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Changes in the Dynamics of Postural and Locomotor Control as a Result of Varying Task Demands

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CHANGES IN THE DYNAMICS OF POSTURAL AND LOCOMOTOR CONTROL AS A RESULT OF VARYING TASK DEMANDS

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ABSTRACT

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The aim of this study was to examine changes in postural and locomotor control under varying task demands. Three experiments were designed to address the impact that fast walking had on standing posture over time, slow walking had on gait dynamics over time, and the extent to which gait speed interacts with the ability to walk randomly.

For experiment I, the aim was to identify the time course in which postural adaptation occurred while walking at faster than preferred speeds. Postural motion was assessed at specific intervals over a 35-min walking trial. Findings revealed that walking at a faster speed increased the amount, variability, and structure (Approximate Entropy-ApEn) of postural motion compared to baseline assessments. Subsequent trials following baseline assessments revealed a leveling-off for specific center of pressure (COP) variables and decline in path length, although heart rate (HR) and rate of perceived exertion (RPE) increased over the entire walking trial.

In experiment II, the aim was to examine changes in stride-to-stride variability over time while walking at slower than preferred speeds. The results revealed an increased stride-to-stride variability and signal regularity (lower ApEn) during walking at 80% preferred walking speed (PWS) compared to PWS. After 10-15 mins a decrease stride-to-stride variability and increase in signal irregularity was seen. Changes leveled-off for the remainder of the session.

Experiment III was designed to examine the effect that intentionally increasing variability (random) had on gait dynamics. Participants were asked to vary their gait while walking on a treadmill at three different speeds. The results revealed gait speed was a significant factor in the amount of variability (CV, range), with higher levels

produced during the slower speed than at PWS and the faster speed. Higher levels of complexity (higher SampEn) were seen in stride time and knee joint motion during the random condition irrespective of gait speed.

Overall, young adults are able to walk at speeds faster or slower than preferred as well as increase gait variability when instructed. These changes in postural and locomotor dynamics reveal that a healthy motor control system can quickly adapt to the task demands imposed upon it.

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CHAPTER I

INTRODUCTION

Standing posture and locomotion are functional tasks in human movement that involve coordination of multiple body segments and sensory input. Variability in human movement was at one time thought of as unwanted "noise"; however, it has recently been identified as an inherent feature of motor control and has shown to be ubiquitous in all complex systems (Newell & Corcos, 1993; Newell, Deutsch, Sosnoff, & Mayer-Dress, 2006). There are multiple levels of control in the healthy motor control system to produce coordinated movement. As a consequence, there are an infinite number of ways in which an individual may solve a movement problem providing order and structure to the seemingly random movements. This ability to coordinate the redundant biomechanical degrees of freedom is a fundamental characteristic of motor control (Bernstein, 1967). While it is virtually impossible to perform the same movement using the same pattern each and every time, too little variability can suggest a system that is unable to adapt to changes in the environment or task (Newell & Slifkin, 1998). Similarly, too much variability in the motor output can also signify problems with control and/or indicate unskilled performance (Kelso, 1993, 1995). Optimal movement has been shown to be inherently stable with the capacity to exhibit high levels of variability amid changes in the task. This ability to generate higher levels of variability is believed to be part of a complex system that affords the flexibility to respond to changes in the environment and/or task (Morrison, Hong, & Newell, 2007).

Assessing movement variability has been especially helpful when trying to identify changes or adaptations in the control of posture and gait. The maintenance of human posture and gait is a complex process that requires the coordination of many systems throughout the body. The sensorimotor system uses information from visual, vestibular, and proprioceptive sources in order to maintain optimal control in posture and locomotion (Simoneau, Ulbrecht, Derr, & Cavanagh, 1995). The constant feedback provided by the sensorimotor system to reduce error during movement is a result of the interactions between the organism (individual), environment, and the goal of the task

(Horak & Nashner, 1986; Newell & Corcos, 1993). Manipulation of one or more of these constraints can change the quality of the motor output revealing the impact of the remaining constraints on patterns of control and coordination (Bernstein, 1967; Horak & Nashner, 1986; Newell & Corcos, 1993). Some of the more common task manipulations to assess standing posture include reducing or eliminating vision (Collins & De Luca, 1995; Day, Steiger, Thompson, & Marsden, 1993; Manchester, Woollacott, Zederbauer-Hylton, & Marin, 1989; Nardone, Tarantola, Galante, & Schieppati, 1998; Nardone, Tarantola, Giordano, & Schieppati, 1997; Simoneau, et al., 1995; Van Emmerik, Remelius, Johnson, Chung, & Kent-Braun, 2010; Woollacott, Debu, & Mowatt, 1987), the use of compliant or perturbing support surfaces under the feet (Aruin, Forrest, & Latash, 1998; Aruin & Latash, 1995; Aruin, Ota, & Latash, 2001; Bunday et al., 2005; Fransson, Gomez, Patel, & Johansson, 2007; Horak & Nashner, 1986; Nashner, 1976), and administering a variety of different fatigue protocols (Biewener, Farley, Roberts, & Temaner, 2004; Bove et al., 2007; Caron, 2003, 2004; Corbeil, Blouin, Begin, Nougier, & Teasdale, 2003; Fox, Mihalik, Blackburn, Battaglini, & Guskiewicz, 2008; Gribble & Hertel, 2004; Nardone, et al., 1998; Nardone, et al., 1997; Simoneau, Begin, & Teasdale, 2006).

Manipulations used to assess locomotor control include changes as a result of injury (Georgoulis, Moraiti, Ristanis, & Stergiou, 2006; Rhea, Wutzke, & Lewek, 2012), the use of split-belt walking with each limb performing at a different speed (Bastian, 2008; Bruijn, Van Impe, Duysens, & Swinnen, 2012), or a different direction (one leg moving forward, one moving backward), (Choi & Bastian, 2007), and non-preferred gait speeds (faster or slower than preferred) (Beauchet et al., 2009; Bruijn, et al., 2012; Chung & Wang, 2010; Dingwell & Marin, 2006; Jordan, Challis, & Newell, 2007; Kang & Dingwell, 2008). Initial responses to most perturbations during both standing and locomotion are usually accompanied by rapid reactions to readjust posture and improve stability. Depending upon the nature of the perturbation, the postural system utilizes feedback and feed forward processes to offset and counteract the destabilizing effect (Fransson, Kristinsdottir, Hafstrom, Magnusson, & Johansson, 2004; Horak & Diener, 1994; Horak, Nashner, & Diener, 1990; Simoneau, et al., 1995). The resultant motor output typically results in greater variability amongst the dependent measures.

Numerous changes in the dynamics of posture and gait are a result of the functional task. These changes can be manifested at a variety of different levels of the system, from the cell to the muscle and individual segments. Physical activity is one such task that is commonly used to challenge postural dynamics. Whether the amount of exertion during the activity is pushed to a level of fatigue to test the boundaries of the system (Corbeil, et al., 2003; Fox, et al., 2008; Nardone, et al., 1997) or enough to moderately perturb the system (Simoneau, et al., 2006) to mimic daily activity, the impact on postural motion has been known to last up to 10 min following cessation of the activity (Nardone, et al., 1998; Nardone, et al., 1997). For example, localized fatigue in the distal muscles of the lower limb (e.g., tibialis anterior, gastrocnemius) has led to an increase in postural sway of both the anterior-posterior (AP) and medio-lateral (ML) axes as well as an increase in sway velocity (Caron, 2003; Corbeil, et al., 2003; Dingwell & Cavanagh, 2001; Vuillerme, Anziani, & Rougier, 2007; Yaggie & McGregor, 2002) resulting in greater demands being placed on the postural control system to assist in regulating balance (Corbeil, et al., 2003). Whole body physical activities such as running and cycling performed at or near maximal heart rate, also induce increases in postural sway (Nardone, et al., 1998; Nardone, et al., 1997; Vuillerme & Hintzy, 2007) as do moderate levels of exertion. Nevertheless, even under conditions of maximal exertion, the changes in center of pressure (COP) motion appear transient, gradually returning to baseline levels over time (Nardone, et al., 1998; Nardone, et al., 1997). The resultant movement pattern is a direct reflection of the task and successive trials have been shown to change the motor output often resulting in lower levels of variability (Horak & Moore, 1993; Horak, et al., 1990). How the postural control system will behave following low-level disturbances encountered everyday in the form of walking can provide insight into the time-related adaptive responses during both bipedal stance and gait.

Walking is arguably the most common form of locomotion for healthy individuals and is viewed as a relatively stable action with some inherent variability in the underlying dynamics (Hausdorff, 2005, 2007). When an individual is walking at his or her preferred pace (~4.4 km/h), the resultant gait pattern has a smaller magnitude of variability (per coefficient of variation, CV, and standard deviation, SD) amongst the spatial and temporal parameters (e.g., stride interval, stride length, step width, etc.), indicating a very repeatable movement pattern with the greatest amount of mechanical energy conservation (Beauchet, et al., 2009; Dingwell & Cavanagh, 2001; Donker, Beek, Wagenaar, & Mulder, 2001; Jordan, et al., 2007). In contrast, walking at a slower or faster than preferred pace requires greater amounts of active control to adjust to the task demands, thus providing a challenge to the neuromuscular system (Jordan, et al., 2007; Jordan & Newell, 2008).

While numerous studies have investigated the impact that gait speed has on the dynamics of posture and gait, the findings of these investigations are primarily a result of the mean and variances (SD and CV) calculated from a small number of strides to assess the impact of a particular intervention or as a result of aging and/or disease (Beauchet, et al., 2009; Dubost et al., 2006; Schniepp et al., 2012; Simoneau, et al., 2006). Increased use of non-linear analyses has provided additional insight into the time-dependent structure of the motor system as it relates to certain characteristics of postural control. The use of Approximate Entropy (ApEn) and Sample Entropy (SampEn) analyses have been useful in furthering the understanding of the complex processes associated with the coordination of posture and gait as a result of different task constraints (Harbourne & Stergiou, 2003; Kavanagh, Morrison, & Barrett, 2006; Morrison, et al., 2007). The use of instrumented treadmills has made it possible to collect repeated strides over a longer duration and not only provides larger datasets for use with non-linear analyses, but also offers the ability to identify non-stationary properties of gait as a result of the varying speeds (Chiu & Wang, 2007; Chung & Wang, 2010; Hausdorff, 2004, 2005). Investigations in the non-stationary properties of posture and gait may further reflect the adaptive capacity and preference for a particular movement pattern within the sensorimotor system (Hausdorff, 2005).

In dynamical systems, the preference for a particular movement pattern is either directly or indirectly related to the goal of the task (Duysens & Van de Crommert, 1998; Horak & Nashner, 1986). A healthy neurologic system will often self-organize to adjust to subtle or overt perturbations during standing or changes in the gait speed while walking to produce a motor output that satisfies some internal or external criterion (Newell, Challis, & Morrison, 2000a; Newell, Deutsch, & Morrison, 2000b). Conversely, the ability to produce highly variable (random) movement has proved to be difficult to achieve (Newell, et al., 2000a; Newell, et al., 2000b). When the goal of the task was to move randomly, whether it be the index finger (s) and different segments of the upper limb (index finger, hand, lower arm, whole arm) in a single plane (Deutsch & Newell, 2004; Newell, et al., 2000a) or the whole body in the form of postural sway (Morrison, et al., 2007), participants had a difficult time accomplishing the task. Both Newell, et al. (2000) and Deutsch & Newell (2004) found that while individuals were able to move more variably, as reflected by a higher CV and SD and greater signal irregularity (higher approximate entropy – ApEn) compared to the preferred movement pattern, they were unable to produce a stochastic output that mimicked that of white noise (highly random). These findings suggest that the structural and functional constraints of the system reduces the ability to intentionally produce random movement in a single joint or multiple joints within a single plane of motion (Deutsch & Newell, 2004; Newell, et al., 2000a). Morrison, et al. (2007) investigated changes in COP dynamics during three postural sway conditions (standing still, preferred sway, and random sway) and changes in the muscles associated with controlling sway (tibialis anterior, soleus). While no differences in the structure (as measured by the changes in ApEn) and modal frequency were seen at the COP level between standing still and preferred sway the random sway condition elicited greater irregularity in the COP dynamics as well as decreased synchrony between AP and ML axes. Conversely, in the muscles used to control sway the inverse response was seen in which the greater irregularity in the COP dynamics resulted in greater regularity in the muscles of the lower leg. This finding suggested that when attempting to move randomly, changes in the complexity of the motor system are limited and organization of the degrees of freedom required that the different structures that produce movement work more independently to perform the task (Morrison, et al., 2007).

While there has been extensive research on the impact of different disturbances on the postural control system the time-dependent changes seen as a result of a consistent low-level perturbation (standing following walking and continuous walking at slow speeds) have not been explored. Furthermore, in an attempt to improve our understanding of the neuromechanical constraints to movement organization, a portion of this study required participants to intentionally make their gait highly variable in an attempt to achieve random motion. Overall, the three experiments that make up this thesis were designed to investigate the impact that simple changes in the tasks (gait speed and variability) have on the adaptive capabilities of the motor control system.

Experiment I: Changes in postural sway as a function of prolonged walking

Statement of the Problem

The primary aim of this study was to identify the time course in which postural adaptation occurs while walking at faster than preferred speeds. The amount of postural sway (COP) was assessed at distinct time periods over the course of 35 min walking trial. Participants were asked to walk on a treadmill at three different speeds; preferred walking speed (PWS), 120% of PWS, and 140% of PWS for a total of 35 min. COP data were collected during bilateral quiet stance under two vision conditions (eyes open and eyes closed) for 60 seconds every 5 min over the course of the walking task. Levels of exertion were measured using a heart rate monitor and a modified (1-10) Borg-scale rate of perceived exertion (RPE).

Research Hypotheses

It was predicted that 1) faster walking would induce initial changes in postural motion followed by a rapid adaptation and return to baseline levels, and 2) the greatest changes in postural sway motion would be seen while walking at the fastest gait speed (140% PWS).

Independent Variables

The independent variables in this study were

Time intervals (7)

Gait speeds (PWS, 120% PWS, 140% PWS)

Visual conditions (eyes open [EO], eyes closed [EC]).

Dependent Variables

The dependent variables in this study were

Measures of exertion:

Mean HR (bpm)

Max HR (bpm)

Percentage of maximal HR (% MHR)

Rate of Perceived Exertion (RPE) – 1-10 scale

Postural sway variables:

COP excursion (mm) – Anterior-posterior (AP) and medio-lateral (ML) axes

Experiment II: Temporal Changes in stride-to-stride variability during slow walking

Statement of the Problem

The primary aim was to explore how variability in the stride interval changed over successive periods of time (e.g., from the first 5 min to the next 5 min, and so on) while walking at speeds slower than preferred. Participants walked on a treadmill at their preferred walking speed (PWS), 80% of PWS, and 90% of PWS for 30 min. The coefficient of variation (CV) and Approximate Entropy (ApEn) of stride time was calculated over each 5-min time block to determine if young adults could reduce stride time variability as a result of adapting to walking at the slower speeds over time.

Research Hypotheses

We hypothesized that 1) walking at a slower than preferred speed would increase the variability and signal regularity of the stride interval initially as an attempt to adjust to the slower pace, and 2) as walking continued at the slower speed a decrease in variability and signal regularity would emerge as a result of adaptation to the task over time.

Independent Variables

The independent variables in this study were

Time periods (6)

Gait speeds (PWS, 90% PWS, 80% PWS)

Dependent Variables

The dependent variables in this study were

Measures of exertion:

Mean HR (bpm)

Rate of Perceived Exertion (RPE) – 1-10 scale

Gait variables:

Time series for stride data

Experiment III: The impact of intentionally increasing stride-to-stride variability during gait

Statement of the Problem

The primary aim of this study was to compare the gait dynamics (stride time and knee joint range of motion) in young adults when asked to produce highly variable (random) and preferred walking patterns at three different gait speeds. Participants walked on a treadmill at three different speeds: preferred walking speed (PWS), 80% PWS, and 120% PWS over 20 min for a total of four 5-min time blocks. The time blocks consisted of two blocks of just walking referred to as control (1^{st} and 4^{th}) and two blocks of intentionally varying their strides (random walking - 2^{nd} and 3^{rd}). Levels of exertion were reported by the participants at 4.5 min of each 5-min walking condition using a modified Borg (1-10) rating of perceived exertion (RPE).

Research Hypotheses

We hypothesized that 1) young adults would be able to intentionally increase the variability of gait dynamics as instructed, irrespective of gait speed, and 2) the ability to increase stride variability would be more difficult during the faster than preferred speed and less difficult at PWS.

Independent Variables

The independent variables in this study were

Movement task (normal and random walking)

Gait speed (PWS, 80% PWS, 120% PWS)

Dependent variables

The dependent variables in this study were

Measures of exertion:

Rate of Perceived Exertion (RPE) – 1-10 scale

Gait variables:

Time series for stride data

Knee kinematics (electro-goniometer measures)

Knee joint flexion/extension angles (°)

Inclusion\Exclusion criteria

Participants consisted of young healthy adults 18-35 years of age. All participants completed a physical activity readiness questionnaire (PAR-Q), and a medical history questionnaire. Subjects were excluded if they reported any neurological and/or cardiovascular disorder, loss of consciousness or concussion within the last year, and/or lower limb musculoskeletal injury/surgery within 1 year. Prior to participation all subjects provided written informed consent. The study was approved by the University Institutional Review Board and all experimental procedures complied with the guidelines.

Operational Definitions:

- Postural control is defined as the ability to control the body's center of mass (COM) over its base of support (BOS) to achieve functional tasks and prevent the body from falling (Winter, 1995).
- Center of Pressure (COP) excursion is defined as the amount of motion (in millimeters) that occurs around the BOS as a person stands on a force plate. This is measured in two planes, anterior-posterior (AP) and medio-lateral (ML).
- Complexity of a system is dependent upon the number of system elements and the functional interactions between them (Vaillancourt & Newell, 2002). The term has evolved from the fields of biology and physics to be applied to the concept of motor control. While the term complexity is loosely defined in the field of human movement much of the literature defines complexity as the unpredictability (irregularity) of sequences of a time series (Morrison, et al., 2007). As an

example, a sine wave is highly predictable and largely dependent upon the previous sequence in the time series revealing low complexity, whereas in a random signal, the properties of the time signal are independent of each other and unpredictable revealing high complexity. (Morrison, et al., 2007; Stergiou, Harbourne, & Cavanaugh, 2006).

- Approximate Entropy (ApEn) is an analysis that computes the conditional probability of the signal by providing a measure of the likelihood that any given data point (n) in the time series that is close for m observations, remains close on the next incremental comparisons ($m + 1$). This is measured by the level of repetition that occurs between m and m+1 vectors within a tolerance range of the standard deviation (r) of a time series. This analysis produces a value between 0 and 2 with values closer to zero indicating higher repeatability of the vectors and a more regular signal. Higher ApEn values represent lower repeatability of the vectors m and $m + 1$ and represent greater irregularity (decreased structure) in the time series. Increases in ApEn have been interpreted as an increase in the signal's time domain complexity (Pincus, 1991).
- Non-stationarity is stated to occur when statistical properties of a time series (COP motion in postural sway, stride time) differ from one segment along the series to the next (Newell, et al., 2006; Terrier & Deriaz, 2012).

Assumptions

- The balance plate used was calibrated and accurately recorded the amount of AP and ML displacement;
- The pressure plate on the treadmill accurately recorded and calculated the specific gait instances to determine the stride time intervals;
- Participants performed the protocols as instructed by the researcher.

Limitations

Although the researchers have tried to reduce the number of limitations in this study, it is impossible to control for everything. One limitation that occurs is the fact that the sample is that of convenience. Participants consist of college students that have volunteered to participate within the university. Another limitation of the study is the use of a treadmill for collecting the data. Although there is good reliability when comparing treadmill and overground locomotion, the treadmill tends to reduce the stride to stride variability due to the constant motion of the belt providing an increase in stability (Dingwell, Cusumano, Cavanagh, & Sternad, 2001).

CHAPTER II

REVIEW OF LITERATURE

The following review of literature focuses on changes in postural control dynamics as it relates to the constraints imposed upon it. This is organized by first identifying the sensory input, the strategies adopted to control standing posture, and the control of locomotion. The second section discusses the factors that impact posture and gait from a constraints perspective providing the theoretical framework for the projects involved in this dissertation. The third section will discuss changes as a result of adaptation and intention. Finally, the discussion revolves around the data analysis used for assessing posture and gait.

Due to the influence of individual, environmental, and task constraints, a preferred postural coordination pattern emerges as a result of the self organizing principles of a dynamical system. The time course of changes in the variability measures of posture and locomotion have not been assessed, nor have the changes that occur in gait when the goal of the task is to increase gait cycle variability (random). The ability to control movement dynamics of posture and gait by adapting to the task or by intentionally increasing variability will provide further insight into the coordination of a highly complex system.

Balance and postural control

Postural control during standing has been defined as the ability for individuals to maintain their center of mass (COM) within the limits of the center of pressure (COP) without having to change the base of support (i.e., take a step or lift feet from surface) (Winter, 1995). In the healthy sensorimotor system the control of posture is highly complex and reliant upon the interaction between the motor cortex, cerebellum, and basal ganglia, along with feedback provided from visual, vestibular, and somatosensory input (Chow, Lauk, & Collins, 1999; Hausdorff, 2007; Prieto, Myklebust, Hoffmann, Lovett, & Myklebust, 1996). The interactions between the sensory input enables human beings to stand upright for long periods of time, reach for an object while standing in one place, and provide the basis for locomotion. While the same structures interact to coordinate standing and locomotion, there are slight differences in function. For clarification purposes, this review will discuss sensorimotor information as it relates primarily to the control of posture during stance followed by an explanation of the most common postural control strategies adopted to maintain balance. The control of gait will be discussed further in this section to establish the different functions of the same sensory input (vision, somatosensory, vestibular) to accommodate the changes in balance with each step.

