THE EFFECT OF TREADMILL WALKING ON THE STRIDE INTERVAL
DYNAMICS OF CHILDREN

by

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for the degree of Master of Applied Science
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Abstract

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The stride interval of typical human gait is correlated over thousands of strides. This statistical persistence diminishes with age, disease, and pace-constrained walking. Considering the widespread use of treadmills in rehabilitation and research, it is important to understand the effect of this speed-constrained locomotor modality on stride interval dynamics. To this end, and given that the dynamics of children have been largely unexplored, this study investigated the impact of treadmill walking, both with and without handrail use, on paediatric stride interval dynamics. An initial stationarity analysis of stride interval time series identified both non-stationary and stationary signals during all walking conditions. Subsequent scaling analysis revealed diminished stride interval persistence during unsupported treadmill walking compared to overground walking. Finally, while the correlation between stride interval dynamics and gross energy expenditure was investigated in an effort to elucidate the clinical meaning of persistence, no simple linear correlation was found.

Dedication

To my grandmother, whose inspiration I will carry with me always.

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List of Symbols

Symbol	Description
fGn	Fractional Gaussian Noise
fBm	Fractional Brownian Motion
DFA	Detrended Fluctuation Analysis
α	Scaling estimate provided by DFA
1/ f boundary	The division between fGn and fBm
OW	Overground Walking
UTW	Unsupported Treadmill Walking
STW	Supported Treadmill Walking
RAT	Reverse Arrangements Test
z_A	Stationarity Test Statistic
K4b ²	Portable Metabolic System
CV	Coefficient of Variation
$\dot{V}O_2$	Mass-specific Gross Oxygen Consumption
VO_2	Mass-specific Gross Oxygen Cost
HR	Heart Rate
r	Pearson's Correlation Coefficient
ρ	Spearman's Correlation Coefficient

Chapter 1

Introduction

1.1 Motivation

During self-paced overground walking, healthy young adults exhibit stride interval persistence extending over thousands of strides (Hausdorff et al., 1996). These dynamics are thought to provide important insight into locomotor control, diminishing with age and in the presence of certain central nervous system diseases (Hausdorff et al., 1997, 1999, 2000; Chau and Rizvi, 2002). Metronomically-paced walking (i.e., stepping to the constant beat of a metronome) alters stride interval dynamics even more drastically, producing anti-persistent time series in lieu of the persistent norm (Hausdorff et al., 1996; Delignieres and Torre, 2009). Given that this locomotor modality imposes a pace-constraint similar to treadmill walking, the observed metronomic effect raises some concern within rehabilitation and scientific communities. In these settings, treadmills are often implemented as part of an intervention or to facilitate gait analysis. From a clinical perspective, a treatment regime that fails to preserve the stride interval dynamics of overground gait would seem to oppose the natural neuromuscular rhythms of healthy locomotion.

To date, the stride interval dynamics of children have been largely unexplored. In the only known paediatric investigation, elevated dynamics were identified and suggested to

reflect a less mature locomotor system (Hausdorff et al., 1999). Considering that developing neuromotor systems may be more susceptible to external gait influences (Forssberg, 1999), and with the emergence of rehabilitation techniques requiring treadmill use (Hesse, 2008; Angulo-Barroso et al., 2008), it is of pertinent clinical importance to study the implication of treadmill walking on paediatric stride interval dynamics.

Accurate quantification of stride interval dynamics requires careful selection of the most appropriate scaling analysis technique, with the choice of method determined by underlying signal properties. This necessarily requires that unknown time series behaviour, particularly stationarity, be investigated before quantification and comparison of stride interval dynamics can effectively proceed.

Finally, if quantification of stride interval dynamics is to become truly useful in gait analysis, the elusive clinical meaning must first be identified. Interestingly, the energetics of locomotion demonstrate similar sensitivities to age, pathology, and pace-constrained walking (Waters and Mulroy, 1999; Morgan et al., 2002; Malatesta et al., 2003; Zarrugh and Radcliffe, 1978), suggesting that stride interval dynamics and energy expenditure may be inherently tied. It is therefore of further relevance to study the association between stride interval persistence and energy efficient locomotion, in an effort to elucidate the clinical relevance of stride interval dynamics.

1.2 Research Question & Objectives

The following research question directed this thesis:

Are the stride interval dynamics of children altered during treadmill walking compared to overground walking?

To answer our research question, and in light of the motivation above, the objectives of this thesis were:

1. To assess the stationarity properties of paediatric stride interval time series;

2. To investigate the effect of treadmill walking, both with and without handrail support, on paediatric stride interval dynamics; and
3. To determine the extent to which stride interval persistence and gross energy expenditure may be linearly dependent.

1.3 Roadmap

Following this introductory chapter, **Chapter 2** introduces the concept of fractal processes and stride interval persistence, and discusses the use of scaling analysis techniques to quantify these dynamics. An overview of the literature pertaining to stride interval dynamics is also presented. In satisfying objective 1, **Chapter 3** includes a full investigation of paediatric stride interval stationarity. Results are discussed in terms of their relevance to the analysis of stride interval dynamics and from a locomotor control perspective. **Chapter 4** addresses objective 2, providing a comprehensive examination into the effect of treadmill walking, both with and without handrail support, on paediatric stride interval dynamics. In fulfillment of the last objective, **Chapter 5** examines the correlation between stride interval persistence and locomotor energy expenditure in children. Finally, **Chapter 6** summarizes the main contributions of this thesis.

The body of this thesis (Chapters 3, 4 and 5) consists of a compilation of three journal articles (one in press and two under review at the time of writing) arising from a single study. Data acquisition (including study protocol and measurement equipment) is described in greatest detail in Chapter 4 (section 4.3.1), such that sections 3.3.1, 5.3.1 and 5.3.2 then represent repeated content. In addition, chapter introductions (section 3.2, 4.2 and 5.2) contain some repeated background information.

Chapter 2

Background

2.1 Fractals in Nature

The fractal phenomenon is one of self-similarity; a property in which a part of an object or process resembles the whole. Interestingly, fractals occur commonly in nature either in the spatial organization of an object or through the temporal fluctuation of a process. To this end, many biological systems have been found to exhibit spatial self-similarity including: branching patterns of bronchial tubes, coronary arterial-venous trees, the structural lining of the intestine and placenta and the structural configuration of bone marrow cells (West et al., 1986; Kassab et al., 1993; Goldberger et al., 1990; Naeim et al., 1996). Likewise, temporal fractals have been identified in physiological and behavioural processes such as: the inter-beat intervals of the heart, fetal breathing dynamics, electromyographic signals, center-of-pressure displacement during stance and, of particular relevance to this research, the stride-to-stride fluctuations of human gait (Kobayashi and Musha, 1982; Szeto et al., 1992; Gupta et al., 1997; Collins and Luca, 1993; Hausdorff et al., 1995; Delignieres and Torre, 2009). Specifically, temporal fractals are described by statistical self-similarity, whereby the statistical properties of part of a time series are proportional to those of the whole (Bassingthwaight et al., 1994).

2.1.1 Stride Interval Dynamics

For many years it was thought that fluctuations in the stride interval, the time between consecutive heel strikes of the same foot, of healthy human gait were strictly random. However, a little over a decade ago, Hausdorff et al. (1995) presented evidence to suggest that stride intervals fluctuate in a complex, fractal-like manner. In other words, a stride interval measured at any point in time was found to depend (statistically) on stride intervals that occurred thousands of strides previously (Hausdorff et al., 1996). This correlation was scale-free and decayed in a power-law fashion, weakening with steps further away in time.

2.2 Quantification of Fractal Dynamics

Fractal analysis of a temporal process focuses on the statistical self-similarity of a signal, with the result providing an estimate of the correlation structure permeating over multiple time scales. A number of scaling analysis techniques exist, which can be applied with varying degrees of effectiveness depending on the very nature of the time series to be examined, either fractional Gaussian noise (fGn) or fractional Brownian motion (fBm) (e.g., see Eke et al., 2000, 2002; Delignieres et al., 2006). For example, for short series, the maximum likelihood estimation method provides good results for fGn, while the scaled windowed variance method seems to work better for fBm (Delignieres et al., 2006; Cannon et al., 1997). On the other hand, for signals at the boundary between fGn and fBm, detrended fluctuation analysis and modified power spectral density seem to work reasonably well with both kinds of series (Delignieres et al., 2006).

2.2.1 Detrended Fluctuation Analysis

Traditionally, detrended fluctuation analysis (DFA) has been used to analyze stride interval dynamics. This technique was originally developed for analysis of non-stationary

heartbeat time series (Peng et al., 1994), and was subsequently applied to stride interval data given its non-stationary appearance. Interestingly however, claims of non-stationarity were purely observational; the stationarity properties of stride interval time series were never tested directly. This thesis addresses this oversight to assess the appropriateness of DFA, and other methods, for quantification of paediatric stride interval time series.

DFA, implemented as described in (Peng et al., 1995), provides an estimate of the strength of correlations in a signal in terms of the scaling exponent, α . Although this technique is unable to attest for the presence of genuine long-range, fractal-like correlations, Delignieres and Torre (2009) recently confirmed long-range dependence in stride intervals using autoregressive fractionally integrated moving average modeling. Thus, one might assume *a priori* that gait is fractal. Nonetheless, processes analyzed with DFA alone are more accurately described in terms of statistical persistence, where the results of DFA are interpreted as follows: $0.5 < \alpha \leq 1.0$ indicates that statistically persistent correlations are present, with the strength of correlation increasing as α approaches 1.0; $\alpha = 0.5$ indicates the data are completely uncorrelated (random); $\alpha < 0.5$ indicates that anti-persistent correlations exist, and $\alpha > 1.0$ indicates the presence of Brownian noise (Hausdorff et al., 1995).

2.3 Significance of Persistence

2.3.1 Trends in the Literature

Stride interval persistence is thought to provide important insight into human locomotor control, with significantly diminished persistence having been identified in the stride interval dynamics of the elderly, children with Spastic Diplegia, and adults with amyotrophic lateral sclerosis, Huntington's and Parkinson's disease (Hausdorff et al., 1997, 2000; Chau and Rizvi, 2002). When considering various measures of gait and balance,

only α was able to distinguish between fallers and non-fallers in a population of older adults with higher-level gait disorder (Herman et al., 2005). Furthermore, compared to healthy young adults, the stride dynamics of children appear to be more highly correlated (Hausdorff et al., 1999). While these elevated dynamics are suggested to reflect a less mature neuromotor system, this conjecture was based on a single study (Hausdorff et al., 1999). This thesis contributes to the body of knowledge pertaining to the stride interval persistence of children, furthering the understanding of developmental dynamics.

2.3.2 Effect of Pace-constrained Locomotion

Interestingly, certain pace-constrained locomotor tasks have also been found to alter stride interval dynamics. For example, Jordan et al. (2007) showed that α increases at walking speeds both faster and slower than preferred, and Delignieres and Torre (2009) demonstrated a change from persistent to anti-persistent behaviour when walking was paced to the constant beat of a metronome. Furthermore, while Frenkel-Toledo et al. (2005) found stride interval persistence to decrease during handrail-supported treadmill walking compared to walking overground with a walker, Chang et al. (2009b) found α to be significantly higher during handrail-supported treadmill walking compared to regular overground walking.

