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VARIABILITY AND LONG RANGE CORRELATIONS IN HUMAN WALKING AND RUNNING

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Abstract

Fluctuations in the stride interval of human walking contain long range correlations that decay in a fractal-like manner (Hausdorff et al., 1995). Using Detrended Fluctuation Analysis (DFA), this thesis examines the structure of variability of the gait cycle in human locomotion. Three experiments were carried out to address several issues: 1) are long range correlations present in the fluctuations of the running gait, 2) are long range correlations present in gait variables other than the stride interval, 3) what is the influence of speed of locomotion on the scaling behavior of the gait cycle fluctuations, and, 4) what is the relationship between stability and long range correlations? Experiment 1 examines the fluctuations in a range of kinematic and kinetic gait cycle variables in walking at 60 through to 140% of preferred walking speed, while Experiment 2 investigates gait cycle fluctuations in running from 80 to 120% of preferred running speed. The results reveal the presence of fractal-like scaling behavior in all variables investigated. For many of both the walking and running gait variables, long range correlations follow a U-shaped function with speed and were minimized at preferred speeds of locomotion. Thus, at preferred speeds, there are a larger number of timescales present in the motor out put which is suggestive of greater adaptability. Experiment 3 examines the relationship between local dynamic stability, stability of the movement pattern, and the long range correlations of the gait cycle while walking and running at speeds close to preferred transition speeds. The results suggest that the scaling behavior of gait cycle fluctuations relates to the stability of the gait cycle. Collectively the findings indicate that DFA is revealing about the number of degrees of freedom available under given conditions, or

conversely the degree of constraint that results from a set of conditions. An alternative but not mutually exclusive possibility is that the correlations are related to the degree of active control associated with locomotion under different circumstances. Thus, as the speed of locomotion moves increasingly away from preferred speeds, structure is introduced to the variability as a result of: a) increasing constraints, b) decreasing degrees of freedom and c) increasing levels of control.

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Chapter 1

Introduction

Processes with long-term correlations, or 1/f-like processes, have been observed in a large number of different systems ranging from the fields of physics to sociology (Kaulakys & Meskauskas, 1998). These types of processes are ubiquitous in nature yet they are not easily explained (Bak, 1996; Wagenmakers, Farrell, & Ratcliff, 2004). Of particular interest to movement scientists have been the long range correlations found in human movement time series such as the inter-tap interval in synchronization studies (e.g., Chen, Ding & Kelso, 1997; Yoshinaga, Miyazima & Mitake, 2000), center of pressure trajectories during standing (e.g., Duarte & Zatsiorsky, 2000, 2001), and the inter-stride interval in human locomotion (e.g., Hausdorff, Peng, Ladin, Wei, & Goldberger, 1995; Hausdorff et al., 1996, West & Griffin, 1998, 1999; West & Scarfetta, 2003).

The primary focus of this dissertation is to examine the fractal nature of human locomotion and how this relates to preferred movement patterns. Specifically, a variety of kinematic and kinetic gait cycle variables will be examined with a particular focus being placed on the structure of the variability of the time series associated with these variables. This theoretical and experimental approach is of interest for several reasons. Firstly, recent studies have shown that there are long range correlations present in the inter-stride interval of human locomotion (e.g., Hausdorff et al., 1995; Hausdorff et al., 1996). The distributions associated with stride interval time series are not normal, indicating that previous research relying on normal distribution statistics needs to be revisited. Secondly, the long range correlations present in the stride interval time series suggest that there may be systematic changes as a function of walking speed. However, this phenomenon has only been investigated over a small range of walking speeds and the implications of the changes recorded for the control of locomotion are unclear. We are, therefore, interested in extending this research on 1/f processes in locomotion to cover the kinematic and kinetic variability of running and walking over a wide range of gait speeds. Thirdly, the structure contained within the stride interval time series may also be present both in other time series that can be derived from foot falls, (for example, the stride length and step impulse time series) as well as at other levels of the locomotor apparatus (for example, joint angles and limb trajectories). A thorough investigation and description of these time series may provide us with further insight into the control and coordination of locomotion in humans.

A subsidiary goal of the thesis is to investigate the variability of gait cycle parameters and how this changes during the walk to run transition. Several studies have characterized the walk to run transition as a non-equilibrium phase transition (Diedrich & Warren, 1995; Kao, Ringenbach, & Martin, 2003). These studies focused on the amount of variability present at the joint level of hip-ankle and hip-knee interactions. This dissertation will examine the postulation that the transition between walking and running reflects the transition between 2 attractors and the associated loss of stability in this process. In addition to the traditional measures of the amount of variability, the structure of variability in a range of kinematic and kinetic gait cycle parameters will be assessed as a means to evaluate the hypothesis that gait transitions are associated with a loss of stability. The remainder of this chapter of the thesis is divided into 7 sections and is given over to a brief statement of the core issues in locomotion and movement variability.

1.1 Fundamental Characteristics of Walking and Running

Human gait typically takes the form of either walking or running, the latter being distinguished from the former by a flight phase during which neither foot is in contact with the ground. The duty factor for each foot is the proportion of the gait cycle for which that foot is in contact with the ground. For walking the duty factor is greater than 0.5, indicating that for some part of the gait cycle both feet are on the ground at the same time. In contrast, the duty factor for running is less that 0.5, meaning that for portions of the gait cycle neither foot is on the ground. The duty factor for both walking and running decreases with increasing speed (Alexander, 1984).

Walking and running can also be distinguished by way of center of mass and energy changes throughout the gait cycle. During walking, the kinetic energy of the center of mass is high while potential energy is low and vice versa. For example during the double support phase of the gait cycle, when the center of mass is at its lowest, the potential (or gravitational) energy of the center of mass is low whereas its kinetic energy is high. During the middle of stance, when the center of mass is at its highest point, the potential energy of the center of mass reaches its peak while kinetic energy is at its lowest. Margaria (1976) compared walking to an egg rolling end over end on a flat surface, as its gravitational potential energy increases, its kinetic energy decreases (and vice versa). Conversely, during running, kinetic and potential energy are both either high or low at the same time. Thus, the modeling of walking is typically treated as an inverted pendulum, with the body being a fixed point mass on top of the upside down pendulum (e.g., Alexander, 1976), where as running has been modeled using a mass-spring approach (e.g., Blickhan, 1989; McMahon & Cheng, 1990). The within-subject variability of these fundamental characteristics of walking and running is rarely considered.

1.2 The Influence of Speed on Parameters of the Gait Cycle

In general, step and stride time decrease with increasing speed while step and stride lengths and rates increase (Grillner, Halbertsma, Nilsson & Thorstensson, 1979; Hirokawa, 1989; Nilsson & Thorstensson, 1987; Oberg, Karsznia & Oberg, 1993). Grillner et al. (1979) showed a decrease in duration of step cycle and support phase of walking and running with increased gait speed. The relation between durations of both the step cycle and support phase with velocity correlated well with a power function for all subjects suggesting a non-linear change with speed.

Nilsson and Thorstensson (1987) investigated walking and running in human subjects across a range of speeds and step frequencies. For both walking and running at preferred step frequency, mean stride cycle duration decreased linearly with increasing speed and stride length increased with increasing speed. Inspection of their Figure 4a suggests that the function for stride length versus velocity for walking is curvilinear whereas in the case of running there is a linear increase with speed, up until higher speeds (>6m/s) at which point stride length remains approximately constant. Hirokawa (1989) investigated walking at preferred, slow and fast speeds in 53 men and 39 women and found that there was an increase in step length and cadence proportional to increases in walking speed. In addition, it was noted that males tend to increase step length to a greater extent, whereas females tend to increase cadence to a greater extent with increasing velocity.

These findings are supported by Oberg et al. (1993) who also showed that step length was lower and step rate higher in women as compared with men at slow, normal and fast speeds. While this group did not specifically analyze the effect of walking speed on step frequency and step length, their data indicate that both variables increase with increasing walking speed. LaFiandra, Wagenaar, Holt and Obusek (2003) showed that an increase in walking speed (range = 0.6-1.6m/s) leads to an increase in pelvic, trunk, and thoracic rotation, as well as increased hip excursion, all of which would contribute to increased step length with increasing gait speed. Lastly, Keller et al. (1996) investigated changes in ground reaction force with speed and found that vertical ground reaction force increased linearly with increased walking and running speed in both female and male subjects.

It has been shown by several different researchers that people tend to naturally walk at a speed that minimizes metabolic energy expenditure (e.g., Bobbert, 1960; Margaria, 1976; Zarrugh, Todd, & Ralston, 1974). However, there appears to be no economically preferred running speed, rather the energy cost of running per unit mass remains essentially constant over increased speeds for a given distance (Margaria, Cerretelli, Aghemo, & Sassi, 1963). These findings for walking may be explained in terms of the pendular behavior of the lower limb during walking (Holt, Hamill, & Andres, 1990, 1991). A pendulum has a natural or preferred frequency at which it will swing that is sometimes called the eigenfrequency. This eigenfrequency is dependent on both the length and mass of the pendulum, as well as the distribution of the mass along the length of the pendulum. Pendulum models are commonly employed in the study of human walking and are particularly relevant for the swing phase of gait (e.g., Mochon & McMahon, 1980). It has been shown that the predicted eigenfrequency of a limb does not differ significantly from the preferred or naturally selected step frequency during walking (Holt et al., 1990,1991) with the preferred frequency typically considered to be both most stable frequency (Kugler & Turvey, 1987).

According to Winter (1980), the power for walking is supplied rhythmically with temporal consistency. As such, Holt et al. (1990) developed a force-driven harmonic oscillator model to investigate whether preferred frequency of walking could be predicted by the least amount of energy required to drive a harmonic oscillator. A force-driven harmonic oscillator requires a temporally consistent rhythmic energy supply (or a periodic forcing function). Additionally, for a given force-driven harmonic oscillator, there is a resonant frequency which requires the least amount of force to maintain its oscillation. Based on the length of their particiants legs, Holt et al. (1990) predicted a preferred period for walking. They used two slightly different methods of prediction, the only difference between the two being that in the 2nd method a gravitational constant was multiplied by 2 (as suggested by Kugler & Turvey, 1987 for quadruped locomotion). The first method of prediction consistently overestimated the preferred walking frequency, whereas there was no statistical difference between the actual and predicted periods using the 2nd method. There was a small (statistically not significant) underestimation of preferred walking frequency by the 2nd method that was attributed to treating the limb as a rigid body of fixed length. On the whole, these results suggest that in walking, the metabolic cost is minimized by taking advantage of the dynamical properties of the leg which in turn reduces the required muscle force.

Following this study, Holt et al., (1991) investigated the metabolic cost of walking at the preferred period, the predicted preferred period and periods above and below than these periods. When speed was kept constant, there was no significant difference between the preferred and predicted period of oscillation. These periods also coincided with minimum metabolic expenditure, whereas at periods above and below preferred and predicted, the metabolic cost of walking increased. These finding support the idea of a preferred walking speed which is related to the dynamic properties of the leg.

In contrast to walking, the energetic cost of running remains constant with increasing speed (Margaria et al., 1963). It has been shown that there is a linear relationship between both the rate of oxygen consumption and speed and the metabolic transport cost and speed (Kram & Taylor, 1990; Margaria et al., 1963). In contrast to these findings are those of Daniels (2002) who showed that at slow speeds (i.e. less than ~2m/s) there is a decrease in both the transport cost and mass specific rate of oxygen consumption with increasing running speed. Thus, it seems that running at very low speeds requires more energy than running at relatively higher speeds. The most likely explanation for the differences in the energetic cost functions for walking and running (at higher speeds at least) is that during running it is possible to exploit the passive elastic components of the lower limb (such as the Achilles tendon and plantar fascia) to a greater extent (Alexander, 1991).

1.3 Gait Transitions

There is a considerable amount of debate in the literature regarding the causes of or triggers for the walk to run transition which is often considered in terms of an optimization problem. The main accounts of the cause of gait transition triggers include a) the mechanical limit to the speed of walking (Alexander, 1976, 1984); b) minimization of the metabolic cost of locomotion (e.g., Margaria 1976; Hoyt & Taylor, 1981), c) minimization of mechanical stress (e.g., Farley & Taylor, 1991; Hreljac, 1995), and d) that gait transitions can be considered to be non-equilibrium phase transitions (e.g., Diedrich & Warren, 1995, 1998).

Alexander (1976) showed that there is a speed it is not possible to walk above, assuming that the hip joint moves on the arc of a circle centered on the foot. This model shows that walking is not possible as speeds above $(gL)^{1/2}$ where g is gravity and L is the length of the leg. So for a leg length of 0.8m and gravity of $9.8m/s^2$, the maximum walking speed is around 2.8m/s. Race walkers are able to walk considerably faster than this as they rotate their pelvis to a greater degree, thus the assumption of the hip rotating on the arc of a circle is violated (Alexander, 1984). However, studies show that the preferred transition speed (PTS) from walking to running is closer to 2m/s (e.g., Hreljac, 1995).

Thus, it has been assumed that there is some variable that is being optimized that can account for the walk to run transition. One such variable is energy expenditure. It has been shown that the metabolic cost of walking at speeds higher than that of the transition speed is greater than that of running (for example Hoyt & Taylor 1981; Zarrugh et al., 1974). However, it has also been noted that the PTS is actually lower than is metabolically optimal (Hreljac 1993, 1995). Thus we can at least conclude that there are factors other than energy expenditure that the locomotor system is constrained to optimize.

Another possible candidate for optimization is some type of kinematic variable. One problem with this is identifying which of the many kinematic variables to choose (Winter, 1983). Using three criteria, Hreljac (1995) was able to select four candidate kinematic variables, which with the addition of a 4^{th} criterion was narrowed down to more or less one variable – angular acceleration of the ankle. During walking at high speeds there was a much larger acceleration of the ankle as compared with running at the same speed. It was noted that large amounts of muscle activity were required to dorsiflex the foot around toe off to avoid dragging the toes on the ground. Not only did subjects report discomfort in these muscles but it also was likely that the muscles were working fairly close to their maximum capacity. Thus comfort, minimization of unnecessary muscle stress (and therefore likelihood of injury) may also be factors in the walk to run transition (Hreljac, 1995).

Studies by Farley and Taylor (1991) support this contention. They showed that horses switch from a trot to a gallop at speeds where it is energetically more efficient to trot, but at which the peak force on the muscles tendons and bones is significantly reduced in galloping as compared with trotting. Furthermore, when the horses carried weights, they made the transition at a lower speed but at the same level of peak force. It seems likely that prevention of injury would be the main motive for this occurrence. However, this explanation for the walk to run (W-R) transition while compelling, fails to account for the run to walk (R-W) transition. Prilutsky and Gregor (2001) hypothesized that muscle activation played a role in determining the PTS. They were able to show that there were higher levels of activation in the swing related muscles tibialis anterior, biceps femoris (long head) and the rectus femoris during walking as compared with running at preferred running speeds. Additionally during running at preferred walking speeds there was a relatively higher amount of support-related activation of extensor muscles during the stance phase. Thus, it seems likely that the W-R transition may be related to excessive activation of swing related muscles while the R-W transition maybe be related to excessive activation of stance related muscles. While there are constraints on the speed that it is possible to walk (Alexander 1976), it appears that both musculoskeletal factors and energy conservation play a role in determining the PTS.

Lastly, it has been suggested by a number of authors that gait transitions take the form of non-equilibrium phase transitions (e.g., Brisswalter & Mottet, 1996; Diedrich & Warren, 1995, 1998; Kao, Ringenbach & Martin, 2003; Seay, Haddad, van Emmerik, & Hamill, 2006; Schoner, Jiang & Kelso, 1990). The primary focus of this research is on the variability associated with gait transitions. Thus, while the hypothesis that gait transitions take the form of non-equilibrium phase transitions is not pursued in this dissertation, these studies are relevant to the experiments carried out in this dissertation and will be discussed in more detail in latter sections of the dissertation.

1.4 Traditional Approaches to the Study of Movement Variability

Variability in human movement has historically been equated with noise in the sensorimotor apparatus (Newell & Corcos, 1993; Newell & Slifkin, 1998). This is in part due to information processing models of motor control (e.g., Fitts 1954), that have their roots in Shannon and Weaver's (1949) information theory. Under this model, signals are thought to be composed of information, or the desired movement, with noise (presumably white Gaussian noise) or "variability" added. Additionally, the fact that the invariance of movement has traditionally held more interest to scientists than movement variance contributes to the over all assumption that variability in human movement is a (white) noise process which in an ideal setting would be minimized or eliminated altogether (Newell & Slifkin, 1998). Traditional measures of movement variability, such as the standard deviation, coefficient of variation and root mean square error provide information regarding the *amount* of variability, however, they do not reflect the sequential dependencies in measures of variability over time.

1.5 The Fractal Nature of Movement Variability

In order to investigate whether or not variability in human movement is a white noise process, or some other kind of deterministic/stochastic process, it is necessary to employ measures that capture the pattern in these time dependent variations. For example, techniques such as approximate entropy (ApEn – Pincus, 1991), spectral analysis (e.g. Hausdorff et al., 1995) and detrended fluctuation analysis (DFA - Peng et al., 1994) provide information regarding regularity, spectral content and time dependent correlations that may (or may not) be present in the time series.

Increasingly, with the use of techniques such as these, it is being recognized that many biological signals contain sequential dependencies that can be described by non-white noise processes such as Brownian motion, colored noise and fractals (e.g., Goldberger, Rigney, & West, 1990; Hausdorff et al., 1995; Lipsitz & Goldberger, 1992).

The term fractal was coined by Mandelbrot in 1975 to describe processes which display self similarity, have no single time scale and cannot be described using traditional descriptive statistics (i.e. the mean and standard deviation do not provide a stable description of a fractal distribution). The essential feature of fractals is that at ever higher levels of magnification, new details are revealed. The new details look the same as the details seen at lower resolutions – the structures at different time scales (or at different spatial scales) are related to each other in a statistical sense. Specifically, statistical self similarity means that the statistical properties of the process of interest (in this case variance) measured at one resolution (time scale) are proportional to the statistical properties of the same process measured at a coarser resolution (Bassingthwaighte, Liebovitch, & West, 1994).

Because of the self similarity and scaling, the statistical properties of fractal processes are not the same as those of non-fractal processes. There is no single "true value" that the measurements will converge on - the measured value will be determined by the resolution at which it is measured and will change with every resolution examined. The mean will either increase or decrease as larger amounts of data are considered, the direction of change will depend on the relative contribution of the largest elements of the fractal. As increasing amounts of data are considered, the standard deviation will tend to increase (rather than decrease as would happen in non-fractals) due to an ever increasing number of small fluctuations. The mean and standard deviation do not provide a stable description in fractal processes as a whole, but can be used to characterize how the different scales of the fractal relate to each other. In the case of stride interval variability, the variance of the stride interval time series increases with increased time series length, thus the time series as a whole can be characterized by how the variance depends on the number of strides measured.

Research using DFA has shown that correlations exist in the inter-beat interval time series of heart rate (Peng et al., 1993). These correlations change with both aging and disease such that a more regular inter-beat interval in heart rate data is indicative of an unhealthy or damaged heart (Peng et al., 1993). Gilden, Thornton and Mallon (1995) have shown that the time series of errors associated with target interval replication fluctuate both temporally and spatially in a 1/f-like fashion. Chen et al. (1997) also reported a 1/f-like process exhibited by human synchronization errors in a finger tapping task. It is generally accepted that 1/f type processes may provide us with insight into the understanding of how movement is coordinated and controlled (Bak, 1996; Chen et al., 1997; Gilden et al., 1995; Wagenmakers et al, 2004), including human locomotion (Hausdorff et al., 1995; Hausdorff et al., 1996; West & Griffin, 1998, 1999).

1.6 Treadmill vs Over-ground Locomotion

As the proposed experiments incorporate the use of a treadmill for collection of locomotion data, it is worthwhile to discuss briefly the differences between treadmill and over-ground walking and running, particularly with reference to changes in variability. While van Ingen Schenau (1980) has demonstrated mathematically that there is no mechanical difference between locomotion on a treadmill versus over-ground when the mechanical variables are described with respect to the surface on which the subject locomotes, studies have shown significant differences for a variety of gait measures.

One of the first comprehensive studies to investigate the differences between over-ground and treadmill running was that of Nelson, Dillman, Lagasse and Bickett, (1972). In this study, the running of 16 experienced runners was examined at three different speeds (approximately 12, 17.5 and 23km/hr) and three different slopes (horizontal, up hill 10%, down hill 10%). It was shown that stride length increased and stride rate decreased during treadmill running at the highest speed for both the horizontal and up hill conditions. In addition, the time of non-support during up hill running at the two slower speeds was reduced in treadmill running. Vertical velocity and the variance of both vertical and horizontal velocities were reduced for all subjects while running on a treadmill as compared with over-ground running. It was suggested that two modifications of running occur on a treadmill as compared to over-ground. Firstly, that the foot tends to be placed further in front of the subject's center of gravity. Secondly, in order to get the foot further in front of the center of gravity while maintaining adequate stride rate, the recovery and heel strike occur more rapidly than in over-ground running. Hence it can be concluded that there are slight differences in running over-ground versus running on a treadmill.

It should be noted that the subjects in this study had a very limited amount of experience running on the treadmill before data collection occurred. Since the publication of this study, it has been shown that subjects may need up to one hour to fully accommodate to running on a treadmill (Wall & Charteris, 1980, 1981). Thus, it is possible that the differences between running on a treadmill versus running over-ground may be minimized given adequate experience with treadmill running.

In a cinematographic analysis of 24 experienced treadmill runners, Elliot and Blanksby (1972) showed that stride length decreased, stride rate increased, and the flight phase decreased during treadmill versus over-ground running. On the other hand, White, Yack, Tucker and Lin (1998) showed no significant difference in walking speed, stride length, cadence and patterns of vertical ground reaction force for both normal slow and fast walking. Overall, they concluded that the phasic pattern of vertical ground reaction force could be considered the same for treadmill and over-ground walking. While there was no explicit consideration of variability of the gait parameters investigated in this study, examination of the standard deviations for each measure indicates that variability is slightly less for force measures and slightly higher for timing measures for treadmill versus over-ground walking.

Arsenault, Winter and Martiniuk (1986) investigated EMG patterns in the lower extremity of 8 subjects and found the EMG profiles were similar between treadmill and over-ground trials. EMG amplitudes were slightly larger and less variable during treadmill locomotion, however, it was concluded that the profiles were similar enough that the treadmill can be considered as a valid lab instrument for the study of gait. Nigg, de Boer and Fisher (1995) found substantial but inconsistent differences in lower extremity kinematics variables during treadmill and over-ground walking and advised caution when extrapolating findings from the treadmill to free locomotion.

Alton, Baldey, Caplin and Morrissey (1998) measured both temporal gait variables and leg joint kinematics in males and females walking on a treadmill and over-ground at their preferred speeds. In females the only difference between treadmill and over-ground walking was an increase in maximum hip flexion angle with treadmill walking. In males the only difference was an increase in cadence and maximum knee flexion angle for treadmill walking. When the results for both female and male subjects were combined, an increase in hip range of motion, maximum hip flexion and cadence as well as a decrease in stance time were observed for treadmill running. Again, standard deviations were not explicitly considered, but these values appear to be similar for both treadmill and over-ground walking. As with Nelson et al. (1972) subjects were given very little time to become familiar with walking on the treadmill (around 3 min practice was given) thus differences between the two conditions may be due in part to lack of experience with walking on a treadmill.

Finally, using nonlinear time series analysis techniques, Dingwell, Cusumano, Cavanagh & Sternad (2001), investigated the kinematic variability and dynamic stability of walking on a treadmill versus walking over-ground. Walking variability was quantified using the standard deviation of stride time. Local dynamic stability (i.e. the system's sensitivity to very small local perturbations, such as the natural fluctuations in the stride interval time series seen in normal walking) was quantified using state space reconstruction and Lyapunov exponents. It was shown that walking on a treadmill is associated with significantly <u>reduced</u> variability (particularly in the lower extremity) and significantly <u>greater</u> local dynamic stability than over-ground walking. Thus, the authors conclude that the use of motorized treadmills for the study of neuromotor control of walking where variability and stability are being considered may lead to misleading inferences about over-ground walking.

1.7 Organization of the Dissertation

The problem that this dissertation focuses on is that of the structure of variability during the gait cycle and how this changes with different gaits (walking and running), different speeds (as percent of preferred walking or running speed) and the transition between walking and running. While fractal distributions are extremely common in biological systems, often the physiological mechanisms underlying these distributions are unclear. In order to better understand the underlying mechanisms of the basic gaits of human locomotion, a more full description of the range of speeds and gaits over which these long range correlations occur is essential.

The following chapter provides a comprehensive review of the literature associated with variability and locomotion and concludes with future research directions. Chapters three and four contain Experiments 1 and 2. These experiments are designed to test the hypothesis that there is a functional relationship between the strength of long range correlations and gait speed that will reflect the degree of adaptability in system organization, with Experiment 1 focusing on the walking gait and Experiment 2 on the running gait. It is predicted that at preferred gait speeds there will be a reduction in the constraints associated with the task of locomotion that will be reflected in a reduction in the strength of the long range correlations in a variety of gait parameter time series at the preferred speed of locomotion.

Experiment 3, chapter five, focuses on gait transitions and seeks to answer two primary questions. Firstly, is there a loss of stability associated with gait transitions, as is suggested by studies that have treated gait transitions as non-equilibrium phase transitions (e.g. Diedrich & Warren, 1995), if so, is this reflected in the strength of the long range correlations? Secondly, how do the long range correlations at the different levels of measurement (for example head vs. ankle) relate to each other and is there a particular level at which they remain constant with respect to changes in speed? With regard to the first question, it is hypothesized that there will be a loss of stability in the region of the transition from walking to running (and vice versa) and that this will be reflected in both an increase in the rate of divergence of maximum finite time Lyapunov exponents and an increase in the strength of the DFA scaling exponent. The increase in the strength of the long range correlations would be indicative of an increase in the level of control required at these relatively more unstable speeds. With regard to the second question, it is hypothesized that the pattern of results from DFA will not be identical across the different levels of measurement. In particular it is hypothesized that the DFA of vertical head displacement will be particularly by speed in that it will be subject to increasing active control to maintain stability. Lastly, chapter 6 provides an overall discussion and conclusions for the experimental chapters.

Chapter 2

Variability of Human Locomotion as a Function of Speed

Human locomotion, even over a level, smooth surface, is a complex task requiring the coordination of a large number of muscles acting over a collective of joints. The neuromuscular system faces the challenge of controlling the center of mass against large gravitational and forward momentum forces when much of the time the body is supported by only one limb (Winter 1983). In spite of this challenging control problem, the gait cycle is subject only to a very low level of variability (Hausdorff et al., 1996; Patla, 1985; Winter, 1984). This finding is generally taken to indicate that subjects have learned a very repeatable kinematic pattern (Winter, 1984), with the increases in variability assumed to be indicative of problems in the control of locomotion (e.g., Gabell & Nayak, 1984; Owings & Grabiner 2004).

The variability of the gait cycle is increasingly being examined in order to better understand the mechanics and control of human locomotion (e.g., Beauchet, Dubost, Herrmann & Kressig; 2005; Dingwell & Marin 2006; Frenkel-Toledo et al., 2005; Heiderscheit, 2000; Li, Haddad & Hamill, 2005; Masani, Kouzaki & Fukunaga, 2002; Sekiya, Nagasaki, Ito & Furuna, 1997). While there have been a significant number of studies focusing on variability of gait cycle parameters (e.g., stride interval, step frequency, ground reaction force etc), in many cases, these studies have contradictory results. There is a lack of comparison of different gait parameters within a study, and many studies use some sort of constraint to natural gait patterns to isolate the particular gait parameter they are interested in studying. Furthermore, the majority of such studies have examined only the amount and not the structure of variability (Newell & Corcos, 1993). It is becoming increasingly apparent (as will be discussed presently) that global measures of variability are insufficient for adequate assessment of gait variability (e.g., Hausdorff et al., 1997).

Traditionally, variability in human movement has been treated as noise superimposed upon a signal, where the signal is the intended movement and the noise is variation about this intended movement (Newell & Slifkin 1998). As such, the focus of this approach has been to quantify the amount of variation associated with the movement property of interest. Typically variability is indexed by either the standard deviation (in absolute terms) or by the coefficient of variation (standard deviation divided by mean). However, although the amount of variability present in the gait cycle provides some information, any sequential dependencies that may exist in the cycle to cycle variation are neglected. In order to investigate the time dependent nature of gait cycle variability, it is necessary to employ both time and frequency domain techniques that can be used to analyze the sequential dependencies of variability. There is a large body of literature which provides evidence that that the cycle to cycle variation seen in a wide variety of physiological systems is non-trivial and may offer insight into the organization and control of these systems (e.g., Chen, Ding & Kelso, 1997; Duarte & Zatsiorsky, 2000, 2001; Hausdorff, Peng, Ladin, Wei, & Goldberger, 1995; West & Scarfetta, 2003; Yoshinaga, Miyazima & Mitake, 2000). For example, the strength of long range correlations in the inter beat interval of the heart has been shown to be a powerful predictor of mortality among patients with chronic congestive heart failure (Ho, et al., 1997).

Recently, Detrended Fluctuation Analysis (DFA) and spectral analysis have been applied to stride interval time series data from human walking. The goal of this approach is to improve the understanding of the organization, regulation and interactions of the locomotor system (e.g. Hausdorff et al., 1995; Hausdorff, Peng, Wei & Goldberger, 2000; Terrier & Schutz 2003). The focus of this chapter is to review and summarize the collective findings of the current literature on the influence of speed on intra-subject variability of human locomotion. These findings and those relating to the structure of gait cycle variability will be discussed with a view to understanding their relevance to the control of locomotion. Lastly, variability as it relates to the transitions of walking and running will be discussed.

2.1 The Influence of Speed on the Amount of Gait Cycle Variability

Several studies have examined the influence of speed on the variability of both walking and running. The results of these studies have, however, been inconsistent, both for walking and for running. This is likely to be in part due to use of different methods for both assessing variability (standard deviation vs. coefficient of variation) and for collecting data (treadmill vs. over-ground locomotion; constrained vs. unconstrained locomotion). The results of these studies as well as the different approaches taken to the study of variability in human locomotion are now discussed.

2.1.1 Decreasing variability with increasing speed

One trend seen in the gait variability literature is that of decreasing variability with increasing gait speed. Maruyama and Nagasaki (1992) investigated the temporal variability of walking over a range of speeds (2 to 6km/hr) and step rates (from 60 to 140step/min). They showed that temporal variability of gait was dependent on step rate and that for any given step rate both SD and CV of stride, step, stance, swing and double stance times of walking decreased with increased walking velocity. However, locomotion in this study was constrained in that participants were required to walk in time with a metronome to control for step rate. While free walking data were collected, only changes in the variability of metronomic walking with speed were reported.

Diedrich and Warren (1995) showed that as running speed increased from very slow running speeds (i.e. speeds that people would naturally walk at) there was a significant decrease in the standard deviation of the relative phase of both the hip-ankle and knee-ankle. No significant effects were observed for either stride frequency or stride length with increasing speed, indicating that variability of relative phase of the joints may be more sensitive to changes in speed. Hausdorff et al. (1996) did not explicitly examine the distributional statistics of the stride interval time series in terms of increasing speed, however, the data collected for this study are available online at www.physionet.org. A 2 (metronome) by 3 (speed) ANOVA of these data reveals that there is a significant decrease in absolute (standard deviation) variability with speed.

The findings on the influence of gait speed on variability in human locomotion generally show a decrease in variability with increased velocity. This is in keeping with the work of Newell and others (Newell, Carlton, Carlton, & Halbert, 1980; Newell, Hoshaizaki, Carlton, & Halbert, 1979) who examined the effects of speed on the variability of discrete aiming movements. It was shown that higher velocity movements (even to the same movement time) are temporally more consistent (less variable) than slower movements. This implies that the faster one walks (and this presumably holds for running too), the less variation there will be in temporal measures of movement outcome.

