

THE EFFECT OF KNEE OSTEOARTHRITIS ON  
THE VARIABILITY AND FRACTAL DYNAMICS OF HUMAN GAIT

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**FACULTY OF GRADUATE STUDIES AND RESEARCH**  
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## Abstract

Knee osteoarthritis (OA) is the most common joint disease seen in the older adult population, and has a greater influence on gait compared to the effects of aging alone. Spatiotemporal parameters can be used to quantify age-related changes in gait, however gait variability (i.e., the magnitude of stride-to-stride fluctuations with respect to time) and the fractal-scaling index (FSI) (i.e., correlated self-similar patterns of gait) can provide better information about the rhythmicity of the gait cycle over time. Recent studies have shown a relationship between knee OA and gait variability for certain spatiotemporal parameters, but the evidence is still preliminary and requires further investigation. Therefore, the purpose of this study was to investigate the relationship between bilateral knee OA and the temporal aspects of gait variability in a group of older adults, and to compare these results to a healthy, age- and sex-matched control group. Each participant completed a ten-minute walking test at a self-selected speed with a tri-axial accelerometer placed on his or her lower back. The device collected the vertical, mediolateral, and anteroposterior accelerations associated with gait, which were then used to analyze the stride time and step time of each gait cycle. Gait variability was assessed by calculating the standard deviation of each participant's stride time and step time. As well, the FSI (*a scaling component*) of the entire series of stride times was calculated for each participant. The degree of bilateral asymmetry for step time and step time SD for each participant was also calculated using an asymmetry index (ASI). While significant differences were found between groups for stride time and step time, no significant differences were found for any of the gait variability parameters, stride time FSI, or bilateral asymmetry measurements. Although the differences in this study were

not statistically significant, knee OA does appear to increase gait variability, decrease stride time FSI and increase the bilateral asymmetry of gait variability compared to age and sex-matched older adults with healthy knees. This study also demonstrated the effectiveness of a tri-axial accelerometer as a non-invasive measurement device to analyze gait variability in older adults with and without bilateral knee OA.

*Keywords:* knee osteoarthritis, gait variability, fractal dynamics, accelerometry

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## **Dedication**

I would like to dedicate my thesis to

My parents, Lisa Heeg and Alain Clermont, and the rest of my family

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## CHAPTER 1: Introduction

Human locomotion is a physically extraordinary process that is crucial for one's independence. The ability of the muscular, skeletal and neural control systems to work simultaneously and adapt in response to physical limitations and environmental conditions is truly remarkable. Throughout its lifespan, the human body goes through a substantial number of physical and biological changes. These changes can impact any of the locomotor systems, and will have a direct effect on a person's walking patterns. As such, human gait is said to be dynamic between three different time periods – childhood, adulthood and old age – such that the only time period of relatively stable walking patterns occurs within the approximate age-range of 15 to 60 years. At the onset of 60 years of age, changes in gait patterns become apparent, and are often influenced by the effects of aging itself. However, pathological conditions, such as knee osteoarthritis (OA), can further intensify these age-related gait disturbances, and have a greater detriment on an individual's independent living (Whittle, 2007).

Knee OA is the most common joint disease seen in the older adult population over the age of 55 years, and has a significant impact on personal suffering and the economic cost of providing health care (Minor & Kay, 2009). It is a degenerative disease that involves the loss of articular cartilage, and is associated with the development of bone spurs and thickening of subchondral bone (Ickinger & Tikly, 2014; Minor & Kay, 2009). Given that the knee plays an important role in the power and absorption phases of the gait cycle, the pathophysiological mechanisms associated with knee OA can significantly compromise the gait pattern.

Understanding the effects of age and pathology on human locomotion can be quite complex, but scientific research can provide additional information about human movement and its physical domains. Locomotion and its dynamic state can be studied in a number of different ways, but gait analysis is one of the most common methods used to describe the manner or style of walking under a wide range of circumstances. Gait analysis is a collective term that involves a number of different approaches, including simple descriptive studies (e.g., kinematic analysis), inverse dynamics analysis and mathematical modeling. With the use of measurement devices such as accelerometers, foot switches, gait mats, gyroscopes, goniometers, video-based tracking systems and force plates, the characteristics of gait can be quantified and described objectively. Although these devices are all beneficial for conducting scientific research, some may not be practical for performing clinical assessments. Therefore, with its small size, affordable cost, and wireless capabilities, accelerometry is gaining popularity as a research tool because of its potential applicability to clinical environments (Kavanagh & Menz, 2008).

For gait analysis to be understood by both researchers and clinicians, a common terminology must be implemented to describe the general patterns of gait. The most common types of parameters researched in gait analysis have been the spatiotemporal parameters – stride time, stride length, step time, step length, cadence, and speed – which can be used to assess performance, age-related changes and the effects of physical therapy. Research has demonstrated that older adults have shorter stride lengths, longer stance phases, reduced speed, and increased double support time as compared to younger adults. Furthermore, the degenerative effects of knee OA have been shown to intensify

these age-related changes in spatiotemporal parameters (Al-Zahrani & Bakheit, 2002; Astephen, Deluzio, Caldwell, Dunbar, 2008; Harris et al., 2008; Ko, Ling, Schreiber, Nesbit, & Ferrucci, 2011; Mandeville, Osternig, & Chou, 2008). It has been suggested that the age and disease-related alterations in the spatiotemporal characteristics of gait occur in response to adaptations made to maintain walking security and balance.

Although human gait appears to be a relatively constant motion, there are complex fluctuations that occur within the gait cycle, even under healthy conditions and in controlled surroundings. The magnitude of these stride-to-stride fluctuations and their changes with respect to time is referred to as gait variability. To quantify gait variability, the standard deviation (SD) and/or coefficient of variation (CoV) of one or more of the spatiotemporal parameters can be calculated and used as a comparison between groups, conditions, or lower limbs to assess gait asymmetry. Gait variability can provide important information about the rhythm and pattern of the gait cycle over time that would remain undetected when using more traditional measures such as average speed, mean stride time, and mean step length (Hausdorff, 2007).

Recently, several studies have shown that the gait cycle has fractal-like properties (i.e., self-similarity) and the measurement of the fractal-scaling index (FSI) is another parameter that can be used to quantify gait variability. The FSI shows whether the gait fluctuations occurring in a time series of data demonstrate either a random pattern or a self-similar fractal pattern that is correlated across different time scales. By using a mathematical procedure known as a detrended fluctuation analysis (DFA), a fractal-scaling component, alpha ( $\alpha$ ), can be determined in order to quantify the fractal properties of one or more of the spatiotemporal gait parameters. The DFA appears to be

a more appropriate measure to assess subtle changes in gait variability, as demonstrated in a study by Hausdorff et al. (1997b). This study found significant differences in the FSI of stride times between young and older adults; however, no significant differences in the CoV were found, suggesting that the FSI is more sensitive to age-related changes in the temporal patterns of gait.

The determination of gait variability can provide information that can be used to quantify the pathologic and age-related alterations in gait. This information can be used to provide an objective assessment of disease severity and evaluate any possible therapeutic interventions. A small amount of variability in the gait cycle is considered to be healthy, and can represent adaptability and efficient gait control with respect to unstable (i.e., changing) environmental conditions (Hausdorff, 2005). However, as people age and/or experience various pathologies, variability in the spatiotemporal parameters increases, as demonstrated with higher values for the SD and CoV (Callisaya, Blizzard, Schmidt, Mcginley, & Srikanth, 2010; Hausdorff, Rios, & Edelberg, 2001c; Kang & Dingwell, 2008). Also, research has demonstrated that more random, uncorrelated fractal patterns occur in diseased older adults compared to young and healthy adults (Hausdorff et al., 1997b; Hausdorff, 2009). The literature suggests that more random (i.e., less correlated) patterns are also associated with a decrease in lower-extremity muscle strength and range of motion, as well as rhythmic impairments in the neuromuscular control system (Hausdorff, 2007; Kang & Dingwell, 2008; Kiss, 2010b). Other studies have found differences in gait variability and the FSI in subjects with neurodegenerative disorders such as Parkinson's disease, Alzheimer's, and Huntington's disease, which demonstrates the important role of the central nervous system in

coordinating the rhythmic aspects of the gait cycle. Research is now beginning to examine the effects of physical pathologies (e.g., stroke and lower limb arthritis) on gait variability and the fractal dynamics of gait.

A study by Kiss (2011) examined the relationship between knee OA and gait variability and found that the participants with moderate and severe knee OA had significantly greater variability in step length, stance time, cadence, and double-support time for the affected knee ( $p < 0.03$ ) as compared to a healthy, age-matched control group. Additionally, sex and speed were factors for increased variability in certain parameters. These findings provide support for the idea that there is a relationship between increased gait variability and knee OA; however, the research in this area is still preliminary, and has not yet investigated the effect of knee OA on the fractal dynamics of gait.

Furthermore, a recent study found that accelerometry can be used to provide a valid determination of gait variability and the FSI in both young and older adults (Kobsar, Olson, Paranjape, Hadjistavropoulos, & Barden, 2014). Because of their ease of use and non-invasive nature, tri-axial accelerometers have the potential to become an important research tool to assess gait variability in clinical settings.

Based on the lack of studies in the research literature and the high number of individuals who develop knee OA, there is a pressing need to conduct further research to examine the relationship between knee OA and gait variability. Therefore, the purpose of this study is to investigate the relationship between knee OA and gait variability in a group of older adults using tri-axial accelerometry, and to compare these results to a healthy, age- and sex-matched control group. It was hypothesized that (i) gait variability

(i.e., stride time SD and combined step time SD) would be significantly greater in the knee OA group compared to the healthy older adult group, while the stride time FSI ( $\alpha$ ) would be less. It was also hypothesized that (ii) the bilateral asymmetry between limbs for step time SD in the knee OA group (as determined by the ASI) would be greater than the difference between limbs in the healthy control group.

## CHAPTER 2: Literature Review

### 2.1 The Knee

One of the largest and more complex freely movable synovial joints of the body is the tibiofemoral joint, also known as the knee joint. Considered a hinge joint, the knee can move freely in one plane with minimal movement in any other; however, at full flexion, it has the capability for slight rotation and lateral gliding (Saladin, 2007). Lovejoy (2007) describes the knee as “essentially two noncongruent rigidly-paired ‘balls’ perched atop corresponding virtually frictionless meniscal lined ‘sockets’” (p. 326), where the ‘rigidly-paired balls’ refer to the femur’s lateral and medial condyles and the ‘meniscal lined sockets’ are the lateral and medial menisci (Saladin, 2007).

The menisci are cartilages in the knee that absorb shock and prevent the femur from moving laterally on the tibia. Within the knee’s patellofemoral, lateral tibiofemoral, and medial tibiofemoral compartments, there are at least 13 sacs filled with synovial fluid, known as bursae that facilitate muscle or joint action (Felson, 2006; Saladin, 2007). Healthy articular cartilage is composed of a 2mm layer of hyaline cartilage with synovial fluid to make the joint motion nearly frictionless (Saladin, 2007). Within the articular cartilage, an interterritorial matrix exists, composed of a fibrillar collagen network with aggregating proteoglycans, aggrecan and many other small proteoglycans that help in lubrication and molecular movement in the extracellular matrix (Goldring & Goldring, 2010). The collagen fibers are made of type II collagen fibres, including type XI collagen within the fibril and type IX collagen in the fibril surface (Goldring & Goldring, 2010).

Four main ligaments – fibular collateral, tibial collateral, anterior cruciate, and posterior cruciate – prevent the knee from moving in unwanted directions and resist potentially disruptive forces. For example, the anterior cruciate ligament prevents hyperextension as the knee is fully extended and the fibular collateral ligament prevents the knee from rotating when the joint is extended (Lovejoy, 2007; Saladin, 2007). A fifth knee ligament, the anterolateral ligament, has been recently identified at the anterolateral aspect of the knee, which likely functions as a stabilizer for internal rotation (Claes et al., 2013). The large anterior muscle of the thigh, quadriceps femoris, and the medial hamstring component, semimembranosus, are active components for the knee's motion as their tendons stabilize the joint. Therefore, increased strength in these muscles is important for injury prevention (Saladin, 2007).