2.3.3 Clinical Relevance

Based on current findings, it is been hypothesized that stride interval persistence largely originates supraspinally, with little dependence on peripheral influences (Chang et al., 2009b; Hausdorff et al., 1996; Hausdorff, 2007; Gates and Dingwell, 2007). Furthermore, in a recent review of gait dynamics, Hausdorff (2007) suggests that the optimal α may occur at a certain appropriate mid-range value, somewhere between perfect correlation (i.e., $\alpha = 1.0$) and uncorrelated random noise (i.e., $\alpha = 0.5$). Unfortunately, while quantification of stride interval dynamics may eventually become a useful clinical tool

to track gait maturity, identify effects of disease, monitor disease progression, and guide treatment regimes (Hausdorff et al., 1999; Hausdorff, 2007), until the precise meaning and physiological origins of stride interval dynamics can be identified, the clinical utility of α remains unclear.

2.4 Locomotor Modalities in Rehabilitation and Research

2.4.1 Overground versus Treadmill Walking

Motorized treadmills are commonly used in clinical and laboratory settings, to facilitate the monitoring of gait measures and to reduce the amount of space required for testing and/or training (Siler et al., 1997; Matsas et al., 2000). However, the extent to which overground walking can be likened to treadmill walking is debatable, with contradictory results in the literature for many gait measures (Riley et al., 2007; Stolze et al., 1997; Alton et al., 1998). To this end, in addition to the distinct walking surface, its dimensions and movement, and the visual constraints associated with treadmill walking (Stolze et al., 1997; Nigg et al., 1995; Dingwell et al., 2001), handrail support may or may not be implemented. On the other hand, overground walking provides more familiar optic flow feedback and the option to freely modulate gait. Evidently, in some cases, the differences between these distinct locomotor modalities affect proprioceptive, vestibular, visual, and exteroceptive afferent information, and consequently influence locomotor control (Deshpande and Patla, 2005; Sorensen et al., 2002; Patla, 2003; Dickstein and Laufer, 2004; Brady et al., 2009; Dixon et al., 2000).

2.4.2 Metronomic versus Treadmill Walking

Stride interval persistence appears to be inherent to a healthy locomotor system. Thus, the apparent interference of metronomically-paced walking on the generation of natural (i.e., overground) stride interval dynamics has caused some to question the impact of treadmill walking, which imposes a similar pace-constraint. From a rehabilitation perspective, an intervention that fails to strengthen or preserve the dynamics of typical locomotion would seem to oppose the natural neuromuscular rhythms of healthy human gait. To this end, while adult stride interval persistence is not significantly different between overground and hands-free treadmill walking (Chang et al., 2009b), the effect of treadmill walking on the gait dynamics of children is unknown. Considering the importance and widespread application of such a locomotor modality in many emerging paediatric rehabilitation techniques (Cherng et al., 2007; Angulo-Barroso et al., 2008; Mutlu et al., 2009), it is of critical value to study the implications of treadmill walking on developmental stride interval dynamics. The research presented within this thesis addresses this important issue.

2.5 Stride Interval Dynamics and The Energetics of Locomotion

Human locomotion is shaped by a preference for energy efficient movements (Patla and Sparrow, 2000), while α is thought to reflect a certain extent of locomotor control (Hausdorff, 2007). Interestingly, the aforementioned sensitivities of stride interval dynamics to age, pathology, and pace-constraints, have similarly been identified in measures of energy expenditure obtained during locomotion. To this end, increased energy requirements have been linked with pathology, immature gait, advanced ageing, and forced-rate stepping (Waters and Mulroy, 1999; Morgan et al., 2002; Malatesta et al., 2003; Zarrugh

and Radcliffe, 1978), while Browning and Kram (2005) show that gross cost of transport is minimized near a subject's preferred walking speed. These seemingly congruent observations have prompted some hypotheses of a close association between stride interval persistence and measures of energy expenditure. Ultimately, examination of such a dependence could serve to further our understanding of stride interval persistence, and may help to elucidate the clinical meaning of α . This thesis includes a first look at the possible association between energy expenditure and stride interval dynamics in children.

Chapter 3

An Investigation of Stride Interval Stationarity in a Paediatric Population

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Chapter 3 addresses the first objective of this thesis. In this chapter, stride interval time series obtained from a paediatric population are tested for weak stationarity. Results of this study serve to guide selection of an appropriate scaling analysis technique for subsequent quantification of stride interval persistence (Chapter 4).

Note: Section 3.2 contains some repeated background information and the methodological details of section 3.3.1 are described more comprehensively in section 4.3.1.

3.1 Abstract

Fluctuations in the stride interval of human gait have been found to exhibit statistical persistence over hundreds of strides, the extent of which changes with age, pathology and speed-constrained walking. Thus, recent investigations have focused on quantifying this scaling behaviour in order to gain insight into locomotor control. While the ability of a given analysis technique to provide an accurate scaling estimate depends largely on the stationary properties of the given series, direct investigation of stride interval stationarity has been largely overlooked. In the present study we test the stride interval time series obtained from able-bodied children for weak stationarity. Specifically, we analyze signals obtained during three distinct modes of self-paced locomotion: (i) overground walking, (ii) unsupported (hands-free) treadmill walking, and (iii) handrail-supported treadmill walking. Using the reverse arrangements test, we identify non-stationary signals in all three walking conditions and find the major known cause to be due to time-varying first and second moments. We further discuss our findings in terms of locomotor control and the differences between the locomotor modalities investigated. Overall, our results advocate against scaling analysis techniques that assume stationarity.

3.2 Introduction

The stride interval, defined as the time between consecutive heel strikes of the same foot, has been increasingly studied in recent years. In particular, much interest has lied in quantifying the statistical persistence of stride interval time series which are known to be correlated up to thousands of strides (Hausdorff et al., 1996). Since the original discovery of these persistent fluctuations (Hausdorff et al., 1995), scaling estimates have shown sensitivity to ageing, pathology, and speed-constrained gait, as well as the potential to differentiate between fallers and non-fallers (Hausdorff et al., 1997, 1999, 2000; Chau and Rizvi, 2002; Jordan et al., 2007; Herman et al., 2005). Thus, through careful quantifi-

cation of the underlying scaling behaviour, stride interval analysis may provide us with a deeper understanding of the locomotor control system and could eventually become a useful clinical tool.

The ability of a particular analysis technique to provide an accurate scaling estimate depends largely on the nature of the given series. When dealing with fractal processes, typically assumed a priori to describe stride interval time series (Delignieres and Torre, 2009), it is first necessary to classify signals as either fractional Gaussian noise (fGn) or fractional Brownian motion (fBm) to ensure that the most relevant analysis technique can be applied (Eke et al., 2002). Indeed, the different properties of fGn and fBm processes, where the first is stationary and the second is non-stationary, necessarily require the application of different estimation techniques to ensure meaningful scaling estimates (Delignieres et al., 2006). Unfortunately, matters are further complicated when dealing with series at the $1/f$ boundary, where it becomes difficult to distinguish between fGn and fBm processes using current methods (Eke et al., 2000; Delignieres et al., 2006).

Until recently, investigators were unaware of this need to classify signals as either fGn or fBm prior to the application of scaling techniques. Thus, the interpretation of previous work must be approached with caution (Delignieres et al., 2006). To this end, it would appear that the stationarity of stride interval time series has not been directly and systematically analyzed to date, even though such knowledge would enable a more informed choice of scaling method. Instead, the use of detrended fluctuation analysis (DFA) to quantify the statistical persistence in stride interval time series has seemingly become an automatic and habitual practice, likely due to the earliest reports which made use of this technique (Hausdorff et al., 1996, 1995). Fortunately, DFA does offer the advantage that it is less affected by non-stationarities (Peng et al., 1994) (i.e., time-varying changes in the statistical properties of a process), known to be common in physiological data (Stanley et al., 1999), than alternative methods sometimes adopted such as spectral analysis and rescaled range analysis (Peng et al., 2000). Nonetheless,

careful quantification of stride interval stationarity may provide insight to either justify or refute the choice of DFA as the most appropriate analysis technique for estimating stride interval persistence.

Some qualitative observations have been made surrounding the stationarity of gait. Hausdorff et al. (1995) observed that stride intervals recorded over 3500 strides remained quite stationary, falling between 1.0 and 1.2 seconds for an entire hour-long walking trial. On the other hand, a subsequent investigation by the same group (Hausdorff et al., 1996) revealed the appearance of “variations in the local average with time” for certain individual time series. They further suggested that these apparent non-stationarities may be the result of a loss of focus on the walking task at hand, or due to the absence of external constraints that might otherwise regulate stride interval behaviour. In support of this latter idea, a stationarity analysis of human gait kinematics performed by Dingwell and Cusumano (2000) identified mild non-stationarities in the lower extremity joint angles of some individuals during overground walking, but found them to largely disappear during subsequent treadmill walking. The authors attributed suppression of the non-stationarities, identified as very low frequency drifting, to the externally imposed speed-constraint that is inherent in treadmill walking.

Analysis of stride interval data from a paediatric population is of particular interest since the statistical persistence present in the stride interval time series of children have been largely unexplored. An initial study by Hausdorff et al. (1999) suggested that quantification of the stride-to-stride fluctuations of children may serve to improve the “early detection and classification of gait disorders in children”. In support of this idea, an investigation by Chau and Rizvi (2002) revealed decreased stride interval correlations in children with spastic diplegia when compared to the able-bodied children, of similar age, reported on by Hausdorff et al. (1999). Therefore, in line with the overall effort to establish the quantitative assessment of stride interval persistence for clinical purposes, we investigate the stationarity of paediatric stride interval time series as an important

first step.

In particular, we investigate series obtained in two distinctly different gait environments. We analyze stride interval time series emerging from overground walking in a level hallway; a gait environment similar to that of everyday walking where the individual is free to modulate his or her walking pattern at will. For completeness and comparison, we also analyze time series obtained from treadmill walking (both with and without handrail support). This locomotor modality, often implemented in clinical and research settings, imposes external-constraints including constant optic flow feedback and speed fixation, the latter of which has been suggested to suppress non-stationarities in gait kinematics (Dingwell and Cusumano, 2000).

3.3 Methodology

3.3.1 Data Collection

Stride interval time series were obtained from 31 asymptomatic children (20 female, 11 male) with a mean age of 7.0 ± 1.6 years. Each child completed a total of three, 10-minute walking trials including: (i) overground walking (OW), (ii) unsupported treadmill walking (UTW) (without handrail support) and (iii) supported treadmill walking (STW) (with side-handrail support). Condition sequences were pseudo-randomized, ensuring that each of the six possible permutations was completed once every six participants. Subjects rested for at least seven minutes before each walking trial. Furthermore, subjects were instructed to walk at their own comfortable walking speed as if “walking to school” or “going for a walk in the park”. The preferred UTW and STW speeds were established after the rest period, immediately prior to the start of each respective trial. With the subject walking on the treadmill (GK200T, Mobility Research, USA) at a relatively slow speed, the speed was increased in 0.1 mph increments until the subject reported that his or her preferred speed had been reached. The speed was then increased by at least 0.5

mph and subsequently decreased in 0.1 mph increments until the subject again reported that his or her preferred speed had been reached. This procedure was then repeated and the mean of the four reported speeds was taken as the preferred walking speed for the given treadmill walking condition.