The inter-subject variability findings of Winter (1984) are also consistent with this suggestion. Inter-subject variability of the joint moments of force at the hip and knee decreases with increasing speed, which is likely to be driven at least in part by decreasing intra-subject variability. There was also an increase in covariation between these two joints with speed. This may result from neuromuscular processes which actively constrain independent movement of the joints with increasing speed (Winter, 1984), or it may be due to the neuro-mechanical limits of the locomotor system. For example, Winter points out that at the fastest cadence, the flexor/extensor moments generated at the hip and knee joints are approaching the maximum of their dynamic range. These possibilities are not mutually exclusive and both suggest mechanisms for the reduction in variability seen at the level of the stride interval. Winter (1984) also notes that low variability at the level of joint moments of force does not necessarily imply a consistent motor pattern due to the infinite number of joint moments of force that could generate identical joint angle histories. However, when the results of the above studies are considered as a whole, it appears as if the decrease in joint moments of force variability associated with increased speed may be at least in part responsible for the decrease in stride interval variability seen under these conditions.

2.1.2 Increased variability with increased speed

Contrary to the studies just discussed, there have been several reports of an increase in variability with increasing gait speed. Belli, Lacour, Komi, Candau and Denis (1995) investigated the temporal variability of running over a range of speeds and showed that CV of stride time during running increased with velocity. Participants in this study ran at speeds corresponding to 60, 80, 100 and 140% of their maximum oxygen uptake for 1 – 3min. Absolute step time variability was not affected by running speed until 140% of maximum oxygen uptake, at which point SD increased significantly. CV of step time also increased with velocity, with significant increases occurring between both 60 and 100% of maximum oxygen uptake and 100 and 140% of maximum oxygen uptake.

In terms of walking, Masani, Kouzaki and Fukunaga (2002) examined the variability of ground reaction force during treadmill locomotion. They showed that there was an increase in the CV of both vertical (Fz) and medio-lateral (Fx) ground reaction force as walking speed increased from 3 to 8km/hr. Fx reflects the lateral sway of the body during locomotion and is generally accepted as being a measure of postural stability (e.g., Donelan, Shipman, Kram & Kuo 2005). Fz on the other hand reflects movement in the vertical direction. Increased variability in Fz indicates that there is an increasing amount of variability in vertical displacement of the center of mass with increasing speed. This combined with increased variability in lateral movement implies that walking at 8km/hr is significantly less stable than walking at preferred walking speeds.

The speeds examined by Belli et al. (1995) essentially correspond to preferred running speeds and faster, whereas those examined by Diedrich and Warren (1995) correspond to preferred and slower. This likely explains the apparently contradictory results from these two studies and highlights the importance of examining a broad range of speeds. The combined results of Belli et al. (1995) and Diedrich and Warren (1995) imply that the variability function for running follows a u-shaped pattern with speed, with the preferred running speed having the lowest amount of variability.

2.1.3 Non-linear Variability Functions with Increasing Speed

In addition to linear changes in gait cycle variability with speed, there have been a number of studies which indicate there is a curvilinear change in variability with speed. The data of Hausdorff et al. (1996) previously discussed show that absolute variability decreased with walking speed. However, when standard deviation was normalized by mean stride interval, a curvilinear pattern of change with speed emerged. There is a significant decrease in relative variability from the slow to normal speed, the decrease from the fast to normal speed, however, failed to reach significance. Van Emmerik, Wagenaar, Winogrodzka and Wolters (1999) investigated SD of stride interval of walking from very slow (0.2 m/s) to moderate (1.4 m/s) walking speeds. Overall the pattern of change with speed was curvilinear, with the most dramatic decreases in variability occurring from 0.2 to 0.6 m/s. SD continued to decrease with increasing speed, however the change between 0.6 and 1.4m/s appears more linear.

Testing three different speeds in both men and women, Hirokawa (1989) found that CV of the step length was minimized at normal walking speeds as compared with slow and fast walking speeds. It should be noted, however, that statistical significance was not reported for any measures in this study. In two similar studies, Yamasaki, Sasaki, Tsuzuki and Torii (1984) and Yamasaki, Sasaki and Torii (1991) investigated walking at different speeds in both males and females. The results of these studies showed that for both sexes moderate walking speeds yielded the lowest amount of variability (as indexed by both SD and CV) in step length and duration. In both studies, an average walking of speed of approximately 5.4 km/hr was estimated to yield the lowest amount of variability for both men and women. Sekiya, Nagasaki, Ito and Furuna (1997) also showed a U-shaped relation between SD of step length and walking velocity with the minima occurring close to the speed of preferred walking. Diedrich and Warren (1995) showed that for walking there was a U-shaped function for the standard deviation of relative phase in both hip-ankle and knee-ankle couplings, with the minimum SD occurring around preferred walking speed. For both the stride frequency and length of walking, similar U-shaped trends were observed, however, they did not reach significance.

In the study by Masani et al. (2002) previously discussed, the results for Fy, or the propulsive force, showed that variability was reduced at the preferred walking speed. From this finding it was concluded that the neuromuscular locomotor system is optimized in terms of propulsive forces rather than in terms of stability. While this conclusion is questionable (given the necessity of maintaining an upright position to effectively generate propulsive forces) it assumes that variability can be equated with stability and this assumption many not be valid.

For example, Dingwell and Marin (2006) also found a significant quadratic trend for variability (SD) and walking speed. In this case, stride variability was calculated from the first difference time series of displacement data. The first difference time series was normalized to stride duration and the mean SD for a trial was calculated as the average SD at the intervals 0 through 100% of stride time. Finite time Lyapunov exponents were calculated to provide a measure of local dynamic stability, or, in other words, a measure of how resistant the locomotor system is to very small, internally generated perturbations. The results from this analysis showed that, unlike the variability measure, local dynamic stability decreased with increasing speed.

As such it appears that there is not a 1:1 mapping between variability and local dynamic stability. Nevertheless, the results of these studies offer an intuitive explanation of gait variability being minimized around the same speed at which people are most accustomed to walking, which in turn coincides with the speed at which metabolic energy expenditure is minimized (Bobbert, 1960; Zarrugh, Todd & Ralston, 1974). Whether this effect is due to the larger amount of time spent walking at preferred or whether it is reflective of some overall stability associated with preferred modes of behaviors remains to be seen.

In the study by Maruyama and Nagasaki (1992) previously mentioned, while it was shown that variability of the temporal gait cycle variables decreased with increasing speed, it was also shown that CV in the duration of step, stance, swing and double stance times exhibited a minimum in a mid-range of step rates (approximately 80 – 125 steps/min) for all speeds. As speed increased there was an increase in the step rate at which minimum CV occurred. The step rate at which minimizes CV for a given walking speed is close to the step rate selected for that speed under natural walking conditions suggesting that free walking is optimized in terms of cadence variability. The findings of this study are consistent with those of tapping tasks which indicate a preferred frequency of tapping as indicated by minimization of variability (Nagasaki & Nakamura, 1982; Sternad, Dean & Newell, 2000).

The frequency related findings for walking may be explained in terms of the pendular behavior of the lower limb during walking (Holt, Hamill & Andres, 1990, 1991). Mechanical systems have an "eigen" frequency, or a preferred frequency for which the amount of energy required to sustain oscillation is minimal. In the case of a pendulum (or limb), this eigenfrequency is dependent on both the length and mass of the pendulum, as well as the distribution of the mass along the length of the pendulum. Additionally, the eigenfrequency is typically considered to be the most stable frequency for the pendulum to oscillate at (Kugler & Turvey, 1987).

Holt et al. (1990) developed a force-driven harmonic oscillator model to investigate whether preferred frequency of locomotion could be predicted by the least amount of energy required to drive a harmonic oscillator. The results suggested that in walking, the metabolic cost is minimized by taking advantage of the dynamical properties of the leg which in turn reduces the force producing contribution of muscle. Following this, Holt et al. (1991) showed what when speed was kept constant, there was no significant difference between the preferred and predicted period of oscillation, and, that these periods coincided with minimum metabolic expenditure. These findings support the idea of a preferred stride frequency for walking which is related to the dynamic properties of the leg and are consistent with the previously described experimental findings for stride frequency.

2.1.4 Summary of Findings on Amount of Gait Cycle Variability

The previous sections demonstrate several patterns of findings regarding the influence of speed of locomotion on variability of gait cycle parameters. This is likely to be due in large part to methodological differences across the different studies. Maruyama and Nagasaki (1992) for example required their participants to walk in time to a metronome. Under such conditions, locomotion becomes to some extent a timing task where the goal is to minimize the difference between foot falls and metronome beeps, thus, the time series of stride intervals is in effect a time series of inter-response intervals. At the very least this constraint will likely induce a reduction in the amount of step to step variation of gait cycle parameters, and, has significant implications for the control of gait parameters (e.g., Hausdorff et al., 1996).

Hirokawa (1989) and Sekiya et al. (1997) examined over-ground walking along relatively short walk ways (~10m) along which it is possible that true steady state locomotion was not achieved. Furthermore, because of the short length of the walkway, only a limited number of consecutive foot falls could be collected. The usefulness of SDs and CVs calculated under these circumstances of limited data is questionable. As such, in these studies it is difficult to get a clear and accurate assessment of variability of the gait cycle during normal walking or running. To fully understand the role that variability plays in the control of locomotion, particularly with respect to speed, it is necessary to have participants walk and run for at least several minutes at a variety of different speeds without introducing additional temporal or spatial constraints.

Despite the apparent inconsistency in the previously discussed results, when unconstrained walking over a range of speeds is considered, the dominant pattern of variability change with speed is curvilinear. As previously mentioned, the combined results of Diedrich and Warren (1995) and Belli et al. (1995) suggest that running at or around preferred running speeds minimizes the variability of the gait cycle. The data of Hausdorff et al. (1996) and the results of Van Emmerik et al. (1999) highlight the relatively larger influence of gait variability at slow speeds as compared with preferred or moderately faster speeds. The fastest speed investigated by Van Emmerik et al. (1999) was approximately 5km/hr, which corresponds to the average preferred speed in Hausdorff et al. (1996). The "fast" speed in from Hausdorff et al. (1996) was approximately 6km/hr. The results of Yamasaki et al. (1984) and Yamasaki et al. (1991) indicate that variability, at least in the stride interval and length of Japanese men and women does not increase substantially until speeds greater than 6km/hr are reached. Holt et al. (1995) have demonstrated a U-shaped curve for head stability with increasing stride frequency, with the standard deviation of head trajectory in the vertical plane remaining constant from around 75 to 100% of predicted preferred stride frequency.

The results of Masani et al. (2002) show that both vertical and medio-lateral ground reaction forces are not optimized in terms of preferred walking speeds. It was, however, shown that force in the anterior-posterior (A-P) direction does follow the same pattern of results as stride interval and length, reaching a minimum at speeds associated with normal walking. The second half of the A-P ground reaction force curve is regarded as a propulsive force and thus may be related to step length - Masani et al. (2002) showed that there were linear increases in both the size of the 2nd peak of the A-P force and step length. Thus, it is possible that the variability function of step (stride) length is driven by variability of the A-P ground reaction force.

Cumulatively these results indicate that the neuromuscular system is able to effectively optimize walking in terms of the variability of stride interval and length over a range of speeds. Whether this is a result of training at preferred speeds or whether this signifies improved stability at these speeds remains to be seen. In any case, it is apparent that a broad range of both preferred and non-preferred walking speeds needs to be investigated in order to capture the full range of speed related variability changes. Another consideration that will be addressed in a later section of this chapter is the use of measures that examine the structure and not just the amount of variability present in the fluctuations of the gait cycle.

2.2 Gait Transitions and Locomotor Variability

There have been a number of studies in the last decade that have examined the hypothesis that gait transitions take the form of equilibrium phase transitions. The motivation for this approach comes primarily from the HKB model (Haken et al., 1985) where a change from anti-phase to in-phase finger movement was observed with increasing frequency of finger oscillation. Because the walk to run transition occurs with increasing speed, or increasing frequency of leg oscillation, researchers have applied the same principals as Haken and colleagues (1985) to the study of gait transitions. Of particular relevance to this dissertation is that this approach to the study of gait transitions involves the investigation of variability of the gait cycle, as this is typically regarded as being an indication of a loss of stability.

The SD of relative phase is a primary measure of coordinative stability (e.g., Diedrich & Warren, 1995, 1998; Haken et al.,1985; Kao, Ringenbach & Martin, 2003;

Seay, Haddad, van Emmerik, & Hamill, 2006). The theoretical motivation for this comes from the notion of critical fluctuations – small variations in the coordination pattern that increase in magnitude as the stability of the coordination pattern decreases (Haken et al., 1985). Relative phase is a reflection of system coordination in that it reflects how the movement of relevant parts of the system (e.g., fingers or limb segments) is coupled. Thus variability in relative phase is regarded as an index of instability in the coordination pattern. Researchers have also used the variability of stride time and length to examine the stability of the gait cycle during gait transitions (e.g. Brisswalter & Mottet, 1996; Diedrich & Warren, 1995).

In general, the results of these studies indicate that there is increased in gait cycle variability when walking and running at speeds associated with gait transitions. Diedrich and Warren (1995) found that the SD of relative phase between the hip and ankle increased in the transition region in both the walking and running gaits. A similar pattern of change was seen for the knee-ankle relative phase in the running gait, but not in the walking gait. During walking, there was an increase in variability of stride frequency in the transition region, but there was no change in variability stride length during walking or in stride frequency and length during running. These results were replicated in a later study by the same authors (Diedrich & Warren, 1998).

Kao et al. (2003) found that hip-ankle relative phase became less variable after the transition to running occurred, although they did not find any change in the amount of variability in the knee-ankle relative phase associated with gait transition. Brisswalter and Mottet (1996) found a decrease in CV of stride duration immediately following the transition to running which is consistent with the results of Diedrich and Warren (1995).

Seay et al. (2006) did not find an increase in variability of relative phase associated with gait transitions. However, this study examined the relative phase of segment angles (i.e. the angle a leg segment made with a horizontal reference line) rather than joint angles.

In the decerebrate cat, sensory information associated with the hip joint has been shown to be relevant for modulation of the gait cycle (e.g., Grillner & Rosignol 1978; Kriellaars, Brownstone, Noga & Jordan 1994). Bernstein (1967) has suggested that one strategy for the reduction of degrees of freedom is for the central nervous system to control joint position rather than specific muscle activity. Therefore, the angle two limb segments make with each other is likely to be a more functionally relevant measure than the angle a limb segment makes to either a horizontal or vertical reference. As such the latter measure fails to capture the entirety of the coordination pattern and this may be why no significant speed effects on variability were detected.

Although it appears that the gait cycle does become more variable in the region of gait transition speeds, it is unclear from these studies whether gait transitions are driven by a loss of stability of coordination patterns. This may be the result of methodological differences just discussed. Regardless, measures such as SD provide only a global, static representation of the amount of variability present in a time series. There is growing literature suggesting that the information contained within the temporal variations of subsequent strides may be revealing about the control of locomotion (e.g., Dingwell & Cusumano, 2000; Hausdorff et al., 1996; Stergiou, Moraiti, Giakas, Ristanis & Georgoulis, 2004).

2.3 The Structure of Variability in Locomotion

The studies previously mentioned made use of distributional statistics of a gait variable to ascertain the amount of variability for a given gait parameter. It is becoming increasingly apparent that fluctuations in the gait cycle are not random (Hausdorff et al., 1995; Kurz & Stergiou 2006; West & Griffin, 1998; 1999). Rather they are self-similar and exhibit long range dependence, such that, any given stride interval is dependent on the stride interval at remote previous times (Hausdorff et al., 1995). The strength of these correlations appears to be dependent on several factors, including age (Hausdorff et al., 1999), neuromuscular health (e.g. Hausdorff et al., 1997; Kurz & Stergiou, 2006) and possibly gait speed (Hausdorff et al., 1996). Thus an increasing amount of research has focused on the time dependent nature of cycle to cycle variation seen during locomotion. These studies are now discussed.

2.3.1 Self Similarity in the Stride Interval of Human Walking

Detrended fluctuation analysis (DFA) is a technique created by Peng et al. (1993) for the calculation of long-range correlations in physiological time series. Long range (power law) correlations in a time series imply the presence of 1/f noise since the power spectrum is the Fourier transform of the autocorrelation function. DFA (which will be described in more detail in the following chapter) is a modified random walk analysis that makes use of the fact that a long range correlated time series can be mapped to a self similar process by integration (Hausdorff et al., 1996). The integrated time series is examined using a windowing process, the log of the variance for a given observation window (F(n)) is plotted against the log of the window size (n) and the linear slope is calculated, yielding an “alpha” value. White noise corresponds to an alpha of 0.5, or a spectral slope of 0; pink noise corresponds to an alpha of 1 or a spectral slope of 1. An alpha value between 0.5 and 1 is indicative of long range correlations such that any given stride interval is dependent on the stride interval at remote previous times and that the dependence of stride intervals decays in a power law, fractal-like manner with time. Alpha values in this range also indicate that fluctuations are self similar, in that fluctuations at one time scale are statistically similar to fluctuations at all other time scales.

Research using DFA has shown that fluctuations are present in the stride interval (where stride interval is defined as the time between the heel strike of one foot and the successive heel strike of the same foot). During walking in healthy young adults, alpha falls between 0.5 and 1.0. For example, in 9 min walking trials, the average value for the DFA scaling exponent was 0.76 (Hausdorff et al., 1995). Subsequent research using a trial length of 1 hour shows that these long range correlations extend for over 1000 strides (Hausdorff et al., 1996). To differentiate statistically between a long range scaling process and a process without correlations, Hausdorff and colleagues generated surrogate data sets by randomly shuffling the original time series. In this way, the distributional statistics (i.e. mean, standard deviation and higher moments) are the same for both the original time series and the corresponding surrogate time series, however, the sequential ordering was destroyed. It has been suggested that the long range correlations observed in healthy population during unconstrained walking may be adaptive in that the presence of multiple time scales prevents mode-locking (Hausdorff et al., 2000).

Using a different experimental protocol for generation and analysis of the stride interval time series, West and Griffin (1998, 1999) confirmed the findings of Hausdorff and colleagues in healthy young individuals. Instead of using force sensors and heel strike to calculate the inter-stride interval, West and Griffin used a goniometer and generated the time series using successive maximal positive extensions of the same knee. The time series were analyzed using relative dispersion (R) and data aggregation, such that the relative dispersion (or CV) was calculated over successively larger numbers of data points (n). The logged values of R and n were plotted and slope calculated. The value of the slopes were in agreement with the alpha values found by Hausdorff and colleagues, (1995, 1996) confirming the presence of long range correlations in the stride interval of human walking.

2.3.2 Frequency Constrained Locomotion

As discussed previously, many studies investigating gait cycle variability require participants to constrain their step frequency, typically by walking in time to a metronome. One draw back to this approach is that it artificially reduces the variability of the stride interval. Another, more subtle issue is that it has been shown that when subjects walk in time with a metronome, long range correlations in the stride interval break down (alpha ≈ 0.5), such that the time series approximates uncorrelated noise (Hausdorff et al., 1996).

Using a GPS system to record step frequency, step length and walking speed time series, Terrier, Turner and Schutz (2005) showed that there were long range correlations present in all of these variables. Interestingly, when walking in time to a metronome, only the long range correlations of the step frequency time series became anti correlated, the remaining variables were unaffected by the timing constraint. The authors concluded that this was due to the feedback loop between the planned (at the level of the spinal cord) movement and the actual movement causing a constant shift about the mean value of step frequency.

A related explanation for these findings lies in the overall goal of the different tasks. In unconstrained over-ground walking there is no explicit requirement to walk with a specific time interval between strides. However, when walking in time to a metronome, the task becomes a timing task and the time series of stride intervals is in effect a time series of inter-response intervals. It is well established that there is a negative covariation between adjacent time intervals during repetitive response tasks (e.g., Wing & Kristofferson, 1973) and as such it should be anticipated that the long range correlations in the stride interval time series break down when people are required to walk in time to a metronome. Because there is no explicit constraint on either the step length or walking speed in the study of Terrier et al. (2005), the long range correlations of these time series are unaltered. One would expect that had the step or stride length been constrained that a similar breakdown in long range correlations would occur in the time series of this variable.

2.3.3 Variables Influencing the Size of alpha

The long range correlations in walking data also break down with aging and disease (Hausdorff et al., 1997) with the long range correlations being significantly lower in elderly (alpha = 0.68) as compared with young adults (alpha = 0.87), and lower still in the case of Huntington's disease patients (alpha = 0.60). Conversely, in the stride interval of young children, the long-range correlations decay more slowly with time than they do in the stride interval of young adults (Hausdorff, Zemany, Peng, & Goldberger, 1999). Thus, it appears that across the life span there is a general trend for a decrease in long range correlations in walking. The mechanisms for these changes in long range correlations with aging are likely to be independent, given that the direction of the shift in size of the scaling exponent is constant with increasing age, where as there is a u-shaped function for walking ability with age. One modeling approach suggests that the decrease in strength of long range correlations seen from childhood to adult hood is related to an increase in neuronal connectivity, whereas the decrease in strength associated with aging and disease is related to the unavailability of some neuronal centers (Ashkenazy, Hausdorff, Ivanov & Stanley, 2002).

With a view to confirming the robustness of the long rang correlations in the stride interval, Hausdorff and colleagues (1996) investigated unconstrained walking at slow, normal, and fast walking paces. The average alpha for these respective speeds was 0.98, 0.90 and 0.97. The differences in alpha with speed was significant, which indicates that there may be systematic changes in the size of alpha with walking speed.

Thus, it is apparent that long range correlations are present in a variety of gait cycle parameters and that they are affected by age, disease state, speed of locomotion and timing constraints. Generally, it appears that although a healthy, unconstrained locomotor system will generate time series that have DFA scaling exponents between 0.5 and 1, values of alpha which approach these limits may be less optimal. For example young children have relatively high alpha values, meaning that from stride to stride they are more regular in time. Conversely, in aging and disease alpha values are relatively low, demonstrating a high degree of randomness. This may relate back to the active degrees of freedom within a given system – in the case of excessive regularity, there may be a reduction in the available degrees of freedom, whereas in the case of excessive randomness the problem may lie in the inability to constrain the active degrees of freedom (vis. Vaillancourt & Newell, 2002). The alpha values seen at preferred, fast and slow walking reported by Hausdorff et al (1996) indicate that at slow and fast speeds, the stride interval time series are significantly more regular than at the preferred walking speed. This suggests that at the preferred walking speed there may be more degrees of freedom available for the control of locomotion.

2.4 Filling in the Gaps – Future Research Directions

In principle, the strength of the long range correlation is representative (at least in a statistical sense) of the degree of influence that any given stride has over upcoming strides. It follows that a high alpha value indicates a high degree of constraint from stride to stride, while a low alpha value indicates the opposite. Hence it seems likely that there is some optimal level of correlation that relates to the overall adaptability of locomotion. The next step in this line of research would be to examine the strength of long range correlations over a range of walking and running speeds to investigate the hypothesis that the speeds naturally selected for walking and running are relatively more adaptable than faster and slower speeds.

Another issue that is raised by all of the studies discussed thus far is the priority of kinematic or kinetic variables in analysis of variability. While there are a vast number to chose from (cf. Winter, 1984) similar arguments are often made for the selection of different variables. For example, Winter (1984) regarded the moments of force about a joint as being the final desired motor pattern at that joint because they represent the net effect of all agonist and antagonist muscle activity. Conversely, Hausdorff et al. (1996) regarded the stride interval as being the final controlled output of the neuromuscular control system. Given the apparent reduction in variability at the distal segments during locomotion (Winter, 1984), and the finding that the stride interval is less variable than the parts (e.g., stance time, swing time) that comprise it (Maruyama & Nagasaki, 1992) it is reasonable to argue for the primacy of the stride interval as a gait variable.

Masani et al. (2002), however, suggest that ground reaction force is a more appropriate global parameter for assessing/characterizing gait than kinematic measures such as stride time/length because the kinetic level reflects the cause of movement. This is the same argument that Winter (1984) advanced for the use of moments of force about a joint as a primary variable. Diedrich and Warren (1995) observed that the variability of relative joint angle is more sensitive to changes in speed than stride length or frequency. However, their finding of no change in variability of stride length or frequency with changing speed is contrary to the majority of reported findings in the gait literature (see previously discussed articles). Another possible variable is head displacement, given the obvious importance of stabilizing the head with respect to the environment. Holt, Jeng, Ratcliffe and Hamill (1995) have reported that head stability is greatest at the preferred step rate. In any case, it can be concluded that in order to have a full understanding of the role that long range correlations play in the control of locomotion, a variety of kinetic and kinematic variables should be investigated.

One goal of this dissertation is to investigate the pattern of change of the DFA scaling exponent with speed of locomotion in both walking and running with a view to understanding what this means in terms of control of human locomotion. Results from Hausdorff et al. (1996) suggest that there may be a U-shaped pattern of change in alpha with speed. The pilot data of this dissertation (Jordan et al., 2005) indicate the same trend in running. U-shaped curves, centered around preferred behavior suggest that the variable in question is being optimized. For example, the energetic cost curve for walking follows a U-shaped curve, the minimum of which is close to the preferred walking speed (e.g., Hreljac, 1993; Margaria et al., 1963), and this indicates that energy expenditure is optimal at preferred walking speeds. There is some evidence that variability of one or more gait parameters follows a U-shaped curve with increasing walking speed (e.g., Dingwell & Marin, 2006; Hirokawa, 1989; Sekyia et al., 1997; Yamasaki et al., 1991). Therefore, the focus of the first two experiments will be on changes in both the amount and structure of variability in the walking and running gaits with speed.

Another goal of this dissertation is to examine the effects of walking and running at and around preferred transition speeds on the strength of long correlations. There is sufficient evidence in the literature to suggest that walking and running at these speeds introduces a degree of instability to the coordination pattern (e.g., Brisswalter & Mottet, 1996; Diedrich & Warren, 1995; 1998). Thus, the focus of the third experiment is on the relationship between changes in the amount and structure of variability, as well as changes in both local dynamical stability (Dingwell & Cusumano, 2000) and coordination pattern stability (Diedrich & Warren, 1995, 1998). Additionally in this study, the difference both in strength of long range correlations and local dynamic stability at the head and ankle were examined. The functional roles of the ankle and head may be different during locomotion (the head being more important for maintaining postural control, the ankle being more important for forward propulsion of the body). Thus, it is expected that there will be differences in the pattern of change in both stability and strength of long range correlations between the head and ankle. Collectively these experiments are designed to further the understanding of the significance of long range correlations in the gait cycle of human locomotion, particularly as they related to variability and stability of locomotion over broad range of speeds.

Chapter 3

Walking Speed Influences on Gait Cycle Variability

3.1 Introduction

Walking is one of the most practiced of all motor skills, thus it is not surprising that there is a very low level of variability (e.g. coefficient of variation ~3%) associated with many biomechanical measures of this particular task, (e.g., Hausdorff et al., 1996; Winter 1983). Despite this, it has been shown that the variability contains fractal-like structure in the form of persistent long range correlations in the stride to stride fluctuations of walking (Hausdorff et al., Terrier, Turner & Schutz, 2005). This finding has led to the idea that the fractal nature of these fluctuations maybe revealing about the mechanisms underlying neural control of locomotion. Research shows that long range correlations in the stride interval of walking break down and become anti-correlated when subjects walk in time to a metronome (Hausdorff et al., 1996; Terrier et al., 2005); that long range correlations break down in older adults and in disease states such as Huntington's due, it is proposed, to a loss of complexity of the neuromuscular system associated with aging and disease (Hausdorff et al., 1997); and that the strength of long range correlation decreases monotonically across the life span (Hausdorff et al., 1999).

Haudsdorff's group has consistently examined the interval between successive heel strikes of the same limb (e.g., Hausdorff et al., 1995; Hausdorff et al., 1996; Hausdorff et al., 1997). However, it has also been shown that the scaling behavior of gait cycle fluctuations is present in the step frequency and length time series as calculated from head displacement recorded via a GPS system (Terrier et al., 2005) and in the stride interval as calculated from peak knee extension (West & Griffin, 1998, 1999). There has been limited comparison or discussion of differences (or similarities) of the strength of long range correlations across these different levels of examination. The finding that the time series of step length, step frequency and instantaneous walking velocity also contain fractal structure (Terrier et al., 2005) demonstrates that long range correlations are not limited to the stride interval.

In addition to the findings just discussed, there is evidence that suggests the strength of the long range correlations in walking may be speed dependent (Hausdorff et al., 1996). Thus far there has been no systematic investigation of the speed – long range correlation function in walking. Individuals exhibit a preference for a particular walking speed, which occurs at or close to the speed at which energy consumption (per unit distance) is minimized (Margaria, 1976). There is also evidence which suggests that preferred walking speeds are associated with a minimum of stride interval variability (Yamasaki, Sasaki, & Torii 1991; Yamasaki, Sasaki, Tsuzuki & Torii, 1984). Results from both walking (Holt, Hamill & Andres, 1990), and leg swinging (Doke, Donelan & Kuo, 2005) studies provide evidence that the metabolic cost of walking is minimized by taking advantage of the dynamical properties of the leg, which in turn reduces the required force contribution of muscle. It has also been shown that when walking at a constant speed, preferred and predicted periods of oscillation are not significantly different, and, that these periods coincide with minimum metabolic expenditure (Holt, Hamill & Andres, 1991). While these findings are specific to changes in frequency with constant speed, they nevertheless support the idea of a preferred walking speed (PWS) which is related to the dynamic properties of the leg. Thus, the amount and structure of variability may be related to the pendular behavior of the lower limb during walking.

This experiment examines the hypothesis that there is a reduction in long range correlations at the preferred walking speed and that this may be related to the dynamics of walking. Based on the findings of Hausdorff et al. (1996), it is hypothesized that the long range correlations will follow a U-shaped pattern of change with speed. It is also predicted that the presence of long range correlations will not be limited to the stride interval (Terrier et al., 2005) and that the influence of speed would be similar across the kinematic and kinetic variables of the gait cycle examined herein.

3.2 Methods

3.2.1 Subjects

Eleven female volunteers from The Pennsylvania State University between the ages of 22 and 30 years of age (average age = 24.9 ± 2.4 years; average height = 164.9 ± 5.1 cm; average mass = 57.2 ± 3.1 kg) were recruited for the study. All participants provided informed consent and all procedures were approved by the Institutional Review Board of The Pennsylvania State University.

3.2.2 Apparatus

The apparatus consisted of a Kistler Gaitway treadmill with 2 embedded force plates (see Figure 3.1). Vertical ground reaction forces (VGRF) were measured by uni-dimensional piezoelectric force sensors and sampled at a rate of 250Hz. The treadmill had a speed range of 0.8 to 20 km/hr, with the smallest increment in speed being 0.1km/hr.

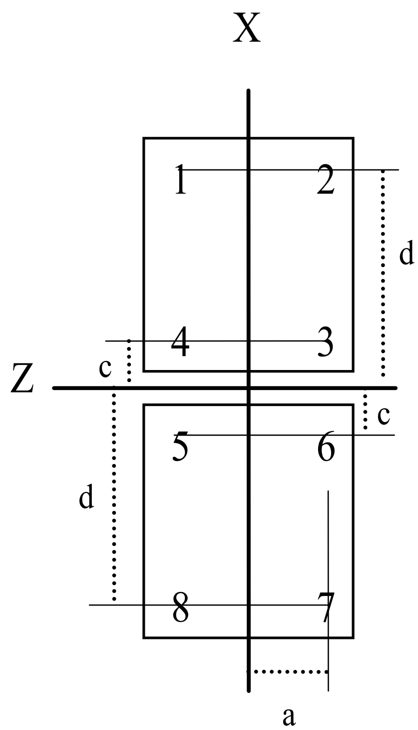

Figure 3.1: Schematic of the force plates and pressure sensors.

3.2.3 Tasks and Procedures

On a day prior to data collection, participants spent 45 min walking on the treadmill at speeds they felt comfortable to become familiar with the treadmill. The first 10 min of each experimental session was used as a warm up/treadmill adaptation period. The preferred walking speed (PWS) was established during this period by initially having the participant walk at a relatively slow speed, and then the investigator increased the speed in 0.1km/hr increments until the subject reported that they were walking at their PWS. The speed was then increased by approximately 1.5km/hr and then decremented by 0.1km/hr until the PWS was re-established. This procedure was repeated - in the majority of participants a similar speed was arrived at on the first 2 attempts to establish PWS and in this case the average of these speeds was taken to be the PWS. If there was not a good match between speeds (greater than 0.4km/hr different) the process was repeated until a close match was achieved.

