ADAPTABILITY OF HUMAN GAIT: EXPLORING RED NOISE AUDITORY STIMULI AND GAIT FLUCTUATION PATTERNS

CECILIA ROSE POWER

A THESIS SUBMITTED TO THE FACULTY OF GRADUATE STUDIES IN PARTIAL FULILLMENT OF THE REQUIREMENTS FOR THE DEGREE OF MASTER OF SCIENCE

GRADUATE PROGRAM IN KINESIOLOGY AND HEALTH SCIENCE

YORK UNIVERSITY

TORONTO, ONTARIO

AUGUST 2019

ABSTRACT

Using dynamical systems approaches to examine gait, it has been found that variability structure is important for understanding gait, and that stimuli can influence gait variability. The present study was performed to test for an adaptability limit and to analyze stepping strategies used for entraining gait. The study used Detrended Fluctuation Analysis to analyze gait variability, characterized by a Fractal Scaling Index (FSI), compared to auditory stimuli with FSI values between 1.00 to 1.25. Stepping strategy was analyzed through kinematics including cadence, stride time, and stride velocity, and comparing stride times to stimuli onset times. Fourteen participants completed one baseline trial and six stimuli-cued walking trials. For gait FSI, differences were found for both stimuli and sex. However, entrainment error only had differences for stimuli, and kinematic variables only showed differences between sexes. Overall, the study showed that a possible entrainment limit exists at a stimuli FSI value of 1.20.

ACKNOWLEDGEMENTS

This thesis is dedicated to every individual who has played a part in the process of completing it. First and foremost, I'd like to thank my supervisors, Dr. William Gage and Dr. Janessa Drake. Your guidance and wisdom have helped me become a better scientist and researcher, and I am beyond grateful to have had your support throughout the duration of my studies at York. Will, I am especially grateful to you for being my primary supervisor. Your feedback and recommendations have helped me think more critically, and you continuously inspire me to work hard.

Next, I'd like to extend my gratitude to my family. From helping me transition my life to a new province to proof-reading all of my documents, you have shown me endless love and support. Thank you for always pushing me to do my best and for always having my best interests in mind, even when I didn't know what my best interests were.

Last, but not least, I'd like to thank all my past and present colleagues within the Gage and Drake laboratories at York, as well as any other researchers who have played a role in the progress of this thesis. To my fellow lab mates, I appreciate the time we have had together, and I am grateful for the friendship we share. I'd especially like to thank Vincenzo DiBacco and Heather Johnston. Vince, I could always count on you for thought-provoking conversation and support when I needed it. Heather, I am forever grateful to you for teaching me how to use SPSS and for your statistics input. As well, thank you for always having my back and being a great friend. Finally, I'd like to thank Dr. Jeevaka Kiriella for your guidance on coding in Matlab, your troubleshooting for creating stimuli, and your helpful tips for finding literature. If it weren't for your help and for the help of everyone involved, this research would not have progressed as smoothly as it did.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

LIST OF ACRONYMS AND ABBREVIATIONS

DST	Dynamical Systems Theory
ISI	Inter-Stride Interval
DFA	Detrended Fluctuation Analysis
FSI	Fractal Scaling Index
PN	Pink Noise
WN	White Noise
RN	Red Noise
PD	Parkinson's Disease
RAS	Rhythmic Auditory Stimulation
IBI	Inter-Beat Interval
PWS	Preferred Walking Speed
HC	Heel Contact
FFT	Fast Fourier Transform
RMS	Root Mean Square
ANOVA	Statistical Analysis of Variance

1.0 INTRODUCTION

Recent studies have used a nonlinear dynamical system approach to examine gait pattern production, and the variability or noise-like patterns that exist within gait. As explained by Stergiou & Decker (2011), there is a "dynamical systems theory" (DST) which states that, to produce stable movement, the system must account for "environmental, biomechanical, and morphological constraints" and that a certain level of variability will elicit a change in gait patterns to increase stability. The DST additionally describes how dynamic systems evolve over time and how non-linear characteristics can arise within a dynamical system (Stergiou, 2016). Non-linearity is characterized by having outcomes which are not "straight lines" or "linear", but instead a change in the input results in a non-proportional change in the outcome (Stergiou, 2016). Therefore, the input results in a non-linear change in the outcome. Thus, the DST provides a basis for the importance of understanding the variability of gait patterns and the factors producing it.

One way of characterizing variability is by determining if the signal is stochastic or if it has deterministic origins (Bruijn, Meijer, Beek, & van Dieën, 2013; Kiriella, 2017; van Emmerik, Ducharme, Amado, & Hamill, 2016). As described by Stergiou (2016), a stochastic dynamical system is one such that, given its current state, can have many possible outcome behaviours, whereas a deterministic dynamical system will have only one possible future outcome for its given current state. Within the realm of non-linear, deterministic dynamical systems, systems can also be described as chaotic. Chaos exists within deterministic systems and is characterized by a sensitivity to initial conditions whereby a change in initial conditions leads to a change in the outcome (i.e. the "butterfly effect"; Stergiou, 2016). In addition, chaos is more complex than the two extremes of randomness and periodicity, but has a predictability that lies between the two extremes (Stergiou, 2016). The concepts of stochastic versus deterministic origins, non-linearity, and chaos within the DST have been used in analysing the variability or the "noise" that exists within a system across time. While many authors state that, in gait, variability or noise can be linked to instability, it has also been found that certain levels of are normal in a healthy system and can be indicative of neuromechanical mechanisms of gait (Buzzi, Stergiou, Kurz, Hageman, & Heidel, 2003; Decker, Cignetti, & Stergiou, 2010; Dingwell & Cusumano, 2000; Hausdorff, 2007).

Evidence shows that gait factors such as the time between consecutive same foot heel strikes, or inter-stride intervals (ISI), possess long-range correlations or $1/f$-scaled noise patterns (Hausdorff, Peng, Ladin, Wei, & Goldberger, 1995). Long-range, power-law, correlations mean that the fluctuations within gait parameters, like ISI, during one stride are correlated to a wide time range of past or future strides and these fluctuations decay in a "scale-free" or "fractal-like" pattern (Hausdorff, 2005; Hausdorff et al., 1995). Since this discovery, the fractal index of gait parameters under varying conditions have been examined using methods such as detrended fluctuation analysis (DFA). For example, it has been found that walking on a motorized treadmill decreases the Fractal Scaling Index, FSI, as calculated with DFA, but increases stability compared to walking on a natural ground surface (Terrier & Dériaz, 2011). Also, different walking speeds can elicit different FSI values for gait factors like ISI, stride length, step interval, and step length (Jordan, Challis, & Newell, 2007). Overall, it is a common finding that although different conditions can elicit changes in the fractal patterns of gait, in young healthy adult populations the average gait FSI value will typically fall within the region of noise patterns with $1/f$ scaling, also known as pink noise (PN), outlined in Table 1.

Along with the existence of natural fractal patterns, research has found that sensory stimuli with fractal patterns can be used to entrain gait. This has been especially useful in the case of pathological gait since authors such as Hove, Suzuki, Uchitomi, Orimo, & Miyake (2012), and

Table 1: Fluctuation values (FSI) and $1/f^{\beta}$ scaling factors for white noise (WN), pink noise (PN), and red noise (RN). All FSI values and β-values are derived from (Hausdorff et al., 1995).

	FSI value	β-value
WN	FSI = 0.5	0
PN	$0.5 \leq$ FSI ≤ 1.0	1
RN	FSI = 1.5	2

Uchitomi, Ota, Ogawa, Orimo, & Miyake (2013) have found that training using fractal auditory stimuli could increase the DFA-calculated FSI value for individuals with Parkinson's Disease (PD) from lower, closer to white noise (WN) than normal values to match healthy FSI values. Therefore, these studies found that the fractal auditory stimuli could be used to increase PD patients' gait stability. Along with this, several authors have also used auditory signals with noise-like fluctuations, or auditory fluctuating imperatives, to entrain gait in healthy individuals to test the complexity and adaptability of the locomotor system. Authors such as Hunt, McGrath, & Stergiou (2014), Kiriella (2017), and Rhea, Kiefer, D'Andrea, Warren, & Aaron (2014) have found that using stimuli with varying FSI values (i.e. WN, PN, or red noise (RN) values as shown in Table 1) can entrain gait FSI values. However, in entraining healthy gait using auditory fluctuating stimuli, FSI values do not on average reach a value of 1.0 or higher even when using RN stimuli with an average FSI of 1.3 (Hunt et al., 2014; Kiriella, 2017). This inability to entrain gait to reach FSI values of 1.0 or higher provides evidence of a possible entrainment limit whereby individuals can only entrain their gait to stimuli with FSI values lesser than the FSI limit value. Therefore, the objective of this study is to determine if it is possible to entrain gait to have FSI values of 1.0 or higher, and to determine if there is a specific FSI value limit for entrainment as stimuli FSI values approach RN.

2.0 LITERATURE REVIEW

2.1 GAIT VARIABILITY AND THE NEUROMUSCULAR SYSTEM

The variability of gait has gained attention in recent studies due to its relation to the capabilities of the locomotor system and stability. As mentioned by Balasubramanian, Clark, & Fox (2014), neural control of walking must account for three main factors in producing successful walking patterns: generation of stepping movements, whole body equilibrium or stability, and adaptability of the system. Adaptability has been linked to the variability, spatio-temporal parameters, and fractal patterns in gait, as well as events such as falls (Goldberger et al., 2002; Hak et al., 2013; Rhea & Kiefer, 2014). Furthermore, studies using nonlinear methods to analyze spatio-temporal and fractal components of gait have been conducted, and one recurring question in several papers has been whether variability in gait is due to either random processes or some other nonlinear interactions in the neuromuscular system (Buzzi et al., 2003; Dingwell & Cusumano, 2000; Hausdorff, 2007).

A study by Hausdorff et al. (1995) found that the variability in gait strides is not random, but possesses long-range correlations or that gait ISI FSI values have $1/f$ scaling or PN patterns. Hausdorff et al. (1995), as well as others have found that in healthy individuals ISI fluctuations have an average FSI value between 0.7 – 0.8 as characterized by DFA (Decker et al., 2010; Hausdorff et al., 1995; Rhea, Kiefer, D'Andrea, et al., 2014). FSI values are values which quantify the serial correlation in the rate of decay of a series (Damouras, Chang, Sejdić, & Chau, 2010). Therefore, long-range correlations in gait ISI variability suggest that the variability has persistence such that one stride is linked to many past and future strides and that the dependence decays in a fractal type of pattern (Hausdorff et al., 1995; Rhea, Kiefer, Wittstein, et al., 2014). Furthermore, as mentioned by authors such as Decker et al. (2010), and Dingwell & Cusumano (2000), locomotor patterns such as the long-range correlations and FSI values above 0.5 indicate that these patterns are not necessarily due to random processes, but that there are both deterministic and stochastic influences on locomotion. Overall, through the analysis of the variability in gait patterns, a better understanding of how the locomotor system generates movements has been attained.