Prior to data collection, subjects were given at least five minutes to become accustomed to treadmill walking and the measurement equipment. This practice period continued until the child reported feeling comfortable with the setup and visually appeared to be walking naturally. Children were recruited through the staff and community programs of Bloorview Kids Rehab (located in Toronto, Ontario, Canada) and the institutional research ethics board approved the study.

During walking trials, heel strike was measured bilaterally via two ultra-thin, force sensitive resistors (Model 406, Interlink Electronics, USA) fastened to the sole of the subject's shoe underneath the heel. Heel contact with the walking surface was reflected by a change in voltage which was sampled at 250 Hz and recorded to the data acquisition card (CF-6004, National Instruments, USA) of a personal digital assistant (Axim x51v, Dell, USA) worn on the subject's abdomen via a waist harness¹.

For each walking trial, data collection was initiated (pre-walk) and terminated (post-walk) while the subject was standing still. Given that the protocol called for analysis of a 10-minute walking period, approximately 10.5 minutes of data were recorded for each trial. This served to ensure that, after removal of start-up and ending effects, full 10-minute gait recordings would be available for analysis.

¹During treadmill walking trials subjects were attached to an overhead safety support (LGJr200, Mobility Research, USA) but were fully weight bearing. Participants also wore a portable metabolic cart (K4b², Cosmed, Italy) throughout the duration of the trials as part of a separate investigation. This system includes a face mask, heart rate monitor, data collection unit and battery, the latter two of which were attached to the subject's back via the waist harness. The total weight of the study equipment worn by subjects was 2.5 kg.

3.3.2 Stationarity

Stationary Series

Assume that X_t is a real-valued random variable representing the observation made at stride interval t and that a series $\{X_t\}$ denotes a family of these real-valued random variables. Without loss of generality, we index the observations such that $t \in \mathbb{Z}$, where $\mathbb{Z}$ is the set of integers. A series $\{X_t\}$ is considered to be *strongly* (or *strict-sense*) stationary if its statistical properties are shift-invariant (Papoulis, 1991), i.e.,

$$f_{X_{t_1}, X_{t_2}, \dots, X_{t_n}}(x_1, x_2, \dots, x_n) = f_{X_{t_1+h}, X_{t_2+h}, \dots, X_{t_n+h}}(x_1, x_2, \dots, x_n) \quad (3.1)$$

where x_i denotes a particular realization of the random variable X_{t_i} , $i = 1, 2, \dots, n$ and $h \in \mathbb{Z}$. More generally, a series is *weakly* (or *wide-sense*) stationary if only its first two moments do not vary with time, such that the mean is constant, i.e.,

$$E(X_{t_1}) = E(X_{t_1+h}) \quad (3.2)$$

and the covariance between two observations made at different times depends only on their time lag and not on their temporal location, i.e.,

$$Cov(X_{t_1}, X_{t_2}) = Cov(X_{t_1+h}, X_{t_2+h}) \quad (3.3)$$

Reverse Arrangements Test

The reverse arrangements test (RAT) is a non-parametric test used to evaluate the weak stationarity of a time series (Bendat and Piersol, 2000). Specifically, the test searches for monotonic trends in the mean square values that are calculated along non-overlapping intervals of a particular signal of interest. The mean square value, given by,

$$E(X^2) = E(X)^2 + Var(X) \quad (3.4)$$

captures the first two moments of the time series for assessment of weak stationarity. The RAT is often used to evaluate the weak stationarity of physiological and biomedical

signals (Alves and Chau, 2008; Chau et al., 2005; Nhan and Chau, 2009; Harris et al., 1993; Hampson et al., 2005; Bilodeau et al., 1997; Novak et al., 1996).

Considering a sample realization of the previously defined time series, $\{x_1, x_2, \dots, x_n\}$, the reverse arrangements test is implemented as follows:

- 1) The sample is divided into M equal and non-overlapping intervals, I_i , where $i = 1, 2, \dots, M$.
- 2) For each interval, the mean square value y_i is calculated, i.e., $y_i = (1/n)\sum_{k \in I_i} x_k^2$, where n is the number of points within each interval and $k = 1, 2, \dots, n$.
- 3) The total number of reverse arrangements, A , present within the sequence of mean square values $y_1, y_2, \dots, y_M$, are counted. A reverse arrangement occurs when one mean square value is greater than a subsequent mean square value, i.e., when $y_i > y_j$ for $i < j$.
- 4) The resulting value, A , is compared to the value that would be expected from a realization of a weakly stationary random process. In the case that the sample time series under consideration is weakly stationary, the expected value of A has a normal distribution with mean $\mu_A = N(N-1)/4$ and variance $\sigma_A^2 = N(N-1)(2N+5)/72$ (Bendat and Piersol, 2000). The null hypothesis that $\{y_i\}$ is weakly stationary is rejected if A falls outside the critical values defined by a significance level of α .

The critical values can be determined from calculation of the stationarity test statistic, z_A , where,

$$z_A = \frac{A - \mu_A}{\sigma_A} \quad (3.5)$$

and $z_A \sim N(0, 1)$. At a significance level α , the critical values are given by $z_{\alpha/2}$ and $z_{1-\alpha/2}$ such that, for a standard normal deviate at a 5% level of significance, we have $z_{\alpha/2} = -1.96$ and $z_{1-\alpha/2} = 1.96$. Comparison of the stationarity test statistic with the critical values at the significance level of interest should be interpreted as follows:

- $|z_A| < z_{1-\alpha/2}$ The null hypothesis that the time series is weakly stationary is accepted.
- $z_A \leq z_{\alpha/2}$ The number of reverse arrangements is less than the number expected of a stationary signal, implying that an upward trend in mean square sequence is present.
- $z_A \geq z_{1-\alpha/2}$ The number of reverse arrangements is greater than the number expected of a stationary signal, implying that a downward trend in the mean square sequence is present.

3.3.3 Data Analysis

Stride Interval Analysis

To extract stride intervals from the heel strike recordings, the initiation and completion of each walking trial was manually selected to remove the extraneous static portion of the recordings (i.e., when the subject was standing stationary). The first 10 seconds of each trial was then eliminated to ensure that the subject had finished accelerating from rest to his or her preferred walking speed and the subsequent 10 minutes of data were used for analysis.

A step function of zeroes and ones, denoting heel contact and heel off respectively, was then generated from each of the voltage signals. Stride intervals were isolated based on an automatic stride interval extraction algorithm adapted from Chau and Rizvi (2002). Briefly, this involved identifying candidate stride times (i.e., all changes in the step function from 1 to 0) and then selecting the most probable event times based on a mean stride estimate. The mean stride estimate is taken as the mean stride interval from a subset of stride intervals, after having eliminated the outliers that may otherwise skew the mean calculation.

From the set of probabilistic stride intervals extracted, we subsequently eliminated the

strides that fell outside 0.01 and 99.99% of a gamma distribution fit, considering these strides as unphysiologically long or short. Ultimately, the number of stride intervals comprising each time series ranged between 446 and 706, depending on the cadence of the participant.

Stationarity Testing

Given that results of the reverse arrangements test are sensitive to window size, M , we tested the stationarity at window sizes ranging from 10 to 45 strides, in 5 stride increments. The minimum window size was constrained to include at least 10 stride intervals and the maximum window size was constrained to include at least 10 windows (for all series but one). In this way, we maintained an adequate number of data points as required to estimate a single statistical parameter (Chau et al., 2005) when calculating both the mean squared value within each interval and the total number of reverse arrangements.

In general, time series lengths were not exact multiples of the chosen window sizes. Therefore, strides that were not included within the intervals for analysis were equally omitted from both ends of the signal. After selecting an appropriate window size for further analysis, we subsequently examined the effect of our choice of trimming location by comparing results when trimming the outstanding stride intervals from the beginning, the end and equally from both ends, of the signals.

To determine which, if either, of the first two moments could be identified as possible contributors to the identified non-stationary signal, we divided the signal into non-overlapping intervals of the chosen length, and computed the mean and variance of the time series within each of these windows. The null hypothesis of time invariance was then tested with regression analysis. Finally, time series for which time-varying means and/or variances were identified were further classified as either having an increasing or decreasing trend. In this case, rejection of the null hypothesis due to a significantly non-zero positive or negative slope indicated the presence of an increasing or decreasing

trend, respectively.

Unless indicated, all tests were performed at a 5% level of significance and left and right foot data were considered separately.

3.4 Results

3.4.1 Effect of Window Size

A total of 186 stride interval time series (31 from each condition for both the right and left feet) were acquired for analysis of stationarity using the reverse arrangements test. Results showing the percentage of non-stationary time series identified at each window size, during the three walking conditions under study, are presented in Fig. 3.1. In general, the least number of non-stationary signals were identified during STW while the most non-stationary signals were identified during UTW.

Figure 3.1: The effect of window size on the percentage of non-stationary time series identified for each walking condition. The first and second bars of each pair depict results of the right and left foot, respectively. OW = overground walking; UTW = unsupported treadmill walking; STW = supported treadmill walking.

A representative boxplot showing the distribution of test statistic values for all right foot time series emerging from UTW trials is presented in Fig. 3.2. The median of the test statistic values clearly fell within the stationary range (i.e., $|z_\alpha| < 1.96$) and were essentially the same for the given window sizes. The same result held for time series emerging from treadmill walking conditions and for left foot data. A Kruskal-Wallis test found that the mean ranks of stationary test statistics were not significantly different among window sizes ($p > 0.96$) in all cases (i.e., for left and right foot data and all walking conditions). Thus, the test statistic alone does not suggest a window size preference for further analysis.

Figure 3.2: The effect of window size on the stationarity test statistic, z_α , for right foot data generated during unsupported treadmill walking. Horizontal dashed lines define the boundary between stationarity and non-stationarity, i.e., $|z_\alpha| < 1.96$.

As evident in Fig. 3.1, the percentage of non-stationary time series identified in

each walking condition tends to decrease with increasing window size. Since at larger window sizes a given time series is divided into fewer segments, a shorter mean square sequence is generated. With a shorter sequence, fewer comparisons between subsequent mean square values can be made, reducing the number of opportunities for detection of a reverse arrangement. This constraint also reduces the variation in the number of reverse arrangements likely to be identified between signals, as is apparent from the progressive shortening of the boxplot whiskers seen in Fig. 3.2. The reduction in the number of non-stationary signals detected at larger window sizes also agrees with intuition, since even a slow varying trend would begin to appear stationary when observed at sufficiently large window sizes. Thus, the reverse arrangements test is least reliable at the largest window sizes, where non-stationarities due to fast varying trends may be masked. Considering the tradeoff between maintaining an adequate number of stride intervals within each window for calculation of each mean square value and generating a sufficiently long mean square sequence for identification of non-stationarities, we chose a mid-range window size of 25 stride intervals for all subsequent analysis. At this window size, a total of 74 signals (36 - right foot, 38 - left foot) were identified as non-stationary.

