{body}$) passed the Bp before the swing foot and a negative time difference indicates it followed after the swing foot crossing (Figure 4.1-2).

**Statistical Analysis**

Differences between groups for preferred speed parameters were tested for significance (p<0.05) using a one-way ANOVA, and for the fixed-speed condition parameters with a 2-factor (group, speed) mixed model ANOVA. No a priori predictions were made concerning group differences at individual speeds, but individual t-tests to assess group differences at each speed were also performed as
an exploratory analysis. F-values with degrees of freedom, p-values, and 95% confidence intervals for group mean differences were reported.

**Results**

**Functional Assessment**

There were no group differences in age, height, or filament pressure threshold (Table 4.1). Compared with controls, the MS group was heavier, had diminished motor drive (slower foot tap speed), and higher vibration threshold and levels of acute (VAFS) and chronic (FSS) fatigue (Table 1). In the 20 m hallway walk test, the MS group walked more slowly in both “normal” and “brisk” speed conditions (Table 4.1).

**Preferred Walking Speed Condition**

In the preferred walking speed condition, no significant differences existed between the groups, but the MS group tended to select slower walking speeds compared with controls (Table 4.2). Despite this difference, both groups walked with similar stride length, and swing time. Compared with controls, the MS group had longer dual support time (Figure 4.3), greater percent of cycle spent in dual support time, lesser percent of cycle in swing time, and walked with wider strides (Table 4.2). There was a trend for the MS group to walk with a slower cadence. These results were not influenced after adjusting (analysis of covariance) for fatigue levels measured during the first laboratory visit.
The timing measures indicating entrance into the unstable equilibrium of swing showed that during the preferred speed trials, the $\text{CoM}_{\text{head}}$ crossed the Bp at nearly the same moment as the swing foot in both groups. Likewise, the $\text{CoM}_{\text{body}}$ crossed the Bp after the swing foot with a similar temporal margin in both groups (Figures 4.1-4.2, Table 4.2).

<table>
<thead>
<tr>
<th>Preferred Speed</th>
<th>MS</th>
<th>Controls</th>
<th>F(1,36)</th>
<th>p</th>
<th>95% CI</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gait speed (m/s)</td>
<td>1.26 (.23)</td>
<td>1.39 (.21)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Gait speed (statures/s)</td>
<td>0.75 (.15)</td>
<td>0.82 (.13)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Cadence (steps/min)</td>
<td>108.9 (8.3)</td>
<td>113.7 (10.2)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stride length (cm)</td>
<td>1381 (185)</td>
<td>1459 (122)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stride width (cm)</td>
<td>14.8 (2.5)</td>
<td>12.4 (1.9)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Dual support time (ms)</td>
<td>403 (66)</td>
<td>346 (51)</td>
<td>8.88</td>
<td>0.005</td>
<td>18 96</td>
</tr>
<tr>
<td>Swing time (ms)</td>
<td>709 (57)</td>
<td>721 (57)</td>
<td>0.51</td>
<td>0.480</td>
<td>-49 24</td>
</tr>
<tr>
<td>Dual support (% cycle)</td>
<td>36 (4)</td>
<td>32 (3)</td>
<td>10.79</td>
<td>0.002</td>
<td>1 6</td>
</tr>
<tr>
<td>Swing time (% cycle)</td>
<td>64 (4)</td>
<td>68 (3)</td>
<td>0.09</td>
<td>0.772</td>
<td>-13 10</td>
</tr>
<tr>
<td>$\text{CoM}_{\text{head}}$-swing foot (ms)</td>
<td>1 (29)</td>
<td>-9 (24)</td>
<td>1.48</td>
<td>0.232</td>
<td>-7 28</td>
</tr>
<tr>
<td>$\text{CoM}_{\text{body}}$-swing foot (ms)</td>
<td>-38 (16)</td>
<td>-39 (18)</td>
<td>0.09</td>
<td>0.772</td>
<td>-13 10</td>
</tr>
</tbody>
</table>
Figure 4.2. MS and Control time difference (ms) relative to swing foot passing $B_p$ for the head (A) and CoM (B) at preferred, fast, medium and slow walking speeds. Positive numbers indicate segment passing beyond $B_p$ before swing foot; negative numbers indicate segment passing beyond $B_p$ after swing foot. Bars: standard error.

**Fixed Walking Speed Conditions**

For the fixed speed trials, gait speeds were not different between groups and both groups walked with similar cadence and stride length (Table 4.3). There were no group by speed interactions ($p>0.05$). Across fixed speeds, there were significant
group effects such that, compared with controls, the MS group walked with longer
dual support time (Figure 4.3) and shorter swing time in absolute and relative
(percent of cycle) time measures, and walked with wider strides.

![Figure 4.3. Dual Support time as percent of cycle time plotted against gait speeds (m/s) of preferred pace walking trials (open markers), and fast, medium and slow conditions (closed markers). MS: diamonds (solid line); Controls: squares (dashed line). Error bars: standard error (horizontal: for preferred speed conditions; vertical: within group variability).](image)

There were no group by speed interactions (p>0.05) for timing measures reflecting the entrance into the unstable equilibrium of swing. However, there were significant group effects such that the MS group allowed the CoM\textsubscript{head} to pass beyond the B\textsubscript{p} earlier relative to the swing foot than controls for the medium and fast speeds. Additionally, compared with controls, the MS group CoM\textsubscript{body} passed the B\textsubscript{p} sooner after the swing foot crossing, at the faster walking speed (Figures 4.1-4.2, Table 4.3).
Table 4.3. Fixed speed results for MS and Control groups walking at fast, medium, and slow speeds. Speed (m/s and statures/s), stride length, stride width, dual support and swing (time and % cycle), CoM_{head} and CoM_{body} passing the physical boundary relative to swing foot. Reported: Mean ± Standard Deviation, F value (degrees of freedom), p value, and 95% Confidence Interval (CI).

<table>
<thead>
<tr>
<th></th>
<th>Slow Speed (0.6 m/s)</th>
<th>Medium Speed (1.0 m/s)</th>
<th>Fast Speed (1.4 m/s)</th>
<th>Group Main Effect</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>MS</td>
<td>Controls</td>
<td>MS</td>
<td>Controls</td>
</tr>
<tr>
<td>Gait speed (m/s)</td>
<td>0.61 (.01)</td>
<td>0.61 (.01)</td>
<td>1.00 (.01)</td>
<td>1.00 (.01)</td>
</tr>
<tr>
<td>Gait speed (statures/s)</td>
<td>0.36 (.02)</td>
<td>0.37 (.02)</td>
<td>0.60 (.03)</td>
<td>0.61 (.03)</td>
</tr>
<tr>
<td>Cadence (steps/min)</td>
<td>36.3 (2.5)</td>
<td>35.6 (2.9)</td>
<td>48.8 (3.0)</td>
<td>47.9 (2.7)</td>
</tr>
<tr>
<td>Stride length (cm)</td>
<td>101 (7)</td>
<td>104 (9)</td>
<td>124 (8)</td>
<td>128 (6)</td>
</tr>
<tr>
<td>Stride width (cm)</td>
<td>15.0 (3.6)</td>
<td>11.6 (2.1)</td>
<td>14.9 (2.9)</td>
<td>11.9 (2.1)</td>
</tr>
<tr>
<td>Dual support time (ms)</td>
<td>744 (73)</td>
<td>715 (64)</td>
<td>475 (38)</td>
<td>458 (39)</td>
</tr>
<tr>
<td>Swing time (ms)</td>
<td>913 (91)</td>
<td>979 (104)</td>
<td>764 (72)</td>
<td>804 (59)</td>
</tr>
<tr>
<td>Dual support (% cycle)</td>
<td>45 (4)</td>
<td>42 (3)</td>
<td>39 (3)</td>
<td>36 (2)</td>
</tr>
<tr>
<td>Swing time (% cycle)</td>
<td>55 (4)</td>
<td>58 (3)</td>
<td>62 (3)</td>
<td>63 (2)</td>
</tr>
<tr>
<td>CoM_{head}-swing foot (ms)</td>
<td>-41 (60)</td>
<td>-57 (64)</td>
<td>-7 (34)</td>
<td>-19 (33)</td>
</tr>
<tr>
<td>CoM_{body}-swing foot (ms)</td>
<td>-113 (26)</td>
<td>-123 (27)</td>
<td>-52 (16)</td>
<td>-58 (15)</td>
</tr>
</tbody>
</table>
Discussion

The intent of this study was to determine whether previously observed changes in the spatiotemporal parameters of gait in MS resulted from a slower preferred gait speed or instead reflect adaptations due to MS not associated with walking speed differences. The results of this study show for the first time that a longer dual support time is an adaptation favored by those with MS independent of walking speed, suggesting that this gait modification is not related to differences in preferred walking speeds (Benedetti et al., 1999; Gianfrancesco et al., 2011; Kelleher et al., 2010; Martin et al., 2006; Sacco et al., 2010) and instead represents a fundamental adaptation in people with MS. However, the lengthening of the dual support phase may not always increase gait stability, as our data also demonstrate the existence of destabilizing swing phase dynamics, especially occurring at walking speeds other than preferred.

Individuals with MS walked with a slower preferred gait speed, spent more time in dual support and adopted a wider stance during gait, in a manner consistent with earlier reports on MS. In fixed speed conditions the MS group also walked with a longer dual support time and a wider stride compared with controls. These adaptations may represent part of a protective gait strategy employed by individuals with MS to counter instability during walking due to deficits in motor drive, peripheral sensation, or higher levels of symptomatic fatigue (Chung et al., 2008; Mevellec et al., 2003; Thoumie & Mevellec, 2002; Thoumie et al., 2005), which were present in the current MS group (see Table 4.1). By adopting a protective gait strategy to compensate for functional limitations related to MS, individuals may be
attempting to improve the stability of gait. However, spending more time in dual support can affect other aspects of gait timing, including swing time.

Lengthening dual support time in an effort to increase stability during gait will shorten swing time when speed is fixed. An individual must then either shorten step length or increase swing foot velocity to maintain cadence. Here at preferred speed individuals with MS walked with longer dual support time and a similar step length, but trended toward assuming a slower cadence, and as such absolute swing timing was similar to controls. In contrast, when walking at specific speeds, individuals from both groups adopted similar stride lengths and cadences, yet individuals in the MS group shortened swing time in order to spend more of the gait cycle with both feet on the ground (i.e., in dual support). In the fixed speed conditions, shortening swing time while keeping step length equal to those of non-MS controls forced the individuals with MS to move the swing foot faster, which may present a challenge to stability.

Individuals with motor drive deficits such as in MS, may be particularly at risk for encountering gait disturbances during swing, as the swing phase of gait is not a passive movement (Whittlesey, Van Emmerik, & Hamill, 2000). A shorter swing time is associated with increased potential for falls in older adults (Mills & Barrett, 2001), and the breakdown of walking stability associated with MS may be partially explained by the alteration of swing timing, especially when activities of daily living dictate a certain walking speed. Indeed the prevalence of falls for those with MS increases with severity of disease, symptomatic fatigue, heat sensitivity, being in a hurry, or having divided attention, all of which are related to increased
swing variability when walking (Cameron & Lord, 2010; Hamilton et al., 2009; Nilsagard et al., 2009). However, walking more slowly is not a panacea for avoiding excessive gait variability as walking at speeds slower than preferred also increase swing variability (Kang & Dingwell, 2008). Our results show that even at slow walking speeds the swing phase must be reliably orchestrated to ensure that the body enters the unstable equilibrium appropriately, and that this entry into the unstable equilibrium is changed in people with MS.

In the fixed speed conditions, the MS group had the CoMbody positioned closer to the anterior stability boundary (Bp) when the swing foot crossed this boundary compared with controls (Figure 4.1B). However, what may be relevant to gait stability is the CoMhead timing of the MS group. In the fixed speed conditions the MS group had the CoMhead consistently shifted towards crossing the physical boundary earlier compared with controls (Figure 4.1A). This dynamical approach to understanding the coordination of body segments during swing shows that with a linear fit applied, the MS group begins to lead the body past the physical boundary with the CoMhead rather than the swing foot at a slower speed, in comparison to controls (Figure 4.2: MS=1.20 m/s; control=1.43 m/s). This tendency to lead into the unstable equilibrium of swing with the head at slower speeds may have consequences for maintaining overall gait integrity for those with MS.

The narrowing of timing between intersegmental boundary crossings may act to reduce the ‘safety’ margin available to individuals with MS for the placement of corrective steps during the controlled fall of gait, particularly if a slip or trip occurred to challenge balance and potentially lead to a fall. A safety margin in the
swing phase of gait may be ensured when the swing foot is in front of the stance foot when the \( \text{CoM}_{\text{head}} \) and \( \text{CoM}_{\text{body}} \) are entering the unstable equilibrium. In this way an individual can be somewhat protected from gait disturbances by ensuring that the swing foot is among the first segments of the body to move beyond the \( B_p \), thereby keeping the swing foot beneath the body, albeit in the air. Swing time can be lengthened commensurate with shortening dual support when underlying strength deficits are overcome with resistance training (Gutierrez et al., 2005; White et al., 2004). This finding may have significant implications for gait rehabilitation and training in MS.

In a recent study, almost 60% of persons with MS reported falls in the past 6 months, with most falls occurring during transfers, ambulation and standing (Matsuda et al., 2011). The two most prominent self-reported reasons for these falls were trips/slips and fatigue. The current results indicate functional adaptations in gait in people with MS in the form of a wider stance base and longer dual support time. However, the formation of swing, during which trips and slips are most likely to occur is clearly shown here to be affected in MS. Intervention methods that focus on the timing of the swing foot in relation to the position the body \( \text{CoM} \) may be beneficial in minimizing the risk of slips and trips and overall fall risk. The training of the proper coordination or timing between the swing foot and the rest of the body (\( \text{CoM}_{\text{body}}, \text{CoM}_{\text{head}} \)) moving beyond the physical boundary may improve gait stability and adaptations by emphasizing the importance of keeping the swing foot out and under the body when taking a step. In addition, gait rehabilitation strategies that focus on individuals with MS walking at speeds other than preferred could enhance
the breadth of walking speeds available to individuals with MS during their daily living.

**Study Limitations**

A limitation of the current study is the relative heterogeneity of the MS group. Although the participants with MS were all ambulating independently, the range in EDSS scores (2.5-6) is still considerable and a more focused gait assessment stratified by EDSS status may be beneficial.

**Conclusions**

In conclusion, systematically longer dual support times in gait of people with MS may be part of a protective strategy that aids in avoiding gait disturbances in people with MS. However, one of the unanticipated consequences of this longer dual support time is the changing of the swing foot timing, especially in relation to the entry of the body into the unstable equilibrium. Additionally, slowing gait speed alone does not obviate gait disturbances induced by disability due to MS, as results from the current study show that gait changes related to MS are present across a wide range of walking speeds.

**Acknowledgements**

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References


CHAPTER V
CENTER OF MASS CONTROL AND HEAD MOVEMENT IN THE SWING PHASE OF
GAIT IN ADULTS WALKING AT PREFERRED SPEED AND STEPPING OVER AN
OBSTACLE

Abstract

In the swing phase of walking gait, the body is exposed to a dynamically
unstable state. From the point when the center of mass (CoM) passes the stance foot
toe until heel-strike the body is in a controlled anterior fall, which is also referred to
as the unstable equilibrium of gait. This study developed three novel aspects related
to the formation of swing in young healthy adults (n=20) as they walked at their
preferred gait speed over unobstructed ground and while stepping over an obstacle.
First, a description of swing was developed that takes into account the motion of the
body CoM in relation to stability boundaries formed by the based of support at the
feet. Second, a new method to investigate coordination during swing described the
coupling of the swing foot with the CoM as estimated by the temporal stability
boundary approach measure time-to-contact (TtC) during the movement towards the
stance foot toe and heel-strike locations. Third, a method was developed that
measured the contribution of head movements in assessing the field of view and the
step landing area during the formation of swing.

At preferred speed participants walked at 1.33 m/s with a swing time of 0.38
s, and swing phase relative durations were: early-swing (CoM behind stance foot)
22%; mid-swing (CoM above stance foot) 40%; and late-swing (CoM in the unstable
equilibrium) 38%. In the coupling of time-to-contact estimates of the approach to the

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stance foot toe the swing foot surpassed the CoM at 27% of swing. The coupling ratio in the approach to the heel strike location was 0.26 at the point that the CoM entered the unstable equilibrium. A fixation of the field of view determined by head orientation occurred at 29% of swing and the proximal field of view limit was located 7 cm closer to the body than the heel-strike location.

In the obstacle condition, the trail foot step was formed with longer early and mid swing phases and a shorter late swing compared to the step with the lead foot. While the lead swing foot passed the stance foot toe before the CoM, when stepping over the obstacle with the trail foot, the opposite occurred as the trail swing foot passed the stance foot toe after the CoM, which obviated the swing foot coupling to the CoM. Leading into the unstable equilibrium with the CoM may predispose the body to gait disturbances and loss of stability. Furthermore, on the approach to the heel-strike location, a larger coupling ratio (0.36) in the step of the trail foot over the obstacle shows that the CoM will arrive with a closer temporal margin to the swing foot compared to the step with the lead foot (0.26), leaving little time to adapt the gait cycle to any unforeseen disturbances.

A fixation of the field of view determined by head gaze orientation occurred at 20% of swing for the lead foot, and 20% of swing for the trail foot. The proximal field of view limit was located closer to the body than the heel-strike location (83 cm lead foot; 68 cm trail foot).

The phases of swing, the coordination between the swing foot and the CoM, and the dynamics of the field of view were observed to be different in the step with the lead foot over an obstacle compared with the step with the trail foot,
demonstrating the sensitivity of these novel measures to altered patterns of swing that go beyond swing duration or step length. The greatest threat to stability in the obstacle condition may be late swing of the step that brings the trail foot over the obstacle. However, this condition of remaining in an unstable state during the step over an obstacle with the trail foot may be somewhat ameliorated by the fact that the entire step path is inside the field of view, where potential gait disturbances can be visually inspected during swing.