## 2.2 Arthritis

Arthritis, a chronic and debilitating disease associated with joint inflammation, swelling, limited movement, stiffness, and pain, is the leading cause of disability for those over the age of 55 years (Minor & Kay, 2009). The two most prevalent forms of arthritis are rheumatoid arthritis and osteoarthritis. Rheumatoid arthritis is caused by an overexpression of the rheumatoid factor antibody, and primarily leads to synovial membrane inflammation and swelling in the hands, wrists, feet, knees, and cervical spine (Minor & Kay, 2009). Osteoarthritis is the most common form of arthritis, affecting over three million Canadians and nearly 27 million Americans, and the prevalence of OA is estimated to increase due to the current population's greater longevity and higher obesity rates (Minor & Kay, 2009; Thomson, 2011; Ickinger & Tikly, 2014).

### 2.2.1 Osteoarthritis

Commonly diagnosed in the knees, hands, hips, and the cervical and lumbar spine, OA is known as a degenerative disease that affects the whole joint unit. This includes loss or deterioration of cartilage, thickening of subchondral bone, and the development of osteophytes and subchondral cysts (Ickinger & Tikly, 2014; Minor & Kay, 2009). Associated with aging, the cartilage cells – chondrocytes – undergo increased stress, altering the composition, content, and structural organization of collagen and proteoglycan. In OA, alterations in the composition and structure of the cartilage matrix can be associated with the deterioration of mineral properties, and result in a reduced structural integrity of the articular surface and hyaline cartilage. The consistent pattern in the OA progression begins with an upregulation of synthetic activity, an increase in extracellular matrix proteins, and an enhancement in chondrocyte cell proliferation, leading to chondrocyte clustering. Then, increased catabolic activity and production of degradative proteinase genes are associated with proteoglycan loss and type II collagen degradation (Goldring & Goldring, 2010).

Recent literature has identified possible pathomechanisms associated with the progression of OA. Among the matrix metalloproteinase (MMP) enzymes (e.g. MMP-1, MMP-3, MMP-9 and MMP-14), MMP-13, or collagenase 3, has been recognized as the major enzyme involved in OA related type II collagen degradation. MMP-13 can be upregulated by different pathological factors; for example, hypoxia-inducible factor 2 $\alpha$  (HIF-2 $\alpha$ ), a major transcription factor, is highly expressed in OA cartilage, and can cause an increase in MMP-13 activity. Hypoxia-inducible factor 1 $\alpha$  (HIF-1 $\alpha$ ) is associated with cartilage anabolism, and the balance between HIF-1 $\alpha$  and HIF-2 $\alpha$  is important for

articular cartilage homeostasis. However, an overexpression of HIF-2 $\alpha$  due to OA will affect that balance and have a catabolic effect. An overexpression of HIF-2 $\alpha$  will also suppress chondrocyte autophagy, a protective mechanism in normal cartilage for intracellular organelles and molecule turnover, leading to cell death. In healthy conditions, the interaction between the ALK-5 receptor and transforming growth factor- $\beta$  (TGF- $\beta$ ) has an anabolic effect on articular cartilage; however, the ALK-1 receptor is initiated under osteoarthritic conditions, and interacts with TGF- $\beta$  to induce MMP-13 activity. Aggracanases (e.g. ADAMTS-5, ADAMTS-4 and ADAMTS-9) also have catabolic properties in the OA progression (Goldring & Goldring, 2010); specifically, ADAMTS-5, accompanied by MMP-13, can cause a breakdown in the articular cartilage aggrecan, leading to type II collagen degradation. Finally, an increase in transmembrane heparin sulfate proteoglycan, SYNDECAN-4, suppresses the protective noncollagenous matrix protein, osteopontin, and ultimately upregulates MMP-3 to interact with ADAMTS-5 and induce a catabolic effect on proteoglycan (van den Berg, 2011).

Osteoarthritis can be diagnosed with clinical evaluation, radiography, and magnetic resonance imaging (MRI). Clinical or symptomatic evaluation is performed by a specially trained physician who will diagnose the individual based on previous history, current symptoms and criteria from the American College of Rheumatology (Jiang et al., 2012; Nelson & Jordan, 2012). Often combined with symptomatic evaluation, radiography examines osteoarthritic features – joint space narrowing, subchondral cysts, and osteophytes – with the use of X-ray. The level of severity can be determined based on criteria from the Kellgren-Lawrence (K-L) grading scale, where a score of 0 = no osteophytes; 1 = possible osteophytes only; 2 = definite osteophytes and possible joint

space narrowing; 3 = moderate osteophytes and/or definite joint space narrowing; and 4 = large osteophytes, severe joint space narrowing and/or bony sclerosis (Nelson & Jordan, 2012; Ickinger & Tikly, 2014; Jiang et al., 2012). However, the use of radiography has been criticized for being insensitive to early diagnosis since 10% of cartilage may have been already lost by the time radiography detects any characteristics of OA (Ding, Jones, Wluka, & Cicuttini, 2010). As a result, the use of MRI is becoming a more attractive tool to identify OA progression by directly examining the entire joint structure and identifying early structural changes that radiography and symptomatic evaluation may miss, such as cartilage defects and volume, subchondral bone expansion, bone marrow lesions, and meniscal lesions (Ding et al., 2010; Menashe et al., 2012). The Osteoarthritis Research Society International (OARSI) has developed a scoring system that incorporates the articular cartilage's grade and stage, where the grade's score is based on the level of articular cartilage depth, and the stage evaluates the horizontal extent of cartilage damage involved in the OA progression. The combined score between the grade and stage will determine the severity level for the articular cartilage damage in OA (Madry, Luyten, & Facchini, 2012). Although MRI can be very advantageous for OA assessment, the cost of an MRI for diagnosis is prohibitive, as the cost can be 5-22 times greater than radiography and symptomatic evaluation (Menashe et al., 2012). As each diagnostic tool has its own benefits and disadvantages, the combination of these assessments becomes an ideal method to provide a better understanding of the OA condition and progression.

Osteoarthritis can be characterized as either primary OA (i.e. genetics and normal wear and tear of aging) or secondary OA (i.e. injury, obesity, prior bone diseases,

rheumatoid arthritis and some metabolic disorders) (Ickinger & Tikly, 2014). Age is recognized as the most common risk factor for the disease (Minor & Kay, 2009). Studies have also identified occupations with heavy lifting and increased amounts of joint flexion to be potential risk factors for OA in weight-bearing joints (Conaghan, 2002; McAlindon et al., 1999; Maetzel et al., 1997; Coggon et al., 2000). The risk of OA increases with greater obesity levels during physically demanding situations (McAlindon et al., 1999). Previous history of joint injury is also considered a high risk factor for the disease, as demonstrated by a 36 year follow up study by Gelber et al. (2000) in which a 13.9% incidence rate of OA was found for those with previous trauma to the knee or hip joint compared to a 6% incidence rate in those without a history of injury.

#### 2.2.1.1 Knee Osteoarthritis

Knee OA is a degenerative disease that is a consequence of many years of ‘wear and tear’ on the knee joint, and “is best conceptualized not as an acquired disease but as an evolutionary process” (Conaghan, 2002, p. 330; Saladin, 2007). It is the most common joint disease and has a significant impact on personal suffering and health related organizations. In 2005, approximately one quarter of the population aged 55 and over experienced knee pain over the course of the year, and half of those individuals were ultimately diagnosed with symptomatic knee OA (Felson, 2006). As well, the increase in longevity and higher obesity rates within our society are significant factors for the increased incidence of knee OA (Uhlig, Slatkowsky-Christensen, Moe, & Kristian, 2010). A meta-analysis conducted by Jiang et al. (2012) suggested that every 5 kg/m<sup>2</sup> increase in body mass index (BMI) corresponds to a 35% higher probability of experiencing knee OA. Although BMI testing is not the most accurate measurement of

adipose tissue, the World Health Organization (WHO) defines overweight and obesity categories as a BMI  $\geq 25$  kg/m<sup>2</sup> and 30 kg/m<sup>2</sup>, respectively (Jiang et al., 2012).

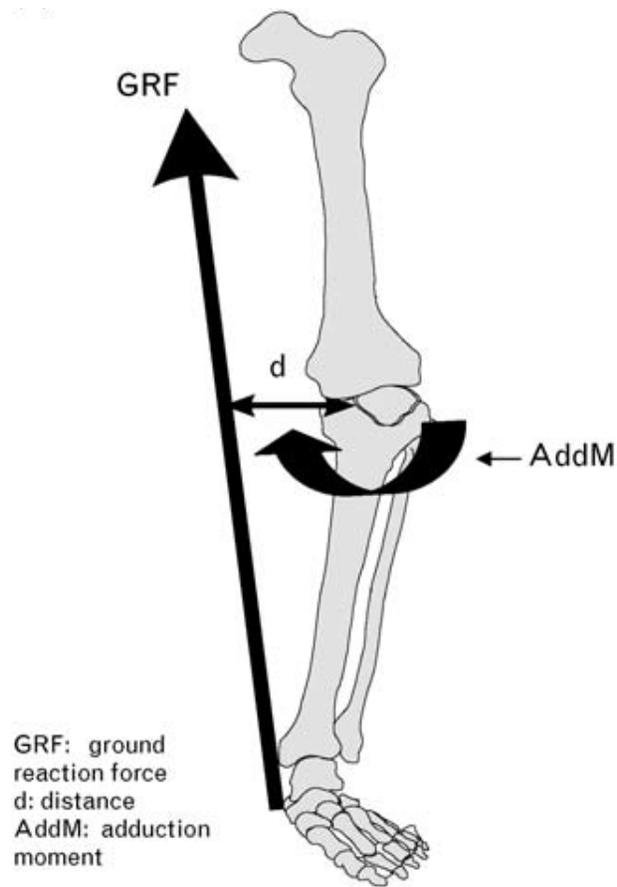
Other significant risk factors associated with knee OA include: biomechanical malalignment, being female, genetics, previous knee injury, low vitamin D and C intake, and the presence of OA in other anatomical locations (Ickinger & Tikly, 2014).

Combined with a family history of knee OA, smoking has also been suggested as a risk factor for increased loss in articular cartilage volume (Ding, Cicuttini, Blizzard, & Jones, 2007). The increasing prevalence of knee OA not only affects the quality of life for individuals suffering from the disease, but also has an economic impact on society in terms of health care demands, therapeutic advances, and the higher probability of absenteeism from work (Uhlrig et al., 2010).

Knee OA can be recognized as a disease of the entire joint, rather than merely the degeneration of cartilage (Block & Shakoor, 2010). With the same pathomechanisms as general OA, it can affect various structures within the three knee compartments – patellofemoral, medial tibiofemoral, and lateral tibiofemoral – where hyaline articular cartilage is lost, and bony remodeling occurs with weakness of periarticular muscles and capsular stretching (Felson, 2006; Ickinger & Tikly, 2014). As well, inflammation of the synovial membrane and laxity of the ligaments may occur that can cause trauma to the bone (Conaghan, 2002; Felson, 2006).

The loss of articular cartilage often becomes evident at the age of 40 years, and progressively worsens over time. If a large enough area of cartilage is lost or bony remodeling prevails, then biomechanical malalignment will develop and cause an increase in the degree of focal loading that can structurally deteriorate the knee and cause

potential joint failure (Felson, 2006; Block & Shakoor, 2010). In a static position, malalignment of the lower-limb mechanical axis (i.e., the angle formed by the femur and the tibia) is directly related to the magnitude of medial knee loading, and is associated with progression of the disease (Block & Shakoor, 2010; Tanamas et al., 2009). Although static loads have implications for the increased risk of knee OA, dynamic activity is a better representation of the activities of daily living, and may have a greater effect on the presence of knee OA. A biomechanical model (see Figure 1) can illustrate how, during walking, the ground reaction force creates a varus torque (i.e. adduction moment) on the knee joint with respect to the distance between the force and the mechanical axis. If this distance is increased due to malalignment, then the magnitude of the varus torque will increase, and will consequently affect the loading on the medial compartment of the knee joint.