During this process of establishing the PWS, the participants were prevented from viewing the speed at which they were walking. Participants performed one 12 min trial at each of the following percentages of preferred walking speed: 80, 90, 100, 110 and 120%. Speeds were presented randomly and participants were given at least 2 min and up to 10 min to recover between trials. Approximately 650 stride intervals were captured per 12 min trial.

3.2.4 Analysis of Force Plate Data

Custom written MATLAB software was used to compute and process the VGRF data. The center of pressure and ground reaction force data were filtered in forward and reverse directions with a 2nd order Butterworth filter with a cut off of 15Hz. A 20N threshold was used for the identification of the start and end of each foot fall - signal below this threshold was considered to be noise.

Eight gait cycle variables were investigated: stride interval and length, step interval and length, and, from the VGRF profile, the impulse, first and second peak forces, and the trough force. Stride interval was defined as starting with the onset of the heel strike of one foot and finishing with the onset of the next heel strike of the same foot. Step interval was defined as starting with the heel strike of one foot and finishing with the subsequent heel strike of the other foot. Step length was calculated by multiplying the step interval by the average belt speed for that interval and adding (or subtracting as appropriate) the difference in distance between the two heel strikes used for the calculation of step interval. Stride length was calculated similarly, multiplying stride interval by the average belt speed and adding (subtracting) the difference in distance between the successive heel strikes of the limb. Impulse is the area under the force time curve, first and second peak force were taken as the maximum vertical ground reaction force produced during the force absorption and generation phases of stance respectively. Trough force was the force at the minima between the first and second peaks (see Figure 3.2).

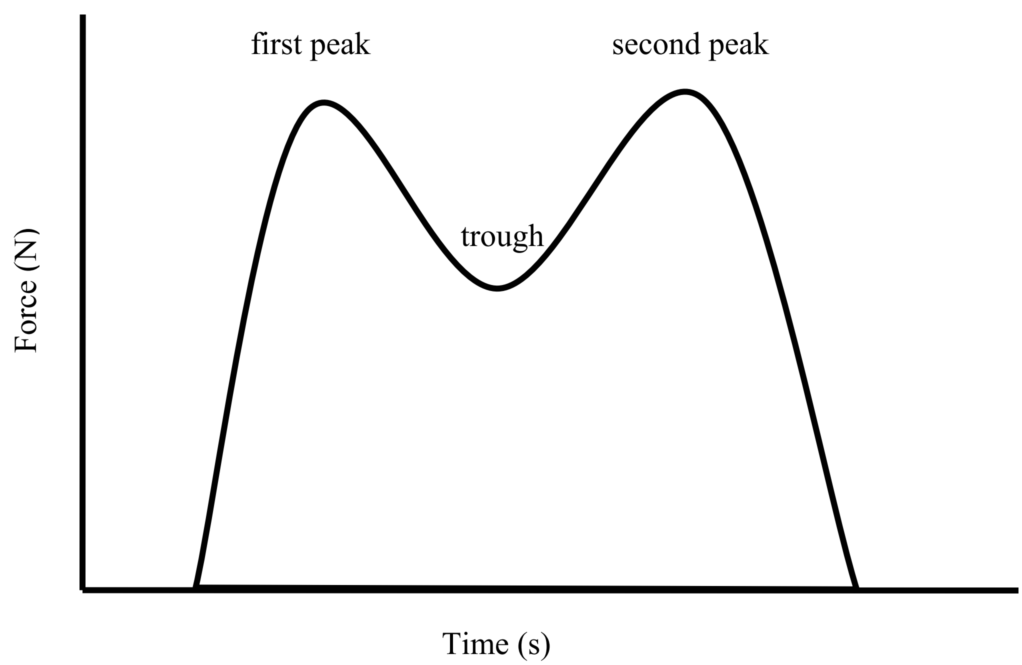

Figure 3.2: Schematic of foot contact impulse. The area under the force-time curve is vertical impulse.

3.2.5 Data Analysis

The mean, standard deviation (SD), coefficient of variation (CV), and the strength of long range correlations (alpha) were calculated for the time series of each dependent variable. The SD and CV index the amount of variability of the time series and do not reflect the structure of stride to stride fluctuations. Long range correlations on the other hand provide a measure of the structure of variability of the time series. Stronger correlations indicate a more predictable time series whereas weaker correlations indicate a less predictable time series where any given stride interval is less dependent on the stride intervals preceding it.

Long range correlations of the time series data were calculated using detrended fluctuation analysis (DFA, Peng et al., 1993). Briefly, this method forms an accumulated sum of the time series, sections it into windows ranging in length from 4 to N/4 data points (where N is the total number of data points in the time series) and the log of the size of the fluctuations for a given window size is plotted against the log of the window size. The slope of this line – alpha – is the value returned by the DFA algorithm. This method avoids the spurious detection of correlations that are artifacts of nonstationarity in the time series (Chen, Ivanov, Hu, & Stanley, 2002). An alpha value of 0.5 corresponds to a white noise process; alpha greater than 0.5 and less than (or equal to) 1.0 indicates persistent long range correlations; alpha less than 0.5 indicates persistent long range anti-correlations.

The mean, SD, CV, and alpha for each variable were calculated for each trial. The effects for each dependent variable were then examined using a 2 way (leg by speed) repeated measures ANOVA. Non-linear regression was performed on the DFA results to test for the presence of U-shaped curves. Post hoc analysis was carried out using the Tukey post hoc test and results are reported as significant if $p < 0.05$.

3.3 Results

Table 3.1 provides an overview of the ANOVA results. Persistent long range correlations were present in all of the gait cycle variables examined. With the exception of alpha of step interval, there were no significant differences in mean, SD, CV or alpha between the right and left legs for any of the gait cycle variables investigated. In the case of step interval, the long range correlations for the right leg were slightly but significantly higher than for the left leg (0.68 vs 0.71 respectively).

Table 3.1: Main effects of speed on all dependent variables from Analysis of Variance

	Mean		SD		CV		DFA	
	F(4,40)	p	F(4,40)	p	F(4,40)	p	F(4,40)	p
Stride Interval (s)	217.56	<.05	14.83	<.05	6.37	<.05	3.12	<.05
Stride Length (m)	323.24	<.05	1.54	n.s.	5.95	<.05	3.12	<.05
Step Interval (s)	216.03	<.05	22.89	<.05	9.75	<.05	5.01	<.05
Step Length (m)	497.94	<.05	1.92	n.s.	10.43	<.05	1.50	n.s.
Vertical Impulse (N.s)	155.13	<.05	8.99	<.05	3.18	<.05	5.16	<.05
Force at First Peak (N)	87.21	<.05	7.84	<.05	2.0	n.s.	0.14	n.s.
Force at Second Peak (N)	105.91	<.05	10.57	<.05	6.55	<.05	0.77	n.s.
Force at Trough (N)	271.81	<.05	7.94	<.05	14.22	<.05	0.47	n.s.

Walking speed had a significant effect on all gait cycle variables, with linear decreases occurring for stride and step interval, impulse, and trough force (Figure 3.3). Linear increases in mean values occurred for stride and step length, peak first and second peaks. In all cases post-hoc testing revealed significant differences between each pair wise combination.

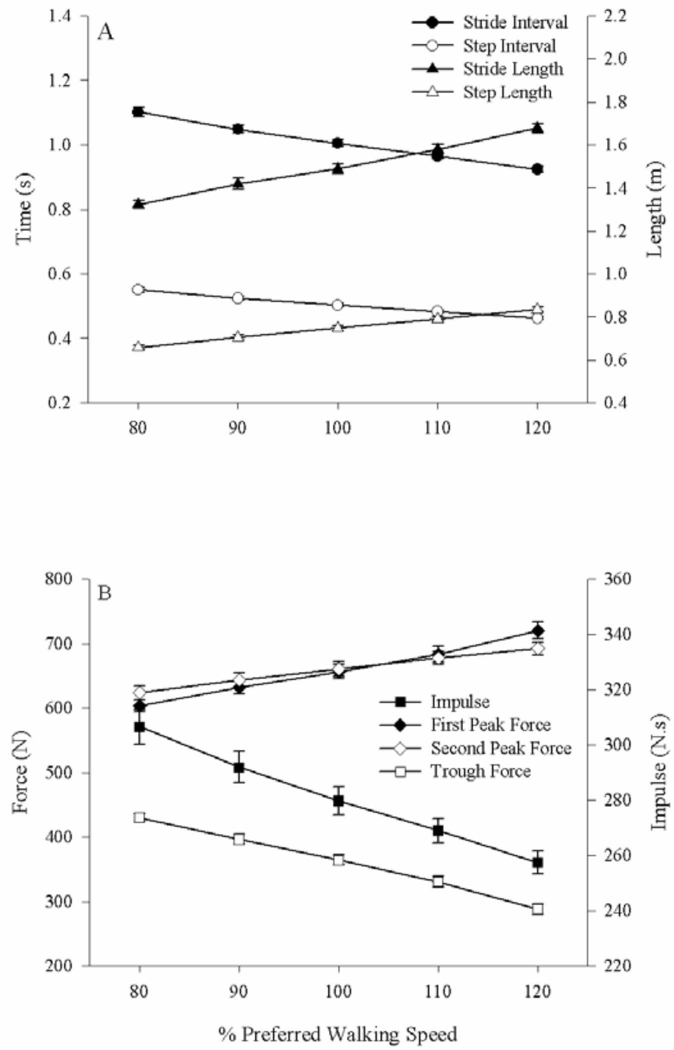

Figure 3.3: Group mean values for (A) stride interval, step interval, stride length and step length versus walking speed; and for (B) impulse, first and second peak forces and trough force versus walking speed.

3.3.1 Amount of Variability

There were significant decreases in the SD of stride interval, step interval, and impulse. As can be seen in Figure 3.4A, the most dramatic decrease for these variables was between 80 and 90% of PWS. Post-hoc testing showed that this difference was significant for all 3 variables. The SD of the forces at first and second peaks as well as trough force increased with speed. Post-hoc testing showed significant increases from 80, 90 and 100% of PWS to 120% of PWS for these 3 variables.

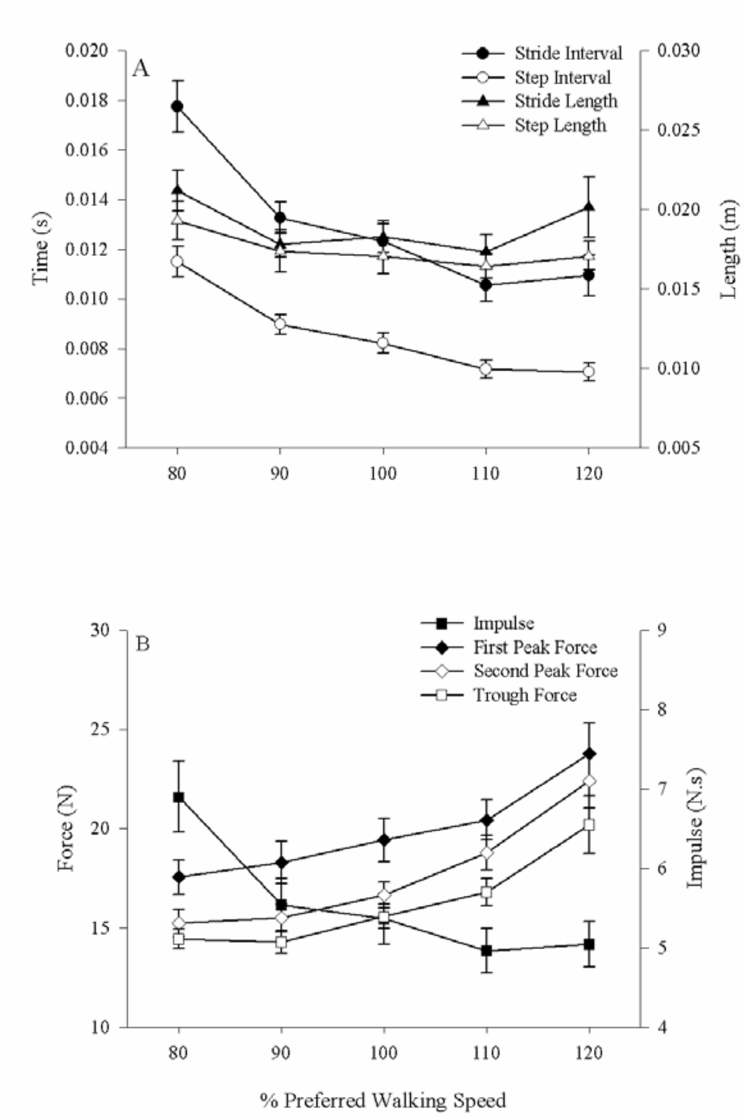

Figure 3.4: Group standard deviation of (A) stride interval, step interval, stride length and step length versus walking speed; and (B) impulse, first and second peak forces and trough force versus walking speed.

In Figure 3.5 it can be seen that there were decreases in CV for step and stride interval, step and stride length and impulse, again with the slope of the curve being steepest between 80 and 90% of PWS in all cases. For step and stride interval, and stride length, post-hoc tests revealed significant differences between 80% of PWS and all other speed conditions. CV of step length decreased significantly between 80% and the three fastest walking speeds, where as for impulse there was a significant decrease from 80 to 110% of PWS only. CV of both second peak force and trough force increased, with significant increases occurring from the slowest three speeds to the fastest speed for second peak force. In the case of trough force, there was a significant increase in CV from all speeds to 120% of PWS, as well as from 90 to 110% of PWS.

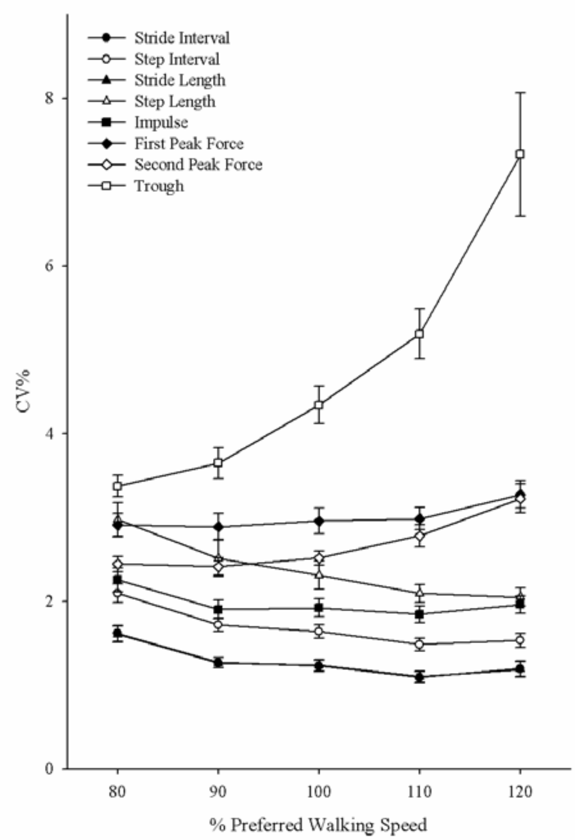

Figure 3.5: Group CV for all variables versus walking speed.

3.3.2 Structure of Variability

Significant speed effects were seen for step and stride interval, stride length, and impulse. These variables and step length all followed a U-shaped pattern of change with increments in walking speed (Figure 3.6). For stride interval and length there was a significant reduction in the strength of long range correlations from 80 to 110% of PWS. For step interval, there was a significant decrease from 80 to both 100 and 110% of PWS. In addition, for impulse, there were also significant differences between 80 and both 110 and 120% of PWS as well as between 90 and 110% of PWS.

Second order polynomial curves were fit to the DFA scaling exponents for the 5 gait cycle variables that demonstrated a curvilinear change with speed. For stride and step interval, stride and step length, and impulse, adjusted r-squared values of 88%, 97.5%, 80.8%, 98.8%, and 75.4% were observed, respectively. For all variables except impulse, the minimum fell between 100 and 110% of PWS. In the case of impulse, the minimum was between 110 and 120% of PWS. While only step interval and step length had significant quadratic components, these adjusted r-squared values indicate that the data are well fit by a 2^{nd} order polynomial, and in all cases r-squared values for linear fits were smaller.

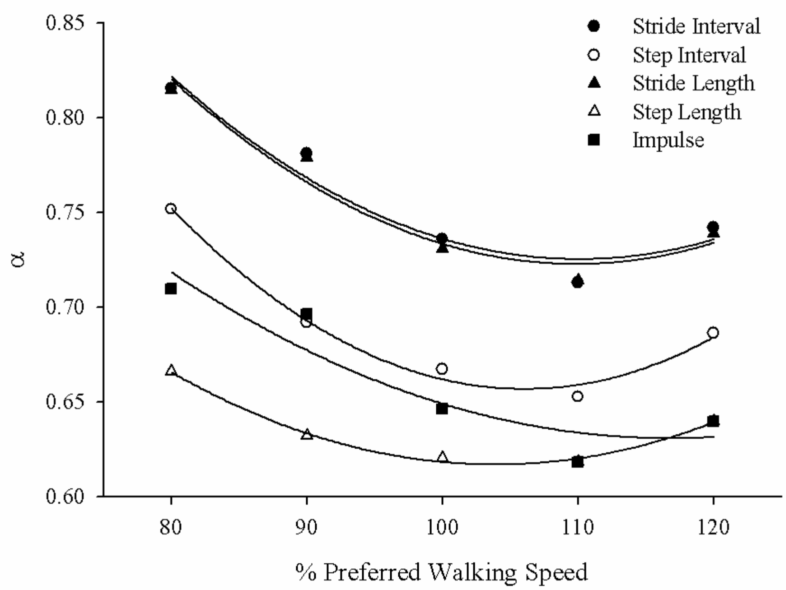

Figure 3.6: Group alpha values for stride interval, step interval, stride length, step length and impulse versus walking speed.

3.3.3 Follow-up Experiment

Observation of Figure 3.6 suggests that had a broader range of speeds been examined, the U-shaped curve for alpha would be more pronounced. In order to examine this possibility we performed a post-hoc data collection using the same protocol but over an increased range of walking speeds: +/-20 and 40% of PWS (i.e. 60, 80, 100, 120, and 140% of PWS). Data were collected from 10 subjects (age = 27.4 ± 4.0 years; height = 166.6 ± 3.8cm; weight = 62.2 ± 5.8 kg) walking for 6 min at each speed, the top speed of 140% made collecting data for a longer period of time prohibitive. In all other respects, the methods were identical to that employed in the initial data collection. The pattern of results for the post-hoc data collection was consistent with that of the original data. Because the motivation for the additional data collection was to clarify the initial findings for alpha, we present only the results from DFA. The DFA scaling exponents for the additional data across the 5 variables demonstrated U-shaped curves with speed that was similar to that of the initial data collection. Figure 3.7 shows the alpha of stride interval versus speed for both experiments separately.

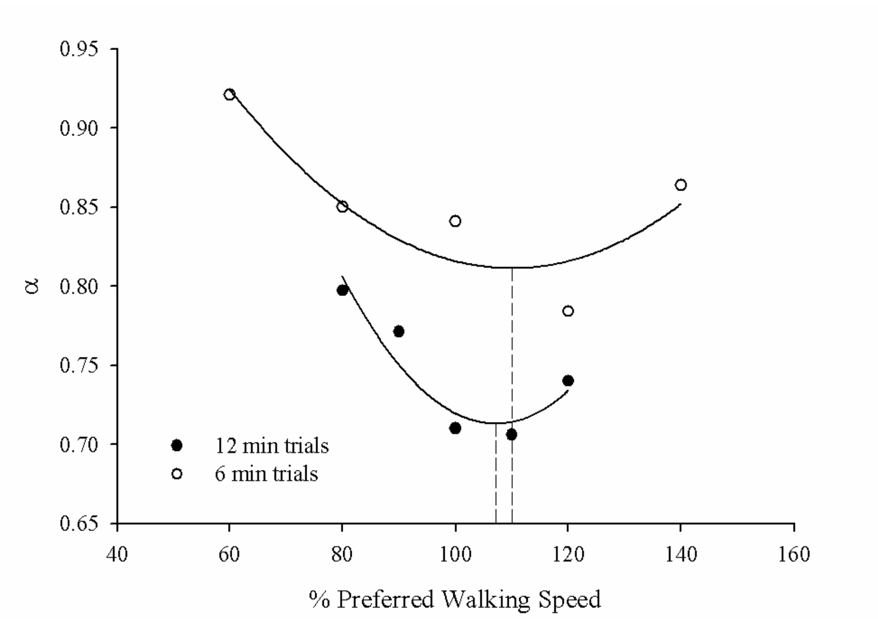

Figure 3.7: Values for alpha for stride interval for both initial and post-hoc data collections.

3.4 Discussion

In this study we examined the long range correlations in multiple gait cycle variables during walking over a range of speeds. The two main findings are that: 1) long range correlations are present in all of the gait cycle variables assessed in this study; and 2) there are distinct, U-shaped patterns of change in the strength of the correlations with speed for 5 out of 8 variable that are centered close to the preferred walking speed.

The hypothesis that PWS would have the lowest strength long range correlation is supported in 5 of the 8 gait cycle variables investigated. Stride and step interval, stride and step length, and impulse all exhibit a curvilinear change with speed, the minima consistently falling between 100 and 110% of PWS. Significant quadratic trends were seen for step interval and step length but not for stride interval and stride length (although the p-values for these variables did come close to reaching significance). On average, the strength of the long range correlations is reduced in step interval and length when compared with the stride interval and length, suggesting that the dynamics of the step may be more complex than that of the stride. Further analysis revealed that step interval and length were significantly more variable than the respective stride measures. It is possible that the increased step to step variability is compensatory, reducing the over all stride variability.

Although U-shaped curves are evident for 5 of the 8 gait cycle variables (Figure 3.6), walking at slower than preferred speeds has a more dramatic effect on the strength of long range correlations than walking at faster speeds. Figure 3.7 shows the DFA results of the stride interval for both sets of data – a similar pattern of results was observed for step interval, stride and step length and impulse. U-shaped curves with very similar values for minima are apparent for both sets of data. The alpha exponents for the post-hoc data are on average higher than those for the original data collection. This is may be due to the difference in trial length – participants walked for twice as long in the original experiment as those in the post-doc experiment (5*12 min versus 5* 6 min, respectively); effects due to fatigue and boredom are likely to be more pronounced during longer trials. Although the effects of speed are more pronounced at slower than preferred speeds, it is clear from Figure 3.7 that the strength of long range correlations increases at faster speeds. We suggest two possible explanations for the overall U-shaped trend seen for strength of long range correlations with increasing speed.

Reduced strength of long range correlations in a statistical sense indicates that any given stride is less influenced or dependent upon all preceding strides (Hausdorff et al., 1996). As such, one interpretation of these results is that the PWS is less constrained, and hence more readily adaptable than speeds faster and slower than preferred. The reduced strength of long range correlation implies an increase in complexity, or an increase in the number of available dynamical degrees of freedom (Slifkin & Newell, 1999). This also suggests that the PWS is more readily adaptable than other walking speeds.

The reduction in long range correlations at the PWS may be a result of improved stability associated with walking at resonance. When individuals walk freely (i.e. at their PWS), they naturally select a stride frequency that is the same as the predicted eigenfrequency of the leg (Holt et al., 1990, 1991). It has also been demonstrated that a rhythmic movement carried out at resonance has greater cycle to cycle reproducibility and stability (Rosenblum, & Turvey, 1988) than movements at other frequencies. Although the studies of Holt and colleagues (1990, 1991) manipulated step frequency, the speed that individuals walked at was their preferred speed. As walking speed increases, step frequency increases concomitantly in such a way that efficiency is optimized (Zarrugh et al., 1974). It is clear then that walking at resonance only happens at the PWS, even though there are preferred step frequencies for all other speeds. Thus, the

PWS is that at which the locomotor system should be most stable and the reduced strength of long range correlations at the PWS may reflect this enhanced stability.

One potentially concerning observation is that these results appear to be in conflict with those of Goodman, Riley, Mitra, & Turvey (2000), who showed that the number of dynamical degrees of freedom required to capture the dynamics of pendulum swinging was reduced at the resonant frequency. This suggests that oscillatory movement at resonant frequency improves the predictability of the movement output, which contrasts with the results for walking. However, there are a number of differences between the current study and that of Goodman et al. (2002), the most obvious of which is that the task of walking is inherently more unstable and complex than that of swinging a pendulum about the wrist (Winter, 1983). While stability may be maximized in both cases under the condition of preferred parameterization of the task, in the case of pendulum swinging the only way in which the task changes at non-preferred frequencies is that more force must be applied to swing the pendulum faster or more damping to swing the pendulum slower. The consequences of the central nervous system not compensating for the loss of stability in the wrist-pendulum system are negligible. This is clearly not the case in locomotion (Newell & McDonald, 1994). While the changing the speed of locomotion may require additional force production or damping as appropriate, the destabilizing effects of this internal perturbation also need to be accounted for. As participants are forced to walk at speeds increasingly different from preferred, it becomes necessary to more actively control the movement out put which may increase the degree of structure present in the gait cycle fluctuations. At slower speeds this may be particularly apparent as the mediolateral excursions of the center of mass increase (Orendurff et al., 2004) - keeping the center of mass over the support limb may therefore require a greater degree of active regulation.

It is apparent that U-shaped curve for DFA has a fairly shallow basin that spans the speeds of 90 to 120% of PWS (Figures 3.6 and 3.7). In terms of energetics, the basin of attraction for walking is shallow, with individuals being able to walk at speeds from ~4 to 6km/hr without a significant increase in energy expenditure (Zarrugh, 1974). Holt, Jeng, Ratcliffe and Hamill (1995) demonstrated a U-shaped curve for head stability with increasing stride frequency that also had a fairly shallow basin, with the standard deviation of head trajectory in the vertical plane remaining constant from around 75 to 100% of predicted preferred stride frequency. These and the current results, therefore, may reflect the locomotor system's ability to optimize walking and in general be able to adapt walking effectively over a range of speeds.

When significant changes were observed for CV, there was a decrease in the majority of variables (stride interval, step interval, stride length, step length, impulse). The implication of this result is that both the kinematic and kinetic gait cycle variables become more consistent as speed increases (Winter, 1983). The only variables where there was an increase in CV with speed were the second peak and trough of the VGRF. For both of these variables there was a large increase from the preferred to the fastest walking speed (see Figure 3.5). This is consistent with the findings of Masani, Kouzaki and Fukunaga (2002) who showed increased CV for both the first and second peaks of vertical ground reaction force, although the variability of trough force was not investigated. The CV of the second peak in the vertical ground reaction force was remarkably low compared with the first peak and trough. This may be because this peak is related to the toe off component of the gait cycle which, the locomotor system would arguably be concerned with controlling (Li & Hamill, 2002).

In summary, the results of this experiment show that long range correlations are present in a range of gait cycle variables during unconstrained walking on a treadmill. In 5 out of 8 gait variables investigated, the DFA scaling exponents followed a U-shaped curve as a function of walking speed, the minima of which fell between 100 and 110% of PWS. These findings are consistent with the proposition that reduced strength of long range correlations at preferred locomotion speeds is reflective of enhanced stability and adaptability at these speeds.

Chapter 4

Running Speed Influences on Gait Cycle Variability

4.1 Introduction

Human locomotion primarily takes the form of either walking or running, with both gaits involving coordination of multiple mechanisms and couplings of the neuromuscular system, including the motor cortex, cerebellum, basal ganglia and feed back from vestibular, visual and peripheral receptors. Although in healthy individuals, the level of variability present in the stride interval of the gait cycle is typically very low (coefficient of variation ~ 3%), it has become increasingly apparent that the variability associated with the gait cycle contains considerable structure that is revealing about the control of locomotion. In the last decade, detrended fluctuation analysis (DFA) has become a widely-used method for detection of long range correlations in noisy, nonstationary time series. In the case of human walking, long range correlations in the stride interval have been shown to extend over thousands of strides and are robust with respect to walking velocity (Hausdorff et al., 1996). Additionally, it has recently been shown that the fluctuations contained within the step length and step frequency time series of walking also exhibit fractal behavior (Terrier et al., 2005).

The results of Experiment 1 demonstrate that the long range correlations present in the stride interval time series show systematic changes with walking speed such that the strength of correlation increases at speeds both faster and slower than preferred. While this finding is consistent with research that indicates that preferred walking speeds are associated with minimal energy expenditure, (e.g., Bobbert, 1960; Margaria, 1976;

Zarrugh, Todd & Ralston, 1974), there appears to be no economically preferred running speed, with the energy cost of running per unit mass remaining essentially constant with increasing speed for a given distance (Margaria, Cerretelli, Aghemo & Sassi, 1963). Similarly, studies investigating systematic changes in gait cycle variability during running gait with speed have shown linear patterns of change (e.g. Belli et al., 1995) whereas there is evidence to suggest that the variability of walking changes as a U-shaped function with speed (e.g. Yamasaki et al., 1984, Yamasaki et al., 1991). Thus the question remains, what is the pattern of change of long range correlations with speed in the running gait?

Decerebrated cats transition from a walking to a running gait with either an increase in the amplitude of stimulation delivered to the locomotor region of the mid brain or an increase in the belt speed of the treadmill the cat moves upon (Shik, Severin & Orlovsky, 1966). Thus, it appears that the main difference between walking and running with regard to higher centers of the central nervous system is related to the excitability of the mid brain. Recently it has been shown in humans that the primary difference between walking and running with respect to timing of muscle activation relates to the differences in the duration of stance for these two gaits (Capellini, Ivanenko, Poppele & Laquaniti, 2006). The common core hypothesis of Zehr (2005) suggests that rhythmic locomotor tasks such as walking running and swimming share common central neural control mechanisms. Thus, while there are differences between walking and running related to the timing of muscle activations, overall mechanics, and feedback from the periphery, it is likely that the same basic neural circuitry is used for the generation of walking and running patterns (Capellini et al., 2006; Zehr, 2005).

It is hypothesized based on the results of Experiment 1 that long range correlations will follow a U-shaped pattern of change with speed. As with walking, it is predicted that the presence of long range correlations will not be limited to the stride interval, and, that the influence of speed on the scaling behavior of gait cycle fluctuations will be similar across the kinematic and kinetic variables investigated.

4.2 Methods

4.2.1 Subjects

Eleven female volunteers from The Pennsylvania State University between the ages of 22 and 27 years of age (average = 24.5 ± 1.8 years) were recruited for the study. The average height and mass of the subjects was 165.3 ± 4.0 cm, and 57.7 ± 3.6 kg, respectively. The subjects were recreational runners who ran a minimum of 15 miles per week. All subjects provided informed consent and all procedures were carried out according to the ethical guidelines laid down by the Institutional Review Board of The Pennsylvania State University.

4.2.2 Apparatus

The apparatus consisted of two piezoelectric Kistler force plates embedded in a Gaitway instrumented treadmill, one in front of the other. Vertical ground reaction force (VGRF) was measured by 8 force sensors located in the corners of each force plate and sampled at a rate of 250Hz. The Gaitway software collected the force data from the 8 force sensors. The treadmill had a speed range of 0.8 to 20 km/hr, with the smallest increment in speed being 0.1km/hr.

4.2.3 Tasks and Procedures

Prior to data collection, participants were given a 45min familiarization session on the treadmill. During this time the participants were able to walk or run on the treadmill at speeds they felt comfortable with. Experimental sessions involved manipulation of 5 levels of running speed (80, 90, 100, 110 and 120% of preferred speed). The first 10 min of each experimental session was used as a warm up/treadmill adaptation period. The preferred running speed was established during this period by initially having the participant run at a relatively slow speed (based upon observation by the experimenter during the 45 min adaptation session). The investigator slowly increased the speed in 0.1km/hr increments until the subject reported that they were running at a speed they would feel comfortable running at continuously for approximately 45 min. The speed was then increased by approximately 1.5km/hr and then decremented by 0.1km/hr until a comfortable running speed was re-established. This procedure was then repeated. In the majority of participants there was a good match between these two speeds on both the first attempt to establish preferred running speed – in this case the average of these speeds was taken to be the preferred running speed. If a similar speed could not be established during this procedure, (greater than 0.4km/hr different) the process was repeated until a close match was achieved. During this process of establishing the preferred running speed, the participants were prevented from viewing the speed at which they were running.