Sensory input

Sensory input from the visual, somatosensory, and vestibular systems act together to provide stable conditions for balance and locomotion. Manipulating one or more of these sensory systems (such as eyes open and eyes closed for vision) has provided substantial information regarding individual contributions and the effect of their interactions in the control of posture. Standing posture is accomplished by exposing the individual to different task constraints (i.e., change from firm to foam surface) or comparing healthy individuals to those with pathology (such as diabetic neuropathy and visual deficits) to determine the extent of input each system has on balance (Horak, Nashner, Diener, 1990).

Vision provides the strongest amount of sensory information for postural control. Sighted persons rely heavily on vision to provide feedback to the central nervous system (CNS) to maintain stability during standing or walking (Simoneau, et al., 1995). The visual system allows individuals to perceive their position in space, velocity, and acceleration relative to another object in the field of vision in response to vestibular receptors, which send information about head velocity to the vestibular nuclei that is then projected onto the oculomotor nuclei (Johanssen, 1991; Latash, 1998). When visual information is removed by closing the eyes during quiet stance, all indices of postural stability are worsened for both young and older healthy adults. This is reflected by an increase in COP postural sway excursion, velocity, and frequency (Simoneau, et al., 1995). This heavy preference on vision facilitates a greater reliance on feedback control compared to feedforward control in maintaining balance. Distorted vision is considered a primary determinant for risk of falls in the elderly during standing and walking (Lord &

Menz, 2000), coupled with reduced somatosensory information which enhances this risk. Standing on a compliant surface with eyes closed reduces the ability to control postural sway, indicating a strong link between vision and somatosensory information (Simoneau, et al., 1995). Simoneau et al. (1995) investigated the role of somatosensory input on postural control and found that the link between the two systems was so strong that when both were eliminated during quiet stance, postural motion increased 150% in healthy young adults (Lord & Menz, 2000; Simoneau, et al., 1995; Vuillerme, Burdet, Isableu, & Demetz, 2006a).

Somatosensory information is relayed through sensory receptors that respond to specific stimuli providing the body with information related to the environment. These receptors include proprioceptors identifying where the body is in space in relation to the environment, mechanoreceptors that react to mechanical stress/strain, and nociceptors, which relay damage to body tissues-leading to pain. The major role of the combined input from these sensory receptors, particularly the proprioceptors and mechanoreceptors, is to relay information about disruptions in the structures. The muscle spindles provide information related to length and velocity of a muscle during postural disturbances, and the golgi tendon organs are sensitive to muscular tension (Kandel, Schwartz, & Jessell, 2000). When a postural muscle like the gastrocnemius is lengthened quickly in response to a perturbation (backwards tilt of the surface), the muscle spindles respond by shortening the involved muscle, thus producing an increase in sway to maintain upright balance. Somatosensory and vestibular input appears to be important in the choice of which postural strategies to use to maintain equilibrium when exposed to a destabilizing event (Horak, et al., 1990; Simoneau, et al., 1995).

Vestibular input provides information on our position in space and identifies changes in velocity and acceleration of our head relative to the direction of the field of gravity (Johanssen, 1991; Latash, 1998). Its origin is seen in the labyrinthine receptors that flow to the vestibular nuclei of the brain stem. The medial and superior vestibular nuclei have an important role in oculo-motor control through input from the vestibulo-ocular reflexes that control eye motion in relation to head motion (Kandel, et al., 2000). The vestibular system is responsible for a large number of postural reflexes that allow for upright standing in humans and projects to the cervical, upper thoracic, and lower lumbar levels.

Individuals with vestibular damage tend to sway excessively or fall when the surface is perturbed compared to those with an intact system (Horak, Shupert, Dietz, & Horstmann, 1994). However, an individual with profound loss of bilateral vestibular function still has near normal function when the other systems are intact. This sensory integration provides for a flexibly organized system that can detect and respond quickly to destabilizing perturbations to maintain postural control in a variety of conditions (Horak & Diener, 1994).

Postural sway strategies

Postural sway is defined as the minute movements that occur in the anterior-posterior (AP) and medio-lateral (ML) direction as the COM oscillates within the boundaries of the COP. Even while standing still, there is a high degree of motion that occurs to establish equilibrium within the system (Newell, Slobounov, Slobounova, & Molenaar, 1997b). As a result of the constant search for equilibrium to adjust the COM over the smaller COP, many researchers have referred to movement of the body to control the amount of postural sway as an inverted pendulum. This theory indicates that the COP is directly correlated with the horizontal acceleration of the COM in either the AP or the ML directions (Winter, Patla, Ishac, & Gage, 2003; Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). In this theoretical framework, the COM is the passively controlled variable and the COP is actively controlled, providing either a stabilizing or destabilizing condition dependent upon the position in which the COP is in relation to the COM. If the COP is ahead of the COM, the latter is accelerated posteriorly to maintain balance; if COP is to the left or right of COM, then it accelerates it according to where the imbalance is (i.e., to the left or the right). This type of strategy is based upon the inverted pendulum assumption that the ankle joint moves as a single stiff segment regulating COP motion in both AP and ML directions (Winter, 1995).

The muscles most active in maintaining COP in the AP direction are the ankle plantarflexors (gastrocnemius and soleus) and dorsiflexors (anterior tibialis and extensor digitorum). For the ML direction the hip abductors/adductors are active in healthy adults, allowing a very synchronous activation of sensory information to provide the control of balance (Winter, et al., 2003; Winter, Prince, Frank, Powell, & Zabjek, 1996). The maintenance of upright standing is produced by small continuous movements that are facilitated by either feedback reflexes from the CNS, elastic properties from the system, or open-loop activity of motor units (Morrison, et al., 2007; Winter, et al., 2003; Winter, et al., 1996). Through the integration of the sensory input received and the adjustments that the CNS makes through closed-loop/open-looped control strategies, the COP is able to reach the limits of stability within individualized parameters. When the COM goes too far beyond the limits of the base of support, the individual may need to use another strategy to maintain balance, such as the use of the hips or to take a step (Collins & De Luca, 1993).

The hip sway strategy has been shown to be utilized when an individual feels threatened in his or her ability to maintain posture (Horak & Nashner, 1986). This strategy is characterized by movement centered on the hip joint to counteract horizontal sheer force against the support surface when responding to larger and faster changes in the COM. An example of this is the postural response experienced when standing on a stationary bus and it unexpectedly begins moving (Gatev, Thomas, Kepple, & Hallett, 1999; Nashner, Shumway-Cook, & Marin, 1983). The elderly tend to present with this type of strategy regardless of the absence or presence of vision due to compromised mechanoreceptors and proprioception resulting from the aging process. This strategy has been associated with a greater freezing of the degrees of freedom, leading to an increased risk of falls in this population (Accornero, Capozza, Rinalduzzi, & Manfredi, 1997). These strategies have been widely accepted in the literature; however, it is important to note that while rotation of the ankle strategy takes place primarily about the ankle joint, it is not exclusive as there is some involvement of motion at the hip joint (Ouiller, Marin, Stoffregen, Boostma, & Bardy, 2006).

Locomotor control

In dynamic situations such as locomotion, postural control becomes a bit more challenging. Locomotion involves a sequence of forward falls, thus requiring the central nervous system (CNS) to adjust segmental posture to prevent the individual from becoming unbalanced (Winter, 1995). The process of walking consists of a large number of neuromuscular responses in a coordinated pattern that is controlled not individually, but rather more synergistically as a unit. The balance and coordination required to perform this task can be best witnessed in an infant beginning the process of learning to walk exhibiting a larger base of support necessary in the early stages to maintain stability and quickly narrowing as coordination improves (Thelen, Kelso, & Foley, 1987). As in standing posture, the COM and COP interact during locomotion to provide stability within the frontal plane of the whole body. During steady state walking the COM must stay within the medial border of the foot to maintain balance (Winter, 1995). The complex coordination required for stepping is divided into two distinct phases of gait, swing and stance. The swing phase of gait is the stage in which knee flexion occurs and the foot is not in contact with the ground. The stance phase of gait occurs when the foot is in contact with the ground and extension occurs at both the knee, hip, and ankle to propel the body forward (Winter, 1983). This coordination is believed to be controlled by motor circuits that lie in the spinal cord and the brainstem and further refined by the higher levels of the brain. Animal studies have demonstrated this automatic response to forward locomotion in the absence of cortical involvement. Decerebrated cats and dogs are able to continue a rhythmic motion in the hind legs, indicating that walking is automatic and can be performed with little sensory feedback from the cerebral hemispheres. This movement pattern, along with automatic breathing and chewing, is thought to be controlled by neural circuitry termed central pattern generators (CPG) (Duysens & Van de Crommert, 1998). CPG circuits are predetermined neural activity responsible for producing timed sequences of repetitive continuous voluntary movement involving synergist muscles. Leg and ankle control in the left and right legs during walking is one example of such movement. Although these appear to be fairly well-developed in lower species animals, specific neurons comprising a CPG in spinal cord segments have not yet been identified in humans (Duysens & Van de Crommert, 1998; Kandel, et al., 2000).

During locomotion, sensory input is continuously modulated to provide information on the surrounding environment while moving and maintaining balance. Proprioceptive input is constantly relayed to the somatosensory and motor cortex to provide information on the interaction between the body and the environment. One of the primary functions of proprioceptive information is to react to unexpected changes to the postural system and provide compensatory responses to control the COM and joint motion. As a result of the constant proprioceptive input, the locomotor system is able to adapt to changes in the environment (Dietz, 2002).

Supraspinal input from the cerebellum and the basal ganglia are believed to have a large impact on gait. Although not completely understood in their relation to gait, research investigating the impact of damage or disease to these regions has given us some understanding of their role in motor planning and regulation (Kandel, et al., 2000). The cerebellum is responsible for planning the precise movements necessary to coordinate the intra- and inter-limb motion and modulate the reflex patterns necessary to navigate through the environment (Morton & Bastian, 2003). The basal ganglia appear to be responsible for coordination of gait stride and rhythm that become abnormal with dysfunction such in as Parkinson's and Huntington's diseases, often characterized by the breakdown of long-range correlations in stride-to-stride variability that is seen in healthy populations (Hausdorff et al., 1996). Coordination of the redundant degrees of freedom within the body to produce goal-directed movement is established through a relatively stable system that is able to adapt to a high degree of variability (Hausdorff, Yogev, Springer, Simon, & Giladi, 2005; Morrison, et al., 2007; Newell, et al., 2000b).

Factors influencing coordination of posture and locomotion

Performing identical movement patterns is virtually impossible, although when performed under similar conditions, the movement outcome can be the same as a result of the goal of the task (Kelso, 1993; Newell & Slifkin, 1998). For example, if the same person attempts to hammer a nail into a board, it is likely that with each swing of the hammer he or she will hit the head of the nail; however; the path in which the hammer travels will vary slightly from one swing to the next. According to Bernstein (1967), movement variability is a result of the continual process of controlling and/or releasing the various degrees of freedom to produce dynamic movement. Changes in movement coordination are influenced by the extrinsic and/or intrinsic factors that inhibit or allow motion to occur referred to as constraints. The three categories that interrelate to influence movement output include the organism (or individual), the environment, and the task (Newell, 1986; Newell & Corcos, 1993). Postural behavior during standing and locomotion is influenced by the various constraints placed upon it and plays a functional role in the ability to adapt to various perturbations (Ouiller, et al., 2006).

Task constraints have a great impact on movement dynamics and are frequently manipulated in research settings. For standing posture, changing the task may be accomplished by performing quiet stance with vision (eyes open) or without vision (eyes closed) (Caron, 2004; Day, et al., 1993; Manchester, et al., 1989; Simoneau, et al., 1995), changing the surface on which one is standing to provide a perturbed surface versus a firm stable surface (Horak, Diener, & Nashner, 1989; Horak & Nashner, 1986; Nashner, 1976), standing on one leg (Fox, et al., 2008), and increasing postural sway during stance (Morrison, et al., 2007). In locomotion, this may be accomplished by having an individual perform a cognitive function while walking at preferred walking speed (PWS) (Dubost, et al., 2006; Ijmker & Lamoth, 2012), walking on a split-belt treadmill with one leg moving forward the other backward (Choi & Bastian, 2007), or walking at speeds that are faster or slower than preferred (Beauchet, et al., 2009; Bruijn, van Dieen, Meijer, & Beek, 2009; Chung & Wang, 2010; Jordan, et al., 2007).

The effect of speed on gait dynamics has been well-documented (Brach, Berlin, VanSwearingen, Newman, & Studenski, 2005; Brisswalter, Fougeron, & Legros, 1998; Danion, Varraine, Bonnard, & Pailhous, 2003; Jordan, et al., 2007; Jordan & Newell, 2008; Kang & Dingwell, 2008). The changes that develop have provided important information about the locomotor system and its ability to adapt to different task demands. All individuals have a preferred walking speed (PWS) that is performed with minimal energy expended to produce that speed (Holt, Jeng, & Ratcliffe, 1995) and a preferred combination of stride/step length and stride/step frequency (Danion, et al., 2003; Latt, Menz, Fung, & Lord, 2008; Murray, Drought, & Kory, 1964) . While most people are quite capable of walking at speeds other than preferred, the influence of the non-preferred speed changes the resultant movement pattern. When either faster or slower than PWS, variability of stride time, structure of the signal, and long-range correlations can change significantly in both young and older adults (Hausdorff, et al., 1996; Jordan, et al., 2007; Kang & Dingwell, 2008). Typically, the healthy neurologic system adapts to the slower gait speed by increasing stride times and stride-to-stride variability and decreases in stride length. Adapting to a faster gait speed elicits a decrease in stride time and increases in stride length, and stride-to-stride variability when compared to PWS, thus indicating that changes in gait speed impacts movement coordination. Long-range correlations are assessed using detrended fluctuation analysis (DFA) to identify self-similar fluctuations that occur in a time series of stride interval data. While walking at PWS even over a long period of time (1 hour or more), there is a high level of self-similarity in the stride-to-stride fluctuations that becomes weaker at non-preferred speeds (Hausdorff, et al., 1996). When an individual is forced to walk faster than their preferred pace, options available to solve the coordination problems become limited and their gait becomes much more constrained (Jordan, et al., 2007). Conversely, walking at speeds slower than preferred elicits even greater stride-to-stride variability than seen in the faster speed in both young and older adults (Beauchet, et al., 2009; Jordan, et al., 2007; Winter, 1983). Increased variability at the slower speeds are due to the greater motor control strategies needed to keep the COM over the support limb and maintain balance (Hausdorff, et al., 1996; Jordan, Challis, & Newell, 2006; Jordan, et al., 2007). Non-preferred speeds impact the cyclic pattern of gait and walking at faster speeds has been used to induce moderate fatigue to the postural system (Simoneau, et al., 2006).

Individual (or organismic) constraints refer to the unique structural and functional characteristics of the organism which strongly influence behavior. Structural constraints produce changes seen in the motor output as a result of aging(Hausdorff et al., 2001; Newell, Sloibounov, Slobounova, & Molenaar, 1997a; Newell, van Emmerik, Lee, & Sprague, 1993), injury (Docherty, Valovich McLeod, & Schultz, 2006; Georgoulis, et al., 2006; Rhea, et al., 2012), various diseases affecting our ability to derive visual or somatosensory information properly (Dingwell & Cavanagh, 2001; Hausdorff, Cudkowicz, Firtion, Wei, & Goldberger, 1998a; Hausdorff et al., 1997b; Owings & Grabiner, 2004b; Simoneau, et al., 1995; Vaillancourt & Newell, 2002), and/or whole body fatigue in the form of walking, running, and cycling (Nardone, et al., 1998; Nardone, et al., 1997; Simoneau, et al., 2006; Vuillerme & Hintzy, 2007). Functional constraints refer to the way that a system functions to achieve the goal of a given task. The interactions of the structural and functional constraints impact the motor control strategies used to solve movement problems that can lead to adaptive behaviors. These persistent movement patterns are particularly well established in posture and locomotor tasks (Ouiller, et al., 2006). To coordinate movement the biomechanical, physiological, and neurologic structure of the individual work together to self-organize and produce optimal movement output for that person. When an individual moves beyond the boundaries of stability a greater freezing of the degrees of freedom occurs in an attempt to modulate the postural system in an unstable position increasing the functional constraints within the system (Newell, et al., 1993). Physical activity has been reported to immediately impact postural stability and under maximal exertion up to 10 min after activity has ended (Nardone, et al., 1998). It is partially due to changes in the somatosensory feedback and increased heart rate that result in deficits during quiet stance following exercise. Several studies have identified the affects of muscle fatigue on quiet stance immediately following some fatiguing event (Gribble & Hertel, 2004; Pline, Madigan, & Nussbaum, 2006; Simoneau, et al., 2006; Vuillerme, et al., 2007; Vuillerme & Pinsault, 2007; Vuillerme, Pinsault, & Vaillant, 2005; Yaggie & McGregor, 2002). Typical responses to the postural disturbances produced by physical exertion are characterized by increases in the amount and/or frequency of COP excursion as an attempt to stabilize the body. However, even under maximal exertion (measured by stationary cycling or treadmill running above the anaerobic threshold), the resultant motor output has been shown to be transient lasting only about 10 min following cessation of the activity (Bove, et al., 2007; Nardone, et al., 1998; Nardone, et al., 1997). Nonetheless, the time course of changes in postural control as a result of a common functional task such as fast walking is still unknown.

Environmental constraints are typically those things that are external to the organism (individual) and are relatively time independent (Newell, 1986). These can include physical barriers such as gravity, natural light, ambient temperature and those things that are not manifestations of the task. Often, these constraints consist of performing a particular task in a different environment such as walking on a smooth sidewalk in which there are few to no obstacles compared to walking on a more natural surface with varying obstacles (i.e., tree roots, branches, and embedded rocks). The ability to adapt to perturbations characterized as changes in the whole body coordination results from not just a neural, biomechanical, or muscular control but from the interaction of all of these structures to shape the postural behavior in relation to the constraints placed upon it (Ouiller, et al., 2006).

Postural coordination as a result of adaptation and intention

How the postural control system modulates the levels of variability in order to produce coordinated movement has been the subject of many investigations over the past couple of decades. Postural and locomotor adaptations and other responses to task demands have been researched.

Postural adaptations to task demands

Changes in postural control as a result of task demands have been the subject of many studies in a variety of populations such as young adults (Collins & De Luca, 1995; Fox, et al., 2008; Gribble & Hertel, 2004; Horak & Nashner, 1986; Morrison, et al., 2007; Newell, et al., 1997b; Simoneau, et al., 1995; Vuillerme, Burdet, Isableu, & Demetz, 2006b; Vuillerme & Hintzy, 2007; Vuillerme, Nougier, & Prieur, 2001; Vuillerme, et al., 2005), older adults (Accornero, et al., 1997; Doumas & Krampe, 2010; Lord & Menz, 2000), individuals with lower extremity injury (Docherty, et al., 2006; Lysholm, Ledin, Ödkvist, & Good, 1998), and persons with neurological disorders (Horak, et al., 1990; Schieppati, Hugon, Grasso, Nardone, & Galante, 1994; Schieppati & Nardone, 1991). The ability to adapt to functional task constraints is an integral part of the postural system. The early work of Nashner (1976) investigated the different reflexive postural responses as a result of changes in the support surface. In his classic work on the impact of a postural "set," it was noted that functional stretch reflexes (FSR) were activated in response to particular perturbances at specific times and, dependent upon the type of reflex initiated, they could be altered to be useful, be of no use, or inappropriate. For instance, when neurologically healthy adults were subjected to a series of postural disturbances at the base of support, the initial response of the FSR was inappropriate producing greater and more destabilizing levels of sway. However, following successive trials, the FSRs responded more quickly and appropriately to the perturbation resulting in lower levels of sway due to the task performed and previous exposure to the perturbances (Nashner, 1976). This ability to adapt to postural disturbances as a function of successive trials was investigated further, and it was found that the response to the postural disturbances were dependent upon magnitude and direction as well as whether it was anticipated or not. Horak et al. (1986) tested the theory that a central motor program (referred to as "central set") was responsible for producing stereotypical responses to successive perturbations. The findings revealed that when a particular magnitude of perturbation was greater than expected, the central set would overestimate the response necessary to maintain postural control. At the same time, when the magnitude of the perturbances were smaller than expected, the response was underestimated producing an increase in sway to maintain control. Subsequent trials of the same magnitude or direction would gradually reduce the amount and/or sway velocity. This response indicated the occurrence of adaptive mechanisms as a result of repeated exposure to a particular disturbance (Horak & Nashner, 1986). Postural disturbances can occur as a consequence of either unexpected external forces (e.g., the sudden movement of the support surface) or of an individual's voluntary movement (e.g., fast walking) (Horak & Macpherson, 2011). Previous work by Simoneau and colleagues (2006) reported that fatigue induced by fast walking resulted in an immediate increase in postural sway. However, these initial increases in COP motion gradually declined over three fatigue blocks, indicating that subjects were able to adapt their balance control to accommodate the fatigue effects.

Locomotor adaptation to task demands

Locomotor control is a complex movement that is performed with very little conscious thought given to the process (Hausdorff, et al., 2005). When a person is walking at his or her preferred pace the pattern is very stable with little variation within the gait cycle (Dingwell, et al., 2001; Hausdorff, 2004). This stable pattern appears directly related to the interactions between the motor circuits in the spinal cord and brainstem with further refinement from the higher level structures to provide the correct stride length and timing for optimal efficiency (Danion, et al., 2003; Holt, et al., 1995; Terrier & Deriaz, 2012).

However, walking for a prolonged period of time (Yoshino, Motoshige, Araki, & Matsuoka, 2004) or at speeds that are faster or slower than preferred (Beauchet, et al., 2009; Bruijn, et al., 2009; Chiu & Wang, 2007; Chung & Wang, 2010; Helbostad &

Moe-Nilssen, 2003; Jordan, et al., 2007; Jordan & Newell, 2008; Terrier, 2012) can disrupt the rhythmic nature of the gait cycle, and the changes are reflected in the various gait parameters. The spatio-temporal parameters that are most influenced during changes in gait speed include stride/step length, stride/step time, and step width. Stride length is the distance covered from heel contact of one limb to heel contact of the same limb; step length is the distance traveled from heel contact of one limb to successive heel contact of the opposite limb; and step width is defined as the distance between the outer most borders of two consecutive footprints. Stride time (also referred to as stride interval) is the amount of time between heel contact of one limb and subsequent heel contact of the same limb, while step time is the amount of time between heel contact of one limb and subsequent heel contact of the opposite limb for 2 consecutive steps (Brach, et al., 2005; Hausdorff, et al., 1996; Jordan, et al., 2007).