*Figure 1.* Knee Joint Adduction Moment. Reprinted with permission from “Lower limb osteoarthritis: biomechanical alterations and implications for therapy,” by J. A. Block and N. Shakoor, 2010, *Current Opinion in Rheumatology*, 22, p. 545.

Compared to a standing position, ambulation can result in a three-fold increase in knee joint loading, which will increase the ‘wear and tear’ process observed in knee OA (Block & Shakoor, 2010). Eventually, at ‘end-stage’ diagnosis, over 60% of articular cartilage will be lost, and total knee replacement surgery must be performed (Ding et al., 2010). Quadriceps weakness and somatosensory deficits have also been consistent observations associated with knee OA, which can cause an affected knee motion and may be a contribution to aberrant joint loading in the disease (Lewek, Rudolph, & Snyder-Mackler, 2004; Block & Shakoor, 2010).

Since the medial compartment bears the majority of the load during walking, it is the most prevalent compartment affected by knee OA (i.e., 60-80% of cases) (Block & Shakoor, 2010). Nevertheless, the disease can affect all three compartments, and pain primarily originates from the patellofemoral compartment. Bone and synovial inflammation or a stretched joint capsule filled with fluid are the main sources of pain (Felson, 2006). Since the anterior compartment is associated with the most pain, activities with increased levels of knee flexion (e.g. walking, running, sitting, stair climbing, and jumping) are related to higher levels of pain for those with knee OA. The levels of pain and limited motion can be further affected based on the increase in severity level (Felson, 2006; Ickinger & Tikly, 2014).

### 2.3 Human Gait

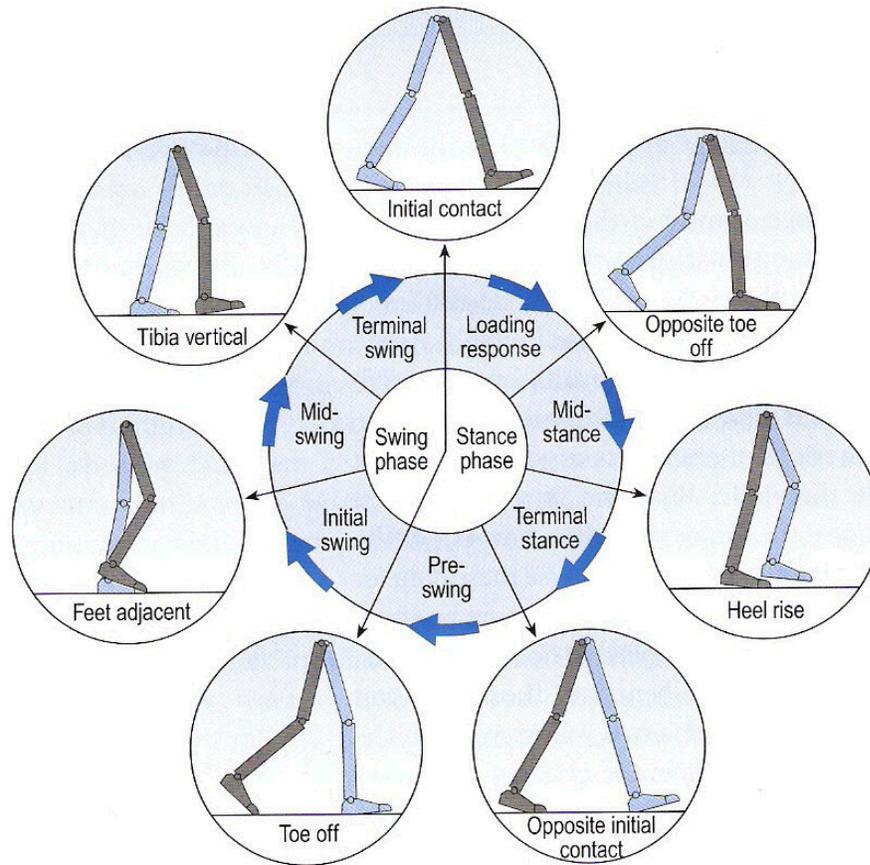
Human locomotion is an extraordinary process that “integrates the input from the motor cortex, cerebellum, and the basal ganglia, as well as feedback from visual, vestibular and proprioceptive sensors to produce carefully controlled motor commands that result in coordinated muscle firings and limb movements” (Hausdorff, 2007, p. 556).

Although the terms are often used interchangeably, *walking* and *gait* have slightly different meanings. Normal human walking and running patterns have been described as “a method of locomotion involving the use of the two legs, alternately, to provide both support and propulsion,” and, specifically with walking, the definition must include: “... with at least one foot being in contact with the ground at all times” (Whittle, 2007, p. 48). The mechanical efficiency of walking has been described using the inverted pendulum model and dynamic walking approach. As the leg encounters a collision phase (i.e., negative work), the body’s centre of mass is redirected over the stance leg for body weight support. The stance leg acts like an inverted pendulum while moving the body’s centre of mass in an arc-shaped motion, and a subsequent push-off phase (i.e., positive work) results in a low mechanical work effect using minimal muscle force (Kuo & Donelan, 2010). Gait has been defined as “the manner or style of walking, rather than the walking process itself” (Whittle, 2007, p. 48), and has been the more studied topic when assessing the locomotor characteristics of humans.

### 2.3.1 Gait Analysis

The study of gait, known as *gait analysis*, has been examined by way of descriptive studies, mathematical analysis, and mathematical modeling throughout its history (Whittle, 2007). To be universal, gait analysis must incorporate a specific terminology, which can be used in any form of methodology that studies the dynamics of gait. When the leg moves forward, a *step* is made, and the distance or time interval between the heel of the trailing leg to the heel of the leading leg is known as the *step length* or *step time*, respectively. The time interval between two successive identical events in the walking pattern is known as the *gait cycle*, and is equivalent to two

subsequent steps, or one *stride*. Within the gait cycle, seven major events have been identified, and the cycle usually begins with (1) the first contact of one foot, followed by (2) opposite toe off, (3) heel rise, (4) opposite initial contact, (5) toe off, (6) feet adjacent, and (7) tibia vertical (see Figure 2). The duration of one gait cycle is defined as the *stride time*, which is divided into *the stance phase* (i.e. when the foot is in contact with the ground) and *swing phase* (i.e. when the foot is being repositioned through the air). The stance and swing phases consist of an approximate gait cycle duration of 60% and 40% for each limb, respectively. As mentioned, one foot must always be in contact with the ground for walking to take place; therefore, a *double support* phase exists when both feet are on the ground between the initial contact of the ipsilateral foot and the subsequent toe off of the contralateral foot. This phase lasts for about 20% of each gait cycle and decreases as walking speed increases (Hausdorff, 2007; Kirtley, 2006; Whittle, 2007).



*Figure 2.* Positions of the Legs during a Single Gait Cycle by the Right Leg (gray).

Reprinted with permission from *Gait Analysis: An Introduction*, by M. W. Whittle, 2008, p. 52.

To further describe an individual's gait pattern, identifying the following measurement variables can be useful. The *cadence* is the number of steps per minute, and can be measured with a stopwatch and counting the number of individual steps taken during a selected time period. The resulting cadence can be used to calculate the average stride time (i.e. stride time (s) = 120 / cadence). Walking speed can be determined by timing the subject as he or she walks a specific distance, or by utilizing the cadence and stride length values (i.e. speed (m/s) = stride length (m) \* cadence (steps/min) / 120). The above mentioned variables – cadence, step length, stride time, stride length, stance time, swing time, double support time, and walking speed – are known as the *spatiotemporal parameters* of gait, and are common comparative variables for screening, assessing, and monitoring the efficacy of therapy and normalization of other gait measurements (Kirtley, 2006; Whittle, 2007).

#### 2.3.1.1 Gait Variability

If the multi-level neural control system is operating at a healthy functional level, an individual's gait pattern can be viewed as a relatively constant motion; however, upon closer examination, there are complex fluctuations within gait, even in controlled environmental surroundings (Hausdorff, 2005, 2007). Gait variability refers to the magnitude of stride-to-stride fluctuations in spatiotemporal parameters and their changes over time. This can characterize a dimension of locomotion that is often distinct from traditional measures, such as average gait speed and stride time, which may indicate that human gait is affected even if there are no apparent disturbances in traditional parameters. Gait variability can provide information to help understand the motor control of gait, quantify pathologic and age-related alterations, and provide objective

measurement during clinical assessments of disease severity and therapeutic improvements (Hausdorff, 2007). A two-domain viewpoint on gait variability has been suggested, where certain parameters (e.g. stride time variability) are related to rhythmicity, and others (e.g. step width variability and double-support time variability) can correspond to postural control (Gabell & Nayak, 1984; Lord, Howe, Greenland, Simpson, & Rochester, 2011a; Lord, Baker, Nieuwboer, Burn, & Rochester, 2011b).

Following data collection, the most common methods used to assess gait variability are the SD and the CoV for the spatiotemporal parameters. As illustrated in Equation 1, for a sample of  $n$  observations for a specific gait variable  $X$  (i.e.  $\{x_1, \dots, x_N\}$ ), the SD is defined as an estimate of the average variability of a set of data measured in the same units as the original data (i.e. the square root of variance). The CoV is defined as the ratio of the SD to the mean ( $M$ ) (Equation 2),

$$SD(X) = \sqrt{\frac{\sum_{i=1}^n (x_i - M)^2}{n-1}} \quad (1)$$

$$CoV(X) = \left(\frac{SD(X)}{M}\right) * 100\% \quad (2)$$

The CoV may be advantageous over SD in gait variability measurements as it is a dimensionless unit, and can be comparable to other studies. Even in healthy populations, a small amount of gait variability is considered normal, and is reflective of adaptability, efficient gait and postural control (Hausdorff, 2005). Hausdorff (2009) suggests that the CoV for stride time in healthy adults is approximately 2%. When the regulating systems for gait become impaired as a result of aging and/or disease, muscle function and

postural control are affected leading to increased gait variability (i.e. gait instability) (Hausdorff, 2005, 2007).

Another aspect of gait variability that can be explored is that of gait asymmetry (typically the difference in step parameters between legs), which has been associated with walking efficiency, balance control, and risk of musculoskeletal injury (Patterson, Gage, Brooks, Black, & McIlroy, 2010). Four equations – asymmetry ratio, asymmetry index, log-transformed asymmetry ratio, and asymmetry angle – have been the most common methods used in research to assess gait asymmetry for various spatiotemporal parameters (see Equations 3 to 6).

$$\text{Asymmetry Ratio} = \frac{ST\ Parameter_{Lower\ Limb\ 1}}{ST\ Parameter_{Lower\ Limb\ 2}} \quad (3)$$

$$\text{Asymmetry Index} = \left( \frac{ST\ Parameter_{Lower\ Limb\ 1} - ST\ Parameter_{Lower\ Limb\ 2}}{0.5 * (ST\ Parameter_{Lower\ Limb\ 1} + ST\ Parameter_{Lower\ Limb\ 2})} \right) * 100\% \quad (4)$$

$$\text{Gait Asymmetry} = \left[ 100 * \left[ \ln \left( \frac{ST\ Parameter_{Lower\ Limb\ 1}}{ST\ Parameter_{Lower\ Limb\ 2}} \right) \right] \right] \quad (5)$$

$$\text{Symmetry Angle} = \left[ \left( 45^\circ - \tan^{-1} \left( \frac{ST\ Parameter_{Lower\ Limb\ 1}}{ST\ Parameter_{Lower\ Limb\ 2}} \right) \right) * 100\% \right] / 90 \quad (6)$$

Patterson and colleagues (2010) compared the four equations to determine the clinical usefulness of each asymmetry measurement, and found each equation to be highly correlated and similar in discriminative ability. Although analyzing gait variability can provide a meaningful representation of many gait cycles, the fractal-like nature of gait can further indicate long-range, self-similar patterns to assess the stability of the gait cycle across different time scales.