**Introduction**

Walking gait for humans is a cyclic process consisting of two distinct phases, where during a stride cycle we spend about 60% in stance and 40% in swing with each leg to locomote overground (Winter, 1983). A portion of the stance phase taken with each leg overlaps in bipedal walking to form the dual support phase of gait. Dual support is taken in a stride stance, a postural configuration where one foot leads the body and one foot trails it, with the Center of Mass (CoM) positioned between them. When a foot or the feet are touching the ground, the perimeter of contact describes the base of support, and for stride stance, it is the area between the toe of the lead foot and the heel of the trail foot (Figure 5.1A). Dual support typically spans about 30% of a stride cycle and “feels” safe to us, as we are in a relatively stable state in a posture that affords a long anterior-posterior base of support. Accordingly, we then must spend roughly 70% of the gait cycle with one foot or the other off the ground in swing and it may be during the alternating swing of the feet that we are most vulnerable to gait disturbances.
The first challenge to stability in swing is the unipedal stance posture, as switching from dual support entails a sharp reduction in both the medial-lateral and the anterior-posterior length of the base of support. Much of swing is inherently stable in the anterior-posterior direction as after toe-off the CoM is quickly thrust into a position between stance foot heel and toe that define the unipedal base of support (Figure 5.1B). In contrast, the latter portion of swing may present the greatest challenge to stability in gait. The ongoing progression of the body causes the CoM to transit into an unstable state beyond the anterior limit of the base of support in what is often called a “controlled fall” (Winter, 1983). The anterior base of support limit lies at the stance foot toe and is referred to here as the physical boundary (B_p). When the CoM passes beyond this physical boundary the body enters a late swing unstable equilibrium (Figure 5.1C). In the late swing unstable equilibrium the CoM is voluntarily initiated into a controlled fall toward the intended step-landing area, referred to here as the anticipated boundary (B_a). To recapture the CoM within a new physical boundary, the swing foot must be moved appropriately toward, and into contact with, the anticipated boundary such that the stable dual support stride stance is reestablished, and the unstable equilibrium of late swing is terminated (Figure 5.1D). Much of the research on walking gait has focused on the stance phase, dual support, and the kinetic and kinematic patterns of toe-off and heel-strike (Winter, 1983), but less is known about the entry into the unstable equilibrium and recapture of the CoM during swing and its impact on gait stability.
Figure 5.1. Center of Mass (CoM) trajectory during walking gait (thin S-curve line) and example CoM position (bulls-eye) during dual support stance (A&D) and swing (B&C) phases of gait. Vertical bars: physical boundary (dark solid lines in A, B & D) and anticipated boundary (thin line in C). Light grey lines depict posterior physical boundary.

As there are no foot-ground interactions to form kinetic signatures in swing, a better understanding of how swing is constructed may be gained by studying temporal dynamics. In addition to identifying when during swing, the CoM is above the stance foot physical boundary or beyond it (Figure 5.2), pertinent signatures of
coordination may be found in the approach to and ultimate contact with the boundaries that define the limits of the late swing unstable equilibrium. The temporal dynamics of this approach to the base of support can be investigated by the time-to-contact (TtC) with these boundaries. TtC is a continuous estimation of the instantaneous time it may take to make contact with a perceived or physical boundary, given distance, velocity and acceleration of the body. The original TtC concept emerged from vision research that sought to characterize the visual perception of a distance closure (gap closure) as a directly perceivable temporal variable called *Tau* (τ) (Lee & Lishman, 1977). In the postural control domain, the TtC concept was later adopted (Riccio, 1993) to explain proprioceptive-based temporal estimates of the approach to postural stability boundaries. In human posture research, TtC describes the approach of the body’s center of pressure (CoP) or CoM towards and away from the physical base of support (Slobounov et al., 1998). TtC was developed as a method to understand postural activity when both feet remain in place and has evolved from descriptions of quiet standing (Haddad et al., 2006) to behavior during more challenged postures (Fortho et al., 2006; Van Emmerik et al., 2010), the postural phase of gait initiation (Remelius et al., 2008) and the CoM activity preceding a step due to a postural perturbation (Hasson et al., 2008; Hasson et al., 2009). However, TtC holds potential for use in tasks that are more dynamic where the physical boundary changes as in locomotion.
Figure 5.2. Apportioning swing: Early swing (toe-off to CoM above heel of stance foot); Mid-swing (CoM above stance foot); Late swing (CoM passing stance foot toe to heel-strike). CoM depicted as a target at beginning and end of mid-swing.

Estimates of when the body as a whole will contact a boundary may be described with TtC, but as the body is composed of many segments that are coincidently moving, a description of how the motions of these segments are coupled is warranted if underlying coordination patterns are to be better understood. A *coupling-gap* is a description of how two motion-gaps interact. This ratio (K) describes how two motion-gaps (TtC of the swing foot and TtC of the CoM) closing in towards a single boundary (Figure 5.1C) are coupled (Lee et al., 2001).
Figure 5.3. Proximal Visual Intersect (PVI) with the transverse plane (A), formed by a 65-degree declination from the neutral plane of the head (A inset). Anterior progression of the PVI and the feet during walking gait (B top), anterior PVI velocity during gait (B bottom) showing systematic slowing of the PVI after toe-off.

Coupling of τ-gaps has been used to describe the first step of walking, where the support of the body has been simplified as being located at the center of pressure. Austad & van der Meer (2007) computed τ-gap to a new physical boundary in the first step of walking, and showed that closure rates change during early development. Austad & van der Meer used tau-gap coupling in the initiation of gait and the approach to a physical boundary, but how gaps are coupled during steady state gait has not been investigated.

During walking, the step landing area and the movement toward this subsequent anticipated boundary location may be organized by information gained from a visually perceived tau-gap. This tau-gap closure to the anticipated boundary of swing may provide important cues related to the approach of the body towards that location. Perception of self-motion extracted from visual information is a form
of proprioception that relies on *optic flow* (Patla, 1991; Marigold, 2008). Optic flow is the pattern of apparent motion of features in the environment caused by relative motion of the observer. Optic flow for sighted individuals is shaped by overall body movement, and compensatory head movements serve to functionally stabilize the visual field and allow for perception of stationary features on the ground (Berg & Mark, 2005; Bradshaw & Sparrow, 2001; Buckley et al., 2008; Gibson, 1958; Marigold & Patla, 2008; Owen & Lee, 2004; Patla et al., 1999; Reynolds & Day, 2005; Warren et al., 1986). The visual perception of closure towards an anticipated boundary is the underlying impetus for the *Tau-gap* theory that characterizes boundary approach as a temporal variable. In this way a tau-gap estimate of self-motion may reinforce the idea that the anticipated boundary of swing is actively selected, and moved toward in a purposeful manner. Here the intersection between the inferior limit of the visual field and the ground in front of the body is termed the *proximal visual intersect* (PVI; Figure 5.3A) and any point more proximal to the body from this PVI is not in the field of view. However, the inferior limit of the field of view is not necessarily the center of optic expansion, or the center of the visual flow field. The value in describing the PVI lays in the ability to describe when during swing the PVI is making its slowest anterior progression, which may yield the most distinct visual cues for the perception of self-motion overground (Figure 5.3B).

The purpose of this study was to develop techniques that characterize the swing phase of gait as a sequence of boundary approaches using time-to-contact and tau-gap temporal estimates. To do this, a new definition of early, mid, and late swing was put forth, beginning at toe-off and demarked by the body CoM passing
the heel and toe boundaries of the stance foot in the transit to the anticipated boundary, ending at heel-strike (Figure 5.2). Once it was better understood when the CoM was positioned behind, above and in front of the stance foot in swing, three main assessments of the dynamics of swing were conducted: 1) the time-to-contact of the swing foot and CoM and their coupling during the approach to the unstable equilibrium (Figure 5.1B); 2) the time-to-contact of the swing foot and CoM and their coupling from entry to the unstable equilibrium until the recapture of the CoM at heel-strike (Figure 5.1C); and 3) the contribution of head movement in the formation of swing by describing how the anterior velocity profile of the PVI helps in assessing the field of view and the next heel-strike location. These characterizations of gait were made from the analysis of gait patterns of young healthy adults walking at their preferred speed over unobstructed ground. Participants were also asked to step over an obstacle in their path in order to understand how features in the environment may influence physical and anticipated boundary approaches and selection of step-landing areas.

**Methods**

**Participant Characteristics**

Twenty young healthy adults participated in this study (10 males and 10 females). The group was a sample of convenience, selected from the student body of graduate and undergraduate Kinesiology members, without neurological or orthopedic problems. The group ranged in age from 20 to 38 years (mean 28.3±5.0), in height from 159 to 183 cm (170±8.1), and in weight from 54.4 to 95.3 kg.
(68.6±12.4). All participants signed an informed consent document approved by the Institutional Review Board of the University of Massachusetts, Amherst.

Walking Gait Protocol

Participants walked along a 10 m walkway featuring a 1 meter wide black rubber floor mat, and grey curtains hung at either end of the walkway to minimize visual distraction during the collection. The kinematic capture volume consisted of the middle 5 m of the walkway. Instructions for preferred walking speed were to walk at a normal unhurried pace, and to “look where you are going” during the transit of the walkway. During obstacle trial collections, a 10 cm diameter tan cardboard tube was placed across the center of the walkway and the participant was advised to please avoid the obstacle to prevent tripping over it. Each condition was performed four times. Participants wore lab-supplied running shoes during data collections.

Experimental Setup for Gait Collections

The Qualisys track manager software was used to synchronize eight Oqus cameras (Qualisys AB, Sweden) with a 1.2 x .6 m force plate at 240 Hz (AMTI, MA). The force plate served to identify bilateral heel-strike and toe-off events and was located in the center of the volume. Segment kinematics were tracked with fifty 10 mm retro-reflective markers. A pointer rod was used to create virtual landmarks for reference to joint centers.
A 13-segment 3D biomechanical model, based on established anthropometric data (Clauser et al., 1969), was constructed from the kinematic data in order to estimate the position of the CoM\textsubscript{body}, the orientation of the head, and the distal end of each foot during gait collections (Figure 5.2). The head was located at pointer events demarking the bilateral poria, a location that closely approximates the medio-lateral axis of the vestibular system. The head segment was oriented to align with the neutral plane of the head, as defined by the Frankfurt Plane (Figure 5.3A; inset): from the poria to the bilateral occipital ridges (Johnson, 1950). The head was tracked with a rigid cluster of five markers worn on the crown. The torso, pelvis, and upper/lower arm and leg segments were built from anatomical landmark events and tracked with marker clusters. The feet were built from landmarks at the tibial medial malleolus and fibular lateral malleolus, and 1\textsuperscript{st} and 5\textsuperscript{th} metatarsal markers, and tracked with the metatarsal markers and a heel marker cluster. A 6 m optical trap provided gait speed estimates on each gait trial.

**Data Analysis**

**Gait Parameters**

Gait speed, step length, cycle time, swing time and positions of the CoM and feet were extracted with Visual 3D software (C-Motion, MD). For the preferred speed walking data, gait parameters were computed from one bilateral gait cycle per trial, averaged over the four trials. In the obstacle crossing data, gait parameters were computed for the stride cycle of the lead and trail foot, averaged over the four trials. For all conditions, swing was defined between toe-off and heel-strike events.
identified from the interaction with the force plate, and divided into three parts by adding the base of support context to the definition. Early swing spanned the time between toe-off and the CoM crossing the posterior limit of the stance foot, mid-swing spanned the time that the CoM was positioned between the posterior and anterior limits of the stance foot, and late swing spanned the time from CoM crossing the toe of the stance foot to heel-strike (Figure 5.2). The heel location was measured at 1.5 cm anterior to the heel marker to account for footwear and marker diameter. The toe of the stance foot defined the physical boundary of stance and was demarked by the 1st metatarsal foot marker.

**Time to Contact Computation**

The CoM time-to-contact of the approach towards the boundaries of swing was computed by dividing the first derivative of CoM position (velocity, Equation 5.1) by the distance between the position of the CoM and the physical boundary ($B_p$) or anticipated boundary ($B_a$).

$$TtC_{CoM}(s) = \frac{B_{p,a} - CoM(m)}{velocity_{CoM}(m/s)} \quad \text{Equation 5.1}$$

Where $B_{p,a}$ can either be the physical ($B_p$) or anticipated ($B_a$) boundary.

The swing foot time-to-contact was computed similarly. The $B_p$ and $B_a$ locations were obtained from kinematic record post-hoc. For unobstructed preferred speed conditions, results were collapsed across one bilateral cycle from each of the four trials. In the obstacle condition, the dependent variables were
computed for the step over the obstacle with the lead foot and trail foot, collapsed across the four trials. The TtC of the CoM toward the Bₐ was reported at the crossing of the CoM beyond the Bₚ (entry into unstable equilibrium; Figure 5.1C).

**Coupling of Time to Contact**

Coupling ratios were computed to describe the relationship between temporal closures and coordination of the CoM and swing foot with the boundaries of swing. The Bₚ approach was described by the coupling ratio Kₚ (Equation 5.2), and the point in swing when the Kₚ becomes less than 1.0 was reported. This point indicates that TtCₗₜₜₜ has dropped below TtCₗₜ in the approach to the Bₚ, signifying that the swing foot will arrive before the CoM enters the unstable equilibrium and the swing foot will lead the body in the approach to the next anticipated boundary.

$$\frac{TtC_{\text{swing foot}}}{TtC_{\text{CoM}}} = K_{p,a}$$ \hspace{1cm} \text{Equation 5.2}

Where Kₚ,a can either be the coupling ratio for the approach to the physical (Kₚ) or anticipated (Kₐ) boundary.

The coupling ratio Kₐ (Equation 5.2) describes the approach to the anticipated boundary and was reported at the point when the CoM enters the unstable equilibrium (crosses Bₚ). The Kₐ ratio at the point of entry into the unstable equilibrium describes the magnitude of differences in contact estimates of the swing foot and the CoM. A high Kₐ ratio indicates the CoM arriving shortly after the swing
foot, whereas a low $K_a$ ratio indicates a longer estimated time between swing foot arrival at the $B_s$ and the eventual CoM arrival. More time (i.e., a low $K_a$ ratio) between swing foot arrival and CoM arrival may allow time for a more reliable placement of the swing foot at the $B_s$ before weight acceptance, thus potentially reducing the chances of a slip.

**Proximal Visual Intersect Computation**

In humans, the lower limit of the field of view arc is anatomical, formed by the orbital ridge of the scull at a declination of approximately $65°$ below the neutral axis of the head (Figure 5.3A) (Trevarthen, 1968). Thus, the head orientation and position constrains the orientation of the field of view. The proximal visual intersect (PVI) describes the intersection between the transverse plane and a vector angled downward $65°$ from the neutral horizontal axis of the head, in the anterior direction along the midline of the body. The PVI was hypothesized to describe the limit in the anterior direction of the field of view of the ground such that no features of the environment more proximal to the body can be visually perceived without additional pitch of the head (see Appendix B-C).

As an individual walks in the anterior direction, the PVI generally moves forward as well (Figure 5.3). The anterior progression of the PVI was hypothesized to describe the anterior sweep of the proximal limit of the field of view across the ground (Figure 5.3B), just as the anterior progression of the CoM describes the forward motion of the body. During walking the CoM cyclically increases and decreases in speed in the anterior direction during a gait cycle, and the PVI behaves
in a similar manner. However, the anterior velocity of the PVI is also affected by the vertical motion of the head, and in particular, the pitch of the head. The head cyclically pitches fore and aft during a gait cycle. The combination of head translation and pitch yields a distinct PVI progression signature such that after toe-off of the swing foot, the anterior velocity of the PVI progression slows significantly (Figure 5.3B). The PVI anterior velocity minima characteristically occur during the early swing portion of the gait cycle, before the entry into the late swing unstable equilibrium.

In tau-gap theory the origin of visual perception of self-motion is closure toward features of the environment. Cues that afford the perception of self-motion may be easiest to extract from the field of view when the visual field is nearly static, such that there is a center of optic expansion or flow near to the area that is most directly being approached. The anterior velocity minima of the PVI during gait may describe the portion of the gait cycle when the most salient perception of self-motion becomes available. For each bilateral swing phase of gait, the TtC to the anticipated step landing area by the swing foot was calculated and reported at the PVI velocity minimum. This temporal gap was then compared to the actual time between the PVI velocity minimum and heel-strike. If the temporal estimate of contact is similar to the actual timing of swing, there may exist a link between the planning of a swing phase and its execution. The location of the PVI at the velocity minima was reported as a distance from the PVI to the $B_a$, and the timing of the PVI as a percentage of the swing cycle.
Statistical Analysis

Differences between steps over the obstacle with the trail and lead foot speed were tested for significance (p<0.05) using a one-way ANOVA. F values with degrees of freedom and 95% confidence intervals for group mean differences were reported.

Results

Preferred speed walking

The preferred walking speed of the young healthy group was 1.33 m/s, with a cycle time of 1.10 s, and a step length of 0.73 m (Table 5.1). The preferred dual support time was 0.20 s and swing time was 0.38 s. The apportioning of the swing phase of gait demonstrated that the CoM was behind the stance foot in early swing for 21.5% of swing, the CoM was above the stance foot in mid-swing for 39.8%, and the CoM was in front of the stance foot in the unstable equilibrium for 38.7% of swing (Table 5.1; Figure 5.2). The CoM crossed the Bp at 61.3% of swing, while the swing foot crossed the Bp at 54.4% of swing (Table 5.1). When the swing foot crosses the Bp before the CoM, the swing foot “has taken the lead” from the CoM in the movement towards the Bp at the point Kg=1, and this occurred at 27.2% of swing (Table 5.1; Figure 5.4). At the entry into the unstable equilibrium by the CoM (passing the Bp), the Kg ratio between swing foot and CoM to the anticipated boundary (Ba) was 0.26 during unobstructed gait at preferred speed (Table 5.1; 5.5).
Figure 5.4. A) Point in swing that the coupling ratio $K_p = 1$; ratio of $TtC_{CoM}$ divided by $TtC_{swing foot}$ to physical boundary ($B_p$), B top) Time to Contact of the swing foot and CoM, and B bottom) coupling ratio $K_p$ between swing foot and CoM on the approach to the $B_p$ during swing from one participant walking at preferred speed. Shown is point in swing when coupling ratio $K_p$ drops below one (dashed line).

Figure 5.5. A) Coupling ratio $K_a$ at CoM entry to the unstable equilibrium; ratio of $TtC_{CoM}$ divided by $TtC_{swing foot}$ to the anticipated boundary ($B_a$), B top) Time to Contact of the CoM and swing foot, and B bottom) coupling ratio $K_a$ between $TtC$ of CoM and swing foot during the approach to $B_a$. 
Table 5.1. Gait results (mean, standard deviation) from preferred unobstructed walking.

<table>
<thead>
<tr>
<th>Preferred Speed</th>
<th>mean</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>walking speed (m/s)</td>
<td>1.33</td>
<td>(.18)</td>
</tr>
<tr>
<td>step length (m)</td>
<td>0.73</td>
<td>(.03)</td>
</tr>
<tr>
<td>cycle time (s)</td>
<td>1.10</td>
<td>(.09)</td>
</tr>
<tr>
<td>stance time (s)</td>
<td>0.72</td>
<td>(.05)</td>
</tr>
<tr>
<td>dual support time (s)</td>
<td>0.20</td>
<td>(.06)</td>
</tr>
<tr>
<td>Swing time (s)</td>
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<td>(.02)</td>
</tr>
<tr>
<td>early swing (% swing)</td>
<td>21.5</td>
<td>(3.6)</td>
</tr>
<tr>
<td>mid swing (% swing)</td>
<td>39.8</td>
<td>(4.6)</td>
</tr>
<tr>
<td>late swing (% swing)</td>
<td>38.7</td>
<td>(3.6)</td>
</tr>
<tr>
<td>CoM crosses B&lt;sub&gt;p&lt;/sub&gt; into UE (% swing)</td>
<td>61.3</td>
<td>(3.6)</td>
</tr>
<tr>
<td>swing foot crosses B&lt;sub&gt;s&lt;/sub&gt; (% swing)</td>
<td>54.4</td>
<td>(4.6)</td>
</tr>
<tr>
<td>K&lt;sub&gt;s&lt;/sub&gt;-1 (% swing)</td>
<td>27.2</td>
<td>(10.7)</td>
</tr>
<tr>
<td>K&lt;sub&gt;s&lt;/sub&gt; ratio to B&lt;sub&gt;s&lt;/sub&gt; (at CoM crossing B&lt;sub&gt;p&lt;/sub&gt;)</td>
<td>0.256</td>
<td>(.05)</td>
</tr>
<tr>
<td>PVI velocity min (% swing)</td>
<td>29.1</td>
<td>(7.4)</td>
</tr>
<tr>
<td>PVI at velocity min to B&lt;sub&gt;s&lt;/sub&gt; (m)</td>
<td>0.07</td>
<td>(.30)</td>
</tr>
</tbody>
</table>

The pause of the field of view described by head movement (PVI velocity minimum) followed a systematic progression in swing (Figure 5.6) and during unobstructed preferred speed walking occurred at 29.1% of swing, and was located 0.07 m closer to the body than the heel-strike location assessed post-hoc (B<sub>s</sub>) (Table 5.1; Figure 5.3, Figure 5.6).