3.4.2 Effect of Trimming Location

A Kruskal-Wallis test found that stationarity test statistics were not significantly influenced by the choice of trimming location for either foot and all walking conditions ($p > 0.74$). Therefore, our arbitrary choice to trim from both ends of the time series is of no consequence.

3.4.3 Sources of Non-stationarity

As summarized in Fig. 3.3, the majority of identified non-stationarities could be attributed to a change in the mean stride interval over time. To this end, very few signals demonstrated time-dependant variance alone, though some had both non-stationary

means and variances. In particular, it was found that for both treadmill walking modalities, 75% of unstable means were due to an increasing mean trend, indicating that stride intervals increased over time. On the other hand, during overground walking just over half (55%) of the unstable means tested positive for an increasing mean trend.

Figure 3.3: Sources contributing to non-stationarity of the time series, as a percentage of the non-stationary signals identified within each particular walking condition. Data presented is for the right foot. OW = overground walking; UTW = unsupported treadmill walking; STW = supported treadmill walking.

3.5 Discussion

3.5.1 A Locomotor Control Perspective

The identification of non-stationary signals within each walking condition is somewhat indicative of the complexities underlying locomotor control. Even during self-paced overground gait, time-varying changes in the stride interval time series occur within a 10-minute period and are largely due to a time-dependent mean. It is possible that these changes in time occurred due to fatigue, boredom, loss of concentration, conscious modulation of gait to regain interest in the task (i.e., to entertain oneself), or in anticipation of task completion (e.g., Wall and Charteris, 1980; Hausdorff et al., 1999). On the other hand, a certain amount of non-stationarity may be inherent to the locomotor control system; a reflection of its effort to integrate complex information coming from proprioceptive, vestibular and visual sensors (Dietz, 2002). Some support to this second possibility is provided through the identification of non-stationary signals during treadmill walking. This locomotor modality is characterized by a constant-speed constraint and static visual feedback, two external cues that are known to influence human locomotor control (Orendurff et al., 2004; Hirasaki et al., 1999; Jordan et al., 2007; Prokop et al., 1997; Warren et al., 2001). Intuitively, one might expect the constant speed constraint of treadmill walking to induce a more stable stride interval time series as compared to unconstrained overground ambulation. On the contrary, our results suggest that treadmill walking without handrail support induces at least as many, if not more, non-stationarities. However, during the more constrained treadmill walking task in which side-handrail support was implemented, far fewer non-stationary time series were identified.

A number of changes to the afferent systems involved in treadmill walking may have contributed to the increased number of non-stationary signals identified during UTW. From a behavioural perspective, subjects were largely naive to treadmill walking, conceivably resulting in an initially cautious gait. Although given the opportunity to become

accustomed to this locomotor modality at the outset of the study, we expect that some unobvious and unreported anxiety may have remained. This would likely have diminished over the 10-minute trial, contributing to a time-dependent change in a subject's stride interval as he or she became more familiar and confident with treadmill walking. A number of studies comparing overground and treadmill walking have also identified biomechanical differences in gait (Alton et al., 1998; Stolze et al., 1997; Nigg et al., 1995; Murray et al., 1985; Van Ingen Schenau, 1980; Riley et al., 2007; Savelberg et al., 1998; Dingwell et al., 2001; Matsas et al., 2000). These are largely explained in terms of mechanical differences in the walking surface, the loss of optic flow feedback, the constant-speed constraint, and behavioural adjustments to a less familiar task. It has been found that as the locomotor system adapts to these differences, an initial period of treadmill familiarization is required before repeated measurements of certain gait parameters cease to change significantly within a walking session (Wall and Charteris, 1980; Lavcanska et al., 2005; Van de Putte, 2006; White et al., 2002; Matsas et al., 2000). Of particular relevance to our results, both Wall and Charteris (1980) and Matsas et al. (2000) found that stride time increases with treadmill habituation. In line with this, of the 12 non-stationary signals identified as having a time-dependent mean during UTW (Fig. 3.3), 75% could be attributed to an increasing trend in the mean, suggesting that these children were still habituating to this locomotor modality.

The reduction in the number of non-stationary signals identified during STW likely occurred due to the additional proprioceptive information available for regulation of locomotor control. In line with this result, Dickstein and Laufer (2004) found that additional somatosensory input provided by light fingertip touch during treadmill walking facilitates spatial orientation and reduces body sway. From a purely anthropometric perspective, by grasping the handrails, the reasonable range for each child's foot placement (and hence stride length) is effectively reduced, restricted by the extent of his or her arm reach. This limitation on stride length imposes an additional constraint on the user's stride interval

on top of the constant speed constraint of the treadmill. Seemingly, the handrails act to somewhat anchor somatosensory feedback, perhaps contributing to fewer non-stationary signals during supported treadmill walking.

We also consider the influence of gait maturity on our results. Since the various afferent systems contributing to regulation of locomotor control may not have reached full maturity in children, the capacity of these systems to efficiently generate stable movement patterns may be reduced (Stolze et al., 1997). This idea may also have contributed to the identification of more non-stationary signals during UTW than during OW in our paediatric study. Conversely, during STW, the additional locomotor constraints discussed above may have sufficiently augmented locomotor regulation so as to overcome the suggested age-sensitivity to treadmill walking. To this end, it would be interesting to assess whether or not the same trend, suggesting an increased number of non-stationary signals during UTW, would also appear in an investigation of stride interval stationarity in an adult population.

3.5.2 Relevance to Analysis of Stride Interval Dynamics

This investigation reveals that the stride interval time series emerging from paediatric gait under varying degrees of locomotor constraint is often, though not always, non-stationary. While the stride interval time series of 11 children (age range 5-9 years old; 5 males) were found to be weakly stationary for all walking trials, other children produced non-stationary signals under at least one walking condition and two children (ages 5 and 8 years; both male) produced non-stationary series for all three walks. Finally, considering the possibility of maturation effects within our sample population, we also note that non-stationary signals were identified in at least one walking condition across the entire age spectrum under analysis. Thus, we have confirmed, at least for a paediatric population, the often assumed notion that stride interval time series exhibit non-stationary behaviour in many cases. Given this finding, when estimating the fractal behaviour of gait, we

emphasize the importance of implementing scaling analysis techniques that are robust to non-stationarities.

DFA is one method that was developed to account for the non-stationary behaviour of series generated from certain DNA sequences (Peng et al., 1994). This method, often applied to stride interval time series and other physiological processes, has since been tested with simulated fGn and fBm data, alongside numerous alternative techniques, with findings depending largely on the nature and length of the series under analysis (Eke et al., 2000; Delignieres et al., 2006). While other methods seem to produce more accurate estimates when dealing strictly with fGn or fBm processes, DFA or a modified power spectral density approach typically provide more robust estimates for series near the $1/f$ boundary (Delignieres et al., 2006). Considering that the scaling behaviour of human stride interval time series is generally considered to fall within this range (Hausdorff et al., 1995), and given the identification of both stationary and non-stationary signals in this investigation, DFA and modified power spectral density would seem to hold the most promise for estimation of statistical persistence within stride interval data.

Nonetheless, both methods still present considerable challenges to gait researchers. While Delignieres et al. (2006) fittingly suggest that the low variability associated with the modified power spectral density method render it most appropriate for comparisons between mean scaling exponents, this variability is significantly increased for series containing less than 512 data points. Within the scientific and rehabilitation communities where stride interval quantification is seemingly of interest, it is not uncommon for the population under study to have gait difficulty, rendering acquisition of a sufficiently long time series frequently impossible. When considering DFA analysis, seemingly most appropriate when the goal is not to compare but rather to quantify the persistence of a series sample (due to its low bias), there are other issues to consider. For example, depending on the signal's underlying correlation properties, certain non-stationarities are still known to influence the scaling estimate (Hu et al., 2001; Chen et al., 2002). In addi-

tion, use of DFA requires that the investigator choose a box size fitting range to be used in the analysis; a choice that can have a significant influence on the scaling estimate. These issues are often handled differently by researchers, complicating the interpretation of results and the ability to draw comparative conclusions across studies (Hausdorff, 2007) and highlighting the need for a widely-acceptable and standardized approach for use in the gait literature.

It has been suggested that an integrated approach be adopted for scaling analysis, in which multiple methods be consistently implemented (Rangarajan and Ding, 2000). This approach would not only facilitate comparison between studies, but may also improve with-in study estimates. Where the results of one method may otherwise lead to false conclusions, inconsistencies revealed by another technique would enable one to more accurately determine the true scaling behaviour of a given time series (Rangarajan and Ding, 2000). Of course, such an approach is only of use if the scaling methods are appropriately chosen based on the underlying signal properties, and only then if the associated user-selected parameters are consistently implemented. For example, when evaluating the statistical persistence of stride-interval fluctuations, the pair of estimators most commonly compared are those derived from DFA and spectral analysis (Hausdorff et al., 1995, 1996). Given that spectral analysis is sensitive to non-stationarities, we question the utility of making such a comparison and suggest instead the use of alternative techniques that, like DFA, are less affected by non-stationary behaviour.

3.6 Conclusion

In order to better understand the human locomotor control system, it is important to carefully characterize gait dynamics. In particular, to determine the appropriateness of various scaling analysis techniques for quantification of the correlation properties of stride interval time series, it is important to know the extent to which the signal of

interest is stationary. This study assesses for the first time the extent to which stride interval time series, obtained from a paediatric population, are weakly stationary. We reveal non-stationary signals in all walking conditions, including the most constrained locomotor modality in which children walked on a treadmill with handrail support. We have thus confirmed that, as is true for many physiological signals, paediatric stride interval time series are often non-stationary. Therefore, when seeking to quantify the statistical persistence of stride interval fluctuations, scaling analysis techniques that are less affected by non-stationarities should be implemented. To this end, much has been done as of late to facilitate the selection and implementation of these techniques when dealing with physiological data. However, this body of work has largely focused on simulated data and other physiological processes, with little empirical investigation of stride interval time series specifically. Evidently, if scaling analysis of stride interval time series is to gain true clinical value, a methodological effort to standardize approaches is needed.

Chapter 4

The Effect of Treadmill Walking on the Stride Interval Dynamics of Children

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This chapter investigates the impact of treadmill walking on paediatric stride characteristics, namely stride interval persistence, to address the second objective of this thesis. Based on the results of stationarity testing presented in Chapter 3, and to facilitate comparison with literature findings, detrended fluctuation analysis is implemented for quantification of persistence.

Note: Section 4.2 contains some repeated background information and the stride interval extraction procedure in section 4.3.2 is reiterated from section 3.3.3.

4.1 Abstract

Treadmills are commonly implemented in rehabilitation and laboratory settings to facilitate gait analysis and training. However, while this locomotor modality is often used with children, its effect on paediatric stride interval dynamics is unknown. This study investigated the stride interval persistence of 30 asymptomatic children after completion of three to six 10-minute walking trials comprised of: (i) overground walking (OW), (ii) unsupported treadmill walking (UTW), and (iii) handrail supported treadmill walking (STW). The primary outcome measure was mean scaling exponent, α , a quantifier of stride interval persistence obtained from detrended fluctuation analysis. Mean preferred walking speed, number of strides taken, stride interval duration and stride interval coefficient of variation were also assessed. Compared to OW, stride interval persistence was not significantly different during STW but was significantly lower during UTW, largely due to the younger children. Preferred speed, number of strides and stride interval duration differed between overground and treadmill walking, and older children demonstrated reduced stride interval variability compared to younger children. The observed treadmill and age effects on stride parameters may be due to a combination of differing locomotor constraints between overground and treadmill walking and developmental differences in sensory processing, cerebellar plasticity, and corticospinal involvement in locomotion.

