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THE INFLUENCE OF ASYMMETRY ON THE
METABOLIC COST OF LOCOMOTION

A Dissertation Submitted in Partial Fulfillment
of the Requirements for the Degree of
Doctor of Philosophy

Shane Patrick Murphy

College of Natural and Health Sciences
School of Sport and Exercise Science
Biomechanics

December 2019

This Dissertation by: Shane Patrick Murphy

Entitled: *The Influence of Asymmetry on the Metabolic Cost of Locomotion*

has been approved as meeting the requirement for the Degree of Doctor of Philosophy in College of Natural and Health Sciences in School of Sports and Exercise Science, Program of Exercise Science

Accepted by the Doctoral Committee

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Date of Dissertation Defense _____

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Research and Sponsored Projects

ABSTRACT

Murphy, Shane P., *The Influence of Asymmetry on the Metabolic Cost of Locomotion.* Published Doctoral Dissertation, University of Northern Colorado, 2019.

In this dissertation, the measurement and impact of asymmetrical locomotion were investigated. In the first study, ten able-bodied individuals were asked to run on a treadmill from which interlimb symmetries of joint level kinematics and kinetics were measured. To obtain a stable measure of interlimb symmetry, an average of 15 strides were needed. However, no differences were found between averages from bins of consecutive and inconsecutive strides. Further, no differences were noted between the average interlimb symmetry and interlimb symmetries calculated from the first, middle, or last, strides. Although there were differences between symmetry calculations, neither measure required a greater number of strides to become a stable measure of interlimb symmetry. In study two, ten able-bodied individuals were asked to walk on a treadmill from which interlimb symmetries of joint level kinematics and spatiotemporal parameters were calculated. The interlimb symmetries became stable with an average of 8 strides. No systematic differences between subsets of three, five, or eight strides were noted. Further, no differences were noted between subsets when utilizing consecutive or inconsecutive strides. Finally, although it required eight strides to achieve a stable mean symmetry index, no differences were noted between the average interlimb symmetry index of the first three, five, and eight strides for all measures. In study three, the metabolic cost of

walking asymmetrically was explored for ten able-bodied individuals. Walking with a unilaterally added 2kg mass at the ankle resulted in an increased metabolic cost of walking compared with normal walking. The asymmetrical swing times were calculated and replicated without the mass via an audible metronome that when matched to initial foot strikes resulted in asymmetrical swing times. This temporally asymmetrical swing time also resulted in an increased metabolic cost of walking compared with normal walking. Additionally, walking to a symmetrical metronome with the added mass increased the metabolic cost of walking. Forcing temporal symmetry when walking with a unilaterally added mass and forcing temporal asymmetry when walking without a unilaterally added mass were found to result in metabolic penalty compared with unmanipulated walking with and without a unilaterally added mass.

The findings of this dissertation indicate that 15 and 8 strides should be collected when studying interlimb symmetries during running and walking, respectively. However, whether the strides are collected consecutively or whether these strides are collected early or late within a trial does not appear to effect results. Further, there does not appear to be a statistical difference between the strides required to achieve a stable mean and fewer strides in able-bodied locomotion. Lastly, forcing an unnatural temporal gait pattern will result in a metabolic penalty during walking. Without interlimb mass differences, an asymmetrical gait pattern results in a greater metabolic cost of walking than a symmetrical gait pattern. More importantly for persons with a unilateral amputation; when interlimb mass differences are present, a symmetrical gait pattern results in a greater metabolic cost of walking than an asymmetrical gait pattern.

DEDICATION

In Memory of Clifford Junior Kinsey
(1929 – 2019)

To my grandfather –

As the keystone of our family and an inspiration to those around you, how truly exceptional of a life you lived. From your passion to your pragmatism, you were a sounding board for all of life's challenges. I could not have overcome this present one without your love and support. You will always be deeply missed.