Figure 5.6. Proximal Visual Intersect velocity across a portion of a stride cycle (stance-foot heel-strike to swing-foot heel-strike). Average and standard deviation shown for all preferred pace walking trials by 20 participants (grey shaded region).
Obstacle condition

When stepping over an obstacle with the lead foot, walking speed averaged 1.36 m/s with a step length of 1.60 m and a cycle time of 1.18 s. The lead foot swing time was 0.50 s and dual support time was 0.44 s (Table 5.2). For the lead foot stepping over the obstacle swing was split into early (20.2%), mid (30.6%) and late (49.2%) phases.

When stepping over an obstacle with the trail foot, walking speed averaged 1.32 m/s with a step length of 1.59 m, and a cycle time of 1.22 s. The trail foot swing time was 0.44 s and dual support time was 0.44 s (Table 5.2). For the trail foot stepping over the obstacle swing was split into early (23.3%), mid (34.6%) and late (42.1%) phases.

On the approach to the unstable equilibrium with the lead foot, the CoM moved beyond the Bp at 50.8% of swing, but the swing foot moved beyond the Bp earlier at 40.1% of swing. Thus swing foot “took the lead” from the CoM when Kp=1 in the movement towards the Bp at 15.0% of swing (Table 5.2, Figure 5.7). At the entry into the unstable equilibrium on the step with the lead foot (CoM passing the Bp), the Ks ratio between swing foot and CoM was 0.26 (Table 5.2).

On the approach to the unstable equilibrium with the trail foot, the CoM moved beyond the Bp at 57.9% of swing, but the swing foot moved beyond the Bp after the CoM, at 59.5% of swing. Thus, during the step over the obstacle with the trail foot, Kp never equals 1.0 during the approach to the Bp (Table 5.2; Figure 5.8). At entry into the unstable equilibrium on the step with the trail foot (CoM passing
the $B_p$), the $K_a$ ratio between swing foot and CoM was 0.363, which was higher compared with the step over the obstacle with the lead foot (Table 5.2; Figure 5.8).

Table 5.2. Gait results (mean, standard deviation) from Stepping over an obstacle with the lead and trail foot.

<table>
<thead>
<tr>
<th>Obstacle condition:</th>
<th>Lead Foot</th>
<th>Trail Foot</th>
<th>F(1,19)</th>
<th>p=</th>
</tr>
</thead>
<tbody>
<tr>
<td>walking speed (m/s)</td>
<td>1.36 (.17)</td>
<td>1.32 (.17)</td>
<td>0.63</td>
<td>0.433</td>
</tr>
<tr>
<td>step length (m)</td>
<td>1.60 (.15)</td>
<td>1.59 (.15)</td>
<td>0.01</td>
<td>0.930</td>
</tr>
<tr>
<td>cycle time (s)</td>
<td>1.18 (.09)</td>
<td>1.22 (.10)</td>
<td>1.80</td>
<td>0.188</td>
</tr>
<tr>
<td>stance time (s)</td>
<td>0.68 (.06)</td>
<td>0.78 (.06)</td>
<td>9.36</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>dual support time (s)</td>
<td>0.44 (.06)</td>
<td>0.44 (.06)</td>
<td>0.43</td>
<td>0.234</td>
</tr>
<tr>
<td>Swing time (s)</td>
<td>0.50 (.04)</td>
<td>0.44 (.03)</td>
<td>35.32</td>
<td>&lt;0.001</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th>early swing (% swing)</th>
<th>mid swing (% swing)</th>
<th>late swing (% swing)</th>
<th>CoM crosses B (% swing)</th>
<th>swing foot crosses B (% swing)</th>
<th>$K=1$ (% swing)</th>
<th>$K$ ratio to B (at CoM crossing B)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>20.2 (.39)</td>
<td>30.6 (.43)</td>
<td>49.2 (.45)</td>
<td>50.8 (.45)</td>
<td>40.1 (.43)</td>
<td>15.0 (.97)</td>
<td>0.259 (.069)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>0.363 (.080)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th>PVI velocity min (% swing)</th>
<th>PVI at velocity min to B (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>20.4 (.74)</td>
<td>0.83 (.42)</td>
</tr>
<tr>
<td></td>
<td>20.0 (.45)</td>
<td>0.68 (.50)</td>
</tr>
</tbody>
</table>

Figure 5.7. Time to Contact of the swing foot and CoM (top) and coupling ratio $K_p$ between swing foot and CoM (bottom) on the approach to the Physical boundary (stance foot toe) during swing. From one participant stepping over an obstacle with the lead foot. Note coupling ratio $K_p$ drops below one at 15% of swing.
Figure 5.8. Time to Contact of the swing foot and CoM (top) and coupling ratio $K_p$ between swing foot and CoM (bottom) on the approach to the physical boundary (stance foot toe) during swing. From one participant stepping over an obstacle with the trail foot. Note coupling ratio $K_p$ does not drop below one.

In the visual selection of a step landing area when stepping over an obstacle with the lead foot, when the PVI velocity minima occurred at 20.4% of swing, and were on average 0.83 m closer to the body than the heel-strike location ($B_a$). In the visual selection of a step landing area in the step over an obstacle with the trail foot, the PVI velocity minimum occurred at 20.0% of swing, and was 0.68 m closer to the body than $B_a$ (Table 5.2; Figure 5.9).
Figure 5.9. Proximal Visual Intersect location at anterior velocity minimum relative to the body during a step at preferred speed (left), and over an obstacle with the lead foot (center) and he trail foot (right).

**Discussion**

This study described phases of swing by relating the center of mass position to the posterior and anterior limits of the stance foot. This added boundary context in the definition of swing phases allowed the assessment of when the CoM had passed the anterior limit of the stance stability boundary to enter a controlled anterior fall, referred to as the unstable equilibrium of gait. The movements toward and across the boundaries of swing were first characterized by their spatio-temporal patterns, which revealed that early swing was the shortest portion of swing in all step conditions, and that mid swing was longer than early swing, but shortened in obstacle conditions. The controlled anterior fall of late swing was approximately as long as mid swing in unobstructed gait but late swing was longer in the obstacle condition for both the step with the lead foot and the step with the trail foot. The coupling of time-to-contact estimates to the anterior stance foot limit
(Bp) illustrated how the swing foot overtook the CoM earlier in swing during the step with the lead foot over an obstacle, compared with unobstructed walking. However, during the step over an obstacle with the trail foot, the swing foot did not overtake the CoM on the approach to the Bp, which may heighten the instability of the unstable equilibrium. In the approach to the anticipated boundary, the coupling ratio between the swing foot and the CoM was larger in the trail foot step over the obstacle, indicating a closer arrival of the CoM after the swing foot is placed at heel-strike, which may decrease the available time for proper foot placement. In the obstacle condition, the field of view paused earlier in swing and head orientation was such that more of the environment immediately anterior to the body was inside the field of view limits during this pause, which may enhance exproprioception (Figure 5.9).

In this study, strong emphasis was placed on approach to the unstable equilibrium of gait. When Kp drops below one, the swing foot will be out in front and under the CoM when the body enters the unstable equilibrium, which may aid in the stability of gait. At toe-off, the swing foot has a low anterior velocity, yet by 27% of swing the swing foot has gained enough anterior velocity that it will cover the distance to the physical boundary before the body CoM. The earlier in swing that the CoM overtakes the swing foot on the approach to unstable equilibrium, the more likely the swing foot will be in front of the body when it enters the unstable equilibrium even if a perturbation is encountered. With the swing foot in front of the body, a walker may be afforded more “protection” from exposure to the unstable equilibrium as the swing foot is positioned beneath and in front of the body through
the late swing. It is known that in falls associated with trips, it is an external force that disrupts the progress of the swing foot (Mills & Barrett, 2001). The protection afforded by an early $K_p=1$ may be quantified by the amount of lead-time the swing foot gains relative to the body CoM passing the $B_p$ as keeping the foot in front of the body may allow the swing foot to re-contact the ground in a position that can arrest any unplanned departure of the CoM if necessary.

In the last part of swing, the CoM is “falling” toward the ground until the subsequent heel strike occurs to end swing and recapture the CoM within a new base of support. The magnitude of the coupling ratio ($K_a$) between the swing foot and CoM contact with the anticipated boundary may describe a “safety margin” for the CoM recapture from the unstable equilibrium. A large $K_a$ ratio indicates that the CoM will arrive at the $B_a$ soon after the swing foot does. When the CoM arrives at the $B_a$ soon after the foot is placed on the ground, there may be challenges in maintaining balance during gait, as the acceptance of body mass by the new stance foot may be hastened. A small $K_a$ ratio, such as the 0.26 ratio in this group’s preferred unobstructed walking, gives the walker more time between foot placement and weight acceptance, a situation that potentially would lead to a more stable base-of-support configuration before the CoM arrives, or give more time to evaluate the surface friction conditions under the foot contact via proprioceptive and cutaneous feedback. It is known that falls are often associated with slips, which typically occur soon after heel-strike (Grabiner et al., 2008; Maki & McIlroy, 2006; Mills & Barrett, 2001). Increasing the time between foot placement and weight
acceptance may reduce the chances of a fall resulting from quickly loading the newly arrived stance foot.

In order to better avoid falls from slips or trips, the environment in the transit path is evaluated prior to heel-strike. It is likely that the anticipated boundary has been at least cursorily examined with the visual sense prior to foot contact. This study showed that the vicinity of the B_a was inside the field of view at the pause in anterior progression of the PVI at 29% of swing. The PVI velocity minimum was shown to be a consistent event (Figure 5.6) that occurs on every step before the body crosses into the unstable equilibrium (passing the B_p). The velocity minimum of the PVI gives the walker a potential opportunity to extract salient visual self-motion cues from elements inside the field of view. The anticipated step-landing area plus an additional 7 cm of the ground was inside the field of view at the PVI velocity minima. Thus, the PVI velocity minimum is occurring in a location near the step landing area. It may be an even greater asset in walking over uneven terrain (or over an obstacle) to keep the anticipated boundary visible, where the foot placement must be made more carefully (Marigold & Patla, 2008).

In order to inspect the environment proximal to the body, the head must be oriented in such a way that a clear forward view of the environment is presented (Figure 5.3) since there are anatomical limits to the size of the field of view. To overcome the intrinsic anatomical limits in the field of view, a dynamic strategy of head re-orientation must be used. Additionally, as self-motion cues may be most readily available when features of the environment are quasi-static, the demonstrated dynamics of head motion during gait serve to purposely orient the
head to easily extract information regarding the closure toward features of the environment about to be contacted. While the PVI is in the extreme periphery of the field of view, peripheral vision is of particular importance to picking up optic flow, especially at the intersection of the field of view and the ground, as this interaction between the ground and the feet is a hallmark of overground locomotion. This description of the PVI and of the anterior velocity profile of the PVI are an important first step in better understanding how head kinematics are functionally organized to enhance perception of self-motion during gait. It is the visual perception of closure toward a point in the environment that underpins the tau-gap theory (Lee, 1980), which is hypothesized here to relate to estimating the swing foot’s arrival at the anticipated boundary. When during swing this velocity minimum in the anterior sweep of the visual field is occurring may indicate the need for planning of the current step. When timing of head movements change, it may be an indicator that these motions of the head that remain after the damping effect of the segmental chain are not only a passive result of the decelerations and accelerations of the body, or accelerations that were not “properly” or fully damped. It may be that the kinematics of the body are globally altered such that an earlier fixation of the environment can occur to facilitate step planning as when encountering an obstacle.

In the step over an obstacle with the trail foot, the late unstable equilibrium is shorter than the step over with the lead foot (42% vs. 49%). In the step by the trail foot over the obstacle, the CoM passes the stance foot toe \( B_p \) (57.9% of swing) before the swing foot (59.5% of swing), as revealed by the relative swing timing analysis performed here. It appears that the least stable moments of crossing an
obstacle are in the finishing of the step after it. When the CoM is beyond the physical boundary while the swing foot is still behind it, the CoM is outside the stability boundary and is literally falling forward and down towards the ground with nothing to catch it until the swing foot eventually surpasses the CoM in late swing to make heel contact to recapture the CoM before a fall occurs. With the swing foot already trailing the body into the unstable equilibrium, it is hypothesized that any perturbation to the swing foot or other part of the body at this point will dramatically decrease the chances of successfully completing the step. The larger $K_a$ ratio (0.36) in the step of the trail foot over the obstacle shows that the CoM will arrive at the anticipated boundary with a closer temporal margin to the swing foot compared to both unobstructed preferred walking and the step with the lead foot over the obstacle (0.26), leaving little time to adapt to any unforeseen disturbances.

The actions of swing are necessary for forming walking gait. However, aside from the toe-off and heel-strike events that bracket swing there are few kinetic signatures of the intrinsic dynamics across this phase of gait. As such, researchers have relied on kinematic events to describe the movements of swing. Swing is traditionally divided into 3 parts. These three parts each comprise around 33% of swing and are termed: initial, mid and terminal swing, and are delineated by four events: toe-off, maximum knee angle, lower limb vertical, and heel-strike (Winter, 1983). Here a parsing of swing from toe-off to heel-strike based on when the body CoM moves in the anterior direction from behind to above (22% of swing) and beyond the stance foot (61% of swing) at preferred speed was put forth (Figure 5.2). The advantage of this method of apportioning swing is that it relates to postural
configurations of the whole body relative to the stability boundary as opposed to strictly kinematic references to lower limb angles. The status of the CoM as either behind (early-swing), inside (mid supported swing) or beyond (late unstable equilibrium) the base-of-support allows an ecological reference to body motion that relates to CoM stability. The drawback to this method is that a full body marker set must be used to accurately estimate the CoM location, although the location of the pelvis is generally a good surrogate measure for whole body CoM location during gait (Winter, 1983).

This new parsing of the swing phase of walking gait shows a more boundary relevant strategy for characterizing the status of the body relative to the stance foot. This method should prove to give researchers and practitioners better insights into the adaptations in swing that people may adopt when navigating the environment or when coping with challenges to balance. The employment of time-to-contact in steady state gait and during the steps required to clear an obstacle are novel and functionally extend the basis of the techniques beyond the case where the postural configuration is fixed (Haddad et al., 2006; Hasson et al., 2009; Van Emmerik et al., 2010). By expanding the role of time-to-contact and tau-gap to walking gait, coordination within body segmental movements can be studied in greater detail, which may be useful especially in swing where there are few kinetic markers of events. These methods of analysis applied to walking gait show that distinct signatures of swing foot dynamics are related to whole body movement at specific points in the swing cycle. Furthermore, the findings that the orientation of the lower limit of the visual field is such that in walking gait at preferred speed the subsequent
step-landing area or anticipated boundary is visible until at least 29% of swing, which provides additional information that can improve interpretations of what may be functional head motion.

The swing phase of walking gait occupies a great majority of the gait cycle, and most disturbances to gait occur during or immediately after a swing phase (Mills & Barrett, 2001). Since there are few kinetic measures of swing performance, a detailed inventory of kinematic behaviors was overdue. Estimates of coordination based on the motion of the body and the swing foot showed that allowing the body CoM to enter the unstable equilibrium wholly unsupported under normal walking conditions is avoided in normal walking. However, this status does change when gait is perturbed, as when stepping over an obstacle. It is commonly perceived that the challenge of stepping over an obstacle is the step with the lead foot, yet this step with the lead foot closely resembles an unperturbed step. In contrast, it is the step over the obstacle with the trail foot that appears to be the more challenging, considering that the body enters the unstable equilibrium on this step without the swing foot beneath it. The observance that the PVI velocity minimum occurs earlier in swing and with the head oriented such that more of the proximal environment was inside the field of view illustrated the importance of vision when coping with an obstacle for these healthy young individuals. How functional head motion can facilitate the distillation of visual information was described such that the center of optic flow expansion may be more closely aligned with features of the environment that we intend to contact, at a specific point in the gait cycle as opposed to linear peripheral flow fields that may provide less distinct signatures of self-motion.
Conclusions

These novel methods of swing phase portioning, coordination, and head movement descriptions of field of view dynamics are sensitive to changes in gait task. That swing is portioned differently during various step conditions illustrates how task specific adaptations occur. Departures in behavior from unobstructed gait patterns in this young health adult group may serve as guideposts of chronically disturbed gait. If a person with pathological gait was observed behaving as these young healthy adults it may signify the origin of a gait strategy, such as stepping over unobstructed ground as if it were littered with sequential obstacles, or continually stepping to bring the trail foot over an obstacle behind the CoM. Another example may be that a person systematically leaves the swing foot behind the body on the entry into the unstable equilibrium, which may signify an exposure to unstable states. The patterns of the coupling ratios or the movements of the head as described here may yield insights into the breakdown of stability and mobility in populations where challenges to balance and falls are common. By using these measures to identify movement patterns that may decrease overall gait stability, training protocols can be prescribed to improve gait stability. These gait-training efforts can focus on functional pick up of visual information by reinforcing the importance of observing each step closely, and purposeful execution of the swing phase of gait that keeps the swing foot beneath the body, especially in the unstable equilibrium.
References


CHAPTER VI

STRIDE PARAMETERS AND HEAD MOVEMENT DURING THE SWING PHASE OF WALKING IN MULTIPLE SCLEROSIS

Abstract

Previous studies have shown that the swing phase of walking is shorter in people with multiple sclerosis (MS) when walking at fixed speeds matched with their non-MS counterparts, presumably to lengthen the time that the center of mass (CoM) is positioned over the boundaries of dual support. It is unknown how altered swing may impact functional mobility for those with MS. In a cross-sectional study we assessed changes in swing parameters between a MS group (n=19) and a gender and age matched control group (n=19) during preferred and fixed, group matched, slow (0.6 m/s), medium (1.0 m/s) and fast (1.4 m/s) walking speeds. The MS group had mild-to-moderate impairment with an Expanded Disability Status Scale score (EDSS) median of 3.75 (range 2.5 to 6). Four aspects of swing construction were investigated: (1) novel swing phases: toe-off to CoM over stance foot heel (early swing), CoM over the stance foot (mid-swing), and CoM beyond the stance foot toe to heel-strike (late swing unstable equilibrium); (2) the CoM crossing into the unstable equilibrium from toe-off; (3) CoM coordination with the swing foot during approach into the unstable equilibrium and recapture within a new base of support at heel-strike; and (4) orientation and timing of head motion to address how the field of view is assessed in those with MS.