Much of the research on gait has focused primarily on the mean values of a finite number of steps/strides to identify any changes that seen as a result of a particular intervention, aging, and/or disease (Beauchet, et al., 2009; Danion, et al., 2003; Frenkel-Toledo et al., 2005; Helbostad & Moe-Nilssen, 2003). While using the mean values related to changes within the gait cycle can be useful in group comparisons, doing so assumes stationarity of the gait cycle pattern. Previous studies have investigated the stride-to-stride fluctuations and have found that indeed gait is non-stationary and that subtle changes are seen over time (Dingwell & Cavanagh, 2001; Hausdorff, 2004; Hausdorff, Zemany, Peng, & Goldberger, 1999; Jordan, et al., 2006, 2007; Kavanagh, et al., 2006). A movement pattern is considered to be non-stationary when its statistical properties differ as a function of time. Non-stationarity of gait parameters, specifically stride time, has been the focus of recent investigations since changes that occur as a result of different tasks are considered to be part of an adaptive process within the motor system (Fairley, Sejdic, & Chau, 2010; Hausdorff, 2007; Newell, et al., 1997a; Sejdic, Fu, Pak, Fairley, & Chau, 2012; Terrier, 2012; Terrier & Deriaz, 2012; Wall & Charteris, 1980). Early investigations on adaptive locomotor behavior were conducted as a means to identify how long an individual needed to walk on the treadmill to reflect gait patterns seen during overground walking (Wall & Charteris, 1980). In a study by Wall & Charteris (1980), subjects naïve to treadmill use were divided into three groups (slow, preferred, fast gait speeds) and instructed to walk for 10 min. The initial locomotor response was rapid destabilization characterized as a tripping/regaining balance response seen within the first 10 seconds. As the task (walking on the motorized treadmill) became more familiar and habituation occurred, the mean stride time intervals became more regular and more similar to those seen during overground walking. A steady state was reached within 10 min in which the gait pattern was more consistent; however, the time to adapt to the slower speed was greater than at the preferred and faster speeds (Wall & Charteris, 1980). In a more recent study of adaptive locomotor behavior, similar patterns of habituation were seen in a pediatric population naïve to treadmill use under three different walking conditions (Fairley, et al., 2010), thus providing further indication that gait is non-stationary and that adaptation to task demands occur when given sufficient time. These investigations into time-dependent changes of gait dynamics utilized the mean values of different gait determinants change to indicate habituation to the various tasks (Fairley, et al., 2010; Wall & Charteris, 1980).

Human locomotion has been shown to be very rhythmic with some stride-to-stride fluctuations occurring even amongst a preferred walking speed (Dingwell & Cusumano, 2000; Terrier & Deriaz, 2012; Winter, 1983). Examining the stride-to-stride variability of the gait cycle has provided further knowledge regarding the physical and cognitive capabilities to dynamically adapt gait and respond to changes in the environment (Hausdorff, 2005). Much of the research in gait variability suggests that increased variability indicates decreased stability, thus leading to a risk of falls in the elderly (Dingwell & Marin, 2006; Hausdorff, 2004; Hausdorff, Edelberg, Mitchell, Goldberger, & Wei, 1997a; Kang & Dingwell, 2006). While this has been shown to be a factor for older adults, in healthy young adults, variability occurs more frequently when walking at a speed that is slower than preferred, but is not correlated with instability (Dingwell & Cavanagh, 2001). The adaptive response to the non-preferred pace is the result of the need to provide greater active control of the motor output to maintain the non-preferred pace (Jordan, et al., 2007; Jordan & Newell, 2008). For the slower speeds, the increased ML excursions of the center of mass (Orendurff et al., 2004) may account for the need for greater control while walking at the faster speeds may place greater physiological and biomechanical constraints on the system producing greater variability (Jordan, et al., 2007). Changes in the stride-to-stride variability as a response to changes in gait speed are useful in determining the underlying neuromuscular control processes in locomotion (Dingwell & Cusumano, 2000).

Intent to increase variability

Variability is a natural occurrence in all biological systems and part of the intrinsic dynamics of the motor system (Neuringer, Kornell, & Olufs, 2001; Yamada, 1995). Movement variability has been studied largely as a by-product of motor output as a consequence of the interaction of the various constraints (Newell & Corcos, 1993). However, the intent to move in a variable manner as the goal of the task has just recently been the focus of investigation. It has been well-established that it is difficult to generate truly random sequences whether it is numbers, letters, calling out heads or tails, or other tasks even when instructed and sufficiently motivated to do so (Bains, 2008; Figurska, Stanczyk, & Kulesza, 2008; Rosenberg, Weber, Crocq, Duval, & Macher, 1990; Wagenaar, 1972). Recently, a few studies have investigated changes in the motor output when the intent to move randomly was the goal of the task (Newell, Challis, Morrison, 2000; Deutsch & Newell, 2004; Morrison, Hong & Newell, 2007). Early investigations have identified that intentional random movement in both single and multi-directional joints of the upper limb is difficult to achieve (Deutsch & Newell, 2004; Newell, et al., 2000a). Randomly produced oscillations of the index finger(s) and multiple segments of the upper limb (finger, wrist, elbow, shoulder, and entire arm) in a single plane (sagittal) produced a smaller amplitude of motion, greater irregularity of the signal (higher ApEn), and lower dimensional properties compared to preferred oscillations (Newell, et al., 2000a). Using the same upper limb joint segments, a subsequent experiment investigated the impact that segmental coupling had on the ability to move randomly. This time the conditions were to move randomly as a unit (stiff joints from shoulder down) or move the entire upper limb, allowing each segment to move independent of the others. The results exhibited inter-segmental and directional influence on the ability to produce random-like motion. When each segment was able to freely move in conjunction with the other, an increase in movement complexity was observed compared to the rigid motion of the arm as a unit. Additionally, the distal segments were able to produce more randomness than the proximal segments during the free moving condition. However, the findings supported the previous ones by Newell et al. (2000), indicating that only a modest level of randomness was output as a result of the task (Deutsch & Newell, 2004). It was concluded that the boundaries that constrained the motion during the random condition were a consequence of the tightly bound single degree of freedom within a single planar direction, therefore making the release of the biomechanical degrees of freedom more limited than originally believed (Deutsch & Newell, 2004; Newell, et al., 2000a; Newell, et al., 2000b).

Further evidence as to the difficulty of achieving random movement was observed in individuals during a postural sway task (Morrison, et al., 2007). Participants were asked to stand still, sway at a preferred frequency (both AP and ML directions), and sway randomly on a force plate for 2 min each. As in previous studies, a significantly greater degree of irregularity in COP output during the random sway condition existed compared to standing still and preferred sway. However, in order to achieve this, a decoupling of the AP and ML motion was seen. Furthermore, an inverse relationship between the muscles activated to maintain postural control (soleus and tibialis anterior) and the COP dynamics transpired, requiring greater predictability of the muscles to produce greater irregularity of the COP dynamics. The implication was that in order to move randomly, a complexity trade-off must occur (Morrison, et al., 2007). The intent to produce a highly variable (random) gait pattern while walking at a constant speed has not been assessed at this time. The cognitive and dynamical constraints imposed upon the system as a result of the imposed speed of the treadmill and the task (increase variability) will provide greater information about the control mechanisms of a complex system.

Analyses used to assess postural and locomotor dynamics

Time domain

Traditional linear measures focused on the central tendency (means, standard deviations, or SD, minimum, and maximum) have been used extensively in the literature (Beauchet, et al., 2009; Caron, 2003; Corbeil, et al., 2003; Fox, et al., 2008; Grabiner, Biswas, & Grabiner, 2001; Gribble & Hertel, 2004; Hausdorff, 2005; Owings & Grabiner, 2004a, 2004b; Terrier, 2012) and were also used in this project. The group means, SD, minimum and maximum values are useful in providing a snapshot on changes that occur as a result of the intrinsic and extrinsic factors being placed upon the system. The use of these descriptive analyses can be important for understanding and determining the quality of the data (James, 2004). The mean values provide an overall discrete value that relays the average of the total data set. The determination of the coefficient of variation (CV), for example, uses the mean value and the SD and is calculated as the sum of the SD divided by the mean and multiplied by 100 [(SD/mean)*100] as a way to identify a percentage of difference between one factor and another. Similarly, the range is calculated by subtracting the minimum value from the maximum value (max-min) and provides another measure of the amount of variability that is seen in the variables that assess posture and gait.

Frequency domain

Frequency analysis is a useful tool in the study of human movement. This analysis takes the data within the time series and transforms it into sets of frequencies. This type of analysis is used to identify the commonly displayed frequency for a particular motion as a guide to apply a filter to the dataset (Giakas, 2004), and/or to differentiate changes in the frequency of movement between different groups or situations (Bravi, Longtin, & Seely, 2011; Giakas, 2004). The power spectrum is commonly used to extract the power produced at a specific frequency using a Fourier transform, which is a mathematical procedure that uses the sums of sines and cosines to describe complicated analog signals (Bravi, et al., 2011; Giakas, 2004). In order to sample the signal it must be digitized by applying a discrete Fourier transform (DFT), the components of which signal represent specific frequencies (known as harmonics). Most movement patterns retain similar frequencies; for example, the frequency of walking at the preferred speed usually falls around 1 Hz (Winter, 2009). The frequency in which the greatest amount of power is revealed is considered the first harmonic (also primary peak) and provides the most information about the signal (Giakas, 2004). Changes in the frequency in which the primary peak occurs in the power spectrum can differentiate between normal and pathological gait (Giakas & Baltzopoulos, 1997; Stergiou, Giakas, Byrne, & Pomeroy,

2002) and in COP dynamics between random and preferred postural sway (Morrison, et al., 2007).

Non-linear analysis

Traditional variability analysis provides information in relation to the group means and standard deviations; however, they can mask the subtle changes that occur in the neuromuscular system in response to perturbations (Dingwell & Cusumano, 2000). Nonlinear analyses provide researchers another set of tools in which to describe the changes in a physiological time series and have been shown to provide a greater understanding of the motor control processes as a result of specific task demands (Buzzi, Stergiou, Kurz, Hageman, & Heidel, 2003; Dingwell & Cusumano, 2000). Over the past several years, the use of nonlinear analyses has been used to identify the different properties of change in motor control that are not reflected in the traditional analyses. Derived from the chaos theory in physics the use of nonlinear analyses have been widely used to assess changes in various physiological signals and have been useful in describing changes in the cardiovascular system as a result of disease (Hausdorff, Forman, Pilgrim, Rigney, & Wei, 1992; Lipsitz, 2002; Pincus, Gladstone, & Ehrenkranz, 1991). The more commonly used analyses include Detrended Fluctuation Analysis (DFA), and entropy analyses such as Approximate Entropy (ApEn) and Sample Entropy (SampEn) (Bravi, et al., 2011). For the purpose of this dissertation, the DFA analysis will be briefly described; however, the non-linear approaches used in this project to assess the complexity of the postural control system include the use of ApEn (Pincus, 1991, 1995) and SampEn (Richman & Moorman, 2000; Yentes et al., 2013).

Detrended Fluctuation Analysis (DFA)

Hausdorff et al. (1995) was amongst the first to demonstrate that although there were many stride-to-stride fluctuations in gait (even at a self-selected pace) that in general, the gait cycle was very periodic when walking and the stride-to-stride fluctuations exhibited long-range correlations reflected over long durations (Hausdorff, et al., 1997b; Hausdorff, Peng, Ladin, Wei, & Goldberger, 1995). These self-similar patterns were determined by the use of DFA. Over the past few decades, this analysis has been used extensively to evaluate gait dynamics in patients with Parkinson's and Huntington's disease (Hausdorff, Cudkowicz, Firtion, Wei, & Goldberger, 1998b; Hausdorff, et al., 1997b), changes in the beat-to-beat time series in elderly with congestive heart failure (Hausdorff et al., 1994), and the impact of gait speed on gait dynamics (Jordan, et al., 2006, 2007; Terrier & Deriaz, 2012). This analysis is based upon forming a cumulative sum of the time series and then sections that time series into lengths of 4 to $N/4$ data points (N = the total number of data points) and then using the log of the average fluctuation size to plot against the log of the window size (Bravi, et al., 2011; Jordan, et al., 2007). DFA detects whether one time segment in a time series correlates with another segment at other points in time referred to as long range correlations. An alpha of 0.5 corresponds to white noise in which no correlations occur from one segment to another. An alpha of greater than 0.5 or less than 1.0 corresponds to persistent long range correlations and values below 0.5 correspond to anti-persistent correlations. The inference is that during gait, the higher the correlations the greater the dependence each stride is upon the previous one, whereas the anti-persistent correlations indicate greater independence that has been interpreted as more adaptability to respond to perturbations (Bravi, et al., 2011; Hausdorff, et al., 1997b; Hausdorff, et al., 1995; Jordan, et al., 2007).

Approximate Entropy (ApEn)

Approximate entropy is a mathematical algorithm that was originally developed by Pincus (1991) to understand the complex phenomena of physiological systems. ApEn has been one of the most commonly used non-linear techniques as it relates to the motor control processes of human movement, particularly to investigate changes in posture and locomotion as a result of the influence of intrinsic (individual) and extrinsic (task) demands (Georgoulis, et al., 2006; Harbourne & Stergiou, 2003; Morrison, et al., 2007; Newell, et al., 1997a; Newell, et al., 1997b; Newell & Vaillancourt, 2001; Pincus, et al., 1991; Pincus & Goldberger, 1994; Thomas, VanLunen, & Morrison, 2013; Vaillancourt & Newell, 2002).

ApEn computes the conditional probability of the signal by providing a measure of the likelihood that any given data point (n) in the time series that is close for m observations, remains close on the next incremental comparisons (m + 1). This is measured by the level of repetition that occurs between *m* and *m*+1 vectors within a tolerance range of the standard deviation *(r)* of a time series. This analysis produces a value between 0 and 2 with values closer to zero indicating higher repeatability of the vectors and a more regular signal. Higher ApEn values represent lower repeatability of the vectors *m* and *m* + 1 and represent greater irregularity (increased structure) in the time series. Increases in ApEn have been interpreted as an increase in the signal's time domain complexity (Hausdorff, et al., 1992; Lipsitz, 2002; Pincus, 1991; Pincus, et al., 1991; Pincus & Goldberger, 1994). The seminal investigation on the use of ApEn to identify illness and disease was conducted by Pincus, et al, (1991) as a means to identify the deterministic patterns in the EKG readings of infants that were in the neonatal intensive care unit with healthy infants. Traditional variability measures between the two groups were very similar (CV, SD); however, when ApEn analysis were conducted it was determined that the EKGs of the sick infants were highly deterministic as compared to the healthy infants (Pincus, et al., 1991). This finding was then put to the test in several other investigations to identify changes in complexity as a result of heart disease (Hausdorff, et al., 1992) and aging (Goldberger, Peng, & Lipsitz, 2002; Kavanagh, et al., 2006; Lipsitz, 2002). In the past couple of decades, this analysis has been used frequently to identify the changes in posture and gait as a result of the influence of the various constraints (Buzzi, et al., 2003; Georgoulis, et al., 2006; Harbourne & Stergiou, 2003; Morrison, et al., 2007).

Sample Entropy (SampEn)

Sample Entropy is another tool that is used to identify the deterministic structure of a complex physiological system. While not as widely used as ApEn, it has appeared in the literature more frequently in recent years, particularly in the time series analysis of gait where the data sizes are often relatively small (Costa, Peng, Goldberger, & Hausdorff, 2003; Richman & Moorman, 2000; Yentes, et al., 2013). This measure differs from ApEn in that it excludes the counts where a vector is compared with itself, thus avoiding the bias that self-matches introduce in the calculation (Bravi, et al., 2011; Richman & Moorman, 2000). This analysis has been shown to be largely independent of data size and has demonstrated greater consistency with smaller data sets such as those most commonly seen for gait. Typically, higher SampEn values indicate greater irregularity (randomness) in the signal while lower SampEn values indicate a more regular signal. Similar to ApEn changes in SampEn have been interpreted as changes in the complexity of the signal's time dependent structure and have also been useful in determining the complexity of various physiological systems.

CHAPTER III

Experiment I: Changes in Postural Sway as a Function of Prolonged Walking

Title: Changes in Postural Sway as a Function of Prolonged Walking

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Introduction

The ability for the postural system to adapt to transient perturbations from the environment is necessary to maintain stability. This is achieved through continuous regulation of sensory information to optimize balance control. Previous investigations have assessed the impact of different forms of perturbations on various populations including healthy young adults (Fransson, Magnusson, & Johansson, 1998; Nashner, 1976; Nashner & Cordo, 1981), individuals with injury to the lower extremity (Docherty, et al., 2006; Lysholm, et al., 1998), older adults at risk of falling (Doumas & Krampe, 2010; Lord & Menz, 2000), as well as those persons with neurological disorders (Horak, et al., 1990; Schieppati, et al., 1994; Schieppati & Nardone, 1991). Depending upon the nature of the perturbation, the postural system utilizes feedback and feed forward processes to offset and counteract the destabilizing effect (Fransson, et al., 2004; Horak & Diener, 1994; Horak, et al., 1990; Simoneau, et al., 1995). The subsequent postural reactions are reliant upon the magnitude and the direction of the disturbance as well as whether it was anticipated or not.

Postural disturbances can occur as a consequence of either unexpected external forces (e.g., the sudden movement of the support surface), or of an individual's voluntary movement (Horak & Macpherson, 2011). Following most perturbations, the initial response is characterized by a rapid reaction to readjust posture – a reaction which typically results in changes in the amount and/or frequency of center of pressure (COP) motion to maintain stability (Gatev, et al., 1999; Horak, et al., 1990; Loram, Maganaris, & Lakie, 2009; Nardone, et al., 1997; Nashner, 1976). If the individual is exposed to the same perturbation over successive trials, the individual is able to compensate for the perturbation to ensure overall balance and stability is maintain (Horak & Moore, 1993; Horak, et al., 1990; McIlroy & Maki, 1995; Nashner, 1976).

There is no doubt that a person is able to quickly adjust and compensate for most transient disturbances applied within a short time frame. However, less is known about how the postural system adapts to maintain balance following constant, low-level perturbations that would be encountered during everyday life. For example, what impact (if any) would walking at a rapid pace for an extended period of time have on balance assessments after the activity? While physical activity has been shown to impact balance and postural motion (Bove, et al., 2007; Caron, 2003, 2004; Davidson, Madigan, & Nussbaum, 2004; Gribble & Hertel, 2004; Kanekar, Santos, & Aruin, 2008; Nardone, et al., 1997), these studies have fatigued the person to varying degrees in order to observe any effect. Even under conditions of maximal exertion, the changes in COP motion appear transient, gradually returning to baseline levels over time. This decline is, in part, due to the ability of the person to adapt to the fatigue effects by appropriately scaling the amount/frequency of sway (Bove, et al., 2007; Nardone, et al., 1998; Nardone, et al., 1997). While there is no doubt that the postural system can adjust to the perturbation produced by physical activity, the time-dependent features that occur during this adaptation process still need to be assessed.

The aim of the current study was to identify the time course in which postural adaptation occurs while walking at faster than preferred speeds. Postural (COP) motion was assessed at specific intervals during the course of a 35 minute-walking trial. It was predicted that faster walking would produce initial changes in postural motion followed by a rapid adaptation and return to pre-activity levels. The greatest changes in postural sway motion would be seen while walking at the fastest gait speed (140% PWS).

Methods

Participants

Fourteen physically active young adults (7 males, 7 females; age = 24.79 ± 4.23 years) were recruited to participate in this study. All participants completed a physical activity readiness questionnaire (PAR-Q), and a medical history questionnaire. Subjects were excluded if they reported any neurological and/or cardiovascular disorder, loss of consciousness or concussion within the last year, and/or lower limb musculoskeletal injury/surgery within 1 year. Prior to participation all subjects provided written informed consent. The study was approved by the University Institutional Review Board and all experimental procedures complied with the guidelines.

Procedures

For this study, multiple assessments of each person's postural motion were taken while walking on a treadmill at three speeds. The speeds were preferred walking speed (PWS) and two faster speeds, 120% PWS and 140% PWS. Each person walked for 35 min at each speed. Postural sway data were recorded prior to each walking activity (pre-walking) and at 5-minute intervals, thereafter up to 35 min. Figure 3.1 illustrates the general protocol for each walking session.

Each participant was asked to attend the laboratory on two separate occasions at least 12 hours apart. During the first session, determination of PWS, the PWS condition and one of the faster walking speed conditions were performed. During the second session, the alternate fast walking condition was performed. The session in which the 120% and 140% PWS were performed was counterbalanced between participants. All walking and balance tasks were performed without shoes.

Gait Assessment: During the first session, each person's preferred walking speed (PWS) was determined by having them walk at their self-selected, comfortable walking speed along a 20 ft. GAITRite pressure sensitive walking surface (CIR systems Inc., Havertown, PA). Six trials were performed with the average speed used as their PWS. Following this, each person performed two of the walking conditions (PWS and one of the faster gait speeds) on an instrumented treadmill (h/p/cosmos mercury med 4.0) with an installed force distribution platform (FDM-T zebris Medical GmbH, Germany). A 15-minute rest was given between each walking condition. In the second session, participants completed the final gait speed condition (120% or 140% PWS).