4.2 Introduction

Close examination of the human stride interval, the time between consecutive heel strikes of the same foot, reveals complex fluctuations that are correlated over hundreds of strides (Delignieres and Torre, 2009; Hausdorff et al., 1995). However, the mechanisms responsible for this stride interval persistence remain unclear. Current locomotor theories suggest that the most basic moving rhythms are achieved by specialized neural networks in the spinal cord, with higher neural centers playing an important role in locomotor control as

well (Dietz, 2003; Forssberg, 1999).

Investigation of stride interval persistence among different populations has revealed diminished correlation with advanced ageing and in the presence of certain neuromuscular pathologies (Hausdorff et al., 1997, 2000; Chau and Rizvi, 2002). Within a cohort of individuals with higher-level gait disorder, these dynamics provided the only distinguishing feature among fallers and non-fallers (Herman et al., 2005). Furthermore, when walking is paced to a metronome, stride interval dynamics demonstrate anti-persistent fluctuations as opposed to the persistent behaviour typical of regular overground walking (Delignieres and Torre, 2009). Given that stride dynamics have shown these sensitivities to locomotor maturity, disfunction and constraint, quantification of stride interval persistence seemingly provides important insight into the central mechanisms driving human locomotor control.

The finding of altered stride interval dynamics during metronomic walking (i.e., walking at a step frequency dictated by the constant beat of a metronome) raises some concern within rehabilitation and scientific communities, where treadmills are commonly implemented to promote restoration of gait function (Hesse, 2008; Kurtais et al., 2008; Damiano and DeJong, 2009) and to facilitate gait analysis and training (Siler et al., 1997; Marsh et al., 2006). Treadmill walking can be somewhat likened to metronomic walking in that it imposes a similar constant-tempo constraint. It is therefore imperative to study the effect of treadmill walking on human stride interval dynamics.

Recently, Chang et al. (2009b) suggested that the afferent influences of treadmill walking on the locomotor control system are distinctly different from those of metronomic walking, after treadmill walking was not found to significantly alter the stride interval persistence of healthy young adults. Nonetheless, it would be naive to assume that the same is true for children. To date, a single study has quantified the stride interval persistence of a paediatric population, identifying greater complexity in their stride-to-stride fluctuations when compared to those of adults (Hausdorff et al., 1999). It is suggested

that these elevated dynamics reflect a less mature locomotor control system (Hausdorff et al., 1999). Indeed, there is considerable evidence to suggest that neuromaturation and locomotor development continues well beyond the age of three (Beck et al., 1981; Assaiante, 1998; Forssberg, 1999; Barnea-Goraly et al., 2005; Dusing and Thorpe, 2007), at which time a seemingly coherent locomotor pattern is produced (Hausdorff et al., 1999). As such, while the stride interval dynamics of adults appear robust to treadmill walking, it is conceivable that paediatric stride dynamics may be more susceptible to the external influences presented by this locomotor modality.

From a rehabilitation perspective, a treatment regime that fails to promote (or at the very least preserve) a certain extent of stride interval complexity, would seem to oppose the natural neuromuscular rhythms of healthy human gait. Given that many emerging rehabilitation techniques make use of treadmills to train children (Cherng et al., 2007; Dodd and Foley, 2007; Damiano and DeJong, 2009), our primary research objective was to investigate the implication of treadmill walking, both with and without handrail support, on the natural stride interval dynamics of children. We hypothesized that locomotor constraints including the constant average speed requirement (Dingwell et al., 2001) and static visual flow (Prokop et al., 1997) imposed by treadmill walking, would diminish paediatric stride interval dynamics. We further expected that implementation of handrail support would enhance sensory feedback in some ways by compensating for the reduced visual cues (Dickstein and Laufer, 2004). Therefore, we expected the handrail-supported treadmill walking condition to restore stride interval dynamics toward values measured during overground walking.

4.3 Methodology

4.3.1 Data Acquisition

Subjects

A total of 31 children were recruited through the staff and community programs at Bloorview Kids Rehab (located in Toronto, Ontario, Canada). Data from one subject was discarded due to technical difficulties with gas exchange measurement, which is part of a secondary investigation reported elsewhere (Fairley et al., 2009b). Of the remaining 30 children (11 male) forming the sample for this analysis, mean age was 7.1 ± 1.6 years (range: 4-10 years), height was 1.249 ± 0.115 m and body mass was 24.1 ± 4.9 kg. Participants were asymptomatic, with normal or corrected-to-normal vision and no history of orthopedic, neurological, respiratory, or cardiovascular illness. The study was approved by the institutional research ethics board and all subjects and their caregivers provided informed, written assent and consent, respectively.

Experimental Protocol

Each subject participated in one study session, completing a minimum of three, 10-minute walking trials, under the following conditions: (i) overground walking (OW), (ii) unsupported (hands-free) treadmill walking (UTW), and (iii) supported treadmill walking (STW). Condition sequences were pseudo-randomized, ensuring that each of the six possible sequences was completed every six participants. A random subset of subjects repeated at least one of the walking trials (immediately following the same study session), while wearing only a portion of the measurement equipment. These repeat trials were carried out to determine if the additional metabolic equipment donned by subjects had an effect on the primary outcome measure (stride interval persistence). The three initial walking trials completed by all 30 subjects are herein referred to as the primary walking trials, while additional trials are referred to as the repeat walking trials.

At the outset of a study session, the subject was introduced to the protocol and outfitted with the measurement equipment. Pre-exercise resting heart rate was then recorded at one minute intervals, with the subject resting in a seated position, long enough to obtain steady-state readings for at least three minutes. Steady-state heart rates were identified when consecutive (1 min. apart) readings differed by less than five beats per minute (Siconolfi et al., 1982).

After this initial rest period, the subject was given at least five minutes to become familiar with treadmill walking and the measurement equipment. Handrail height was adjusted such that the base of the cylindrical side-rails was located at the level of the subject's radial styloid process, while he or she stood on the treadmill with arms relaxed at his or her sides. The subject walked until the investigator visually identified gait to appear natural and until the subject reported feeling comfortable with the setup. Following the equipment familiarization period and between each walking trial, to mitigate the effects of fatigue, subjects rested for at least seven minutes and long enough to ensure that heart rate had returned to within the pre-exercise resting range.

Prior to handrail-supported treadmill walking trials, subject's were instructed to "keep [his or her] hands on the rails for the entire 10 minutes of walking". If the subject lost hand contact with either rail during the trial, he or she was reminded to "please put [his or her] hand(s) back on the rail(s)". For all trials, subjects were instructed to walk at their comfortable speed as if "walking to school" or "going for a walk in the park". Comfortable treadmill speeds were determined immediately prior to each treadmill walking trial. Initially, the subject walked on the treadmill (GK200T, Mobility Research, USA) at a relatively slow speed. The speed was then increased in 0.1 mph increments until the subject reported that his or her comfortable speed had been reached. Subsequently, the speed was increased by at least 0.5 mph and to the extent that the subject was pressed to maintain a walk. The speed was then decreased in 0.1 mph increments until the subject once again reported that his or her comfortable walking speed had been reached. This

procedure was repeated and the mean of the four reported walking speeds was taken as the comfortable speed for that particular treadmill walking trial. For overground walking trials, subjects completed a single lap warm-up immediately prior to the trial, to ensure that he or she was familiar with the route and thus facilitate maintenance of a comfortable pace. The overground route consisted of a rectangular circuit (total length = 84.4 m), through a level, linoleum-floored hallway (width = 2.43 m). Lap times were recorded using a stopwatch to allow for subsequent calculation of the subject's comfortable overground walking speed. This speed was taken as the average of completed lap speeds, where lap speed was estimated by dividing lap distance by the time required for lap completion.

Measurement Equipment

Subjects wore an adjustable waist harness which, during treadmill walking trials only, was loosely attached to an overhead bar via safety belts (LGJr200, Mobility Research, USA). The belts did not impede arm swing and subjects remained fully weight-bearing throughout the trials.

During all walking trials, heel strike was recorded bilaterally via two force-sensitive resistors (Model 406, Interlink Electronics, USA). These paper-thin sensors were fastened, one each, to the sole of the subject's shoes underneath the heel and produced a change in voltage upon heel strike. This voltage signal was sampled at 250 Hz and recorded to a data acquisition card (CF-6004, National Instruments, USA), housed in a personal digital assistant (Axim x51v, Dell, USA) that was secured via the waist harness to the subject's abdomen.

A heart rate transmitter (WearLink 31, Polar Electro, Finland) was also worn by subjects throughout the study session and a portable system for pulmonary gas exchange measurement (K4b², Cosmed, Italy) was donned during the three primary walking trials only. The K4b² system included a face mask, flow meter, data collection unit, battery,

and associated cables. The battery and data collection unit were secured to the waist harness on the subject's back.

To monitor handrail contact during supported treadmill walking trials, each treadmill handrail was instrumented with 14 force-sensitive resistors (Model 408, Interlink Electronics, USA). Sensors were oriented lengthwise and fastened around the cylindrical side-handrails in two layers of seven such that the active area of sensors did not overlap. Handrails were slightly tapered, with circumferential dead-space reaching a maximum of approximately 2.8 cm at the end of the rail (toward the back of the treadmill) and a minimum of 2.1 cm toward the front. Contact with each force-sensitive resistor was reflected as a change in voltage. These 28 voltage signals were simultaneously sampled at 1 kHz, recorded to a data acquisition card (USB-6210, National Instruments, USA) and stored in a personal computer for subsequent analysis. Measurement of handrail and heel strike contact was commenced simultaneously and recorded in parallel throughout the duration of all treadmill trials.

The total mass of the equipment worn by subjects was 2.5 kg during the primary walking trials and 1.4 kg during repeat walking trials (without the K4b² system). The experimental setup, showing the equipment worn by subjects during each primary walking trial, is depicted in Fig. 4.1.

4.3.2 Data Analysis

Stride Interval Extraction

Prior to the extraction of stride intervals from heel strike data, gait signals were manually trimmed in order to remove the extraneous portions of the recordings, obtained immediately before the initiation and after the termination of gait. The first 10 seconds of data were then removed in order to ensure the subject had finished accelerating from rest and the subsequent 10 minutes of walking data were used in the analysis. For stride

Figure 4.1: Subject completing a primary supported treadmill walking trial while wearing study equipment.

extraction, the signal was converted into a step function of ones and zeroes denoting possible heel-off and heel contact events, respectively.

Stride intervals were extracted based on a stride interval extraction algorithm adapted from (Chau and Rizvi, 2002). Depending on the cadence of the subject, between 446 and 706 strides comprised each walking trial. In total, 180 stride interval time series were obtained from the three primary trials (30 participants $\times$ 3 conditions $\times$ 2 sides) and 58 were obtained from the repeat trials (12 from OW, 7 from UTW and 10 for STW for both the right and left foot).