2.2 INFLUENCE OF SENSORY STIMULI ON GAIT VARIABILITY

Research using sensory stimuli such as auditory or visual cues have found that gait patterns can change in response to the stimuli. For example, it has been found that the FSI values of parameters such as stride length, stride time, and stride speed can fall below 0.5 when entrained to a rhythmic auditory stimulus (RAS) (Terrier & Dériaz, 2012). However, the patterns of the stimuli, whether rhythmic or fractal, have varying effects on gait parameters. Fractal stimuli, or fluctuating timing imperatives, are such that the times between consecutive stimuli onsets fluctuate. In the case of auditory stimuli, the time between consecutive beeps, or the inter-beep interval (IBIs, fluctuates over a series of beeps. Sample WN, PN, and RN auditory signal IBI graphs are provided in Figure 1. When using music with fractal properties that have been changed to match WN, PN, or RN scaling such as the examples provided in Figure 1, research found that individuals could shift their ISI fractal scaling towards the FSI value of the stimulus, whether it is higher or lower than the baseline (Hunt et al., 2014). This shift in gait FSI is a similar finding to Rhea et al. (2014) where the use of a visual stimulus with long-range correlated fractal patterns significantly increased FSI value from baseline, whereas a stimulus with random, or WN, fractal pattern caused a decrease in the FSI value. Therefore, different stimuli result in different changes to gait patterns.

Due to young, healthy adult gait variability naturally falling within the region of PN, the ability to shift towards either WN or RN is a testament to the adaptability of the locomotor system. However, studies have shown that, although gait FSI values can be entrained and shift towards RN

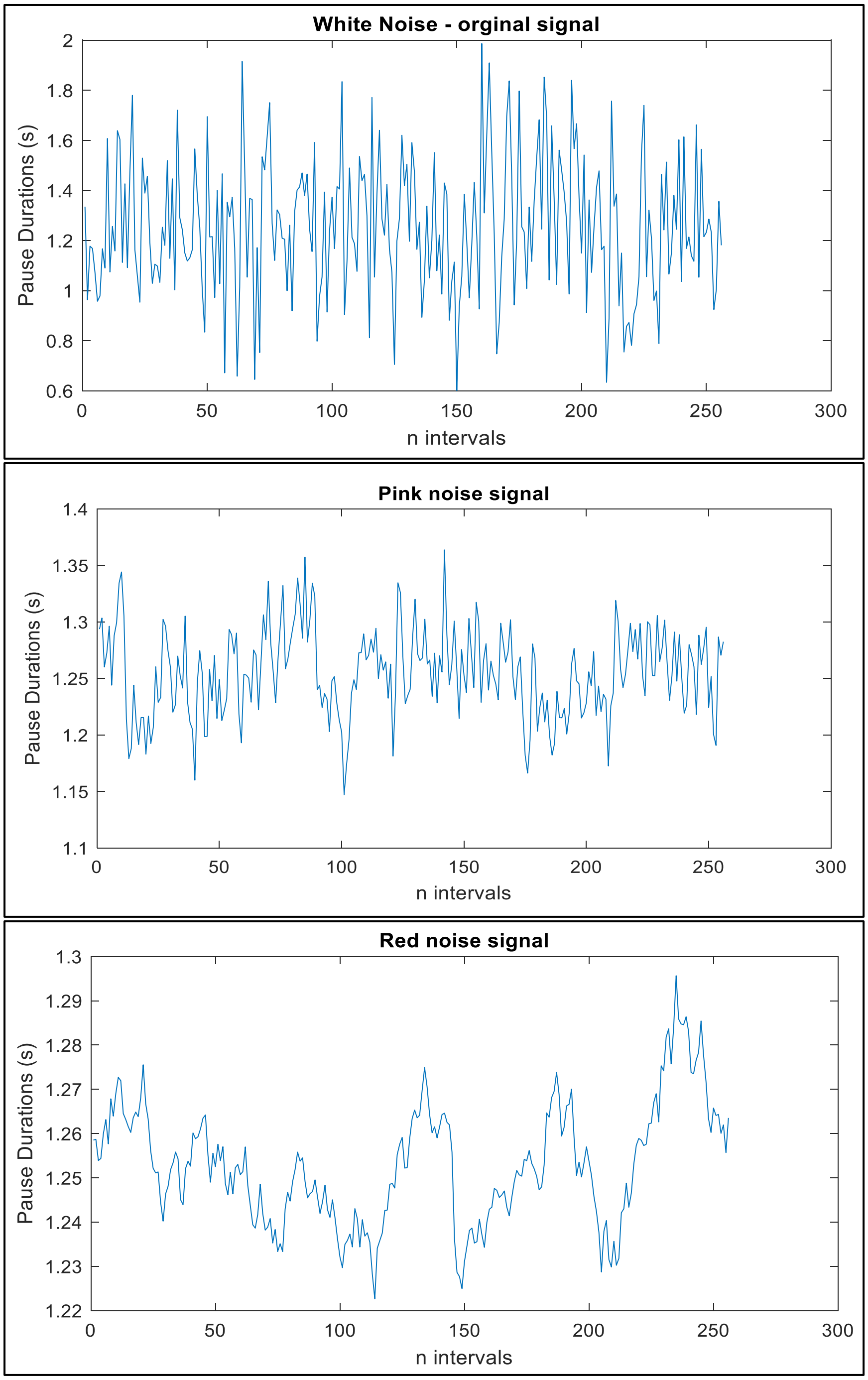


Figure 1: Sample white noise (WN), pink noise (PN), and red noise (RN) inter-beat interval (IBI) graphs. Graphs were created in Matlab. Fractal Scaling Index (FSI) of the WN, PN, and RN imperatives are approximately 0.5, 1.0, and 1.5, respectively. Each interval represents a single IBI, and the pause duration represents the time of each IBI, or the time between consecutive beeps.

scaling using fractal stimuli, they do not reach values of 1.0 or higher on average (Hunt et al., 2014; Kiriella, 2017). The finding that gait FSI does not reach 1.0 is important since a FSI value of 1.0 would indicate that the spatio-temporal gait parameters are optimally correlated or that the $1/f$ scaling is "purely scale invariant" (Hausdorff et al., 1995; Kiriella, 2017). As well, FSI values greater than or equal to 1.0 are said to no longer have the power-law type of correlations where fluctuations away from the average are typically followed by similar fluctuations (Peng, Havlin, Stanley, & Goldberger, 1995; Scafetta, Marchi, & West, 2009). An example of this would be a larger than average fluctuation being followed by another larger than average fluctuation, rather than a larger fluctuation being followed by a smaller fluctuation. Since evidence shows that gait FSI in healthy young individuals does not reach 1.0 or higher, it has been suggested that there is an upper limit to the fractal patterns individuals can produce in their gait or a limit to the stimuli FSI values that individuals can entrain to (Kiriella, 2017). This upper limit is of interest due to research by authors such as Hausdorff (2007) who have shown that as the FSI value decreases and gait becomes more variable, the instability of gait and the risk of falling increases. As well, as mentioned by Kiriella (2017), an entrainment limit would suggest that there is therefore a limit in the adaptability of the neuromuscular system.

2.3 AGING, DISEASE, AND REHABILITATION OF GAIT VARIABILITY

Studies involving older or disease-state populations have found that the fractal-like fluctuations in the gait parameters of individuals in these populations are shifted towards WN, or have lower average FSI values in comparison to young, healthy populations (Buzzi et al., 2003; Goldberger et al., 2002; Hove et al., 2012; Thaut, McIntosh, & Rice, 1997). This decrease in FSI value has been linked to a loss in motor control or adaptability. Since it has been found that gait can be influenced by using stimuli with fractal-like patterns, many authors have sought to use fractal stimuli to rehabilitate gait in older individuals or people with impaired gait such as those with PD or those who are stroke survivors. Based on the results of authors such as Hove et al. (2012), it has been found that healthy $1/f$ scaling values can be restored through rehabilitation using fluctuating stimuli whereas the same result was not achieved when using rhythmic stimuli. Some studies have also looked at the effect of using stimuli-based rehabilitation over a period of days or weeks. In one study, it was found that after a 6-week program using RAS, gait parameters such as velocity and stride length of post-stroke individuals had a significantly higher percentage of improvement compared to those who completed the rehabilitation program without the RAS (Thaut et al., 1997). For those with PD, it was found that over a period of four days, rehabilitation using auditory cues show an increasing trend for ISI FSI values (Uchitomi et al., 2013). Therefore, it can be said that training over an extended period may contribute to a higher level of gait entrainment.

2.4 GAIT ADAPTATION TECHNIQUES AND KINEMATICS

Other than non-linear techniques such as DFA, traditional linear methods can also be used in analyzing gait patterns. Some linear analyses included in previous studies include gait velocity, stride velocity, stride time, and cadence. According to Riley, Paolini, Croce, Paylo, & Kerrigan (2007), both over-ground walking and walking on a motorized treadmill have similar kinematics. These authors found that over several trials kinematic differences were small, but treadmill walking showed more consistency in kinematic measures such as gait velocity and stride time, and that these measures were statistically smaller than over-ground walking (Riley et al., 2007). For example, in one trial average walking speed was 1.41 m/s for treadmill walking and 1.48 m/s for over-ground walking (Riley et al., 2007). While small differences were found in this trial for walking speed, kinematics such as stride time were found to be the same with an average value of

1.06 s (Riley et al., 2007). In a study analysing the effect of RAS on PD patients, participants who received RAS training had increased stride velocity following the training period compared to baseline measures (Thaut et al., 1996). As mentioned by Hak et al. (2013), and Hausdorff (2005), gait measures such as these have also been used in assessing falls-risk such that factors like step time have increased variability and gait speed is decreased in those who have been determined to have a higher falls risk. However, Hausdorff (2005) also mentions that these measures can be altered by factors such as fear and that non-linear measures like FSI can be helpful in better determining falls risk. Stergiou (2016) also mentions that linear techniques contain limited information compared to non-linear techniques in terms of showing changes in neuromuscular system responses to change. While many studies have analysed gait from either linear or non-linear approaches, few have examined the relationship between kinematic measures and the effect of fractal auditory stimuli. Furthermore, few studies have examined the differences in kinematics in response to different FSI values of fractal stimuli. In analyzing the differences in kinematics, it is possible to check for kinematic strategies which allow for adaptation of gait to varying types of fractal stimuli.

Similar to kinematics, another linear analysis which has been used while examining the effect of fluctuating stimuli on gait is determining the synchronization technique that individuals use to match their heel contact (HC) to the onset of the stimulus. Particularly, evidence shows that there tends to be anticipatory behaviour in gait signals such that there tends to be a time difference between gait and the stimulus (Kiriella, 2017; Stephen, Stepp, Dixon, & Turvey, 2008; Stepp & Turvey, 2010). In general, anticipation is found in response to fractal stimuli since, as mentioned by Rhea, Kiefer, Wittstein, et al., (2014), fractal properties are often perceived (with or without awareness) by the individual as long as they are attainable. Marmelat, Torre, Beek, &

Daffertshofer (2014) found that individuals tended to show either an anticipatory or synchronized response when walking with a fractal stimulus which has a FSI near 0.9. Additionally, Kiriella (2017) found that anticipation responses occurred when individuals walked with either WN or PN types of stimuli. These findings are consistent with the notion that anticipation exists where attainable fractal properties are perceived.

Furthermore, research has found that entrainment and anticipation can occur using any stepping strategy (Kiriella, 2017; Marmelat et al., 2014; Rhea, Kiefer, Wittstein, et al., 2014). For example, anticipation can occur using a proactive strategy whereby the time difference between the signals is such that the HC occurs prior to the onset of the stimulus (Kiriella, 2017). Conversely, a reactive type of strategy can occur such that the HC times occur post-stimuli onset times (Kiriella, 2017). In analyzing stepping strategies, most of the literature has used fractal stimuli with FSI values less than 1.0. Therefore, further analysis is needed to examine what type of stepping strategy would occur in response to fractal stimuli with FSI values greater than 1.0.