### 2.3.1.2 Fractal-Like Nature of Gait

A fractal-like scaling index for gait analysis can indicate whether the gait fluctuations during a specific time series demonstrate an uncorrelated, random representation or if they display self-similar fractal patterns over a long-range period of time. According to Hausdorff (2007), a fractal explanation of gait patterns would demonstrate a dependence of a stride interval at any moment toward an interval at further remote times, which would decay over time in a fractal-like, power law fashion. The two most common gait parameters used to assess fractal-like patterns have been stride time and step width (Decker, Cignetti, & Stergiou, 2010). As with gait variability, fractal-like patterns in gait can provide information relating to an individual's locomotor function when some conventional measures – average gait speed and stride time – may not (Hausdorff, 2007).

To quantitatively characterize fractal patterns in gait, a type of spectral analysis and modified random walk analysis, known as detrended fluctuation analysis (DFA), can be used. Hausdorff (2007) describes the DFA as an assessment of a long-range correlated time series that is related to a fractal process through integration, where the various integrated time-series in a gait pattern will be self-similar to the fluctuations at different temporal moments. The different observation windows,  $F(n)$ , scale as a power-law with the number of strides within the window of observation, and will usually increase in accordance to the window size  $n$ .  $F(n) \sim n^\alpha$  results from a linear relationship in a double log graph, and the self-similarity parameter,  $\alpha$ , can be determined by calculating the slope of the line relating  $F(n)$  to  $\log n$ . White noise exists when  $\alpha = .5$ , indicating a completely uncorrelated step pattern, whereas long-range self-similar correlations exist if

$.5 < \alpha \leq 1.0$ . When  $\alpha < .5$ , the stride intervals demonstrate anti-persistent correlations, meaning a small stride interval is likely to be followed by a large one and vice versa throughout the various windows of observation.

Hausdorff et al. (1995) used the DFA to analyze the gait patterns of ten young, healthy men at a self-determined walking speed on a level and obstacle free course. Mean long-range correlations with a scaling component of  $\alpha = 0.76 \pm 0.11$  were found in the study, which demonstrated stride fluctuations that were statistically similar to those at other time scales. In another study, Hausdorff (1997b) found a fractal scaling component of  $\alpha = 0.87 \pm 0.15$  for young and healthy adults. These two values signify the range in the long-range correlation components for healthy gait using the FSI. Hausdorff (2007) described this statistical behavior as a “long-term, non-trivial ‘dependence’ or ‘memory’ in the [neurophysiological] control system,” where fluctuations in the gait pattern can be related to similar variations in stride intervals hundreds of strides earlier in a fractal-like manner (p. 560).

Further studies examined the long-range, fractal-like correlations under speed-controlled walking and running conditions. For example, Hausdorff et al. (1996) studied a group of healthy young men while walking for one hour at slow, preferred, and fast walking speeds, and found fractal-like correlations over thousands of strides in the stride interval time series of  $\alpha = .90, .84$  and  $1.0$  for the three respective speeds. Jordan, Challis, and Newell (2006) also found a similar quasi U-shaped curve of  $\alpha$  values in the stride interval time series ( $p < 0.05$ ) while studying female runners at five different speeds (80-120% of preferred running speed) over 8 minute trials with an average of 660 strides per trial. As illustrated by the findings of these two studies, the quasi U-shaped

curve of long-range correlations with respect to speed suggests a ‘biological stress’ placed on the gait patterns when speeds are above or below the preferred walking or running speed.

Simple devices, such as footswitches, are typically used to measure the large number of strides required to determine the FSI. Along with footswitches, the GAITRite™ walkway and video-based marking systems are common devices used to assess spatiotemporal parameters of gait for the determination of gait variability (Lord et al., 2011a). These measurement tools, however, are primarily used for laboratory purposes, and may not be practical for clinical assessments. However, accelerometry is a new and promising tool for gait analysis that has better potential clinical applications.

#### 2.4 Accelerometry

Various gait analysis tools provide their own benefits to research, but accelerometry has become more of an alternative approach to conventional methods because of its potential applicability in clinical analysis. It has many advantages compared to other assessment tools in gait analysis. For example, its low cost, potential use in practical settings (compared to laboratory restricted equipment), and its small size and wireless capabilities enable subjects to perform more natural movements (Kavanagh & Menz, 2008; Senden, Grimm, Heyligers, Savelberg, & Meijer, 2009). It can also provide a continuous interpretation of numerous strides over periods of days, weeks, and months. To match the desired human gait frequency, which is usually between 0.6-5.0 Hz, bandwidth settings can be set by attaching coupling filter capacitors to the accelerometer’s output (Godfrey et al., 2008).

The most common types of accelerometers used in biomechanical research are strain gauge, piezoresistive, piezoelectric, and differential capacitor. Although the four types are structurally different, they utilize the same principles of Hooke's law ( $F = kx$ ) and Newton's 2<sup>nd</sup> law of motion ( $F = ma$ ) with a mass-spring system to measure acceleration (Godfrey et al., 2008; Kavanagh & Menz, 2008). The mass-spring system will undergo a compression or stretching force depending on the associated movement, and a proportional compression or stretching force will be generated by the spring. Since the mass and spring stiffness components can be controlled, the resultant acceleration can be calculated based on a variation from the principles of Hooke's and Newton's laws (Kavanagh & Menz, 2008), as seen in Equation 7,

$$a = \frac{kx}{m} \quad (7)$$

With the use of an accelerometer attached to the human body, tri-axial accelerometry can examine the rate of body motion in the vertical, anteroposterior, and mediolateral axes. The calibration of an accelerometer is essential for data collection, and a common way to calibrate the system is with gravity. Since gravity has a well-known, constant acceleration of  $-9.81 \text{ m/s}^2$ , the vertical alignment of a stationary accelerometer's output must equal gravity, and a two-point linear calibration will transform the raw values into measurable units of acceleration (Kavanagh & Menz, 2008).

#### 2.4.1 Spatiotemporal Gait Analysis

Based on the inverted pendulum model in terms of the human's centre of mass trajectory, Zijlstra and Hof (1997) have proposed relationships between the lower trunk's

acceleration and spatiotemporal parameters of gait. The relationships also present cyclic patterns in amplitude and timing of lower trunk acceleration during gait. With comparisons to known ground reaction force patterns during the gait cycle, Zijlstra and Hof (1997, 2003) identified acceleration patterns for the vertical, anteroposterior, and mediolateral axes in relation to gait characteristics. The transition from mid-stance to single support phase can be characterized by a transition from vertical acceleration values lower than gravity to values higher than gravity, respectively. The increase in vertical acceleration is generated as the body is falling forward and downward. As the gait cycle continues from single support to double support phases, an upward body movement is seen, and there is a deceleration in the anteroposterior direction. Peak anteroposterior acceleration values during the gait cycle represent heel strike, followed by a deceleration period shortly after foot contact until mid-stance. During the right single support phase, mediolateral acceleration increases to the left direction, and during left single support, it increases to the right. As well, peak mediolateral accelerations can identify heel strike for each foot, individually. Temporal parameters, such as stride time, can be calculated as the period between two successive peak accelerations (Kavanagh & Menz, 2008). Additionally, the amplitude of each acceleration pattern becomes more pronounced as the gait speed increases (Zijlstra & Hof, 2003).

Vertical acceleration values are also useful to calculate step length and mean walking speed. According to Zijlstra & Hof (1997; 2003), step length can be determined based on changes in the height of a person's centre of mass with the use of Equation 8,

$$\text{step length} = 2\sqrt{2lh - h^2} \quad (8)$$

where  $h$  is the height difference of the centre of mass, or the change in vertical acceleration amplitude between highest and lowest values in the step cycle, and  $l$  equals leg length. The mean step length can be divided by mean step duration to find mean walking speed.

Considering the accelerations are based on the local coordinate system of the accelerometer, proper placement of the device must occur for the optimal measurement of segmental acceleration. The literature suggests placing the accelerometer on the L3 spinous process of the lower back to reflect actual lower trunk acceleration during walking. As well, the data is best interpreted when the subject's walking speed is at a steady pace and the gait pattern is cyclical (Kavanagh & Menz, 2008). Steady state walking speed can be achieved within the first few steps in healthy individuals, but for older participants, Lindemann et al. (2008) suggests gait analysis should begin after at least 2.5 metres if their self-selected walking speed is under the habitual 1.2 m/s. It is also recommended to measure the steady state walking motion for at least 20 gait cycles in order to properly assess the individual's spatiotemporal parameters. This is particularly important when measuring gait variability and the fractal scaling index, as long-range, power law correlations for stride data are required for proper measurements (Lindemann et al., 2008).

Although the use of accelerometry to examine spatiotemporal parameters is relatively recent, studies have found good reliability and repeatability in terms of walking speed, cadence, step duration, and step length when comparing accelerometric gait analysis to other motion analysis tools, previous research, or under different experimental conditions (Henriksen et al., 2004; Hartmann, Luzi, Murer, de Bie, & de

Bruin, 2009; Hartmann, Murer, de Bie, & de Bruin, 2009; Senden et al., 2009; Zijlstra & Hof, 2003). Studies have used accelerometers to examine and compare spatiotemporal gait parameters between healthy populations of young and older adults under different walking speeds and environmental conditions (Henrisken, Lund, Moe-Nilssen, Bliddal, & Danneskiold-Samsøe, 2004; Hartmann et al., 2009b; Senden et al., 2009; Shimpl et al., 2011). As well, accelerometry is beginning to gain popularity as a tool for identifying differences in spatiotemporal parameters for pathological conditions, such as Parkinson's disease (Maquet et al., 2010; Yang, Hsu, Snih, Lu, & Chan, 2011) and stroke patients (Mizuike, Ohgi, & Morita, 2009; Lee et al., 2010).

## 2.5 Gait and Aging

In an individual's lifetime, the gait is dynamic between three different time periods – childhood, adulthood, and old age. A child's spatiotemporal parameters continually change throughout their growth period until they reach the approximate age of 15, where their normal adult values are often attained (Whittle, 2007). Following relatively stable natural walking patterns throughout most of adulthood, the onset of age-related changes in gait begins within the range of 60-70 years and are influenced by either “the effects of age itself [or] the effects of pathological conditions, such as [OA] and parkinsonism” (Whittle, 2007, pp. 86-87). In the older adult populations, changes in gait parameters – decreased stride length, increased stance phase, reduced speed, and increased double support time – occur to maintain walking security and balance (Maki, 1997 as cited in Kirtley, 2006; Murray, Kory, & Clarkson, 1969 as cited in Whittle, 2007).

### 2.5.1 Stride-to-Stride Fluctuations

Since the dynamics of gait can be affected by age-related causes, studies of gait variability and the fractal-like nature of gait have primarily focused on the older adult population. The most common spatiotemporal parameters measured for gait variability and aging have been step time, stance time, swing time, stride time, step length, step width, stride length, and gait speed (Lord, Howe, Greenland, Simpson, & Rochester, 2011). Research has demonstrated a relationship between an increased level of variability for the spatiotemporal parameters in gait and an increase in age (Callisaya, Blizzard, Schmidt, McGinley, & Srikanth, 2010; Hausdorff et al., 2001b; Hausdorff et al., 2001c; Kang & Dingwell, 2008). Within the studied parameters, increased stride time variability has been a common reflection of gait instability, and has been linked to decreased physical activity levels (Beauchet et al., 2009; Hausdorff et al., 2001b, 2001c; Herman, Giladi, Gurevich, & Hausdorff, 2005; Kang & Dingwell, 2008; Montero-Odasso et al., 2011). Decreased gait velocity has also been shown to be associated with increased gait variability (Beauchet et al., 2009; Buzzi, Stergiou, Kurz, Hageman, & Heidel, 2003; Herman et al., 2005).