At all speeds, individuals with MS shortened early swing and lengthened mid-swing compared with controls, while maintaining similar late swing durations.
Longer mid-swing may indicate a preference for keeping the CoM over the base of support in people with MS. Across fixed speed the CoM of the MS group entered the unstable equilibrium earlier compared with controls, which may indicate a hastening of swing. The coordination between the CoM and the swing foot was measured by their coupling of temporal approaches. Coupling results showed that both groups moved toward the termination of swing in a similar manner, which may indicate a consistency in the late swing “controlled anterior fall” of gait. Head motions induced a pause in the anterior field of view progression sooner after toe-off that was positioned closer to the body in the MS group compared to controls, which illustrates increased reliance on visual information in gait. Head motion may indicate that the MS is using vision earlier in swing and keeping more proximal environmental features in view longer compared with controls.

When altering swing, early swing may be the most adaptable portion to shorten, thus allowing mid-swing time to be lengthened and late swing to remain intact. Shortening early swing may facilitate adaptation to the use of vision in walking as well. An earlier pause in the field of view sweep across the ground may be driven by alterations in step execution in MS that allow greater visual exproprioception compared with non-MS counterparts. Early swing compression may improve stability by granting longer times over the stance foot and greater visual control over locomotion, but may detract from stability by hastening the preparations for entry into the unstable equilibrium of gait.
Introduction

Multiple sclerosis (MS) is a disease of the central nervous system where the myelin sheaths of the neurons are damaged by an autoimmune response that results in disrupted signal transmission (Noseworthy et al., 2000). Symptoms of MS are widespread and can affect any or all nervous system functions, as detectable myelin scarring tends to be randomly distributed. It is often a problem with gait or vision that brings an individual to first seek medical help, which ultimately can lead to a diagnosis of MS. MS can disturb balance and coordination as damage to sensory and motor pathways may have lead to diminished perception and control of muscles (Frzovic et al., 2000; Noseworthy et al., 2000; Remelius, et al., 2008; Van Emmerik et al., 2010). Most people with MS remain ambulatory for many years following a MS diagnosis, yet the ability to walk is compromised to some extent as over 50% of people with MS experience falls, which can impact the sense of functional mobility (Finlayson, 2006). Mobility and vision rank highest among factors that influence quality of life in those living of MS (Heesen et al., 2008).

It is known that functional impairments related to MS are associated with changes in the parameters of gait (Benedetti et al., 1999; Gianfrancesco et al., 2011; Kelleher et al., 2010; Martin et al., 2006; Remelius et al., 2011; Sacco et al., 2010; Thompson & Nutt, 2007). When left to walk at their own preferred pace, people with MS maintain a similar swing time while lengthening their dual support time and walking more slowly compared with non-MS counterparts. This apparent sparing of swing time may be related to differences in preferred walking speed, as across matched fixed walking speeds, people with MS shortened swing time while
coincidently lengthening dual support time compared with their non-MS counterparts (Remelius et al., 2011). Lengthening dual support time in walking is commonly ascribed to be part of a “nonspecific protective gait strategy” (Benedetti et al., 1999) intended to improve balance control by maintaining a wider and longer base of support for a longer portion of the gait cycle (Gianfrancesco et al., 2011; Kelleher et al., 2010; Martin et al., 2006; Remelius et al., 2011; Sacco et al., 2010; Thompson & Nutt, 2007). However a strategy featuring longer dual support times may have limitations, when it induces a commensurate shortening of swing time if this leads to hastened body movements while the swing foot is in the air (Remelius et al., 2011). Swing altered in such a manner may exacerbate vulnerabilities to gait disturbances, resulting in trips and slips that could lead to an increased risk of falling (Matsuda et al., 2011; Mills & Barrett, 2001) yet the impact of shortened swing time on gait stability and its relationship to the high prevalence of falls in those with MS (Finlayson et al., 2006) remains unclear.

To uncover embedded dynamics in swing and to discern what in swing may be altered in those with MS, a novel partitioning of swing was developed. Swing is defined between toe-off and heel-strike, but here the position of the center of mass (CoM) and the context of the stability boundary, or limits of the foot contact with the ground, were added and used to split swing into three phases. Two events served to partition swing, the point in swing when the CoM moves above the posterior limit of the stance foot (heel), and beyond the anterior limit of the stance foot (toe) in the sagittal plane. Thus, the phases of swing were defined as (Figure 6.1): early (CoM behind the stance foot), mid (CoM between heel and toe of the stance foot) and late.
swing (CoM in front of the stance foot). Identifying which portions of swing are altered may be important for understanding the propensity of accidental falls in those with MS, considering the swing phase of gait itself is often referred to as a “controlled fall” (Winter, 1983) (late swing; Figure 6.1A&B). The controlled anterior fall of gait is defined as the portion of swing where the CoM is in an unstable equilibrium where the CoM has moved beyond the anterior stance foot stability boundary. The controlled fall of gait and the unstable equilibrium definitions coincide with the description of late swing put forth here.

![Figure 6.1. Three swing phases between toe-off and heel-strike from: A) MS, and B) control participants walking at preferred speed. Early swing; mid swing (CoM over stance foot); and late swing.](image)

In addition to understanding how much time is spent inside or outside the stance foot stability boundary, describing how boundaries are approached may also reveal features of stability. A means to characterize boundary approach as
developed from the study of posture is the temporal estimate of time to contact (TtC). TtC is a continuous estimation of the instantaneous time it may take to make contact with a perceived boundary, given distance and velocity of the body. The original TtC concept emerged from vision research that sought to characterize the visual perception of a distance closure (gap closure) as a directly perceivable temporal variable called \( \tau \) (Lee & Lishman, 1977). In the postural control domain, the TtC concept was later adopted (Riccio, 1993) to explain proprioceptive-based temporal estimates of the approach to postural stability boundaries (perimeter of the feet). In human posture research, TtC describes the approach of the body’s center of pressure (CoP) or center of mass (CoM) towards and away from the physical base of support (Slobounov et al., 1998). TtC was developed as a method to understand postural activity when both feet remain in place and has evolved from descriptions of quiet standing (Haddad et al., 2006) to behavior during more challenged postures (Forth et al., 2006; Van Emmerik et al., 2010), the postural phase of gait initiation (Remelius et al., 2008) and the CoM activity preceding a step due to a postural perturbation (Hasson et al., 2008; Hasson et al., 2009). Nonetheless, TtC holds potential for use in tasks that are more dynamic where the stability boundary changes as in locomotion.

Estimates of when the body as a whole will contact a boundary may be described with TtC, but as the body is composed of many segments that are coincidently moving, a description of how the motions of these segments are coupled is warranted if inter-segmental coordination patterns are to be properly understood. This coordination can be described by the coupling ratio of two motion-
gaps (Lee et al., 2001). In locomotion, a coupling ratio can reveal how the motion-gaps of the swing foot and the CoM close in towards a single boundary, such as the anterior limit of the stance foot stability boundary (toe). Coupling of τ-gaps has been used to describe the first step of walking, where the support of the body has been simplified as being located at the center of pressure. Austad & van der Meer (2007) computed τ-gap to a new physical boundary in the first step of walking, and showed that closure rates change during early development. Austad & van der Meer (2007) used tau-gap coupling in the initiation of gait and the approach to a physical boundary, but how gaps are coupled during steady state gait has not been investigated.

To form the patterns of steady state gait, a cyclical transition into the unstable equilibrium of swing is necessary, and coordination of movement into this unstable state is important. However, the recapture of the body from this unstable state at heel-strike is also critical to gait stability. Where and when to place the foot and terminate swing is guided in part by visual information in sighted individuals. Over multi-surfaced terrain, the next step landing area has been visually fixated upon at least one step in advance (Marigold & Patla, 2007) and vision can continue to guide foot motion during the current step (Reynolds & Day, 2005). Swing foot motion toward this next anticipated boundary (heel-strike location) may be organized by information gained from a visual τ-gap perceived while this area is still inside the field of view. Perception of self-motion extracted from visual information is a form of proprioception also referred to as ‘visual exproprioception’ (Lee & Kalmus, 1980), where information of the environment and of the body movement
within and relative to the environment is perceptible (Patla, 1991; Marigold, 2008). Visual exproprioception relies heavily on optic flow, which is the pattern of apparent motion of features in the environment caused by motion of the observer. Optic flow for sighted individuals is shaped by overall body movement, and compensatory head movements serve to functionally stabilize the field of view and allow for perception of stationary features on the ground (Berg & Mark, 2005; Bradshaw & Sparrow, 2001; Buckley, MacLellan, Tucker, Scally, & Bennett, 2008; Gibson, 1958; Marigold & Patla, 2008; Owen & Lee, 2004; Patla, Prentice, Rietdyk, Allard, & Martin, 1999; Reynolds & Day, 2005; Warren, Young, & Lee, 1986).

By following the motion of the head across the walking gait cycle, it may be possible to discern for how long the anticipated boundary of swing is perceptible and how far in front of the body the field of view is positioned during swing. Here the intersection between the inferior limit of the visual field and the ground in front of the body is termed the proximal visual intersect (PVI; Figure 6.2) and portions of the environment more distal to the body from this PVI are in the field of view. However, the inferior limit of the field of view is not necessarily the center of optic expansion, or the center of the visual flow field. The value in describing the PVI is in the ability to describe when during swing the field of view is making its slowest anterior progression and what portions of the environment are inside the field of view. Furthermore, in this quasi-static part of the anterior sweep of the field of view, perception of salient visual cues of self-motion may be most easily extracted and used to derive a per step assessment of \( \tau \)-gap.
The purpose of this study was to compare characteristics of swing construction in people with MS to their non-MS counterparts. These comparisons were made while walking at preferred speed and across a range of fixed speeds, as group comparisons can be confounded when speeds are not equal. Four aspects of the swing phase of walking were investigated: (1) Swing construction was compared as the duration of each of three swing phases by partitioning swing (toe-off to heel-strike) into early mid and late phases based on the transit of the center of mass (CoM) past the posterior and anterior stance foot boundaries. (2) The point at which the body enters the unstable equilibrium of swing and begins its “controlled anterior fall” occurs at the conclusion of the mid-swing defined here. Entry into the unstable equilibrium is considered here as an important marker of swing construction and the timing of each group’s entry was compared. It was
hypothesized that the early and mid phases of swing were primarily altered to account for the shortening of swing and that late swing would not be changed by the MS participants. (3) To understand the coordination in swing during approach to the entry into the unstable equilibrium and the recapture from this state at heel-strike, coupling ratios between the temporal approaches to the physical (stance foot toe) and anticipated (heel-strike location) boundaries between the swing foot and the body by each group were compared. (4) This study then compared how and when in swing each group constructed head movements to orient the field of view relative to the body. Here, it was hypothesized that those in the MS group rely on vision in swing to a larger degree compared with their non-MS counterparts by adaptations to head movement in order to bring the PVI closer to the body.

**Methods**

**Participant Characteristics**

Nineteen volunteers with mild-to-moderate impairment due to MS (16 females) participated in this study (Table 6.1). The MS group was asked to evaluate their disability using the self-administered Expanded Disability Status Scale (EDSS (Bowen, Gibbons, Gianas, & Kraft, 2001)(median=3.75; range 2.5 to 6). None of the participants with MS reported exacerbations of symptoms in the previous six months, visual acuity less than 20/200, oculomotor disorders, or were non-ambulatory. The distribution of the MS subtypes was: primary progressive [2], relapsing remitting [15], and secondary-progressive [2]. Participants walked unassisted during data collection, although 3 participants with MS reported
occasionally using a cane for walking during regular daily activities. The time elapsed since diagnosis ranged from 2 to 26 years (mean=10.3; SD=7.8), and participants were taking the following medications: immunomodulators [15 participants], anti-depressants [8], anti-spasticity [6], anti-fatigue [5], anti-convulsive/sedative [2], and anti-vertigo [1].

Control participants (n=19) were age- and gender-matched to those in the MS group (Table 6.1). All participants were otherwise healthy (no orthopedic or other neurological problems besides MS) and sedentary to recreationally active (≤30 min structured exercise a day, three days per week). All participants gave informed consent, as approved by the Institutional Review Board from the University of Massachusetts Amherst, before testing.

### Table 6.1. Demographic, sensory and functional characteristics for MS and control groups. Reported: Mean ± Standard Deviation, F value (degrees of freedom), p value, and 95% Confidence Interval (CI).

<table>
<thead>
<tr>
<th>Participant Characteristics and Functional Assessment</th>
<th>MS</th>
<th>Controls</th>
<th>F(1,36)</th>
<th>p</th>
<th>95% CI</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>51.3 (10.5)</td>
<td>51.8 (11.5)</td>
<td>0.01</td>
<td>0.930</td>
<td>-7.6 6.9</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>167.6 (6.7)</td>
<td>164.8 (6.7)</td>
<td>1.70</td>
<td>0.201</td>
<td>-7.5 1.6</td>
</tr>
<tr>
<td>Body mass (kg)</td>
<td>75.65 (16.6)</td>
<td>67.39 (10.0)</td>
<td>4.96</td>
<td>0.032</td>
<td>-41.13 1.92</td>
</tr>
<tr>
<td>Vibration threshold (volts)</td>
<td>17.7 (13.1)</td>
<td>8.9 (6.4)</td>
<td>6.79</td>
<td>0.013</td>
<td>-15.4 -1.9</td>
</tr>
<tr>
<td>Pressure threshold (gr)</td>
<td>4.64 (10.23)</td>
<td>0.65 (.70)</td>
<td>2.88</td>
<td>0.098</td>
<td>-8.76 0.78</td>
</tr>
<tr>
<td>Fatigue score (VAFS cm)</td>
<td>3.88 (2.39)</td>
<td>2.33 (1.88)</td>
<td>6.71</td>
<td>0.014</td>
<td>-3.18 -0.39</td>
</tr>
<tr>
<td>Fatigue Severity Scale</td>
<td>5.6 (1.2)</td>
<td>2.7 (1.1)</td>
<td>3.2</td>
<td>&lt;0.001</td>
<td>1.9 3.9</td>
</tr>
<tr>
<td>Foot taps (in 10 s)</td>
<td>34.3 (9.1)</td>
<td>48.4 (9.9)</td>
<td>17.70</td>
<td>&lt;0.001</td>
<td>6.6 18.8</td>
</tr>
</tbody>
</table>

Walking Gait Protocol

Participants were tested on two separate visits. In the first visit, all participants underwent a functional assessment, and the MS group completed the
Self-Administered EDSS (Bowen et al., 2001). Peripheral sensitivity to vibration was assessed with a Biothesiometer (Bio-Medical Instrument Co, Ohio) as the voltage setting at which vibration is first perceived; and to light touch pressure with a single filament test protocol (North Coast Medical Inc, CA) as force in grams of the smallest filament perceived. Vibration and touch pressure threshold values were reported as an average from bilateral tests on the ball, arch and heel of each foot (Table 6.1). Motor drive was measured as the average number (across two measurements) of foot-taps each participant was able to perform in 10 s with each foot (Kent-Braun, Walker, Weiner, & Miller, 1998) (Table 6.1). Acute symptomatic fatigue was assessed with a visual analog fatigue scale (VAFS) at the start of the first visit, by marking the current degree of fatigue on a 10-cm scale (Table 6.1). The Fatigue Severity scale (FSS) measured chronic fatigue (Krupp et al., 1989).

During the second visit, kinematic and pressure mat recordings were collected for 4 preferred walking speed trials. These were followed in random order by trials at fixed-speeds: slow (0.6 m/s), medium (1.0 m/s), and fast (1.4 m/s). Participants were provided verbal feedback about their speed relative to the target speed. Individuals were requested to walk at each speed until 4 recordings were obtained at each speed that fell within ±5% of the target speeds. Participants wore lab-supplied running shoes.

**Experimental Setup for Gait Collections**

Qualisys track manager software was used to synchronize eight Oqus cameras (Qualisys, Sweden) sampled at 240 Hz with a 0.5 x 2.0 m pressure mat (RSscan, Belgium) sampled at 60 Hz. The pressure mat was located in the center of a
0.7 x 14 m wooden walkway and served to identify bilateral heel-strike and toe-off events. A 6 m optical trap provided average gait speed estimates on each gait trial.

Segment kinematics were tracked with fifty 10 mm retro-reflective markers. A pointer rod was used to create virtual landmarks for reference to joint centers. A 13-segment 3D biomechanical model, based on established anthropometric data (Clauser, McConville, & Young, 1969), was constructed from the kinematic data in order to estimate the position of the CoM, the orientation of the head, and the proximal and distal end of each foot during gait collections. The head segment of the model was defined by pointer events at the porions. The midpoint between the bilateral porions located the CoM of the head, and the head was oriented towards bilateral pointer events at the inferior orbital ridges to align the segment towards the neutral plane of the head, and tracked with a rigid cluster of markers worn on the crown. The torso, pelvis, and upper/lower arm and leg segments were built from anatomical landmark events and tracked with marker clusters. The feet were built from landmarks at the tibial medial malleolus and fibular lateral malleolus, and 1st and 5th metatarsal markers, and tracked with the metatarsal markers and a heel marker cluster. The proximal and distal ends of the foot segments during stance were used to describe the base of support (base of support) in the unipedal posture of walking gait. The proximal foot (posterior base of support limit) was demarked by a virtual marker constructed 15 mm anterior to the heel marker in stance (accounting for marker and shoe offset from the foot). The distal foot (anterior base of support limit), also referred to here as the physical boundary of stance (Bp), was constructed at the bisection of the 1st and 5th metatarsal markers.
Data Analysis

Swing Timing

Visual 3D (C-Motion, MD) was used to build the model, filter marker data (8 Hz low-pass bidirectional zero lag) and post-process CoM position, bilateral foot position, head orientation, gait speed, and swing time. The proximal (posterior) and distal (anterior) foot locations in stance were used to describe the sagittal plane boundaries of the stance foot. These stance foot definitions allowed swing to be parsed between toe-off and heel-strike into three phases: (1) early swing: from toe-off to CoM passing the posterior heel of the stance foot; (2) mid-swing: where the CoM is positioned between posterior and anterior limits of the stance foot; and (3) late-swing: from when the CoM passes beyond the stance foot to heel-strike (Figure 6.1). The event dividing mid and late swing also demarks the passing of the physical boundary of the stance foot ($B_p$) by the CoM and the entry into the unstable equilibrium, which ends at heel-strike. The unstable equilibrium is the latter portion of swing where the CoM is no longer above the base of support in the anterior-posterior direction. Swing timing in each of these phases (early; mid; late) was reported in absolute times (i.e., from toe-off).