3.0 RESEARCH QUESTIONS AND HYPOTHESIS

The current study was designed to examine the adaptability of the neuromuscular system's ability to produce specific fractal patterns during gait. Furthermore, it was designed to determine if there is an entrainment limit when walking with RN auditory fluctuating timing imperatives, or stimuli, with DFA-calculated FSI values of 1.0 or greater. The purpose of this design was to further explore the relationship between gait and auditory stimuli, and to test for a limit of the neuromuscular system's capacity to adapt gait. Particularly, this was done to determine if ISI FSI values greater than or equal to 1.0 are achievable for healthy populations. FSI values greater than or equal to 1.0 were of interest since, although deviations in ISI fractal scaling away from PN has been linked to pathology and fall risk, RN patterns have a different level of complexity and FSI values at 1.0 would indicate optimal spatio-temporal correlations. A secondary purpose of the study was to compare gait step times to the stimulus beep times during respective cued trials to determine if individuals tend to have proactive or reactive types of anticipatory responses when they are walking to the stimuli. Lastly, this study was done to compare changes in stimulus FSI values to gait kinematics such as stride time, cadence, and stride velocity. This was to determine if there are significant differences in kinematics when walking with or without stimuli which have FSI values of 1.0 or higher. Regarding past research and the design of this study, the study addresses the following research questions:

(1) When walking to stimuli with FSI values of 1.0 or higher, are individuals able to entrain gait to also have FSI values of 1.0 or higher?

(2) Will there be a FSI value limit for stimuli which individuals are able to entrain to, after which they will no longer show entrainment?

(3) In comparing step times to stimulus beep times, will individuals show a reactive response such that there is a time lag in the step times compared to the beep times? Or, conversely, will there be an anticipatory response such that the step times are in advance of the beep times?

(4) Lastly, will there be any differences in gait kinematics across trials with or without auditory stimuli with FSI values of 1.0 or higher?

The hypotheses of the study were that, (1) gait FSI values will equal values of 1.0 or greater in response to stimuli which have FSI values close to 1.0, and that (2) gait FSI values will differ across trials such that there will be a limit after which individuals will no longer be able to entrain their gait to the auditory stimuli. These hypotheses were developed based on work by authors such as Kiriella, (2017) since it was found that individuals were able to entrain gait to stimuli with lower

FSI values within WN or PN ranges, but were unable to entrain to stimuli with higher, RN FSI values. Next, it was also hypothesized that (3) stride HC times and stimuli onset times will be different such that there is a proactive anticipatory response. This was again based from work by Kiriella (2017) which showed that, on average, participants tend to step in advance of stimulus beep times, with the accuracy of this strategy decreasing as stimulus FSI values increase. Lastly, it was also hypothesized that (4) there will be a difference in kinematic measures across trials. Particularly, it was expected that there would be differences between trials without auditory stimuli and trials with auditory stimuli.

4.0 METHODS

4.1 OVERVIEW OF PROTOCOL

Approval to conduct this study was obtained through York University's Office of Research Ethics: Certificate #STU 2018 – 123, *Adaptability of Human Gait: Effect of Training with Red Noise Auditory Stimuli on Gait Fluctuation Patterns*. The protocol of this study was designed to determine if gait fractal scaling can be entrained to equal a FSI value of 1.0 or greater as calculated using DFA and using auditory fluctuating stimuli with RN fractal properties. This study included one lab session which followed the procedure outlined in Figure 2 and lasted approximately 1 – 2 hours. At the beginning of the lab session, basic participant anthropometric information such as age, sex, height, weight, and ankle width were collected for use with Visual3D (C-motion, USA) modelling. Then, participants were affixed with infrared reflective markers for motion capture, and safety devices including the treadmill shutoff cord and a harness affixed to the ceiling. Once the participant was ready, an initial preferred walking speed (PWS) test was performed following the method used in previous studies to determine the participant's natural gait speed (Kiriella, 2017; Marmelat et al., 2014). First, the participant began walking on the motorized treadmill at the

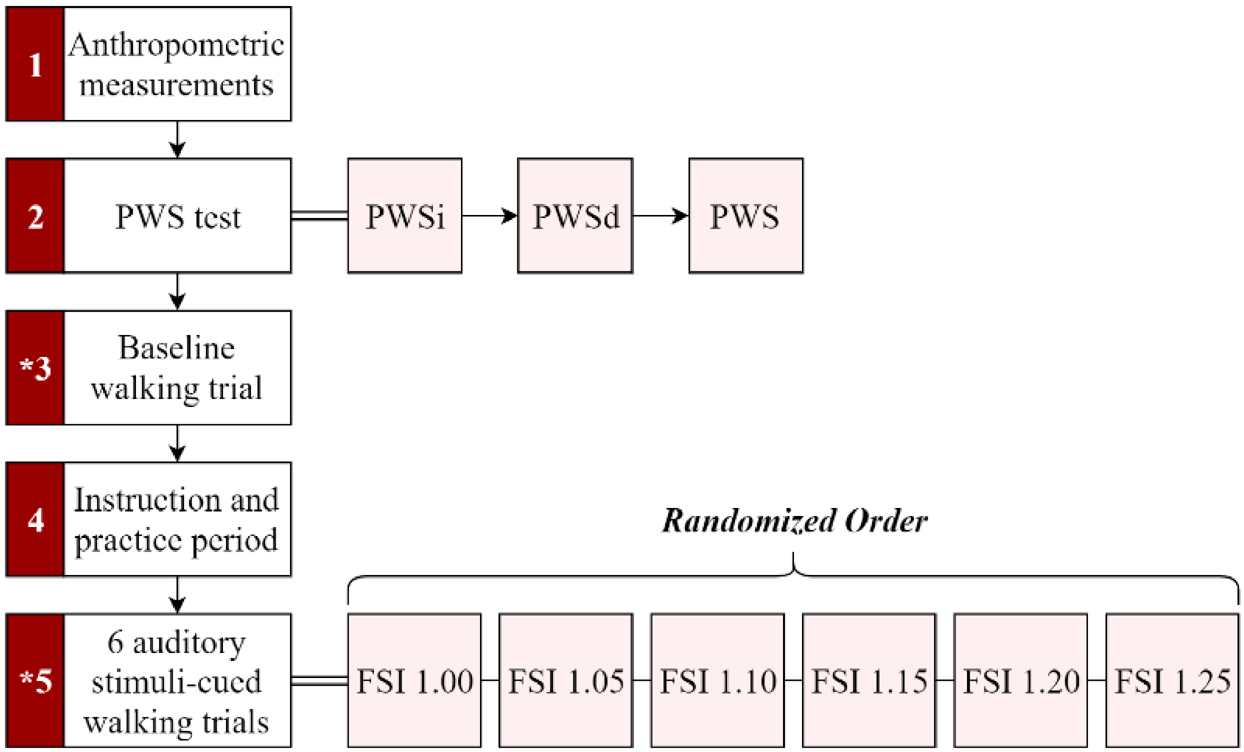


Figure 2: Procedural sequence flow chart. Sequence occurs numerically beginning with block 1 and ending with block 5. Rest periods were given between each block, between each walking trial, and whenever otherwise needed by the participant. Asterisk (*) in blocks 3 and 5 indicate motion capture collection. Stimuli (represented by their FSI values) used individually for each trial within block 5 were performed in random order.

lowest belt speed (0.50 m/s) and the belt speed was gradually increased by the researcher at equal increments (0.10 m/s) until the participant verbally indicated that they felt they were walking at a comfortable pace. This speed was defined as the increasing PWS (PWS_i). Then, the belt speed was increased until the participant was walking at a considerably fast speed, approximately 1.0 m/s greater than the PWS_i or at a speed where the participant was on the verge of running. Following this, the researcher began gradually decreasing the belt speed at equal increments (0.10 m/s) until the participant again verbally indicated that they felt that they were walking at a comfortable pace. This speed was then defined as the decreasing PWS (PWS_d). Throughout the remainder of the study, the participant would walk at their overall PWS which was calculated by taking the average of the PWS_i and the PWS_d. Once the PWS was determined, baseline gait measurements were taken in order to create the auditory stimuli to be used in the remainder of the study. In order to collect baseline measurements, participants walked on the treadmill with motion capture for approximately 5 minutes, or at least 256 strides, without any auditory stimuli. 256 strides was chosen as the minimum based on recommendations from past studies for performing a DFA calculation (Hausdorff et al., 1997; Kiriella, 2017; Marmelat et al., 2014).

Following baseline measurement collections, auditory fluctuating stimuli were created using participant mean and standard deviation ISI values which were calculated from their baseline gait measurement. A total of six stimuli were created for each participant with each metronome having FSI values equal to approximately 1.00, 1.05, 1.10, 1.15, 1.20, and 1.25 respectively. After the stimuli were created, the participants were instructed on how to match their right heel strike to the beat of the stimulus and were given a brief period for practice. Specifically, they were instructed to step on the beat of the stimulus to the best of their ability throughout the entire trial. Once they were satisfied with their practice, participants then completed six approximately 256-stride stimulus-cued walking trials. To minimize the potential effect of fatigue, cued walking trials were performed in a randomized order such that the stimulus FSI values did not occur in a sequential order, and rest breaks were given between each trial. Like the baseline uncued walking trial, motion capture data was collected for each of the cued walking trials. As well, the stimulus beep times were collected simultaneously during the cued trials through the Vicon motion capture software as an analog EMG signal in order to compare beep times to heel strike or stride times.

4.2 INFORMED CONSENT AND PARTICIPANTS

A total of 14 participants (7 male, 7 female) with an average age of 25.3 (±2.7) years drawn from the general York University community were included in this study. Participants provided written informed consent prior to testing and were assigned a random individual 5-digit code number. All data files collected were marked only with the corresponding code numbers to protect participant identity. Exclusion factors included any history of neurological disorders or injuries; recent (within the previous year) history of musculoskeletal disorders, injuries, or significant pain; any history of auditory impairments; and/or an inability to walk unassisted for periods of 20 minutes or more.

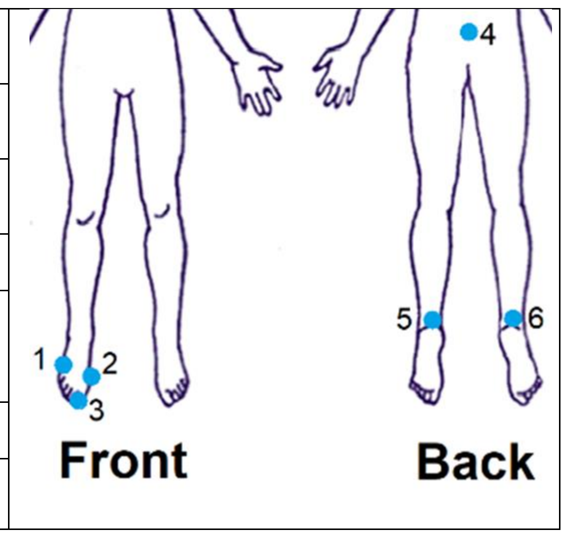
4.3 EQUIPMENT AND MEASURES

4.3.1 ANTHROPOMETRICS AND MOTION CAPTURE

For measuring individual anthropometrics, a weight scale (Seca, Chino, CA, USA) was used to determine participant height and weight and a measurement ruler was used to determine foot length and width. Motion capture was achieved using seven Vicon Mx series motion cameras (MX40, Vicon, Denver, CO, USA) and six reflective markers placed in strategic anatomic locations based on the recommendations of Robertson (2012), as outlined in Figure 3. The right

Figure 3: Marker name and corresponding anatomic location of each marker for use with 3D motion capture. A total of 6 markers are included.