It has been shown that an individual's fractal properties of gait become impaired with age as well. In a study by Hausdorff et al. (1997b), the FSI displayed a significant difference ( $p < 0.003$ ) in stride interval fluctuations between disease-free older adults ( $M_{AGE} = 75.7 \pm 3.2$  years) and young healthy adults ( $M_{AGE} = 24.6 \pm 1.9$  years), where the respective  $\alpha$  scaling exponents were  $0.68 \pm 0.14$  and  $0.87 \pm 0.15$ , respectively. Although the FSI was significantly different between the two groups, the CoV and average stride time values were not. This indicates that the DFA may be a more sensitive measurement

to assess subtle, age-related changes in gait. It has been suggested that the positive relationship between decreased gait stability (referring to the stability of the gait pattern) and age may be caused by a decrease in lower-extremity muscle strength, function, and range of motion (Hausdorff et al., 2001b, 2001c; Hausdorff, 2005, 2007; Kang & Dingwell, 2008; Kiss, 2010b), as well as “an inconsistency of the central neuromuscular control system’s ability to regulate gait and maintain a steady walking pattern” (Montero-Odasso et al., 2011, p. 7).

### 2.5.2 Falls and Frailty

The various age-related factors responsible for increased gait variability can lead to a decrease in dynamic balance, which has motivated researchers to examine the relationship between gait variability, frailty, and falls. Falls have become an increasingly serious health issue with significant economic and social consequences. Fractures, hospitalization, mobility restrictions, fear of falling, and a loss of independence are a few negative outcomes following incidents of falls, which can lead to institutionalization and mortality (Hausdorff, 2007). Research has found a relationship between higher levels of gait variability and a previous history of falls (Hausdorff et al., 1997a; Hausdorff et al., 2001b), the risk of future falls (Hausdorff et al., 2001c; Kang & Dingwell, 2008; Verghese, Holtzer, Lipton, & Wang, 2009) and fear of falling (Herman et al., 2005). As well, Montero-Odasso et al. (2011) have suggested that increased variability in stride time, stride length, and step width is significantly related to frailty, with greater stride time variability being the most affected parameter. The understanding of gait variability’s relationship to falls and instability may help identify potential risk factors and improve physiological function through therapeutic interventions prior to

injury (Hausdorff, 2005). For example, Hausdorff et al. (2001b) found a significant reduction in stride time variability following a six-month exercise program, which was associated with increased knee extension strength and higher scores on a physical performance test.

Clearly, a number of age-related physiological deficits can increase gait variability and affect the fractal-like nature of spatiotemporal gait parameters. These deficits have an obvious negative effect on balance and stability with respect to frailty and falls. Due to the interrelationships of physiological capacity, neuropsychological status, and health-related quality of life, human pathological conditions have also been shown to have a significant impact on stability and falling (Hausdorff et al., 2001b).

## 2.6 Pathophysiology and Gait

### 2.6.1 Neurodegenerative Diseases

A number of neurodegenerative diseases have been studied in terms of gait variability and the FSI of gait. For example, Parkinson's disease has been related to disturbances in stride length and speed (Ebersbach et al., 1999; Morris, Iansak, Matyas, & Summers, 1994; Morris, Iansak, Matyas, & Summers, 1996), and research has now extended this knowledge to indicate increased variability in stride length and stride time for those experiencing the disease (Blin, Ferrandez, & Serratrice, 1990 as cited in Hausdorff, 2007; Hausdorff et al., 1998). Gait variability for those with Parkinson's disease has also been shown to be associated with increased disease severity (Hausdorff, 2007), and the literature suggests the increased variability "reflects alterations in rhythm generation and motor programming" (p. 569). Additionally, a recent review article by Hausdorff (2009) outlined the fractal-like scaling properties of those suffering from

Parkinson's disease and suggested that the stride-interval fluctuations are altered with the disease, especially as the pathology progresses.

Other neurodegenerative disorders that can affect gait pattern stability include Huntington's and Alzheimer's disease (Hausdorff, 1997b; Hausdorff et al., 1998; Sheridan, Solomont, Kowal, & Hausdorff, 2003). Hausdorff (1997b) studied the relationship between disease and stride-interval fluctuations in subjects suffering from Huntington's disease. The results demonstrated a significantly lower  $\alpha$  level in the Huntington's disease subjects compared to the disease-free control group (Huntington's:  $\alpha = 0.60 \pm 0.24$ ; controls:  $\alpha = 0.88 \pm 0.17$ ;  $p < 0.005$ ). The lower  $\alpha$  value for the individuals with Huntington's disease suggests a more random fractal-like scaling factor for the stride-interval fluctuations.

Researchers believe that alterations in the fractal properties of stride-interval correlations are likely associated with impairments in the central nervous system (that may occur as a result aging, for example), which provides the rationale for investigating the effects of neurodegenerative disorders on the fractal-like properties of gait. Although knee OA is a joint pathology, many of the gait disturbances seen in knee OA are similar to those seen in older adults and in subjects with various neurodegenerative disorders, such as the ones mentioned previously. This suggests there may be a relationship between the fractal-like nature of gait and knee OA.

### 2.6.2 Knee Osteoarthritis

As mentioned previously, age-related changes in gait occur inevitably, even in healthy individuals. Given that knee OA has its own effect on gait dynamics, the combination of age-related gait impairments with other pathophysiological mechanisms

will intensify the gait complications. The knee experiences a varied range of motion throughout the gait cycle and serves as a power producer and contact energy absorber during the stance phase (Lovejoy, 2007; Saladin, 2007; Whittle, 2007). Consequently, if the knee is affected by OA, then a compromised gait pattern will reflect the degenerative disease.

#### 2.6.2.1 Joint Kinematics

Research has primarily concentrated on the joint kinematics associated with knee OA. Studies have found a reduced range of motion of the affected knee and alterations in knee flexion, extension and adduction moments (Al-Zahrani & Bakheit, 2002; Astephen, Deluzio, Caldwell, & Dunbar, 2008; Deluzio & Astephen, 2007; Harris et al., 2008; Kaufman, Hughes, Morrey, Morrey, & An, 2001; Ko, Ling, Schreiber, Nesbit, & Ferrucci, 2011; Landry, McKean, Hubley-Kozey, Stanish, & Deluzio, 2007; Messier, Loeser, Hoover, Semble, & Wise, 1992 as cited in Al-Zahrani & Bakheit, 2002). In addition, knee OA also affects the spatiotemporal parameters of gait.

#### 2.6.2.2 Spatiotemporal Parameters

The simplest and most researched gait parameter in knee OA – gait speed – has been shown to be slower than that of healthy control groups (Al-Zahrani & Bakheit, 2002; Astephen et al., 2008; Deluzio & Astephen, 2007; Harris et al., 2008; Kaufman et al., 2001; Ko et al., 2011; Mandeville et al., 2008; Yakhdani et al., 2010). There is also evidence to suggest that gait speed progressively decreases as the disease severity increases (Astephen et al., 2008). A decreased stride length, reduced cadence, prolonged stance time, and increased stride time have also been shown for individuals experiencing knee OA (Al-Zahrani & Bakheit, 2002; Astephen et al., 2008; Harris et al., 2008; Ko et

al., 2011; Mandeville et al., 2008), and Astephen et al. (2008) demonstrated that stride time and stance time further increased with OA severity. Similar to the inevitable aging effects on gait, these changes in spatiotemporal parameters can be seen as compensatory strategies to improve dynamic stability and balance during walking.

#### 2.6.2.3 Falls

Combined with the intensified gait impairments, knee OA has been associated with a significantly greater incidence of falls (Sturnieks et al., 2004), and considered as an important risk factor for an increased risk of falling (Campbell, Borrie, & Spears, 1989; Levinger et al., 2011). The decreased strength in knee extension, proprioceptive deficits and standing balance have been suggested to be important physical reasons for the increase in falls caused by the disease (Blake et al., 1988; Campbell et al., 1989; Levinger et al., 2011; Sturnieks et al., 2004). For those falling with knee OA, further injuries may ensue, such as hip and other nonvertebral fractures (Arden et al., 2006).

#### 2.6.2.4 Gait Variability

In recent years, researchers have begun to examine the relationship between lower-limb arthritic conditions and gait variability. Callisaya et al. (2010) examined variability in step time, step length, step width, and double support time for those with a self-reported medical history of chronic pathologies (e.g. hypertension, diabetes, stroke and arthritis). Using the GAITRite™ system, the authors found that a self-reported history of lower-limb arthritis was the only condition associated with greater gait variability (i.e. step time variability) at a preferred walking speed. These findings may be due to increased levels of pain or decreased muscular strength in the arthritic group that interfered with step timing. Kiss (2010a, 2010b, 2011) has taken a more precise approach

to lower-limb arthritis and gait variability by studying individuals with hip and knee OA. Compared to a healthy, age-matched control group, higher variability values (SD) in cadence, step length, step width, duration of support phase, and double support time were found for those experiencing unilateral hip OA (Kiss, 2010a; 2010b). In the study by Kiss (2010a), walking speed influenced the degree of variability for those with hip OA on the affected side, where speeds higher or lower than the participants' self-selected walking speed displayed greater SD values. Variability in cadence, double support time, and the step length, step width, and duration of the support phase of the affected hip also increased as severity (i.e. healthy, moderate, and severe) increased.

Kiss (2011) also examined the relationship between knee OA and gait variability in cadence, step length, step width, stance phase and double support time using a treadmill and an ultrasound-based motion analysis system. Compared to the healthy, age-matched control group, subjects with moderate (K-L grade 3) and severe (K-L grade 4) unilateral knee OA demonstrated significantly higher variability values (CoV) in affected-side step length and stance time, as well as cadence and the duration of the double-support phase ( $p \leq 0.03$ ). These differences in variability were significantly higher when the walking speed differed from the participants' self-selected speed ( $p \leq 0.04$ ). Sex was also a factor for some gait variability measures, such that females with moderate knee OA demonstrated higher CoV values in cadence, and women with severe knee OA had greater step length variability. Since being female is considered to be a significant risk factor for knee OA, these results may be of importance to understand any additional implications in gait disturbances for females with knee OA.

Furthermore, the degenerative nature of knee OA can affect gait variability over time. DeCaria, Petrella, Wolfe, Chesworth, and Montero-Odasso (2011) found a significant increase in stride time variability over a six-month period ( $CoV = 2.0 \pm 0.8\%$  to  $2.6 \pm 0.8\%$ ;  $p = .04$ ) for individuals suffering from knee OA. These findings are consistent with the results of Kiss (2011) in which significant increases in gait variability were found between moderate and severe unilateral knee OA groups. As a result, interventions for knee OA early in the diagnosis may help to delay these progressive gait impairments.

Although knee OA has shown a positive relationship with certain spatiotemporal parameters of gait variability, the research is minimal, and does not include the FSI or bilateral asymmetry index (ASI). As mentioned, the FSI can provide an understanding of the long-range, self-similar patterns of gait that can be useful for further understanding gait stability. The instruments that have been used for researching knee OA and gait variability may, in some instances, be impractical for clinical purposes. Tri-axial accelerometry has been shown to provide many benefits to research, and may be a better method for the clinical assessment of gait variability. As a result, this research project was designed to assess gait variability and the FSI with a tri-axial accelerometer to provide a more comprehensive picture of knee OA and gait variability that can be used for future clinical intervention.

## CHAPTER 3: Methods

### 3.1 Study Design

A matching-only posttest-only control group design was used for this research project. The sample ( $N = 34$ ) consisted of two groups. There was one experimental group ( $n = 17$ ), which consisted of subjects with bilateral knee OA. The second group was a control group ( $n = 17$ ), which consisted of healthy older adults matched according to age (i.e., no significant difference between age) and sex (i.e., 10 females and 7 males).

On the day of the experiment, the participants were informed of the aims of the study and any known associated risks, followed by completion of a participatory written consent form as approved by the Regina Qu'Appelle Health Region and University of Regina Ethics Review Board. Height and mass were then measured to calculate body mass index (BMI). The Knee Injury and Osteoarthritis Outcome Score (KOOS) questionnaire was also administered to all participants. The KOOS is a reliable and valid health-status instrument for the assessment of pain, stiffness, difficulty in activities of daily living, and difficulty with exercise and knee-related quality of life (Collins, Misra, Felson, Crossley, & Roos, 2011). This questionnaire was administered to further identify and quantify the presence of knee OA and provide another comparison variable between the two groups.