Quantification of Stride Interval Persistence

To estimate the statistical persistence of the stride interval time series, detrended fluctuation analysis (DFA) was implemented as described elsewhere (Peng et al., 1995). This method produces a scaling estimate, α , for each signal, which is interpreted as follows: $0.5 < \alpha \leq 1.0$ indicates that the signal is statistically persistent; $\alpha < 0.5$ means the signal is anti-persistent; $\alpha = 0.5$ indicates random, uncorrelated behaviour; and $\alpha > 1.0$ indicates the presence of Brownian noise (Hausdorff et al., 1995). Given that the results of DFA are sensitive to the range of box sizes used, we carried out the analysis proposed by Damouras et al. (2009) for determination of the box size range producing the most stable α -estimates. We began the procedure with a wide box size range of $[n_{min} = 4, n_{max} = 128]$, where the upper box size was modified to include at least two boxes in the shortest stride interval time series (446 strides in length). Then, considering the 90 identified left and right foot ranges separately, we calculated a single range containing 95% of the individual ranges to use in the DFA analysis. Ultimately, the optimal box size fitting ranges identified for right and left foot series were $[n_{min} = 13, n_{max} = 64]$ and $[n_{min} = 16, n_{max} = 60]$, respectively.

Handrail Contact

To ensure subjects complied with the STW protocol, the duration of handrail contact was quantified using the voltage signals acquired from sensor handrails during both treadmill trials, trimmed to correspond with the same 10-minute interval in which stride interval persistence was assessed. Since voltage decreased with increasing force on a sensor, data points recorded during STW that fell below a threshold value were taken to indicate that contact with that respective sensor was in effect. For each sensor, a separate baseline threshold was defined, taken as the lower 95% confidence limit of voltages recorded by the corresponding sensor during the UTW trial (where there was no handrail contact). Handrail contact was computed as the percentage of time, during each 10-minute trial,

that contact was made with at least one sensor on the right, the left, or both handrail(s).

Statistical Analysis

In all cases, right and left foot data were considered separately and $p = 0.05$ was used to determine significance unless otherwise indicated. In addition, all data were first tested for normality using the Chi-squared goodness-of-fit test (Bendat and Piersol, 2000).

Effect of Additional Measurement Equipment. Considering the additional weight and potential unfamiliarity imposed by the subject-worn metabolic equipment, we tested for any effects of this equipment on our primary outcome measure, α . For each walking condition, paired t-tests were implemented to compare mean α -values generated during primary and repeat trials using only data from the subset of subjects that performed both tasks.

Effect of Age. Given the age range of our sample, we also tested for possible maturation effects. Based on the findings for α reported by Hausdorff et al. (1999), we divided our sample into two groups: younger, 4- to 7-yr-old children ($n = 18$); and older, 8- to 10-yr-old children ($n = 12$). For each gait parameter (stride interval persistence, preferred speed, number of strides, stride interval and stride interval coefficient of variation), unpaired t-tests or Wilcoxon rank sum tests (the non-parametric equivalent) were employed, as appropriate, to compare results obtained for younger and older age groups during each respective primary walking condition.

Effect of Walking Condition. To assess the effect of walking condition on each gait parameter, we compared mean values obtained from the three primary walking trials, first for the entire cohort and then separately for younger and older age groups. Based on results of normality testing, either one-way repeated measures ANOVAs or Friedman's tests (the non-parametric equivalent) were employed. If significant differences were found, paired t-tests or Wilcoxon signed rank tests (the non-parametric equivalent) were performed, with a Bonferroni adjusted significance level of $p = 0.0167$, to compare

two groups at a time.

4.4 Results

For each walking condition, no significant differences between right and left foot mean α -values ($p > 0.08$) were detected by paired t-tests. Furthermore, for all gait parameters, statistically similar results were obtained for both right and left foot data. Hence, the results that follow are reported for right foot data only.

4.4.1 Stride Interval Persistence

Stride interval persistence of right foot time series generated during primary walking trials is depicted in Fig. 4.2. Group means and standard deviations of α are presented in the last row of Table 4.1.

Effect of Additional Measurement Equipment. Considering each walking condition separately, no significant differences were identified between the mean α -values calculated from the primary and repeat walking trials ($p > 0.2$). This result held true for both right and left foot data, and suggests that the additional mass and/or unfamiliarity imposed by the subject-worn metabolic equipment did not significantly alter the statistical persistence of stride interval time series.

Effect of Age. During overground walking, a significant difference was identified between the mean α -values of the younger (4- to 7-yr-old) and older (8- to 10-yr-old) children ($p = 0.035$). On the other hand, α -values generated from time series obtained during both treadmill walking conditions were similar among groups.

Effect of Walking Condition. Considering the agglomerate, 4- to 10-yr-old age group, a one-way repeated measures ANOVA identified a significant difference in mean α -values between walking conditions ($p = 0.002$). Subsequent pairwise comparisons revealed the mean stride interval persistence of UTW to be significantly lower than the mean value

Figure 4.2: Box plots of scaling estimates, α , for all (A), younger (B) and older (C) children. The data presented are for the right foot and were obtained during the three primary walking trials.

obtained during OW ($p < 0.001$). No other significant differences were detected. When considering the effect of walking condition on each age group separately, a significant difference between OW and UTW was only identified in the younger children ($p < 0.001$).

4.4.2 Other Gait Parameters

Mean preferred walking speed and a number of stride characteristics of the analyzed time series, calculated from primary walking trials, are presented in Table 4.1. While the preferred speed of subjects was faster overground than on the treadmill ($p < 0.001$), there were no significant differences in speed between the two treadmill walking conditions ($p > 0.028$). For all conditions, the average preferred speed of the younger children was slower than that of the older children ($p < 0.02$). More strides were taken during OW than during the two treadmill conditions ($p < 0.001$) which is not surprising given the

faster overground pace. However, significantly more strides were also observed during UTW in comparison to STW ($p < 0.001$), suggesting that gait adjustments were made (e.g., shorter stride length, higher cadence, postural changes). At least some of the gait adjustments can be attributed to changes in stride interval duration (and hence stride lengths), which were significantly different across all three walking conditions ($p < 0.001$). Compared to the younger children, the stride interval of the older children was significantly longer during UTW only ($p = 0.047$). Finally, while no differences in stride interval coefficient of variation (CV) were detected among the three conditions ($p > 0.3$), the older children had a significantly lower CV than the younger children during OW and UTW ($p < 0.028$).

4.4.3 Handrail Contact

Overall, we found children to be very compliant with the handrail supported treadmill walking task. Children maintained contact with at least one rail for no less than 97.5% of the trial duration, and had simultaneous contact with both rails for at least 82.3% of the trial. Given that α was slightly more variable during handrail STW, albeit not significantly so, we also tested for a possible correlation between stride interval persistence (α) and handrail contact durations (right, left and both). No significant correlations were identified ($p > 0.06$).

4.5 Discussion

4.5.1 Effect of Unsupported Treadmill Walking on Persistence

This study demonstrates that comfortably-paced treadmill walking, without handrail support, significantly alters a child's natural stride interval dynamics. In particular, stride-to-stride fluctuations become less structured during UTW compared to OW. A number of differences between these two locomotor modalities may have contributed to

Table 4.1: Preferred walking speed and characteristics of right foot stride interval time series obtained from the three primary walking trials: overground walking (OW), unsupported treadmill walking (UTW) and supported treadmill walking (STW).

Parameter	Age Group	Condition		
		OW	UTW	STW
Preferred Speed ^a , m/s	All	1.13 ± 0.20	$0.73 \pm 0.14^{\dagger}$	$0.76 \pm 0.12^{\dagger}$
	Younger	1.05 ± 0.19	$0.66 \pm 0.12^{\dagger}$	$0.72 \pm 0.10^{\dagger}$
	Older	$1.24 \pm 0.16^{\ddagger}$	$0.83 \pm 0.11^{\dagger\ddagger}$	$0.82 \pm 0.12^{\dagger\ddagger}$
Number of Strides ^a	All	596 ± 51	$538 \pm 39^{\S}$	$494 \pm 38^{\dagger}$
	Younger	603 ± 55	$548 \pm 36^{\S}$	$499 \pm 37^{\dagger}$
	Older	585 ± 45	$522 \pm 38^{\S}$	$486 \pm 40^{\dagger}$
Stride Interval ^b , s	All	1.00 ± 0.02	$1.11 \pm 0.02^{\S}$	$1.21 \pm 0.02^{\dagger}$
	Younger	0.99 ± 0.02	$1.08 \pm 0.02^{\S}$	$1.19 \pm 0.02^{\dagger}$
	Older	1.02 ± 0.02	$1.14 \pm 0.02^{\S\ddagger}$	$1.23 \pm 0.03^{\dagger}$
Stride Interval Coefficient of Variation ^b , %	All	6.20 ± 0.45	5.81 ± 0.40	6.00 ± 0.61
	Younger	7.24 ± 0.59	6.67 ± 0.56	7.08 ± 0.93
	Older	$4.65 \pm 0.42^{\ddagger}$	$4.53 \pm 0.22^{\ddagger}$	$4.37 \pm 0.28^{\ddagger}$
Scaling Estimate ^a , α	All	0.86 ± 0.12	$0.73 \pm 0.11^{\dagger}$	0.79 ± 0.16
	Younger	0.90 ± 0.08	$0.72 \pm 0.12^{\dagger}$	0.80 ± 0.16
	Older	$0.81 \pm 0.15^{\ddagger}$	0.76 ± 0.09	0.78 ± 0.17

^a Values are mean $\pm$ SD.

^b Values are mean $\pm$ SE.

[†] Significantly different than OW ($p < 0.001$).

[§] Significantly different than STW ($p < 0.001$).

[‡] Significantly different than younger children ($p < 0.05$).

this change. The requirement for maintenance of a constant average speed within the spatial limitations imposed by the treadmill belt necessarily restricted gait adjustments to be more instantaneous in nature. Logically, with stride intervals changing on a more moment-to-moment basis as opposed to more gradually over the long-term, it seems reasonable that α would decrease, with less dependence between stride intervals taken further apart in time. In support of this, Delignieres and Torre (2009) have identified an even greater change in stride interval behaviour, from persistent to anti-persistent, when walking is more tightly constrained; paced on a stride-to-stride basis to the constant beat of a metronome.

The lack of optic flow feedback during treadmill walking may also have contributed to the reduction in paediatric stride interval dynamics during UTW (Stolze et al., 1997). Human locomotor control relies heavily on vision, and optic flow in particular, to control walking under varying environmental conditions (Warren et al., 2001). Furthermore, the ability to distinguish between and successfully implement visual cues is thought to improve with increasing age and walking experience, with greater automaticity of control gained with practice (Schmuckler and Gibson, 1989). To this end, while we found reduced α during UTW, largely due to the younger children studied, α was not affected in an adult population performing an analogous UTW task (Chang et al., 2009b). Therefore, unlike adults, children may not have been able to adapt to the static visual feedback during UTW due to their incomplete neuromaturation and more limited opportunity for locomotor practice in varying environments. Interestingly, in an obstacle avoidance study, Pryde et al. (1997) observe adult-like avoidance strategies beginning in 8-yr-old children, and suggest that visual-sensory motor control strategies are not yet mature in the younger, 5- to 7-yr-old children studied. This particular age-sensitivity exactly resonates with our results, lending support to the idea that stride interval persistence of the younger, 4- to 7-yr-old children in particular may have been more affected during treadmill walking due to unfamiliar visual constraints.