	Marker Name	**Marker Anatomic Location**
1	RLAT	Right distal-lateral aspect of fifth metatarsal
2	RMED	Right distal-medial aspect of first metatarsal
3	RHAL	Right anterior hallux
4	SAC	Sacrum; mid-way between left and right posterior superior iliac spine (PSIS) landmarks
5	LCAL	Left posterior calcaneus
6	RCAL	Right posterior calcaneus



foot marker placement included the calcaneus, hallux, first metatarsal, and the fifth metatarsal. Other markers placements included the left calcaneus and the sacrum. The cameras were placed surrounding the treadmill in the collection space and were used to collect gait kinematic events like ISI, cadence, and gait velocity, which are outlined in in more detail in section 4.4.3. A motorized treadmill (Bodyguard Fitness, Quebec, Canada) was used to ensure the participants walked at consistent gait speeds, or steady-state gait.

4.3.2 AUDITORY STIMULI

Auditory stimulus signals were produced via Matlab R2018b (The Mathworks, Natick, USA) following the method proposed by Kasdin (1995), and Kiriella (2017) and were projected through external speakers (MLi 699, MidiLand, Germany) plugged into the computer. The first step in creating each stimulus was to create a Gaussian WN signal using the "wgn()" function and the participant's baseline stride time average and standard deviation. Stride time, or ISI, average and standard deviation were used as the IBI average and standard deviation to make sure the pace of the stimuli beeps matched the pace of the participant's PWS. Next, the WN signal was altered into a RN signal by using a Fast Fourier Transform (FFT), dividing by frequency amplitudes, and then using an inverse FFT. Following the creation of the RN signal, FSI value was checked to ensure it fell within an appropriate range. Signal FSI was checked by performing a DFA in Matlab using the method described by authors such as Bruijn et al. (2013), Chau (2001), and Peng et al. (1995). First, the ISI time series of length N was integrated and divided into equal length intervals. Then, a line of best fit was added and average fluctuation about this line was calculated using a root mean square (RMS) calculation for each interval, n. Next, the steps following the initial integration were repeated for varying interval, n, lengths. Once this was complete, the log values of the average RMS values were plotted against the log n values and the scaling component, FSI, was found by calculating the slope of the line. If the resultant FSI value of the RN signal was appropriate, the signal was then transformed into a stimulus using the “audiowrite()” function in Matlab, as done by Kiriella (2017). Each signal played during trials included a total of 256 data points or “beeps” which were each 10 ms in duration, and were collected as analog signals using a custom BNC cable connected to an analog to digital converter and transmitted to the Vicon software.

4.4 DATA PROCESSING

4.4.1 FRACTAL SCALING INDEX DATA PROCESSING

FSI values of the auditory stimuli were determined prior to motion capture collection using the method outlined in section 4.3.2. After collecting motion capture data, gait FSI values were also calculated using ISI data for each walking trial using the method described by Kiriella (2017), and Zeni, Richards, & Higginson (2008). The first step in calculating the gait ISI was to calculate velocity data by taking the first derivative of the right calcaneus (RCAL) marker displacement data using the “First_Derivative” pipeline in Visual3D (v5, C-motion, USA). Next, instances of HC were determined from the derived velocity data using the “Event_Threshold” pipeline to find frames just prior to descending zero crossings in the axis of progression. After finding HC times, ISI data series were created using the “Metric_Time_Between_Events” pipeline to determine the times between consecutive instances of HC. Finally, ISI data was exported to Matlab where DFA analysis was performed as described in section 4.3.2. After determining the FSI values of both the ISI time series and the stimuli for respective trials, average ISI (i.e. average stride time) and entrainment error was calculated. Average entrainment error was calculated by taking the difference between the stimulus FSI and the gait FSI values for respective trials and then taking the absolute value of the difference.

4.4.2 HEEL CONTACT TIMES AND STIMULUS BEEP TIMES PROCESSING

Since analog stimulus data and motion capture data were collected simultaneously at different sampling rates (1000 Hz and 100 Hz respectively), motion capture data first needed to be upsampled for comparison. To do this, calculated RCAL marker velocity data (as described in section 4.5.1) was upsampled to 1000 Hz using the “interp()” function in Matlab. After upsampling the velocity data, instances of HC were determined by finding zero crossings using the “find()” function in Matlab. After determining HC times, beep times from analog stimulus data were also determined. Stimulus data was exported into Matlab where the “findpeaks()” function was used to determine the times of peaks which where larger than a specified threshold value. Finally, after both HC time series data and stimulus beep time series data were created, the time difference between them was calculated by subtracting the HC times from respective beep times. This was done to determine if HC times occurred before or after respective beep times.

4.4.3 KINEMATIC DATA PROCESSING

To analyze kinematics, displacement data from both the right calcaneus (RCAL) marker and the left calcaneus (LCAL) marker were initially filtered in Visual3D software. Filtering was completed using a dual-pass, low-pass Butterworth filter with a cut-off frequency of 12 Hz. 12 Hz was chosen as the cut-off frequency due to literature which stated that walking kinematic data possesses a maximum frequency of 12 Hz (Stergiou, 2016). After filtering, left and right HC time data were created using the respective LCAL and RCAL filtered displacement data following the same method used in section 4.5.1. This included using both the “First_Derivative” and “Event_Threshold” pipelines. Left and right HC data was then transferred to Microsoft Excel (365, Microsoft Corporation, Washington, USA) where cadence was calculated using the following equation to determine the average amount of steps were taken per minute:

$$Cadence\ \left(\frac{steps}{min}\right) = \frac{\#\ steps}{\left(HC_f time - HC_i time\right) \div 60\ sec})$$

In this equation, "$\#\ steps$" was equal to the total number of left and right HC instances within the trial, minus one. One was subtracted from the total number to account for the last HC being the last point in a step cycle. Next, "$HC_f time$" corresponds to the time, in seconds, of the final HC within the trial. Similarly, "$HC_i time$" was the time, in seconds, of the initial HC within the trial. Both $HC_f time$ and $HC_i time$ could be either a left HC or a right HC time depending on which occurred first or last within the trial.

Using the filtered RCAL marker velocity data that was previously created for analyzing cadence, average walking velocity was also determined for each trial. Only the right foot was analyzed since this is the main foot of interest for all FSI data as well. Additionally, velocity measures were taken from the anterior/posterior direction of movement only since this is the plane of progression. From Visual3D, the velocity data was exported to a separate Excel file and, then, to determine the average walking velocity the filter function in Excel was used to remove all negative (i.e. posterior direction) velocities from the data. Once all data included only positive (i.e. anterior or forward direction) velocities, averages were taken for each trial to estimate the average walking velocity.

4.5 STATISTICAL ANALYSIS

Statistical analyses were primarily performed using the IBM SPSS statistics software (Version 25, IBM corporation, New York, USA) with an alpha of 0.05 used throughout all the analyses to indicate statistical significance. Prior to statistical testing, all data samples were tested for normality or skewness using a Shapiro-Wilk test of normality. Datasets which presented a left skew were corrected using a logarithmic transformation function in SPSS. The primary statistical test used for dependent measures included two-way repeated measures analysis of variance (ANOVA). ANOVAs performed compared the dependent measure means with the first factor of stimulus FSI (or trial) and the second factor of sex. In the case of statistically significant differences, Tukey post-hoc tests were used to analyze the factor of stimulus FSI and independent samples t-tests were used to analyze the factor of sex. The dependent variables used in the ANOVA tests include gait FSI value, FSI entrainment error, HC and beep time differences, cadence, stride time, and average walking speed. For the dependent variable of gait FSI value, additional outlier checks were performed including the construction of box and whisker plots.

5.0 RESULTS

5.1 FRACTAL SCALING INDEX DATA

For the dependent variable of gait FSI, two-way repeated measures ANOVA showed no significant interaction effect between the factors of sex and stimulus FSI [$F(6,13) = 0.349$, $p = 0.909$]. However, there were statistically significant differences between gait FSI values for sex [$F(1,13) = 8.338$, $p = 0.005$] and stimuli [$F(6,13) = 8.934$, $p < 0.001$]. Mean and standard deviation values per stimulus FSI value (i.e. per trial) are presented in Table 2. Table 2 was constructed using all data collected and outlines these values for all participants, for male participants, and for female participants. Post-hoc independent samples t-test for the factor of sex showed statistically significant differences between the gait FSI values of males and females [$t(96) = -2.394$, $p = 0.019$]. Next, results of Tukey post-hoc analysis are presented in Table 3 for the factor of stimuli. Significant differences were found between baseline, uncued trials and trials using stimuli with FSI values of 1.05 or 1.15. Significant differences were also found between trials with stimulus values of 1.00, 1.05, or 1.15, and trials with stimulus FSI values of 1.20 or 1.25. Lastly, trials with

Table 2: Average gait Fractal Scaling Index (FSI) values per trial across all subject groups. FSI is calculated with Detrended Fluctuation Analysis (DFA). Values are presented as by subject group and auditory stimulus (stimulus) FSI (i.e. trial). Baseline measurements are presented in the "None" column.

		Stimulus FSI						
		None	**1.00**	**1.05**	**1.10**	**1.15**	**1.20**	**1.25**
Subjects	**All (n=14)**	0.802 (±0.19)	0.984 (±0.12)	1.029 (±0.11)	0.892 (±0.22)	1.003 (±0.23)	0.747 (±0.19)	0.679 (±0.11)
	Male (n=7)	0.837 (±0.26)	1.003 (±0.12)	1.087 (±0.10)	0.991 (±0.18)	1.072 (±0.21)	0.781 (±0.22)	0.718 (±0.09)
	Female (n=7)	0.767 (±0.10)	0.966 (±0.13)	0.972 (±0.10)	0.793 (±0.22)	0.936 (±0.25)	0.714 (±0.15)	0.641 (±0.12)

Table 3: Post-hoc Tukey analysis p-values for the factor of stimulus Fractal Scaling Index (FSI) on the dependent measure of gait FSI. P-values comparing groups are listed and statistically significant differences (p<0.05) are indicated by an asterisk (*).