Heart rate (HR) was monitored throughout the session using a heart rate monitor (Polar, Inc.). The age of each participant was used to determine their individual age-predicted maximal heart rate and was used in the following formula to identify physiological effort while walking (ACSM, 2005).

$$220\text{-}age = MHR;\ MHR\,/actual\ HR = \%\ MHR\ (exertion) \qquad (3.1)$$

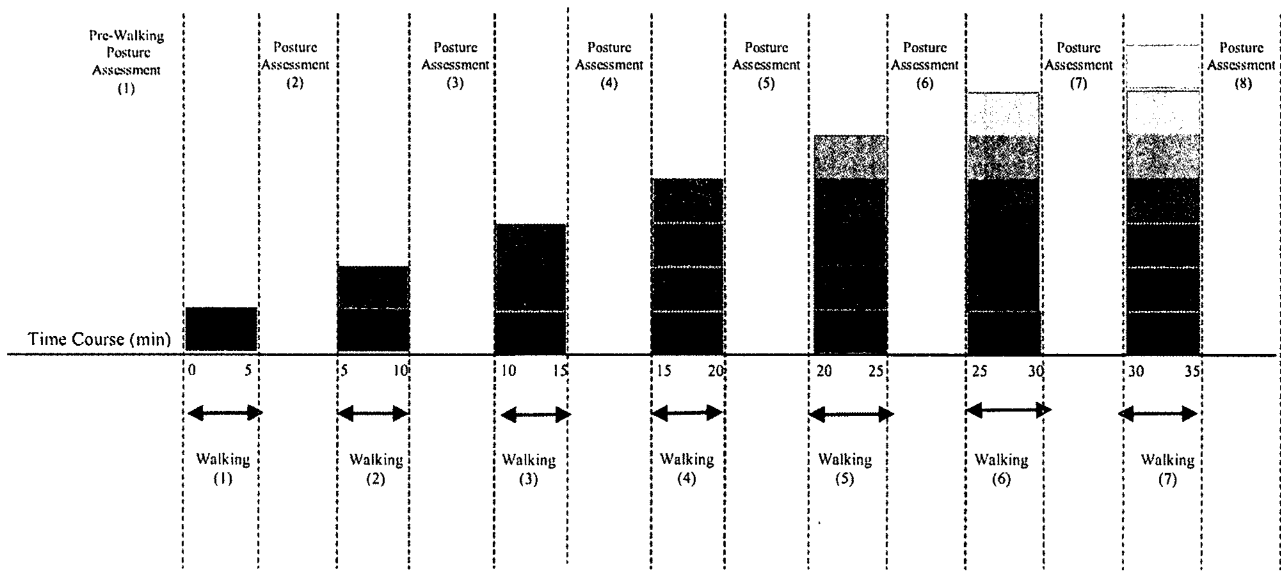

Figure 3.1. Illustration of the walking protocol depicting posture assessments (*top*) and time intervals (*bottom*) for each 5-minute period throughout the entire walking session. The boxes within each walking interval illustrate the cumulative impact of continuous walking on posture for each individual. Each posture assessment performed involved measuring COP motion both with the eyes open (EO) and with eyes closed (EC). A single 60 s trial was performed for each vision condition.

Subjects were asked to rate their perceived exertion (RPE) at 4.5 min of each 5-minute gait interval throughout the entire 35-minute period (a total of 7 measures) with a modified Borg-10 point scale (Borg, 1982). This scale ranges from 1 ("little or no exertion") to 10 ("maximal effort").

COP Assessment: Measures of the center of pressure (COP) were assessed using a Bertec force plate (Model 5050, Bertec Corp., Columbus, OH) at a sampling frequency of 1000 Hz. For all posture assessments, participants immediately stepped off of the treadmill onto a pliable surface (15.2 cm high, medium density foam pad) positioned on the force plate. Participants were instructed to adopt a comfortable bilateral stance on this surface with their feet hip distance apart. Lines were drawn on the foam surface to provide information as to each person's relative foot position (these corresponded with markings on the force plate beneath it). Each participant was instructed to stand comfortably with the toes in contact with the markings on the foam. Following each walking period, participants used these markings as a guide to position their feet for each posture assessment. The foam surface was used to provide greater postural challenge during standing. Individuals performed two 60s standing trials: one with eyes open (EO) and one with eyes closed (EC). Immediately following each assessment, participants would step back onto the treadmill and continue walking at the same speed. During the course of each walking condition, COP data was collected at discrete periods, namely, prior to the beginning of the walking activity (denoted as baseline), at 5-minute intervals for the duration of the task, and immediately following the final walking period. As highlighted in figure 3.1, this generated a total of sixteen posture trials per gait speed condition. The order with which the vision conditions were performed was counterbalanced across subjects.

Data Reduction and Analysis

All COP and HR data were processed using custom designed software programs in Matlab version 7.8 (R2009a, Mathworks, Inc., Natick, MA). Prior to analysis, the COP data were down-sampled from 1000 Hz to 100 Hz, and filtered using a 2nd order low-pass Butterworth filter (cutoff frequency 30 Hz).

COP Measures: The dependent measures calculated included: COP excursion (e.g., mean, standard deviation (SD), and maximal sway range), COP velocity, and total COP motion (path length and 95% ellipse area). COP velocity involves the total displacement of the COP in both the medio-lateral (ML) and anterior-posterior (AP) directions, divided by the length of the trial (COP velocity = total excursion/time). Path length identifies the total length of the COP excursion and is approximated by the sum of the distances between two consecutive points on the COP path in both the A-P and M-L axes. The 95% confidence ellipse sway area (ESA) was calculated using the equation by Prieto et al. (1996), in which the area of the 95% bivariate confidence ellipse is expected to include 95% of the points within the sway pathway (Prieto, et al., 1996).

Assessment of the degree of regularity of the COP data was performed using Approximate Entropy (ApEn) analysis. This analysis measures the conditional probability of the signal by providing a measure of the likelihood that any given data point (n) in the time series that is close for m observations, remains close on the next incremental comparisons ($m + 1$). This is measured by the level of repetition that occurs between m and $m+1$ vectors within a tolerance range of the standard deviation (r) of a time series. This analysis returns a value between 0-2 with lower values reflecting vectors of length m are more likely to be close (within the tolerance range) to the next incremental comparisons ($m + 1$) thus indicating greater regularity (less structure) in the time series. A perfect sine wave or a straight line with no deviation should produce an ApEn score close to zero. Higher ApEn values represent lower repeatability of the vectors m and $m + 1$ and represent greater irregularity (increased structure) in the time series. Increases in ApEn have been interpreted as an increase in the signal's time domain complexity (Pincus, 1991).

Statistical Analysis

Kolmogrov-Smirnov tests for normality were conducted on the data. Log_{10} transformations were required on several of the dependent variables to achieve a normal distribution prior to statistical analyses. Due to the consistent difference between vision and no-vision conditions, the postural data was divided into two datasets (eyes open and eyes closed). This allowed us to identify the effects of the intervention irrespective of vision.

A two-way analysis of variance (ANOVA) with repeated measures was conducted on the HR, RPE, and COP data (without pre-walking values). For HR and RPE data the within subjects factors were gait speed (3 levels) and walking time (7 levels). For the COP data, the within subjects factors were gait speed (3 levels) and posture trials (7 levels) to determine the impact that repeated walking sessions had on posture. Pairwise comparisons using Bonferroni corrections were used for determining differences where significant main effects and interactions were observed.

For each of the COP variables an average value of postural stability was calculated from assessments conducted prior to walking (pre-walking assessments). This value was used for comparison with successive posture trials over the entire walking time. To determine the impact that walking at the faster gait speeds had on postural motion, individual pairwise comparisons were conducted using pre-walking assessments and an average value of the 7 assessments taken at each gait speed. All analyses were conducted using Statistical Analysis Software (SAS Institute Inc., Cary, NC, USA) with significance levels set at $p < 0.05$.

Results

Measures of Heart Rate (HR) and Rate of Perceived Exertion (RPE)

Table 3.1 contains descriptive statistics of the group averages in mean HR, RPE scores, percent of maximal heart rate (% MHR), and COP variables for each gait speed condition. A significant gait speed by time interaction was observed for both mean HR ($p < 0.001$) and % MHR ($p < 0.001$). Pairwise comparisons revealed that walking at the faster gait speeds had the greatest impact on HR (both mean HR and % MHR) whereas, walking at PWS did not produce any significant changes. Heart rate increased 5% after walking at 140% PWS for 25 min (p =0.002) and 11% at 35 min compared to the first 5-10 min (p = 0.001).

Table 3.1. Average values (Mean ± SD) for gait speed effects of exertion and COP variables. COP measures are under both vision conditions

Measure of Exertion	Gait Speed		
	PWS	120% PWS	140% PWS
Mean Borg Value	2.3 ± .76[a,b]	3.2 ± 1.1[a,c]	4.8 ± 1.7[b,c]
Mean HR (BPM)	87.7 ± 6.8[a,b]	98.1 ± 11.4[a,c]	113.9 ± 16.9[b,c]
% Max HR	45.1 ± 3.3[a,b]	50.3 ± 5.6[a,c]	58.2 ± 8.7[b,c]

COP Variable		Gait Speed			
		Pre-walking	PWS	120% PWS	140% PWS
Path Length (mm)	EO	170.0 ± 33.5[a]	167 ± 32.8	166.8 ± 42.8	184.1 ± 54.5[a]
	EC	245.3 ± 56.3[a]	247.3 ± 86.2	235.8 ± 86.2[a]	258.2 ± 106.0
95% ellipse (mm2)	EO	568.7 ± 217.2[a,b,c]	654.5 ± 361.5[a]	809.3 ± 585.2[b]	803.8 ± 571.5[c]
	EC	1015.3 ± 495.3[a,b,c]	1231.0 ± 851.1[a]	1351.5 ± 1060.0[b]	1508.8 ± 1203.4[c]
Mean M-L (mm)	EO	13.3 ± 3.4	13.1±4.9	13.7±5.4	14.8±5.3
	EC	15.5 ± 3.2[a,b,c]	17.5±7.6[a]	17.5±7.7[b,d]	19.0±8.3[c,d]
Mean A-P (mm)	EO	18.8 ± 4.5[a,b]	20.1 ± 6.5	20.8 ± 7.6[a]	20.2 ± 7.5[b]
	EC	29.3 ± 7.6	29.8 ± 10.9	30.0 ± 12.2	31.1 ± 12.1
SD M-L (mm)	EO	4.8 ± 1.4[a]	4.7 ± 1.7	5.0 ± 1.7	5.1 ± 1.6[a]
	EC	6.0 ± 1.1[a,b,c]	6.1 ± 2.2[a,d]	6.3 ± 2.4[b]	6.6 ± 2.3[c,d]
SD A-P (mm)	EO	6.8 ± 1.3[a,b,c]	7.6 ± 2.3[a]	8.3 ± 3.2[b]	8.0 ± 3.1[c]
	EC	9.6 ± 2.5 [a,b,c]	10.3 ± 3.5[a]	10.8 ± 3.8[b]	11.8 ± 4.3[c]
Sway Range M-L (mm)	EO	25.8 ± 6.4[a]	25.5 ± 8.1[b]	27.4 ± 9.2	28.3 ± 9.0[a,b]
	EC	34.8 ± 13.2[a]	34.9 ± 13.0	35.7 ± 15.2	37.7 ± 13.7[a]
Sway Range A-P (mm)	EO	36.6 ± 6.5[a,b,c]	40.3 ± 12.4[a]	42.6 ± 15.7[b]	41.9 ± 14.9[c]
	EC	55.2 ± 14.6[a,b,c]	59.6 ± 20.5[a]	61.2 ± 21.7[b]	63.0 ± 23.3[c]
ApEn A-P	EO	.07 ± .02 [a,b,c]	.06 ± .02[a]	.06 ± .02[b,d]	.06 ± .02[c,d]
	EC	.08 ± .03[a,b]	.07 ± .02[a]	.09 ± .02[c]	.09 ± .02[b,c]
ApEn M-L	EO	.06 ± .02[a]	.06 ± .02	.06 ± .02[a]	.06 ± .02
	EC	.07 ± .02[a,b]	.06 ± .02[a,c,d]	.07 ± .02[b,c]	.07 ± .02[d]

Note. Mean values in the same row sharing the same superscript letters are significantly different from each other according to Bonferroni adjustment $p < 0.05$. ApEn = Approximate Entropy values, EO = eyes open, EC = eyes closed.

Table 3.2. Main effects and interactions of the 2-way ANOVA for gait speed and time on all dependent variables under both vision conditions

Measure of Exertion	Gait Speed Effects		Time		Speed*Time Interaction	
	$F_{(2, 26)}$	p	$F_{(6,78)}$	p	$F_{(12,156)}$	p
Mean Borg Value	35.42	< 0.001*	9.82	0.002*	1.56	0.11
Mean HR	41.25	< 0.001*	8.42	0.005*	8.98	< 0.001*
% Max HR	37.94	< 0.001*	8.66	< 0.001*	9.16	< 0.001*

COP Variable		Gait Speed Effects		Time (walking trials)		Speed*Time Interaction	
		$F_{(2, 26)}$	p	$F_{(6,78)}$	p	$F_{(12,156)}$	p
Path Length	EO	3.39	0.049*	0.29	0.94	1.98	0.03*
	EC	1.36	0.27	4.12	0.01*	1.54	0.11
95% ellipse	EO	2.53	0.99	0.48	0.82	0.63	0.81
	EC	3.03	0.06	0.84	0.54	0.90	0.55
Mean M-L	EO	3.64	0.04*	0.49	0.82	0.49	0.92
	EC	4.58	0.02*	1.75	0.12	0.63	0.81
Mean A-P	EO	0.17	0.85	1.25	0.29	1.42	0.16
	EC	0.78	0.47	1.44	0.21	0.71	0.74
SD M-L	EO	2.91	0.08	0.7	0.65	0.8	0.65
	EC	4.73	0.02*	1.91	0.09	0.83	0.62
SD A-P	EO	1.14	0.33	0.3	0.93	1.54	0.11
	EC	1.8	0.18	2.28	0.04*	0.97	0.48
Sway Range M-L	EO	4.12	0.03*	0.42	0.86	0.51	0.90
	EC	2.79	0.08	0.99	0.44	0.78	0.67
Sway Range A-P	EO	0.38	0.68	1.32	0.26	1.48	0.14
	EC	1.31	0.28	1.88	0.09	0.77	0.67
ApEn A-P	EO	4.03	0.03*	0.88	0.52	1.49	0.13
	EC	24.09	< 0.001*	2.65	0.02*	2.52	.04*
ApEn M-L	EO	1.84	0.17	1.03	0.41	1.09	0.37
	EC	20.71	< 0.001*	1.57	0.17	1.39	0.24

Note: * denotes statistical significance according to Bonferroni adjustment at the $p < 0.05$ level. EO=eyes open, EC=eyes closed conditions.

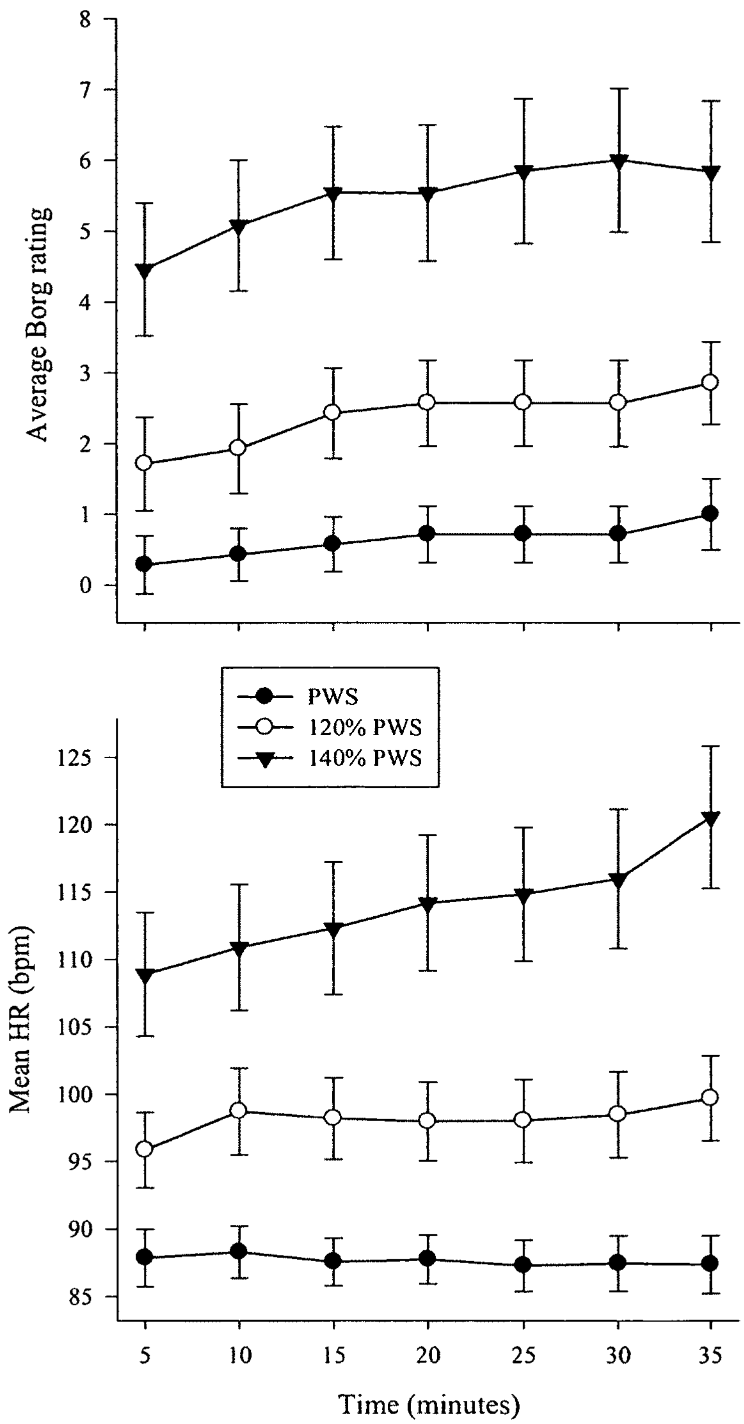

Figure 3.2. Mean values for heart rate (HR) and rate of perceived exertion (RPE) over time for each of the three gait speeds. HR increases occurred as a function of gait speed. Changes in RPE were seen as a function of gait speed and time.

Walking at 120% PWS produced a 2% increase in HR following 25 min of walking compared to the first 5 min ($p < 0.001$). ANOVA results for the effects of gait speed and time on all of the dependent variables are presented in Table 3.2.

Significant main effects for gait speed ($p < 0.001$) and time ($p = 0.002$) were found for RPE values. Higher RPE's were reported while walking at the fastest gait speed compared to 120% and PWS (*p's* < 0.001). Lower RPE scores were reported at PWS than at the faster gait speeds (120%; $p = 0.007$, 140%; $p < 0.001$). RPE was reported to be 12% greater following 15 min of walking ($p = 0.03$) and 15% greater after 30 min compared to the first 5 min. These changes in HR and RPE as a function of gait speed and time are illustrated in figure 3.2.

Posture Analysis

Figure 3.3 provides representative COP traces for a single subject taken at pre-walking (0 min) and during the final posture assessment (35 min) for each gait speed condition. The mean values for ML and AP excursion over time are illustrated in Figure 3.4.

The Impact of Gait Speed on COP variables

Gait speed had the greatest impact on changes in the COP variables. Average values (mean ± SD) for all of the dependent COP variables prior to walking (pre-walking) and for each of the three gait speed conditions are presented in Table 3.1.

A significant change in the mean (EO; $p = 0.04$, EC; $p = 0.02$), standard deviation (EC; $p = 0.018$), and range of COP excursion (EO; $p = 0.03$) in the ML direction was observed as a function of gait speed. Pairwise comparisons revealed that changes in various parameters of ML motion (e.g., variability (+ 9%, $p = 0.015$), mean (EC; +11%, $p = 0.026$), and range (+ 12%, $p = 0.031$) of ML COP motion) were greatest during the fastest walking speed compared to PWS (Table 3.2).

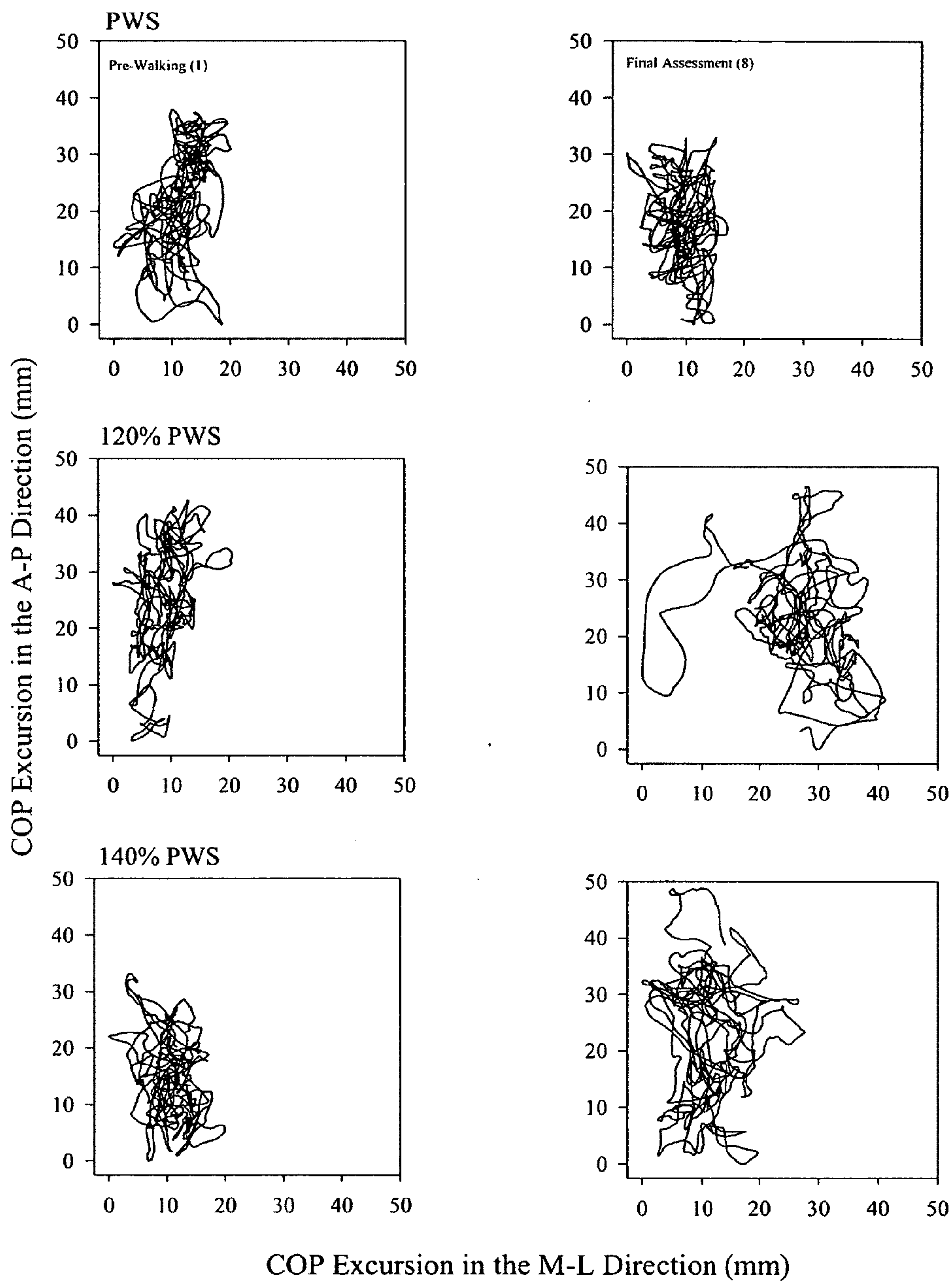

Figure 3.3. Representative examples of COP motion during the three gait speed conditions (PWS, 120% PWS, and 140% PWS). COP traces are shown for the initial (pre-walking) assessment and at completion of the walking protocol (35 min.). These examples of the COP motion shown were attained from a single participant standing on a foam surface with eyes open.