Once the preferred running speed was established, the participants performed 8 min trials at 80, 90, 110 and 120% of this speed. Speeds were presented randomly with approximately 650 stride intervals captured per trial. Subjects were given at least 2 min and up to 10 min to recover between trials.

4.2.4 Analysis of Force Plate Data

Custom written MATLAB software was used to compute and process the vertical ground reaction force (VGRF) data. The center of pressure and ground reaction force data were filtered in forward and reverse directions with a 2nd order Butterworth filter with a cut off of 30Hz. There was noise in the force plate data, some of it introduced by vibration of the treadmill; the cutoff frequency was selected to maintain the subject generated signal but to eliminate signal due to the treadmill's high frequency vibrations. A threshold of 30N was used for the identification of the start and end of each foot fall - signal below this threshold was considered to be noise.

The gait parameters investigated were as follows: stride interval, step interval, stride length, step length, duration of foot contact, step impulse, peak VGRF, and time to peak VGRF. Stride interval was defined as starting with the onset of the heel strike of one foot and finishing with the onset of the next heel strike of the same foot. Step interval was defined as starting with the heel strike of one foot and finishing with the subsequent heel strike of the other foot. Step length was calculated by multiplying stride interval by the average belt speed and adding (subtracting) the difference in distance between the successive heel strikes of the limb. Stride length was calculated similarly, multiplying stride interval by the average belt speed and adding (subtracting) the difference in distance between the successive heel strikes of the limb. The duration of contact for a particular foot was taken to be the time from heel strike to toe off for that foot. Impulse is the area under the force time curve, peak VGRF was taken as the maximum vertical ground reaction force produced during stance phase, with time to peak VGRF being the time taken to reach peak VGFR from initial contact (heel strike).

4.2.5 Data Analysis

The mean, standard deviation (SD), coefficient of variation (CV), and the strength of long range correlations were calculated for the time series of each dependent variable. The SD provides a description of the absolute amount of variability in a given time series whereas the CV provides a description of the amount of variability in the time series normalized by the mean of the time series. These measures of variability refer only to the amount of variability of the time series and do not reflect the structure of stride to stride fluctuations.

Long range correlations on the other hand provide a measure of the structure of variability of the stride interval time series. Stronger correlations would indicate a more predictable, regular time series whereas weaker correlations would indicate a less predictable time series where any given stride interval is less dependent on the stride intervals preceding it. Long range correlations of the time series were calculated using detrended fluctuation analysis (DFA; Peng et al., 1993). Briefly, this method first forms an accumulated sum of the time series, sections it into windows ranging in length from 4 to N/4 data points (where N is the total number of data points in the time series) and the log of the size of the fluctuations for a given window size is plotted against the log of the window size. The slope of this line – alpha – is the value returned by the DFA algorithm. One of the main advantages of using this method over others (such as spectral analysis) is that it avoids the spurious detection of correlations that are artifacts of nonstationarity in the time series (Chen, Ivanov, Hu & Stanley, 2002). An alpha value from the DFA of 0.5 corresponds to a random walk; alpha greater than 0.5 and less than (or equal to) 1.0 indicates persistent long range correlations (e.g., a long stride interval is likely to be followed by a long stride interval whereas a short stride interval is more likely to be followed by another short stride interval); alpha less than 0.5 indicates persistent long range anti-correlations (e.g., a long stride interval is likely to be followed by a short stride interval and vice versa); and an alpha of 1.5 indicates brown noise.

The mean, SD, CV, and DFA scaling exponent for each variable were calculated for each trial. The effects for each dependent variable were then examined using a 2 way (leg by speed) repeated measures ANOVA. Post hoc analysis was carried out using the Tukey post hoc test. In addition, a non-linear regression was performed on the DFA results to test for the presence of U-shaped curves.

4.3 Results

A typical stride interval time series is shown in Fig 4.1. For all variables recorded over all speeds in the running trials, there was no significant difference ($p > .05$) between the left and right legs with the exception of DFA of stride interval. The long range correlations were slightly weaker on average in the right leg (alpha = 0.823) than in the left leg (alpha = 0.829). Table 4.1 provides a summary of the statistical results for all running variables as a function of speed, and significant p-values indicate a significant main effect for speed.

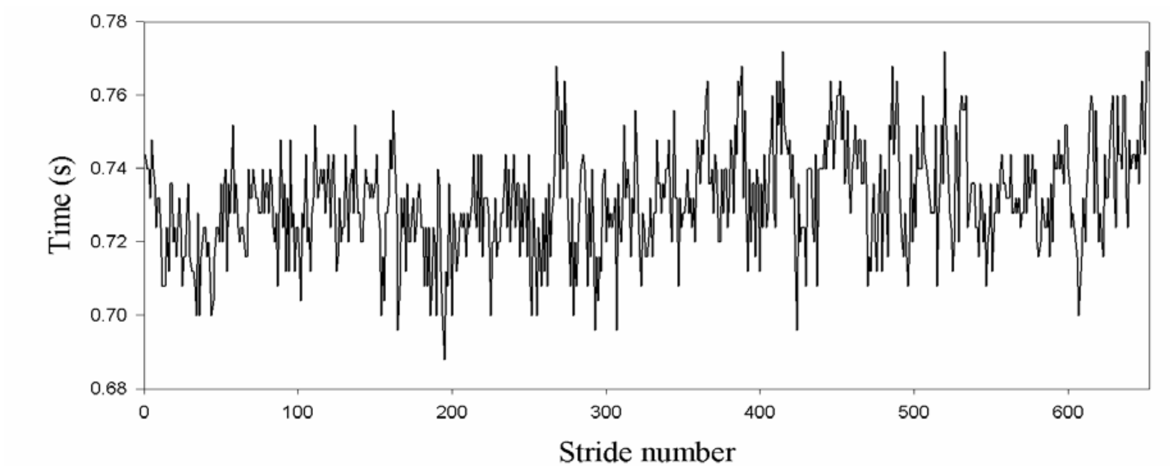

Figure 4.1: Representative stride interval time series for running at the preferred speed.

4.3.1 Mean Values

There was a significant effect of speed on the mean of all gait cycle variables (Figure 4.2). Significant decreases in average values occurred for stride interval, step interval, impulse, duration of contact and time to peak VGRF. For the remaining variables there were increases in average values with speed.

Table 4.1: Main effects of speed on all dependent variables from Analysis of Variance

	Mean		SD		CV		DFA	
	F(4,40)	p	F(4,40)	p	F(4,40)	p	F(4,40)	p
Stride Interval (s)	43.40	0.01	6.71	0.01	3.65	0.05	3.02	0.05
Stride Length (m)	525.28	0.01	2.10	ns	9.86	0.01	2.32	ns
Step Interval (s)	43.28	0.01	9.46	0.01	4.36	0.01	3.91	0.01
Step Length (m)	524.17	0.01	0.29	ns	15.30	0.01	2.12	ns
Vertical Impulse (N.s)	35.67	0.01	10.18	0.01	5.13	0.01	2.52	ns
Duration of Contact (s)	325.80	0.01	11.69	0.01	5.09	0.01	6.15	0.01
Active Peak VGRF (N)	61.90	0.01	2.82	0.05	5.31	0.01	6.33	0.01
Time to Active Peak VGRF (s)	68.35	0.01	2.26	ns	0.49	ns	2.13	ns

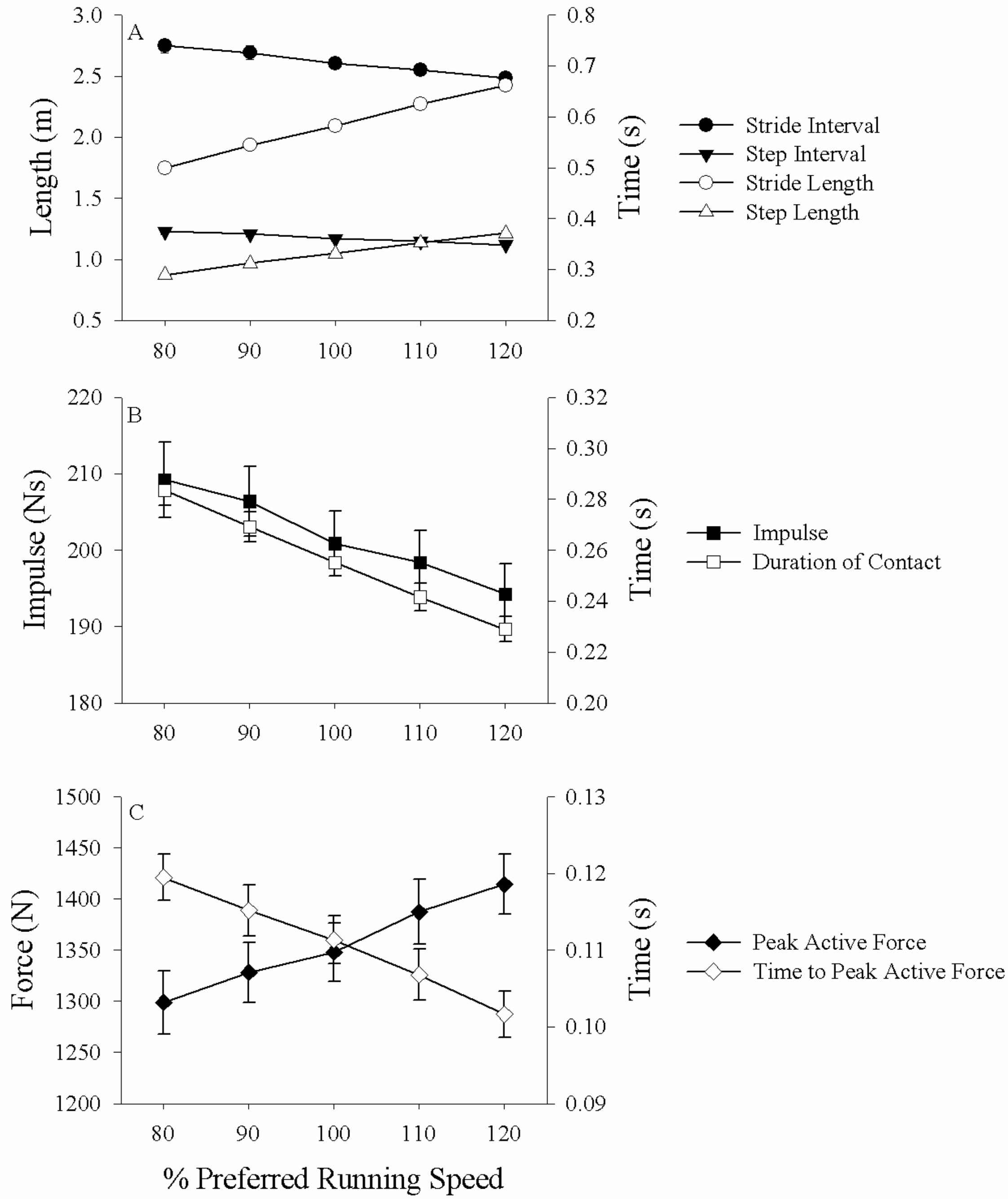

Figure 4.2: (A) Mean step and stride intervals and lengths for all speeds; (B) Mean impulse and duration of contacts for all speeds; (C) Mean peak active force and time to peak active force for all speeds.

4.3.2 Amount of Variability

SD in general followed a similar pattern of change with speed as the mean values, (Figure 4.3), with decreases occurring for stride interval, step interval, impulse, duration of contact and time to peak VGRF. In addition, SD of peak active VGRF also decreased with speed. While there was a trend for increasing variability with speed for increases in stride and step length with speed, this failed to reach significance. When variability was normalized by mean values, CV decreased significantly with speed for all variables, except time to peak VGRF where there was no change (Figure 4.4).

4.3.3 Structure of Variability

For stride interval, step interval, stride length, step length and impulse the strength of long range correlations tended to follow a U-shaped trend with speed (Figure 4.5). Post-hoc tests revealed significant decreases in strength between 80 and 100% for both stride and step interval. In order to confirm the U-shape of the curves, second order polynomials were fit to the DFA results. Significant ($p < 0.5$) r-squared values were obtained for step interval (88%), step length (95.9%) and impulse (87.7%). The curves for stride interval (72.2%) and length (62.7%) came close to reaching significance with p values of 0.075 and 0.098 respectively. In the case of both duration of contact and peak active force, the strength of long range correlations decreased with increasing speed.

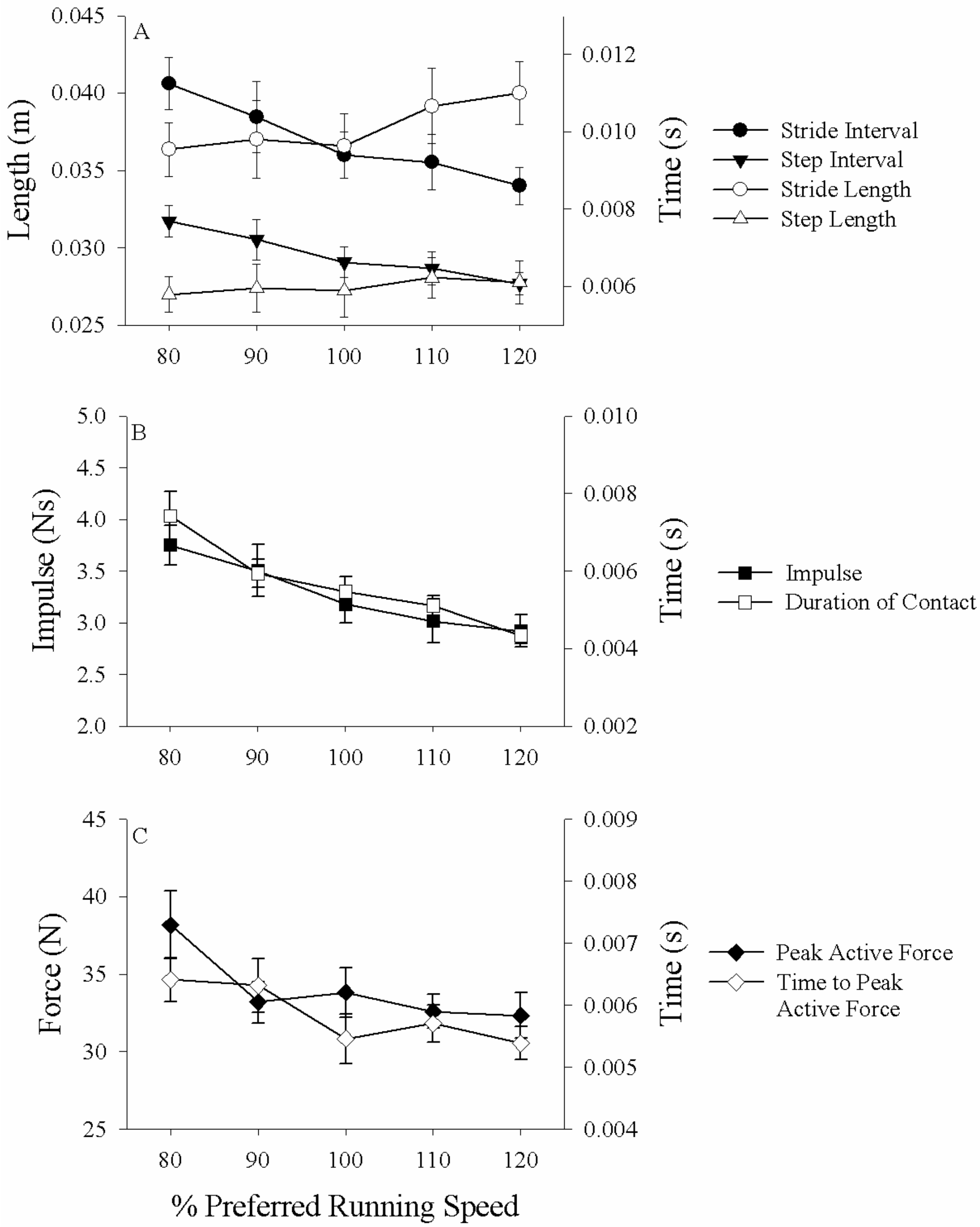

Figure 4.3: (A) SD of step and stride intervals and lengths for all speeds; (B) SD of impulse and duration of contacts for all speeds; (C) SD of peak active force and time to peak active force for all speeds.

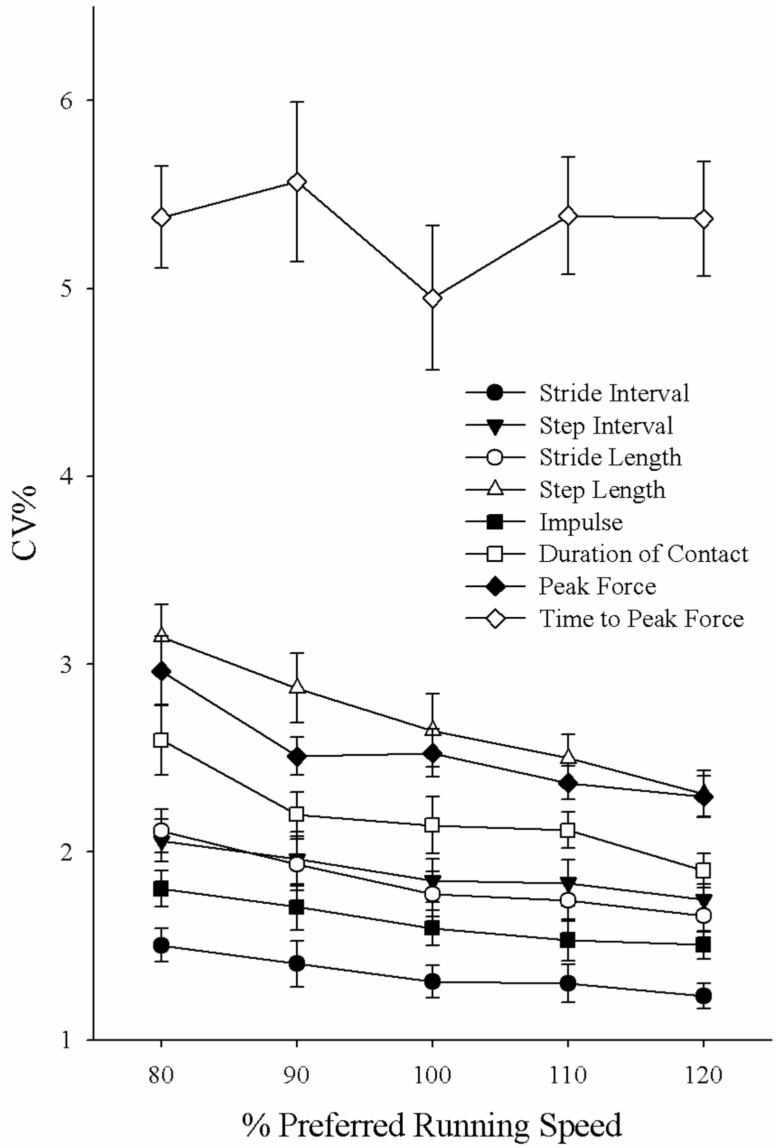

Figure 4.4: CV of all gait cycle variables versus speed

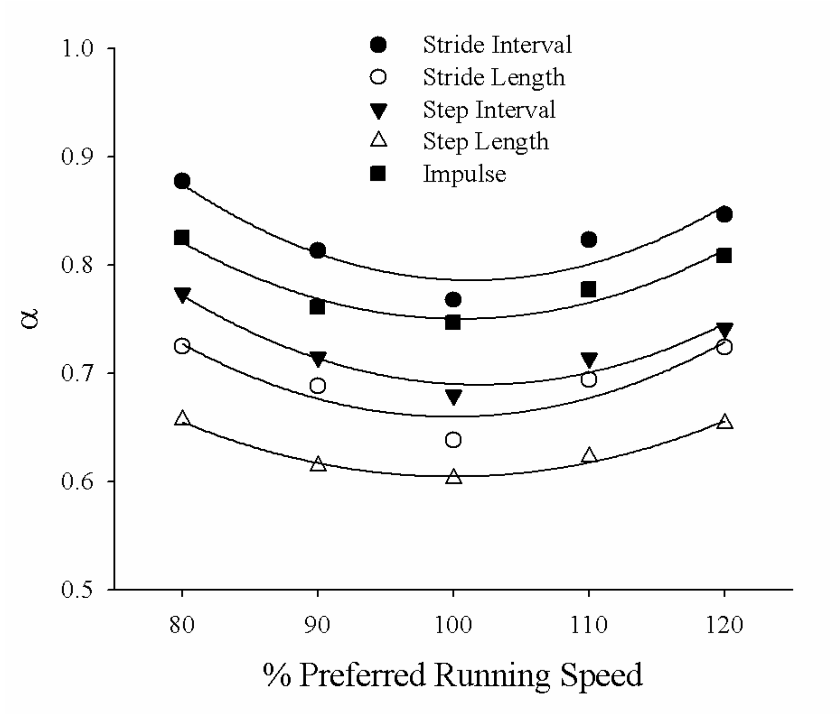

Figure 4.5: Second order polynomial fit for DFA versus speed for stride interval, step interval, stride length, step length and impulse.

4.4 Discussion

This is the first study to examine the long range correlations in both kinematic and kinetic gait variables during running. The results show that long range correlations are present in temporal (stride and step time), spatial (stride and step length) and kinetic (impulse and peak VGRF) variables of the gait cycle. The values for DFA obtained from this treadmill experiment are generally smaller than those previously reported by Hausdorff and colleagues (1996) for walking over ground in healthy young adults. However, for all the gait parameters of all participants across all speeds, the DFA scaling exponents generally took on a value between 0.5 and 1 and on average fell well within this range (see Figure 4.5). Furthermore, these long range correlations are robust with respect to speed and show a similar pattern of change with speed for many of the gait parameters measured.

The U-shaped DFA curves for stride interval and length, step interval and length and impulse all had minimum strength of long range correlations at the preferred running speed. It was hypothesized that these DFA results are related to the dynamics associated with preferred modes of behavior and the enhanced adaptability that is present in the scaling of a multiple degree of freedom system with running speed. Statistically speaking, a reduction in the strength of long range correlations means that there is a reduction in the overall time dependence of a variable. In other words, in the case of stride interval, any given stride interval is less dependent on the stride intervals preceding it. It appears that there is an increased degree of flexibility or adaptability at the preferred running speed in that a range of variables (e.g. stride interval, step length, impulse) have significantly less sequential dependencies at this speed. This is consistent with the hypothesis that the system may be more stable and better able to recover from perturbations given the flexibility afforded by running at the preferred running speed.

It has been suggested that the long range correlations present in the stride interval of human walking relate to the integrity of the neuromuscular apparatus, with a breakdown in long range correlations being associated with both aging and disease (Hausdorff et al., 1997). Prevailing definitions of physiological "complexity" relate to both the number of elements or component parts, and their interactions (e.g., Lipsitz & Goldberger, 1992; Pincus, 1994; Vaillancourt & Newell, 2002). Examination of physiological complexity alone is insufficient for the understanding of intentional behavior such as that of locomotion where the motor output complexity is influenced in large part by the task dynamics (cf. Vaillancourt & Newell, 2002). Measures of predictability or regularity, such as DFA, when considered within the context of a given task, provide a measure of system adaptability rather than just a description of the system per se (e.g., Hausdorff et al., 1996; Hausdorff et al., 1999; Newell & Corcos, 1993; Slifkin & Newell, 1999; West & Scarfetta, 2003).

Thus in the current study we have shown that at preferred running speeds the structure of the variability of the stride interval becomes less regular when compared with speeds faster and slower than preferred. The increased regularity seen at non-preferred speeds may relate to the reduced availability of degrees of freedom away from preferred running speed. Slifkin and Newell (1999) suggested that isometric force production is optimized in terms of dynamical degrees of freedom in a mid-range of force production and that the extrema of very small and very large force production requirements acted as boundary conditions or constraints for the task. In the case of locomotion, increased running speed beyond the preferred speed reduces the number of dynamical degrees of freedom. One example of this is that with increasing speed muscles are working outside their normal range and closer to their maximum capacity (e.g. Hreljac, 1995; Neptune et al., 2005; Prilutsky & Gregor, 2001).

Dynamical degrees of freedom refer to the number of first order differential equations necessary for complete description of the behavioral dynamics and can be estimated by measures of dimension (e.g., Daffertshofer, Lamoth, Meijer, & Beek 2004; Dingwell & Cusumano, 2000; Mitra, Amazeen & Turvey, 1998). Though it is assumed that the integer dimensions of locomotion remains constant with speed (e.g. Dingwell & Marin 2006), given the fractal nature of locomotion it is more appropriate to think in terms of fractional rather than integer dimension (Bassingthwaighte, Liebovitch & West, 1994). In terms of adaptability, increased dimensionality (or dynamical degrees of freedom) implies that there exist a greater number of viable solutions to the coordination problem of running.

Contrasting with this interpretation are the findings of Goodman et al. (2000) who showed that during pendulum swinging in the upper limb the number of dynamical degrees of freedom was decreased and predictability was minimized at the preferred or resonant frequency. While pendulum models are generally not applicable to the running gait, it is reasonable to suspect that the dynamical properties of the multilink lower limb to a large extent determine the naturally adopted stride frequency during running. Stride frequency was not directly manipulated in this experiment, however, there was a significant increase in stride frequency with speed. It is known that for a given speed, participants naturally adopt the most efficient stride frequency (Zarragh et al., 1974), thus manipulation of speed cannot be expected to have the same influence on gait variables as manipulation of stride frequency. Furthermore, locomotion is a multiple degree of freedom problem that occurs against a background of other control processes, such as posture and balance control (Newell & McDonald, 1994; Winter 1983). In contrast to this, pendulum swinging studies generally are carried out while participants sit in a chair with back support and movements are constrained to motion of a single mechanical degree of freedom.

The speed that a given distance can be comfortably covered by an individual will depend on their training status and will be subject to change as training status improves. Additionally, it has been shown that there is no energetically optimal running speed

(Margaria et al., 1963). Furthermore, as running speed increases there is a concomitant increase in knee flexion angle (Mann & Hagy, 1980). Consequently, the effective length of the “pendulum” motion of the lower limb is neither constant across the gait cycle nor across different speeds. Thus, in the case of running, interpreting the influence of energetics and limb dynamics on the changing strength of long range correlations with speed is not as straight forward as it is for walking, where the effective limb motion does not change length to such an extent.

That long range correlations are present in a range of gait cycle variables and that they generally follow a similar trend with increasing speed maybe be a reflection of the inter-relatedness of these variables. For example, by definition a stride is the addition of two consecutive steps, thus it is not surprising that a similar pattern of change is present for both step and stride dynamics. Changes in running velocity occur through modulation of both stride interval and length. An increase in running velocity could be achieved via lengthening the stride alone, shortening the interval between strides alone, or some combination of the above. As can be seen in Figure 4.2A, mean stride length increases whereas mean stride time decreases with increasing speed, thus during unconstrained running there is simultaneous modulation of the temporal and spatial aspects of the stride to achieve a given running speed. Similar interactions can be seen between other variables – for example, a shorter step interval reduces the total time the foot spends in contact with the ground, which will affect the impulse, as will the peak VGRF applied to the ground. It is clear that changes in one variable with speed do not happen in isolation. It is possible that the common pattern of changing strength of long range correlations in these different gait cycle variables across the different speeds is a reflection of the changing influence of different inputs (e.g. proprioceptive feedback - Hreljac, 1995; Raynor, Yi, Abernethy & Jong, 2002) to the locomotor apparatus.

In contrast to the U-shaped curves seen for DFA, the mean, SD and CV of all gait variables changed in a linear fashion with increments in speed. With the exception of time to peak active force, there was a significant decrease in relative variability with speed for all variables. SD of stride and step length tended to increase with increasing running speed, however, this pattern of change (which did not reach significance) was probably driven by the significant increase in mean stride and step length with running speed. As anticipated the CV is very low (typically under 3%) for all gait variables, again with the exception of time to peak active force. Generally, therefore, it appears that the kinematic and kinetic variables associated with running become less variable in the sense of the dispersion of the distribution with increased running speed.

One interesting question that arises out of the current findings is why do the long range correlations for some of the gait variables follow U-shaped curves where as for other variables the pattern of change is different? It is reasonable to assume that certain gait variables are more influenced by factors external to the neuromuscular system (such as footwear) than others, and that these factors may mediate changes in long range correlations with speed. For example, it has been shown that softer shoes result in an increased time to peak active force (Clarke, Frederick & Cooper, 1983). The softness of footwear, however, is unlikely to have a significant impact on variables such as stride interval and length. Conceptually this is similar to the findings of Terrier et al., (2005) who show that use of a metronome influences the long rang correlations in step frequency but not step length.

In summary, these results show that long range correlations are present during running in a range of gait cycle parameters and that in the majority of variables recorded the strength of correlation followed a U-shaped function with running speed. This set of findings supports the general hypothesis that the long range correlations associated with human locomotion are ubiquitous. The proposition that the preferred running speed is associated with an increase in dynamical degrees of freedom and enhanced flexibility was also supported. This is not only the case for the traditionally examined stride interval, but also for spatial and kinetic variables such as stride length and impulse.

5.1 Introduction

It has been suggested that walking and running at speeds at and around preferred transition speeds results in destabilization of the gait cycle (e.g., Brisswalter & Mottet, 1996; Diedrich & Warren, 1995, 1998). The results of Experiments 1 and 2 have shown that there are long range correlations present in the stride interval of both running and walking gaits, the strength of which appears to be speed dependent. In particular, it has been shown that the strength of long range correlations is increased both at fast walking speed and at slow running speeds. The changes in the size of the DFA scaling exponent alpha with speed of locomotion in this dissertation have been interpreted as being reflective of the enhanced adaptability and stability associated with locomoting at preferred speeds. Thus, the purpose of this study is to investigate walking and running at and around preferred transition speeds on both the amount and structure of kinematic and kinetic variability of the gait cycle.

The concepts of stability and variability are linked, at least in part, via the concepts of self organization and attractor dynamics. The HKB model (Haken, Kelso, & Bunz, 1985) provides a classic example of this – increasing variability in the phase relationship of the fingers (i.e. the standard deviation of the relative phase of the fingers) as frequency of finger oscillation is increased is considered to be demonstrative of critical fluctuations in the order parameter, which are in turn associated with loss of stability of the coordination mode (in this case the anti-phase coordination pattern). In the case of locomotion, several groups of researchers have used the standard deviation (SD) of relative phase to examine the stability of both inter- and intra-limb coordination at and around gait transition speeds (Diedrich & Warren, 1995; Kao et al., 2003; Seay et al., 2006).

Diedrich and Warren (1995) examined the variability of discrete relative phase of the hip-ankle and knee-ankle joint angles as participants walked and ran over a range of constant speeds encompassing both the preferred walk-run and run-walk transition speeds. In this protocol participants did not transition between walking and running, rather walking trials proceeded from slowest to fastest speed, running trials from fastest to slowest, and the variability of each gait in the region of the known transition speed was examined. In general there was greater relative phase variability prior to gait transitions than in the transition region. In the case of running, SD of relative phase continued to decrease with increasing speed after the transition region, whereas for walking SD of relative phase began to increase again as gait speed was reduced after the transition region. Interestingly, the SD of ankle-hip relative phase appears to be greater in the running gait than walking over all speeds, implying that had the transition from walking to running occurred, there would have been an increase in the variability of relative phase after the transition, rather than the predicted decrease. Conversely, for the run to walk transition there would be a marked decrease in relative phase variability of these joints after the transition. However, the running and walking values of ankle-knee relative phase variability cross over immediately after the transition region suggesting that transitioning between walking and running would reduce variability of relative phase of these two joints.

Diedrich and Warren (1995) did not find a significant pattern of change in the variability (SD) of stride length or frequency. However, Brisswalter and Mottet (1996) investigated the variability (CV) of stride duration in the region of the walk to run transition and found increased variability during their three speed conditions preceding the transition. Additionally, they showed a decrease in CV of stride duration immediately following the transition to running.