		Stimulus FSI						
		None	**1.00**	**1.05**	**1.10**	**1.15**	**1.20**	**1.25**
Stimulus FSI	**None**	-	0.084	**0.012***	0.804	**0.039***	0.980	0.493
	1.00	0.084	-	0.993	0.787	1.000	**0.008***	**0.000***
	1.05	**0.012***	0.993	-	0.352	1.000	**0.001***	**0.000***
	1.10	0.804	0.787	0.352	-	0.602	0.289	**0.024***
	1.15	**0.039***	1.000	1.000	0.602	-	**0.003***	**0.000***
	1.20	0.980	**0.008***	**0.001***	0.289	**0.003***	-	0.941
	1.25	0.493	**0.000***	**0.000***	**0.024***	**0.000***	0.941	-

stimulus FSI values of 1.10 were found to be statistically significantly different from trials with stimulus FSI values of 1.25. These findings, as well as the independent samples t-test findings are demonstrated in Figure 4. For the significant main effect of sex, it is evident in Figure 4 that average gait FSI values are consistently larger for males than they are for females. For the significant main effect of stimulus, groupings of significant differences are represented by the star

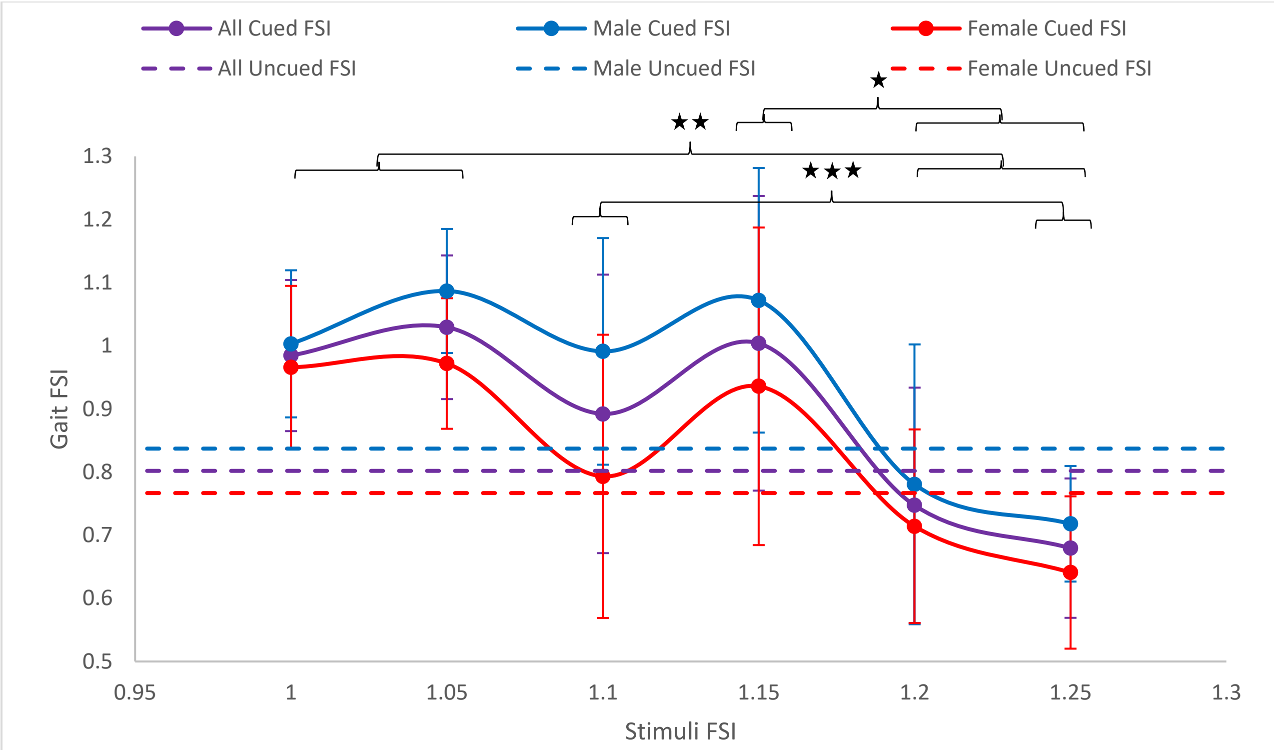


Figure 4: Gait Fractal Scaling Index (FSI) in response to varying stimulus FSI values. Baseline, uncued gait FSI measurements are represented by dashed lines. Standard deviation bars are included for cued gait FSI. Stars (★) indicate statistically significant differences.

(★) groupings in Figure 4. So, it is evident that the gait FSI for the stimuli FSI values of 1.00, 1.05, and 1.15 are larger than for stimuli FSI values of 1.20 or 1.25, and the gait FSI for the stimuli value of 1.10 is greater than for the stimuli FSI value of 1.25.

However, as also depicted in Figure 4, there were relatively large standard deviation bars, especially for the trials with stimulus FSI values of 1.10 and 1.15. So, box and whisker plots were constructed in SPSS and a manual evaluation of the data was done to check for the presence of outliers. As seen in Figure 5, one box and whisker plot which shows the presence of an outlier value is the baseline, or ".00" trial since there is one data point (labelled as point 57) with an approximate gait FSI value of 1.37 that lies outside the bounds of the whiskers. Next, the second box and whisker plot that shows the presence of an outlier is the stimulus FSI 1.15 trial. In this

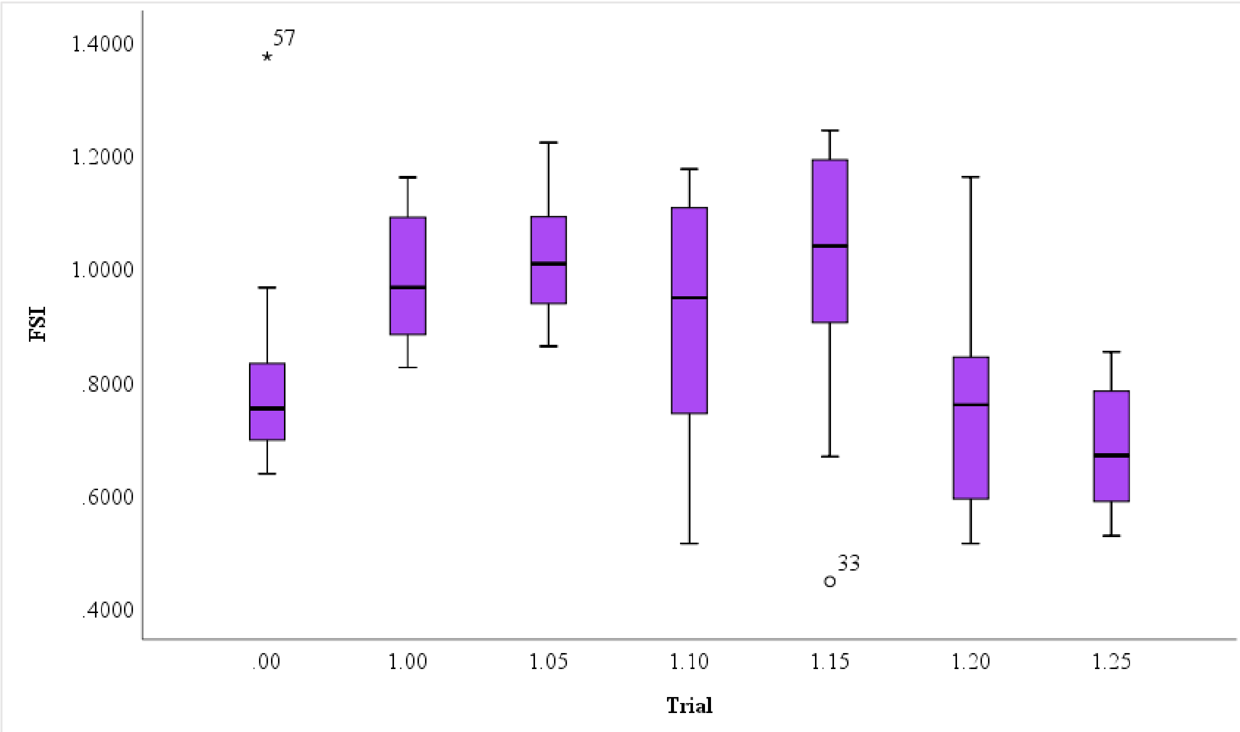


Figure 5: Box and whisker plots for gait Fractal Scaling Index (FSI) values per each stimulus FSI level (i.e. trial). Cued trials are represented by their respective FSI values and baseline, uncued trials are represented by the stimuli FSI value of ".00".

trial, one data point (labelled as point 33) lies outside the whisker bounds with an approximate value of 0.45. Therefore, trials which are potentially affected by the presence of an outlier include the baseline trial and the trial with stimuli FSI of 1.15.

Given the presence of potential outliers, a secondary ANOVA and post-hoc analysis was performed for the dependent variable of gait FSI with the outlier values removed. The results of the ANOVA analysis showed no significant interaction effect between sex and stimuli [$F(6,13) = 0.799$, $p = 0.574$], significant differences for the variable of sex [$F(1,13) = 6.386$, $p = 0.013$], and significant differences for the variable of stimuli [$F(6,13) = 16.799$, $p = 0.000$].Although this result is the same as before outliers were excluded from the analysis, post-hoc analyses show other differences between stimuli values. Conversely to ANOVA results, the results of the post-hoc

Table 4: Post-hoc Tukey analysis p-values for the factor of stimulus Fractal Scaling Index (FSI) on the dependent measure of gait FSI after exclusion of outlier values. P-values comparing groups are listed and statistically significant differences ($p<0.05$) are indicated by an asterisk (*). Significant differences not found when outliers were included are indicated by two asterisks (**).

		Stimulus FSI						
		None	**1.00**	**1.05**	**1.10**	**1.15**	**1.20**	**1.25**
Stimulus FSI	**None**	-	**0.001****	**0.000***	0.182	**0.000***	0.898	0.775
	1.00	**0.001****	-	0.979	0. 594	0.911	**0.000***	**0.000***
	1.05	**0.000***	0.979	-	0.146	1.000	**0.000***	**0.000***
	1.10	0.182	0.594	0.146	-	0.078	**0.009****	**0.003***
	1.15	**0.000***	0.911	1.000	0.078	-	**0.000***	**0.000***
	1.20	0.898	**0.000***	**0.000***	**0.009****	**0.000***	-	1.000
	1.25	0.775	**0.000***	**0.000***	**0.003***	**0.000***	1.000	-

independent samples t-test for the factor of sex showed no statistically significant differences between the gait FSI values of males and females [$t(92) = 1.975$, $p = 0.051$]. For the post-hoc Tukey analysis, significant differences were found between baseline, uncued trials and trials using stimuli with FSI values of 1.00, 1.05, or 1.15, as shown in Table 4. Additionally, significant differences were found for trials with stimuli FSI values of 1.00, 1.05, 1.10, and 1.15 compared to trials with stimuli FSI values of 1.20, and 1.25. So, compared to when the outliers were included in the analysis, additional differences were found between baseline and stimuli values of 1.00, as well as between stimuli values of 1.10 and 1.20. Overall, the results of the statistical analyses following the exclusion of the outlier values are presented in Figure 6. Even though post-hoc analysis showed no significant effect of sex, male gait FSI values are again consistently higher than female gait FSI values. This is seen in Figure 6, although male and female gait FSI values are often closer together than they were in Figure 4, particularly for the baseline values, and stimuli FSI values of 1.15 or 1.20.