Your grandson,

Shane Patrick Kinsey Murphy

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CHAPTER I

INTRODUCTION

Healthy human locomotion is generally characterized as a symmetrical coordination of the two lower extremities (Whitall & Caldwell, 1992). During walking and running the limbs are out of phase but rely on symmetrical movement patterns to support and propel the body forward (Inman, Ralston, & Todd, 1981). Historically, biomechanical gait analyses have relied on a single limb to characterize movement patterns of healthy individuals. Although this is generally still an acceptable method of analyzing healthy gait, any unilateral deviation from an individual's norm may result in interlimb differences of preferred movement patterns (Sadeghi, Allard, Prince, & Labelle, 2000). In turn, clinicians and researchers alike have used interlimb symmetry indices to help establish rehabilitative goals or quantify the magnitude of deviation from a given research condition (Carpes, Mota, & Faria, 2010; Czerniecki & Morgenroth, 2017; Herzog, Nigg, Read, & Olsson, 1989; Hoerzer, Federolf, Maurer, Baltich, & Nigg, 2015; Kumar et al., 2014; Nigg, Vienneau, Maurer, & Nigg, 2013; Robinson, Herzog, & Nigg, 1987).

Humans locomote symmetrically by incorporating both limbs to support and propel the body forward. This symmetry illustrates at a basic level how the body can adapt to a level of functional asymmetries and still walk or run (Sadeghi et al., 2000). Although measures of *global* symmetry are useful to quantify how the overall body

adapts to perturbations, measures of *local* asymmetries give insight to how joint level adaptations occur (Sadeghi, 2003).

   For discrete measures, interlimb symmetry during walking and running has been defined as a perfect agreeance between limbs, where there is zero percent difference between measures (Robinson et al., 1987). This symmetry index (SI) is widely used with a number of kinematic and kinetic measures to quantify the direction and magnitude of incongruence between limbs (Carpes et al., 2010; Nasirzade, Sadeghi, Mokhtarinia, & Rahimi, 2017; Sadeghi et al., 2000). The SI represents the difference between measures as a percentage, where the magnitude of asymmetry is relative to the other limb and the sign (positive or negative) of the value gives insight to the direction to which limb presented with greater asymmetry. There are a number of variations to the original proposed equation to improve on the limitations of dealing with some discrete measures (Cabral et al., 2016; Carpes et al., 2010). Although some measures are helpful in improving limitations, measures of symmetry may become less clinically intuitive (Kumar et al., 2014; Nigg et al., 2013; Zifchock, Davis, Higginson, & Royer, 2008).

   One alteration to the SI that still utilizes perfect agreeance between limbs but quantifies asymmetry magnitudes regardless of direction is the absolute symmetry index (ASI). The ASI is limited by removing the quantification of which limb is asymmetrical, but gives an improved indication of asymmetry magnitude by ignoring which limb is asymmetrical (Carpes et al., 2010). For example, the SI may indicate 0% asymmetry when equal and opposite asymmetries are present, where the ASI would indicate the absolute percentage and eliminate any possible negations across multiple strides.

Although the ASI eliminates this limitation, it is less applicable to populations such as unilateral amputees, where a consistent directional asymmetry is expected for a given measure.

As symmetry indices are widely used, it is important to best understand how changes in methodologies used during gait analyses may result in different findings (Kumar et al., 2014). Specifically, questions around the number of strides required, whether these strides need to be consecutive, and if there are any interaction with the specific symmetry index used are yet to be addressed explicitly within the literature. For example, three to five strides of inconsecutive strides during overground walking and running have commonly been analyzed (Carpes et al., 2010; Nasirzade et al., 2017). However, recent improvements to instrumented treadmills allow for the analysis of many more strides that can be collected consecutively. Beyond the known differences to walking and running between treadmill and overground locomotion, the difference in analyzing consecutive strides may affect the results of a study. These and other answers are explicitly important to those trying to better understand locomotion in populations where unilateral deviations that make locomotion more difficult, and in turn collecting fewer strides is advantageous.