Coupling of Time to Contact

In order to compute the coupling ratio of temporal approaches, time-to-contact (TtC) with the boundaries of swing (Figure 6.3B-6.4B top panels) was computed by dividing the distance between the position of the CoM and the physical boundary ($B_p$) or anticipated boundary ($B_a$) by the CoM velocity (Equation 6.1). The
swing foot time-to-contact was computed similarly. The \( B_p \) and \( B_a \) locations were obtained from kinematic record post-hoc.

\[
TiC_{\text{CoM}}(s) = \frac{B_{p,a} - CoM(m)}{\text{velocity}_{\text{CoM}}(m/s)} \quad \text{Equation 6.1}
\]

Where \( B_{p,a} \) can either be the physical \((B_p)\) or anticipated \((B_a)\) boundary.

Coupling ratios were computed to describe the coordination of the temporal closures \((TtC_{\text{CoM}} \text{ and } TtC_{\text{swing foot}})\) toward two boundaries of swing \((\text{Figure 6.3B-6.4B bottom panels})\). For describing coordination of the move into the unstable equilibrium \((B_p \text{ approach})\), the point in swing when the coupling ratio \(K_p\) (Equation 2, \text{Figure 6.3}) becomes less than 1.0 was reported.

\[
\frac{TtC_{\text{swing foot}}}{TtC_{\text{CoM}}} = K_{p,a} \quad \text{Equation 6.2}
\]

Where \(K_{p,a}\) can either be coupling ratio for the approach to the physical \((B_p)\) or anticipated \((B_a)\) boundary.

The point in swing when \(K_p\) drops below 1.0 indicates that the temporal estimate of contact with the \(B_p\) of the swing foot has dropped below that of the CoM in the approach to the \(B_p\). Once \(TtC_{\text{swing foot}}\) is shorter than \(TtC_{\text{CoM}}\), the swing foot will theoretically arrive at the boundary before the CoM. For describing coordination of the move toward recapture of the CoM from the unstable equilibrium \((B_a \text{ at heel-strike})\), the coupling ratio \(K_a\) (Equation 6.2) was reported at the CoM crossing of the \(B_p\) \((\text{Figure 6.4})\). The \(K_a\) ratio at the point of entry into the unstable equilibrium by
the CoM describes the magnitude of differences in contact estimates of the swing foot and the CoM. A high $K_a$ ratio indicates the CoM arriving shortly after the swing foot, whereas a low $K_a$ ratio indicates a longer estimated time between swing foot arrival at the $B_a$ and the eventual CoM arrival. More time between swing foot arrival and CoM arrival may allow time for a more reliable placement of the swing foot at the $B_a$ before weight acceptance, thus potentially reducing the chances of a slip.

Figure 6.3. A) Point in swing when the coupling ratio $K_p = 1$; $K_p$ is the ratio of $TtC_{CoM}$ divided by $TtC_{swing foot}$ to the $B_p$ (physical boundary), B) Time to Contact of the swing foot and CoM (top) and coupling ratio $K_p$ (bottom) between swing foot and CoM on the approach to the $B_p$ during swing from one participant walking at preferred speed ($K_p = 1$ vertical dashed line).
Figure 6.4. A) Coupling ratio $K_a$ at CoM entry to the unstable equilibrium; ratio of $TtC_{CoM}$ divided by $TtC_{swing foot}$ to the $B_a$ (anticipated boundary); B) Time to Contact of the CoM and swing foot (top), and coupling ratio $K_a$ (bottom) between $TtC$ of CoM and swing foot during the approach to $B_a$.

**Proximal Visual Intersect Computation**

In humans, the lower limit of the field of view arc is anatomical, formed by the orbital ridge of the scull at a declination of approximately 65° below the neutral axis of the head (Figure 6.2) (Trevarthen, 1968). Thus, the head orientation constrains the limits of the field of view. The proximal visual intersect (PVI) describes the intersection between the transverse plane and a vector angled downward 65° from the neutral horizontal axis of the head in the anterior direction along the midline of the body. The PVI was hypothesized to describe the limit in the anterior direction of the field of view of the ground such that no features of the environment more proximal to the body can be visually perceived without additional pitch of the head (see Appendix B & C).
As an individual walks in the anterior direction, the PVI generally moves forward as well (Figure 6.2B). The anterior progression of the PVI was hypothesized to describe the anterior sweep of the proximal limit of the field of view across the ground, just as the anterior progression of the CoM describes the forward motion of the body. While walking forward, the CoM cyclically increases and decreases in speed during a gait cycle, and the PVI behaves in a similar manner. However, the anterior velocity of the PVI is also affected by the vertical motion of the head, and in particular, the pitch of the head (see Appendix D). The combination of anterior and vertical head translation and pitch yields a distinct PVI progression signature such that after toe-off of the swing foot, the anterior velocity of the PVI progression slows (Figure 6.2B). The PVI anterior velocity minimum occurs during the early swing portion of the gait cycle, before entry into the unstable equilibrium at the start of late swing.

In tau-gap theory the visual perception of self-motion is derived from sensing closure toward features of the environment. Cues that afford the perception of self-motion may be easiest to extract from the field of view when the visual field is quasi-static, such that optic expansion or flow is centered on the area that is most directly being approached. The anterior velocity minimum of the PVI during gait may describe the portion of the gait cycle when the most salient perception of self-motion becomes available. The point in swing where this anterior sweep of the proximal field of view limit is slowest was reported in absolute time from toe-off. Furthermore, the location of the PVI at the anterior velocity minimum relative to the
CoM position (positive number indicates the distance PVI lies ahead of the CoM) was reported.

**Statistical analysis**

Differences between groups for preferred speed parameters were tested for significance (p<0.05) using a one-way ANOVA, and for the fixed-speed condition parameters with a 2-factor (group, speed) mixed model ANOVA. A one-way ANOVA was used to test for significant differences within each of the fixed-speed conditions post-hoc. F values with degrees of freedom and 95% confidence intervals for group mean differences were reported.

**Results**

**Functional Assessment**

There were no group differences in age, height, or filament pressure threshold (Table 6.1). Compared with controls, the MS group was heavier, had diminished motor drive (slower foot tap speed), and higher vibration threshold and higher levels of acute and chronic fatigue (VAFS, FSS; Table 6.1) (Remelius et al., 2011).

**Swing Phase Timing**

As reported earlier (Remelius et al., 2011), when walking at preferred speed, total swing time was not different between groups. However, direct comparisons available from fixed walking speeds showed that swing time was shorter in the MS
group compared with controls when both groups walked with similar gait speeds (Table 6.3; Figure 6.5).

At every walking speed, the MS group shortened early swing and lengthened mid-swing in comparison with controls (Tables 6.2-6.3; Figure 6.1; Figure 6.5). In contrast, at every walking speed, late swing was the same between groups. Additionally, late swing was the same at every speed within each group (Tables 6.2-6.3; Figure 6.5).

![Figure 6.5. Timing of the three swing phases (ms) and the entry to the unstable equilibrium between toe-off and heel-strike from MS and non-MS participants walking at preferred and fixed speeds: early; mid (CoM over stance foot); and late.]

**Entry to the unstable equilibrium**

When walking at preferred speed, there were no group differences in when the CoM passed the anterior physical boundary of the stance foot (sum of early and mid swing) and entered the unstable equilibrium (CoM crossed Bp; Table 6.2). Yet,
when walking speeds were fixed, the MS group entered the unstable equilibrium by crossing the $B_p$ with the CoM sooner after toe-off compared with controls (Table 6.3).

**Table 6.2.** Swing timing parameters of preferred walking speed condition, for MS and control group. Reported: Mean ± Standard Deviation, F value (degrees of freedom), p value, and 95% Confidence Interval (CI).

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Preferred Speed</th>
<th>Controls</th>
<th>F(1,36)</th>
<th>p</th>
<th>95% CI</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gait speed (m/s)</td>
<td>1.26 (.23)</td>
<td>1.39 (.21)</td>
<td>3.54</td>
<td>0.061</td>
<td>-0.01 0.28</td>
</tr>
<tr>
<td>Swing time (ms)</td>
<td>354 (29)</td>
<td>361 (27)</td>
<td>1</td>
<td>0.480</td>
<td>-25 12</td>
</tr>
<tr>
<td>early swing (ms)</td>
<td>54 (23)</td>
<td>85 (14)</td>
<td>28</td>
<td>&lt;0.001</td>
<td>-44 -19</td>
</tr>
<tr>
<td>mid-swing (ms)</td>
<td>150 (30)</td>
<td>130 (21)</td>
<td>6</td>
<td>0.018</td>
<td>3 37</td>
</tr>
<tr>
<td>CoM into unstable equilibrium (ms)</td>
<td>204 (23)</td>
<td>215 (27)</td>
<td>2</td>
<td>0.158</td>
<td>-28 5</td>
</tr>
<tr>
<td>late swing (ms)</td>
<td>151 (18)</td>
<td>145 (17)</td>
<td>1</td>
<td>0.359</td>
<td>-6 17</td>
</tr>
<tr>
<td>toe-off to $K_s=1$ (ms)</td>
<td>41 (22)</td>
<td>47 (19)</td>
<td>1</td>
<td>0.438</td>
<td>-19 8</td>
</tr>
<tr>
<td>$K_s$ ratio @ CoM crossing $B_p$</td>
<td>0.21 (0.03)</td>
<td>0.22 (0.04)</td>
<td>0.69</td>
<td>0.412</td>
<td>-0.03 0.01</td>
</tr>
<tr>
<td>toe-off to PVI velocity min (ms)</td>
<td>62 (36)</td>
<td>85 (42)</td>
<td>3</td>
<td>0.077</td>
<td>-49 3</td>
</tr>
<tr>
<td>CoM to PVI distance @ PVI vel min (mm)</td>
<td>511 (115)</td>
<td>692 (187)</td>
<td>11</td>
<td>0.002</td>
<td>-292 -71</td>
</tr>
</tbody>
</table>
Table 6.3. Swing timing parameters of slow, medium and fast walking speed conditions for MS and control groups. Reported: Mean ± Standard Deviation, F value (degrees of freedom), p value, and 95% Confidence Interval (CI).

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Slow Speed (0.6 m/s)</th>
<th>Medium Speed (1.0 m/s)</th>
<th>Fast Speed (1.4 m/s)</th>
<th>group: across fixed speeds</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>MS</td>
<td>Controls</td>
<td>MS</td>
<td>Controls</td>
</tr>
<tr>
<td>Gait speed (m/s)</td>
<td>0.61 (.01)</td>
<td>0.61 (.01)</td>
<td>1.00 (.01)</td>
<td>1.00 (.01)</td>
</tr>
<tr>
<td>Swing time (ms)</td>
<td>460 (52)</td>
<td>493 (54)</td>
<td>385 (38)</td>
<td>401 (28)</td>
</tr>
<tr>
<td>early swing (ms)</td>
<td>11 (32)</td>
<td>48 (26)</td>
<td>48 (21)</td>
<td>81 (20)</td>
</tr>
<tr>
<td>mid-swing (ms)</td>
<td>302 (21)</td>
<td>298 (19)</td>
<td>182 (15)</td>
<td>172 (18)</td>
</tr>
<tr>
<td>CoM into unstable equilibrium (ms)</td>
<td>313 (37)</td>
<td>346 (33)</td>
<td>230 (27)</td>
<td>253 (22)</td>
</tr>
<tr>
<td>late swing (ms)</td>
<td>147 (32)</td>
<td>147 (40)</td>
<td>155 (25)</td>
<td>148 (25)</td>
</tr>
<tr>
<td>toe-off to Kp=1 (ms)</td>
<td>25 (21)</td>
<td>32 (17)</td>
<td>37 (20)</td>
<td>47 (17)</td>
</tr>
<tr>
<td>Ka ratio @ CoM crossing Bp</td>
<td>0.12 (0.02)</td>
<td>0.12 (0.01)</td>
<td>0.18 (0.02)</td>
<td>0.18 (0.02)</td>
</tr>
<tr>
<td>toe-off to PVI velocity min (ms)</td>
<td>89 (26)</td>
<td>111 (39)</td>
<td>73 (31)</td>
<td>96 (28)</td>
</tr>
<tr>
<td>CoM to PVI distance @ PVI vel min (mm)</td>
<td>435 (210)</td>
<td>706 (146)</td>
<td>536 (84)</td>
<td>728 (194)</td>
</tr>
</tbody>
</table>
Coordination measures of swing: coupling of swing foot to CoM

While walking at preferred speed, there was no difference between groups in when the coupling ratio to the physical boundary (K_p between TtC_{swing foot} and TtC_{CoM}) dropped below one (toe-off to K_p=1; Table 6.2). Across fixed speeds walking speeds, the K_p ratio dropped below one sooner after toe-off in the MS group compared to controls (toe-off to K_p=1; Table 6.3).

At every walking speed, in the approach to the anticipated boundary (B_a) there were no group differences in the K_a coupling ratio between TtC_{swing foot} and TtC_{CoM} (K_a ratio @ at CoM crossing B_p; Tables 6.2-6.3).

Visual cues of swing

When walking at preferred speed, the velocity minimum in the proximal visual intersect (PVI) trended towards occurring earlier in swing in the MS group compared with controls (toe-off to PVI velocity min; Table 6.2). When walking at fixed speeds, the PVI velocity minimum occurred earlier in swing in the MS group compared with controls (toe-off to PVI velocity min; Table 6.3).

The PVI describes what portions of the environment are inside the field of view and the location of the PVI at the pause in the anterior progression (PVI velocity minimum) was closer to the body CoM in the MS group compared with controls across all speed conditions (CoM to PVI distance @ PVI velocity min; Tables 6.2-6.3; Figure 6.6).
Figure 6.6. Distance from CoM to Proximal Visual Intersect (PVI) of the field of view with the transverse plane at PVI velocity minimum while walking at medium speed: MS (A), and Control (B).

**Discussion**

This study sought to compare the construction of swing in a volunteer group with mild-to-moderate impairment due to MS and their non-MS counterparts. By parsing swing into phases that related the transit of the CoM over the posterior and anterior boundaries of the stance foot, the context of body movement in relation to the stability boundaries was added to create a novel definition of swing. As hypothesized, it was observed that the MS group spent less time with the CoM behind the stance heel (early swing) and spent more time over the stance foot (mid swing) compared with their non-MS counterparts, while keeping the controlled fall
or unstable equilibrium (late swing) unaltered in all walking speed conditions. The lengthening of mid swing may be further evidence of a strategy to keep the CoM above the base of support. While not the case in preferred speed walking, across fixed speed conditions the MS group shortened the temporal approach to the unstable equilibrium, illustrating that it is the shortening of early swing, despite a lengthening of mid swing, that is the source of shorter swing times across fixed speeds in those with MS compared with controls. On the approach to the unstable equilibrium (CoM crossing Bp), the coordination between the TtCswing foot and TtCCoM expressed by the coupling ratio Kp coincided with the timing of the entry into the unstable equilibrium: Kp=1 occurred at a similar time after toe-off for both groups at preferred speed, and earlier after toe-off in the MS group across fixed speeds compared with controls. The shortening of the time to the unstable equilibrium entry by the MS group may be driving the earlier overtaking of the CoM by the swing foot in preparation of the controlled fall in those with MS compared to controls. In movements toward terminating the unstable equilibrium and acquiring a new stance, both groups coupled TtCCoM with TtCswing foot in a similar fashion across all walking speeds, as Ks ratios were not different between groups. As Ks ratios and the timing of late swing were similar between groups, those with MS may orchestrate late swing in a similar manner compared with their non-MS counterparts. Finally, the timing of the anterior velocity minimum of the proximal visual intersect (PVI) occurred earlier after toe-off in those with MS compared with their non-MS counterparts. Additionally, the orientation of the field of view during swing as prescribed by head motion and depicted by the PVI paused at a location closer to the
body in those with MS compared with their non-MS counterparts. These measures of visual information availability show that the MS group is using vision earlier in swing and keeping more proximal environmental features in view longer compared with controls.

As reported earlier, in the preferred walking condition, gait speed tended to be slower, yet swing time remained similar between groups, while across fixed walking speeds gait speeds were similar between groups, yet the MS group walked with shorter swing times compared with controls (Remelius et al., 2011). An initial conclusion may be that preferred gait speed is chosen by those with MS in part to orchestrate a consistent swing phase of gait considering the similarity of swing time compared with controls. However, this report showed that even when swing time was the same between groups, as in the preferred speed conditions, individuals with MS systematically shortened early swing and lengthened mid swing compared with controls. This lengthening of mid swing may be an additional aspect of the protective gait strategy that favors increased time over the stability boundaries. In this way people with MS may be keeping the body above the base of support for as long as possible even during unipedal support, an argument similarly ascribed to the lengthening of dual support times (Benedetti et al., 1999). Perhaps this gait strategy may be better described as an effort to keep above the base of support in both dual and unipedal supported stance configurations. While spending more time over the A/P base of support may enhance overall stability while walking in the plane of progression, it is important to note that during normal gait the CoM typically does not transit over the stance foot in the medio-lateral direction but
oscillates between the medial boundaries of the feet (Winter, 1983). However, the pattern of positioning of the CoM over the anterior posterior stance foot boundaries by the MS group may allow for a managed entry into the unstable equilibrium, despite the CoM being medial to the medial boundary of the stance foot. Upon entry into the unstable equilibrium, late swing unfolded in a similar manner between groups and across all walking speeds. Similar late swing timing between groups may indicate that the controlled fall into the unstable equilibrium of swing is a momentum driven action in contrast to a motion driven principally by muscular force. Swing is not a passive movement by the swing leg (Whittlesey, Van Emmerik, & Hamill, 2000), yet the controlled fall through the unstable equilibrium by the CoM may be a momentum driven phenomenon as a whole. If late swing is relatively constant, it may be the differences in the events leading up to the entry into the unstable equilibrium of late swing that are responsible for the propensity for those with MS to fall.

At preferred speed the total time from toe-off to the physical boundary crossing (CoM entering the unstable equilibrium) was similar between groups despite the alterations to both early swing and mid swing by those with MS compared with controls. In contrast, across fixed speeds the alterations to early swing and mid swing by those with MS did provoke an earlier entry into the unstable equilibrium compared with controls. When walking at speeds other than preferred, those with MS appear to hasten the entry into the unstable equilibrium. Having less time to orchestrate the body movements into the unstable equilibrium may lead to less reliable foot placement at the end of the controlled fall, as the
temporal window to plan and organize the entry and eventual recapture of the CoM from this unstable state is reduced for those with MS compared to controls. A trip that occurs in swing (i.e., the swing foot is perturbed) will be afforded less time for recovery, as the body has begun the transit into the unstable equilibrium sooner after toe-off in those with MS compared with controls.