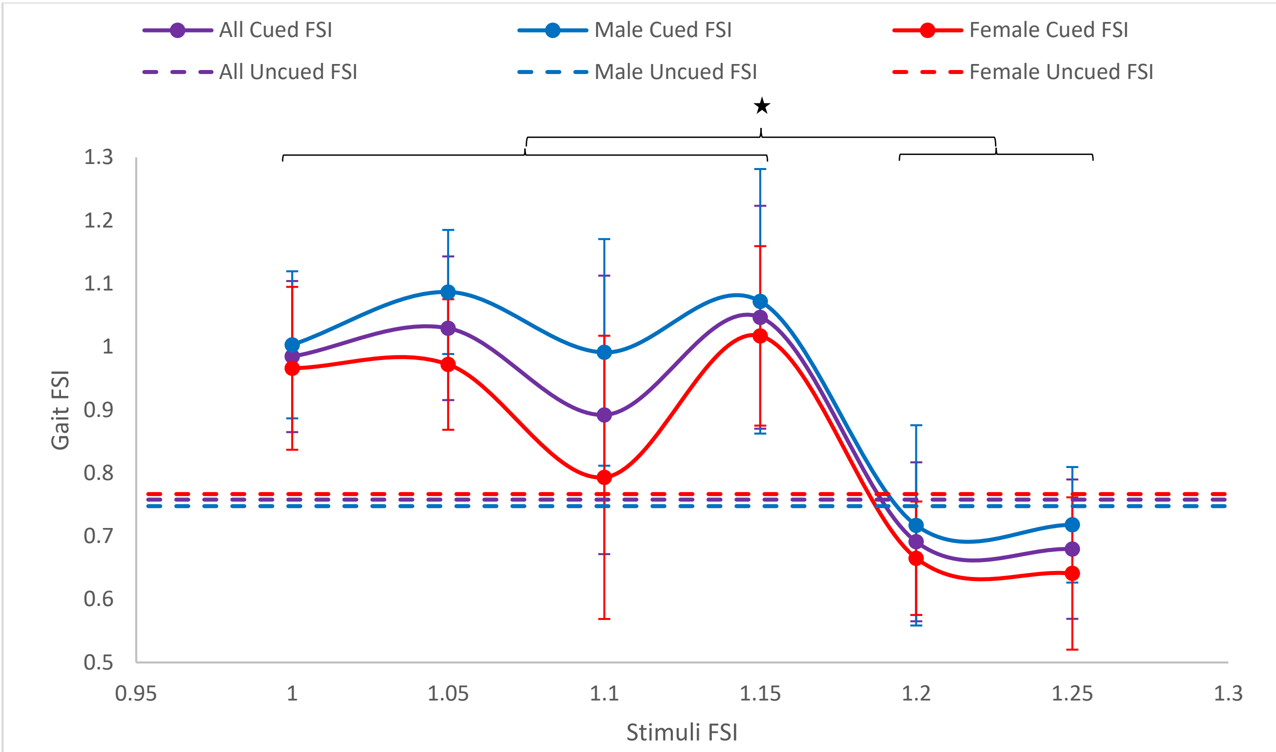


Figure 6: Gait Fractal Scaling Index (FSI) in response to varying stimulus FSI values after extraction of outlier values. Baseline, uncued gait FSI measurements are represented by dashed lines. Standard deviation bars are included for cued gait FSI. Stars (★) indicate statistically significant differences.

To further analyse the gait FSI data, a bar graph was constructed to compare gait FSI standard deviation values after removing outlier values. As seen in Figure 7, most trial standard deviations lie between the range of 0.10 to 0.15. However, the trials with stimuli FSI values of 1.10 and 1.15 have standard deviations which are larger than 0.15. This would suggest that, in using stimuli with FSI values between 1.10 to 1.15, there start being individual differences in terms of entrainment such that the gait FSI values will have greater between-individual differences. So, some people would continue entraining gait in an increasing FSI pattern up until the stimuli FSI value of 1.20 whereas some people would have lesser increases or decreasing gait FSI patterns after stimuli FSI values of 1.10. Additionally, since standard deviations in the 1.20 and 1.25 stimuli FSI trials go back to the range found for baseline and 1.00 and 1.05 stimuli FSI trials, this suggests

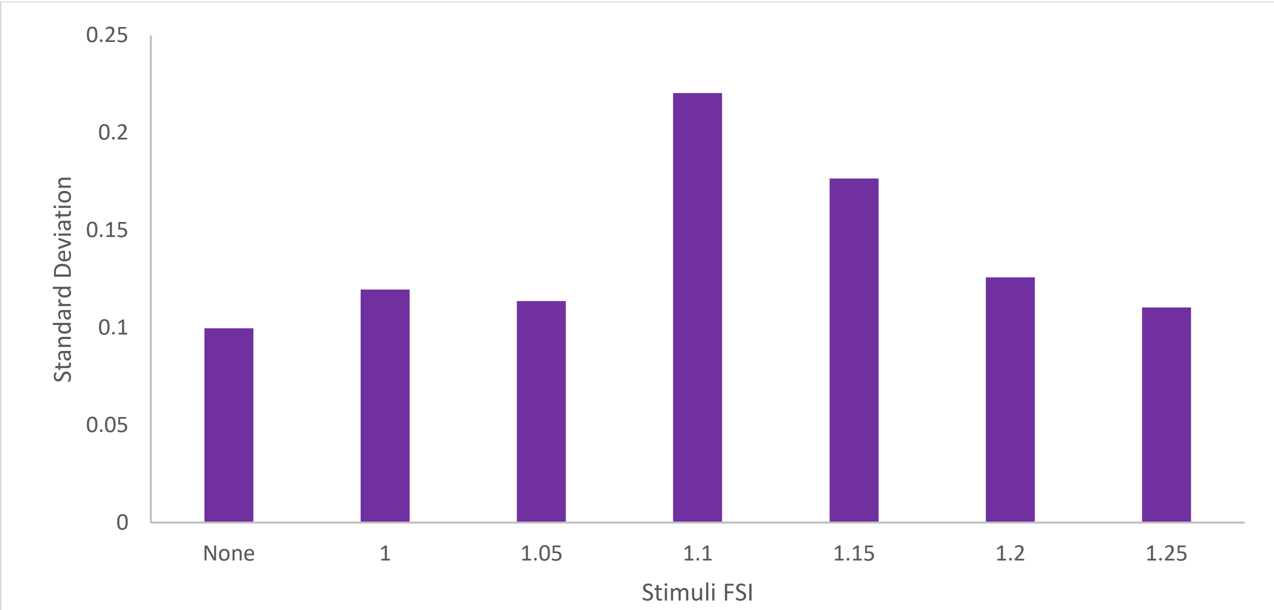


Figure 7: Gait Fractal Scaling Index (FSI) standard deviation in response to varying stimuli FSI values. Baseline, uncued walking trial standard deviations are represented by "none" stimuli FSI.

that entrainment ability for all individuals is again more similar after a stimuli FSI value of 1.20. This suggests that an adaptability limit exists at a stimuli FSI value of 1.20 because standard deviations decreased again, and gait FSI values decayed towards and below baseline values as seen in Figures 6 and 7.

The next ANOVA analysis was performed for the dependent measure of entrainment error and showed no significant interaction effect between stimulus FSI and sex [$F(5, 11) = 0.406$, $p = 0.843$] and no significant effect of sex were present in the data [$F(1,11) = 2.736$, $p = 0.102$]. However, a significant main effect of stimulus FSI values on gait FSI values [$F(5,11) = 7.625$, $p < 0.001$] was found. Mean and standard deviation gait FSI values per trial are presented in Table 5. Additionally, Table 5 shows mean and standard deviation gait FSI values for all subjects, for male subjects only, and for female subjects only. Although not statistically significant, there is a trend

Table 5: Average entrainment error values per trial across all subject groups. Error is calculated as the absolute difference between gait Fractal Scaling Index (FSI) values and stimulus FSI values. Values are presented as by subject group and auditory stimulus FSI (i.e. trial).

		Stimulus FSI					
		1.00	**1.05**	**1.10**	**1.15**	**1.20**	**1.25**
Subjects	**All (n=14)**	0.109 (±0.05)	0.095 (±0.06)	0.224 (±0.21)	0.189 (±0.20)	0.456 (±0.19)	0.573 (±0.11)
	Male (n=7)	0.101 (±0.04)	0.081 (±0.06)	0.135 (±0.16)	0.149 (±0.17)	0.422 (±0.22)	0.527 (±0.09)
	Female (n=7)	0.117 (±0.05)	0.110 (±0.06)	0.313 (±0.22)	0.228 (±0.23)	0.490 (±0.15)	0.610 (±0.12)

in the data whereby the entrainment error values are greater for females than for males. This corresponds to the previous findings whereby males had higher gait FSI values than females since the higher male FSI values were closer to the FSI values of the stimuli. Results of Tukey post-hoc analysis are presented in Table 6 for the factor of stimulus FSI on the dependent measure of entrainment error. For stimulus FSI values between 1.00 and 1.15, no significant differences between trials were found. Similarly, there was no significant difference between trials using

Table 6: Post-hoc Tukey analysis p-values for the factor of stimulus Fractal Scaling Index (FSI) on the dependent measure of gait entrainment error. P-values comparing groups are listed and statistically significant differences (p<0.05) are indicated by an asterisk (*).

		Stimulus FSI					
		1.00	**1.05**	**1.10**	**1.15**	**1.20**	**1.25**
Stimulus FSI	**1.00**	-	0.986	0.990	1.000	**0.042***	**0.004***
	1.05	0.986	-	1.000	0.988	**0.006***	**0.000***
	1.10	0.990	1.000	-	0.992	**0.007***	**0.001***
	1.15	1.000	0.988	0.992	-	**0.039***	**0.004***
	1.20	**0.042***	**0.006***	**0.007***	**0.039***	-	0.969
	1.25	**0.004***	**0.000***	**0.001***	**0.004***	0.969	-

stimulus FSI values of 1.20 and 1.25. However, trials with stimulus FSI values between 1.00 and 1.15 were found to be statistically significantly different than trials with stimulus FSI values of either 1.20 or 1.25.

5.2 HEEL CONTACT TIMES AND STIMULUS BEEP TIMES

For the analysis of the differences between HC times and stimulus beep times, two subjects were excluded due to incomplete stimulus data. ANOVA analysis ($n = 12$) of the mean difference between HC times and stimulus beep times showed no significant interaction effect between sex and trial [$F(5,11) = 0.606$, $p = 0.695$], no significant main effect of sex [$F(1,11) = 0.690$, $p = 0.409$], and no significant main effect of stimulus FSI (i.e. trial) [$F(5,11) = 0.715$, $p = 0.615$]. On average, participants showed positive differences between stimulus beeps and HC times across all trials 80.8% to 90.9% of the time. Therefore, evidence suggests that HC occurred before stimulus beep onset most of the time.

5.3 KINEMATIC DATA

The first kinematic variable analysed was cadence. Two-way repeated measures ANOVA showed a no significant interaction effect between trial and sex was found [$F(6,13) = 0.002$, $p = 1.000$]. Similarly, no significant effect of stimulus FSI (i.e. trial) was found [$F(6,13) = 0.16$, $p = 1.000$]. However, there was a significant main effect of sex on cadence [$F(1,13) = 28.314$, $p < 0.001$]. Average cadence (in steps/min) per trial and per subject group are outlined in Table 7. Post-hoc independent samples t-test comparing the males and females showed a statistically significant difference [$t(96) = -5.685$, $p < 0.001$] between males and females. As seen in Table 7, males had consistently higher cadence values across trials.

Next, the average walking speed was analyzed. Similar to cadence, ANOVA results

Table 7: Average cadence values per trial across all subject groups. Cadence is reported in steps per minute (steps/min). Values are presented as by subject group and auditory stimulus (stimulus) Fractal Scaling Index (FSI; i.e. trial). Baseline measurements are presented in the "None" column.