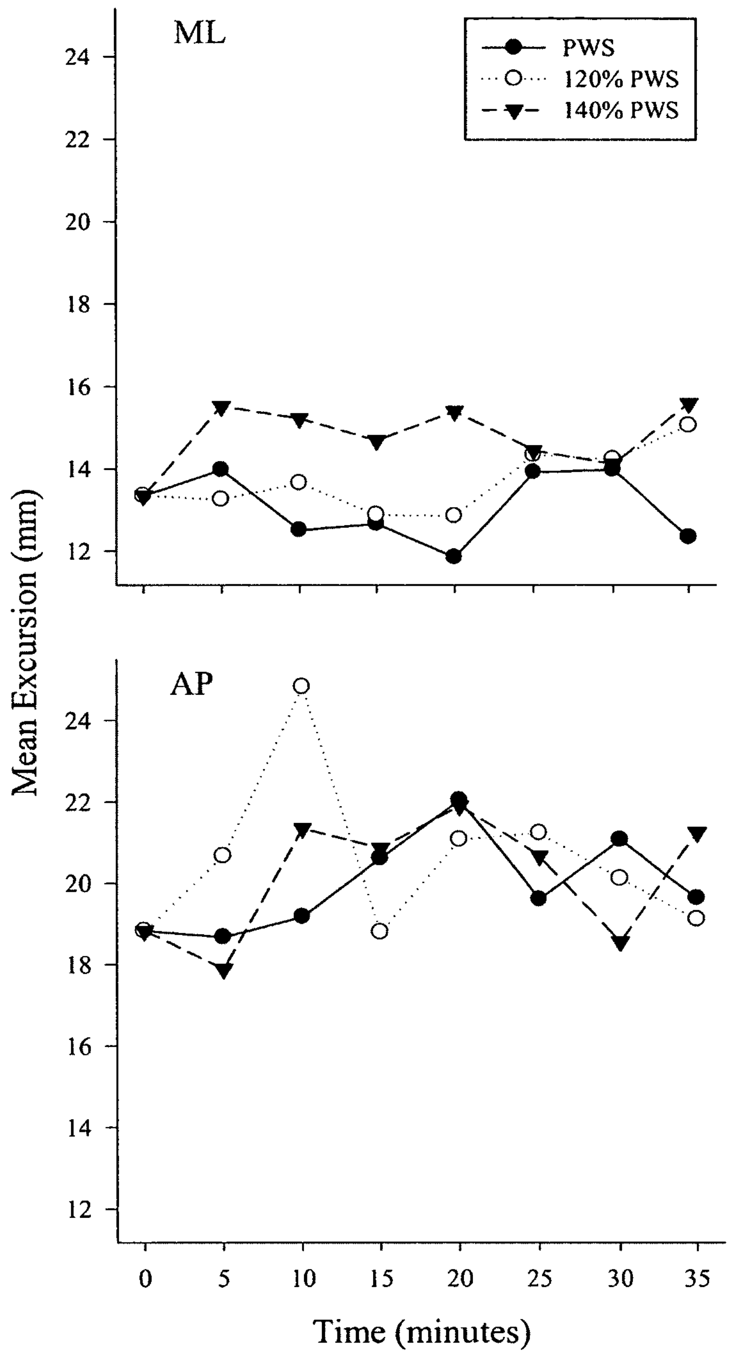

Figure 3.4. Changes over time in mean COP excursion for both ML and AP directions under the EO condition. Changes were seen in the mean ML as a function of gait speed.

Table 3.3. Pairwise comparisons of pre-walking values and average values of the 7 posture trials for all of the dependent COP variables.

COP variable		PWS		120% PWS		140% PWS	
		% change	*p*	% change	*p*	% change	*p*
Path Length	EO	- 1%	0.46	- 2%	0.23	+ 9%	< 0.001*
	EC	+ 1%	0.69	- 4%	0.03*	+ 5%	0.06
95% ellipse	EO	+ 13%	0.001*	+ 42%	< 0.001*	+ 41%	<0.001*
	EC	+ 21%	<0.001*	+ 33%	<0.001*	+ 49%	<0.001*
Mean M-L	EO	- 2%	0.56	+ 3%	0.43	+ 12%	0.004
	EC	+ 12%	0.002*	+ 13%	0.01*	+ 25%	<0.001*
Mean A-P	EO	+ 7%	0.05	+ 11%	0.008*	+ 8%	0.03*
	EC	+ 1%	0.73	+ 2%	0.56	+ 7%	0.09
SD M-L	EO	- 1%	0.66	+ 1%	0.09	+ 7%	0.04*
	EC	+ 7%	0.03*	+ 10%	0.009*	+ 17%	<0.001*
SD A-P	EO	+ 10%	0.001*	+ 21%	<0.001*	+ 19%	<0.001*
	EC	+ 8%	0.01*	+ 12%	<0.001*	+ 18%	<0.001*
Sway Range M-L	EO	+ 1%	0.63	+ 6%	0.07	+ 11%	0.003*
	EC	+ 4%	0.31	+ 6%	0.15	+ 14%	0.001*
Sway Range A-P	EO	+ 12%	<0.001*	+ 16%	<0.001*	+ 16%	<0.001*
	EC	+ 8%	0.04*	+ 11%	0.002*	+ 16%	<0.001*
ApEn A-P	EO	- 11%	0.001*	- 20%	<0.001*	- 10%	0.004*
	EC	- 17%	0.001*	+ 2%	0.53	+ 12%	<0.001*
ApEn M-L	EO	- 4%	0.30	- 15%	<0.001*	NC	0.99
	EC	- 12%	0.001*	+ 14%	<0.001*	+ 2%	0.46

Note: * denotes statistically significant difference from pre-walking assessment values $p < 0.05$. NC=no change from pre-walking assessments to average value at that gait speed.

Posture assessments taken following the repeated walking sessions indicated a significant gait speed-by-time interaction for path length with eyes open (p =0.03). Pairwise comparisons revealed that while walking at PWS path length decreased over time ($p = 0.03$) compared to the other gait speeds. No differences occurred between the repeated trials when Bonferroni corrections were applied.

Changes in postural stability were observed between the pre-walking assessments and those taken during walking. Significant differences between the pre-walking values and the average values for each gait speed were found in most of the COP variables. A greater amount of postural sway occurred to maintain stability during assessments after walking at the faster gait speeds than during PWS. Significant differences between walking at PWS compared to the pre-walking values were present in only a few of the COP variables. Significant increases in 95% ellipse sway (p <0.001) and standard deviation of COP excursion in the AP direction ($p < 0.05$) were seen for all gait speeds under both vision conditions. Table 3.3 summarizes the percent of change and p values between pre-walking and post-walking assessments at each gait speed under both vision conditions.

Impact of Walking Time on COP variables

A significant effect for time was seen for path length ($p = 0.009$) and the standard deviation of COP excursion in the AP direction ($p = 0.044$) under the EC condition. A 13% decrease in path length was seen after walking 25 min compared to the first walking trial (5 min) ($p = 0.036$). These changes were seen among all of the gait speeds but most apparent during PWS. No differences in the variability of AP COP excursion means were found when Bonferroni corrections were applied. The percent of change and p values for significant time and interaction effects are shown in Table 3.4.

Table 3.4. Pairwise comparisons of significant changes over time as a function of time and gait speed for exertion measures and COP variables

Measure of Exertion	Gait Speed by Time Interactions							
	Time		PWS		120% PWS		140% PWS	
Mean Borg Value	% change	*p*	% change	*p*	% change	*p*	% change	*p*
T5 v T15	+ 11%	0.03*	-	NS	-	NS	-	NS
T5 v T30	+15%	0.04*	-	NS	-	NS	-	NS
Mean HR (BPM)								
T5 v T25			-	NS	-	NS	+5%	0.04*
T5 v 30			-	NS	-	NS	+ 6%	0.03*
T5 v 35			-	NS	-	NS	+ 11%	0.01*
T10 v T25			-	NS	-	NS	+ 3%	0.04*
T10 v T35			-	NS	-	NS	+ 9%	0.04*
T25 v T35			-	NS	+2%	0.005*	-	NS
% Max HR								
T5 v T25			-	NS	-	NS	+5%	0.04*
T5 v 30			-	NS	-	NS	+ 6%	0.03*
T5 v 35			-	NS	-	NS	+ 9%	0.02*
T10 v T25			-	NS	-	NS	+ 3%	0.04*
T10 v T35			-	NS	-	NS	+ 7%	0.02*
T25 v T35			-	NS	+2%	<0.001*	-	NS

Table 3.4 (continued)

COP variable	Gait Speed by Time Interactions							
	Time		PWS		120% PWS		140% PWS	
	% change	*p*	% change	*p*	% change	*p*	% change	*p*
Path Length EC								
T5 v T25	- 13%	0.03*	- 21%	NS	- 3.5%	NS	- 13%	NS
ApEn AP EC								
T20 v T25	- 11%	NS	- 9%	NS	- 13%	NS	- 11%	NS

Note: * denotes statistically significant Bonferroni adjusted for multiple comparisons at $p < 0.05$ level. T=time intervals (i.e., T5 = following 5 min of walking, T10 = 10 min of walking). NS = not statistically significant.

Regularity of COP Motion

The respective differences in mean ApEn values for both AP and ML postural sway across the different gait speed conditions can be seen in Table 3.1. For COP motion in the AP direction, a significant gait speed-by-time interaction was found for ApEn values under the EC condition ($p = 0.04$). Pairwise comparisons showed that gait speed had the greatest impact on the interaction effect. This is indicated by greater signal regularity (lower ApEn) of COP motion in the AP direction during PWS than during 120% and 140% PWS (p's < 0.001). Compared to pre-walking values, signal regularity increased on average by 17% (lower ApEn) in assessments taken during PWS ($p = 0.001$) and decreased by 12% (higher ApEn) during 140% PWS ($p < 0.001$). Under the EO condition, a significant gait speed effect ($p = 0.030$) was seen in ApEn in the AP direction. Signal regularity decreased 14% (higher ApEn) while walking at 140% PWS compared to assessments taken during 120% PWS ($p = 0.046$). Changes in ApEn were seen across all of the gait speeds compared to pre-walking assessments with the greatest change seen at 120% PWS (- 20%, $p < 0.001$).

For COP motion in the ML direction, a significant change in ApEn values across the different gait speed conditions was observed ($p < 0.001$) under the EC condition (Table 3.2). Post hoc analysis indicated greater signal regularity (lower ApEn) of COP motion during PWS compared to the two faster gait conditions ($p < 0.001$). Under PWS conditions, ApEn tended to decrease over time following the pre-walking assessments. In contrast, ApEn values tended to increase from the initial assessments during the other two gait conditions.

Discussion

The aim of this study was to identify the time course in which postural adaptation occurs while walking at faster than preferred speeds. The results of this study indicate that two different, yet important, factors influenced postural stability. The first factor was the impact of walking at faster gait speeds. As expected, fast walking was shown to have the greatest impact on postural stability, with increases in the amount, variability, and structure of the COP signal. The second factor, the impact of successive walking trials over a prolonged period of time (35 min), was more subtle. Although walking at the faster gait speeds was reflected by a linear increase in measures of exertion (HR and RPE) over the entire gait activity, increases in postural motion were most evident within the first 5 min of walking. Over the remaining period, COP motion remained relatively constant for the remainder of the walking period. Overall, the results of the current study indicate that while walking at a faster pace produces a moderate increase in the degree of exertion the effects of this task were transitory on postural motion.

Fast Walking induces Alterations in Postural Control

The results demonstrated that walking at faster than preferred gait speeds was more physically demanding than walking at PWS. The effect of fast walking on posture was reflected by an increase in COP motion immediately after the beginning of the walking activity. Assessment of HR and RPE measures during each speed condition revealed that the fastest walking speed was reported to be more physically demanding than the other two conditions. For example, HR and RPE measures increased significantly from the PWS task, (HR: 45 ± 3.3% of age predicted maximal heart rate, and RPE: 2.34 ± 0.76) to the fastest gait speed task, (HR: 55% ± 9.5, and RPE: 4.82 ± 1.7). In line with the increasing levels of exertion during the faster walking speeds, changes in COP excursion were also seen during the 120% PWS and 140% PWS conditions. Generally, walking at a faster speed produced increases in the amount (path length, range, 95% sway area), variability (SD of sway), and structure (increased ApEn) of the COP motion. These increases are consistent with previous research which has reported that increasing levels of exertion systematically influences postural motion (Bove, et al., 2007; Lepers, Bigard, Diard, Gouteyron, & Guezennec, 1997; Nardone, et al., 1997; Pline, et al., 2006; Simoneau, et al., 2006).

The Impact of Walking Time on Postural Control

The results indicated that faster walking produced initial increases in the amplitude (path length, COP range), structure (ApEn), and variability (SD of sway) of postural motion in both the AP and ML directions. Changes in the COP motion within a single plane of direction (e.g., COP range and SD in AP direction) were only observed in comparison to pre-walking values and were not impacted by the subsequent walking trials over the remainder of the gait task. Changes in path length were seen relative to pre-walking values and were also impacted by the walking trials. While the general pattern was for an increase in postural motion as a function of the increasing task demands, those increases do not necessarily reflect a loss of stability. Indeed, Schieppati et al. (1994) reported that small increases and/or decreases in sway are both representative of an optimal level of postural control and reflect the ability to adapt to the demands of the postural task (Schieppati, et al., 1994).

As the postural task was repeated, a decline in path length was seen even as measures of exertion (HR, RPE) increased, indicating a period of adaptation to the impact of the subsequent walking trials. These results are consistent with Simoneau and colleagues (2006), who reported an initial increase in COP motion immediately following fast walking. Similar to our results, these authors noted that the initial increase was transient, leveling off after a short period of time. Together these findings indicate that the increased speed requirement of the task has an immediate impact on postural motion (as reflected by an increase in sway motion and variability). However, the impact of this task requirement is transient, with individuals demonstrating the ability to quickly adapt to the constraints inherent in the activity. Our results are also consistent with previous studies which have shown that individuals are able to quickly adjust to predictable task demands (within a few trials) and produce an appropriately scaled response (Horak, et al., 1990; Nashner, 1976). Overall, the rapid adjustments in balance observed over the course of the prolonged walking activity illustrate the capability of the postural control system to quickly adapt and compensate to the stresses placed upon it. While fast walking had an immediate effect on postural motion, this effect was transient and persons were able to compensate for this quickly.

Interestingly, this plateau effect observed for the COP measures was not reflected by a similar pattern of results for the measures of exertion. For example, measures of heart rate continued to increase over the entire time course of the fast walking activities with mean HR exhibiting changes from the first 5 minute period through to the end of the task (120% PWS, 96 to 99.7 BPM; 140% PWS, 109 to 120.5 BPM). Despite increases in HR and RPE, postural motion reached a steady state after the first 5-10 min. Together these results indicate that walking at a moderately fast pace has a differential effect on different systems within the body. While this dissociation between postural motion and exertion was observed during the fast walking conditions, it is unlikely that such a relation persists for more strenuous activity. Indeed, previous research has reported a strong relation between specific COP measures (sway path length) and maximal oxygen uptake during treadmill running (Bove et al., 2007).

Visual Impact on Postural Motion

While visual feedback has a strong impact on postural stability (Ledin, Fransson, & Magnusson, 2004; Schieppati, et al., 1994; Vuillerme, et al., 2006b; Vuillerme, et al., 2001), the pattern of COP changes observed in the current study were seen regardless of the presence/absence of vision. Consistent with other studies (Bove, et al., 2007; Caron, 2004; Vuillerme & Hintzy, 2007), an increase in the amount of sway was seen with the absence of vision however; a similar trend was seen over the two conditions. One suggestion could be that while the degree of postural motion initially increased with the moderate degree of exertion, the effects of fast walking were not strong enough to require increased input from other non-affected postural mechanisms. It is more likely that the initial effects of walking at a faster speed could be compensated for easily and quickly.

Conclusion

Overall, the results indicate that individuals were able to quickly adapt to the effect fast walking had on balance. Walking at a faster speed produced increases in various parameters of postural motion however; these changes were transient, leveling off early in the gait task and in some cases even declining by the end of the activity. These adaptations were observed despite an increase in measures of exertion (HR, RPE) across the time course of the walking task. Together these findings highlight that the postural system is able to quickly adapt to the task demands of walking at a faster pace to maintain an optimal level of balance.

CHAPTER IV

Experiment II: Temporal Changes in Stride-To-Stride Variability during Slow Walking

Title: Temporal Changes in Stride-To-Stride Variability during Slow Walking

Authors: Kathleen S. Thomas, Daniel M. Russell, Bonnie L. Van Lunen, Sheri R. Colberg, Steven Morrison

Introduction

Walking is the most common form of locomotion in humans (Duysens & Van de Crommert, 1998; Winter, 1983). While individuals can easily adapt to changes in gait speed with little conscious effort, each individual has a preference for a particular speed, step length, and stride period (Choi & Bastian, 2007; Reisman, Block, & Bastian, 2005).

When an individual is walking at his or her preferred pace (~4.4 kmh), the resultant gait pattern has a smaller magnitude of variability amongst the specific spatial and temporal parameters (e.g., stride interval, stride length, and step width) (Beauchet, et al., 2009; Dingwell, et al., 2001; Hausdorff, 2004; Jordan, et al., 2007). In contrast, when forced to change their pattern to accommodate speeds that are either faster or slower than preferred, individuals require greater active control to adjust to the task demands (Jordan, et al., 2007; Orendurff, et al., 2004) consequently resulting in an increase in variability within the gait parameters (Beauchet, et al., 2009; Bruijn, et al., 2009; Chiu & Wang, 2007; Chung & Wang, 2010; Hausdorff, 2005).

Previous studies investigating the impact of speed on gait have used the mean and variance values calculated from a small number of strides to characterize changes in variability as a result of some type of intervention or as a consequence of aging and/or disease (Beauchet, et al., 2009; Grabiner, et al., 2001; Schniepp, et al., 2012). While this approach has been useful in differentiating changes as a result of speed, this assumes that the gait cycle is stationary and does not take into account time-varying changes in the variability. Changes in motor control are time-dependent and alternate approaches to analysis of the time series can identify the non-stationary properties within physiological signal. The presences of non-stationarity is inherent in physiological time series and is considered to be part of an adaptive process within the sensorimotor system (Newell, et al., 1997a). Non-stationarity is defined as differing statistical properties of a time series at different segments along the sequence (Newell, Deutsch, Sosnoff, & Mayer-Dress, 2006). The use of instrumented treadmills has made it easier to collect repeated strides for a longer duration, thus providing the ability to identify the impact that time has on gait changes (Bruijn, et al., 2009; Chiu & Wang, 2007; Jordan, et al., 2007; Owings & Grabiner, 2004b). While most of these studies investigated time dependent changes of variability and structure over a longer duration, the pattern in which variability itself changes at specific increments over the course of the walking task has not been investigated. Changes in the stride interval as a function of time when the gait speed remains constant is of interest as it can reveal the non-stationary dynamics of the gait cycle.

One of the first studies to investigate non-stationarity of the gait cycle was conducted as a means to identify the amount of time necessary for young adults unfamiliar with treadmill walking to habituate their gait to mimic overground walking (Wall & Charteris, 1980). In that study, Wall and Charteris, (1980) reported that a rapid accommodation to walking on a treadmill (~ 10 sec) was followed by a longer period of habituation in which the mean stride interval became more similar to overground walking in individuals unfamiliar with treadmill use. As the task of treadmill walking at three speeds (slower, normal, and faster) became more familiar, the mean stride time increased for up to 10 min (particularly at the slower speed) before leveling off and settling into the newly acquired gait pattern. Similar findings were revealed in the gait dynamics of children naïve to treadmill walking during overground, supported (holding onto handrails) treadmill, and unsupported treadmill walking (Fairley, et al., 2010), which further suggest that gait is non-stationary and takes time to adapt to differing task demands. Although the previous studies (Fairley, et al., 2010; Wall & Charteris, 1980) used changes in the mean stride time to indicate habituation to the different tasks, stride-to-stride variability can indicate changes in the dynamics of gait that may further reflect the adaptive capacity of the locomotor system (Hausdorff, 2005). The changes in signal regularity have provided additional information regarding the ability to adjust to changes in a given task such as walking at different speeds. Higher levels of regularity of a particular signal are often interpreted as a decrease in complexity, indicating less adaptability to changes in the environment (Pincus, 1995).