In order to more accurately replicate the protocol used in the original HKB experiment (Haken et al., 1985), Kao et al. (2003) investigated hip-ankle and knee-ankle joint angle coordination during gait transitions using a continuous change in treadmill speed. Counter to the findings of Diedrich and Warren (1995), no change in the standard deviation of hip-ankle and knee-ankle continuous relative phase (CRP) prior to either the walk to run (W-R) or run to walk (R-W) transition was observed. It was shown, however, that variability of the hip-ankle relative phase was larger before than after the walk to run transition. Seay et al. (2006) used a similar method to that of Kao et al. (2003) and investigated variability in CRP of both intra- and inter-limb coordination using limb segment angles. They found no increase in coordination variability before the transition from walking to running was made. Rather, their results showed that CRP of running at the slowest speed is greater than that of walking at the fastest speed. Inspection of their Figures 3 and 5 also indicate that in general variability of CRP decreases with increasing gait speed. Collectively, these observations suggest that it is increasing speed that fundamentally causes a reduction in coordination variability rather than gait transitions per se.

Thus, it is unclear from these studies whether gait transitions are driven by a loss of stability of coordination patterns. This may be because the methods employed are inconsistent both in terms of how data were collected (joint angles vs. segment angles), and in terms of how data were analyzed (continuous vs. discrete relative phase, CV of stride interval). Because it is the movement about the joints that is controlled by the neuromuscular apparatus, the joint angles as defined by limb segments provide a more functionally relevant source of data than, for example, the angle a limb segment makes with some external reference (such as vertical or horizontal references – Diedrich & Warren, 1995; Seay et al., 2006). Furthermore, measures such as SD provide only a global, static representation of the amount of variability present in a time series. There is a large body of literature which has shown that there is information contained within the temporal variations of subsequent strides that may be revealing about the control of locomotion (e.g., Dingwell & Cusumano, 2000; Hausdorff et al., 1996; Jordan et al., 2005; Stergiou, Moraiti, Giakas, Ristanis & Georgoulis, 2004).

The examination of local dynamic stability through Lyapunov exponents is becoming increasingly common in the gait literature (e.g., Dingwell & Cusumano, 2000; Dingwell, Cusumano, Cavanagh, & Sternad, 2001; Stergiou et al., 2004). Local dynamic stability refers to the ability of a given system to withstand very small perturbations, such as those evident in the stride interval of locomotion. Specifically, the size of the Lyapunov exponent quantifies the rate of divergence of initially nearby trajectories in state space. In a perfectly stable system, there will be little or no divergence of nearby trajectories with time, whereas an unstable system will have a very high rate of divergence. Thus, calculation of finite time Lyapunov exponents (referred to here in as Local Stability Exponents, or LSEs) provides an index of stability that is presumably related to the stability of the coordination pattern but that also considers the time dependent structure of stride to stride fluctuations.

The current experiment investigates the relationship between long range correlations (DFA) and gait cycle stability as indexed by both finite time Lyapunov exponents and SD of relative phase. Experiments 1 and 2 reported in Chapters 3 and 4 have shown that at the preferred walking and running speed the long range correlations for the majority of the measured gait cycle variables are weaker than at speeds slower and faster than preferred. Based on these findings, and the hypothesis that the changing strength of long range correlations with speed is reflective of both the stability and adaptability of the gait cycle, it is expected that the size of alpha will be greatest when individuals are required to *walk* at speeds *faster* than the preferred W-R transition speed, and when they are required to *run* at speeds *slower* than the preferred R-W transition speed. The loss of stability associated with these speeds will be reflected in an increase in the size of the finite time Lyapunov exponents at the PTS, as well as an increase in the variability of intra-limb coordination (relative phase).

5.2 Methods

5.2.1 Subjects

Twelve female volunteers from The Pennsylvania State University, (average age = 26.2 +/- 2.9 years) were recruited for the study. The average height and mass of the subjects was 167.3 ± 3.8 cm, and 62.4 ± 5.6 kg, respectively. All subjects provided informed consent and all procedures were carried out according to the ethical guidelines laid down by the Institutional Review Board of The Pennsylvania State University.

5.2.2 Apparatus

Footfall data were collected using a Kistler Gaitway treadmill with two embedded force plates. Force data were collected at 250Hz from 8 force sensors located in the corners of the two force plates. The treadmill had a speed range of 0.8 to 20 km/hr, with the smallest increment in speed being 0.1km/hr. A Motion Analysis system with 3 visible red cameras (Motion Analysis Corp.) was used to record 3D marker positions. Eight passive reflective markers (see Figure 5.1) were placed on the left side of the participant on the lateral aspects of the acromion process (shoulder), the lateral epicondyle of the humerus (elbow), the styloid process of the ulna (wrist), the ASIS, the greater trochanter (hip), the lateral condyle of the femur (knee), the lateral maleolus of the fibular (ankle) and the head of the 5th metatarsal (toe). A ninth marker was attached to a head band that was secured to the participants head such that the marker was on the top of the head, midway between the ears. In addition to these markers, clusters of marker (four markers per cluster) were attached to the lateral aspects of the upper arm, lower arm, pelvis (lower back), thigh, shank and foot. Motion analysis data were sampled at a rate of 125Hz.

5.2.3 Tasks and Procedures

Participants came into the lab on three separate occasions – the first day was for a 45 min treadmill familiarization/adaptation session where data were not collected (Wall Chateris 1980, 1981). The second and third days involved data collection.

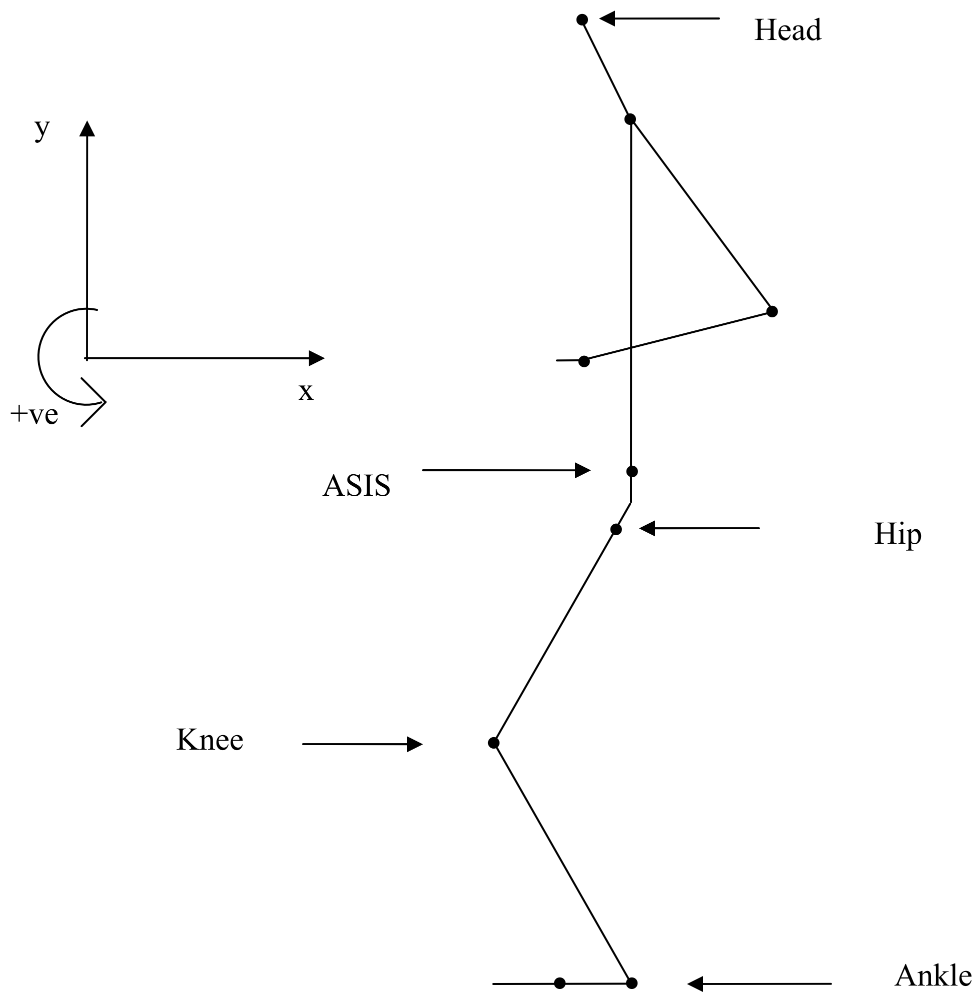

Figure 5.1: Marker locations

At the beginning of each of these days, participants were given 5 - 10min to warm-up on the treadmill, after which the preferred speed for switching gaits was calculated. For the W-R transition, participants began walking at 5.5km/hr (~1.5m/s) and the treadmill speed was increased in 0.1 km/hr increments every 10sec with participants being instructed not to resist the switch to running if it feels more comfortable. This process was repeated three times and the average transition speed taken. Once the transition speed was determined, 7 speeds corresponding to 90, 95, 97.5, 100, 102.5, 105, and 110% of the W-R transition speed were calculated. The R-W transition speed was calculated using the same method, with participants initially running at 9km/hr (2.5m/s) and treadmill speed being decreased by 0.1km/hr every 10s. Speeds corresponding to 90, 95, 97.5, 100, 102.5, 105, and 110% of the R-W transition speed were calculated.

On one of the two experimental days the effects of switching from a walking to a running gait was investigated, with the effects of switching from a running to a walking gait being investigated on the remaining day. On each of the experimental days 14 trials were carried out. The first 7 trials on a given day were "SWITCH" trials. For the walk to run condition, trials were presented in order from slowest to fastest. At the beginning of each trial, participants were given 30 s to decide whether they were more comfortable walking or running at that speed. Once their preferred gait had been established, participants locomoted for 5 min at that speed while data were collected. For the run to walk condition, the same protocol was employed, with the subjects starting at the fastest speed (110% of preferred transition speed) and locomoting at progressively slower speeds with each trial. At the remaining 7 trials of the day were "MAINTAIN" trials. The same speeds were investigated but the participant was instructed to either walk at all speeds (on the W-R transition day) or run at all speeds (on the R-W transition day). Half of the subjects started with the W-R protocol and half started with the R-W protocol.

During the SWITCH trials, all participants except one transitioned at either their predicted transition speed or at a speed 2.5% faster or slower than preferred (i.e. a neighboring speed). For each condition, 5 trials were used in data analysis: the trial occurring at the actual transition speed, the next two fastest and next two slowest trials. One participant transitioned from running to walking at two speeds slower than predicted, therefore in this case there were only 4 trials analyzed for both the run to walk condition and the run all condition for that participant. Averaged data from the remaining participants on that trial was used in place of the missing data. The trial for which the transition occurred is referred to as T, the 2 trials prior to the transition are T-1 and T-2, the 2 trials post transition are called T+1 and T+2, with T±1 being closest to the transition trial and T±2 being furthest from the transition trial. T-2 and T-1 are referred to as "pre-transition" trials, trials T through T+2 are referred to as "post transition" trials.

5.2.4 Data Analysis

Force Plate Data: Custom written MATLAB software was used to compute and process the vertical ground reaction force (VGRF) data. The center of pressure and ground reaction force data were filtered in forward and reverse directions with a 2nd order low pass Butterworth filter. Cutoff frequencies for the walking and running data respectively were 15Hz and 30Hz. There was noise in the force plate date, some of which was introduced by the vibration of the treadmill. This noise was damped to a greater extent during the walking trials because of the relatively greater percentage of time that was spent in stance. As such, thresholds of 20 and 30N were used for the walking and running date respectively - signal below these thresholds was considered to be noise. The stride interval for both walking and running was calculated from the VGRF. Stride interval was defined as starting with the onset of the heel strike of one foot and finishing with the onset of the next heel strike of the same foot.

The mean, standard deviation (SD), coefficient of variation (CV), and the strength of long range correlations were calculated for the time series of each dependent variable for each trial. The SD provides a description of the absolute amount of variability in a given time series whereas the CV provides a description of the amount of variability in the time series normalized by the mean of the time series. These measures of variability refer only to the amount of variability of the time series and do not reflect the structure of stride to stride fluctuations.

Long range correlations on the other hand provide a measure of the structure of variability of the stride interval time series. Stronger correlations would indicate a more predictable, regular time series whereas weaker correlations would indicate a less predictable time series where any given stride interval is less dependent on the stride intervals preceding it. Long range correlations of the time series were calculated using detrended fluctuation analysis (DFA; Peng et al., 1993). This analysis is performed on the integrated time series, which is sectioned into window ranging in length from 4 to N/4 data points (where N is the total number of data points in the time series). The slope (alpha) of the line relating the log of fluctuation size for a given window size to the log of the window size is the value returned by the DFA algorithm. This method avoids the spurious detection of correlations that are artifacts of nonstationarity in the time series (Chen, Ivanov, Hu & Stanley, 2002). An alpha value from the DFA of 0.5 corresponds to a white noise process; alpha greater than 0.5 and less than (or equal to) 1.0 indicates persistent long range correlations, and alpha less than 0.5 indicates persistent long range anti-correlations. The effects for mean, SD, CV, and DFA scaling exponent for each variable were examined using a 2 way (leg by speed) repeated measures ANOVA. Post hoc analysis was carried out using the Tukey post hoc test.

Kinematic Data: The raw joint trajectory data were first post processed using Motion Analysis software (Eva Real Time 4.1.0; Motion Analysis Corp). This process involved filling in any gaps left in a markers time series which were a result of the marker being obscured. This was necessary because during walking the arm swings back and forth past the leg, temporarily obscuring the makers on the ASIS and the greater trochanter. When available, marker clusters were used to fill the gaps using a rigid body assumption. When marker clusters were not available (the cluster that was used to fill the ASIS marker was attached to the posterior aspect of the pelvis and was not always continuously visible to the cameras), a cubic spline curve was used to fill the gaps. Following the post processing, custom written MATLAB software was used to filter the data in forwards and backwards directions using a 2nd order low pass Butterworth filter with a 6Hz cut off frequency. Custom written MATLAB software was then used to calculate the trajectories of the 8 main markers, as well as the joint angle displacement of the hip, knee, and ankle joints. Only movement in the sagittal plane was examined. Hip flexion, knee extension and ankle dorsiflexion are all defined as positive (see Figure 5.1).

To assess the strength of long range correlations at the extreme ends of the body, a peak picking algorithm was used to generate a time series of trajectory peaks in the vertical direction for the head and ankle markers. DFA were performed on these time series. The head peaks twice every gait cycle, whereas the ankle peaks once, and therefore only every second peak of the head trajectories were used. In addition, the mean, SD and CV of each time series of peaks were calculated. Four 3 way (marker by condition by trial) repeated measures ANOVAs were used to examine the influence of these factors on the strength of long range correlations, and on the mean and variability of the time series of peaks. To examine specifically the influence of gait (walking or running), marker (head or ankle) and speed (from slow to fast) on the alpha, mean, SD and CV of the peak time series, 4 additional 3 way ANOVAs were used. Post-hoc analysis was carried out using the Tukey post hoc test.

Local dynamic stability of the ankle, hip, ASIS and head marker trajectories in the vertical direction was examined using finite time Lyapunov exponents (Rosenstein, Collins & DeLuca, 1993). These exponents are referred to herein as Local Stability Exponents, or LSEs, and are denoted where appropriate by lambda*. The first stage of the process of calculating the LSEs requires construction of the appropriate state space for each original time series and its time delayed copies (Takens, 1981):

$$\mathrm{y}(t) = [\Delta x(t), \Delta x(t+\tau), \ldots, \Delta x(t + (\mathrm{d}_E - 1)\tau)], \qquad (5.1)$$

where $\mathrm{y}(t)$ is the d_E – dimensional state space vector, $\Delta x(t)$ is the original time series, τ is the time delay and d_E is the embedding dimension. The time delays for each marker for both running and walking were calculated from the first minimum of the average mutual information function such that adjacent delay coordinates shared a minimal amount of information. A global false nearest neighbors analysis was performed to establish an appropriate embedding dimension.

Lyapunov exponents provide a measure of the average exponential rate of divergence of originally nearby trajectories in state space, thus quantifying the sensitivity of a system to local perturbations (Kantz & Schreiber, 1997; Rosenstein et al., 1993).

Negative Lyapunov exponents indicate local stability, while positive exponents indicate the opposite. The maximum Lyapunov exponent for a system is defined by:

$$d(t) = De^{\lambda t}, \tag{5.2}$$

where D is the initial distance between neighboring points and d(t) is the average separation in state space after time t. lambda is only well defined as $D \to 0$ and $t \to \infty$, however, physiological time series generally do not approach these limits. However, Local Stability Exponents can be reliably estimated for physiological data using the log-transform of both sides of the above equation:

$$\ln[dj(i)] \approx \lambda^*(i\Delta t) + \ln[Dj], \tag{5.3}$$

where dj(i) is the Euclidean distance between the jth pair of nearest neighbors after i time steps. lambda* in turn were estimated from the slope of the linear fit to the line defined by:

$$y(i) = 1/\Delta t <\ln[dj(i)]> \tag{5.4}$$

where < > denotes the average over all values of j (Rosenstein et al., 1993).

Because stride interval will be different for different participants and speeds, the time axis of this curve is rescaled for every trial by multiplying by the stride interval for the appropriate trial and the slope was calculated between 4 and 10 strides (Dingwell & Cusumano, 2000; Dingwell & Marin, 2006).

The mean size of the Local Stability Exponent was calculated for each speed in each condition. As with the peak data, the effect of marker location (ankle, ASIS, head and hip), condition (SWITCH or MAINTAIN) and trial (T-2, T-1, T, T+1 and T+2) on the size of Local Stability Exponent were examined using a 3 way ANOVA. To examine the influence of gait (walking or running), marker (ankle, ASIS, head or hip) and speed (from slow to fast) on the size of the Local Stability Exponent, and additional 3 way ANOVAs was used. As before, the Tukey post hoc test was used for post hoc analysis. For all ANOVAs and post-hoc tests, results are reported as being significant when $p<0.05$.

In order to examine the overall stability of intra-limb coordination, a point estimate of relative phase was calculated for the combinations of ankle hip and ankle knee joint angles. The hip, knee and ankle angles were defined as sagittal projections of the trunk, thigh, shank and foot segments (see Figure 5.1). A peak picking program was used to locate peaks in hip extension, keen flexion and ankle plantar flexion. Peak ankle plantar flexion was compared with peak knee flexion and peak hip extension. The time between successive peaks of hip extension was defined as 360° and time to peak ankle plantar flexion was calculated as a proportion of this cycle in degrees. Similarly, the time between peaks of knee flexion was defined as 360° and time to peak ankle plantar flexion was calculated as a proportion of this cycle in degrees. The mean, standard deviation and coefficient of variation for each 5min trial were calculated and the effect of condition (SWITCH or MAINTAIN) and trial (T-2, T-1, T, T+1 and T+2) on these variables was examined using 2-way ANOVAs for the W-R and R-W data.

5.3 Results

Because of technical problems during data collection, one participant's motion analysis data could not be used. The average speed for the W-R transition was 7.1 +/- 0.3km/hr, the average speed for the R-W transition was 6.9 +/- 0.5km/hr.

5.2.1 DFA of Stride Interval – Peak Time Series

W-R Condition: The 3 way (marker by condition by trial) ANOVA performed yielded no significant main effects. There were significant 2-way interaction effects for marker vs. condition, $F(1,10) = 19.45$, and for marker vs. trial, $F(4,40) = 2.82$. Figure 5.2 illustrates the 3-way interaction which was also significant ($F(4,40) = 5.90$) and revealed that for the head marker, T-2 and T-1 had significantly smaller alpha values than T, T+1 and T+2 in the SWTICH condition. As can be seen in Figure 5.2B, while the size of alpha of the head marker increased with speed for both the SWITCH and MAINTAIN trials, effectively the only increase in the SWITCH trials occurred between T-1 and T, where as the increase in the size of alpha for the MAINTAIN trials was more incremental. Post-hoc tests revealed that for the head marker, in the SWITCH condition there were significant differences between both of the pre-transition and all of the post transition trials, but not within the pre-transition or post transition trials. Conversely, for the MAINTAIN trials there were significant increases from T-2 to T through T+2 and from T-1 to T4 only. For the ankle marker, there were no significant differences between any of the trials in the SWTICH condition whereas there was a steady increase from T-2 to T+2 in the MAINTAIN condition, with T-2 having a significantly smaller alpha value than T through T+2 and T -1 having a significantly smaller alpha than T+1 and T+2. As is evident in Figure 5.2A, on the whole, the size of alpha was smaller in the SWITCH trials than in the MAINTAIN trials, with the exception of alpha at T-2 and T-1.

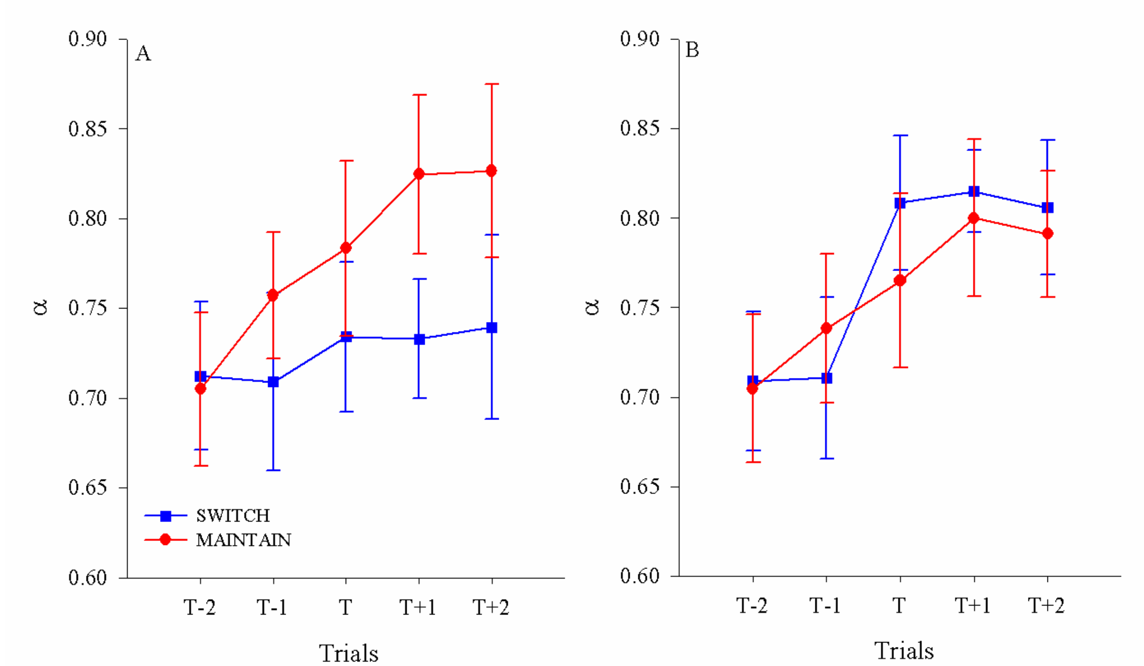

Figure 5.2: Group alpha values for (A) the peak time series of the ankle; and (B) of the head. Comparison of SWITCH and MAINTAIN trials for the W-R condition

R-W Condition: It is clear from Figure 5.3 that the results for the run-walk condition were less clear than that of the walk to run condition. There were no main effects for marker, condition or trial, (F(1,10) = 2.63, p = 0.14; F(1,10) = 0.91, p = 0.36; F(4,40) = 0.56, p = 0.69 respectively). There was a significant interaction effect, F(1,10) = 7.79, for marker vs. condition. Post-hoc tests revealed that there was no difference in the size of alpha for the ankle marker between the 2 conditions (alpha = 0.74 for both SWITCH and MAINTAIN conditions) whereas there was a significant increase in the size of alpha values for the head marker from the SWTICH condition (alpha = 0.76) to the MAINTAIN condition (alpha = 0.80). The three way interaction illustrated in Figure 5.3 was not significant (F(4,40) = 2.36, p = 0.07). Figure 5.3 suggests, however, that alpha was smaller post-transition than pre-transition in the SWITCH trials, and tended to increase in the MAINTAIN trials as speed decreased.

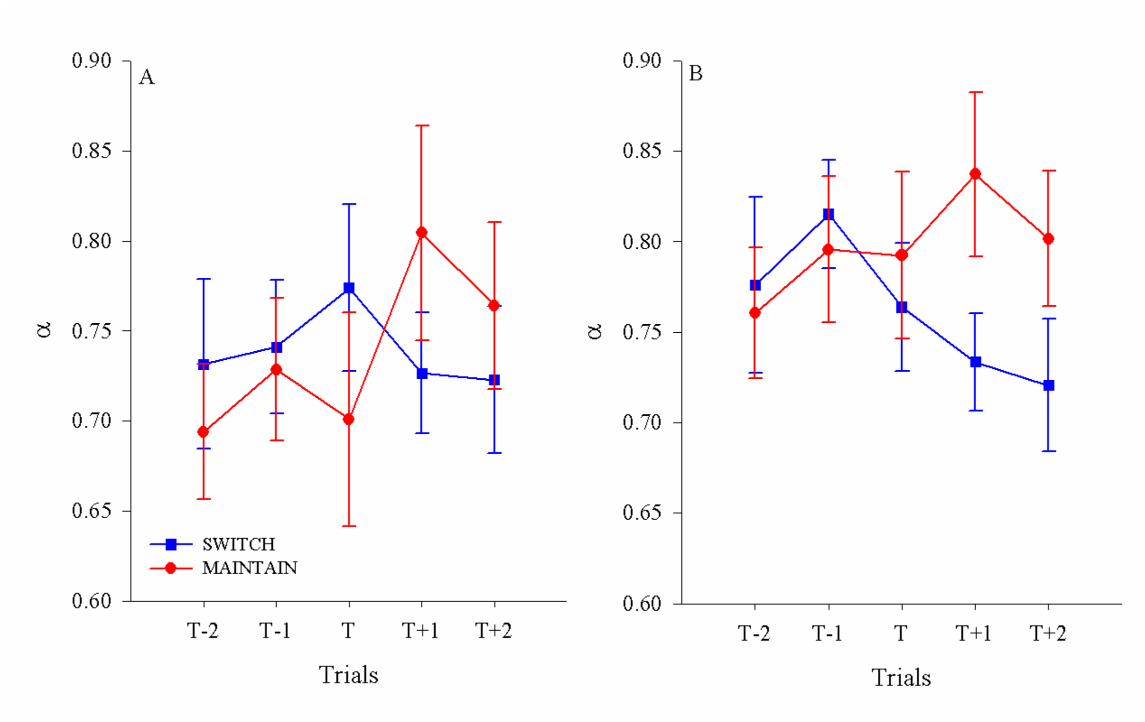

Figure 5.3: Group alpha values for the peak time series of (A) the ankle and (B) the head. Comparison of SWITCH and MAINTAIN trials for the R-W condition

Walking vs. Running Gait: There were no main effects for marker, gait or speed for any of the MAINTAIN walk or run trials. There was a significant marker by gait interaction, $F(1,10) = 7.81$, such that the alpha of the ankle marker was larger in the walking gait than the running gait, whereas the alpha of the head marker was larger in the running gait than the walking gait. Figure 5.4 shows a significant 2-way interaction between speed and gait, $F(4,40) = 2.87$, with the alpha for the walking gait tending to increase and alpha for the running gait tending to decrease with speed.

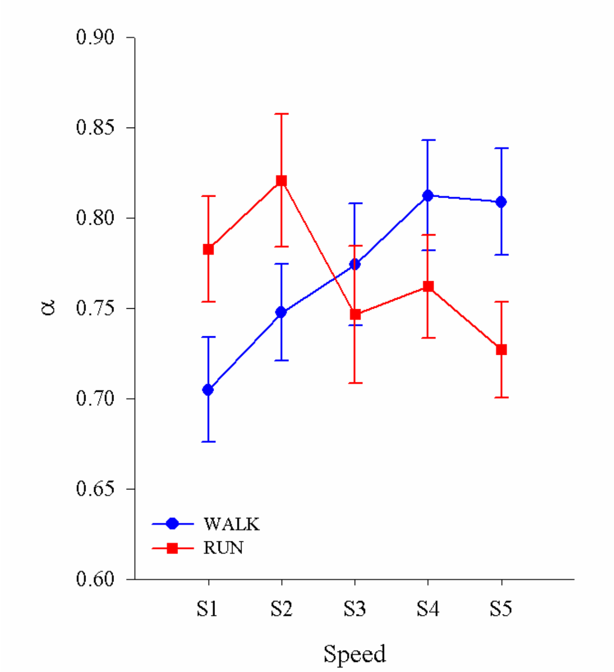

Figure 5.4: Group alpha values averaged for the head and ankle peak-to-peak time series - comparison of the walking to the running gait with increasing speed

5.3.2 DFA of Stride Interval – Treadmill Data

In general, the results from the treadmill data correspond closely to that of the peak data. There were no significant main or interaction effects in either the W-R condition or the R-W condition for the detrended fluctuation analysis of the stride interval calculated from the treadmill data. When the walk MAINTAIN and run MAINTAIN trials were compared across leg and speed, there was a significant gait by speed interaction, $F(4,44) = 2.88$, with alpha of the walking stride interval tending to increase with speed whereas alpha of the running stride interval tended to decrease with speed. There was also a significant three way interaction, $F(4,44) = 3.01$. Post hoc tests revealed that there was a significant increase in the size of alpha with increasing walking speed in both the right and left legs. There was also a significant decrease in the size of alpha with increasing walking speeds. The slowest walking speed had a significantly smaller alpha than the slowest running speed, there was no difference in the size of alpha for the two gaits at speed 2 (the 2nd slowest speed), and for the 3 remaining speeds alpha was significantly greater in the walking than running condition.

5.3.3 Mean, SD and CV of Stride Interval - Peak Time Series

W-R Condition: There was no difference in average time between peaks (i.e. stride interval) at the ankle or hip marker, $F(1,10) = 3.56$, $p = 0.09$. There was a significant condition effect, $F(1,10) = 56.76$ in that the time between peaks during the SWITCH condition was on average smaller than that of the MAINTAIN condition. A significant speed effect revealed that the overall effect of increasing speed was a reduction in the average time between peaks. A significant condition by speed interaction is illustrated in Figure 5.5A, $F(4,40) = 63.72$, with post-hoc tests showing that the speed effect was driven by the difference between T-T+2 for the SWITCH and MAINTAIN conditions. The average time between peaks for T through T+2 for the MAINTAIN condition was significantly longer than for the corresponding SWTICH trials. The SD of the time between peaks was significantly reduced in the SWITCH condition as compared with the MAINTAIN condition, $F(1,10) = 6.82$. Aside from this, there were no other significant main or interaction effects for either SD or CV, although there was a tendency for the ankle to be less variable than the head, both in terms of absolute ($F(1,10) = 4.63$, $p = 0.06$) and normalized variability ($F(1,10) = 4.08$, $p = 0.07$).

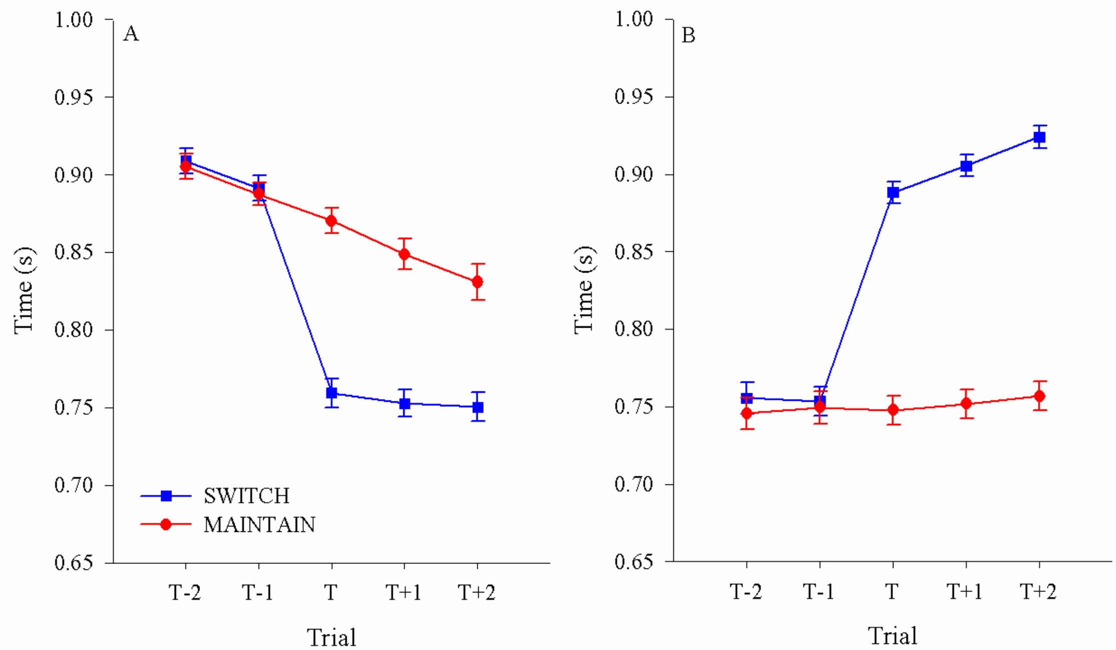

Figure 5.5: Group mean stride interval calculated from the peak time series in (A) the W-R condition and (B) the R-W condition. Comparison of SWITCH and MAINTAIN trials.