In a healthy population, a certain level of local asymmetry has previously been quantified for walking and running in spatiotemporal, kinematic, and kinetic measures (Carpes et al., 2010; Nasirzade et al., 2017; Sadeghi et al., 2000). However, as normative values of asymmetry are measure-dependent, no universal threshold of asymmetry is established. Further, individual asymmetries may start to be overlooked when data is averaged over a larger sample (Ammann & Wyss, 2015). When asymmetries are present,

these asymmetrical movement patterns are not inherently detrimental and may give an indication to some functional level of adaptability (Ducharme et al., 2018; Haddad, van Emmerik, Whittlesey, & Hamill, 2006; Sadeghi, 2003; Xia, Ye, Gao, Lu, & Zhang, 2016). These asymmetries may also be present as a byproduct of individual limbs having distinct mechanical tasks to complete: propelling the body forward versus supporting the mass of the body (Sadeghi et al., 2000). This theory of laterality includes mechanical and neurophysiological adaptations that allows the body to acutely compensate, within strides, to unexpected gait asymmetries (Kozlowska, Latka, & West, 2017; Sadeghi et al., 2000). These theories of laterality and adaptability suggest that the presence of asymmetries may exceed just structural limitations and give insight to control mechanisms meant to overcome asymmetrical perturbations.

Endurance athletes, such as cyclists and runners, are commonly studied to better understand the effects of asymmetrical movement patterns, as they perform a large number of bilaterally symmetrical gait cycles (Carpes et al., 2010; Gilgen-Ammann, Taube, & Wyss, 2017). Asymmetries are commonly approached as a possible cause of musculoskeletal injuries or detrimental to performance (Bredeweg, Buist, & Kluitenberg, 2013; Gilgen-Ammann et al., 2017; Louw & Deary, 2014; Vincent et al., 2014; Zifchock, Davis, Higginson, McCaw, & Royer, 2008). Findings on the relationship between biomechanical asymmetries and injury have been equivocal with some research generally supporting (Gilgen-Ammann et al., 2017; Louw & Deary, 2014; Subotnick, 1981; Valovich McLeod et al., 2011; Vincent et al., 2014) and others refuting (Bredeweg et al., 2013; Tenforde, Ruder, Jamison, Singh, & Davis, 2018; Warren, Smith, & Chimera, 2014; Zifchock, Davis, & Hamill, 2006; Zifchock, Davis, Higginson, McCaw, et al.,

2008) the relationship. In cases where asymmetry levels are consistent in spite of unilateral injury, it has been argued that the risk of injury may be related to bilateral risk factors where the direction of the asymmetries may be related to which limb developed the injury (Zifchock et al., 2006; Zifchock, Davis, Higginson, McCaw, et al., 2008). Further, some ambiguity between running symmetry and injury may stem from the limited number of strides analyzed, approximately five, used to quantify these gait asymmetries (Tenforde et al., 2018; Zifchock et al., 2006; Zifchock, Davis, Higginson, McCaw, et al., 2008). When reported, other studies observing running asymmetries have still only utilized five to ten strides, with little concern to considering the effect that analyzing inconsecutive strides may have on the results (Beck, Azua, & Grabowski, 2018; Herzog et al., 1989; Pappas, Paradisis, & Vagenas, 2015). These studies may not have yet achieved a stable measure of symmetry, in which the measure may be more susceptible to changes in findings with more strides, and in turn findings may have changed if observing a greater number of strides.

During walking and running, gait parameters such as stride length and stride frequency are optimized to minimize energetic costs (Alexander, 2002). As asymmetries increase, the resulting metabolic cost of asymmetrically running increases, further supporting that individuals will naturally adopt symmetrical gait patterns to minimize metabolic energy expenditures (Beck et al., 2018). However, runners with asymmetrical leg lengths have been shown to move more asymmetrically at the same metabolic demand (Seminati et al., 2013). This suggests that structural symmetry may not be essential to maintain metabolic costs of locomotion, but can start to effect metabolic costs

when major asymmetrical structural differences are present, such as individuals with unilateral amputations (Czerniecki & Morgenroth, 2017; Yen, Schmit, & Wu, 2015).