Indeed, it has been shown that variability increases while walking both slower and faster than PWS and that walking slower elicits greater stride time variability than the faster speeds. Jordan et al. (2007) identified that greater variability (as a percent of CV) occurred at 80% and 90% PWS compared to PWS, 110% and 120% of preferred (Jordan, et al., 2007). Consequently, this study design employs slower gait speeds of 80% and

90% PWS to determine if young adults were able to reduce stride time variability as they become more habituated to walking at the slower gait speed.

The aim of this study was to explore how variability in the stride interval changed over successive periods of time (e.g., from the first 5 min to the next 5 min, and so on) while walking at speeds slower than preferred. It was hypothesized that walking at a slower speed would increase the variability and signal regularity of the stride interval initially as an attempt to regulate the gait pattern to accommodate the slower pace. Additionally, it was predicted that as walking continued at the slower speed a decrease in variability and signal regularity would emerge as a result of adaptation to the task despite walking at a constant pace.

Methods

Participants

Fourteen physically active young adults (5 males and 9 females; age 24.47 ± 5.32) were recruited to participate in this study. All participants completed a physical activity readiness questionnaire (PAR-Q), and a medical history questionnaire. Subjects were excluded if they reported any neurological and/or cardiovascular disorder, loss of consciousness or concussion within the last year, and/or lower limb musculoskeletal injury/surgery within 1 year. Prior to participation all subjects provided written informed consent. The study was approved by the University Institutional Review Board and all experimental procedures complied with the guidelines.

Procedures

The stride interval was collected from each person walking on a treadmill at three speeds. The speeds were preferred walking speed (PWS) and two slower speeds, 90% PWS and 80% PWS. The order of the gait speeds were counterbalanced between subjects.

All of the testing was completed in a single session that consisted of establishing the PWS for each individual followed by continuous walking for 30 min at each of the three gait speeds. A 15-minute rest was given between each walking session to eliminate any fatigue effects. Each person's PWS was determined by having him or her walk at a comfortable, self-selected walking speed along a 20 ft (610 cm), pressure sensitive walking surface (CIR systems Inc., Havertown, PA). Six trials were recorded, and the average speed over the trials established the PWS and was used to calculate the speeds of 90% and 80% of PWS.

Heart rate (HR) was recorded using a Polar® monitor (Polar, Inc.). The mean HR value was used to determine the physiological effort at each gait speed and was used for analysis. A rating of perceived exertion (RPE) was obtained at 4.5 min of each 5-minute time block throughout the entire 30-minute walking period using a modified Borg, 10-point scale. The modified Borg scale ranges from 1 being "little or no exertion" to 10 being "maximal effort" (Borg, 1998).

Each individual performed a warm-up and familiarization period on the treadmill at their PWS for up to 2 min prior to data recording. Approximately 1500 strides over 30 min were collected for each walking condition on a treadmill (h/p/cosmos mercury med 4.0) with installed pressure plate (FDM-T Zebris Medical GmbH, Germany). The data were processed in six 5-min time blocks (0-5 min, 5-10 min, 10-15 min, 15-20 min, 20-25 min, and 25-30 min) at each gait speed condition. Specific gait events were established using WinFDM-T© software (Zebris Medical GmbH, Germany). Stride interval was defined as the time between the heel strike of one foot and the subsequent heel strike of that same foot. Stride interval data were derived using MyoResearch XP Master Edition1.07.09 (Noraxon U.S.A. Inc, Scottsdale, AZ) software. All walking sessions were performed without shoes.

Data Reduction and Analyses

Gait data was processed using custom software programs developed with Matlab version 7.8.0 (The Mathworks, Inc., Natick, MA). The mean, standard deviation (SD), and coefficient of variation (CV) were determined for each 5-minute time block and used for analysis. The CV was computed as SD divided by the mean and multiplied by 100 to quantify the amount of variability in each time series.

The degree of regularity of the stride interval was computed using Approximate Entropy (ApEn) analysis. This analysis is used to determine the degree of predictability or regularity in a physiological time series. Typically, a more regular signal produces an

ApEn value close to zero while a more irregular signal produces an ApEn value closer to 2. Changes in ApEn have been interpreted as changes in the signal's time domain complexity and have been used to determine the complexity of various physiological systems (Morrison, et al., 2007; Pincus, 1991; Pincus & Goldberger, 1994).

Statistical Analysis

Normality of the stride interval data was determined using Kolmogrov-Smirnoff and skewness and kurtosis tests prior to analysis. Where necessary, a $\log_{10}$ transformation was performed to achieve a normal distribution prior to analysis. A Mixed General Linear Model (GLM) with repeated measures was used for analysis on each of the dependent variables. The within-subject factors were speed (3 levels) and time block (6 levels). Pairwise comparisons were conducted where significant main effects and interactions were corrected using Bonferroni adjustments. All analyses were conducted using Statistical Analysis Software (SAS, version 11.0; SAS Institute Inc., Cary, NC, USA) with significance levels set at $p < 0.05$.

Results

Effect of gait speed and walking duration on HR and RPE

Participants walked for 30 min at each of the three gait speed conditions: 80% PWS (3.3 ± 0.3 kmh), 90% PWS (3.7 ± 0.3 kmh), and PWS (4.1 ± 0.3 kmh). Figure 4.1 illustrates changes in HR and RPE as a function of changes in walking speed and time block. The results revealed a significant effect of gait speed for HR ($F_{2,25} = 133.33, p < 0.0001$) and RPE values ($F_{2,26} = 15.05, p < 0.0001$). Post hoc analyses revealed that walking at 80% PWS (94 ± 3.0 bpm) elicited a lower mean HR compared to 90% PWS (97 ± 3.1 bpm) and PWS (99 ± 3.0 bpm; p's < 0.0001). There was no difference in mean HR at 90% PWS compared to PWS. Lower RPE values were reported during the slower gait speeds (e.g., 90% PWS; 2.1 ± 1.1, $p < 0.0001$; 80% PWS; 2.2 ± 0.7, $p = 0.0009$) compared to PWS (2.6 ± 1.0). A significant time block effect was seen for mean HR ($F_{5,65} = 6.51, p < 0.0001$) and RPE values ($F_{5,65} = 5.37, p = 0.0003$). Post hoc comparisons revealed significant increases in both HR and RPE values in the 3rd block compared to the 1st block (p's < 0.05). HR and RPE values remained higher over the subsequent time

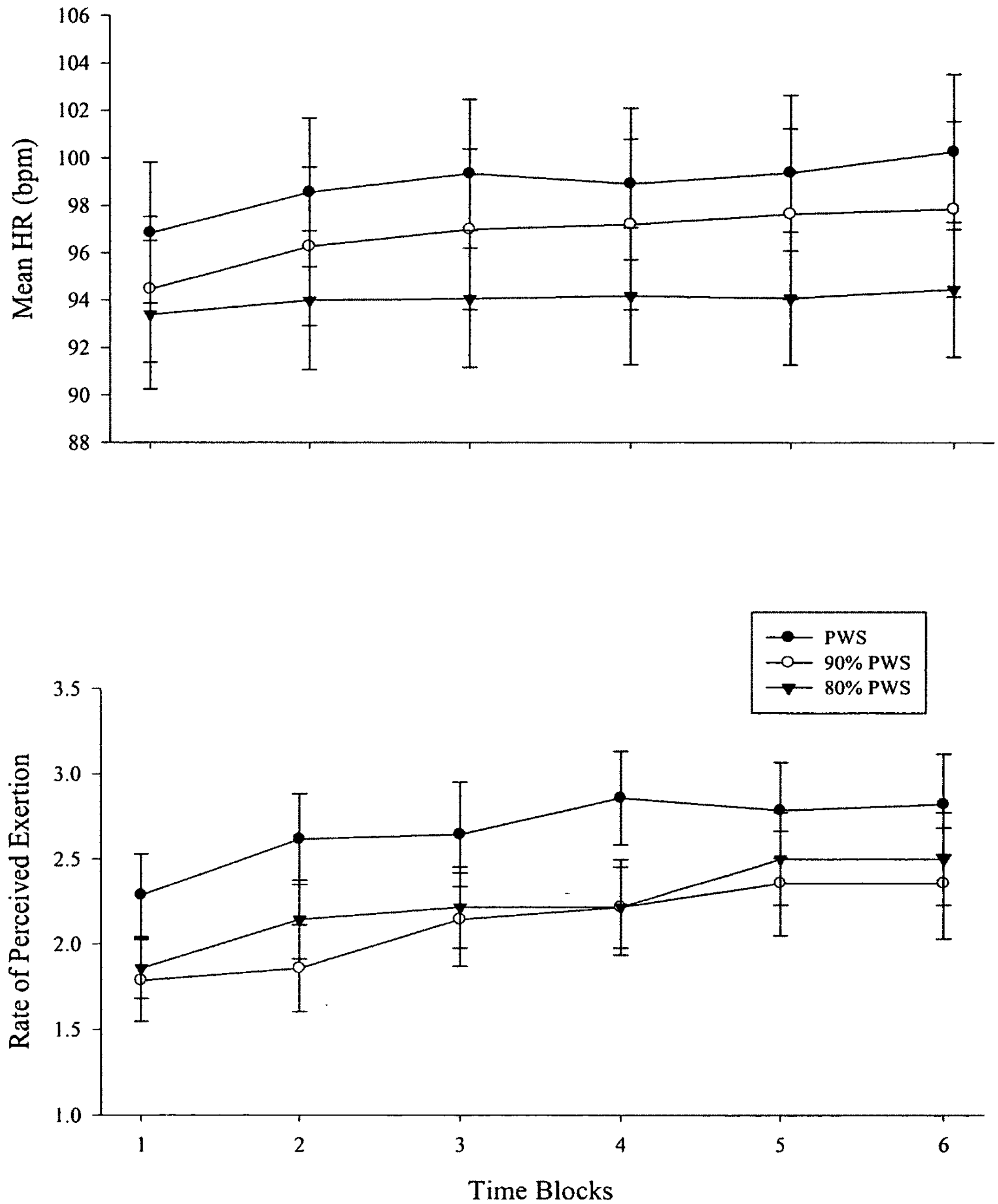

Figure 4.1. Changes in the mean HR and RPE as a function of gait speed and time. Each time block is equal to 5 min (1=0-5 min, 2=5-10 min, 3=10-15 min, 4=15-20 min, 5=20-25 min, 6=25-30 min).

blocks (4-6) (p's < 0.05) compared to the initial block (Figure 4.1). No significant interactions for gait speed and time block were seen in either HR or RPE.

Stride interval changes as a function of gait speed and walking duration

Figure 4.2 provides representative raw data for stride interval duration under each of the gait speed conditions and time blocks across the entire walking task. Figure 4.3 illustrates the impact of gait speed and time block on the mean, CV, and ApEn of the stride interval. A significant main effect for the mean of stride interval was found for gait speed ($F_{2, 26} = 1199.83, p < 0.0001$) and block ($F_{5, 65} = 2.53, p = 0.04$). For the speed effect, stride interval was longer when walking at 80% PWS compared to the faster gait speeds (p's < 0.0001). For the effect of time, a decrease in mean stride interval was seen in the 4th time block ($p = 0.04$) as compared to the initial block (0-5 min). Although there was no significant interaction effect ($p = 0.12$), the largest decline in the mean stride interval (Figure 4.3) was observed in the slowest gait speed at the 4th time block.

A significant interaction for gait speed and time block was revealed in the CV of stride interval ($F_{10,125} = 2.35, p = 0.014$). This is seen in time blocks 1 and 2 while walking at the slowest gait speed (80% PWS) (Figure 4.3). Further pairwise comparisons revealed a significantly greater CV in time blocks 1 and 2 compared to blocks 3 ($p = 0.006$), 4 ($p = 0.04$), and 5 ($p = 0.01$) while walking at 80% PWS only. No changes in CV of stride interval occurred from blocks 3-6 regardless of the gait speed.

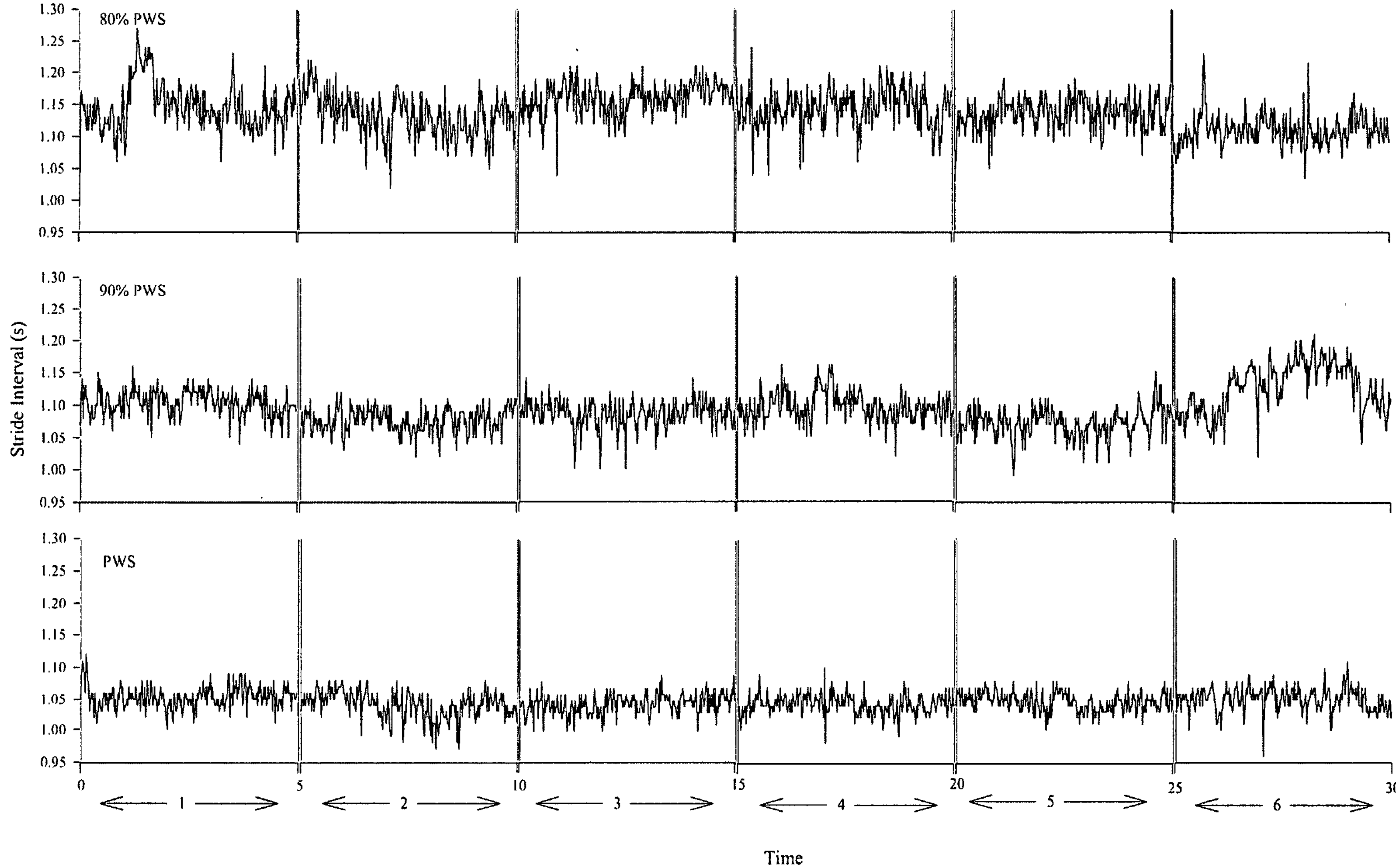

Figure 4.2. Representative stride interval time series of the left leg for each of the gait speed conditions (PWS, 90% PWS, and 80% PWS). Each 5-min segment consists of ~ 245 strides. The entire 30 min session is equal to ~ 1470 strides total. Time blocks are defined by the arrows below the time (in min) range.

Results of the ApEn analysis on the stride interval revealed significant main effects for gait speed ($F_{2,26} = 99.27, p < 0.0001$) and time block ($F_{5,65} = 3.36, p = 0.009$). Post hoc analyses for gait speed revealed a more regular signal (lower ApEn) while walking at 80% PWS compared to the other two gait conditions (p's < 0.0001). For the effect of time, greater signal regularity was observed in the 2nd time block (5-10 min) compared to the 3rd block (15 min) ($p = 0.02$); however, no significant differences were seen between the 2nd time block and blocks 4-6. Interestingly, there were no significant differences between the 1st time block and any of the subsequent periods. No changes in signal regularity were observed between blocks 3-6 (Figure 4.3).

Discussion

The aim of this study was to identify the impact that slow walking had on gait dynamics over time. In general, walking at a pace 20% slower than preferred elicited changes in the amount, variability, and structure of the stride interval compared to PWS. Furthermore, these increases in the amount, variability, and structure of the stride interval were only seen in the initial time blocks (0-5, 5-10 min) while walking at 80% PWS. The changes in variability and structure that were seen while walking at the slowest gait speed condition decreased over time (10-15 min), and eventually reached a steady state pattern for the remainder of the task. Overall, these results indicate that while the locomotor system can easily adjust to walking at a speed 20% slower than preferred, it takes time to adapt and settle in to the task.

Changes in gait dynamics as an effect of speed

Walking at slower speeds produced differential changes in the measures of gait (stride-to-stride interval) and exertion. As expected, at slower gait speeds, mean stride interval increased compared to PWS. These findings are consistent with previous investigations that have reported similar increases in stride interval when walking slower (Beauchet, et al., 2009; Chiu & Wang, 2007; Chung & Wang, 2010). Likewise, a slower speed also resulted in increases in the level of stride-to-stride variability (Beauchet, et al., 2009; Bruijn, et al., 2009; Jordan, et al., 2007) compared to PWS. In the current study,

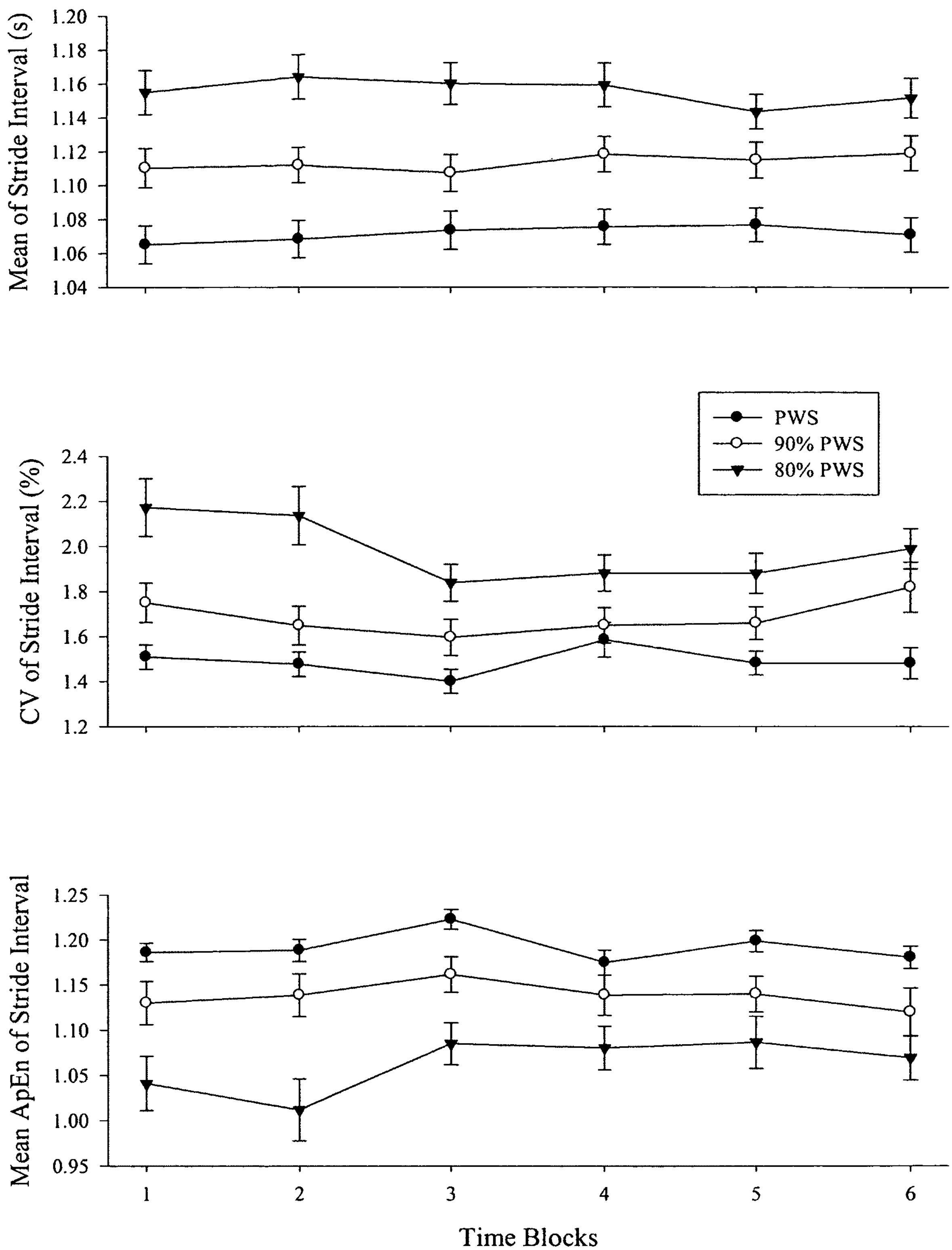

Figure 4.3. The mean (*top*), % CV (*middle*) and ApEn (*bottom*) over the specific time blocks for each of the gait speeds. Changes over time were only seen at 80% of PWS compared to the other gait speeds

the amount of stride-to-stride variability while walking at PWS was low (CV ~ 1.5%), which is consistent with other studies indicating a very repeatable movement pattern (Jordan, et al., 2007; Winter, 1995). As gait speed decreased, an increase in stride-to-stride variability was observed. For example, a decrease in the gait speed of 10% resulted in only a modest increase in stride-to-stride variability with a CV of 1.7%, whereas decreasing the speed by 20% resulted in a significant increase in variability (CV = 2.2%) compared to PWS.