While the timing of the CoM into the unstable equilibrium provides pertinent information regarding the support of the body in swing, it is the movement of the swing foot in relation to the body that ensures the success of swing and perhaps especially of the controlled anterior fall of gait. For a more comprehensive description of swing, the coupling of the CoM displacement to the motions of the swing foot towards the physical and anticipated boundaries of swing were described. The coupling ratio $K_p$ demonstrates the preparedness of the person walking to enter the unstable equilibrium, as the sooner after toe-off the $K_p=1$ the more likely that the swing foot will be anterior to the body during the controlled fall. Having the swing foot in front of the CoM during late swing may allow more time to adapt to unexpected perturbations, and potentially make the swing foot more readily available to arrest the body in the case of a gait disturbance. In those with MS, across fixed speeds the entry into the unstable equilibrium occurred earlier in swing, yet the point when the swing foot took the lead from the CoM ($K_p=1$) also occurred earlier, possibly signifying a systematic compression of events leading up to the entry to the unstable equilibrium, such that the controlled fall can be entered without critical deficiencies in coordination. However, it appears that this attainment of an earlier $K_p=1$ by those in the MS group did not prevent a reduction
in the temporal margin between the CoM and swing foot passing the physical boundary, as was observed previously in the MS group (Remelius et al., 2011). Even though the swing foot takes the lead earlier in swing for those with MS as demonstrated here, this may not be enough to ensure a comparable temporal margin of the swing foot leading the body into the unstable equilibrium. When the swing foot is closer to the CoM in the entry into the unstable equilibrium the walker may be more susceptible to gait disturbances. Remelius et al. (2011) reported reductions in the temporal margin between the swing foot passing the stance foot toe and the COM entering into the unstable equilibrium by those with MS. When the CoM enters into the unstable equilibrium with the swing foot further in front of the stance foot there may be improvements in the ability to endure a gait disturbance in swing. By reporting coupling ratios of late swing, more information on the recapture from the unstable equilibrium was available for the description of movement as a whole during swing. What these coupling relations indicate is that changes in swing by those with MS are made primarily in the lead up to the transition into the unstable equilibrium, (the sum of early and mid swing times) compared with controls, which may increase the challenges to gait stability compared with controls. The alteration of the first two phases of swing may be attributed to factors that can be changed, moving the body over the stance foot more quickly.

Regardless of how late swing is initiated, it is important to gauge where and when it will be terminated. In the visual perception of the upcoming step landing area, the velocity minima of the proximal visual intersect occurred earlier in swing in the MS group compared with controls, indicating a potentially heightened need
for retrieving the information regarding the immediate vicinity earlier in the step for planning. This may be coinciding with the earlier initial ballistic throw of the body up over the stance foot and out in front of the stance foot and into the unstable equilibrium. Perhaps an earlier assessment of the potential step-landing environment facilitates a quicker execution of the swing of walking gait. Additionally, the field of view was oriented closer to the body and revealed more of the subsequent step-landing area, and in the general area of the environment more proximal to the body for the MS group compared to controls while the PVI was at the swing anterior velocity minimum. This directing of the field of view closer to the body can improve visual exteroception of the environment and increase the availability of visual proprioceptive information form sight of the body's segments within the field of view. Furthermore, the area of the step-landing area was less and less visible as walk speed increased, whereas the distance between the PVI and the CoM did not increase as walking speeds increased. There may be a set distance ahead that the field of view is oriented toward regardless of walk speed, but that can be “adjusted” to adapt to functional limitations such as those due to MS.

**Implications for MS and rehabilitation**

The findings of this study showed that those with MS have adapted the patterns of swing even at preferred walking speed, when swing times are the same as their non-MS counterparts. These adaptations included shortening early swing, perhaps to lengthen the mid swing phase and prepare for the entry into the unstable equilibrium, as well as a functional earlier visual assessment of the ground along the
transit path, with the field of view directed more proximal to the body. Many of the adaptations made in forming a gait strategy are likely adaptive and not the result of a conscious decision, but an underlying ongoing shift in patterns that relates to favorable outcomes. Keeping the body above the base of support may be the best strategy for coping with disturbances to balance. In addition, the adaption of body postures and movements (such as increased head and trunk pitch) that allow for greater visual input in those with sensory loss may be the best compensation available to augment the sensory system. Keeping the body above the posterior and anterior boundaries of the base of support while augmenting visual sensory information are adaptive traits that may respond to gait retraining, where the innate instruction to “look where you are going” and to keep your feet under you while walking can be reinforced in order to lessen the chances of experiencing a slip or trip that results in a fall.

**Study Limitations**

A limitation of this study is the relative heterogeneity of the MS group. The estimation of the proximal visual intersect was based on literature accepted norms of approximately -65 degrees from neutral, and individual variations form this average were not accounted for by direct measurement. A means to measure the true visual arc of each individual is explored in Appendix B-C.
Conclusions

During walking gait individuals with MS resume a stature above the base of support quickly after toe-off (and the end of dual support) and they seek to stay in this posture for longer compared with controls. Furthermore, people with MS tend to orient their field of view downward to a greater extent and do so earlier in swing than their non-MS counterparts. These patterns may represent additional facets of a protective gait strategy that keeps the center of mass of the body over the base of support during unipedal postures of walking gait, while identifying where when and how to reform a dual support stance with as much time to plan as possible. Since one cannot remain over the unipedal stance boundaries and keep forward progression, these individuals with MS moved into the unstable equilibrium and recaptured the CoM within the stability boundaries of dual support with hastened swing foot movements. As swing is shorter in those with MS at matched fixed walking speeds compared with controls, care must be exercised as the swing foot must be moved faster during swing, and a perturbation during swing could have a more immediate impact, resulting in a loss of balance and falls.

Acknowledgements

The authors thank the study participants. The national Multiple Sclerosis Society funded this study (RG 3974A2).
References


CHAPTER VII

SUMMARY AND FUTURE DIRECTIONS

Summary

People with multiple sclerosis (MS) often have problems with balance that have manifested as difficulty with mobility and in particular, difficulty with walking. Research has shown that people with MS walk with a slower preferred speed, shorter strides, and a longer dual support time, presumably to improve balance during gait. However, as over 50% of people with MS experience falls in activities of daily living, these adaptations to gait may not be entirely effective. While the longer time spent in dual support may appear to help balance, lengthening dual support may induce changes in other parameters of gait such as swing time, resulting in an overall reduction in gait stability. The current project consisted of three studies, which investigated how those with MS are adapting stride parameters, and in particular swing foot, CoM and head dynamics during swing. Study 1 was designed to investigate if differences in preferred walking speed are driving observed longer dual support times in those with MS through a comparison of walking patterns in non-MS individuals across a set of matched fixed walking speeds. Furthermore, if longer dual support was due to functional impairment from MS, assess induced changes in the swing phase of gait. In study 2, a novel definition of swing that included the context of the center of mass (CoM) location relative to the stance foot base of support boundaries was put forth. Then, the coordination of the approach to the physical anterior boundary of the stance foot (toe) and the heel-strike location
(anticipated boundary) was assessed with coupling ratios computed between the swing foot and CoM temporal approaches towards these boundaries. Finally, how the field of view may be oriented was described by measuring head movements during gait. In study 3, differences in the formation of swing based on the methods developed in the second study were assessed between individuals with and without MS at their preferred walking speed and across fixed walking speeds.

The first study experiment revealed that regardless of gait speed, people with MS walk with a longer dual support time and increased stride width. That those with MS lengthened dual support at all walking speeds reinforces earlier findings that were obtained from gait studies at preferred gait speed only (Benedetti et al., 1999; Kelleher et al., 2009; Martin et al., 2006). Combined with wider strides, those with MS may be striving to create a more stable base of support and to stay over it for a longer time. However, in fixed speed walking conditions lengthening dual support forced a shortening of swing time in those with MS compared to their non-MS counterparts. Furthermore, the alterations to swing went beyond simply shortening swing time, as those with MS shortened the temporal margin between when the swing foot and CoM crossed the stance foot toe, compared with controls. The CoM passing the stance foot toe is a marker that the body has entered the unstable equilibrium of gait, and an altered relationship to the swing foot during this transition may signify a susceptibility to gait disturbance in swing. In those with MS the body CoM passed the stance foot toe (entered the unstable equilibrium) sooner after the swing foot crossed the stance foot toe compared with controls. This alteration in timing coordination between the swing foot and the center of mass
may describe conditions that potentially afford less protection from perturbations during swing. Changes in swing timing may affect coordination patterns that limit the effectiveness of a gait strategy favoring longer dual support times. As trips most likely occur during swing, and slips occur most often at the termination of swing, understanding changes in swing may be particularly important to helping those with MS maintain their mobility.

The second study was performed with a group of twenty healthy young adults to develop new methods in order to better understand three aspects of swing. First, swing was partitioned into three phases (between toe-off and heel-strike), separated by the added context of the CoM transit over the posterior and anterior limits of the stance foot. Second, the coordination of the CoM with the swing foot was described with a coupling ratio during the approach to the unstable equilibrium and during the recapture of the CoM from the unstable equilibrium with the acquisition of heel contact. The coupling ratio was formed between temporal estimates of contact of the swing foot and CoM with the boundaries of the unstable equilibrium, computed as time-to-contact (TtC). Third, motions of the head during swing were used to describe the anterior progression of the field of view. These methods were applied to the young healthy adults while walking at preferred speed and during the steps taken while clearing an obstacle.

When young adults walked at preferred speed, they split swing into early swing (21.5% behind stance foot heel), mid swing (39.8% between stance foot toe heel and toe) and late swing (38.7% in front of stance foot toe). On the approach to the unstable equilibrium the swing foot took the lead from the CoM (K_p=1) at 27.2%
of swing. On the approach to the termination of the unstable equilibrium (at the heel-strike location), the coupling ratio $K_a$ between swing foot and CoM was 0.26 as the CoM entered the unstable equilibrium. Finally, head movements during gait engendered a pause in the field of view anterior progression at 29.1% of swing, which was located closer to the body than the subsequent heel strike location.

When stepping over an obstacle, the step with the trail foot was made with a shortened late swing phase compared with the lead foot step, and this was due to longer early and mid swing phases. The coupling ratio $K_p$ between $T_{TC_{swing\ foot}}$ and $T_{TC_{COM}}$ to the anterior stability limit at the stance foot toe ($B_p$) during the step over the obstacle with the lead foot reached 1.0 at 15% of swing. In contrast, $K_p$ did not reach 1.0 during the step over the obstacle with the trail foot, as the swing foot arrived at the $B_p$ after the CoM, which may indicate an unstable condition. The coupling ratio $K_a$ for the approach to the anticipated boundary (heel-strike location) was lower (0.259) during the step over the obstacle with the lead foot, compared with the step with the trail foot (0.363), which equates to a smaller gap between swing foot arrival and CoM arrival for the trail foot. Getting the swing foot to the ground earlier before CoM arrival may be important to establishing a safe base of support over which to move the CoM during gait. In obstacle conditions, the PVI occurred earlier in swing (20%) and was closer to the body than in unobstructed gait, signifying a greater reliance on visual information in negotiating the obstacle. Measures developed in this study showed systematic differences in the swing phase patterns between the step over the obstacle with the lead foot and trail foot, which
illustrates how these measures were sensitive gait task, and that differences may be observed in swing construction between those with MS when compared to controls.

The third study involved a description and identification of changes in the formation of the swing phase during walking at preferred and fixed imposed walking speeds in people with MS compared with controls. This study investigated 1) identify adaptations in timing to the three new phases of swing, 2) the timing of CoM entry into the unstable equilibrium, 3) the coordination between the swing foot and body CoM (coupling ratios), and 4) the head orientation during gait adopted. Those in the MS group systematically shortened early swing and lengthened mid swing compared with controls, even at preferred speed where no difference in total swing time duration was observed. Thus, the MS group spent more of swing over the posterior-anterior limits of the stance foot compared with controls. The longer mid swing times in the MS group may be a part of a gait strategy to keep the CoM above the base of support. Despite a longer mid swing, the sum of early and mid swing times illustrated that the MS group moved the body CoM into the unstable equilibrium sooner after toe-off in the fixed speed conditions, yet the MS group maintained similar late swing times compared with the control group. This sparing of late swing across all walking speeds is particularly interesting considering that the young healthy adults did significantly alter late swing when stepping over an obstacle. However, the MS group reached $K_p=1$ earlier in swing compared with controls at fixed speeds, which was a feature of the young healthy group’s step over the obstacle with the lead foot. This earlier $K_p=1$ which may indicate that walking for those with MS may equate to a series of steps over perceived obstacles.
Coincident with the shortening of early swing, the MS group also paused the anterior sweep of the field of view earlier after toe-off and closer to the body when compared to controls. The field of view limits derived from head orientation show that vision may be relied upon more heavily by those with MS, supported by their lower measured sensitivity to touch perception in the vibration testing results compared with controls. Substituting vision for other diminished sources of sensory input may be an adaptation that occurs in many activities of daily living, including locomotion. Vision is often impaired in those with MS, yet even diminished visual perception may be beneficial in counteracting more severe peripheral sensation loss. It appears that swing is adapted functionally by those with MS to both increase the time that can be spent over the posterior anterior limits of the stance foot and provide earlier and more proximal field of view availability for visual exproprioception compared to non-MS controls. However, the shortening of early swing, and total swing time at fixed speeds may lead to less adaptable swing foot motion and reliable placement at heel-strike.

**Future Directions**

The results from the current series of investigations have provided direction for future work. First, while care was taken in the selection of individuals for these studies, the MS study population was somewhat heterogeneous as EDSS scores had a fairly wide range between 2 and 6. The MS population as a whole is quite heterogeneous, therefore it is not unusual that a sample of people from this group was as well. There may be additional insights gained from splitting the MS group
into sub-groups based on EDSS score, symptom type and or severity, such as sensitivity to touch and vibration. Somatosensory loss diminishes the internal sense of where segments of the body are in relation to one another and the environment, which can impair balance. Grouping the MS individuals as below or above a threshold for peripheral sensation loss may result in the ability to link head movements and resulting field of view orientation to strategies that may supplement for diminished tactile perception.

Second, the head centric measure for describing the field of view limit dynamics developed here can potentially be expanded by adding the context of eye motion. As the eyes operate from within the head, eye movements and/or fixations that occur before after or during the velocity minimum in anterior progression of the proximal field of view limit may provide additional insights into the “circumstance the visual field is operating within” (Lee 1980) that the literature on gaze tracking has lacked. By adding eye gaze tracking to the head tracking, the eye gaze can be superimposed within the field of view, placing the limits of foveal vision into context of the limits of the field of view. If the two fields are moving together, the eye is quasi-static within the head, which may play an important role in the perception of self-motion, which may be especially relevant in tasks of locomotion where the body is in motion in the anterior direction. However, these measures of head motion relating to the gaze behavior to the global field of view may be expanded to reveal important clues regarding altering heading direction and the perception and adjustment of heading when walking along non-linear paths.
Another possible expansion of the current research is the analysis of approach to stability boundaries during gait that involves medio-lateral control. With the exception of stride width, all analyses in the current project described anterior-posterior motion in the transverse plane. The development of time to contact with a compound boundary such as the anterior and lateral edges of the stance foot could relate to lateral gait stability, control, and potential coupling with anterior displacements of the body in gait. This may describe a coupling within one variable, the body CoM during lateral and anterior motions, or relate the lateral displacement of the CoM to the stride width in pure frontal plane behavior.

Further description of the PVI motion in the sagittal plane is possible as well. A PVI anterior velocity minimum consistently occurred near the toe-off of swing during each step, but the absolute velocity at that point in the PVI anterior progression was not as consistent. The PVI velocity minimum dropped to varying degrees from the local mean, sometimes even crossing zero (especially at slow gait speeds) indicating that vision may be used differently during subsequent steps, especially at different walking speeds. This is not surprising given the “higher order” demands on vision as we commonly do more than one thing at a time, such as gaze upon works of art in a sculpture garden while walking, and we need not attend to every step especially when walking over unobstructed terrain. An array of vectors declined at systematic angles departing from the neutral plane of the head all the way to the -65° lower limit of the field of view may illustrate which area in the anterior field of view does in fact reach a zero anterior velocity during the pause in PVI progression. This identification of the center of local optic expansion during
swing may correlate with the subsequent step length and duration in a way that the PVI (at -65°) alone cannot describe.

Learning more about the use of vision in locomotion is an open research topic. The methods developed here to describe head movement and the resulting field of view orientation can be applied to a wider range of walking gait speeds, and to the other principle locomotion mode of running. Adaptations to stride parameters across the physiological range of walking and running speeds may be better explained if field of view behavior derived from head motion is added to investigations of gait, thus enhancing models of step formation by clearly illustrating the “circumstance in which vision is operating” (Lee 1980).
References


APPENDIX A

INFORMED CONSENT
RESEARCH INFORMED CONSENT FORM

Project Title: Dynamic balance control and fatigue in Multiple Sclerosis: Walking Study


Department of Kinesiology, University of Massachusetts, Amherst, MA.

Project Sponsor: National Multiple Sclerosis Society

By signing this consent form you, ________________, indicate that you willingly agree to participate in this project. The essence of this project is as follows:

Purpose of the Study

The purpose of this study is to assess aspects of balance control during walking at different speeds in people with Multiple Sclerosis. In this study, walking will be assessed using data from special cameras that see reflective markers, and a special platform that measures footfall patterns. These cameras can make high-speed assessments of your walking by very accurately tracking your movements. Results from this study will allow clinicians and researchers to gain a better understanding of the balance control required by individuals with MS to walk in their daily lives. The information that will be obtained can be useful in the evaluation and development of treatment, early diagnosis, and tailored physical therapy programs.

Procedures

The experiment will comprise of a total of three separate sessions:

1. On visit 1 you will be asked to fill out questions regarding your fatigue state and health status, and will be assessed for walking ability (10 meters [32.8 feet] at normal and brisk pace) and sensory sensitivity (timed vibration test with tuning fork). In this session you will also receive training on the equipment by which we assess your muscle strength.
2. On visit 2 you will be asked to walk at four different speeds overground.
3. On visit 3 you will also be tested for your muscle strength in your leg using a strength-testing machine (Biodex).

Each visit is expected to take two hours.

Expected risks or discomfort
The risks involved in the project are not different from what you encounter in normal daily life. We have added safety precautions in the form of harnesses and physical support whenever you need it. We will provide sufficient rest between walking and strength testing measurement to prevent you from experiencing fatigue.

**Expected Benefits**  
No direct benefits to your health are expected to occur by participating in this study. The results will help our understanding of balance control and propensity for falling, especially in individuals with MS.

**Cost and Compensation**  
There is no personal cost involved by participating in this study. By participating in this study you will receive a participant stipend of $50 to cover transportation costs. The University of Massachusetts does not have a program for compensating subjects for injury or complications related to human subjects research but the study personnel will assist you in getting treatment.

**Alternative Procedures**  
There are no alternative procedures. The measurements in this study provide the best and optimal way to measure and assess your balance and muscle strength.

**Subject enrollment/length of study**  
It is expected that 40 participants will be enrolled in this study (20 people with MS and 20 age-matched non-MS participants). This study is expected to last for one year and is part of a larger 4-year study on fatigue and balance control in MS. You will be asked to visit the laboratory on three separate sessions over a 4-6 week period. Each testing session will last for approximately 2 hours.

**Confidentiality**  
Information produced by this study will be confidential and private. If the data are used for publication in the scientific literature or for teaching purposes, no names will be used. Information obtained from this study will not be released to anyone except upon your written request.