Both groups participated in an identical experimental protocol – a ten-minute walking test at a self-selected speed. A tri-axial accelerometer was placed on the L3 spinous process of the lower back of each participant to collect the vertical, mediolateral, and anteroposterior accelerations associated with gait, and the accelerometer's data was imported into a MATLAB program to analyze each subject's stride times and step times.

In addition to the mean stride time, left step time and right step time, two additional measures were used to determine each participant's gait variability: 1) the SD of the mean stride time and combined step time, and 2) the FSI (*α scaling component*) of the entire series of stride times. The degree of bilateral asymmetry for step time and step time SD within each group was also calculated using an asymmetry index (ASI).

### 3.2 Participants

Using G\*Power (Faul, Erdfelder, Lang, & Buchner, 2007), an *a priori* power analysis was used to estimate the necessary sample size for the desired statistical power of 0.8. The *a priori* power analysis computed a sample size of 50 to achieve the desired statistical power. Participants were recruited via contact with city of Regina medical clinics (e.g., the Aspen Medical Centre), orthopedic surgeons, the University of Regina Centre on Aging & Health participant directory, email, and poster recruitment strategies. Thirty-four older adults were eventually included in the study, and were placed into either the experimental or control groups based on the inclusion and exclusion criteria. The experimental group ( $n = 17$ ) consisted of patients who had received a diagnosis of knee OA. Inclusion criteria consisted of an age equal to or greater than 55 years, a medical diagnosis of knee OA, and the ability to walk for ten minutes without the use of an assistive device or experiencing excessive pain that would limit or prohibit the assessment of gait. Exclusion criteria included any recent history of surgery affecting the legs or lumbar spine, OA in any other joint of the lower extremity, any other neuromuscular disorders, history of stroke, cardiovascular disease, or any other medical condition or physical impairment (e.g., diabetes, recent/chronic head injuries, low blood pressure, vestibular conditions, and inner ear problems) that would affect their gait,

balance, and/or ability to walk at a steady pace for ten minutes. Given that all of the experimental group participants had bilateral knee OA, their lower limbs were classified as either ‘more affected’ or ‘less affected’. This classification was done subjectively by asking each participant about the severity of his or her knee OA symptoms. Based on the participants’ radiographic assessments of the knees, the K-L grading scale was used by a healthcare professional to describe the knee OA severity level for the experimental group. The healthy, age- and sex-matched control group ( $n = 17$ ) consisted of healthy older adults that had no limitations in terms of their walking ability. For this group, the inclusion and exclusion criteria were the same as the experimental group, aside from the presence of knee OA. The control group’s lower limbs were classified as either ‘dominant’ or ‘non-dominant’ by subjectively asking each participant which leg was their dominant leg.

### 3.3 Gait Analysis

Each participant completed an identical walking protocol. The participants walked around an indoor 200-metre oval track for a total of ten minutes. The researcher timed the walk and notified the participants when the ten-minute walk was completed. Each individual was instructed to walk at his or her self-selected walking speed in a consistent manner. A self-selected walking speed was chosen to more accurately represent the typical gait pattern and rhythm of each participant according to their stature and other physical factors such as strength and flexibility. Speeds above or below a person’s self-selected speed have been shown to affect measures of gait variability (Hausdorff et al., 1996; Jordan et al., 2006; Kiss, 2010a; Kiss, 2011); therefore, a self-

selected speed was chosen to eliminate the unwanted effects that controlled speeds could have had on the outcome measurements.

A GENEActiv device (see Figure 3) was used to collect vertical (y-axis), mediolateral (x-axis), and anteroposterior (z-axis) acceleration data during the walking trial. The GENEActiv device is a reliable body-worn accelerometer that measures and tracks everyday living in all environments (GENEActiv, 2012). The size of the device is 43mm \* 40mm \* 13mm, and weighs 16g. A 100 Hz sampling frequency was used for the study, which is an accepted frequency for determining measures associated with gait variability (Hartmann et al., 2009a; O'Sullivan, Blake, Cunningham, Boyle, & Finucane, 2009). The device was attached to a belt and worn firmly on the lower back near the body's centre of mass (i.e., the L3 vertebra) of each participant for the walking protocol. To identify the starting point of the data collection period, the participant paused and stood still for approximately five seconds before the trial began. Following the walking trial, the GENEActiv device was stopped, and data was imported into the GENEActiv software program for data analysis.

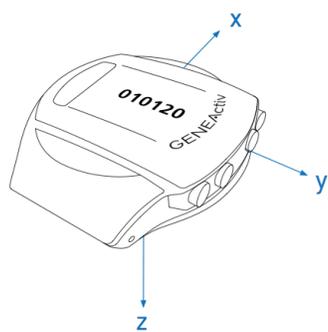


Figure 3. GENEActiv Device

### 3.4 Data Analysis

The vertical, mediolateral, and anteroposterior acceleration data were extracted from the device using the GENEActiv software. To calculate valid measurements of gait variability and the FSI, the length of the walking trial is very important. Hausdorff et al. (1997b) suggested a minimum of five minutes of data to allow for true testing of the long-range correlation measurements, and a further study has shown high levels of reliability for an eight-minute walking trial (Pierrynowski et al., 2005). Generally, there is a positive relationship with the length of the walking trial and the accuracy of the FSI (Hausdorff, 2007); therefore, a total of nine minutes of data was used for this analysis. It has also been reported that steady-state walking speeds can be achieved in 2.5 metres for older adults (Lindemann et al., 2008). As a result, the first 15 seconds of data were excluded to make sure the participants achieved a steady-state walking speed, and accurate measurements of variability and fractal-scaling coefficients could be determined.

Custom MATLAB (MATLAB, Natick, MA) software routines, developed in a previous student's thesis project (Kobsar et al., 2014), were used to filter and detect the pattern of peak accelerations in the anteroposterior direction, which corresponded to the heel strikes of each foot (see Figure 4). A zero lag, 4<sup>th</sup> order Butterworth low-pass filter with a cutoff frequency of 10 Hz was applied to the raw acceleration data. The peak negative anteroposterior acceleration values were used to compute the step and stride times (i.e., time between peaks) for each leg throughout the walking trial. This method has been shown to provide better results than using the peak positive anteroposterior or vertical accelerations (Kobsar & Barden, 2012). The mediolateral acceleration data was

used to identify which leg (left or right) was associated with the corresponding filtered anteroposterior acceleration peak (see Figure 5).

Figure 4. Anteroposterior Acceleration

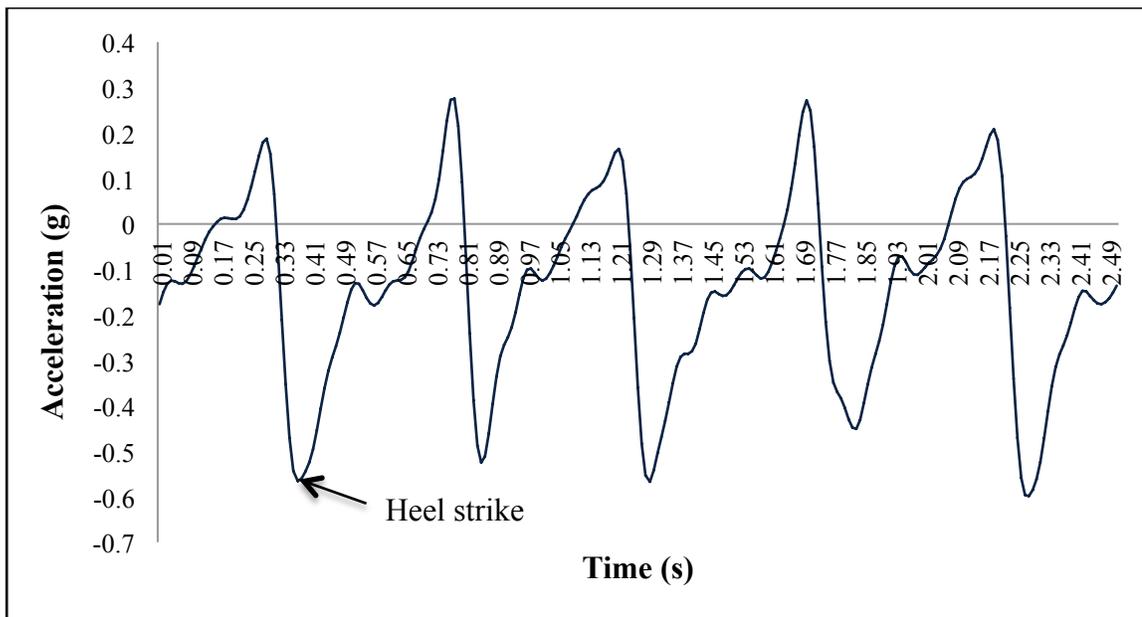
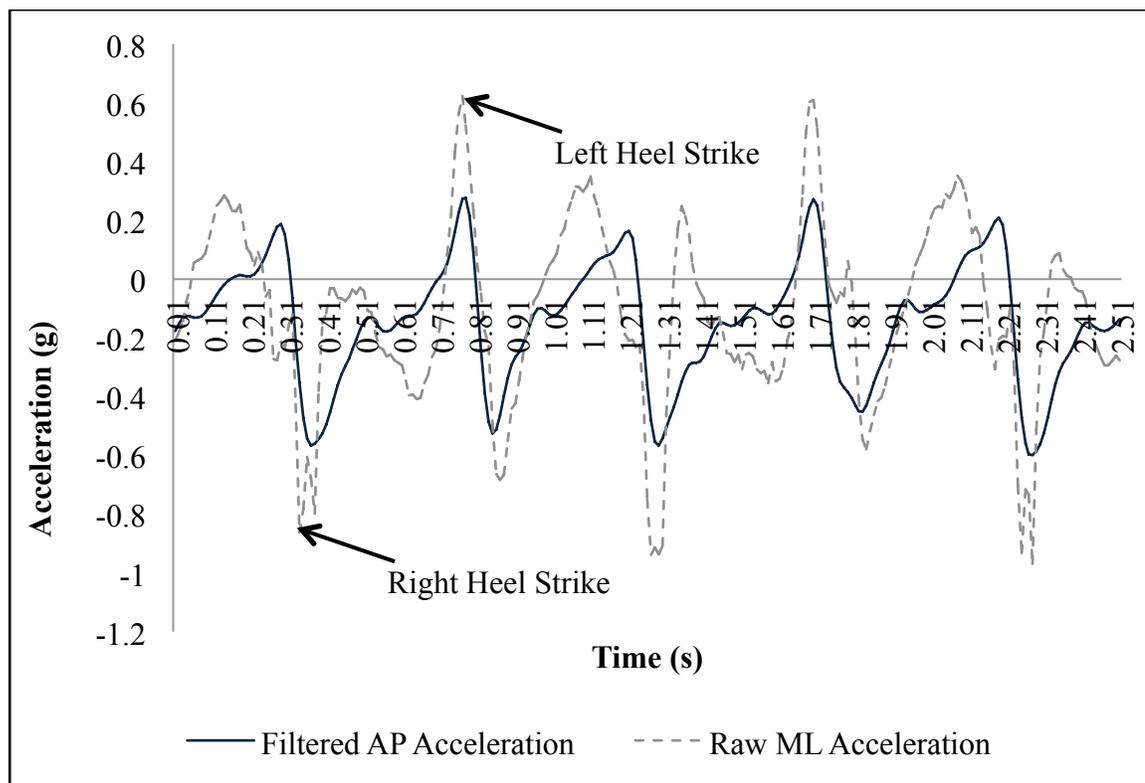


Figure 5. Mediolateral Acceleration



Using MATLAB, individual left and right average step times were calculated and then classified based on limb category (i.e., dominant, non-dominant, more affected, and less affected). Individual left and right step times were added together to identify the mean stride times for each participant. The SD for each temporal parameter (i.e., step time and stride time) was calculated to quantify the level of gait variability. A detrended fluctuation analysis was then used to determine the FSI ( $\alpha$ ) of stride time.