The decrease in the measures of exertion (e.g., RPE) during the slower gait conditions compared to PWS confirms that slower walking was perceived as requiring less exertion than a preferred pace. Despite the finding that walking slower required less exertion, the concurrent increase in gait variability may signify that this condition requires greater active control of the motor output to adjust to the non-preferred pace (Jordan, et al., 2007). A possible explanation could be the result of the increase in ML excursion of COM during slow walking which produces an additional challenge to balance thus increasing variability (Orendurff, et al., 2004).

The impact of gait speed was further evidenced by changes in the structure (ApEn) of the stride interval. Overall, walking at 80% of PWS elicited higher levels of signal regularity compared to that of PWS and 90% PWS. Greater signal regularity (lower ApEn) is indicative of less flexibility in various physiological systems as a result of changes in task demands (Morrison, et al., 2007). It was expected that walking at the slowest gait speed would elicit higher levels of regularity than the other gait speed conditions, particularly in comparison with PWS. These results reflect that the further the gait speed gets from preferred, the more control necessary to adjust to the walking task that are not observed at preferred and speeds similar.

Changes in the gait dynamics as a result of slow walking over time

The unique finding of this study relates to the impact slow walking had on the variability of the gait cycle over time. Specifically, when walking at 80% PWS, the pattern of stride-to-stride variability was greater during the initial periods of the walking activity (0-10 min) before settling into a lower level of variability that was similar to that at the other speeds. These changes over time indicate that when walking at this slower pace, the pattern of stride-to-stride variability exhibits a degree of non-stationarity, especially during the initial walking periods. Interestingly, the time-dependent changes at the slowest gait speed were not observed at PWS or at 90% PWS, indicating that there is a degree of stationarity to gait when walking at or near a preferred pace. Not surprising, the initial attempts to adjust stride interval to the slowest walking speed resulted in an increase in variability (CV) and a decrease in signal regularity (lower ApEn). Increased variability at the slower gait speeds is consistent with previous research (Bruijn, et al., 2009; Choi & Bastian, 2007; Georgoulis, et al., 2006; Jordan, et al., 2007; Kang & Dingwell, 2008), indicating that walking at slower speeds requires a change to the rhythm of the gait pattern to allow for adaptation to the slower speeds. While the aforementioned studies investigated changes in variability as a result of longer walking periods at different gait speeds, they did not explore the pattern in which the variability itself changed from one period of time to the next (Jordan, et al., 2007). For example, in the current study the variability and regularity of the stride interval from the first 5 min of walking was then compared to the next 5 min and so on over 6 successive increments that produced a pattern of variability over the entire 30 min session. Previous investigations that did explore changes over distinct time periods found that there was a period of habituation necessary to adapt to unfamiliar tasks (Matsas, Taylor, & McBurney, 2000; Owings & Grabiner, 2004b; Wall & Charteris, 1980). In the current study, temporal changes in stride-to-stride variability at the slowest gait speed (80% PWS) did not emerge until after 10 min of walking (~500 strides), whereas in previous studies kinematic and temporal changes were seen after 6 min (Matsas, et al., 2000; Wall & Charteris, 1980). The eventual decrease in variability seen after the first 10 min leveled off and was followed by a steady state pattern for the remainder of the task. The stride-to-stride variability at 80% PWS over the remaining time instances (15-30 min) became more similar to that at PWS and 90% PWS, indicating that the locomotor system was able to adjust their gait cycle to the task demands over a short period of time. These results are consistent with reports from several other studies indicating that movement variability may be diminished with task repetition (Fairley, et al., 2010; Nashner, 1976).

Conclusion

Overall, these results highlight that walking at a speed 20% slower than preferred is not difficult to do for a healthy young adults. While there was a higher amount of stride-to-stride variability initially, this accounted for only a small portion of the trial whereas walking at 90% PWS did not have a significant effect. These changes in the pattern of variability suggest that adaptation to the slower gait speed occurs as a result of the repetition of walking at a constant pace. Walking at speeds at/or near preferred produces relatively stationary gait dynamics whereas walking at 20% slower than preferred gait reveals non-stationary properties. The healthy neuromuscular system is able to adapt to walking at a slower than preferred pace, but as the pace decreases, it takes longer to adapt to the change in speed.

CHAPTER V

Experiment III: The impact of intentionally increasing stride-to-stride variability during gait

Title: The impact of intentionally increasing stride-to-stride variability during gait

Authors: Kathleen S. Thomas, Daniel M. Russell, Bonnie L. Van Lunen, Steven Morrison

Introduction

Walking is undoubtedly the most common form of locomotion in humans and because of this; it has been the focus of extensive research over the past several decades. Previous studies have examined the impact of changes in gait as a result of aging and/or disease (Hausdorff, et al., 1997a; Hausdorff, et al., 1997b; Hausdorff, et al., 2001; Hausdorff, et al., 1999; Hollman, Kovash, Kubik, & Linbo, 2007), injury (Georgoulis, et al., 2006), and maturation (Hausdorff, et al., 1999). Similarly, many studies have investigated the effects of different task demands on changes in the complexity of gait (Abernathy, Hanna, & Plooy, 2002; Beauchet, et al., 2009; Beauchet, Dubost, Herrmann, & Kressig, 2005; Bruijn, et al., 2009; Bruijn, et al., 2012; Choi & Bastian, 2007; Dingwell, et al., 2001; Fairley, et al., 2010; Jordan, et al., 2007; Jordan & Newell, 2008; Kokshenev, 2004; Sparrow, Begg, & Parker, 2008). Walking at non-preferred speeds is one of the more commonly investigated tasks related to the complexity of gait dynamics (Abernathy, et al., 2002; Beauchet, et al., 2009; Bruijn, et al., 2009; Chiu & Wang, 2007; Dingwell & Marin, 2006; England & Granata, 2007; Jordan, et al., 2007; Jordan & Newell, 2008). Walking at a preferred pace is a relatively stable action with some inherent variability from stride-to-stride amongst the specific gait parameters, such as stride time, stride length, step time, and step length (Beauchet, et al., 2009; Dingwell, et al., 2001; Hausdorff, 2004; Jordan, et al., 2007). When required to walk at a pace that is faster or slower than preferred, greater active control is required to adjust to that speed and the resultant gait pattern is more variable (Beauchet, et al., 2009; Bruijn, et al., 2009; Chiu & Wang, 2007; Chung & Wang, 2010; Jordan, et al., 2007).

Given that intent can alter gait, an important question is how much individuals can consciously increase the variability of gait. What changes would emerge if the goal of the task was to intentionally increase the variability of the gait cycle while walking at an imposed speed? The intent to produce random movement as the goal of the task has just recently been the focus of investigation (Deutsch & Newell, 2004; Morrison, et al., 2007; Newell, et al., 2000a; Newell, et al., 2000b; Newell & Vaillancourt, 2001).

While it has been established that humans have difficulty generating random strings of letters, symbols and numbers (Figurska, et al., 2008; Rosenberg, et al., 1990; Wagenaar, 1972), few studies have examined the extent at which an individual is able to produce random movements (Morrison, et al., 2007; Newell, et al., 2000b; Shumway-Cook & Horak, 1986). When adult participants were asked to move randomly, whether it be the index finger (s) (Newell, et al., 2000a) or different segments of the upper limb (index finger, hand, lower arm, whole arm) in a single plane (Deutsch & Newell, 2004; Newell, et al., 2000a), participants were able to produce a higher coefficient of variation (CV), standard deviation (SD), and greater irregularity of the signal (higher approximate entropy – ApEn) compared to the preferred movement pattern. However, they were unable to produce a stochastic output that mimicked that of white noise, which was the parameter established for the random condition. Morrison et al. (2007) investigated changes in COP dynamics and the muscles associated with controlling sway about the ankle (tibialis anterior, soleus) during three postural sway conditions (i.e., standing still, preferred sway, random sway). Standing still and preferred sway conditions did not differ in structure and modal frequency at the COP level; however, higher levels of irregularity in the COP dynamics, as well as decreased synchrony in AP and ML axes, were seen during the random sway condition. The relation between the lower leg muscles and COP dynamics exhibited inverse responses and were characterized by higher levels of regularity in the lower leg muscles when the COP dynamics showed higher levels of irregularity. As a result, the different structures that produce movement have to work more independently to perform the task of swaying randomly (Morrison, et al., 2007).

To further examine the ability to produce random movement in whole body motion, this study employed the most commonly utilized form of locomotion (walking) and neurologically healthy young adults walking with a highly variable (random) gait pattern. To our knowledge, the ability to intentionally generate greater variability during gait has not been investigated previously. Of interest is how the healthy neurologic system adapts to volitionally produce increases in variability and regularity while walking at non-preferred gait speeds. For the purpose of this study, the term random is used to refer to the task of moving more variably while walking. The only parameters set to establish this was the specific instruction provided to participants to make each stride different and was characterized by an increase in the amount and structure of variability compared to the preferred walking (control) condition.

The aim of this study was to identify changes in the variability and complexity of gait dynamics in neurologically healthy adults while intentionally increasing gait cycle variability (random condition) under three different speeds compared to the preferred gait pattern. It was hypothesized that: 1) Young adults can increase gait variability when instructed to do so, and 2) Gait speed will have an impact on the amount and structure of variability. Greater variability of gait will be produced during PWS and lower levels of variability will be produced at the faster speed (120% PWS).

Methods

Participants

Ten healthy young adults (1 male, 9 females; age 25.75 ± 3.96) were recruited to participate in this study. The participants completed a physical activity readiness questionnaire (PAR-Q) and a medical history questionnaire prior to participating in the study. Subjects were excluded if they reported any neurological and/or cardiovascular disorder, loss of consciousness or concussion within the last year, and/or lower limb musculoskeletal injury/surgery within 1 year of the study. All subjects provided written informed consent prior to participation in the study. The study was approved by the University Institutional Review Board and all experimental procedures complied with the guidelines.

Equipment and procedures

For each person, stride time intervals were recorded during walking on a treadmill (h/p/cosmos mercury med 4.0) with an installed pressure plate (FDM-T zebris Medical GmbH, Germany). Specific gait events were established using WinFDM-T© software (Zebris Medical GmbH, Germany). To evaluate the effects of the different walking conditions on knee joint motion, two 2D electrogoniometers (SG150, Biometrics, Inc.) were connected to two accelerometers (Delsys Trigno System, Delsys, Inc, Boston, MA). The goniometers were affixed to each leg to record knee joint flexion and extension at a sampling rate of 148 Hz. Subjects were standing while each end-block was attached to the skin using double-sided adhesive tape. The proximal end was attached in line with the greater trochanter of the femur and the distal end was attached in line with the lateral malleolus. Subjects were asked to stand still with head facing forward and knees in full extension to obtain a standing calibration prior to each gait speed condition (Biometrics, 2005; Piriyaprasarth, Morris, Winter, & Bialocerkowski, 2008).

Data collection was conducted in a single lab visit and consisted of establishing the PWS and then walking for 20 min at each of the three speeds. Upon arriving, subjects were asked to complete the PAR-Q and the Medical Health Questionnaire and provide written informed consent. They were then instructed to walk at a comfortable, self-selected speed along a 20 ft (610 cm), pressure sensitive walking surface (CIR systems Inc., Havertown, PA). Three trials were recorded and the average speed was used to establish the PWS and calculate the alternate speeds of 80% and 120% of PWS. The speeds were preferred walking speed (PWS; 4.3 ± .37 Km/h), 80% PWS (3.4 ± .30 Km/h), and 120% PWS (5.16 ± .45 Km/h). Participants rested for 10 min between each gait speed condition to eliminate any fatigue effects. All walking sessions were performed without shoes.

All participants performed a total of four 5-min time blocks that consisted of two blocks of walking referred to as control (1st and 4th) and two blocks of variable strides referred to as random (2nd and 3rd) while walking at each of the three speeds. Speeds were counterbalanced between each subject. For each gait speed participants were instructed to walk as they would normally for the first 5-min block of time. While continuing to walk, participants were then asked to "make each stride as different from the other in length and/or speed while continuing forward progress, keeping feet moving forward and eyes looking straight ahead." This instruction was provided at the beginning of the second 5-min time block and again at the beginning of the third 5-min time block. For the final time block, subjects were instructed to walk normally again. Figure 5.1 illustrates the experimental protocol and provides representative raw data for stride time at each of the three gait speeds.

A rating of perceived exertion (RPE) was obtained at 4.5 min of each 5-minute time block for each walking condition using a modified Borg, 10-point scale. The modified Borg scale ranges from 1 being "little or no exertion" to 10 being "maximal effort" (Borg, 1982, p. 380).

Each individual performed a 2 min warm-up and familiarization period on the treadmill at their PWS prior to data recording. Approximately 240 strides were collected across each 5-min time block

Data Reduction and Analyses

The dependent variables assessed include the mean, range, coefficient of variation (CV), and signal regularity of the stride time and signal regularity of knee joint motion (KJM). Stride time was defined as the time lapse between the heel strike of one foot and the subsequent heel strike of the same foot. The magnitude (or amount) of variability was defined as the CV and range. The range was established by subtracting the minimum stride time from the maximal stride time. CV was computed by dividing the standard deviation (SD) by the mean and multiplying the value by 100 (SD/mean*100) to provide a percentage of variability.

Signal Regularity

The degree of regularity or structure for both the stride time and KJM signal was assessed using Sample Entropy (SampEn). SampEn (m, r, N) is the negative natural logarithm of the conditional probability that two sequences of data similar in m points will be similar at the next point (Richman & Moorman, 2000). This analysis has been shown to be largely independent of data size and has demonstrated greater consistency with smaller data sets such as those most commonly seen for gait (Yentes, et al., 2013). The parameters (m, r) were used in accordance with other gait studies (m=2, r=0.2) (Decker, Cignetti, & Stergiou, 2012; Tochigi, Segal, Vaseenon, & Brown, 2012; Yentes, et al., 2013). Typically, higher SampEn values indicate greater irregularity (randomness) in the signal while lower SampEn values indicate a more regular signal. Changes in SampEn have been interpreted as changes in the complexity of the signal's time domain and have been used to determine the complexity of various physiological systems (Richman & Moorman, 2000; Yentes, et al., 2013).

Statistical Analysis

A Mixed General Linear Model (GLM) with repeated measures was used for analysis on each of the dependent variables. The within-subject factors were speed (3 levels) and walking condition (2 levels – control and random). Pairwise comparisons were conducted where significant main effects and interactions were observed using Bonferroni adjustments. All analyses were conducted using Statistical Analysis Software (SAS, version 11.0; SAS Institute Inc., Cary, NC, USA) with significance levels set at $p < 0.05$.

Results

Participants walked on an instrumented treadmill under 2 different conditions (2 trials of just walking, or control trials, and 2 trials of random walking) at 3 different gait speeds (PWS, 80% PWS, and 120% PWS). No significant differences were observed between the normal conditions (trial 1 & 4), and the random conditions (trial 2 & 3); therefore, the trials were collapsed to provide a single trial for each condition at each speed. Similarly, no differences between the right and left legs were seen in the majority of the dependent variables, and data were combined to reflect both legs. The one exception was the KJM SampEn value in which there was a significant right and left leg difference and this was reported accordingly.

Rates of perceived exertion (RPE)

The impact of the different walking conditions on RPE across each speed is illustrated in Figure 5.2. The results revealed significant main effects for gait speed ($F_{2, 18} = 65.34$, $p < 0.0001$) and condition ($F_{1, 9} = 67.47$, $p < 0.0001$) on RPE values. As gait speed increased, RPE was significantly higher at 120% PWS (3.97 ± 1.1) than at either the slower gait speed (2.21 ± 1.1) or PWS (2.77 ± 1.1, p's < 0.05). For the condition effect, the random walking condition elicited higher RPE values (3.52 ± 1.1, $p < 0.0001$) than normal walking (2.47 ± 0.9).

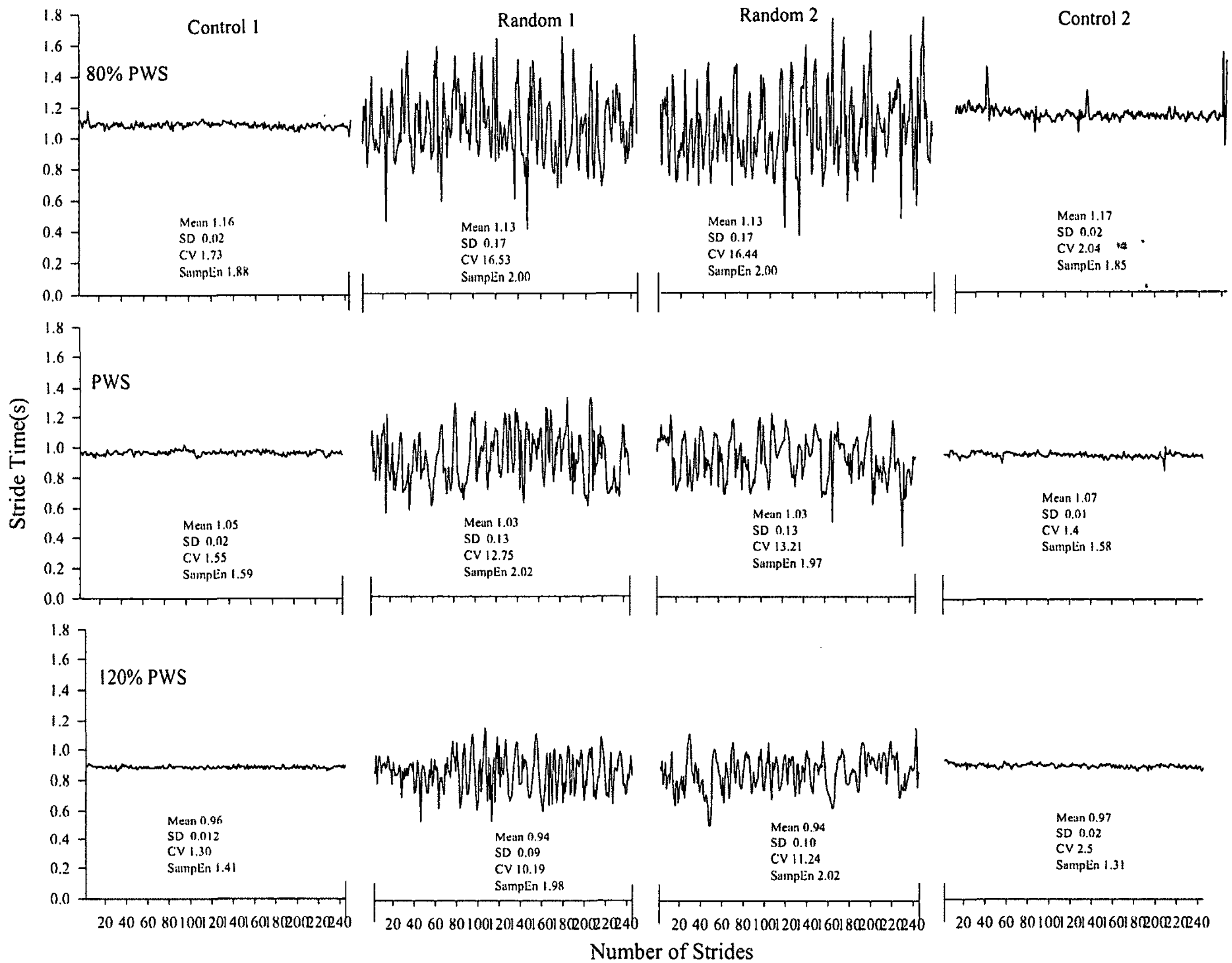

Figure 5.1. Representative time series for stride time illustrating the control and random walking conditions at each of the gait speeds (80% PWS, PWS, 120% PWS). All traces were taken from a single subject. Data from the left leg is shown as there were no right/left leg differences in stride time.

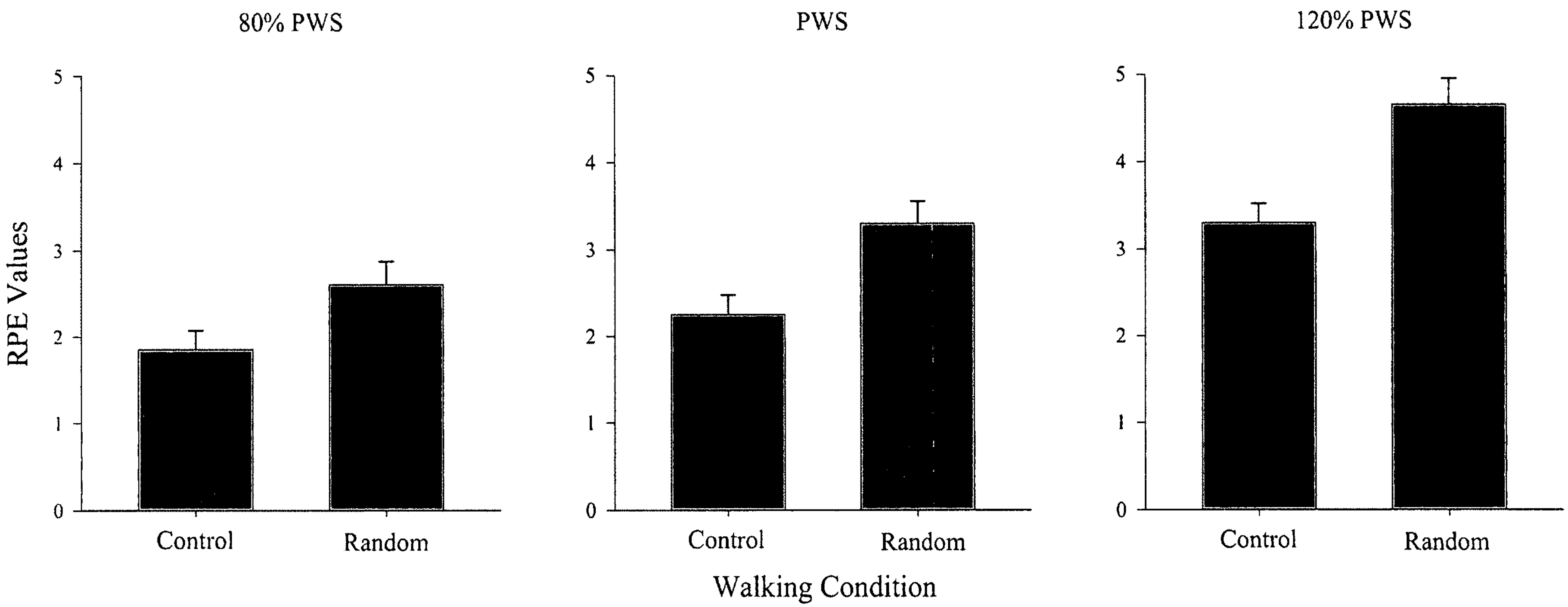

Figure 5.2. Mean and SEM of changes in the RPE as a function of the different walking conditions at each of the gait speeds. The RPE is a subjective measure used to identify how difficult a task is perceived to be. Random condition elicited a much higher RPE irrespective of gait speed than the control condition.