4.5.2 Effect of Supported Treadmill Walking on Persistence

Compared to OW, the stride interval persistence in our paediatric population was not significantly different during STW. Contrarily, adults have demonstrated significantly elevated dynamics during STW (Chang et al., 2009b). Children and adults are thought to perceive sensory information differently, or use the available information to a different extent (Berard and Vallis, 2006). To this end, postural stability studies suggest that, unlike adults, children are unable to precisely couple body sway with somatosensory information provided by fingertip touch (Barela et al., 2003). Thus, whereas adults may use handrail support as an additional somatosensory cue to increase stride interval persistence, children may yet to have developed this precise means of control.

On the other hand, the lack of statistical difference during STW in this paediatric study, compared to the adult study, may in part be due to the larger variability in α within walking conditions. Whereas the standard deviation in α for adults was 0.06, 0.08 and 0.10 for OW, UTW and STW respectively (Chang et al., 2009b), it was 0.12, 0.11 and 0.16 for these same conditions, respectively, in children. This observation suggests that paediatric variability in α may converge, toward adult values, with maturity.

4.5.3 Effect of Developmental Differences on Persistence

When comparing between the younger (4- to 7-yr-old) and older (8- to 10-yr-old) children, a significant difference in stride interval persistence was only identified during the most familiar, overground walking task. This would seem to suggest that treadmill walking acts in some respects to ‘level the playing field’ when it comes to stride interval dynamics. From a clinical perspective, this may indicate that the locomotor control systems of younger children have not sufficiently matured so as to be ‘dynamically’ ready for treadmill walking. Conversely, it is conceivable that treadmill walking could be used to accelerate the development of more robust dynamics.

A Physiological Perspective

Chang et al. (2009b) suggested that the cerebellar vermis may play a critical role in the maintenance of stride interval dynamics. To this end, age-dependent neuroplasticity of the cerebellum may in part account for the observed development differences in stride interval persistence. In a recent investigation of rhythmic tapping performance between early- and late-trained musicians, Watanabe et al. (2007) attributed greater improvement in early-trained children (< 7 yrs old) to a so called ‘sensitive’ period when stimulated neural systems, in particular the cerebellum, are more susceptible to change. Seemingly then, this cerebellar sensitivity could also be responsible for the observed treadmill-induced change in the stride interval dynamics of younger children.

On the other hand, the possibility that an entirely different control mechanism may be governing pediatric stride dynamics, perhaps in the lower spinal circuitry, should not be dismissed. This idea may explain the contradictory effect of treadmill walking on stride interval dynamics between children and adults, and also the difference in stride interval complexity previously identified between children and adults when walking over-ground (Hausdorff et al., 1999). While infant stepping is believed to occur largely under the control of spinal neurocircuitry, adults are thought to have stronger supraspinal connections which contribute to locomotor output (Lamb and Yang, 2000). Furthermore, there is evidence that refinement of corticospinal motor tracts continues into adolescence (Martin, 2005). Conceivably then, stride interval dynamics may be generated by lower spinal mechanisms early in life, with control of dynamics taken over, or greater influence gained, by higher neural centers later in development. This conjecture, paired with the argument that afferent information arising from treadmill walking is first relayed to spinal circuits (Chang et al., 2009b), suggests that treadmill walking does not impact the cortically-mediated stride interval persistence of adults but may alter the primarily spinal stride dynamics of children.

Irrespective of the neural circuitry responsible for stride dynamics, the age-sensitivity

of these complex patterns would certainly seem to suggest that either: (i) neural feedback being sent to control centers change with age (Berard and Vallis, 2006), or (ii) control centers themselves develop with age and thus differ in their ability to receive, integrate, or process the motor output signals (Martin, 2005; Watanabe et al., 2007). It is well established that that human nerve and muscle cells are postmitotic, and therefore present to their greatest extent after early development (Vandervoort, 2002). Subsequently, according to the neuronal group selection theory, a child’s neural circuitry adapts and develops with practice, retaining only the most favourable neuromotor networks (Forssberg, 1999). Thus, quantification of stride interval dynamics may provide insight into neural development and efficiency, reflecting the extent of experience-based selection that has occurred within the circuitry. Stride interval dynamics of older children and adults may be robust to subtle gait influences, such as those imposed by treadmill walking, when the locomotor activity sufficiently approximates the activity upon which experience-based selection of neuromotor networks took place (i.e., overground walking). On the other hand, when presented with a less-familiar task in which experience-based selection may be lacking (e.g. metronomically-paced walking), the selection process is revisited, and stride interval dynamics are altered. In this regard, our findings suggest that where a child has yet to develop advanced neural control, as reflected through diminished stride interval dynamics, sufficient practice may bridge the apparent developmental gap.

Cognitive Involvement

The different effect of treadmill walking on the stride interval dynamics of children and adults may be related to the extent of locomotor effort, perhaps cognitive load in particular, required for treadmill locomotion. When going from overground to treadmill walking, a larger change has been identified in children than in adults toward a more protective gait pattern (i.e., a broader step width and greater outward rotation of the foot) (Stolze et al., 1997). Seemingly then, a more cautious or ‘conscious’ gait may

lead to a decrease in α . To this end, when an adult population is asked to complete an inherently attentionally demanding task such as metronomically-paced walking, the expression of locomotor complexity is indeed altered (Delignieres and Torre, 2009). In addition, other populations that rely significantly on attentional processes to control gait, such as the elderly and individuals with Parkinson’s or Huntington’s disease (Yang et al., 2007; O’Shea et al., 2002; Woollacott and Shumway-Cook, 2002; Delval et al., 2008), also exhibit reduced stride interval persistence (Hausdorff et al., 1997).

4.5.4 Other Stride Parameters

Other gait parameters assessed in this study also showed sensitivity to walking condition. Considering the aggregate, 4- to 10-yr-old age group, children walked the fastest during the most familiar overground condition, taking the most strides and having a shorter stride interval. Compared to overground walking, the finding of reduced preferred speed during both treadmill walking conditions is consistent with findings in the literature (Wall and Charteris, 1980; Jeng et al., 1996; Chang et al., 2009b) and is likely most responsible for the corresponding decrease in stride number and increase in stride interval. In contrast, when average gait speed is maintained between overground and treadmill walking, the closely associated parameters of cadence and stride length typically change in the opposite direction, with an increased cadence and decreased stride length associated with treadmill walking (Stolze et al., 1997).

To this end, while speed was similar in this study during both treadmill walking tasks, fewer strides and a longer stride interval were identified in the handrail supported condition, suggesting that postural or temporal gait adjustments were made. Although it has been found that grasping the handrails does not alter the sagittal plane kinematics of treadmill walking in adults (Siler et al., 1997), it is unknown if the same is true for children. Furthermore, in a study pertaining to treadmill habituation it was found that stride interval increased with familiarization “as the confidence of the subject grew” (Wall

and Charteris, 1980). Thus, some locomotor differences may be attributable to changes in gait due to the level of perceived comfort with the walking task at hand. Considering that even light touch during locomotion provides enhanced spatial orientation and facilitates body stability (Dickstein and Laufer, 2004), subjects may have walked more confidently and comfortably during STW, with a corresponding decrease in stride number and stride interval.

Contrasting between age groups, we found that older children walked significantly faster during all three walking conditions, with an increase in stride interval duration that reached significance during UTW only. Not surprisingly given their larger stature, older children typically walk faster than younger children (Beck et al., 1981). These speed differences, and associated changes in stride cycle duration, are suggested to be primarily due to physical and not neuro-developmental maturation (Zijlstra et al., 1996). On the other hand, the younger children in our study also had significantly greater stride interval CV across all three conditions compared to the older children. This result is also consistent with other paediatric studies (Hausdorff et al., 1999; Zijlstra et al., 1996) and is suggested to be due to decreased walking speed, decreased postural stability at these slower speeds, and possibly other aspects of motor control development (Hausdorff et al., 1999; Zijlstra et al., 1996).

4.5.5 Study Limitations

Since the gait parameters of children are generally more variable than those of adults (Hausdorff et al., 1999; Stolze et al., 1997), a larger sample size and narrower subject age-ranges may be warranted in future studies to better elucidate the developmental profile of stride interval persistence. Furthermore, although different treadmill familiarization periods have been reported for different gait parameters (White et al., 2002; Lavcanska et al., 2005; Van de Putte, 2006; Wall and Charteris, 1980), we only implemented a single clinically relevant timeframe of approximately five minutes which may be insufficient for

the stabilization of stride dynamics. Future research should investigate the necessary familiarization period, both within-day and between-day, to obtain the most stable α -estimate.

4.6 Conclusions

In a group of 30 neurologically healthy children, we found stride interval persistence to diminish significantly from overground values when walking on a treadmill without handrail support. Preferred speed, number of strides and stride interval duration also differed between overground and treadmill conditions. Furthermore, older children exhibited lower stride interval variability and great consistency in stride dynamics across treadmill and overground conditions. The observed treadmill and age effects on stride characteristics may be due to a combination of different locomotor constraints between overground and treadmill walking, and developmental differences in sensory processing, corticospinal contributions to locomotor control and cerebellar plasticity between younger and older children. Future research with more children and finer age demarcations is required to further elucidate the developmental profile of stride interval dynamics.

Chapter 5

Investigating the Correlation between Paediatric Stride Interval Persistence and Gross Energy Expenditure

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This chapter probes the hypothesis of an association between stride interval persistence and the energy expended during locomotion. In fulfillment of the third objective of this thesis, the linear dependence between α and measures of gross energy expenditure obtained from a paediatric sample are explored.

Note: Section 5.2 contains some repeated background information and the methodological details of sections 5.3.1 and 5.3.2 were also discussed previously.

5.1 Abstract

Stride interval dynamics are thought to provide insight into neuromotor control, though their exact clinical meaning has not yet been realized. Since human locomotion is shaped by energy efficient movements, it has been hypothesized that stride dynamics and energy expenditure may be inherently tied, both having demonstrated similar sensitivities to age, disease, and pace-constrained walking. This study tested for correlation between stride interval persistence and gross measures of energy expenditure, including mass-specific gross energy consumption, mass-specific gross energy cost, and heart rate. Metabolic and stride interval data were collected from 30 asymptomatic children while performing three self-paced walking trials including overground walking, hands-free treadmill walking, and handrail-supported treadmill walking. No correlations between stride interval persistence and measures of gross energy expenditure were identified, suggesting that no simple linear dependence exists between these measures.

5.2 Introduction

The human stride interval (i.e., the time between consecutive heel strikes of the same foot) exhibits statistical persistence, with correlations extending over thousands of strides (Hausdorff et al., 1995). This persistence is typically quantified in terms of α , a scaling estimate provided by detrended fluctuation analysis (DFA) (Hausdorff et al., 1995; Hausdorff, 2007). To date, α has been found to change across the age spectrum, when certain neuromuscular disorders exist, and during some pace-constrained walking tasks (Hausdorff, 2007; Delignieres and Torre, 2009; Jordan et al., 2007). In light of these findings, it has been suggested that quantification of stride interval persistence may serve as a useful clinical tool for detection, classification, and monitoring of gait disorders (e.g., Hausdorff, 2007). However, the physiological underpinnings of stride interval dynamics remain unclear.