R-W Condition: Figure 5.5B shows the overall pattern of results for mean stride interval with decreasing speed. As with the W-R condition, there was no significant marker effect for the average time between peaks, F(1,10) = 0.14, p = 0.71. There were significant main effects for both condition (F,(1,10) = 134.71), and speed (F(4,40) = 92.28) such that the average time between peaks was significantly longer in the SWITCH vs. the MAINTAIN condition (0.85 vs. 0.75 s respectively) and that the average time between peaks increased with decreasing speed (from 0.75 to 0.84 s). There was also a significant condition by speed interaction (F(4,40) = 129.61). Figure 5.5B suggests and post hoc tests confirm that there was no difference in the average time between peaks for the two fastest speeds in either the SWITCH or MAINTAIN conditions but that the time between peaks was significantly longer in the SWITCH condition as compared with the MAINTAIN condition for the post-transition trials. There was no significant change in the average time between peaks with changing speed for any of the MAINTAIN trials whereas the three post-transition trials had significantly more time between peaks than the pre-transition trials in the SWITCH condition. Evident in Figure 5.6 is a significant condition by speed interaction effect for CV of time between peaks, $F(4,40) = 2.88$, such that relative variability tended to decrease with decreasing speed in the SWITCH condition, but increase with decreasing speed in the MAINTAIN condition. There were no other main or interaction effects for either SD or CV of time between peaks.

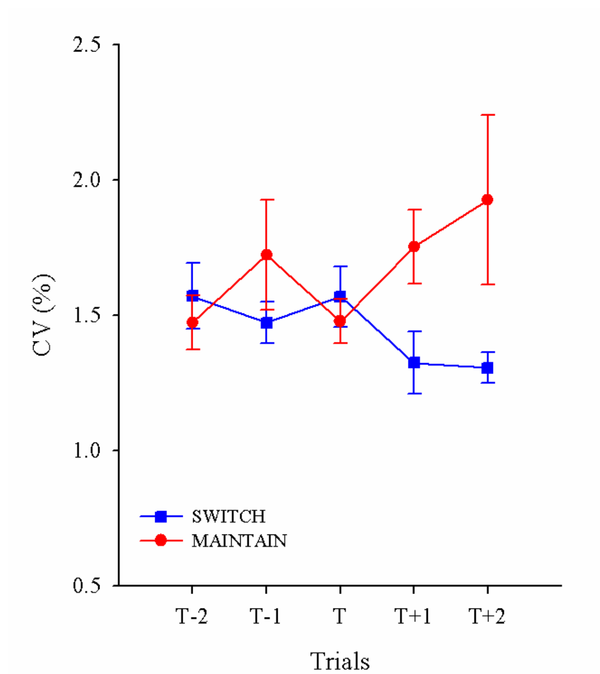

Figure 5.6: Group relative variability of stride interval calculated from the peak time series – comparison of SWITCH and MAINTAIN trials for the R-W condition

Walking vs. Running Gait: As with both the W-R and R-W trials, there was no main effect for average time between peaks for the ankle and head markers, $F(1,10) =$

4.35, p = 0.06. Although the p-value indicates there was nearly a significant difference, the average time between peaks for the ankle and head markers was 0.8092 and 0.8096 s respectively, a difference 20 fold smaller than the sampling rate used for data collection. There was a significant speed effect, $F,(4,40) = 45.95$, with the average time between peaks decreasing with increasing speed, and a significant gait effect, $F(1,10) = 102.03$, such that the average time between peaks was larger in the walking gait (0.87 s) than in the running gait (0.75 s). Figure 5.7 shows that the speed effect was driven by changes in the walking gait. Post-hoc testing of gait by speed interaction, $F(4,40) = 26.06$ confirms this observation. There were no significant effects for either SD or CV.

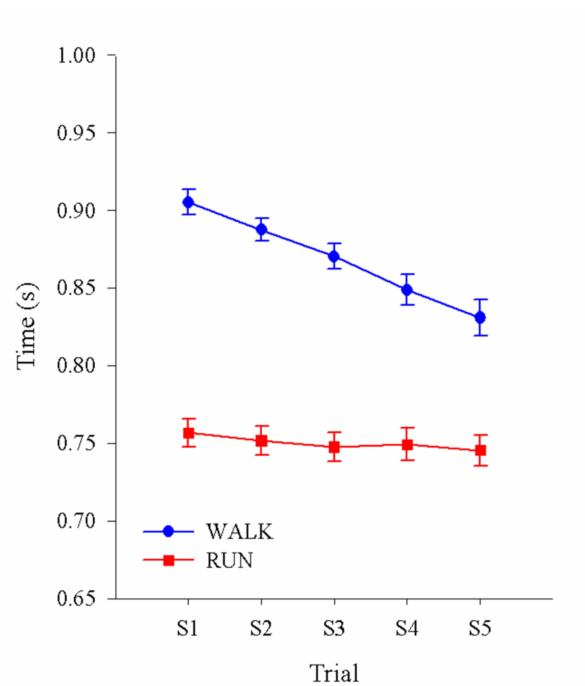

Figure 5.7: Group mean stride interval calculated from the peak time series - comparison of the walking to the running gait with increasing speed

5.3.4 Mean, SD and CV of Stride Interval –Treadmill Data

W-R Condition: The results of the 3-way ANOVA indicated that there was a significant difference between the average stride interval of the left and right legs, $F(4,44) = 7.51$. However, the average time difference between the left and right legs was less that 1 thousandth of a second and therefore smaller than the sampling rate used for data collection. There was a significant main effect for condition, $F(1,11) = 69.25$, with the SWITCH trials having significantly smaller average stride intervals than the MAINTAIN trials (0.81 vs. 0.87s). A significant condition by speed interaction, $F(4,44) = 69.78$, revealed that there was a sharp decrease in stride interval after the transition to running occurred in the SWITCH trials. Post-hoc tests showed that there were significant differences between all 3 post-transition trials in the SWITCH and MAINTAIN conditions. In terms of variability, there was a significant interaction effect between condition and speed for SD, with the MAINTAIN trials increasing in variability with speed, particularly for trials T through T+2, and the SWITCH trials tending to decrease in variability with speed.

R-W Condition: There were main effects for both condition $F(1,11) = 162.12$, and speed $F(4,44) = 144.74$, on the average stride interval, with the SWITCH condition having significantly smaller average stride interval than that of the MAINTAIN condition, and stride interval increasing with decreasing speed. As with the walk to run condition, there was a significant 2-way interaction between condition and speed, with the average stride interval increasing significantly in the SWITCH condition for the post transition trials. There were no difference between the SWITCH and MAITAIN conditions for T-2 and T-1. In terms of variability, there was a significant 2 way interaction between speed and condition for CV, with variability being significantly greater in the MAINTAIN condition than the SWITCH condition at trials T+1 and T+2. Additionally, in the SWITCH condition, T+1 and T+2 were significantly less variable than T-2 – T.

Walking vs. Running Gait: There were significant main effects for gait, $F(1,11) = 121.07$, and speed, $F(4,44) = 48.17$, but not for leg, $F(1,11) = 1.27$, $p = 0.28$. There was a significant interaction between gait and speed which showed that although the main effect for speed was a decrease in mean stride interval with increasing speed, this effect was only significant for the walking gait. Post hoc tests showed that there were significant increases in stride interval at every increment in speed for the walking gait, that the walking gait had for all speeds a significantly larger stride interval, and that the size of stride interval did not change across speed for the running gait. Figure 5.8 shows the pattern of change of SD and CV with speed for the walking and running gaits. A gait by speed interaction for SD revealed that the variability of the stride interval increased with speed in the walking gait but not the running gait. There was a significant difference in the amount of variability at the fastest walking speed compared to the three fastest running speeds. There was also a significant interaction between gait and speed for CV, again with the walking gait tending to increase in variability with increasing walking speed, whereas the running gait was not affected by speed. Post hoc tests revealed that there was a significant increase in CV from the slowest to the fastest walking speed.

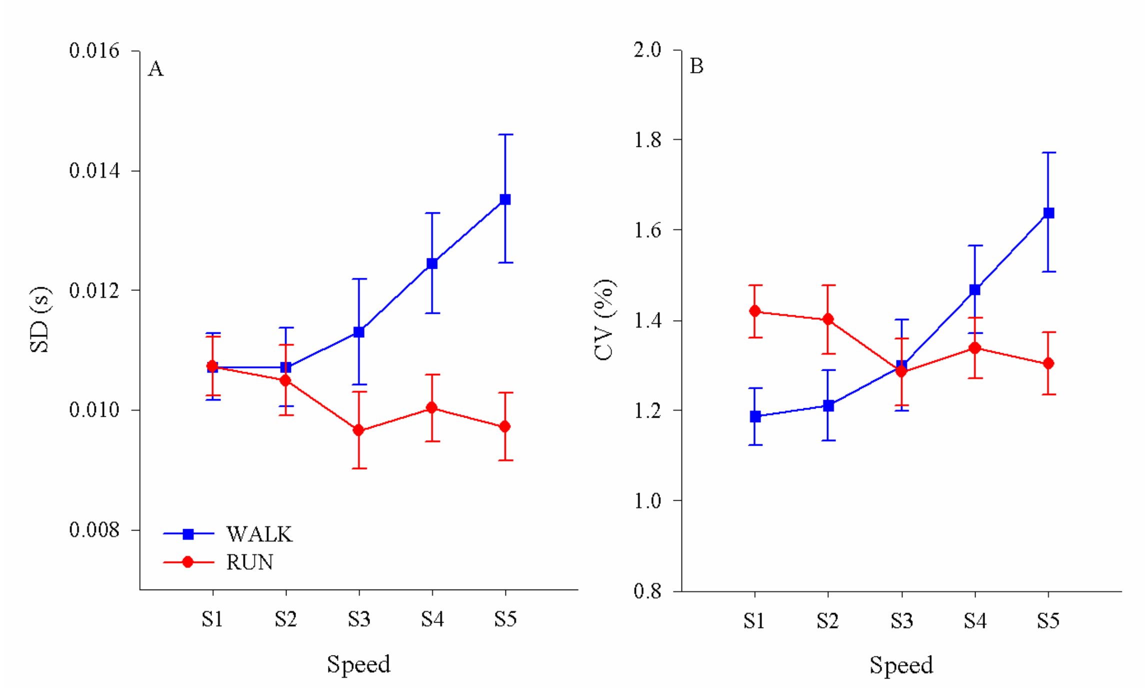

Figure 5.8: Group (A) absolute variability and (B)relative variability of the stride interval calculated from treadmill data – comparison of walking and running gaits.

5.3.5 Local Stability Exponents

The time delays for each marker in both directions for walking and running are given in Table 5.1. As with previous gait literature (Dingwell et al., 2001, Dingwell et al., 2006), an embedding dimension of 5 was found to be satisfactory.

Table 5.1 Time delays calculated for state space reconstruction from the first minimum in average mutual information.

	Walk	Run
Ankle	18	24
ASIS	15	15
Head	18	15
Hip	15	15

W-R Condition: There was no significant main effect for either condition or speed on the strength of the LSE. There was a significant main effect for marker location $F(3,30) = 39.90$, with the ankle marker having significantly larger, and the head having significantly smaller, LSEs than all other markers. A significant interaction effect between condition and marker ($F(3,30) = 16.18$) revealed a tendency for the strength of LSEs to be greater in the SWITCH trials than in the MAINTAIN trials for the head marker, with the reverse being true for the ankle marker. There was no difference in strength of LSEs between the SWITCH and MAINTAIN trials for the ASIS or hip markers. There was a significant three way interaction, $F(12,120) = 6.5$, which is illustrated in Figure 5.9. Post hoc tests showed that for the ankle marker the size of the LSEs was smaller in SWITCH than the MAINTAIN condition at T-1, T+1 and T+2. For the ASIS marker, post hoc tests showed an inconsistent pattern of results, with the LSE of the ASIS at SWITCH T-1 being significantly smaller than that of both MAINTAIN T-1 and SWITCH T. For the head marker, the two pre-transition trials in the SWITCH condition had significantly smaller LSEs than the post-transition trials in the same condition. The post-transition trials in the SWITCH condition had significantly larger LSEs than the equivalent trials in the MAINTAIN condition. Lastly, for the hip marker the post-transition trials tended to have larger LSEs than the pre-transition trials in the SWITCH condition, with the differences reaching significance for T-2 and T as well as for T-2 and all post-transition trials. As with the head marker, the LSEs for the SWITCH condition were larger than those of the MAINTAIN condition, although the difference failed to reach significance for T+1.

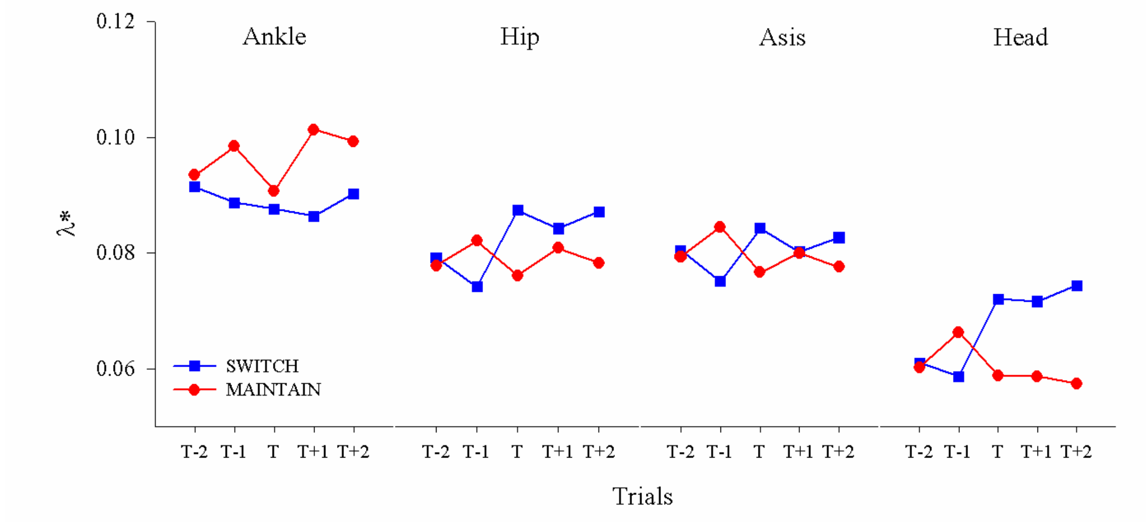

Figure 5.9: Group local stability exponents for the ankle, ASIS, head, and hip. Comparison of SWITCH and MAINTAIN trials for the W-R condition

R-W Condition: Figure 5.10 illustrates the pattern of findings for the R-W condition. As was the case for the W-R condition, there was a significant marker effect, $F(3,30) = 27.91$, with post-hoc tests revealing that the head marker had significantly smaller LSEs than all other markers, with the ankle marker having larger LSEs than the ASIS. There was no difference in the size of LSEs for the hip and ASIS markers. The condition by marker, condition by speed and marker by speed interactions all reached significance ($F(3,30) = 2.94$; $F(4,40) = 2.83$; and $F(12,120) = 4.68$ respectively), however, the three way interaction failed to reach significance ($F,(12,120) = 1.11$, $p = 0.36$). Post-hoc analysis showed that the in the case of the marker by condition interaction, differences in sizes of the LSEs between markers was reduced in the MAINTAIN condition compared to the SWITCH condition. Post-hoc tests for the marker by speed interaction showed that the ankle was not affected by speed but that the head ASIS and hip tended to have smaller LSEs for the post-transition trials than for the pre-transition trials. For the head, there was a significant decrease between T-2 and T as well as between T-1 and T through T5. For the ASIS, there was a significant decrease from T-1 to T and for the hip from T-1 to both T and T+1. Lastly, for the condition by speed interaction, there was an overall trend for the SWITCH condition to decrease and the MAINTAIN condition to increase with decreasing speed. This difference was most apparent at T+2.

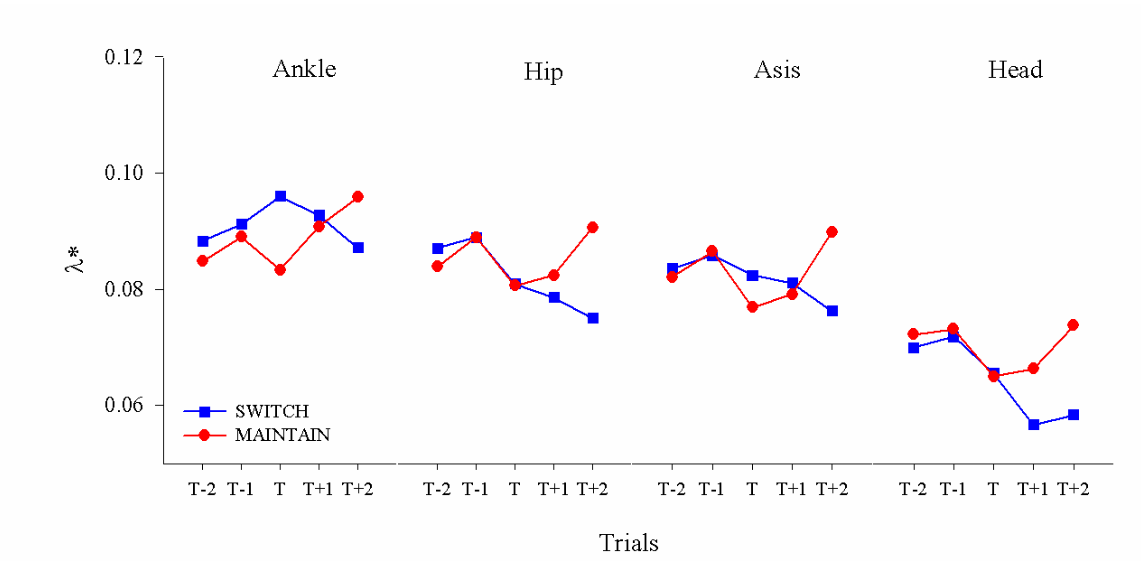

Figure 5.10: Group local stability exponents for the ankle, ASIS, head, and hip. Comparison of SWITCH and MAINTAIN trials for the R-W condition

Walking vs. Running Gait: There was a main effect was for marker, F(1,10) = 31.90, – the same pattern as previously described occurred with the head having the smallest LSEs, the ankle the largest, with no difference occurring between the hip and ASIS markers. Figure 5.11 shows that there is no difference between the walking and running gaits in terms of the size of LSEs when the ankle, ASIS, head and hip are considered together. However, there was a significant interaction between marker and gait, F(3,30) = 10.99, such that the size of the LSEs of the ankle were smaller in the running gait than in the walking gait, the reverse being true for the head marker. There was no effect of gait on either the hip or ASIS markers.

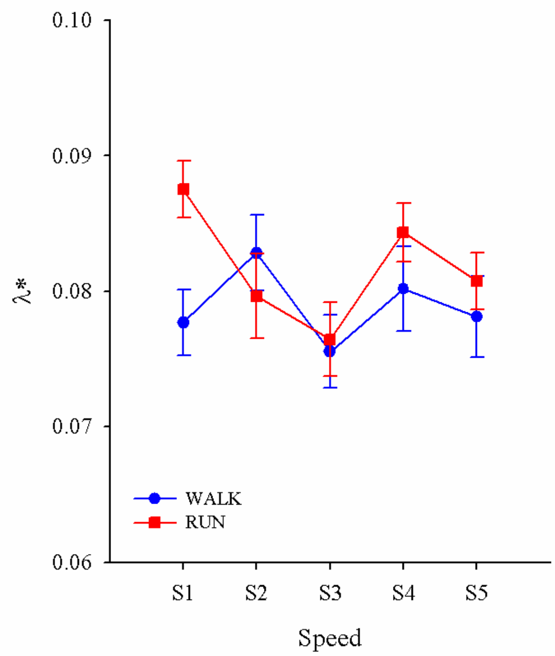

Figure 5.11: Group local stability exponents - comparison of the walking to the running gait with increasing speed

A significant three way interaction, F(12, 120) = 3.17, revealed that for the head and hip markers, the running condition generally yielded larger LSEs than the walking condition, whereas the ankle marker generally had smaller LSES in the running condition. Post hoc tests revealed significant difference between the walking and running gaits for the head marker at the slowest speed and at the two fastest speeds. For the hip marker, the difference between walking and running only reached significance at the slowest speed. For the ankle marker, the decrease in size of LSE from walking to running was significant for the two fastest speeds, and there was also a significant decrease in LSE size in the running gait from the slowest speed to both the 3rd and fastest speed. In the case of the ASIS marker, for the slowest speed the size of LSEs were greater in the running condition than in the walking condition.

5.3.6 Relative Phase of the Hip and Ankle Joints

Figure 5.12A illustrates a clear decease in mean relative phase of the hip and ankle joints associated with changing from a walking to a running gait which reached significance. Figure 5.12B shows the opposite effect when changing from a running to walking gait. For both the W-R and R-W conditions there were significant interaction effects ($F(4,40) = 8.31$ and $F(4,40) = 6.32$ respectively). Subsequent post hoc testing confirmed in the W-R condition there was no difference in relative phase between the hip and ankle at T-2 and T-1 but that there was a significant decrease in relative phase from pre to post transition trials in the SWITCH condition. In addition, the relative phase of trials T through T+2 was greater in the MAINTAIN condition than in the SWITCH condition. The opposite pattern of significant results was evident for the R-W condition, such that relative phase increased in the post-transition trials in the SWITCH condition. There were however no main or interaction effects for the SD of relative phase, for either the W-R or R-W condition, as is evident in Figure 5.13.

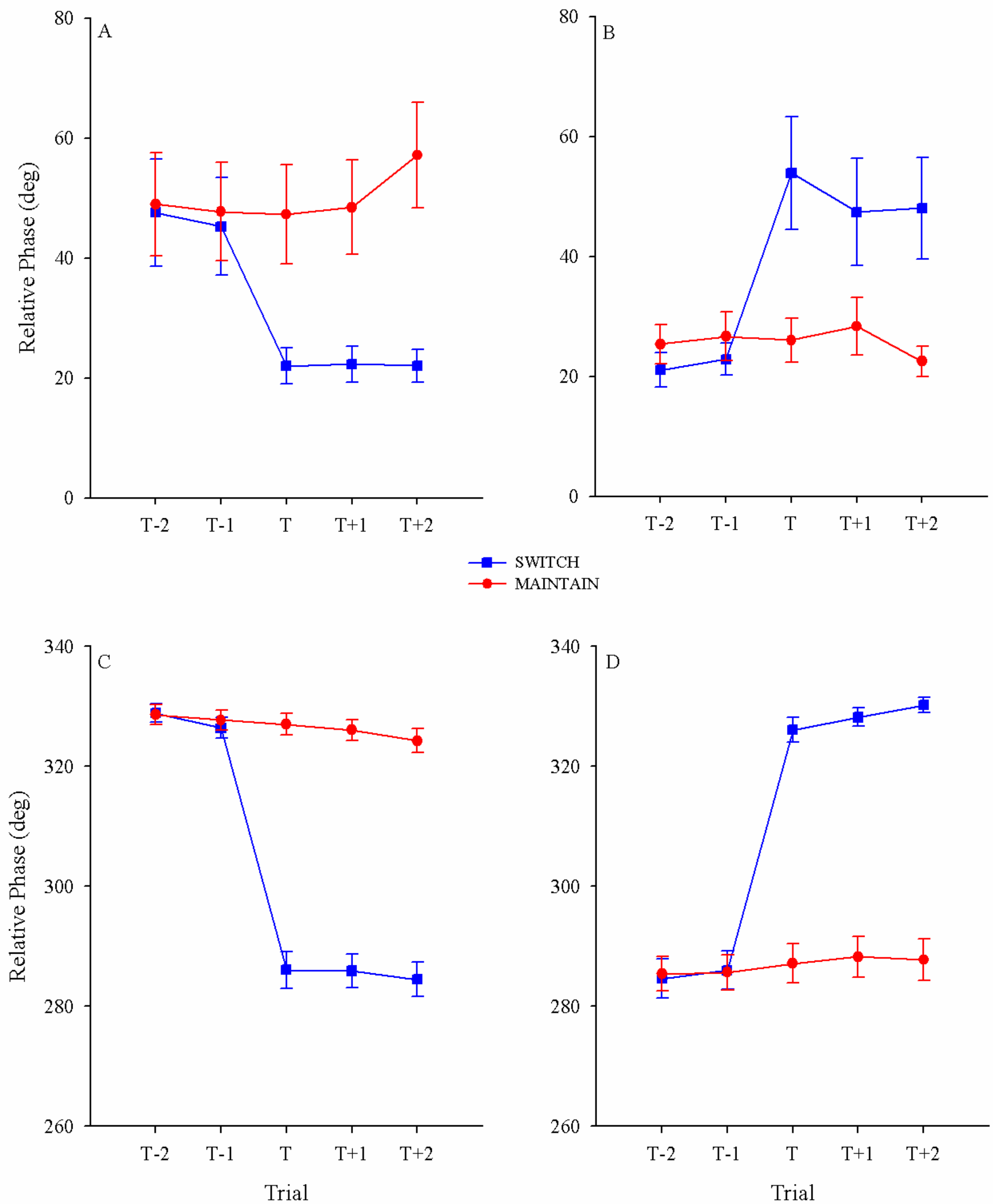

Figure 5.12: Group mean relative phase for the hip and ankle during (A) the W-R condition and (B) during the R-W condition; and for the knee and ankle during (C) the W-R condition and (D) the R-W condition. Comparison of SWTICH and MAINTAIN.

5.3.7 Relative Phase of the Knee and Ankle Joints

The interactions between condition and speed for both the W-R and R-W condition seen in Figure 5.12B and D, reached significance ($F(4,40) = 158.56$ and $F(4,40) = 141.18$ respectively). Post-hoc testing revealed that there was a significant decrease in knee ankle relative phase after the transition to running occurred in the SWITCH trials in the W-R condition, with a significant increase in average relative phase occurring after the transition to walking in the R-W condition. There was no effect for speed on the MAINTAIN trials in either the W-R or the R-W condition. Figure 5.13 B and D show the variability of knee-ankle relative phase for the W-R and R-W conditions respectively. It can be seen that the variability of knee ankle relative phase increased after the transition to running occurred in the W-R condition, an effect that reached significance $F(4,40) = 22.88$. Conversely, there was no change in the variability of the MAINTAIN trials with decreasing speed. For the R-W condition, the variability of knee ankle relative phase decreased significantly after the transition to running in the SWITCH trials, $F(4,40) = 4.85$. Again, decreasing speed did not influence the variability of the knee ankle relative phase.

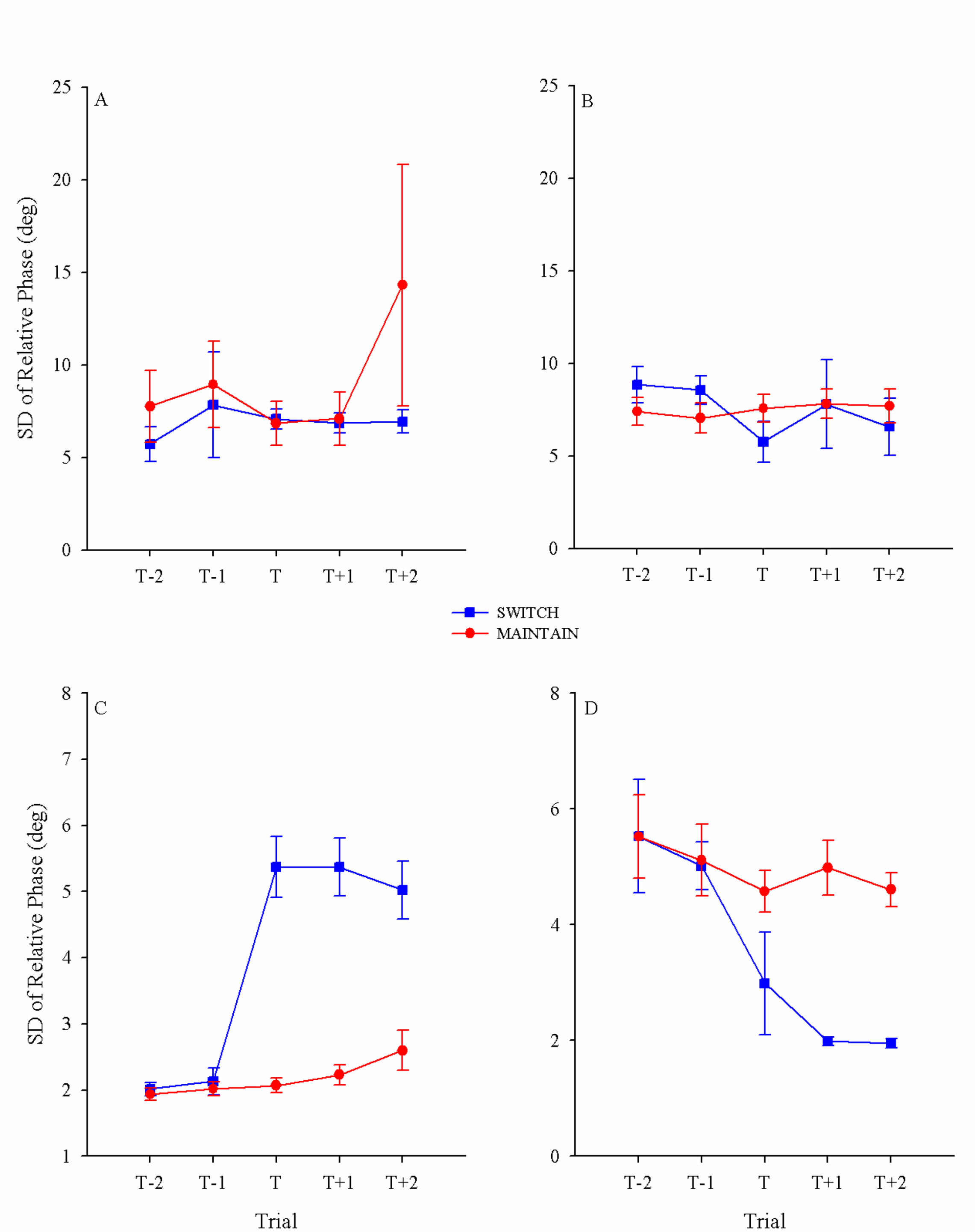

Figure 5.13: Group SD of relative phase for the hip and ankle during (A) the W-R condition and (B) during the R-W condition; and for the knee and ankle during (C) the W-R condition and (D) the R-W condition. Comparison of SWTICH and MAINTAIN.

5.4 Discussion

The results of this study show that slow running is associated with a greater degree of instability than fast walking, both in terms of the local dynamic stability of the head and the coordination of the knee and ankle. This result is mirrored in the detrended fluctuation analysis of the peak to peak interval at the head - the size of alpha at the head is greater in running than walking. These findings are consistent with the proposition that there is a relationship between the stability of locomotion and long range correlations in the stride interval of human walking.

Increased SD of relative phase in the running gait relative to walking is consistent with the findings of Seay et al. (2006) and Diedrich and Warren (1995). The destabilization of knee-ankle coordination observed in the current study, however, is not consistent with the results of Diedrich and Warren (1995). This may be due to differences in the method for determining relative phase. Diedrich and Warren examined the peak ankle plantar flexion as a fraction of the time between successive peaks in knee extension whereas in this study peak knee flexion was used as a reference point. Another non-trivial difference is that Diedrich and Warren calculated hip angle with respect to vertical, rather than with respect to the pelvis as was done here. Regardless of methodological differences, the evidence from the current study as well as that of Seay et al. (2006), and Diedrich and Warren (1995) suggests that slow running is less stable in terms of relative coordination than fast walking.