Changes in stride time as a result of gait speed and walking condition

The impact of gait speed and walking condition for the mean, CV, and range of stride time under each of the speeds can be seen in Figure 5.3. Significant main effects for gait speed ($F_{2,18} = 172.35, p < 0.0001$) and walking condition ($F_{1,9} = 12.2, p = 0.0068$) were seen in the mean stride time. For the speed effect, as gait speed decreased, stride time increased (*p's* < 0.0001). Across all three of the gait speeds the mean stride times during the random walking were significantly shorter than the control condition ($p < 0.05$). No significant differences were seen in mean stride times between the random and control conditions as a result of gait speed.

A significant gait speed by condition interaction for range ($F_{2,18} = 13.8, p = 0.0002$) and CV ($F_{2,18} = 7.47, p = 0.004$) of stride time. Post hoc analyses revealed that the random condition elicited a greater CV and range compared to the control across all three of the gait speeds (*p's* < 0.0001); however, greater CV and range were observed during the slower speed compared to the other speeds (*p's* < 0.05). Interestingly, the CV was similar between PWS and the faster speed in both of the walking conditions (control, random),although the slower speed elicited a greater amount of variability as reflected by the CV ($16 \pm 8\%$) under the random condition only when compared to the faster speed ($11 \pm 6\%$; *p's* < 0.05). This finding suggests that participants were able to increase the stride variability during slower walking compared to walking at the faster pace. Similarly, the slower speed elicited a significantly greater range ($0.97 \pm .45$s) than PWS ($0.72 \pm .42$s) and the faster gait speed ($0.58 \pm .25$s) during the random condition.

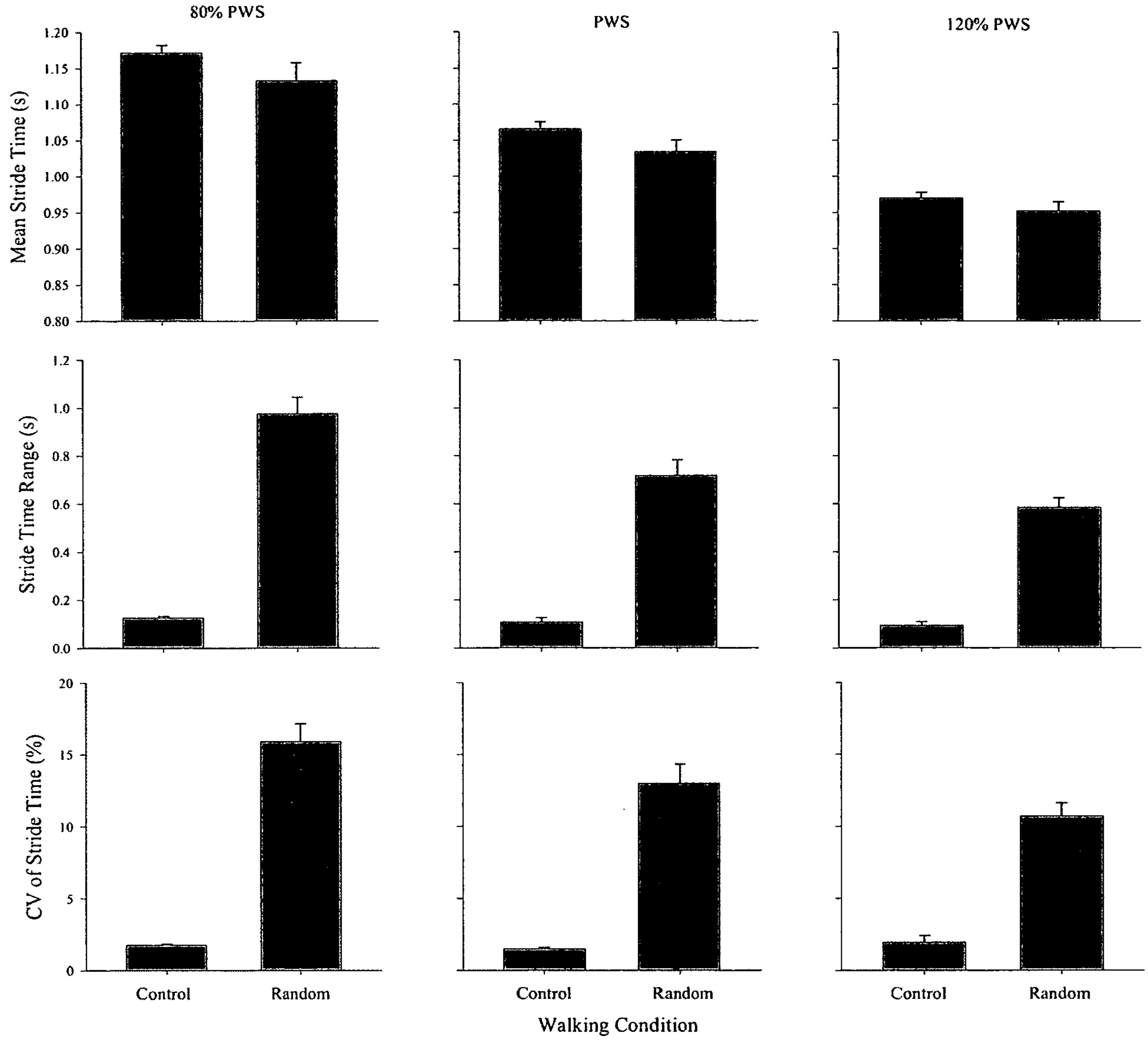

Figure 5.3. Mean (*top)*, range (*middle*), and % CV (*bottom*) for stride time as a function of the different walking conditions at each of the three gait speeds. For all three gait speeds the random walking condition resulted in lower mean stride times and greater variability compared to normal walking. Error bars indicate SEM.

Sample Entropy Analysis

Changes in signal regularity for stride time and KJM as a function of gait speed and walking condition are illustrated in Figure 5.4. A significant interaction effect for gait speed and walking condition ($F_{2, 18} = 36.70, p < 0.0001$) was revealed in SampEn of stride time. Subsequent analysis revealed that walking at PWS and 120% PWS produced higher levels of irregularity of the signal between random and control conditions that was not seen during 80% PWS. Walking at the slower gait speed elicited higher levels of irregularity (80% PWS) compared to the other speeds during the control condition (p's < .05). There was no significant effect for gait speed on structure during the random condition.

Significant main effects for gait speed ($F_{2, 18} = 23.02, p < 0.0001$), walking condition ($F_{1, 9} = 41.19, p < 0.0001$), and leg ($F_{1, 9} = 6.92, p = 0.027$) were revealed in SampEn for KJM. Subsequent analysis revealed significant differences in the time dependent structure resulting in higher SampEn at the faster speed than the other speeds (PWS, 80% PWS). No significant differences occurred between PWS and the slower speed ($p = .44$). Higher SampEn was seen in the random walking relative to the normal walking conditions (p's < 0.05). Higher SampEn was seen in the left leg compared to the right leg. There were no significant interactions between gait speed and condition.

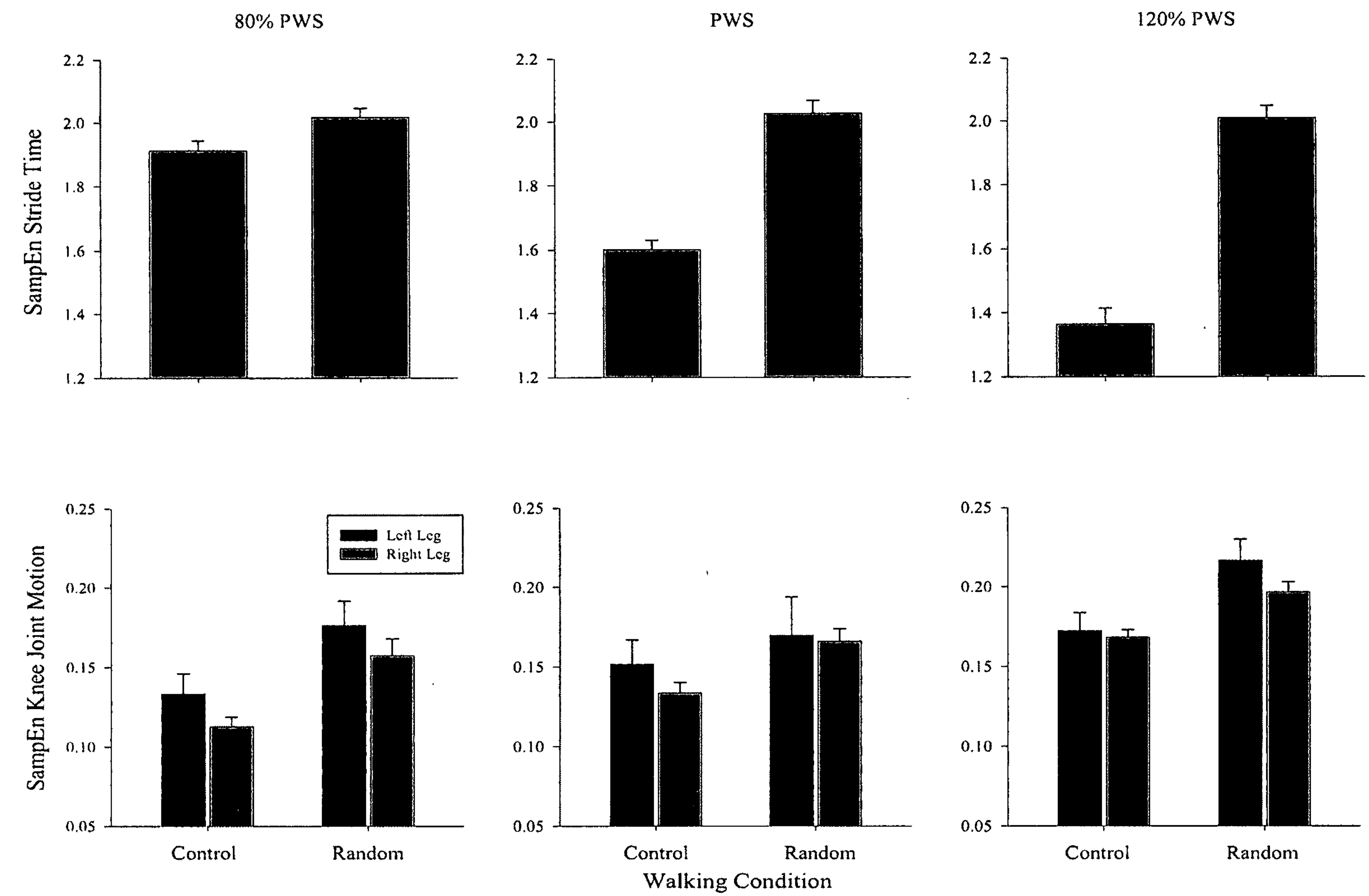

Figure 5.4. Mean and SEM of changes in Sample Entropy (SampEn) for stride time (*top)* and knee joint motion (*bottom)* as a function of the different walking conditions across each of the speeds. SampEn of stride time shows greater irregularity of the signal during random walking at both the PWS and the 120% PWS, whereas there was no difference between the conditions (random, control) at 80% PWS. Irregularity of the knee joint motion signal increased with speed.

Discussion

The main findings revealed that both gait speed and the goal of the task (random walking) impacted the gait dynamics of young adults. Gait speed had a significant impact on the amount of variability produced (as a percent of CV and range); however, it did not impact the complexity of the time series for stride time during the random condition. Interestingly, the time dependent structure of the knee joint motion had higher levels of irregularity at the faster speed. There was not an effect for gait speed at the preferred and slower paces. As expected, walking speed had a significant effect on perceived exertion, variability, and structure in the time series in both stride time and knee joint motion (KJM), as did the random walking condition. The lack of differences between the two control conditions suggests that the participants were able to quickly go from walking random their preferred pattern. These results further illustrate the adaptive behavior of the locomotor system.

Changes in gait speed either facilitated (slower speed) or constrained (faster speed) the ability to perform the task of increasing the magnitude of stride time variability during the random condition; however, participants reported that the random condition was more demanding than the normal condition irrespective of gait speed. In order to accomplish the task of random walking, participants sustained a non-preferred movement pattern for five min while simultaneously maintaining the pace of an imposed speed on a motorized treadmill. It has been suggested that maintenance of non-preferred movement patterns results in greater exertional requirements (Diedrich & Warren, 1995), which is a possible explanation for the higher RPE's reported by participants in this study. Regardless of the gait speed, participants were able to achieve the goal of making their gait more variable, as reflected in amount of variability in stride time and increased complexity in the time series for stride time and knee joint motion.

Overall, healthy young adults were able to walk randomly, as illustrated by an increase in the amount of variability and the time-dependent structure of stride time and KJM compared to the control. Gait speed was a significant factor in the amount of variability produced regardless of walking condition. During the slower gait speed, participants were able to produce greater higher levels of stride-to-stride variability whereas the faster pace elicited a lower amount of stride time variability. The impact of gait speed was not reflected in the time dependent structure of the random condition in either stride time or knee joint motion.

Gait speed and random condition effect

It has been well documented that walking at speeds slower or faster than preferred elicits changes in the temporal parameters of gait. Our findings are similar to previous research showing that walking at speeds that are faster or slower than preferred increased the stride-to-stride variability and changed the time dependent structure of the signal (Beauchet, et al., 2009; Chung & Wang, 2010; Dingwell & Marin, 2006; Jordan, et al., 2007; Jordan & Newell, 2008).

As this is the first study to examine the changes in gait dynamics when the goal of the task was to produce a random gait cycle, it was also of interest to investigate the impact that speed would have on the dynamics of gait. Preferred walking speed is a very rhythmic motion with low levels of stride to stride variability (Danion, et al., 2003; Dingwell & Marin, 2006; Jordan, et al., 2007). Therefore, it was predicted that participants would be able to produce a more variable gait pattern at their preferred walking speed compared to the non-preferred speeds as a result of the more efficient gait cycle (Holt, et al., 1995; Jordan, et al., 2007). Instead, participants in this study were found to produce a greater amount of variability (as reflected in the CV and range) at the slower gait speed compared to either PWS or the faster speed. Interestingly, gait speed did not have any impact on the structure of the signal. While the random condition revealed complexity in the signal compared to the control, gait speed did not have an impact indicating the ability to accommodate the non-preferred speeds to achieve the task. This increase in stride time variability has been a well-documented consequence of slower walking; however, it is usually accompanied with higher levels of signal irregularity as well (Beauchet, et al., 2009; Hausdorff, 2004; Jordan, et al., 2007).

A possible explanation for the varied results between the amount and structure of the variability may be the analyses performed. The CV and range are generated by taking the averages over the entire trial (for this experiment it was five min and approximately 240 strides) and producing a single value that is used to represent the entire time series, thereby providing a snapshot overview of the entire walking session. However, the use of SampEn provides us with the time dependent changes within the time series and a measure of how similar or different the gait pattern was over the entire 5-min period (Buzzi, et al., 2003; Dingwell & Cusumano, 2000; Hausdorff, et al., 1996). Consequently, it was more difficult to increase the amount of stride time variability at the faster speed. The difficulty of walking fast and performing the goal of the task was confirmed by the reported RPEs. The difficulty in producing random movement during the faster gait speed may be explained by the constraint placed upon the requirement to vary movement and the necessity to coordinate those movements at a much quicker pace (Jordan, et al., 2007).

The only left and right leg difference was revealed in the KJM structure of the signal. Higher levels of irregularity were elicited in the left leg than in the right across all conditions. Locomotion requires propulsion to move the body forward and control to stabilize the body during single leg stance. A possible explanation for the findings could be that the left leg is responsible for medio-lateral control and the right leg is responsible for propulsion (Sadeghi, Allard, Prince, & Labelle, 2000). As right/left limb differences were not present in the other variables and laterality of the participants was not determined we can only speculate as to the cause of this finding.

Conclusions

In summary, the present findings reveal that healthy young adults were able to increase the amount of variability and complexity in gait dynamics as instructed, but perceived it as being more physically demanding to achieve regardless of gait speed. While speed impacted gait dynamics during the control condition, the results revealed in the random condition provides us with greater insight into the influence of intent on the motor control processes of locomotion. The amount of variability (measured by the CV and range) produced during the random condition was directly related to the gait speed. The slower speed required less exertional demands than the other speeds and enabled higher CV and stride time range in the random condition compared to both PWS and the faster gait speed. The amount of variability during the faster gait speed was lower (CV, range) compared to the other speeds and coupled with the higher RPEs indicates that it was more difficult to accomplish. However, the time dependent structure revealed a different pattern in response to the random condition and the impact of gait speed. The complexity of the stride time during the random condition was not impacted by gait speed however, higher levels of complexity were elicited in knee joint motion during random walking at 120% PWS that were not present in the other gait speeds. Ultimately, young adults were able to adapt their gait to increase the structure of the stride time and knee joint motion and easily change their gait pattern from variable to a preferred pattern seamlessly irrespective of gait speed.

CHAPTER VI

CONCLUSIONS

Overall, the results of these three experiments reveal that the healthy postural control system is able to adapt relatively quickly to the task demands. In the first experiment we aimed to identify the time course of changes in postural motion as a result of fast walking during the course of a 35-min session. We believed that the fastest gait speed would induce the greatest amount of changes in postural motion compared to the other two speeds (PWS, 120% PWS), but that these changes would be transient returning to baseline levels rather quickly. The results supported our hypotheses as walking 40% faster than preferred elicited the greatest changes in postural motion compared to the baseline assessments than the other speeds. As predicted, the healthy sensorimotor system was able to quickly adapt to the postural changes as a result of the fast walking and within 5-10 min postural motion was back to baseline levels, reaching a steady state in postural motion by the end of the session. This adaptive postural response to the continuous perturbation was present, given that the measures of exertion (HR, RPE) continued to rise in response to the increased physical activity for the remainder of the task. Individuals were able to quickly adapt to the effect fast walking had on balance.

In the second experiment, we explored how stride-to-stride variability changed over successive time blocks while walking at speeds slower than preferred. We predicted that walking at a slower than preferred pace would elicit higher levels of variability and signal regularity compared to PWS, but that over time the amount of variability and regularity of the signal would decline. Our hypotheses were supported by the results of the study. Our findings revealed that walking at 20% slower than preferred elicited a greater amount of stride-to-stride variability and higher levels of regularity than walking at preferred speed and 10% slower than preferred. Even at the slowest walking speed levels of variability were relatively low, indicating that the task did not produce a strong disruption to the gait pattern. However, with continued walking at the slowest pace (80% PWS), the stride-to-stride variability decreased and signal structure increased after 10 min becoming more similar to the values at the other speeds. Conversely, the results of walking at 10% slower than preferred did not change the time course of stride-to-stride variability as walking at the slowest speed did. Participants were able to adjust their stride to accommodate a slower speed, but the time course of adaptation was dependent upon how much slower the pace was from preferred. Walking only 10% slower than preferred did not increase variability when compared to PWS and participants quickly adjusted, particularly in comparison to the slowest speed. These results highlight that the healthy locomotor system can adapt to walking at speeds that are 20% slower than preferred over time.

In our final experiment, our aim was to compare the gait dynamics (stride time and knee joint motion) in young adults while intentionally increasing stride cycle variability (random) during walking at three different gait speeds. It was hypothesized that the participants would be able to increase variability as instructed regardless of the gait speed. It was also hypothesized that walking at PWS would allow participants to generate greater amounts of variability (per CV and range). The results supported our first hypothesis that when instructed to do so, healthy young adults would be able to increase stride cycle variability. Our second hypothesis was not supported as the slower gait speed afforded the ability to produce higher amounts of variability than the other gait speeds. The results of this study revealed that when the goal of the task was to make the stride cycle more random, young adults were able to perform the task, although it was reportedly (higher RPE's) more demanding than the control condition (just walking). The amount of variability during the random condition was strongly related to gait speed. When walking at 80% PWS the amount of stride-to-stride variability was greater than walking at the preferred and faster speed. Walking at the faster speed elicited the lowest amount of variability than PWS and higher levels of irregularity in knee joint motion. These findings, coupled with the reports of higher RPE during the faster gait speed, suggest that the combination of the gait speed and the task was more demanding relative to the other speeds. This was not the case in the structure for the stride time however. Higher levels of irregularity were produced in stride time during the random condition irrespective of gait speed. The ability to increase variability of gait was influenced by gait speed. The walking speed impacted the amount of variability produced in the stride time but the structure of the random condition was not affected by the gait speed.

Overall, the findings of these three experiments revealed that simple changes to a common task can influence the movement patterns of healthy young adults. However, the healthy neuromotor system is quickly able to adapt to the task demands to control postural sway motion induced by walking at a faster pace, reduce stride time variability and increase complexity while walking at non-preferred speeds, and intentionally increase variability in gait cycle.

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APPENDIX I

PUBLISHED WORKS

1. Thomas, K. S., VanLunen, B. L., & Morrison, S. (2013). Changes in postural sway as a function of prolonged walking. *Eur J Appl Physiol, 113*(2), 497-508.

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