Interestingly, physiological cost has demonstrated similar sensitivities to age, disease, and tempo-constrained locomotion with increased energy requirements having been linked to advanced ageing (Malatesta et al., 2003), pathology (Waters and Mulroy, 1999), and forced-rate stepping (Zarrugh and Radcliffe, 1978). To this end, given that stride interval dynamics are thought to reflect a certain extent of locomotor control (Hausdorff, 2007), and that human locomotion is shaped by a preference for energy efficient movements (Patla and Sparrow, 2000), one might hypothesize that energy and scaling measures exhibit a certain degree of dependence.

To probe this hypothesis, we assessed the correlation between stride interval persistence and gross energy expenditure. Notably, the first study to investigate this dependence found no association in both adult and elderly populations (Malatesta et al., 2003). However, stride interval persistence was quantified using time series obtained from just six minutes of walking, a time-frame requiring at least three repeated measures in order to provide a reliable estimate of stride interval complexity (Pierrynowski et al., 2005). Distinct from Malatesta et al. (2003), we utilize longer (10-minute) time series to provide a more reliable quantification of stride interval persistence and, in addition to unsupported treadmill walking (UTW), we consider overground walking (OW) and handrail supported treadmill walking (STW). Furthermore, we assess the correlation between stride interval dynamics and measures of gross energy expenditure for the first time in a pediatric population.

5.3 Methodology

5.3.1 Experimental Protocol

The detailed experimental procedure is reported elsewhere (Fairley et al., 2009a). In brief, data were collected from 30 asymptomatic children (4-10 years old; 11 male) who completed three 10-minute walking trials under the following conditions: (i) overground

walking, (ii) unsupported (hands-free) treadmill walking, and (iii) side-handrail supported treadmill walking. Mean ($\pm$ SD) age, height, and body mass of subjects was 7.1 ± 1.6 years, 1.249 ± 0.115 m and 24.1 ± 4.9 kg, respectively. Subjects walked at their preferred speed, determined separately for each condition, and rested long enough before each trial to allow heart rate to return to pre-exercise resting values. All subjects abstained from eating or drinking anything other than water for at least two hours prior to data collection. This study was approved by the institutional research ethics board.

5.3.2 Measurement Equipment

Stride interval time series were acquired from heel-strike data, obtained from force sensors (Model 406, Interlink Electronics, USA) on the subjects' shoes. Energy expenditure was measured, breath-by-breath, via a portable system for pulmonary gas exchange measurement (K4b², Cosmed, Italy) that was carefully calibrated according to the manufacturer's instructions prior to each study session. Heart rate data were also obtained via a heart rate transmitter (WearLink 31, Polar Electro, Finland). Two portable data collection units, secured to the subject's front and back via a waist harness, were used to store data. The total mass of all subject-worn equipment was 2.5 kg. Additional equipment details are provided in (Fairley et al., 2009a).

5.3.3 Data Analysis

Stride interval extraction and quantification of stride interval persistence was carried out as described in (Fairley et al., 2009a). Ultimately, DFA provided a scaling estimate, α , for each of the 180 stride interval time series (30 participants $\times$ 3 conditions $\times$ 2 feet), to be used in subsequent correlation analysis with measures of energy expenditure.

Energy expenditure was assessed in terms mass-specific gross oxygen consumption ($\dot{V}O_2$; $\text{ml}\cdot\text{kg}^{-1}\cdot\text{min}^{-1}$), mass-specific gross oxygen cost (VO_2 ; $\text{ml}\cdot\text{kg}^{-1}\cdot\text{m}^{-1}$), and heart rate (HR; bpm). These parameters were each computed, as an average of the recorded

breath-by-breath values, over the last three minutes of each walking trial. With subjects having walked for seven minutes prior, this time frame is considered sufficient to ensure that the child reached a steady-state of exercise (McArdle et al., 1986). Data points for which the respiratory exchange ratio exceeded 0.9 were discarded. This ratio is indicative of anaerobic activity and should not be expected during comfortably paced walking (Waters and Mulroy, 1999). In total, 90 values were obtained for each energy measure (30 subjects x 3 conditions).

Prior to statistical analysis, all data were tested for normality using the Chi-squared goodness-of-fit test (Bendat and Piersol, 2000) to inform the choice between Pearson's (r) or Spearman's (ρ) correlation coefficients. Correlations were performed between right (and then left) foot α -values and each measure of energy expenditure, first considering data from each walking condition separately, and then considering all walking conditions together. Mean ($\pm$ SD) $\dot{V}O_2$, VO_2 , and HR were also calculated for each walking condition, and either one-way repeated measures ANOVAs or Friedman's tests (the non-parametric equivalent) were implemented, depending on results of normality testing, to compare among conditions. Where significant differences were identified, paired t-tests or Wilcoxon signed rank tests (the non-parametric equivalent) were used (with a Bonferroni adjusted significance level of $p = 0.0167$) to compare two walking conditions at a time. Except where indicated above, all tests were performed using $p = 0.05$ to determine significance.

5.4 Results

Considering each walking condition separately, right foot stride interval persistence (α) was not significantly correlated with $\dot{V}O_2$ ($\rho^2 < 0.035$, $p > 0.32$), VO_2 ($r^2 < 0.065$, $p > 0.18$), or HR ($\rho^2 < 0.012$, $p > 0.56$). Statistically similar results were obtained for the left foot. Furthermore, there were no significant correlations between α and any of the

measures of energy expenditure when all walking conditions were considered together (r^2 or $\rho^2 < 0.021$, $p > 0.17$).

As summarized in Table 5.1, no significant difference in $\dot{V}O_2$ ($p = 0.67$) was identified among the three walking conditions. However, when oxygen uptake was normalized to both body mass and walking speed to give VO_2 , the energy cost of walking a one meter distance, a significant difference was detected ($p < 0.001$). Subsequent pairwise comparisons (with Bonferroni correction) found that VO_2 calculated during OW was significantly different than both UTW ($p < 0.001$) and STW ($p < 0.001$). A significant difference in HR was also identified among the three walking conditions ($p = 0.045$) which could be attributed, after pairwise comparison (with Bonferroni correction), to a nearly significant difference between OW and UTW ($p = 0.022$). Mean α -values and preferred speeds (which were significantly slower during treadmill walking) are discussed in detail elsewhere (Fairley et al., 2009a).

Table 5.1: Measures of energy expenditure obtained during overground walking (OW), unsupported treadmill walking (UTW) and supported treadmill walking (STW) trials.

Parameter	OW	UTW	STW
$\dot{V}O_2$ (ml·kg ⁻¹ ·min ⁻¹)	17.31 ± 4.63	16.20 ± 3.70	16.69 ± 4.56
VO_2 (ml·kg ⁻¹ ·m ⁻¹)	0.259 ± 0.064	0.381 ± 0.094 [†]	0.370 ± 0.084 [†]
HR (bpm)	120 ± 9	118 ± 9	118 ± 10

Values are mean ± SD.

[†] Significantly different than OW ($p < 0.001$).

5.5 Discussion

This paediatric study found no correlation between α and measures of gross energy expenditure. These results corroborate findings for adults and the elderly (Malatesta et al.,

2003), and suggest that stride interval dynamics are not simply an outward indicator of gross physiological cost. Gross energy measures encompass basal metabolic rate and the energy required for maintenance of an upright body position, in addition to the metabolic demands of making the walking movements prescribed in this study (Malatesta et al., 2003). Therefore, if α is indeed a reflection of neuromotor control (Hausdorff, 2007; Fairley et al., 2009a), these gross energy measures may not be sensitive enough to reflect the underlying neurological processes associated with motor activity alone. Alternatively, given that an association between either energy utilization or dynamic stability has yet to be found (Malatesta et al., 2003; Fairley et al., 2009a; Chang et al., 2009a), stride interval persistence may not simply imply biomechanical efficiency. To this end, recent conjecture proposes that α may be more directly related to the extent of cognitive processes required of a particular gait task (Delignieres and Torre, 2009; Fairley et al., 2009a). Thus, future research may include investigation into the association between α and net energy expenditure (i.e., the incremental cost of walking over resting) or the effect of dual task conditions (i.e., cognitive load) on α .

Our comparison of energy expenditure between walking conditions is limited by the different speeds at which participants walked. Presumably, in addition to biomechanical dissimilarities between overground and treadmill walking (e.g. Alton et al., 1998), the significant differences identified in VO_2 and HR occurred largely due to the speed discrepancy across conditions and participants (Waters et al., 1988). Nonetheless, our decision to have subjects walk at their preferred walking speed, specific to each condition, was justified. Both α and oxygen consumption have been found to follow a U-shaped curve with speed, reaching a minimum at approximately the subject's preferred speed (Jordan et al., 2007; Browning and Kram, 2005). Therefore, it was essential to relate these parameters within the corresponding minima-regions of their respective U-shaped curves, i.e., the regions associated with minimal gross energy expenditure and 'natural' stride interval dynamics.

5.6 Conclusions

This study did not identify any significant correlations between stride interval persistence and measures of gross energy expenditure in a paediatric sample, suggesting that no simple linear association exists. The physiological meaning of stride interval persistence remains to be identified before α can become a truly informative clinical tool.

Chapter 6

Conclusion

6.1 Contributions

As a whole, this work has furthered the body of knowledge pertaining to paediatric stride interval dynamics. The main contributions of this thesis can be summarized as follows:

1. *Quantified paediatric stride interval stationarity.* This was the first known attempt to systematically quantify the weak stationarity of stride interval time series. Specifically, in the paediatric sample studied, non-stationary time series were identified during each of the three common but distinct modes of locomotion investigated (overground walking, unsupported treadmill walking and supported treadmill walking). Overall, results contested the use of scaling analysis techniques that assume stationarity when quantification of stride interval persistence is of interest.
2. *Elucidated the impact of treadmill walking on paediatric stride interval persistence.* Compared to overground walking, the stride interval persistence of children was not found to significantly differ during handrail-supported treadmill walking, but was significantly diminished during unsupported treadmill walking. In particular, this change was largely attributed to the younger children studied. Future research is needed to establish the clinical implications of this treadmill walking effect.

3. *Revealed age-specific effects of gait modality on stride interval persistence.* Within the paediatric population studied, a significant difference in stride interval persistence was identified between younger and older children during overground walking but not during treadmill walking. This suggested that treadmill walking may act to ‘level the playing field’ in terms of stride interval dynamics. Furthermore, when contrasted with adult findings in the literature, this research exposed apparent differences in treadmill stride interval dynamics between children and adults. Further investigation is required in order to delineate the developmental profile of stride interval dynamics.
4. *Found no correlation between energy expenditure and stride interval persistence.* No simple linear dependence was identified between stride interval persistence and measures of gross energy expenditure, including mass-specific gross oxygen consumption, mass-specific gross oxygen cost, and heart rate. The clinical meaning of stride interval dynamics remains to be discovered.

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