The Amputee Coalition of America estimates two million Americans are living with a major extremity amputation, with the population doubling by 2050. The number of persons with transtibial amputations (TTA) has grown with an increased rate of distal limb amputations as a complication of type-2 diabetes, advancements in lifesaving medical treatments, and improved protective military gear; with 42% of combat related amputations occurring at the TTA level (Belatti & Phisitkul, 2013; Epstein, Heinemann, & McFarland, 2010; Krueger, Wenke, & Ficke, 2012). Although the hip and knee joints are largely preserved in the affected limb with a TTA, the loss of the ankle and associated musculature results in a number of mechanical challenges that the individual must overcome to successfully locomote (Czerniecki & Morgenroth, 2017; Hak, van Dieen, van der Wurff, & Houdijk, 2014; Mattes, Martin, & Royer, 2000; Nolan, 2008; Wanamaker, Andridge, & Chaudhari, 2017; Warren et al., 2014). Following surgery, a person with a unilateral TTA are fitted with a light-weight prosthetic limb, creating a mechanical asymmetry with the intact limb. These mechanical asymmetries are thought to contribute to the asymmetrical gait patterns during prosthetic use (Mena, Mansour, & Simon, 1981). Such interlimb temporal deviations include shorter contact times and longer swing times on the prosthetic side compared to the intact limb (Adamczyk & Kuo, 2015; Czerniecki, Gitter, & Weaver, 1994; Sanderson & Martin, 1996).

Individuals with a TTA also exhibit increased metabolic costs during walking and running compared with those without an amputation (Mengelkoch, Kahle, & Highsmith, 2014). In both walking and running, individuals with a unilateral TTA have an increased

metabolic cost (~20-30%) compared with those without amputations, even at reduced

speeds of walking and running in individuals with a TTA (Gailey et al., 1994;

Mengelkoch et al., 2014; Waters & Mulroy, 1999). Equivocal results are often reported in

the literature related to whether prosthetic mass influences metabolic costs of locomotion

(Gailey et al., 1994; Mattes et al., 2000; Smith & Martin, 2013). However, Smith &

Martin (2013) reported that both mass and mass location influenced metabolic cost

during walking for individuals with a unilateral TTA. Additionally, alterations in

spatiotemporal gait characteristics, from preferred, increase metabolic cost (Umberger &

Martin, 2007; Zarrugh, Todd, & Ralston, 1974). Even unilateral deviations, asymmetrical

step time during walking increased metabolic cost compared to symmetrical step time,

supporting spatiotemporal asymmetries could alter metabolic costs (Ellis, Howard, &

Kram, 2013).

These findings suggest that altered limb swing is metabolically costly. When

combining task-by-task contributions of the metabolic cost of walking and normalizing to

100%, the net metabolic cost of walking attributed ~7% solely to limb swing (Arellano

& Kram, 2014). However, when addressing limb swing as a task that can assist in

forward propulsion and a number of other tasks; the metabolic cost of limb swing is

approximately 20% (Gottschall & Kram, 2003). This cost is substantial and coincides

with increased in the metabolic cost of walking in persons with a unilateral TTA (Gailey

et al., 1994; Mengelkoch et al., 2014; Waters & Mulroy, 1999).

It is understood that asymmetrical gait patterns, due to altered limb inertia and

alterations in step time, increase the cost of locomotion (Ellis et al., 2013; Smith &

Martin, 2013). However, it is not clear how these two manipulations interact, and in turn

how each specifically contributes to the increased metabolic costs of locomotion seen in TTAs (Czerniecki & Morgenroth, 2017; Mattes et al., 2000; Rowe et al., 2014; Smith & Martin, 2013). As there are a number of confounding factors in those with unilateral amputations, it is desirable to first understand how these manipulations may affect healthy individuals to best understand the underlying metabolic cost of walking and running with inertial and spatiotemporal manipulations. Healthy controls will provide insights into the underlying relationships among inertial asymmetry, temporal asymmetry, and metabolic cost and will provide a foundation from which future rehabilitation programs could be designed for clinical populations where gait symmetry is often a goal. It is intended that the findings of this project will contribute to the growing body of research aimed at improving the quality of life of those with amputations, with the specific intent on improving the metabolic cost of walking as to make activities of daily living less energetically taxing.