**Voluntary Participation**  
You are under no obligation to participate in this project. You may withdraw your participation at any time without prejudice.

**Request for Additional Information**  
Should you have any questions about your treatment or any other matter relative to your participation in this project, you may call: Richard Van Emmerik at (413-545-0325; email: rvanemmerik@kin.umass.edu) or Jane Kent-Braun at (413-545-9477; email: janekb@kin.umass.edu). If you experience a research related injury at any time during this study, you may contact: Richard Van Emmerik at (413)545-0325 (work) or (413)259-1040 (home). If you would like to speak with someone not
directly involved in the research study, you may contact the Human Research Protection Office at the University of Massachusetts via email at humansubjects@ora.umass.edu; telephone (413) 545-3428; or mail at the Human Research Protection Office, Research Administration Building, University of Massachusetts Amherst, 70 Butterfield Terrace, Amherst, MA 01003-9242.

Subject Statement of Voluntary Consent
When signing this form I am agreeing to voluntarily enter this study. I understand that, by signing this document, I do not waive any of my legal rights. I have had a chance to read this consent form, and it was explained to me in a language which I use and understand. I have had the opportunity to ask questions and have received satisfactory answers. A copy of this signed Informed Consent Form has been given to me.

____________________________________________________
Subject/Parent Guardian's Name (Print or type)

____________________________________________________
Signature

____________________________________________________
Date

If required: Witness (Print or type) to ☐ Discussion ☐ Signature

____________________________________________________
Signature

____________________________________________________
Date

Study Representative Statement:
I have explained the purpose of the research, the study procedures, the possible risks and discomforts, the possible benefits, and have answered any questions to the best of my ability.

____________________________________________________
Study Representative Name (Print or Type)

____________________________________________________
Date

____________________________________________________
Signature

____________________________________________________
Date
RESEARCH INFORMED CONSENT FORM

Project Title: How we move our head to visually anticipate stepping during gait

Investigators: Jebb Remelius, Karthick Sugumar, Richard E.A. Van Emmerik; Department of Kinesiology, University of Massachusetts, Amherst, MA.

Project Sponsor: None

By signing this consent form you, ____________________, indicate that you willingly agree to participate in this project. The essence of this project is as follows:

**Purpose of the Study**
The purpose of this study is to assess the use of peripheral vision during walking gait. In this study, motion of the body and forces of foot contact will be measured using data from a high-speed camera system that can very accurately track your movements and a special force sensing walkway. Results from this study will allow researchers, clinicians, and scientists to gain a better understanding of importance of vision during tasks of daily living. The information that will be obtained can be useful in the overall assessment of how vision is used in other populations with clinical disability and movement disorders.

**Procedures**
The experiment will comprise of a total of one session.

During this visit you will;
1. Be measured for height, weight, and other relevant body measurements needed to correctly assess human movement.
2. Have reflective markers placed on various segments of your body, which will be used to track your movement during walking.
3. Have the limits of your peripheral vision mapped by looking for objects moved into your field of view, while holding your head in a fixed position.
4. Walk along the boardwalk at a normal comfortable speed.
5. Walk along the boardwalk at your normal comfortable speed and step over a small obstacle placed across the walkway.
6. Walk with increased or decreased speed, as instructed by the study administrators.
7. These collections, under the conditions and speeds will allow us to determine the way we use visual information to guide our steps.

The visit is expected to last for approximately two hours (2 hours).

**Expected risks or discomfort**
The risks involved in the project are not different from what you encounter in normal daily life, and is significantly less than you encounter during recreational sports.

**Expected Benefits**
No direct benefits to your health are expected to occur by participating in this study. The results will help our understanding of the use of the visual field during walking.

**Cost and Compensation**
There is no personal cost involved by participating in this study. The University of Massachusetts does not have a program for compensating subjects for injury or complications related to human subject's research but the study personnel will assist you in getting treatment.

**Alternative Procedures**
There are no alternative procedures. The measurements in this study provide the best and optimal way to measure and assess your walking and body movements.

**Subject enrollment/length of study**
It is expected that 18 subjects will be enrolled in this study. This study is expected to last for 2 months. You will be asked to visit the laboratory on one occasion at a time of your convenience. The testing session will last for approximately 1.5 hours.

**Confidentiality**
Information produced by this study will be confidential and private. Participant names will be coded with number designations to ensure confidentiality. All electronic data will be stored on password-protected computers and informed consent documents will be stored in a locked filing cabinet in the motor control laboratory. If the data are used for publication in the scientific literature or for teaching purposes, no names will be used. Information obtained from this study will not be released to anyone except upon your written request.

**Voluntary Participation**
You are under no obligation to participate in this project. You may withdraw your participation at any time without prejudice.

**Request for Additional Information**
Should you have any questions about your treatment or any other matter relative to your participation in this project, you may call: Richard Van Emmerik at (413-545-0325; email: rvanemmerik@kin.umass.edu), Jebb Remelius (413-687-4089; email: jebb@kin.umass.edu) or Karthick Sugumar: email ksugumar@student.umass.edu. If you experience a research related injury at any time during this study, you may contact: Richard Van Emmerik at (413)545-0325 (work) or (413)259-1040 (home). If you would like to speak with someone not directly involved in the research study, you may contact / the Human Research Protection Office at the University of Massachusetts via email/ /at humansubjects@ora.umass.edu <mailto:humansubjects@ora.umass.edu>; telephone (413) 545-3428; or mail at the
Human Research Protection Office, Research Administration Building, University of Massachusetts Amherst, 70 Butterfield Terrace, Amherst, MA 01003-9242.

**Subject Statement of Voluntary Consent**
When signing this form I am agreeing to voluntarily enter this study. I understand that, by signing this document, I do not waive any of my legal rights. I have had a chance to read this consent form, and it was explained to me in a language which I use and understand. I have had the opportunity to ask questions and have received satisfactory answers. A copy of this signed Informed Consent Form has been given to me.

Subject’s Name (Print or type)

__________________________________________________________
Signature

__________________________________________________________
Date

If required: Witness (Print or type) to ☐ Discussion ☐ Signature

__________________________________________________________
Signature

__________________________________________________________
Date

**Study Representative Statement:**
I have explained the purpose of the research, the study procedures, the possible risks and discomforts, the possible benefits, and have answered any questions to the best of my ability.

Study Representative Name (Print or Type)

__________________________________________________________
Date

__________________________________________________________
Signature

__________________________________________________________
Date
APPENDIX B

DEMARKING THE FRANKFURT PLANE
Visual information is valuable to the success of movements underway and for the planning of future movements. During the dynamics of movement, a sighted organism is constantly reorienting the field of view in order to perceive information. The field of view is described as the area of one’s surroundings that is visible at one time. How the field of view is orientated may provide insight into how vision is used to construct movements. In humans, the field of view is frequently referenced to the neutral plane of the head, which allows for comparisons between individuals. For the purpose of this report, the Frankfurt Plane (The American heritage medical dictionary 2007) was chosen to represent an anatomic reference to the neutral plane of the head, and how to delineate the Frankfort Plane is described herein.

The Frankfurt Plane, established at the 1884 World Congress of Anthropology in Frankfurt Germany, is described as the plane of the head that most closely parallels the surface of the earth during quiet upright standing (Figure 1.6 inset). The Frankfurt Plane lies parallel to the bilateral upper margins of the ear canal or porion (passing through the center of the vestibular rings) and the inferior limit of the orbits of the eyes (orbitale). The Frankfurt Plane is frequently used in orthodontics and radiography but has had limited use in biomechanical investigations of human movement. The Frankfurt Plane has been shown to have good reliability in forming a repeatable measurement of the neutral plane of the head that is closely aligned with true horizontal. However, some caution must be exercised in its use as Madsen et al. showed that the Frankfurt Plane can be pitched 1-5 degrees down from true horizontal (Madsen, Sampson, & Townsend, 2008). This difference between true horizontal and the Frankfurt Plane is not an issue in
comparative studies, where between group differences are of interest, as each individual will be affected by the same offset from true horizontal.

Figure B.1. Proximal visual intersect (PVI) with the ground (circled: location of local PVI velocity minima) during early-swing showing fixation on upcoming step-landing area contacted at the subsequent left heel-strike (lighter); inset: Frankfurt Plane and a -65° declination describing the proximal limit of the lower visual field.

In order to identify the Frankfurt Plane of the human head, and track the motion of the plane with a motion capture system, four anatomical reference points are identified. Once the plane is demarked, the declination of the Frankfurt Plane relative to horizontal is reported as an indicator of the postural orientation of the head during quiet upright stance in all 20 participants of Data Set "A", a group of young healthy adults, and 24 members of Data Set “B”, 12 older healthy adults and 12 older persons with multiple sclerosis.
Frankfurt Plane identification

Demarking the Frankfurt Plane was accomplished with four virtual landmarks referenced to three 3D tracking markers also used to track the head. Three tracking markers are necessary to create a virtual landmark as the virtual landmark is located by a 3D offset from the plane formed by the three chosen tracking markers. An adjustable helmet suspension apparatus with five tracking markers was placed on the crown of the participant's head and adjusted with a circumferential strap to securely fix the markers to the scull. The anatomical landmarks of the Frankfurt Plant were identified by palpation of the crest of the inferior orbits and visual identification of the point above the ear canal (or porion) (Figure B.2). Placing the tip of the Visual 3D pointer rod at each of the four identified locations in turn and compressing the spring mechanism of the Visual3D pointer rod demarked these anatomical landmarks. The spring pointer rod has one tracking marker fixed to the base of the handle, and one tracking marker positioned 0.123 meters from the tip of the spring rod. A 3D distance measurement algorithm in Visual 3D detects the initiation of spring compression events and creates a temporal event one frame prior to the onset of compression at the projected location of the tip of the spring pointer rod. In this way, virtual landmarks are formed at the pointer rod tip location the moment the tip has been positioned precisely. In Visual 3D, a 3D offset to the virtual landmark was created by computing distances from the pointer rod tip at the spring compression event relative to the local coordinate system of the plane formed by the bilateral posterior and crown tracking markers on the helmet.
From the four virtual landmarks created in this manner, a Frankfurt Plane segment was created. A least squares fit between the four landmarks was used to orient the local coordinate system of the Frankfurt Plane segment where the positive X direction is lateral to the right, the positive Y direction is anterior, and the positive Z direction is vertical following the right hand rule. This Frankfurt Plane segment was located by the four virtual landmarks and tracked using all five tracking markers on the helmet suspension device. During subsequent kinematic data recordings, the 3D location and orientation of the Frankfurt Plane segment is fixed relative to the helmet suspension device, such that motion of the head can be reported relative to the neutral plane of the head.

![Figure B.2. Marker placement and landmark definition demarking the Frankfurt Plane. Darker markers: segment definition landmarks.](image)

**Participant Instruction**

Participants were asked to stand quietly in the upright “anatomical position” with their gaze directed to the far wall at eye level for two seconds. The goal of this anatomical position posture was described as to stand erect yet relaxed while remaining as till as possible, with the head level, the feet a comfortable width apart, and the arms akimbo. Each participant was asked to stand upright, but not to assume a rigid “standing at attention” character. Kinematic recordings consisted of
two seconds of this quiet upright anatomical stance followed by enough time to position the pointer rod once at each of the four head landmark locations (bilateral porion and bilateral lower orbital ridge). Note: In Data set “A”, the young healthy adults were asked to adopt a 1,000-yard stare, but the anterior visual limit was a curtain hanging from the ceiling approximately 5 m anterior to the participant. In Data Set “B”, participants were asked to look toward the far wall, which was approximately 20 m anterior from the participant, as no curtain was present.

**Results**

**Subject Characteristics**

This procedure was conducted for all 20 participants of Data Set “A”, a group of young healthy adults, and 24 members of Data Set “B”, 12 older healthy adults and 12 older persons with multiple sclerosis. The participant group of young healthy adults was a sample of convenience, selected from the student body of graduate and undergraduate Kinesiology members. The young group, composed of 10 men and 10 women, ranged in age from 20 to 38 years of age (mean 28.3±5.0), ranged from 159 to 183 cm in height (170±8), and ranged from 54.4 to 95.3 kg in weight (68.6±12.4). The participant group of older healthy adults consisted of 12 women, selected as an age matched control group for comparisons to the MS group, ranging from 34 to 69 years of age (mean 51.9±11.5), 109 to 178 cm in height (mean 144±20 cm), and 61.3 to 69.5 kg in weight (mean 64.7±2.4 kg). The participant group of older adults with multiple sclerosis included 12 women ranging from 32 to 65 years of age (mean 51.4±8.3 years), 124 to 193 cm in height (mean 153±23 cm),
and 63.5 to 67.5 kg in weight (mean 65.2±1.4 kg). All participants signed an informed consent document approved by the University of Massachusetts, Amherst.

**Frankfurt Plane Orientation during Quiet Standing**

This report summarizes the orientation of the neutral plane of the head as measured by the Frankfurt Plane during the anatomical position assumed by each participant during the first two seconds of the model calibration data collection.

The Frankfurt Plane was pitched downward from parallel with the ground: the young adult group (range 10.1° to -12.3°) -5.05°±6.72°; in the older adult control group (range 7.5° to -9.6°) -1.29°±6.45°; in the MS group (range 0.8° to -14.3°) -4.13°±4.59° (Figure B.3). MS and control groups were not different (p>0.09). Older control and young group were not significantly different (p>0.06). The young group was not significantly different from the MS group (p>0.68).
Figure B.3: Angle of the Frankfurt Plane relative to lab coordinate system horizontal (0°) of each participant in the young group (triangles) the older adults (diamonds) and the MS group (bars) during two seconds of quiet upright standing. Thick bars represents group mean angle.

**Discussion**

On average, the Frankfurt Plane during quiet stance was pitched forward and declined from horizontal in all three groups. It should be noted that the quiet standing anatomical position participants were instructed to assume was not intended to align the head with true horizontal. Therefore, the variability of the Frankfurt Plane angles relative to horizontal seen across these 44 participants may be due to each individual’s natural postural habits of head orientation during stance, or differences in interpretation of instructions. The quiet upright “anatomical position” the participants were asked to assume is not an adequate instruction to
align the head with true horizontal. Methods of aligning the neutral plane of the head with horizontal include instructing the participant to look into a reflection of their own eyes in a vertically positioned mirror, or by conducting a sequence of choreographed head pitch exercises shown to yield a repeatable neutral alignment. Although the group difference between young and older healthy adults was not significant, the trend toward a significant difference may be due to the curtain installed for collections of data set “A” (young), which considerably shortened the distance at which participants were instructed to fixate upon in the protocol instruction. For each group the alignment of the Frankfurt Plane was declined below horizontal on average between -1° and -5° as reported by Madsen et al. (Madsen et al., 2008), yet the range of values recorded far exceeded the -1° to -5° net declination.
APPENDIX C

MAPPING THE LOWER LIMIT OF THE FIELD OF VIEW IN QUIET STANCE
Describing the size of the lower visual field of view may aid in the understanding of what visual information is available to an individual during dynamic tasks such as gait and how the individual may be using vision to construct these dynamics. Field of view refers to the maximum visual arc or angle at which an individual can detect a stimulus in the visual periphery. Perimetry is the science of measuring the size and shape of the peripheral field of view. Perimetry is generally utilized as a diagnostic tool to quantify visual field loss due to injury or disease, and absolute measures of perimetry map the function of the retina. During absolute measurements of the field of view, the perimetrist will instruct the participant to re-orientate the head during collections to overcome any encroachment of the anatomical features of the scull into the field of view. This is necessary for a complete map of the retina for individuals whose faces have a prominent brow, or high cheekbones. In absolute perimetry measurements, the field of view is referenced to the fixation point of the eye. However, relative measurements of perimetry are also commonly performed. In relative field of view measurement the head is not moved. Keeping the head fixated reduces the visual arc to some degree due to intrusions by formations of the orbits, and reports measurements of the field of view referenced to the head.

A relative field of view perimetry measurement relates directly to the normal circumstance of the field of view in activities of daily living; where a person must re-orient the head in order to intentionally bring the parts of the environment occluded by the anatomy of the scull into the field of view. This is important when describing the field of view as the area of one’s surroundings that is visible at one time. That
the field of view is primarily restricted by head orientation is the impetus for developing an ecological method of mapping the lower limit of the visual field in relation to the head orientation. Perimetry is typically performed with the participant seated and the head situated in a fixation device. However, this type of perimetry testing has a major shortcoming in that it lacks context with regard to common behaviors such as moving overground with walking gait while interacting with features of the environment. If the use of the field of view is to be better understood in context of human movement, an ecological method for field of view mapping is necessary.

The extents of the field of view in the sagittal plane have been measured to encompass a span of approximately 110° (Trevarthen, 1968). Of this 110° approximately 45° of the field of view is superior to the neutral plane of the head (Frankfurt Plane). Of particular interest in this report is the inferior or lower field of view that extends approximately 65° below the neutral plane of the head (Figure 6.1). It is necessary to employ a measurement technique that is in keeping with the context of overground locomotion and postural tasks of daily living. Put forth here was a method to determine the limits of the field of view in context with an upright standing posture, which more closely represents the type of visual information accessible from within the field of view of the environment in unobstructed conditions.
Figure C.1. Visual field limits during quiet stance and level gaze (A) and during stepping with the gaze pitched downward (B). Frankfurt Plane represented at 0° in B and -30° in C. Adapted from Trevarthen (Trevarthen, 1968)

**Mapping the lower field of vision**

The post-processing mapping of the lower field of view was accomplished in Visual 3D by creating a kinematic model segment called the lower field of view segment (LFV) with an angular declination initially set at -65° from the Frankfurt Plane segment as reported by Trevarthen (described in Appendix B). The LFV segment was constructed between the pair of orbitale virtual landmarks created for the Frankfurt Plane segment and two new virtual landmarks superior to the porion virtual landmarks. The offset superior to the porion virtual landmarks was perpendicular to the Frankfurt Plane segment and this vertical offset was computed by multiplying the distance from the orbital landmark to the porion landmark by the tangent of the visual angle initially set at the literature based declination limit of -
65°. In this way, a right triangle was formed where an angle could be input, creating a plane that was oriented inferior to the Frankfurt Plane in pitch.

Once the LFV segment was created with the initial -65° angular offset from the Frankfurt Plane segment, the motion of the LFV was extracted from the 3D motion of the head tracking marker cluster. From the motion of the LFV segment, the projected intersection of the LFV segment and the ground was tracked and from this intersection, the declination from neutral of the lower field of vision was measured. The intersection between the LFV segment and the ground was computed by taking the dot product between a unit-vector projected perpendicular from the floor and a unit-vector projected in the anterior direction from the LFV segment. This intersection between the LVF and the ground forms the dependent variable called the *proximal visual intersect* (PVI). In this way, when person stands facing in the anterior direction, features of the environment anterior to the PVI are a visible part of the field of view, while elements of the environment on the ground posterior to the PVI are not visible and not accessible by the visual system. This intersect describes the lower limit of the visual field. However, the LFV and the resulting PVI must be adjusted to the anatomical limits of an individual’s field of view.