The degree of bilateral asymmetry between limb types for step time and step time SD in each group was calculated using an asymmetry index (ASI) calculation (see Equation 4 and Equations 9 to 12). This measure of gait asymmetry was proposed by Robinson, Herzog and Nigg (1987), and has been used in other studies to assess gait asymmetry (Kim & Eng, 2003; Patterson, Gage, Brooks, Black, & McIlroy, 2010). The magnitude of the ASI indicates the degree of asymmetry and the sign reflects the pattern of asymmetry (left side greater or right side greater). An ASI equal to zero indicates perfect symmetry.

$$\text{Knee OA Step Time ASI \%} = \left( \frac{\text{Step Time}_{\text{More Affected}} - \text{Step Time}_{\text{Less Affected}}}{0.5 * (\text{Step Time}_{\text{More Affected}} + \text{Step Time}_{\text{Less Affected}})} \right) * 100 \quad (9)$$

$$\text{Control Step Time ASI \%} = \left( \frac{\text{Step Time}_{\text{Non-Dominant}} - \text{Step Time}_{\text{Dominant}}}{0.5 * (\text{Step Time}_{\text{Non-Dominant}} + \text{Step Time}_{\text{Dominant}})} \right) * 100 \quad (10)$$

$$\text{Knee OA Step Time SD ASI \%} = \left( \frac{\text{Step Time SD}_{\text{More Affected}} - \text{Step Time SD}_{\text{Less Affected}}}{0.5 * (\text{Step Time SD}_{\text{More Affected}} + \text{Step Time SD}_{\text{Less Affected}})} \right) * 100 \quad (11)$$

$$\text{Control Step Time SD ASI \%} = \left( \frac{\text{Step Time SD}_{\text{Non-Dominant}} - \text{Step Time SD}_{\text{Dominant}}}{0.5 * (\text{Step Time SD}_{\text{Non-Dominant}} + \text{Step Time SD}_{\text{Dominant}})} \right) * 100 \quad (12)$$

### 3.5 Statistical Analysis

All statistical analyses were performed using PASW Statistics 18 (SPSS Inc., Chicago, IL). First, an independent t-test was used to compare subject demographics (i.e., age, height, mass, and BMI), KOOS score, and average speed between groups. Then, an independent t-test was also used to compare the gait characteristics (i.e., mean stride time, stride time SD, stride time FSI, mean combined step time, step time ASI, and step time SD ASI) between groups. The significance level ( $\alpha$ ) was set at 0.05 for all statistical analyses.

## CHAPTER 4: Results

Demographics, KOOS score, and speed comparisons between groups are shown in Table 1. Individual subject characteristics for the control and experimental groups can be seen in Tables 2 and 3, respectively. The two groups were matched for age, which is demonstrated by the non-significant difference between groups. The knee OA group had a greater mass, BMI, and KOOS score. Also, the knee OA group's self-selected walking speed was significantly slower than the control group.

Table 1

*Subject Demographics, KOOS Score, and Speed Comparisons between Groups*

Parameter	Control Group (Mean $\pm$ SD)	Knee OA Group (Mean $\pm$ SD)	<i>p</i>
Age (years)	65.76 $\pm$ 9.51	64.53 $\pm$ 6.33	.659
Height (cm)	166.79 $\pm$ 10.55	166.45 $\pm$ 9.29	.922
Mass (kg)	73.27 $\pm$ 16.21	84.09 $\pm$ 11.00*	.030
BMI (kg/m <sup>2</sup> )	26.12 $\pm$ 3.91	30.43 $\pm$ 3.87*	.003
KOOS	96.00 $\pm$ 8.33	54.41 $\pm$ 16.42*	.000
K-L Grade	---	3.06 $\pm$ 0.75	---
Speed (m/s)	1.45 $\pm$ 0.22	1.31 $\pm$ 0.15*	.035

*Note.* Significant differences between groups ( $p < 0.05$ ) are marked with an asterisk (\*).

Table 2

*Control Group's Subject Characteristics*

Subject Code	Age	Sex	BMI	Dominant Knee
C01	77	Female	32.6	Left
C02	67	Female	22.0	Right
C03	59	Male	24.2	Right
C04	55	Female	20.3	Right
C05	57	Male	26.6	Right
C06	79	Female	22.0	Right
C07	74	Male	28.8	Right
C08	56	Female	25.4	Right
C09	56	Female	24.1	Right
C10	58	Male	25.6	Left
C11	67	Female	31.1	Right
C12	55	Male	33.5	Right
C13	65	Female	26.8	Right
C14	75	Male	23.9	Right
C15	79	Female	22.2	Right
C16	60	Female	30.2	Right
C17	79	Male	24.8	Right

Table 3

*Experimental Group's Subject Characteristics*

Subject Code	Age	Sex	BMI	More Affected Knee	KOOS	K-L Grade
OA01	66	Female	25.1	Right	53	3
OA02	62	Male	30.2	Right	56	3
OA03	60	Male	31.3	Left	47	4
OA04	56	Female	34.6	Right	16	4
OA05	58	Female	34.5	Right	37	4
OA06	70	Female	27.9	Left	64	3
OA07	59	Female	36.8	Left	43	3
OA08	68	Male	27.7	Right	67	4
OA09	82	Female	25.6	Left	60	2
OA10	63	Male	28.2	Left	33	4
OA11	58	Female	30.4	Left	50	3
OA12	64	Female	32.3	Left	80	2
OA13	71	Male	32.1	Left	59	2
OA14	63	Female	26.5	Left	79	2
OA15	66	Male	38.0	Right	64	3
OA16	69	Male	26.7	Left	49	3
OA17	62	Female	29.4	Right	68	3

*Note.* The KOOS score and Kellgren-Lawrence (K-L) grade represents the 'more affected' knee.

On average, participants with knee OA had a significantly greater stride time and step time than the healthy control group. However, there were no significant differences for stride time SD, stride time FSI, combined step time SD, step time ASI, or step time SD ASI between groups, as summarized in Table 4.

Table 4

*Gait Characteristic Comparisons between Groups*

Parameter	Control Group (Mean $\pm$ SD)	Knee OA Group (Mean $\pm$ SD)	<i>p</i>
Stride Time (ms)	1006 $\pm$ 69.95	1050.6 $\pm$ 63.39*	.036
Stride Time SD (ms)	31.62 $\pm$ 13.30	36.81 $\pm$ 11.30	.230
Stride Time FSI ( $\alpha$ )	0.78 $\pm$ 0.17	0.76 $\pm$ 0.14	.689
Step Time (ms)	499.94 $\pm$ 34.86	526.14 $\pm$ 31.35*	.027
Step Time SD (ms)	26.25 $\pm$ 12.57	30.97 $\pm$ 13.41	.298
Step Time ASI (%)	-2.69 $\pm$ 15.54	-0.38 $\pm$ 11.92	.631
Step Time SD ASI (%)	-1.62 $\pm$ 6.14	-3.21 $\pm$ 7.15	.492

*Note.* Significant differences between groups ( $p < 0.05$ ) are marked with an asterisk (\*).

Since speed has been shown to have an effect on gait variability (Hausdorff et al., 1996; Jordan et al., 2006; Kiss, 2010a; Kiss, 2011), and speed was significantly different between groups, separate ANCOVAs were used with speed as a covariate to determine whether there were any differences in the parameters between groups when controlling for speed. Speed was significantly related to stride time, stride time SD, stride time FSI, step time, and step time SD ( $p < .05$ ). However, there were no significant differences

between groups for the dependent variables after controlling for the effect of speed ( $p > .05$ ).

## CHAPTER 5: Discussion

The overall purpose of this study was to investigate the relationship between knee OA and gait variability in a group of older adults using tri-axial accelerometry, and to compare these results to a healthy, age- and sex-matched control group. It was hypothesized that (i) gait variability (i.e., stride time SD and combined step time SD) would be significantly greater in the knee OA group compared to the healthy older adult group, while the stride time FSI ( $\alpha$ ) would be less. It was also hypothesized that (ii) the bilateral asymmetry between limbs for step time SD in the knee OA group (i.e., ‘more affected’ vs ‘less affected’) would be greater than the difference between ‘non-dominant’ and ‘dominant’ legs in the healthy control group. The primary findings of the study were: (i) there were no significant differences between groups for gait variability parameters or stride time FSI; and (ii) the bilateral asymmetry for step time SD was slightly greater in the knee OA group than the control group, but the difference was not statistically significant.

### 5.1 Participant Demographics

The knee OA group was significantly heavier ( $84.09 \pm 11.00$  kg) and had a greater BMI ( $30.43 \pm 3.87$  kg/m<sup>2</sup>) than those with healthy asymptomatic knees ( $73.27 \pm 16.21$  kg;  $26.12 \pm 3.91$  kg/m<sup>2</sup>). Based on the World Health Organization’s International Classification of adult underweight, overweight and obesity according to BMI, the experimental group had eight participants who were considered ‘overweight’ and nine that were considered ‘obese’. In the control group, eight participants were considered ‘normal’ weight, five were considered ‘overweight’, and four were considered ‘obese’. It has been suggested that obesity alters gait mechanics (Aststephen et al., 2008); however, it

is difficult to control BMI in gait studies with knee OA participants since obesity is a known risk factor for the disease (Uhlig et al., 2010; Jiang et al., 2012). Therefore, within the scope of this research design, the significant difference between groups is not surprising and consistent with the literature that has similar participants (Asthephen et al., 2008; Zeni & Higginson, 2009; Kiss, 2011).

The KOOS questionnaire was administered as a comparison measurement to further identify the presence of knee OA in the experimental group and the absence of knee OA in the control group. The significant difference in KOOS scores found between the two groups was expected since the absence of knee OA was part of the inclusion criteria for the healthy control group. The questionnaire was also administered to potentially identify any relationships between the gait variability parameters and KOOS scores. Had there been any significant differences between groups for stride time SD, step time SD, and stride time FSI, it would have been interesting to see whether there was an inverse relationship between KOOS scores and gait variability. Future studies may want to explore this relationship further.

## 5.2 Gait Characteristics

### 5.2.1 Spatiotemporal Parameters

Spatiotemporal parameters were first compared to assess the gait characteristics between groups. In terms of self-selected gait speed, the knee OA group walked significantly slower ( $1.31 \pm 0.15$  m/s) than the healthy control group ( $1.45 \pm 0.22$  m/s), which is consistent with the literature (Al-Zahrani & Bakheit, 2002; Asthephen et al., 2008; Gök, Ergin, & Yavuzer, 2002; Deluzio & Asthephen, 2007; Kaufman et al., 2001; Ko et al., 2011; Mandeville et al., 2008; Zeni & Higginson, 2010; Zeni & Higginson,

2011). The knee OA group also had a significantly greater mean stride time and mean combined step time ( $1050.6 \pm 63.39$  ms; and  $526.14 \pm 31.35$  ms, respectively) than the healthy control group ( $1006 \pm 69.95$  ms; and  $499.94 \pm 34.86$  ms). These gait alterations in participants with knee OA have been attributed to knee pain, impaired knee proprioception, and may be viewed as extra compensatory strategies to reduce joint loading and maintain walking security and balance. A significantly increased stride time in participants with knee OA has been consistently reported in the literature (Mills, Hunt, & Ferber, 2013), but, to our knowledge, there is no evidence of any research that investigated differences in combined step time, specifically. As a result, this study extends previous work by demonstrating that mean step time can be used as another temporal parameter during gait analysis to distinguish between individuals with and without knee OA.

### 5.2.2 Stride-to-Stride Fluctuations

The first objective of this study was to determine whether there was a difference in gait variability between patients with knee OA and older adults with healthy knees. It was hypothesized that stride time SD and combined step time SD would be greater in the knee OA group. Although stride time SD and combined step time SD were both greater in the knee OA group, these differences were not statistically significant.