This dissertation consisted of three studies. In the first two studies, the number of strides required to achieve a stable mean symmetry index and if strides need to be collected consecutively was determined for walking and running. The first study also aimed to understand if factors such as symmetry index and the timing of strides being collected altered outcomes during running. In addition to aims mentioned earlier, the second study determined if any significant difference occurred between subsets of strides throughout the full set of strides, and if the mean symmetry index from first three strides, five strides, and the average number of strides to achieve a stable mean symmetry index differed. The third study examined the effect of asymmetrical gait on metabolic costs by manipulating unilateral inertial and gait temporal parameters.

Study One Hypothesis – Running Symmetry
Stride Threshold

H01 There will be no effect on the lower extremity joint kinetic, joint
kinematic, and spatiotemporal symmetry indices when calculated from
consecutive and inconsecutive strides.

H02 There will be no effect on the lower extremity joint kinetic, joint
kinematic, and spatiotemporal symmetry indices when calculated from
groups of strides (First, Middle, Last, Random).

H03 There will be a difference between the Symmetry Index and Absolute
Symmetry Index, but indices will not differ in the number of strides to
achieve a stable mean symmetry index.

Study Two Hypothesis – Walking Symmetry
Stride Threshold

H01 There will be no effect on the lower extremity joint kinematic and
spatiotemporal symmetry indices when calculated from consecutive and
inconsecutive strides.

H02 There will be no difference between the mean symmetry indices of the
lower extremity joint kinematic and spatiotemporal symmetry indices
when calculated from different size subsets of strides (three strides, five
strides, number of strides to achieve an average stable mean symmetry
index).

H03 There will be a difference between the first three, five, and number of
strides to achieve an average stable mean symmetry index for the lower
extremity joint kinematics and spatiotemporal symmetry indices.

Study Three Hypothesis – Metabolic Cost of
Walking Asymmetrically

H01 Asymmetrically added mass will result in a greater metabolic cost of
walking compared with unloaded walking.

H02 Temporal asymmetries will result in a greater metabolic cost of walking
compared with symmetrical walking.

H03 Walking with temporal symmetry and an asymmetrically added mass will
result in a greater amount of metabolic cost of walking compared with
temporal asymmetric walking and walking with an asymmetrically added
mass.

CHAPTER II

REVIEW OF LITERATURE

Locomotion involves a symmetrical coordination of extremities, allowing for gait analyses to rely on the movement pattern of a single limb to generalize movement patterns (Whitall & Caldwell, 1992). However, a certain level of asymmetry has previously been quantified in healthy individuals during walking and running (Carpes et al., 2010; Nasirzade et al., 2017; Sadeghi et al., 2000). These unilateral deviations can be measured using interlimb symmetry indices, and can provide clinicians and researchers a way to quantify interlimb differences (Carpes et al., 2010; Czerniecki & Morgenroth, 2017; Herzog et al., 1989; Hoerzer et al., 2015; Kumar et al., 2014; Nigg et al., 2013; Robinson et al., 1987; Sadeghi et al., 2000). As a commonly used measure, it is important to know how to best obtain a stable mean symmetry index and be able to confidently report findings.

One population with inherently asymmetrical gait are persons with unilateral lower extremity amputations. Those with unilateral transtibial amputations (TTA) have mechanical differences between the affected and intact limb, resulting in inertial differences and kinematic, kinetic, and spatiotemporal adaptations. This mechanical or inertial asymmetry affects the timing of gait events such as increasing swing times and decreasing stance times on the affected limb (Adamczyk & Kuo, 2015; Czerniecki et al., 1994; Sanderson & Martin, 1996). Individuals with TTA also exhibit increased metabolic costs during walking and running compared with those without amputations (Mengelkoch

et al., 2014). It is understood that asymmetrical gait patterns, due to altered limb inertia and alterations in temporal symmetry, increase the cost of locomotion (Ellis et al., 2013; Smith & Martin, 2013). However, it is not clear how these two manipulations interact, and in turn how each specifically contributes to the increased metabolic costs of locomotion seen in individuals with unilateral TTA (Czerniecki & Morgenroth, 2017; Mattes et al., 2000; Rowe et al., 2014; Smith & Martin, 2013). Included in this review of literature is a description of healthy walking gait, with an emphasis on spatiotemporal parameters, kinetics, joint kinematics, joint kinetics, and metabolic costs. Asymmetrical locomotion, and methods of quantifying joint level asymmetries is included. Lastly, a description of unilateral TTA walking gait and how researchers have replicated amputee gait in healthy individuals is provided.