After the PVI was created at the initial angle of declination of -65° a measurement of head orientation was performed at each of five marked locations on the floor. Five gaze tracking markers were placed on the floor in a line parallel with the midline of the body in the anterior direction. These markers were placed in front of a study participant in the kinematic collection volume who was wearing the
helmet suspension device at 0.25m intervals from 0.25m to 1.25m anterior to the bilateral tibial epicondyle. After a virtual landmark was created at each gaze tracking marker location, the marker was removed, and this was repeated on all gaze tracking markers from distal to proximal. The average anterior posterior difference between the PVI and the gaze-tracking marker locations was computed and the LFV segment declination angle was changed systematically until the average difference between each of the five distances between the PVI and the gaze tracking marker locations was zero. In order to create the landmarks at the gaze tracking marker locations, it was necessary for the participant to understand and comply with the instructions.

**Participant Instruction**

A consecutive kinematic collection to the Frankfurt Plane measurement was made to map the lower visual field. The collection time was open ended to allow participants adequate time in establishing their head orientation at each fixation location. Participants were instructed to look at the marker positioned at the furthest distance from their body (in the anterior direction). Then while maintaining an upright standing body position, slowly pitch their head backward until the marker on the floor was no longer visible, and forward until the marker became visible again. Head pitch instructions were given as upward and downward, but the conventions of pitch can also be instructed as forward pitch and backward pitch, and superior and inferior pitch. The participant was instructed to perform a series of these forward and backward head pitch movements until a pitch orientation was attained that kept the marker consistently “just visible” at the lower limit of their
field of view. The participant was free to shift their eye position within the head, and “check” the visibility of the target by gazing forward and using the peripheral vision, and gazing downward to use foveal vision on the target while maintain head orientation. For those participants wearing glasses, they were instructed to look under the glasses if that increased the amount they were able to pitch their head back and while keeping the marker visible. The participant was asked to maintain a quiet upright standing position and verbally report when the furthest gaze tracking marker from the body (starting at 125mm anterior) was just visible after a series of voluntary changes in the pitch of the head in the forward and backward directions brought the tracking marker to the lower limit of the field of view. Upon the participant’s verbal report that the gaze tracking marker was just visible, a spring pointer event was recorded at the gaze tracking marker location on the floor. The farthest anterior marker was then removed and the process of adjusting head pitch by the participant was repeated on the next most distal gaze-tracking marker until the participant reported that the current gaze-tracking marker was at the lower limit of their field of view. The gaze tracking markers used for visual identification had a coating of silver retro-reflective tape that had a strong and distinct contrast against the black walkway ground surface. During post-processing an iterative process was used to determine the inferior sagittal angle to the lower field of view limit. To arrive at the inferior sagittal angle, the average difference between the PVI and the set of gaze-tracking markers was set at less than ±0.002 m.
Results

Subject Characteristics

This study was conducted on all 20 participants of Data Set “A”, young healthy adults, and 12 older healthy adults and 12 older individuals with MS that were members of Data Set “B”. The participant group of young healthy adults was a sample of convenience, selected from the student body of graduate and undergraduate Kinesiology members. The young group was composed of 10 men and 10 women, and ranged in age from 20 to 38 years of age (mean 28.3±5.0), ranged from 159 to 183 cm in height (170±8), and ranged from 54.4 to 95.3 kg in weight (68.6±12.4). The participant group of older healthy adults included 12 women ranging from 34 to 69 years old (mean 51.9±11.5), 109 to 178 cm in height (mean 144±20 cm) and 61.3 to 69.5 kg (mean 64.7±2.4 kg). The participant group of older adults with multiple sclerosis included 12 women ranging from 32 to 65 years of age (mean 51.4±8.3 years), 124 to 193 cm in height (mean 153±26 cm) and 63.5 to 67.5 kg (mean 65.2±1.4 kg). The MS participants had EDSS that ranged from 2.5 to 6. All participants signed an informed consent document approved by the University of Massachusetts, Amherst.

Lower limit of the field of view relative to the Frankfurt Plane

By systematically adjusting the angular offset used to create the LVF segment orientation from a starting point of 65° declination, the difference in the anterior direction at each spring pointer event between the gaze tracking marker corresponding to the spring pointer event and the PVI was minimized. The inferior
angle of the lower field of view was adjusted until the average of all five distances from the PVI to the corresponding gaze-tracking marker event was ±0.002 m in the anterior direction.

When the lower visual field limit is measured from the Frankfurt Plane the lower visual field ranged from -48.6° to -72.8° (mean -61.7°±6.9°) in the 20 young adults from data set “A”. For the 12 healthy older adults from data set “B”, the angle describing the lower limit of the field of view ranged from -57.9° to -75.4° (mean -69.7°±7.92°), where for 12 adults with multiple sclerosis the angle ranged from -48.5° to -66.1° (mean -60.5°±5.02°) (Figure C.1). When the lower field of view limit is measured from the Frankfurt Plane, the control group of older healthy adults showed a larger declination than the young group (p=0.01) and the MS group (p=0.002), but the young adult and MS group were not different (p=0.60).
Figure C.2: Declination from Frankfurt Plane to the lower limit of the field of view as measured by the average of five self-reported measurements in Young (diamonds), Control (diamonds) and MS (dash) groups. Thick bar in each group represents group mean angle.

By adding the declination from horizontal reported in Appendix B to the lower visual field angle computed here, the declination of the visual field limit can be reported in reference to the horizontal. When the lower visual field limit is measured from horizontal: In the 20 young adults from data set “A” the lower visual field ranged from -43.1° to -84.8° (mean -66.79°±9.35°). For the 12 healthy older adults from data set “B”, the angle describing the lower limit of the field of view ranged from -55.7° to -83.4° (mean -70.30°±9.54°), where for 12 adults with MS the angle ranged from -52.1° to -73.9° (mean -64.67°±6.13°) (Figure C.2). When the lower limit of the field of view was taken as a declination from horizontal there were no group differences in the size of the visual field: controls vs. young (p=0.54); controls vs. MS (p=0.19) and MS vs. young (p=0.49).
Figure C.3: Declination from horizontal to the lower limit of the field of view as measured by the average of five self-reported measurements in Young (diamonds), Control (diamonds) and MS (dash) groups. Thick bar in each group represents group mean angle.

**Discussion**

The lower visual field measurement described here as the proximal visual intersect represents the intersection of the lower limit of the field of view and the ground. Between participant differences in declination from the Frankfurt Plane to the PVI ostensibly are due to differences in the anatomy of the scull. However, without absolute measurements of perimetry it is not possible to discern whether differences between individuals result from the function of the retina or the anatomy of the scull. Differences in the shape of the lower visual field may result in altered head motion during dynamic tasks. An individual with a shallower angle to the lower field of view limit may need a greater amount of head forward pitch than a person with a deeper angle to the lower field of view limit if the perception of
information at similar proximal distances from the body was important. A correlation between the declination to the PVI and head pitch during gait may provide information on the reliance of vision during gait.

Measuring the PVI angle from horizontal diminished group differences to non-significant levels because the group mean angles became closer to one another but also because adding the declination of the Frankfurt Plane to horizontal increased the variability around the mean angle of the PVI for all groups. A participant may have a smaller lower field of view (lower PVI angle from the Frankfurt Plane) but stand quietly with the head pitched upward to a greater extent. This situation will make the span of the lower field of view appear smaller in this participant due to the addition of a positive pitch angle to a negative angle that describes the declination to the PVI from the Frankfurt Plane. In the opposite case, a participant may have a great deal more field of view than other participants while holding their head with a greater downward pitch, artificially increasing the measure of the lower field of view. The correction of the lower limit of the field of view from the Frankfurt Plane to the horizontal does not decrease variability of the declination angle of the PVI.

The PVI describes the lower limit of the field of view with the ecological context intact, as it was based on a standing posture and conducted by bringing objects on the ground into the field of view. The advantage to measuring the lower visual field while standing is that a better understanding of how we use vision during activities where the body is carried in an upright posture is possible.
APPENDIX D

PROXIMAL VISUAL INTERSECT MOTION DURING WALKING GAIT
The intersection between the anatomically limited relative lower field of view and the ground during quiet standing has been identified and termed the Proximal Visual Intersect (PVI). The PVI describes the lower limit of the field of view with the ecological context intact, as it was based on a standing posture and conducted by bringing objects on the ground into the field of view. The context retained when measuring the lower visual field while standing is that a better understanding of how we use vision during activities where the body is carried in an upright posture is possible. However, the true advantage to observing the PVI may be in following the movement of the PVI during dynamic tasks of daily living, such as walking gait. The motion of the PVI may yield great insight into how an individual may be functionally extracting visual information from the lower field of view.

The PVI dynamics that emerge during gait may serve to describe how and when the visual field is being functionally shifted during dynamics of motion (Figure 1.6). This report serves as a descriptive account of how the PVI (developed in Appendix C) emerges from coordinated dynamics within the body. The proximal visual intersect is a resultant measure stemming from the movement of the body, the movement of the head on the body and the lower limit of the field of view. Specifically of interest is how and when the anterior velocity of the PVI reaches a local minimum during a gait cycle (PVIvm) as it transits the ground during walking gait (Figure 1.7).

**Behavior of the Proximal Visual Intersect during Walking Gait**

A plot of the anterior progression of the PVI (Figure D.1) on average shows a quasi-linear displacement for the first third of the time between heel-strikes, then a
pause in translation with little anterior displacement during the middle third of the time between heel-strikes, followed by another period of quasi-linear displacement. The total displacement of the PVI during one stride is approximately equal in length to one stride of walking gait. The quasi-static pause in anterior translation corresponds with slower velocity of the PVI. By taking the first derivative of the position of the PVI, velocity of the PVI is described. Examination of the PVI during walking gait shows a clear systematic drop in anterior PVI velocity that occurs early in the swing phase of gait (Figure D.2).

![Graph](image-url)

**Figure D.1:** Mean PVI anterior progression across a portion of a walking gait cycle from Left Heel Strike to Right Heel Strike. PVIvm timing is shown as red ticks, and toe-off is shown as a cross. Toe-off occurs prior to the temporary slowing of the PVI progression (flatter section of graph over time centered at the PVIvm tick).
Figure D.2: Anterior PVI velocity across a portion of the gait cycle from left heel-strike to right heel-strike from 18 walking steps at preferred speed. PVI anterior velocity minimum event is shown as vertical ticks, and toe-off is shown as crosses. After Toe-off the anterior PVI velocity begins to slow until it reaches the PVI anterior velocity minimum. High anterior PVI velocity relates to a forward scanning of the environment, while low anterior PVI velocity relates to a quasi-fixation of the visual perception.

**Head Motion and Proximal Visual Intersect Motion during Walking Gait**

The anterior progression of the PVI describes the forward sweep of the field of view and a forward motion of the field of view causes expansion of the optic flow field, which is known to be a strong sensory stimulus. There are kinematic factors that influence the anterior progression of the PVI. These factors influence the progression of the PVI in position, velocity and further derivatives. The three primary contributors to where the lower field of view is resting on the ground are: A) anterior motion of the head, B) vertical motion of the head, and C) pitch of the head (Figures D.3-5).
Figure D.3: Factor A) Anterior head displacement yields an equal transit of the PVI while head pitch and vertical elevation remain constant.

Figure D.4: Factor B) Upward vertical head excursion yields an anterior shift of the PVI while head pitch and anterior velocity remain constant.
Figure D.5: Factor C) Increased head pitch yields an anterior shift of the PVI while head vertical position and anterior velocity remain constant.

The emergence of PVI motion can therefore be ascertained by studying the correlation of these three contributing factors directly. A plot of the head anterior velocity during gait (Factor “A” Figure D.6) shows that the PVIvm occurs near the point of minimum head anterior velocity. The anterior velocity of the head decreases steadily since soon before toe-off of the trailing stance foot. The head speeds up and slows down during the gait cycle due to the propulsion of the body by the feet. After heel strike, it is expected that the head will begin to slow in velocity, as the foot striking the ground induces deceleration called the “braking phase” of gait. However, the trace of anterior head velocity only explains part of the emergence of the PVI anterior translation, and does not explain the systematic slowing of the PVI entirely.
Figure D.6: Factor A) Head anterior velocity across a portion of a walking gait cycle from heel strike to contralateral heel strike. PVIvm timing is shown as vertical ticks, and toe-off is shown as crosses. Toe-off occurs after the anterior velocity maximum, and PVIvm occurs prior to the anterior head velocity minimum.

The second contributing factor to the anterior progression of the PVI, vertical displacement of the head (Factor “B” Figure D.4), shows that the head is translating vertically upward after a local medium following heel strike (~10% of the plot in Figure D.7) during toe-off and at the PVIvm. The head is traveling upward in the vertical direction due to the posting of the body up and over the unipedal stance leg in the swing phase of the contralateral foot. The toe-off and PVIvm occur during the peak in vertical upward head velocity (Figure D.8). However, when the head is traveling upward at its fastest velocity, this will cause the PVI anterior velocity to increase, and move faster across the ground in the anterior direction.
Figure D.7: Factor B) Head vertical displacement across the gait cycle at preferred speed starting from Left Heel Strike to Right Heel Strike. PVIvm timing is shown as vertical ticks, and toe-off is shown as crosses. Toe-off occurs during the rise in vertical head excursion, and PVIvm occurs before maximal vertical head excursion. The head is moving downward slightly early after heel-strike, which is due to the loading of the new stance foot.

Figure D.8: Factor B) Head vertical velocity across the gait cycle at preferred speed starting from left heel-strike to right heel-strike. PVIvm timing is shown as vertical ticks, and toe-off is shown as crosses. Toe-off occurs before maximal positive vertical head velocity, and PVIvm occurs after maximal vertical head velocity. The head is moving downward early after heel strike, which is due to the loading of the new stance foot.
An investigation of the third contributing factor to the anterior progression of the PVI, pitch of the head (Factor “C” Figure D.5), shows that the head is rotating downward after reaching a maximum upward pitch orientation near toe-off (Factor “C” Figure D.9) and is rotating downward during the PVIvm before reaching a local maximum downward pitch during swing at ~60%. The toe-off occurs while the head is rotating at quasi-static near zero velocity and the PVIvm occurs during the peak in downward pitch velocity (Factor “C” Figure D.10). The head pitch velocity appears very similar in shape to the PVI velocity plot (Figure D.2). This similarity appears to explain the global behavior of the PVI when given the inclusion of the three factors that can propel the PVI across the ground in the anterior direction (Figure D.1).

![Pitch of Frankfurt Plane during walking](image)

**Figure D.9:** Factor C) Pitch of the Frankfurt Plane of the head across a portion of a walking gait cycle from left heel-strike to right heel-strike. PVIvm timing is shown as vertical ticks, and toe-off is shown as crosses. Toe-off occurs near maximal upward head pitch, and PVIvm occurs during the downward pitch of the head.
Figure D.10: Factor C) Pitch angular velocity of the Frankfurt Plane across a portion of a walking gait cycle from heel strike to contralateral heel strike. PVIvm timing is shown as vertical ticks, and toe-off with crosses. Toe-off occurs at the onset of a decrease in head pitch velocity, and PVIvm occurs near the slowest pitch velocity. The angular velocity of the head produces a similar shaped trace as the anterior velocity of the PVI.

**Origin of Head Motion during Walking Gait**

Head motion during gait is known to be a damped kinematic resultant of the kinetic effects of the cyclical striking and pushing-off of the feet during gait (Latt et al., 2008; Menz et al., 2003). The question this paper seeks to answer is why does the PVI velocity drop after toe-off. The motion of the PVI is a result of the three factors described above (Figure D.11), but questions remain whether these motions of the head result from a voluntary muscular action that forms part of the coordinated motions taken during every step, or does this behavior emerge due to a complimentary process of damped accelerations. While muscular tone and
compensatory contraction of muscles of the body may contribute to the stabilization of the head during gait, allowances for the retention of functional aspects of motion of the head may be preserved by not completely damping accelerations from heel-strike in order to gain meaningful sensory information during distinct parts of the gait cycle.

Figure D.11: Cascade of decelerations within the body due to a heel-strike event during walking gait. (1) heel-strike; 2) deceleration of the pelvis; 3) deceleration of the proximal head and allowance of a forward pitch of the head.

The foot decelerates rapidly upon heel strike, which then decelerates segments of the body in progressively distal segments from the foot until decelerations reach the head. This includes the intermediate segment, the pelvis, which commonly is referenced as an indicator of how the body CoM is progressing. After each heel-strike, the pelvis receives a slightly delayed deceleration of its mass. The maximal deceleration of the pelvis occurs at roughly 25% of the time between subsequent heel-strikes. Following the maximal deceleration of the pelvis, toe-off of
the trailing stance foot occurs. This deceleration of segments more distal to the foot progresses up the torso and to the neck and finally the head. Finally, the head reaches its maximal deceleration in anterior head velocity and downward pitch at nearly 40% of the time between opposite heel-strikes. Due to the anterior deceleration of the head, the nearly zero vertical acceleration of the head, and the head being anatomically attached at the proximal neck, the inertia of the head causes a forward pitch of the head. After toe-off of the stance foot, the head deceleration is at a maximum, and only muscular force resists the forward pitch from continuing unabated. Some downward and forward head pitch may be allowed to functionally occur such that the visual system can fixate the field of view on the upcoming step landing area.

Fixating the field of view may be part of a strategy of monitoring the sequential “falls” that make up human gait. It may be from a fixation on or around areas immediately before the body that the motor systems gain information on self-motion relative to the ground.

It is at the point where upward velocity of the head has begun to slow, and the body is preparing for the shortly beginning “fall” of the head from its highest position in the gait cycle (Figure D.7) that may be most important for the comprehension of self-motion relative to the ground. In order to provide an understanding of the approach pattern toward the ground it makes sense that the PVI is occurring during a quasi-stable downward acceleration pattern (Factor “B” Figure D.12) well before downward acceleration of the head (negative) begins to peak.
Figure D.12: Cascade of decelerations within the body due to a heel-strike event during walking gait: 1) heel-strike; 2) anterior deceleration (negative) of the pelvis; 3) head anterior deceleration (negative) (Factor A) and a head forward or downward (negative) pitch (Factor C). Pitch acceleration shown with 1k degrees/second$^2$ scale and standard deviation of 18 steps at preferred speed. Head vertical acceleration is shown as a dotted line (Factor B). PVIvm timing is shown as dashed vertical line, and toe-off as crosses.

As the head moves upward vertically, the PVI travels forward, but a mitigating factor that may cancel out the addition of anterior translation during a PVIvm is that the total head excursion is relatively small in a typical gait cycle. For example, in Figure D.7 the total excursion of the head is less than 6 cm from one heel-strike to a contra-lateral heel-strike.

The velocity minimum of the Proximal Visual Intersect (PVI) during a walking step cycle is tightly correlated with several features of the gait cycle. The motions of the body segments are coordinated such that the PVI anterior velocity
slows to speeds approaching zero. The behavior of the PVI is driven by motion of the head. The orientation of the head rhythmically cycles during the gait cycle. This relationship can be described regarding where and when the field of view is fixating on particular areas on the ground during walking gait.
References


BIBLIOGRAPHY


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