The lack of significant differences between groups may have been attributed to the small sample size in the study and the heterogeneity of the experimental group. Post-hoc power analyses indicated statistical power values of 0.33 for stride time SD and 0.27 for combined step time SD, or a 33% and 27% chance, respectively, of detecting a difference if one genuinely exists. According to Cohen (1992), any statistical power

value smaller than 0.8 would suggest the high possibility of a Type II statistical error. Differences in knee OA severity (i.e., K-L grade) within the experimental group may also have impacted the results by increasing the heterogeneity in the group, thus violating the homogeneity of variance assumptions of the statistical test. The study's inclusion criteria for the knee OA group did not specify a severity level with respect to the radiographic assessments. Given that this is the first study to assess gait variability and FSI in stride time and step time for patients with knee OA, it was important to initially determine whether a relationship existed between increased stride and step time variability and knee OA, regardless of severity. Within the experimental group, the K-L grades ranged from 2-4 with a mean of  $3.06 \pm 0.75$  (i.e., moderately severe). The literature suggests that knee OA severity levels may have a positive relationship with gait variability. DeCaria and colleagues (2011) found that stride time variability increased over time due to the degenerative nature of knee OA, and Kiss (2011) also found significant increases in select gait variability parameters between the moderate and severe knee OA groups. In addition to increasing the heterogeneity of the knee OA group, the participants with lower severity levels (i.e., K-L grade 2) may have compromised the results. However, if the study were limited to participants with severe knee OA (i.e., K-L grade 4), then all participants might not have been able to complete the walking protocol (i.e., walk for ten minutes without the use of an assistive device or experiencing excessive pain that would limit or prohibit the assessment).

Kiss (2011) found a relationship between knee OA and gait variability in step length, stance time, cadence, and double-support time; however, Moe-Nilssen, Aaslund, Hodt-Billinton, and Helbostad (2010) found that certain parameters for gait variability do

not correlate (i.e., step time vs. step length) and likely represent different constructs. It has been suggested that stance, step, and stride time variability represent rhythmicity and central nervous system impairment, whereas parameters such as step width variability and double-support time variability may be a better reflection of dynamic balance control and sensory impairment (Gabell & Nayak, 1984; Brach, Studenski, Perera, VanSwearingen, & Newman, 2008; Lord et al., 2011a). Given that knee OA affects the physical characteristics associated with changes in gait and has been shown to detrimentally affect dynamic balance compared to healthy older adults (Hinman, Bennell, Metcalf, & Crossley, 2002; Mohammadi, Taghizadeh, Ghaffarinejad, Khorrami, & Sobhani, 2008), spatiotemporal parameters corresponding to dynamic balance control may be more suitable for analyzing the relationship between knee OA and gait variability.

It was also expected that the experimental group's stride time FSI would be significantly less than the healthy control group. The stride time FSI, indicated by the fractal scaling component ( $\alpha$ ), shows whether the stride time variability occurring in a time series of data demonstrates a random pattern or a self-similar fractal pattern across different time scales. A smaller  $\alpha$  value indicates a more uncorrelated or random stride pattern. The knee OA group did display a slightly lower average stride time FSI ( $0.76 \pm 0.14$ ) than the control group ( $0.78 \pm 0.17$ ), but this comparison was not statistically significant ( $p = .689$ ). The FSI values found in this study show that the participants demonstrated long-range self-similar correlations (i.e.,  $0.5 < \alpha \leq 1.0$ ). The average FSI value for the healthy older adult group in this study is comparable to a previous study by Kobsar, Olson, Paranjape, Hadjistavropoulos and Barden (2014) which used a similar

walking protocol ( $\alpha = 0.76 \pm .13$ ,  $M_{AGE} = 76 \pm 5$  years) as well as a study by Gates and Dingwell (2007) ( $\alpha = 0.88 \pm .106$ ,  $M_{AGE} = 57.6 \pm 7.7$  years). Herman and colleagues (2005) found a slightly higher stride time FSI value than the current study for their healthy older adults ( $\alpha = 0.88 \pm 0.22$ ,  $M_{AGE} = 78.54 \pm 5.6$  years), and Hausdorff et al. (1997b) found a lower mean stride time FSI of  $0.68 \pm 0.14$  for their control group ( $M_{Age} = 75.7 \pm 3.2$  years). However, these studies only tested the participants during two-minute and six-minute walking trials, respectively, which may not be enough data to accurately represent the true fractal scaling patterns of healthy older adults. To our knowledge, this is the first study to examine the relationship between knee OA and stride time FSI. As a result, there are no known FSI values for knee OA patients to which this study's findings can be compared.

It has been suggested that a decreased stride time FSI may be linked to a loss of muscular strength, function, and range of motion, but it may be a better reflection of centrally-mediated motor control and the neuromuscular system's ability to regulate and maintain a steady walking pattern (Hausdorff, 2007; Montero-Odasso et al., 2011). Research shows that gait variability and FSI parameters are affected by several neurodegenerative diseases – Parkinson's, Huntington's, and Alzheimer's disease (Blin, Ferrandez, & Serratrice, 1990 as cited in Hausdorff, 2007; Hausdorff, 1997b; Hausdorff et al., 1998; Sheridan, Solomont, Kowal, & Hausdorff, 2003; Hausdorff, 2009), but until now, research on the effect of joint pathology has not been investigated.

Gates and Dingwell (2007) compared the stride time FSI between diabetic patients with significant peripheral neuropathy and control subjects with no history of diabetes or neuropathic illness. They found no significant differences between groups for

stride time FSI. The authors suggest that their findings support the notion that the central nervous system is the primary source of long-range correlations in gait, and individuals with peripheral deficits (i.e., diabetic neuropathy) may be able to accommodate and adapt their gait patterns with respect to their peripheral sensory loss (Gates & Dingwell, 2007). These accommodations and adaptations may also be seen in people with knee OA, a degenerative peripheral joint disease, which may explain the non-significant differences between groups found in this study. Nevertheless, this study is one of the few that has used tri-axial accelerometry to detect stride time variability and FSI in older adults (Paterson, Hill, & Lythgo, 2011; Kobsar et al., 2014). As such, it adds to the growing body of literature that demonstrates the use of accelerometry as an effective method to clinically assess gait variability and FSI in older adults with and without knee OA.

### 5.2.3 Bilateral Asymmetry

Differences in gait characteristics between more affected and less affected limbs with knee OA may indicate that asymmetric gait is common with the disease whereas asymptomatic individuals experience a more symmetrical gait pattern (Creaby, Bennell, & Hunt, 2012). In particular, Creaby and colleagues (2012) found that knee OA patients with unilateral knee OA or bilateral structural disease and unilateral pain have been shown to have between-limb asymmetries with respect to knee flexion angles and moments. The authors suggest that their findings support the idea that pain is associated with gait disturbances rather than the structural aspects of OA. Furthermore, Mills, Hettinga, Pohl, and Ferber (2013) found between-limb kinematic asymmetries to be more prevalent in mild to moderate bilateral knee OA participants compared to unilateral

knee OA and a healthy control group. This is an interesting finding, as it was expected that unilateral knee OA would have had a greater degree of bilateral asymmetry. In fact, the participants with unilateral mild to moderate knee OA had relatively similar gait patterns and joint angles to the control group with respect to bilateral asymmetry. These findings may further support the important relevance of knee OA progression and severity, where unilateral knee OA participants with mild to moderate severity levels may be able to maintain kinematic symmetry for a longer period of time in the knee OA progression compared to those with bilateral knee OA (Mills et al., 2013a).

Gait variability can characterize a dimension of gait that is often distinct from traditional spatiotemporal parameters, and this may also be true for bilateral asymmetry. Prior to this study, gait variability parameters had not been used to assess bilateral asymmetry in knee OA patients, but they have been used to identify bilateral asymmetry in other pathological conditions (e.g., post-stroke patients with unilateral weakness due to hemiparesis) (Balasubramanian, Neptune, & Kautz, 2009). For between-leg comparisons, Balasubramanian, Neptune, and Kautz (2009) found both swing time variability and pre-swing time variability measurements (i.e., CoV) to be greater in the paretic leg compared to the non-paretic leg. These differences also had a positive relationship with severity.

In our study, participant classification for the bilateral knee OA group was similar to the methodology used by Creaby et al. (2012) and Mills et al. (2013a) who found greater bilateral asymmetry in their bilateral knee OA group. Thus, it was expected that the difference in right and left mean step times and step time SDs, calculated with the ASI, would be indicative of bilateral asymmetry, and the differences

would be greater in the knee OA group compared to the healthy control group. However, there were no significant statistical differences in bilateral ASI for either step time or step time SD between groups.

Understanding gait asymmetry can be important in clinical settings due to its possible association with walking efficiency, balance control, and risk of musculoskeletal injury (Patterson et al., 2010). In this research project, the asymmetry index was used to explore bilateral asymmetry between limb types and groups, but there are also other methods used in research and clinical settings to quantify gait asymmetry. There are also no commonly accepted spatiotemporal parameters to assess gait asymmetry in the literature. Many different spatiotemporal parameters have been used to assess gait asymmetry and it's probable that each parameter has a different meaning. While studying hemiparetic post-stroke patients, Patterson and colleagues (2010) suggested that: (i) double-support time asymmetry may be a good reflection of postural control demands during gait; (ii) swing time asymmetry may indicate “insufficient power generated to swing the paretic limb quickly or increased time required for foot placement”, and (iii) stance time asymmetry may reflect balance issues (p. 244).

The knee OA participants in our study were all clinically diagnosed as patients with bilateral knee OA, which was likely responsible for the low level of asymmetry in the group. A unilateral knee OA group would have likely had a greater difference in step time and step time SD between legs, presenting a greater degree of bilateral asymmetry in the group. Mills et al. (2013a) suggested a greater level of bilateral asymmetry in their bilateral mild to moderate knee OA group, but the heterogeneity of our experimental group with respect to knee OA severity may have affected the results. Future research

should identify a specific and/or different knee OA severity level to further understand the bilateral asymmetry measurements with respect to gait variability. Other parameters (e.g. swing time, stance time, and double support time) and methods (e.g., symmetry ratio, gait asymmetry, and the symmetry angle) may also provide further insight into the relationship between gait variability and bilateral gait asymmetry in knee OA patients. Future research should compare step time SDs as well as other gait variability parameters with different asymmetry methodologies in both unilateral and bilateral knee OA patients to gain a better understanding of gait variability asymmetry in this population.

### 5.3 Study Limitations

A limitation to this study is that there were no radiographic assessments for knee OA recorded for the healthy control group participants. This means there is a possibility that some older adults in the healthy control group may have had some unknown structural deterioration associated with knee OA. The KOOS questionnaire was used as a comparison measure between groups, and although the normalized scores were very high for the healthy control group (i.e.,  $96 \pm 8.33$  out of 100), this does not necessarily preclude the absence of knee OA and any subsequent effect it may have had on gait variability. Another limitation is the possible influence of medication in the experimental group. Although medication was recorded, it was not accounted for in the exclusion criteria of the study. The most common type of medication for the knee OA group (i.e., for eight participants) was the pain reliever acetaminophen (i.e., **TYLENOL®** Arthritis). It would have been unethical to prevent these participants from taking their pain medication; however, research suggests that gait adaptations may be compensatory

mechanisms in response to pain, which may help to explain the lack of between-group significant differences in gait variability parameters found in the current study.

## CHAPTER 6: Conclusion

To our knowledge, this is the first study to investigate the relationship between knee OA and the temporal aspects of stride pattern variability. This study also demonstrates the effectiveness of a tri-axial accelerometer as a non-invasive measurement device to analyze gait variability in older adults with and without bilateral knee OA. Although the differences in this study were not statistically significant, knee OA was associated with increased gait variability (i.e., stride time SD and step time SD), decreased stride time FSI and increased bilateral asymmetry of gait variability compared to age and sex-matched older adults with healthy knees. Further studies should (i) investigate these variables with a larger sample size, (ii) explore the relationship between knee OA and the variability of other spatiotemporal parameters, particularly those associated with dynamic balance control, (iii) explore the effect of knee OA on gait variability with a controlled-speed walking trial, (iv) examine gait variability and fractal patterns of stride time in a knee OA group with a homogeneous severity level, (v) investigate gait variability and knee OA longitudinally to see the impact of disease development and progression, (vi) explore the relationship between gait variability and KOOS scores and the questionnaire's specific subcategories (i.e., symptoms, pain, activities of daily living, sport and recreation, and quality of life), and (vii) further investigate bilateral asymmetry associated with gait variability in patients diagnosed with bilateral and/or unilateral knee OA.

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