		Stimulus FSI						
		None	**1.00**	**1.05**	**1.10**	**1.15**	**1.20**	**1.25**
Subjects	**All (n=14)**	100.92 (±9.77)	100.17 (±9.44)	100.26 (±9.46)	100.60 (±9.50)	100.15 (±9.66)	100.15 (±9.32)	100.47 (±)
	Male (n=7)	105.35 (±10.19)	104.70 (±9.47)	104.93 (±9.44)	105.21 (±9.46)	104.91 (±9.53)	104.80 (±9.18)	105.08 (±9.75)
	Female (n=7)	96.48 (±7.56)	95.65 (±7.46)	95.59 (±7.34)	96.00 (±7.53)	95.40 (±7.66)	95.51 (±7.30)	95.87 (±7.11)

showed no significant interaction between trial and sex [F(6,13) = 0.006, p = 1.000], no significant effect of trial [F(6,13) = 0.021, p = 1.000], and a significant main effect of sex [F(1,13) = 11.760, p = 0.001]. Mean and standard deviation results per trial and per subject group for average walking speed are presented in Table 8. Post-hoc independent samples t-test to compare males and females showed a statistically significant difference [t(96) = -2.373, p = 0.020]. As seen in Table 8, male subjects showed larger average walking velocities than female subjects. This is consistent with

Table 8: Average walking speed values per trial across all subject groups. Walking speed is reported in units of metres per second (m/s). Values are presented as by subject group and auditory stimulus (stimulus) Fractal Scaling Index (FSI; i.e. trial). Baseline measurements are presented in the "None" column.

		Stimulus FSI						
		None	**1.00**	**1.05**	**1.10**	**1.15**	**1.20**	**1.25**
Subjects	**All (n=14)**	1.42 (±0.22)	1.44 (±0.21)	1.45 (±0.22)	1.44 (±0.21)	1.45 (±0.21)	1.44 (±0.22)	1.44 (±0.22)
	Male (n=7)	1.51 (±0.26)	1.51 (±0.26)	1.53 (±0.27)	1.51 (±0.25)	1.52 (±0.26)	1.50 (±0.27)	1.52 (±0.27)
	Female (n=7)	1.35 (±0.16)	1.36 (±0.14)	1.38 (±0.14)	1.37 (±0.15)	1.37 (±0.13)	1.37 (±0.15)	1.37 (±0.14)

average treadmill speeds since the chosen treadmill speed, or PWS, was on average higher for males than it was for females.

Lastly, average ISI was also analyzed. Like all other kinematic measures, there was no interaction effect between the two factors [$F(6,13) = 0.004$, $p = 1.000$, no effect of trial [$F(6,13) = 0.016$, $p = 1.000$], and a significant main effect of sex [$F(1,13) = 28.514$, $p < 0.001$]. . The average and standard deviation values are presented in Table 9 per trial and per subject grouping. For ISI, independent samples t-test showed significant differences [$t(96) = 5.705$, $p < 0.001$]. As seen in Table 9, females on average had larger stride times and therefore took more time to complete each step.

Table 9: Average Inter-Stride Interval (ISI) values per trial across all subject groups. ISI values are reported in units of seconds (s). Values are presented as by subject group and auditory stimulus (stimulus) Fractal Scaling Index (FSI; i.e. trial). Baseline measurements are presented in the "None" column.

		Stimulus FSI						
		None	**1.00**	**1.05**	**1.10**	**1.15**	**1.20**	**1.25**
Subjects	**All (n=14)**	1.199 (±0.11)	1.207 (±0.11)	1.207 (±0.11)	1.203 (±0.11)	1.208 (±0.11)	1.208 (±0.11)	1.204 (±0.11)
	Male (n=7)	1.148 (±0.11)	1.154 (±0.11)	1.152 (±0.11)	1.149 (±0.10)	1.152 (±0.11)	1.152 (±0.10)	1.150 (±0.11)
	Female (n=7)	1.250 (±0.09)	1.261 (±0.10)	1.262 (±0.09)	1.256 (±0.09)	1.265 (±0.10)	1.263 (±0.09)	1.258 (±0.09)

6.0 DISCUSSION

In using nonlinear dynamical systems approaches to examine gait, previous research has established important information about variability structure and how it relates to stability and neuromechanical control of movement (Buzzi et al., 2003; Hausdorff, 2007; Stergiou & Decker, 2011). In particular, research that applied techniques such as DFA provided evidence that the variability in gait has deterministic origins and the variability can be altered by factors such as age, disease, or external stimuli (Goldberger et al., 2002; Hausdorff, 2005; Kiriella, 2017). This changein variability is in line with the DST which explains that gait patterns are influenced by environmental, biomechanical, and morphological factors in order to produce stable movement (Stergiou & Decker, 2011). Evidence also shows that, in young, healthy adult populations, natural gait variability in measures like ISI contain a PN type of structure whereas with age or disease, there is a decay in the structure towards more randomness, or WN (Goldberger et al., 2002; Hausdorff et al., 1995; Hove et al., 2012; Kiriella, 2017). This finding has been shown consistently across many studies and was represented in the baseline measures of the present study given that an average FSI value of 0.802 (±0.19) was found for all participants. Overall, evidence suggests that DST approaches to analyzing gait are helpful for collecting evidence about variability.

In studies examining the effect of external stimuli on gait variability, it has been shown that fractal stimuli such as auditory stimuli can be used to entrain gait FSI in varying directions away from baseline values and towards the FSI of the stimulus, and also to change the stability of gait (Hove et al., 2012; Hunt et al., 2014; Kiriella, 2017; Rhea, Kiefer, D'Andrea, et al., 2014). However, studies such as those done by Hunt et al. (2014) and Kiriella (2017) have shown that entrainment typically only occurs when individuals walk with fractal stimuli which have either WN or PN types of structure and that gait FSI values typically do not reach values of 1.0 or greater. Contradictory to the findings of these studies, the current study found that it was possible for individuals (particularly males) to entrain gait to fractal stimuli with FSI values greater than 1.0 (PN), and to produce gait FSI values greater than 1.0 (Figure 2 and Figure 4). However, as mentioned by Kiriella (2017), the inability to entrain gait to RN stimuli was potentially due to the existence of an upper limit of gait adaptability. Given that the present study found that average FSI values of 1.0 or higher were only found during trials where fractal stimuli had FSI values of approximately 1.05 or 1.15, and that entrainment error was significantly larger for trials where fractal stimuli had FSI values of 1.20 or 1.25, this suggests that an entrainment limit does exist at a stimulus FSI value of 1.20. This was also evidenced in the present study through examining the standard deviation values for each stimulus level. Figure 7 illustrates that, after a transition zone in the stimuli FSI levels of 1.10 and 1.15 where standard deviations increased, the standard deviations decreased again at the stimuli FSI level of 1.20. This suggested that all individuals acted more similarly to each other before and after the transition zone and that 1.20 is a possible limit for the general population. In the studies done by Hunt et al. (2014) and Kiriella (2017), they used stimuli which had FSI values of approximately 1.30. Therefore, their results of not finding FSI entrainment larger than 1.00 is likely due to the stimuli FSI being greater than 1.20.

An additional interesting finding of the present study was that, at stimuli FSI values which approximately equal 1.10, entrainment error also starts getting larger, although not statistically different than stimuli FSI values of approximately 1.00, 1.05, or 1.15. As shown in Figures 4 and 6, the otherwise increasing trend between stimuli FSI values of 1.00 and 1.15 has a downward dip at the stimuli FSI value of 1.10. In this is particularly the case for females since gait FSI values approach baseline FSI values at this stimulus FSI value. Therefore, it is possible that for some individuals there is an entrainment limit which occurs at the stimuli FSI value of 1.10. Again, this finding is further evidenced by the transition zone shown by the standard deviations in Figure 7. Since the standard deviations of gait FSI increased within the trials with stimuli FSI values of 1.10 and 1.15 compared to other trials, a transition is suggested where some individuals have an earlier limit than others.

In comparison to non-linear measures such as DFA, it has been said that linear measures such as kinematics contain limited information compared to non-linear techniques in terms of showing changes in neuromuscular system responses to change (Stergiou, 2016). The present study further provided evidence in support of this statement since no significant differences were found across trials (i.e. across stimulus FSI values) for cadence, ISI (i.e. stride time), or average walking speed. Therefore, since significant differences were found for gait FSI values between trials, this suggests that the non-linear measures are more sensitive to detecting changes within the neuromuscular system than the linear measures. The only significant differences found for kinematic measures were in comparing the factor of sex. Overall, this suggests that the strategies individuals use to entrain their gait to changes in stimuli are not kinematic-based strategies, but rather some other type of neuromuscular system related strategy.

Other linear measures of anticipation and synchronization techniques have also been analysed in previous literature. Particularly, research shows that there tends to be a time difference between gait and the stimulus (Kiriella, 2017; Stephen et al., 2008; Stepp & Turvey, 2010). As said by Kiriella (2017), this time difference is such that HC occurs either before stimulus onset or after stimulus onset. In other words, the stepping strategy for entrainment can be proactive or reactive (Kiriella, 2017). In the current study, it was found that between 80 to 90 percent of trials (with no significant differences across trials or sexes) showed a proactive type of response such that HC occurred in advance of stimulus onset. These findings are consistent with the notion that anticipation exists where attainable fractal properties are perceived.

In revisiting the hypotheses of the study, the first hypothesis was that (1) gait FSI values will equal values of 1.0 or greater in response to stimuli which have FSI values close to 1.0. We fail to reject this hypothesis due to average gait FSI values being greater than 1.0 in certain cued trials, especially for male participants since female participants did not on average reach these values. The second hypothesis was that (2) gait FSI values will differ across trials such that there will be a limit after which individuals will no longer be able to entrain their gait to the auditory stimuli. This second hypothesis is also failed to reject on the basis that significant differences were found between trials in terms of gait FSI, and that evidence suggests there is a limit at a stimuli FSI value of 1.20 after which participants no longer show entrainment. Additionally, we also fail to reject the hypothesis that (3) stride HC times and stimuli onset times will be different such that there is a proactive anticipatory response. We fail to reject the third hypothesis since, on average, time differences existed between stimuli onset and HC such that HC times occurred in advance of stimuli onset times 80-90 percent of the time. Conversely to the first three hypotheses of the study, we rejected the final hypothesis that (4) there will be a difference in kinematic measures across trials. The fourth hypothesis was rejected due to finding no statistically significant differences in kinematic measures across trials, only across sexes.

Limitations in the present study included the absence of real-time feedback for the participants, minimal practice, limited kinematic analysis, and the absence of background information regarding participants' physical activity levels and experience. Since participants did not receive any form of feedback regarding the quality of their ability to match their HC to the stimulus, it is possible that this could play a role in the larger entrainment errors for trials using stimuli with larger FSI values. If the participants were given real-time feedback, the entrainment error could decrease and entrainment to larger FSI values could occur. However, since the participants were not given feedback, the task would need to be implicitly learned or there would need to be some sort of perception of entrainment error with or without awareness for entrainment to occur. Next, since participants were given relatively short periods of time to practice the HC matching task before trials began, it is possible that longer practice periods or training sessions could also result in greater entrainment to RN fractal stimuli. Other than entrainment, some limitations can be found in regard to the finding of no significant differences between trials for kinematics measures. For example, analysis of other kinematic variables like stride length, stride width, stance or swing time, percent single support, and percent double support, may provide more insight into kinematic strategies which change in order to adapt gait to varying stimuli. Lastly, it is possible that individuals' backgrounds and experiences could have impacted the results of the study and individuals' ability to entrain their gait to fractal stimuli. For example, individuals who have trained in music or sports which require time-based precision (such as gymnastics, figure skating, or dance) may have different entrainment abilities or adaptability than individuals who have not trained in areas such as these. Therefore, future studies could be done to analyze the effect of feedback and extended practice or training on entrainment to RN stimuli, to compare other kinematic measures to FSI and gait entrainment, or to compare participant background and experience information to their ability to entrain gait to fractal stimuli.