Decreases in head stability during running seen in this study may be attributable to the relatively larger peak in ground reaction force seen for running compared with walking at these speeds (Keller et al., 1996) as well as poorer shock attenuation at shorter stride lengths (Shorten & Winslow, 1992; Mercer, Vance, Hreljac & Hamill, 2002). Despite this, vertical accelerations of the head remain within a narrow range across running speeds (Hamill, Derrick & Holt, 1995) indicating that head stability is an important criterion during running. Also of note is that although head stability is decreased during slow running, the head is still significantly more stable than the ankle. It appears that the effort required to maintain head stability at slow running speeds is reflected in the relatively larger size of alpha during running at these speeds.

In contrast to the head, the LSEs were greatest at the ankle and had a greater degree of local dynamic stability in the running gait than in the walking gait, with ankle stability increasing with running speed. The same pattern of results is evident in the DFA of the peak to peak time series of the ankle, further supporting the link between stability and long range correlations. There was no difference in local dynamic stability or the strength of long range correlations during the SWITCH trials. However, slow running was associated with greater ankle stability than fast walking. Walking at speeds close to and faster than the preferred W-R transition speed causes a decrease in the force producing capabilities of the plantar-flexors, despite increasing muscle activation (Neptune & Sasaki, 2005). Similarly, it has been shown that the mechanical limits of the dorsiflexors are approached under these conditions (Hreljac, 1995). The relatively larger alpha of the ankle during walking, particularly in comparison to running at post W-R transition speeds is consistent with the idea that the degrees of freedom of the muscle groups crossing the ankle joint are limited.

The same pattern of results was apparent in the stride interval calculated from the treadmill data, with alpha increasing significantly with speed during walking trials and decreasing significantly with increasing speed during running trials. Thus, as walking speed moves further from preferred, alpha increases, as running speed moves closer to preferred, alpha decreases. This trend was apparent for both the hip and ankle locations and is consistent with the hypothesis that walking and running at preferred speeds results in a reduction in strength of long range correlations. This result is also consistent with those of Experiments 1 and 2. In a statistical sense, the smaller the value of alpha, the less any given stride interval is related to or dependent upon previous stride intervals. As such the size of alpha may reflect the overall adaptability of the system at a given speed.

Increasing walking speed is achieved through concurrent increases in stride length and stride frequency. As walking speed increases, locomotion becomes more constrained – the stride length for example "saturates" at around 2m/s due to the biomechanical limits of the system (e.g. Hirasaki, Moore, Raphan & Cohen, 1999). In order to increase walking speed above 2m/s stride frequency alone can be increased, thus there is a reduction in the number of independently controllable elements at this speed. In addition to this example, there are numerous other constraints that result from both fast walking and slow running that are not encountered during normal speeds of locomotion. The proposition that there is a reduction in dynamical degrees of freedom at these speeds is consistent with the more regular output seen at these speeds.

The amount of variability (SD) in the stride interval as estimated using ground reaction force data was increased in the W-R condition when participants were forced to maintain a walking gait at speeds they would typically run at. This is consistent with the results of both Brisswalter and Mottet (1996) who showed that walking at speeds close to the W-R transition speed resulted in an increased stride interval variability and those of

Yamasaki, Sasaki, Tsuzuki and Torii (1984) who showed a U-shaped variability function for step duration with speed. While this trend was present when the stride interval was estimated from time between peaks, it failed to reach significance, although there was a significant overall trend for absolute variability to be greater in the MAINTAIN than in the SWTICH trials. Similarly, during the R-W trials relative variability was significantly higher when participants ran at speeds they would normal walk at. Cumulatively, these findings suggest that walking at preferred running speeds and vice versa results in larger amounts of variability (at least in the stride interval). From this it may be inferred that one of the effects of gait transitions maybe a reduction in stride interval variability.

In summary, walking and running are complex postural tasks which require continual control of postural stability and coordination of a large number of degrees of freedom (Bernstein, 1967; Newell & McDonald, 1994; Winter, 1983). This study shows that there is a loss of gait cycle stability as indexed by both SD of relative phase and LSEs associated with speeds at and around preferred transition speeds that is reflected in the relatively larger size of alpha. This instability is particularly evident when individuals run at slow speeds as compared with walking at fast speeds. The pattern of change with speed of the local dynamic stability of the head parallels the speed related patterns of change of both the stability of knee-ankle coordination and alpha of the peak to peak interval of the head. This supports the hypothesis that there is a relationship between stability and long range correlations in the stride interval of human locomotion. The increasing size of alpha in walking and decreasing size in running with speed in this study is consistent with the general proposition that preferred speeds of locomotion have reduced strength of long range correlations and that this is indicative of both increased adaptability and gait cycle stability at these speeds.

Chapter 6

General Discussion

The focus of this dissertation was to examine the long range correlations in the stride interval of human walking and running. This type of fractal structure is common in nature yet it is not easily explained (Bak, 1996; Mandelbrot, 1982; Wagenmakers, Farrell, & Ratcliff, 2004). Despite the fact that there has been a long history of treating variability in human movement as a random noise superimposed on an otherwise regular signal, time dependent structure appears to be the norm rather than the exception to the rule in movement time series (e.g., Chen, Ding & Kelso, 1997; Duarte & Zatsiorsky, 2000, 2001; Newell & Slifkin, 1998; Riley & Turvey, 2002; Yoshinaga, Miyazima & Mitake, 2000) including those of locomotion (e.g., Hausdorff et al., 1995; Terrier, Turner & Schutz, 2005; West & Griffin, 1998, 1999; West & Scarfetta, 2003).

Thus, while is not surprising that there is structure in the variability of the stride interval of human walking given the prevalence of 1/f processes in biological systems, that the correlations last for thousands of strides (Hausdorff et al., 1996) is remarkable. It has been proposed that these correlations emerge as a result of the multiple interacting processes occurring on multiple time scales that are necessary for the functional and adaptive control of locomotion (Hausdorff et al., 2000). That there are systematic changes in the strength of long range correlations with aging and disease (Hausdorff et al., 1997; Hausdorff et al., 1999), and, walking in time to a metronome disrupts the correlations (Hausdorff et al., 1996; Terrier et al., 2005), is consistent with this theoretical claim. Therefore, it appears that the fractal nature of this structure in the variability may relate to the control processes that organize locomotion.

Experiments 1 and 2 investigated the influence of speed on the amount and structure of variability in the walking and running gaits, respectively. Experiment 3 investigated the influence of walking and running at speeds at and around the preferred walk to run and run to walk transition speeds. The effects of speed and gait manipulations on the amount and structure of variability as well as several measures of stability were examined. The major empirical findings as they relate to the primary questions of this dissertation are discussed below.

6.1 The Ubiquity of Long Range Correlations in Human Locomotion

The first question this dissertation examined was the degree to which long range correlations are present in locomotion – i.e. are long range correlations present in a range of kinematic and kinetic gait variables? The results of all three experiments show that the fluctuations in a variety of kinematic and kinetic gait cycle variables exhibit long range dependence that decays in a fractal-like power-law fashion with time. Across the three experiments, 16 different time series of kinematic and kinetic variables were examined from the ground reaction force data. Two additional sets of time series were generated from the peaks in vertical head and ankle trajectories. Long range correlations were consistently present in all of these time series for all participants, in both the running and walking gaits. Thus, long range correlations are intrinsic to the locomotion of healthy individuals and are present in multiple different properties of locomotor output.

Previous studies examining fluctuations of the gait cycle have used several different methods for calculating the time series of gait cycle variables of interest. Haudsdorff's group has consistently examined the interval between successive heel strikes of the same limb (e.g., Hausdorff et al., 1995; Hausdorff et al., 1996; Hausdorff et al., 1997). Terrier et al. (2005) examined step frequency and length calculated from head displacement recorded via a GPS system. West and Griffin (1998, 1999) used peak knee extension to calculate the stride interval time series. In all cases long range correlations were found to be present in the time series, however, until now there has been limited comparison or discussion of differences (or similarities) of the strength of long range correlations across these different levels. The results of Experiment 3 are particularly interesting as they demonstrate that while the strength of long range correlations are similar when examined using heel strike data and peak data of the head and ankle, the pattern of change with speed is substantially different across these levels. This is discussed in more detail in latter sections of this discussion.

Dingwell et al. (2001) have shown that, despite the fact that there is no biomechanical reason to expect locomotion on a treadmill to be different to over-ground locomotion (van Ingen Schenau, 1980), the magnitude of stride to stride variations is reduced and that local dynamic stability is greater in treadmill versus over-ground walking. Given the break down in long range correlations that occur when individuals walk in time to a metronome (Hausdorff et al., 1996; Terrier et al., 2005) it might have been anticipated that the constantly driven speed of the treadmill would eliminate the long range correlations in the stride interval. These experiments demonstrate clearly that this is not the case, although the strength of correlations in these experiments is somewhat weaker than that found previously for over ground walking.

That the long range correlations of the stride interval were abolished by walking in time to a metronome but not by the constant speed of the treadmill shows that this breakdown is task specific. A possible explanation for this finding may be related to the overall goal of the task. Under the constraint of auditory pacing, locomotion becomes a timing task and the time series of stride intervals is in effect a time series of inter-response intervals. It is well established that there is a negative covariation between adjacent time intervals during repetitive response tasks (e.g., Wing & Kristofersson, 1973a, 1973b). The expected result of this is, therefore, that gait cycle fluctuations become anti-persistent, which is what happens (Hausdorff et al., 1996). In contrast to walking in time to a metronome, the task of walking on a motor driven treadmill involves no explicit time keeping goal. Thus, while treadmill locomotion reduces the variability associated with the stride interval time series, the long range correlations remain intact.

Also of note is the observation in Experiment 3 that the strength of long range correlation is on average the same for walking and running, although there were differences in the strength of correlations with speed in these two gaits (which will be discussed presently). This suggests that the influence of walking at preferred running speeds is similar to that of running at preferred walking speeds in terms of the movement dynamic, and is consistent with the argument developed in Experiments 1 and 2 that the strength of long range correlations is related to the overall adaptability of the gait cycle.

6.2 Speed Related Changes in the Strength of Long Range Correlations

Having established the ubiquity of long range correlations in human locomotion, the next question this dissertation examined was: what is the influence of changing speed on the strength of long range correlations? This question is two fold. Firstly, does increasing or decreasing speed change the strength of long range correlations? Preliminary evidence for this possibility comes from the experiments of Hausdorff et al. (1996) which indicated the presence of a U-shaped function for alpha over only 3 different walking speeds. Secondly, if speed does influence the strength of long range correlations, is the effect uniform across the different kinematic and kinetic variables examined? It has been shown that fluctuations in the step frequency time series become random-like while walking in time to a metronome, the long range correlations of the step length time series remain (Terrier et al., 2005). Therefore, it is possible that the long range correlations of the different variables may be differentially effected by changes in speed, particularly at extreme speeds. For example, while both stride length and frequency increase in tandem with walking velocity, at very fast walking speeds the stride length saturates and the only way to gain additional increments in speed is through increasing stride frequency (e.g., Hirasaki et al., 1999).

Figure 6.1 shows the results from the DFA for all three experiments – average alpha values for all participants are plotted against the average speeds corresponding to the % of preferred walking (Experiment 1), preferred running (Experiment 2), W-R and R-W transition speed (Experiment 3). This clearly demonstrates that the strength of long range correlations is influenced by speed. In both Experiments 1 and 2 a U-shaped pattern of change with speed was observed for alpha of the stride interval of both waking and running, with the minimum falling at or close to the preferred speed of locomotion. Furthermore, this U-shaped trend was present in the stride length, step interval and length and impulse time series in both walking and running. This pattern of findings supports the prediction that preferred speeds of locomotion have weaker long range correlations compared to speeds faster and slower than preferred. One of the implications of this finding is that the DFA measure is indicative of the number of available dynamical degrees of freedom.

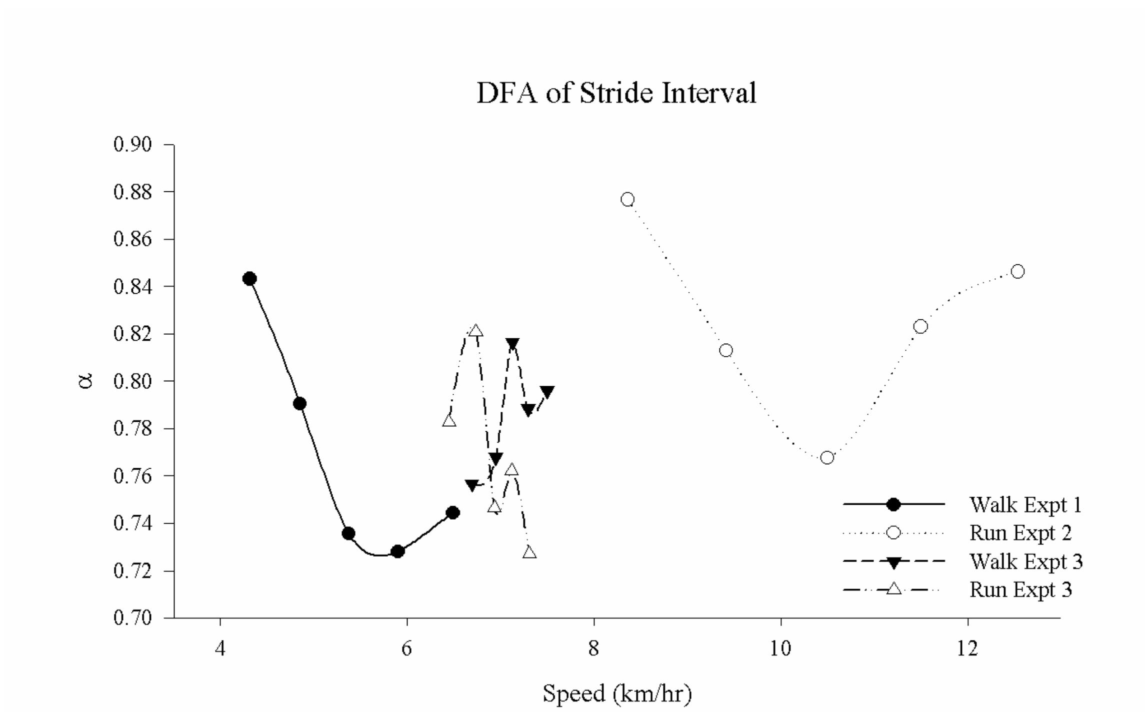

Figure 6.1: Average alpha values of the stride interval across all three experiments plotted against the corresponding average speed calculated for the three experiments.

In both the elderly and in Huntington's patients, the fluctuations of the stride interval exhibit more random-like behavior, and in Huntington's patients, the degree to which the correlations break down is related to the functional impairment of the individual (Hausdorff et al., 1997). This is regarded as being reflective of a breakdown of connections between various elements involved with locomotion. At the other end of the developmental spectrum, young children have significantly larger alpha values than both older children and adults (Hausdorff et al., 1999). The locomotor apparatus is a complex multi-link system with a large number of both mechanical and dynamical degrees of freedom, and this pattern of change in the scaling behavior of fluctuations across the life span (and in disease) may be driven by the process of learning to control the degrees of freedom associated with the system.

Bernstein (1967) proposed a three stage model of learning, with the first stage being a freezing of degrees of freedom, the second a gradual introduction of additional degrees of freedom, and finally in the third stage, exploitation of reactive forces. The age at which acquire an adult-like gait pattern is still an issue for debate (Ganley & Powers, 2005; Sutherland, 1997). According to the results of Hausdorff and colleagues it may be until adolescence before stride to stride control of locomotion is fully mature (Hausdorff et al., 1999). The reduction in size of alpha throughout childhood may reflect the gradual freeing of degrees of freedom, until adulthood, where use of comparatively more degrees of freedom is reflected in a smaller alpha than seen in childhood. In the elderly and disease populations, the problem may lie in an inability to suppress degrees of freedom when necessary due to degradation of both neural and muscular function (e.g., Vaillancourt & Newell, 2002). One interpretation of the relatively larger alpha values at speeds away from preferred, therefore, is that it is reflective of a reduced availability of degrees of freedom and reduced adaptability of the neuromuscular system.

Another interpretation of these findings is that the reduced strength of long range correlations at preferred speeds reflects a lack of control, in that under preferred conditions, there is less need for central contributions to the control of the gait cycle. It has been shown that there is a significant attentional or cognitive cost associated with walking at preferred running speeds (Abernethy, Hanna, & Plooy, 2002). Similarly, studies investigating bimanual coordination have shown that oscillating pendulums at frequencies faster than preferred increases both the variability of the movement pattern and the reaction time to probe secondary task tests (Temprado, Chardenon & Laurent, 2001). Thus, there is an attentional cost to the central nervous system for maintaining non-preferred coordination patterns. The increasing size of alpha at non-preferred speeds of locomotion may similarly be due to the constraints associated with stabilizing a less stable movement pattern.

We know from experience that humans are able to walk at speeds they would prefer to run at and vice versa. Horses can be trained to adopt non-preferred gait patterns over a range of speeds without exhibiting any obvious instabilities (Hoyt & Taylor, 1981), yet when locomoting near typical transition speeds, decerebrated cats are unable to stabilize their gait pattern and alternate between walking and trotting (Shik, Severin & Orlovskii, 1966). Thus, in the mature, neurologically intact animal, there are central mechanisms in place to stabilize a given locomotor pattern under non-preferred conditions. The strength of long range correlation may, therefore, result in part from the increasing influence of central control that becomes a necessary constraint as the gait cycle becomes less stable. Previously it has been suggested that “a more centralized control” of the gait rhythm has the effect of reducing the strength of long range correlations in gait cycle fluctuations (Terrier et al., 2005). However, this speculation was related to the breakdown in long range correlations that occurs when walking in time to a metronome. The gait cycle fluctuations become anti-persistent under these conditions and thus the sequential behavior of the fluctuations can be understood as a lag-one autocorrelation which is typical of time keeping tasks (Wing & Kristofferson, 1973a, 1973b).

As discussed in Experiment 1, the preferred walking speed is the only speed where walking at the resonant frequency of the leg can occur naturally. It has been suggested that walking at the resonant frequency of the leg minimizes the force required to maintain the oscillations of the leg during locomotion (e.g., Holt et al., 1995). Walking faster or slower than the preferred speed will result in an increase or decrease in frequency of leg oscillation. Thus, it is likely that additional muscle force will be required in the form of propulsion or damping to swing the leg faster or slower respectively. In turn this need for additional control at non-preferred speeds may be reflected in the increased size of alpha at these speeds.

The muscle groups that cross the ankle joint appear to become particularly challenged at higher walking speeds. It has been suggested that the increasing force production and decreasing efficiency of these muscles at fast walking speeds may be one of the triggers of the walk to run transition (e.g., Hreljac, 1995; Neptune & Sasaki, 2005). Similarly, Prilutsky and Gregor (2001) observed that there was excessive activation of stance related muscles during slow jogging and of swing related muscles during slow walking that was reduced once the transition to walking (or running) occurred. These findings suggest that there is at least a significant increase in the neural drive to muscles at speeds far from preferred walking or running speeds. The increased size of alpha at these relatively extreme speeds may be related to the increased neural drive. The reduced dynamical degrees of freedom hypothesis is also consistent with this observation in that these muscle groups are approaching the limits of their dynamical range (Hreljac, 1995; Neptune & Sasaki, 2005; Prilutsky & Gregor, 2001).

In summary, the first two experiments showed conclusively that there are U-shaped functions for alpha with gait speed that center on preferred speeds of locomotion. Experiment 3 confirmed that the size of alpha increases as walking speed increases and running speed decreases away from preferred. Two explanations for this phenomenon were proposed: 1) that the size of alpha was reflective of the available degrees of freedom at a given speed; and 2) that the size of alpha was reflective of the degree of control asserted at a given speed. In the literature relating to rhythmic movement, preferred modes of behavior are generally regarded as being more stable than non-preferred modes (e.g., Abernethy et al., 2002; Brisswalter & Mottet, 1996; Temprado, Zanone, Monno & Laurent, 1999). Because of the rhythmic nature of locomotion, and because of the apparent relationship between the size of alpha and preferred walking and running speeds, the idea that the strength of long range correlations may also relate to the stability of walking and running was explored.

6.3 Long Range Correlations and Stability

One way to gain insight into the control of a system is to examine it when it is near a transition or at its most unstable (e.g., Kelso 1995). Therefore, the focus of the third experiment was on the influence of walking and running at and around preferred transition speeds on the strength of long range correlations in the stride interval. The overarching finding for the strength of long range correlations supported the results from the first two experiments – the size of alpha increased with both increasing walking speed and decreasing running speed. Most importantly, inspection of Figure 5.4 from Experiment 3 strongly suggests that switching from a walking to a running gait at the speed calculated to be the preferred W-R transition speed will reduce the size of alpha. Likewise, switching from a running to a walking gait with decreasing speed at the speed calculated to be the preferred R-W transition speed will also reduce the size of alpha. These findings show that extreme speeds constrain the dynamics of locomotion in the same way that extreme levels of isometric force production have been shown to constrain the dynamics of force output (Slifkin & Newell, 1999) and lend further support to the suggestion that the size of alpha is related to the available dynamical degrees of freedom at a given speed of locomotion.

In terms of movement pattern stability there was a decrease in stability of knee-ankle coordination associated with the running gait compared to the walking gait. This was also the case for local dynamic stability of the head and to a lesser degree of the hip. The similarity of the pattern of change of these variables (local dynamic stability of the head and stability of knee-ankle coordination) suggests that there may be a relationship between the stability of coordination patterns and the resistance of the system to local perturbations. It is tempting to speculate that the decrease in local dynamic stability is related to increased local perturbations resulting from increased variability of knee-ankle coordination. However, previous research has shown that although the pattern of change in local stability may be similar to variability, the two are not necessarily related

(Dingwell & Cusumano, 2000). Further research is necessary before any conclusions can be drawn about the relationship between local dynamic stability and stability of the coordination pattern.

The pattern of change in alpha calculated for the peak to peak interval of the head also followed this pattern of changes seen in the stability measures, in that alpha was greater for running than walking. In addition to this overall trend, generally it appeared that for the ankle, local dynamic stability was greater in the running than in the walking condition, with the size of alpha for the peak to peak interval at the ankle being greater in walking than in running. Thus, the long range correlations are influenced by both local dynamic stability and the stability of the coordination pattern. As such it is likely that the scaling behavior of the gait cycle fluctuations is related to both the cognitive cost of stabilizing non-preferred behaviors (Abernethy et al., 2002; Temprado et al., 1999) as well as the reduction in dynamical degrees of freedom that likely occurs at these speeds (cf. Slifkin & Newell, 1999).

Preferred modes of behavior, particularly rhythmic behaviors, are frequently associated with the notion of an attractor in motor control (e.g., Brisswalter & Mottet, 1996; Turvey, 1990). An attractor is regarded as a state (or series of states) that a system gravitates to from initial conditions and returns to following perturbation (e.g., Turvey, 1990). While the results of this dissertation do not address directly the existence of attractors for walking or running, the pattern of change of alpha with speed is consistent with the idea that there is on some level, a driving of individual's towards preferred speeds of locomotion and away from non-preferred speeds. In any case, in a multi-degree of freedom system that faces a large number of constraints such as the locomotor system, there will necessarily be tradeoffs with regard to different aspects of stability, such that the primary goals of maintaining an upright position and forward progression are met (Newell & McDonald, 1994). It may be that alpha reflects this overall movement stability.

6.4 Variability at the Head vs. the Ankle

One interesting result that emerged from Experiment 3 was the finding that there are differences in the strength of long range correlations in the time series of inter-peak intervals (the equivalent of inter-stride interval) at the head and ankle as a function of gait and speed. DFA of the head marker followed the same pattern of results as both the SD of knee-ankle coordination and the local dynamic stability of the head. In contrast, while the pattern of change of alpha and lambda* of the ankle were strikingly similar, the DFA of the ankle was largely unaffected by the variability of knee-ankle coordination. Lastly, head stability was decreased and alpha increased during running whereas ankle stability was improved and alpha was smaller during running.

These anatomical location-dependent differences in the structure of variability may simply be a by-product of the transmission of fluctuations across the different levels of the multi-link locomotor apparatus or they may be functionally relevant. If the latter speculation is correct, these findings may relate to the muscles crossing the ankle joint, the force producing capabilities of which have been shown to be constrained at fast walking speeds. Hreljac (1995) identified angular acceleration of the ankle joint as a likely trigger for the transition from walking to running, based on the observation that during walking at high speeds there was a much larger acceleration of the ankle as compared with running at the same speed. This was also associated with increasing muscle activity in the dorsiflexors to clear the toes of the ground during toe off. Participants reported discomfort in the dorsiflexors and it was suggested that the muscles were working close to their maximum capacity.

Neptune and Sasaki (2005) have reported that despite an increase in muscle activation, there is a significant decrease in the ground reaction force during the propulsive phase of the gait cycle at fast walking speeds. The plantarflexors are the main contributors to the ground reaction force during this phase of the gait cycle, suggesting that their force generating capacity is compromised at high speeds. Further, simulations showed that once the transition to running occurred, there was a dramatic increase in force production for a similar level of muscle activation. The relatively larger alpha of the ankle during walking, particularly in comparison to running at post W-R transition speeds is consistent with the idea that the degrees of freedom of the muscle groups crossing the ankle joint are limited.

In contrast to this, it is likely that running at slow speeds has a particularly destabilizing influence on the head. This is supported by the results of Experiment 3 which show that stability of the head is reduced during running compared to walking. Keller et al. (1996) demonstrated that in running at slower speeds (comparable to the speeds used in Experiment 3) the peaks in vertical ground reaction force were much greater in running than in walking. Data from Experiment 3 confirm this. Therefore, it is possible that the relatively larger forces during slow running, combined with greater vertical excursions of the center of mass under these conditions have a destabilizing influence on the head that is reflected in a relatively larger LSEs. In order to maintain head stability, head position may be controlled to a greater extent, and as such alpha becomes larger.

6.5 Amount vs. Structure of Variability

As discussed in chapter 2 of this dissertation, there are a number of studies that have studied the influence of speed on the amount of variability of the gait cycle, the results of which are inconsistent. It is likely that this is due to a combination of methodological differences, including the range of speeds investigated, whether locomotion was constrained or unconstrained, and how many steps were collected. The importance of investigating a wide range of both preferred and non-preferred speeds has been emphasized in this dissertation. Considering each experiment alone it appears that the changes with speed are either linear or curvilinear – there is no evidence of a U-shaped function. In contrast, when the data from all three experiments are examined together (Figure 6.2) it is apparent that both SD and CV of the stride interval of walking follow a U-shaped trend when a broad range of speeds are considered. The stride length and impulse data of the three experiments is consistent with that of the stride interval. The minimum of these curves fall at a speed slightly faster than preferred walking speed (calculated in Experiment 1 to be on average 5.4 km/hr) but is still within the range of energetically optimal walking speeds (Margaria, 1976).

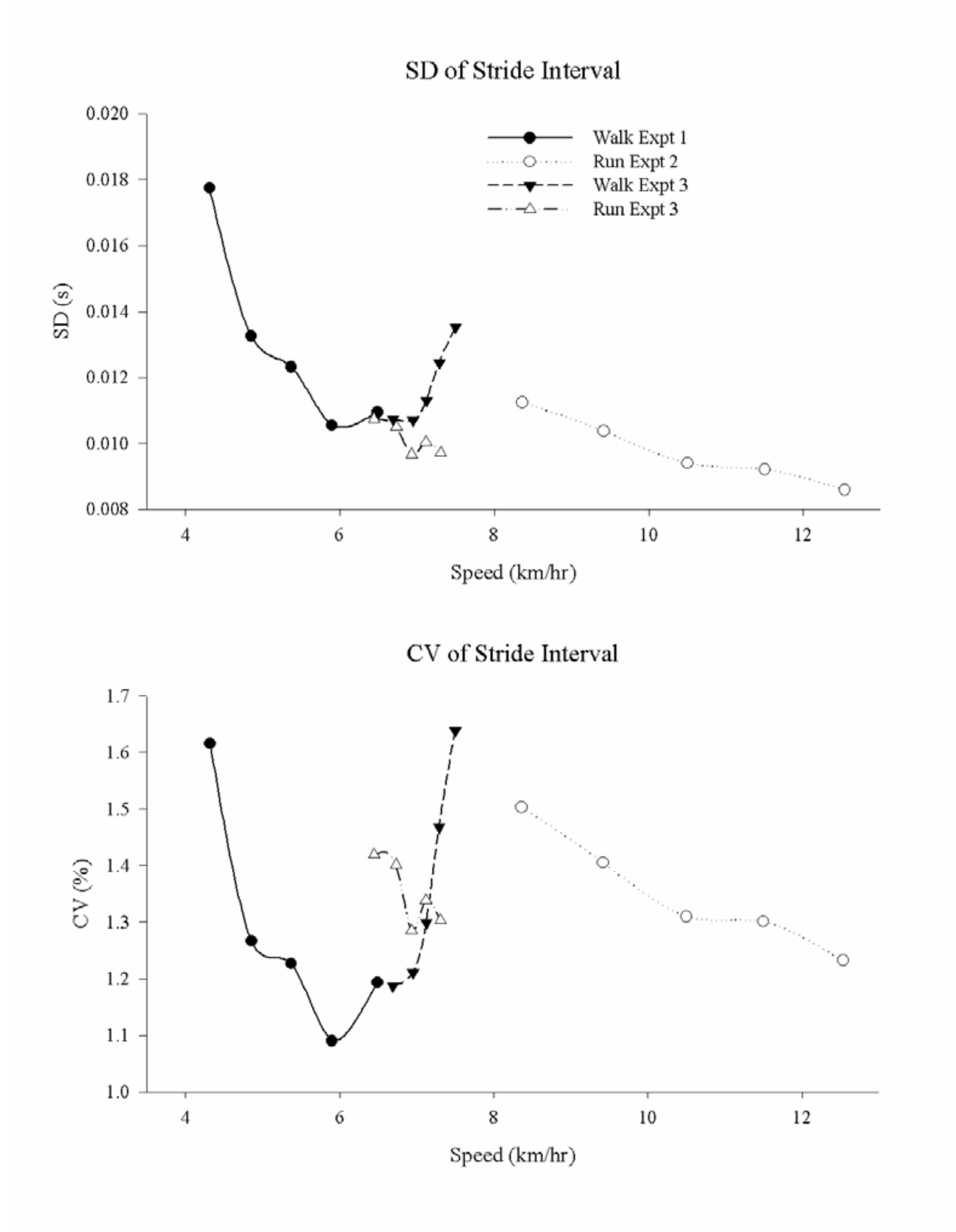

Figure 6.2: Absolute and relative variability of the stride interval across all three experiments.

There is no U-shaped function for variability with speed for running apparent in Figure 6.2. The most likely explanation for this is that the range of speeds examined in this study, while broad, does not cover the range of possible running speeds. The upper end of this range was not examined because the length of time the running trials were collected for in Experiment 2 made running at faster speeds prohibitive. If a U-shaped function for variability does exist for running, it is evident from Figure 6.2 that it is a shallower U-shaped function than that for walking. The variability curves of the stride interval bear a resemblance to that of metabolic transport cost curves for walking and running, where a U-shaped pattern emerges for walking and a relatively flat line emerges for running (Margaria et al., 1963). Recently, there has been evidence to suggest that the metabolic transport cost data for running are better fit by a quadratic function than a linear function and that particularly at slow running speeds there is an increase in the metabolic transport cost (Daniels, 2002).

Comparison of Figures 6.1 and 6.2 reveals a striking similarity between the pattern of change of DFA and CV for the walking gait but not for the running gait. For the walking data, this pattern of results shows that the speed at which variability is minimized is also the speed at which the least degree of structure is present in the time series. This is consistent with the findings of Slifkin and Newell (1999) for isometric force production who demonstrated that the signal to noise ratio (the inverse of CV) and ApEn (a measure of the structure of variability that takes on values between 0 and 2, with increasing values indicating increasing structure) both followed an inverted-U shaped pattern with increasing force production. They concluded that information transmission related to isometric force production occurred when the greatest degree of structure was present in the force output. A parallel interpretation of the walking data is that maximum and minimum speeds constrain the dynamics of walking such that the available degrees of freedom are limited at these extreme speeds. In the range of "normal" walking speeds, these constraints are reduced and there is an increase in the number of independently controlled system elements contributing to system output, increasing the overall adaptability of the system.