7.0 CONCLUSION

In conclusion, the present study showed that individuals can entrain gait to have FSI values of 1.00 or greater. However, this entrainment only exists when using stimuli which have FSI values between 1.00 and 1.15, except for stimuli with FSI values approximately equaling 1.10. Furthermore, this entrainment of gait FSI has been shown to on average only occur for males. When stimuli reach FSI levels approximately equal to 1.10, it has been shown that gait FSI levels tend to approach baseline, uncued values. Additionally, for stimuli with FSI values equal to 1.20 or higher, it has been shown that gait FSI not only approaches baseline measurements, but may also decrease below baseline PN levels and approach WN values. Therefore, the present study provided evidence that individuals can achieve optimal spatio-temporal patterns within their gait (i.e. FSI value of 1.00). Additionally, the findings of this study suggested that an entrainment limit, which was suggested to exist in previous studies such as the one done by Kiriella (2017) does exist. This entrainment limit, or a limit for the neuromuscular system's adaptability, was shown to occur at stimuli FSI values approximately equal to 1.20. However, as evidenced through entrainment errors and standard deviations, there is a transition zone between stimuli FSI values of 1.10 to 1.15 whereby some individuals start reaching a limit earlier than others. The present study also provided evidence that the strategy individuals use to match various fractal stimuli is not kinematic in nature. This is due to the absence of significant differences in kinematic measurements across trials. However, the present study did provide evidence that, for fractal stimuli with FSI values greater than 1.00, the stepping strategy used to match the stimulus is most likely to be proactive such that individuals will step in advance of the beat onset most of the time. Overall, the findings of the present study provided significant implications for understanding the neuromuscular control of gait and stability.

REFERENCES

Balasubramanian, C. K., Clark, D. J., & Fox, E. J. (2014). Walking Adaptability after a Stroke and Its Assessment in Clinical Settings. *Stroke Research and Treatment*, *2014*, 1–21. https://doi.org/10.1155/2014/591013

Bruijn, S. M., Meijer, O. G., Beek, P. J., & van Dieën, J. H. (2013). Assessing the stability of human locomotion: a review of current measures. *Journal of the Royal Society Interface*, *10*(83), 1–24. https://doi.org/10.1098/rsif.2012.0999

Buzzi, U. H., Stergiou, N., Kurz, M. J., Hageman, P. A., & Heidel, J. (2003). Nonlinear dynamics indicates aging affects variability during gait. *Clinical Biomechanics*, *18*(5), 435–443. https://doi.org/10.1016/S0268-0033(03)00029-9

Chau, T. (2001). A review of analytical techniques for gait data. Part 1: fuzzy, statistical and fractal methods. *Gait and Posture*, *13*(1), 49–66. https://doi.org/10.1016/S0966-6362(00)00094-1

Decker, L. M., Cignetti, F., & Stergiou, N. (2010). Complexity and Human Gait. *Revista Andaluza de Medicina Del Deporte*, *3*(1), 2–12.

Dingwell, J. B., & Cusumano, J. P. (2000). Nonlinear time series analysis of normal and pathological human walking. *Chaos: An Interdisciplinary Journal of Nonlinear Science*, *10*(848), 848–863. https://doi.org/10.1063/1.1324008

Goldberger, A. L., Amaral, L. A. N., Hausdorff, J. M., Ivanov, P. C., Peng, C.-K., & Stanley, H. E. (2002). Fractal dynamics in physiology: Alterations with disease and aging. *Proceedings of the National Academy of Sciences, 99* (Supplement 1), 2466–2472. https://doi.org/10.1073/pnas.012579499

Hak, L., Houdijk, H., Steenbrink, F., Mert, A., van der Wurff, P., Beek, P. J., & van Dieën, J. H. (2013). Stepping strategies for regulating gait adaptability and stability. *Journal of Biomechanics*, *46*(5), 905–911. https://doi.org/10.1016/j.jbiomech.2012.12.017

Hausdorff, J. M. (2005). Gait variability: methods, modeling and meaning. *Journal of NeuroEngineering and Rehabilitation*, *2*(19), 1–10. https://doi.org/10.1186/1743-Received

Hausdorff, J. M. (2007). Gait dynamics, fractals and falls: Finding meaning in the stride-to-stride fluctuations of human walking. *Human Movement Science*, *26*(4), 555–589. https://doi.org/10.1016/j.humov.2007.05.003

Hausdorff, J. M., Mitchell, S. L., Firtion, R., Peng, C. K., Cudkowicz, M. E., Wei, J. Y., & Goldberger, A. L. (1997). Altered fractal dynamics of gait: reduced stride-interval correlations with aging and Huntington's disease. *Journal of Applied Physiology (Bethesda, Md. : 1985)*, *82*(1), 262–269. https://doi.org/10.1016/j.measurement.2012.04.013

Hausdorff, J. M., Peng, C.-K., Ladin, Z., Wei, J. Y., & Goldberger, A. L. (1995). Is walking a random walk? Evidence for long-range correlations in stride interval of human gait. *Journal of Applied Physiology*, *78*(1), 349–358. https://doi.org/10.1152/jappl.1995.78.1.349

Hove, M. J., Suzuki, K., Uchitomi, H., Orimo, S., & Miyake, Y. (2012). Interactive Rhythmic Auditory Stimulation Reinstates Natural 1/f Timing in Gait of Parkinson's Patients. *PLoS ONE*, *7*(3), 1–8. https://doi.org/10.1371/journal.pone.0032600

Hunt, N., McGrath, D., & Stergiou, N. (2014). The influence of auditory-motor coupling on fractal dynamics in human gait. *Scientific Reports*, *4*(5879), 1–6. https://doi.org/10.1038/srep05879

Jordan, K., Challis, J. H., & Newell, K. M. (2007). Walking speed influences on gait cycle variability. *Gait and Posture*, *26*(1), 128–134. https://doi.org/10.1016/j.gaitpost.2006.08.010

Kasdin, J. (1995). Discrete Simulation of Colored Noise and Stochastic Processes and 1/f^a Power Law Noise Generation. *Proceedings of the IEEE*.

Kiriella, J. B. (2017). *An Analysis of the Relationship Between Complexity and Gait Adaptability* (doctoral dissertation). York University, Canada. Retrieved from https://yorkspace.library.yorku.ca/xmlui/bitstream/handle/10315/34351/Kiriella_Jeevaka_B_2017_PhD.pdf?sequence=2

Marmelat, V., Torre, K., Beek, P. J., & Daffertshofer, A. (2014). Persistent Fluctuations in Stride Intervals under Fractal Auditory Stimulation. *PLoS ONE*, *9*(3), 1–9. https://doi.org/10.1371/journal.pone.0091949

Peng, C.-K., Havlin, S., Stanley, H. E., & Goldberger, A. L. (1995). Quantification of scaling exponents and crossover phenomena in nonstationary heartbeat time series. *Chaos*, *5*(82), 82–87. https://doi.org/10.1063/1.166141

Rhea, C. K., & Kiefer, A. W. (2014). Patterned variability in gait behavior: How can it be measured and what does it mean? In *Gait Biometrics: Basic Patterns, Role of Neurological Disorders and Effects of Physical Activity* (pp. 17–44). Nova Science Publishers, Inc.

Rhea, C. K., Kiefer, A. W., D'Andrea, S. E., Warren, W. H., & Aaron, R. K. (2014). Entrainment to a real time fractal visual stimulus modulates fractal gait dynamics. *Human Movement Science*, *36*, 20–34. https://doi.org/10.1016/j.humov.2014.04.006

Rhea, C. K., Kiefer, A. W., Wittstein, M. W., Leonard, K. B., MacPherson, R. P., Wright, W. G., & Haran, F. J. (2014). Fractal gait patterns are retained after entrainment to a fractal stimulus. *PLoS ONE*, *9*(9), 1–10. https://doi.org/10.1371/journal.pone.0106755

Riley, P. O., Paolini, G., Della Croce, U., Paylo, K. W., & Kerrigan, D. C. (2007). A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects. *Gait and Posture*, *26*(1), 17–24. https://doi.org/10.1016/j.gaitpost.2006.07.003

Robertson, D. G. E. (2012). *Vicon Workstation Quick Reference Guide*. Ottawa, ON.

Scafetta, N., Marchi, D., & West, B. J. (2009). Understanding the complexity of human gait dynamics. *Chaos*, *19*(2), 1–10. https://doi.org/10.1063/1.3143035

Stephen, D. G., Stepp, N., Dixon, J. A., & Turvey, M. T. (2008). Strong anticipation: Sensitivity to long-range correlations in synchronization behavior. *Physica A: Statistical Mechanics and Its Applications*, *387*(21), 5271–5278. https://doi.org/10.1016/j.physa.2008.05.015

Stepp, N., & Turvey, M. T. (2010). On strong anticipation. *Cognitive Systems Research*, *11*(2), 148–164. https://doi.org/10.1016/j.cogsys.2009.03.003

Stergiou, N. (2016). *Nonlinear Analysis for Human Movement Variability*. (N. Stergiou, Ed.). Boca Raton, FL: Taylor & Francis Group.

Stergiou, N., & Decker, L. M. (2011). Human movement variability, nonlinear dynamics, and pathology: Is there a connection? *Human Movement Science*, *30*(5), 869–888. https://doi.org/10.1016/j.humov.2011.06.002

Terrier, P., & Dériaz, O. (2011). Kinematic variability, fractal dynamics and local dynamic stability of treadmill walking. *Journal of NeuroEngineering and Rehabilitation*, *8*(1), 12. https://doi.org/10.1186/1743-0003-8-12

Terrier, P., & Dériaz, O. (2012). Persistent and anti-persistent pattern in stride-to-stride variability of treadmill walking: Influence of rhythmic auditory cueing. *Human Movement Science*, *31*(6), 1585–1597. https://doi.org/10.1016/j.humov.2012.05.004

Thaut, M. H., McIntosh, G. C., & Rice, R. R. (1997). Rhythmic facilitation of gait training in hemiparetic stroke rehabilitation. *Journal of the Neurological Sciences*, *151*(2), 207–212. https://doi.org/10.1016/S0022-510X(97)00146-9

Thaut, M. H., McIntosh, G. C., Rice, R. R., Miller, R. A., Rathbun, J., & Brault, J. M. (1996). Rhythmic Auditory Stimulation in Gait Training for Parkinson's Disease Patients. *Movement Disorders*, *11*(2), 193–200. https://doi.org/10.1002/mds.870110213

Uchitomi, H., Ota, L., Ogawa, K., Orimo, S., & Miyake, Y. (2013). Interactive Rhythmic Cue Facilitates Gait Relearning in Patients with Parkinson's Disease. *PLoS ONE*, *8*(9). https://doi.org/10.1371/journal.pone.0072176

van Emmerik, R. E. A., Ducharme, S. W., Amado, A. C., & Hamill, J. (2016). Comparing dynamical systems concepts and techniques for biomechanical analysis. *Journal of Sport and Health Science*, *5*(1), 3–13. https://doi.org/10.1016/j.jshs.2016.01.013

Zeni, J. A., Richards, J. G., & Higginson, J. S. (2008). Two simple methods for determining gait events during treadmill and overground walking using kinematic data. *Gait & Posture*, *27*(4), 710–714. https://doi.org/10.1016/j.gaitpost.2007.07.007