Nevertheless, the relationship between the amount and structure of gait cycle variability for running is markedly different. While there is a U-shaped function for alpha and speed, the amount of variability decreases essentially linearly with running speed. There is evidence to suggest that the variability function for running may be U-shaped when faster speeds are examined (e.g., Belli et al., 1995). If this is the case, the implication is that the DFA measure is more sensitive to changes in speed than global measures of variability. In any case, the U-shaped function for DFA with speed in running shows that, despite a linear change in overall variability, there are subtle changes in the interaction between the many different elements of the locomotor apparatus that are evident in the structure of gait cycle variability.

A final comment on the amount versus the structure of variability is appropriate at this time. When the results of three experiments of this dissertation are considered as a whole, as in Figures 6.1 and 6.2, there is a similar pattern of change in both the amount and structure of variability of walking as a function of speed. However, the patterns of change are different in the running gait, with alpha following a U-shaped pattern and SD and CV decreasing with increasing running speed. Furthermore, there are differences between the ankle and head both in terms of local dynamic stability and in terms of the strength of long range correlations that are not evident in either the SD or CV of the time between peaks at either the head or ankle. This underscores the importance of looking to the time dependent structure of time series when trying to understand the nature of control. There is a large body of literature focused on using the amount of variability as a primary outcome measure yet it is becoming increasingly apparent that there is a substantial amount of relevant information that is missed using this approach. In contrast, both the detrended fluctuation measure and the local stability exponent measure capture subtle differences in movement dynamics that are not reflected in the global measures of amount of variability.

6.6 Conclusions and Future Directions

In summary, this dissertation has shown the following: 1) long range correlations are present in a large number of kinematic and kinetic gait cycle variables; 2) long range correlations are present in both the walking and running gait; 3) the long range correlations are reduced at preferred speeds in both walking and running and increase at speeds faster and slower than preferred; 4) there is a relationship between strength of long range correlations and stability such that the strength of correlation increases as locomotion becomes less stable; and 5) the size of the alpha is sensitive to differences in control at the foot and ankle. Collectively the results of these experiments suggest that detrended fluctuation analysis is revealing about the number of degrees of freedom available under given conditions, or conversely the degree of constraint that results from a set of conditions. An alternative but not mutually exclusive possibility is that the size of alpha is related to the degree of active control associated with locomotion under different circumstances. Thus, as the speed of locomotion moves increasingly away from preferred speeds, structure is introduced to the variability as a result of: a) increasing constraints, b) decreasing degrees of freedom and c) increasing levels of control.

One area for future exploration is that of the effect of cognitive load or attention on the scaling behavior of gait cycle fluctuations. Abernethy et al. (2002) have shown that walking at preferred running speeds is costly in terms of attention. This dissertation shows that long range correlations increase when walking at preferred running speeds (and vice versa) and I have suggested that this may be related in part at least to the increased attentional cost associated with these conditions. This possibility could be examined by introducing a secondary task during locomotion, (for example a mental arithmetic task), with the prediction being that this would serve to increase long range correlations in gait cycle fluctuations.

Another issue that warrants further exploration is the influence of grade on the scaling behavior of gait cycle fluctuations. Extreme grades may be regarded in a similar context as extreme speeds in that they serve as boundary conditions that constrain the dynamics of walking. If the pattern of change of alpha with speed is related to dynamical degrees of freedom and constraints as hypothesized in this dissertation, it can be predicted that as participants walk up (or down) grades of increasing magnitude the pattern of change will be similar to that of speed.

Lastly, another area of research that should be examined is whether these findings transfer to over-ground locomotion. Locomotion on a motor driven treadmill constrains participants to an almost constant speed, while locomotion over-ground does not have this constraint. Further, treadmill locomotion increases local dynamic stability and decreases the amount of variability in gait cycle fluctuations (Dingwell & Cusumano, 2000). Therefore, it is reasonable to speculate that the pattern of findings revealed in this dissertation may actually be stronger in over-ground than treadmill locomotion.

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.

Appendix A: Informed Consent/Experiments 1 & 2

Informed Consent Form for Clinical Research Study

Pennsylvania State University

Title of Investigation: Variability in Locomotion

Principle Investigator: Kimberlee Jordan (266 Recreation Bldg., 863-4037)

Other Investigators: Karl Newell (201 Henderson Bldg., 863-2426)

Date ____________

This is to certify that I, ________________________, have been given the following information with respect to my participation as a volunteer in a program of investigation under the supervision of ___________________________.

1. Purpose of the study:

To investigate variations in timing and spacing of successive strides during walking and running at different speeds.

2. Procedures to be followed:

This study involves manipulation of both the speed (80 – 120% of preferred) and type (walking or running) of locomotion. You will be asked to perform both walking and running trials at five different speeds: 80, 90, 100, 110 and 120% of your preferred walking or running speed. The walking and running trials will be conducted over two days with all walking trials being performed on one day and all running trials performed on a separate day. Each session should take no longer than one and a half hours. In addition to these test days, there will be a pretesting session to familiarize you with walking and running on the treadmill. During this session you will be asked to walk and run at self selected speeds on the treadmill for up to 45 minutes

At the start of each experimental day, you will be weighed using the force platforms. Following this, there will be a 15 minute warm up/habituation period of walking (for the walking experiment) or walking and running (for the running experiment) during which time the preferred walking or running speed was established. During the warm up/habituation period, you will not be able to view (via the treadmill display screen) the speed at which you are walking or running, however you will be free to increase or decrease your speed at will using the treadmill controls. Once the preferred walking or running speed has been established, the experimenter will calculate 80, 90 110 and 120% of this speed.

At the end of the warm up/habituation period you will be given up to five minutes to stretch after which time data collection trials will commence. One trial will be performed per walking or running speed and trial speeds will be presented randomly(by chance). During the walking trials you will be walking for between 10 and 15 minutes per speed. During the running trials you will be running for between 6 and 10 minutes, such that approximately 600 stride intervals are captured per trial (the trials carried out at slower speeds will require longer data collection periods). You will be given at least 2 minutes and up to 10 minutes to recover between trials.

3. Discomfort and risks:

The primary risks involved with this study concern the stress to the heart and leg muscles presented by work on a motorized treadmill. According to the American College of Sorts Medicine Guidelines for Exercise Testing and Prescription, the risk of death during a *maximum* effort exercise test is minor (0.5 per 10,000 tests), but does exist. Other potential risks include shortness of breath, fainting, nausea, muscle strain and muscle soreness. The risk of falling is no greater than during over ground running, in fact due to a smooth consistent surface the risk of falling may be reduced in treadmill running. There are no known risks associated with the recording of force, spatial and temporal information using force platforms.

All treadmill work during this study will occur at intensities less than maximum effort so the chance of risk can be considered to be less than stated above. You understand that you will complete a medical screening questionnaire given to you by the investigator before you participate. The purpose of this questionnaire is to rule out any conditions you might have that could place you at greater risk than normal. The investigator will be present at all times and will follow the proper procedures for stopping the treadmill if you experience any of the symptoms listed above. If an emergency situation occurs, access to medical care at Ritenour Health Center or Mount Nittany Medical Center is available via a telephone located in the CELOS laboratory.

4. a. Benefits to me (the participant): Although the work loads examined in this study are below those required to gain a substantial training effect, you may enjoy this opportunity for some exercise. You will also receive financial compensation for your participation in this study.

b. Benefits to society: Society will benefit from the publication of results of this study. The information gained in this study will contribute to the understanding of the control of locomotion.

5. Time duration of the procedures and study:

The data collection will be divided into two sessions carried out over two days. Each session will last up to one and a half hours. Prior to these sessions there will be a

45 minute visit that will allow you to become familiar with walking and running on the treadmill.

6. Statement of confidentiality:

Your participation in this research is confidential. Only the investigator and his/her assistants will have access to your identity and to information that can be associated with your identity. In the event of publication of this research, no personally identifying information will be disclosed.

7. You must be 18 year of age or older to participate in this study.

8. Right to ask questions:

You have been given an opportunity to ask any questions you may have, and all such questions or inquiries have been answered to your satisfaction. You may contact the Office for Research Protections, 212 Kern Graduate Building, University Park, PA16802, (814) 865-1775 for additional information concerning my right as a research participant.

9. Compensation:

$10 per day of testing will be provided to you for compensation of your time, as well as $10 for the initial accommodation session where no data are collected.

10. Injury Clause:

Medical care is available in the event of injury resulting from research but neither financial compensation nor free medical treatment is provided. You are not waving any rights that you may have against the University for injury resulting from negligence of the University or investigators.

11. Voluntary participation:

Your participation in this study is voluntary, and you may withdraw from this study at any time by notifying the investigator. Your withdrawal from this study or your refusal to participate will in no way affect your care or access to medical services. You may decline answers to specific questions.

This is to certify that I consent to and give permission for my participation as a volunteer in this program of investigation. I understand that I will receive a signed copy of this consent form. I have read this form and understand the content of this consent form.

Volunteer Date

I, the undersigned, have defined and explained the studies involved to the above volunteer

Investigator Date

Appendix B: Informed Consent/Experiment 3

Informed Consent Form for Clinical Research Study

Pennsylvania State University

Title of Investigation: Variability in Locomotion

Principle Investigator: Kimberlee Jordan (266 Recreation Bldg., 863-4037)

Other Investigators: Karl Newell (201 Henderson Bldg., 863-2426)

Date ____________

This is to certify that I, _________________________, have been given the following information with respect to my participation as a volunteer in a program of investigation under the supervision of ___________________________.

1. Purpose of the study:

To investigate variations in timing and spacing of successive strides during walking and running at different speeds and during gait transitions.

2. Procedures to be followed:

This study involves manipulation of both the speed (85 – 115% of preferred gait transition speed) and type (walking or running) of locomotion. You will be asked to perform both walking and running trials at seven different speeds: 85, 90, 95, 100, 105, 110 and 115% of your preferred walk to run (W-R) and run to walk (R-W) transition speeds. Data collection will take place over two days, each session should take no longer than two hours. In addition to these test days, there will be a pretesting session to familiarize you with walking and running on the treadmill. During this session you will be asked to walk and run at self selected speeds on the treadmill for up to 45 minutes. If you have been a participant in my previous walking and running experiment you will not be required to undergo the pretesting session as you will already be familiar with walking and running on treadmills.

On the experimental days I will be collecting kinematic data using passive markers and a motion analysis system. This will require that I attach (using double sided tape) 5 reflective markers, one on each of the following places – your trunk, hip, knee, and ankle joints and your foot.

At the start of each experimental day, you will be weighed using the force platforms. Following this, there will be a 5 - 10 minute warm up/habituation period of walking and running following which the preferred W-R and R-W transition speed will be established. During the warm up/habituation period, you will not be able to view (via the treadmill display screen) the speed at which you are walking or running, however you will be free to increase or decrease your speed at will using the treadmill controls. Once the preferred transition speeds have been established, the experimenter will calculate 85, 90, 95, 100, 105, 110 and 115% of these speeds.

On one of the two data collection days, a stepped increment/decrement speed protocol will be employed and you will be allowed to make the transition from walking (running) to running (walking). Following the determination of gait speeds, you will begin walking (running) on the treadmill at the lowest (highest) speed. Data will be collected for 5min at this speed, after which the treadmill speed will be increased (decreased) by 5% of the transition speed and a further 5min of data will be collected. This procedure will be continued until you have completed one five minute walking (running) or running (walking) trial at each speed. You will be asked not to resist the switch to running (walking) if it feels more comfortable. This procedure will then be repeated with decreasing (increasing) increments in speed.

During the other day of data collection, you will be required to refrain from making the transition from walking (running) to running (walking). Following the determination of gait speeds, you will be asked to walk (run) on the treadmill at the lowest (highest) speed. Data will be collected for 5min at this speed, after which the treadmill speed will be increased (decreased) by 5% of the transition speed and a further 5min of data will be collected. This procedure will continue until you have completed one five minute walking (running) or running (walking) trial at each speed. I ask that you complete all of the different speeds using the same gait that you begin data collection with i.e. if you begin the trials walking, you will walk for all 7 trials, if you begin the trials running you will run for all 7 trials. This procedure will then be repeated with decreasing (increasing) increments in speed.

3. Discomfort and risks:

The primary risks involved with this study concern the stress to the heart and leg muscles presented by work on a motorized treadmill. According to the American College of Sports Medicine Guidelines for Exercise Testing and Prescription, the risk of death during a *maximum* effort exercise test is minor (0.5 per 10,000 tests), but does exist. It is also possible that temporary skin irritation may develop as a result of taping reflective markers to the skin. The risk of this is reduced to some extent by the use of hypoallergenic tape. Other potential risks include shortness of breath, fainting, nausea, muscle strain and muscle soreness. The risk of falling is no greater than during over ground running, in fact due to a smooth consistent surface the risk of falling may be reduced in treadmill running. There are no known risks associated with the recording of force, spatial and temporal information using force platforms.

All treadmill work during this study will occur at intensities less than maximum effort so the chance of risk can be considered to be less than stated above. You understand that you will complete a medical screening questionnaire given to you by the investigator before you participate. The purpose of this questionnaire is to rule out any conditions you might have that could place you at greater risk than normal. The investigator will be present at all times and will follow the proper procedures for stopping the treadmill if you experience any of the symptoms listed above. If an emergency situation occurs, access to medical care at Ritenour Health Center or Mount Nittany Medical Center is available via a telephone located in the CELOS laboratory.

4. a. Benefits to me (the participant): Although the work loads examined in this study are below those required to gain a substantial training effect, you may enjoy this opportunity for some exercise. You will also receive financial compensation for your participation in this study.

b. Benefits to society: Society will benefit from the publication of results of this study. The information gained in this study will contribute to the understanding of the control of locomotion.

5. Time duration of the procedures and study:

The data collection will be divided into two sessions carried out over two days. Each session will last up to two hours. Prior to these sessions there will be a 45 minute visit (pre testing session) that will allow you to become familiar with walking and running on the treadmill.

6. Statement of confidentiality:

Your participation in this research is confidential. Only the investigator and his/her assistants will have access to your identity and to information that can be associated with your identity. In the event of publication of this research, no personally identifying information will be disclosed.

7. You must be 18 year of age or older to participate in this study.

8. Right to ask questions:

You have been given an opportunity to ask any questions you may have, and all such questions or inquiries have been answered to your satisfaction. You may contact the Office for Research Protections, 212 Kern Graduate Building, University Park, PA16802, (814) 865-1775 for additional information concerning my right as a research participant.

9. Compensation:

$15 per day of testing will be provided to you for compensation of your time, as well as $10 for the initial accommodation session (pre testing session) where no data are collected.

10. Injury Clause:

Medical care is available in the event of injury resulting from research but neither financial compensation nor free medical treatment is provided. You are not waving any rights that you may have against the University for injury resulting from negligence of the University or investigators.

11. Voluntary participation:

Your participation in this study is voluntary, and you may withdraw from this study at any time by notifying the investigator. Your withdrawal from this study or your refusal to participate will in no way affect your care or access to medical services. You may decline answers to specific questions.

This is to certify that I consent to and give permission for my participation as a volunteer in this program of investigation. I understand that I will receive a signed copy of this consent form. I have read this form and understand the content of this consent form.

____

Volunteer Date

I, the undersigned, have defined and explained the studies involved to the above volunteer

____

Investigator Date

Appendix C: Informed Consent/Experiment 1 (post-hoc data collection)

Informed Consent Form for Clinical Research Study

Pennsylvania State University

Title of Investigation: Variability in Locomotion

Principle Investigator: Kimberlee Jordan (266 Recreation Bldg., 863-4037)

Other Investigators: Karl Newell (201 Henderson Bldg., 863-2426)

Date ____________

This is to certify that I, _________________________, have been given the following information with respect to my participation as a volunteer in a program of investigation under the supervision of ___________________________.

1. Purpose of the study:

To investigate variations in timing and spacing of successive strides during walking at different speeds.

2. Procedures to be followed:

This study involves manipulation of walking speed (60 – 140% of preferred). You will be asked to perform walking trials at five different speeds: 60, 80, 100, and 120% of your preferred walking or running speed. The data collection will last approximately 45min.

At the start of each experiment, you will be weighed using the force platforms. Following this, there will be a 5 – 10 minute warm up/habituation period of walking during which time the preferred walking speed will be established. During the warm up/habituation period, you will not be able to view (via the treadmill display screen) the speed at which you are walking, however you will be free to increase or decrease your speed at will using the treadmill controls. Once the preferred walking speed has been established, the experimenter will calculate 60, 80 120 and 140% of this speed.

At the end of the warm up/habituation period you will be given up to five minutes to stretch after which time data collection trials will commence. One trial will be performed per speed and speeds will be presented randomly (by chance). Trials will last for 6min and you will be given at least 2 minutes and up to 10 minutes to recover between trials.

3. Discomfort and risks:

The primary risks involved with this study concern the stress to the heart and leg muscles presented by work on a motorized treadmill. According to the American College of Sorts Medicine Guidelines for Exercise Testing and Prescription, the risk of death during a *maximum* effort exercise test is minor (0.5 per 10,000 tests), but does exist. Other potential risks include shortness of breath, fainting, nausea, muscle strain and muscle soreness. The risk of falling is no greater than during over ground running, in fact due to a smooth consistent surface the risk of falling may be reduced in treadmill running. There are no known risks associated with the recording of force, spatial and temporal information using force platforms.

All treadmill work during this study will occur at intensities less than maximum effort so the chance of risk can be considered to be less than stated above. You understand that you will complete a medical screening questionnaire given to you by the investigator before you participate. The purpose of this questionnaire is to rule out any conditions you might have that could place you at greater risk than normal. The investigator will be present at all times and will follow the proper procedures for stopping the treadmill if you experience any of the symptoms listed above. If an emergency situation occurs, access to medical care at Ritenour Health Center or Mount Nittany Medical Center is available via a telephone located in the CELOS laboratory.

4. a. Benefits to me (the participant): Although the work loads examined in this study are below those required to gain a substantial training effect, you may enjoy this opportunity for some exercise.

b. Benefits to society: Society will benefit from the publication of results of this study. The information gained in this study will contribute to the understanding of the control of locomotion.

5. Time duration of the procedures and study:

The study will last approximately 45 min of one day.

6. Statement of confidentiality:

Your participation in this research is confidential. Only the investigator and his/her assistants will have access to your identity and to information that can be associated with your identity. In the event of publication of this research, no personally identifying information will be disclosed.

7. You must be 18 year of age or older to participate in this study.

8. Right to ask questions:

You have been given an opportunity to ask any questions you may have, and all such questions or inquiries have been answered to your satisfaction. You may contact the Office for Research Protections, 212 Kern Graduate Building, University Park, PA16802, (814) 865-1775 for additional information concerning my right as a research participant.

9. Compensation:

No financial compensation will be provided.

10. Injury Clause:

Medical care is available in the event of injury resulting from research but neither financial compensation nor free medical treatment is provided. You are not waving any rights that you may have against the University for injury resulting from negligence of the University or investigators.

11. Voluntary participation:

Your participation in this study is voluntary, and you may withdraw from this study at any time by notifying the investigator. Your withdrawal from this study or your refusal to participate will in no way affect your care or access to medical services. You may decline answers to specific questions.

This is to certify that I consent to and give permission for my participation as a volunteer in this program of investigation. I understand that I will receive a signed copy of this consent form. I have read this form and understand the content of this consent form.

____

Volunteer Date

I, the undersigned, have defined and explained the studies involved to the above volunteer

____

Investigator Date

Appendix D: Physical Activity Readiness Questionnaire (PAR-Q)

Physical Activity Readiness Questionnaire - PAR-Q (revised 2002)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

YES	NO		
☐	☐	1.	**Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?**
☐	☐	2.	**Do you feel pain in your chest when you do physical activity?**
☐	☐	3.	**In the past month, have you had chest pain when you were not doing physical activity?**
☐	☐	4.	**Do you lose your balance because of dizziness or do you ever lose consciousness?**
☐	☐	5.	**Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?**
☐	☐	6.	**Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?**
☐	☐	7.	**Do you know of any other reason why you should not do physical activity?**

If you answered

YES to one or more questions

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want — as long as you start slowly and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

NO to all questions

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

DELAY BECOMING MUCH MORE ACTIVE:

- if you are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better; or
- if you are or may be pregnant – talk to your doctor before you start becoming more active.

PLEASE NOTE: If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

Informed Use of the PAR-Q: The Canadian Society for Exercise Physiology, Health Canada, and their agents assume no liability for persons who undertake physical activity, and if in doubt after completing this questionnaire, consult your doctor prior to physical activity.

No changes permitted. You are encouraged to photocopy the PAR-Q but only if you use the entire form.

NOTE: If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

NAME ____

SIGNATURE ____ DATE ________________________________

SIGNATURE OF PARENT or GUARDIAN (for participants under the age of majority) ________________________________ WITNESS ________________________________

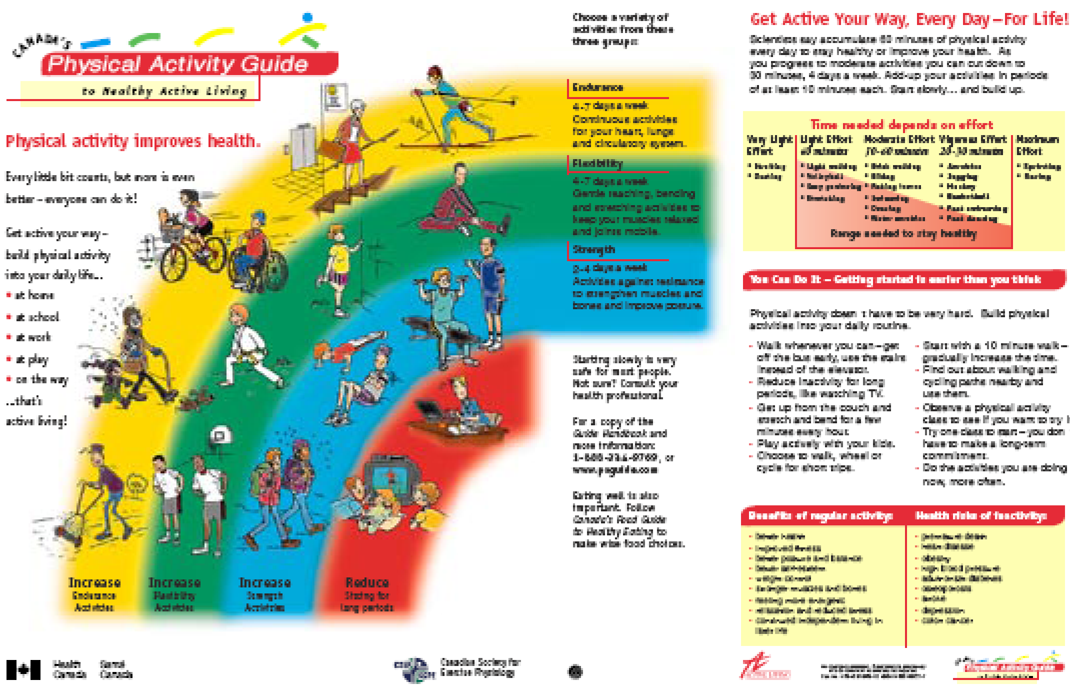
Note: This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.

continued on other side...

...continued from other side

PAR-Q & YOU

Physical Activity Readiness
Questionnaire - PAR-Q
(revised 2002)



Source: Canada's Physical Activity Guide to Healthy Active Living, Health Canada, 1998 http://www.hc-sc.gc.ca/hppb/paguide/pdf/guideEng.pdf

FITNESS AND HEALTH PROFESSIONALS MAY BE INTERESTED IN THE INFORMATION BELOW:

The following companion forms are available for doctors' use by contacting the Canadian Society for Exercise Physiology (address below):

The **Physical Activity Readiness Medical Examination (PARmed-X)** — to be used by doctors with people who answer YES to one or more questions on the PAR-Q.

The **Physical Activity Readiness Medical Examination for Pregnancy (PARmed-X for Pregnancy)** — to be used by doctors with pregnant patients who wish to become more active.

References:

Arraix, G.A., Wigle, D.T., Mao, Y. (1992). Risk Assessment of Physical Activity and Physical Fitness in the Canada Health Survey Follow-Up Study. **J. Clin. Epidemiol.** 45:4 419-428.

Mottola, M., Wolfe, L.A. (1994). Active Living and Pregnancy, In: A. Quinney, L. Gauvin, T. Wall (eds.), **Toward Active Living: Proceedings of the International Conference on Physical Activity, Fitness and Health.** Champaign, IL: Human Kinetics.

PAR-Q Validation Report, British Columbia Ministry of Health, 1978.

Thomas, S., Reading, J., Shephard, R.J. (1992). Revision of the Physical Activity Readiness Questionnaire (PAR-Q). **Can. J. Spt. Sci.** 17:4 338-345.

For more information, please contact the:

Canadian Society for Exercise Physiology
202-185 Somerset Street West
Ottawa, ON K2P 0J2
Tel. 1-877-651-3755 • FAX (613) 234-3565
Online: www.csep.ca

The original PAR-Q was developed by the British Columbia Ministry of Health. It has been revised by an Expert Advisory Committee of the Canadian Society for Exercise Physiology chaired by Dr. N. Gledhill (2002).

Disponible en français sous le titre «Questionnaire sur l'aptitude à l'activité physique - Q-AAP (revisé 2002)».

Supported by: Health Canada Santé Canada

Appendix E: Detrended Fluctuation Analysis

Detrended fluctuation analysis (DFA – Peng et al.,1993) is a reliable method for examining long range correlations in noisy, non-stationary time series. Because physiological time series are bounded, the first step in the DFA algorithm is to integrate the original time series using the following:

$$y(k) = \sum_{i=1}^{k} [B(i) - B(avg)] \qquad \text{(A.E 1)}$$

where *B(i)* is the ith value in the original time series (e.g. the ith stride interval).

The next step is illustrated in Figure A.E. 1- the integrated time series is divided into windows or boxes of a given size and a least squares line is fit to the data. The "trend" is removed by subtracting the data within a give box from the line fit to that data.

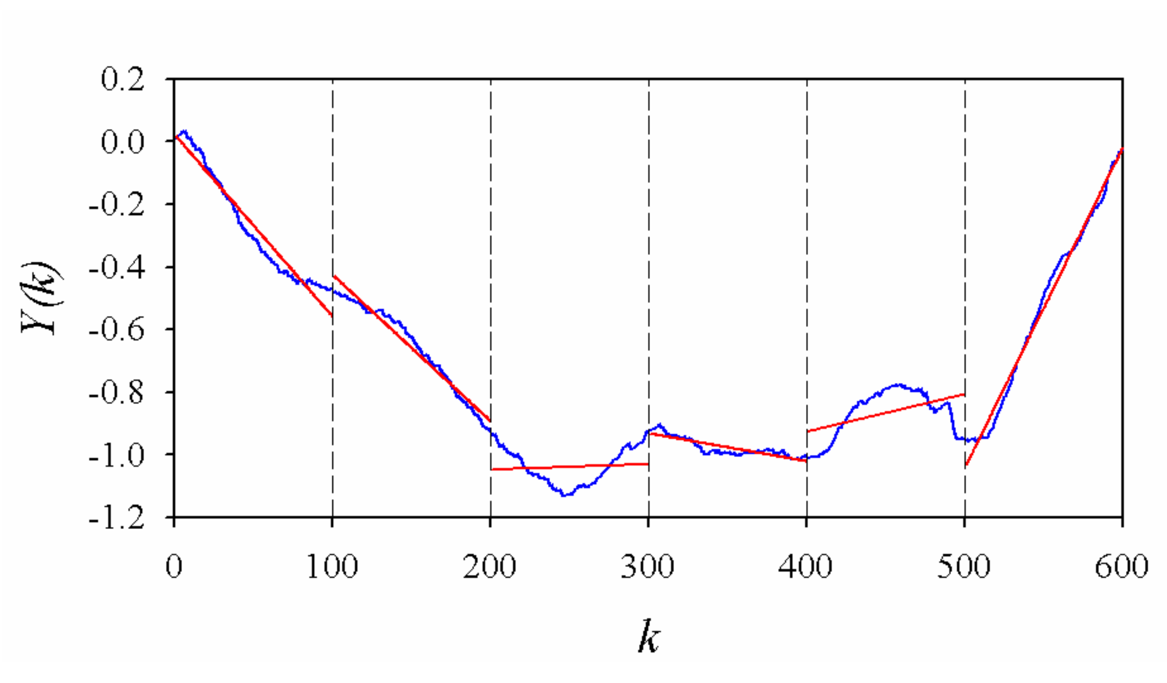


Figure AE.1 The integrated time series given from equation A.E. 1. Dotted lines indicate box sizes of n = 100, with the solid red lines being the least squares fit or "trend" of the data within each box.

The average fluctuation size for a given box size (in this example n = 100) is then calculated. This process is repeated for all time scales, or box sizes ranging from n = 4 to n = N/4, such that the average fluctuation size for each box size is calculated:

$$F(n) = \sqrt{\frac{1}{N}\sum_{k=1}^{N}[y(k) - y_n(k)]^2} \qquad \text{(AE. 2)}$$

where F(n) is the average size of fluctuation for a given box size (n). When F(n) and n are plotted on a double logarithmic graph (Figure A.E. 2), a linear trend indicates power-law scaling of fluctuation size to box size. The slope of this line is the DFA scaling exponent alpha. An alpha-value of 0.5 corresponds to white noise, 0.5 < alpha ≤ 1 indicates power law scaling, alpha = 1 corresponds to pink noise and alpha = 1.5 corresponds to Brownian motion. When alpha >1, correlations still exist but are not of a power-law form.

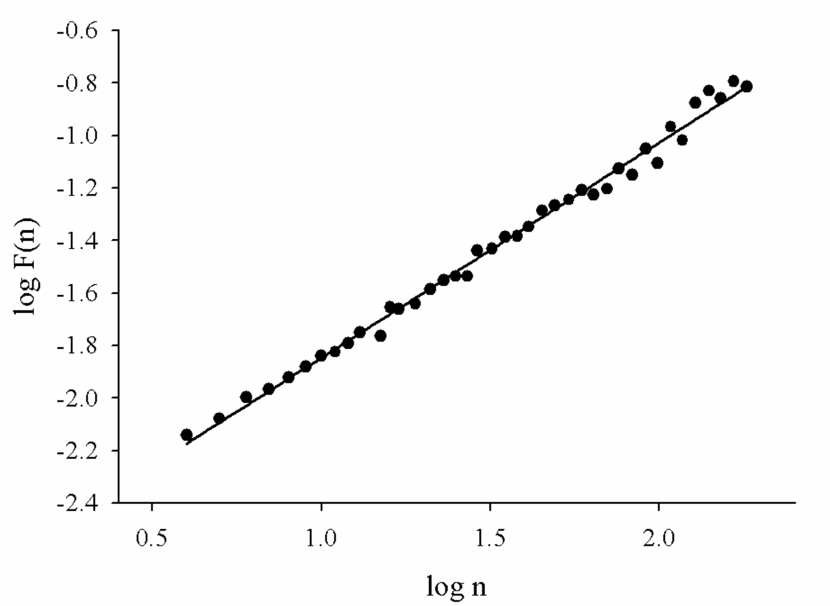


Figure A.E. 2: Slope of line (alpha) relating log of average window size, n, to log of average fluctuation size, F(n).

Curriculum Vitae

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Education

PhD in Kinesiology	Pennsylvania State University, University Park, Pennsylvania, January 2002 – August 2006.
M.S. in Kinesiology	Pennsylvania State University, University Park, Pennsylvania December 2001.
B.PhEd (Hons)	School of Physical Education, University of Otago, Dunedin, New Zealand, December 1998.

Peer-Reviewed Publications

Jordan, K. & Newell, K.M. (2004). Task goal and grip force dynamics. *Experimental Brain Research, 156,* 451-457.

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