Walking Gait

Bipedal walking is characterized with at least one leg in contact with the ground at all times and contains a dual stance phase in which both limbs are in contact with the ground. A single limb will go through a stance phase and swing phase during one stride, with the contralateral limb in the opposite phase, except during dual stance (Figure 2.1) (Inman et al., 1981).
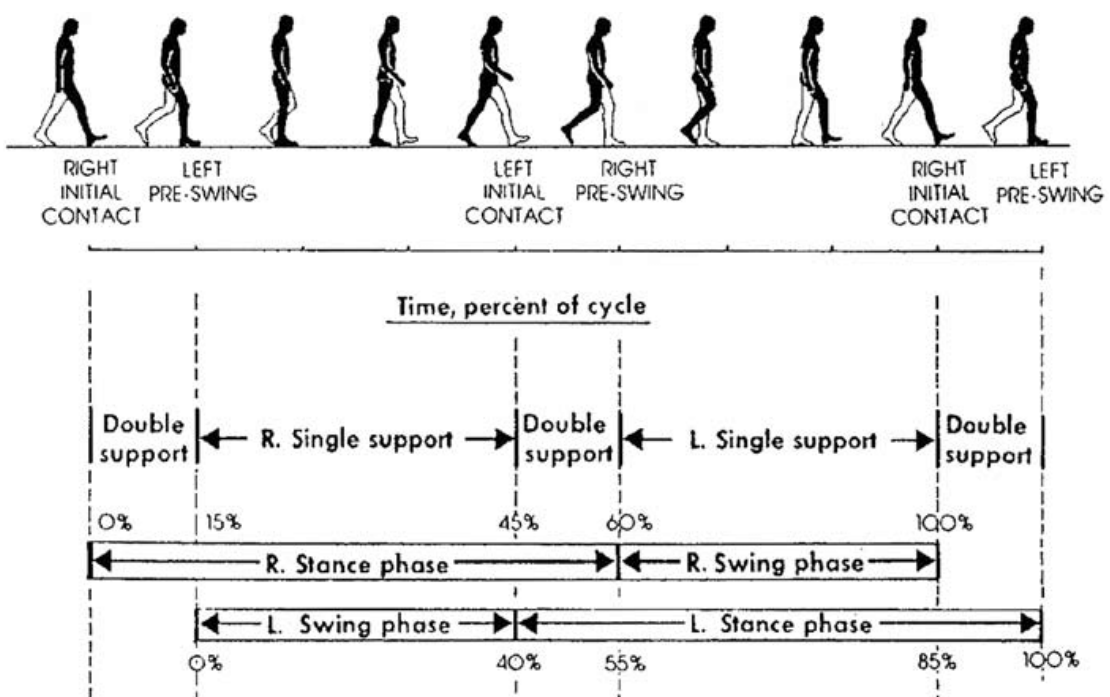
Figure 2.1 Gait Cycle of Normal Human Walking (adapted from Inman et al., 1981).

Walking gait was first conceptualized to contain six major determinants: pelvic rotation, pelvic tilt, knee flexion during stance, foot mechanisms during stance, knee mechanism during stance, and lateral displacement of the pelvis (Saunders, Inman, & Eberhart, 1953). More recently, walking mechanics has been modeled as an inverted pendulum, where the center of mass moves in a parabolic arch during single leg stance as the leg acts as a rigid segment (Figure 2.2) (Farley & Ferris, 1998). Regardless of model, walking is a symmetrical coordination of limbs, in which the legs are out of phase (Whitall & Caldwell, 1992). Although the inverted pendulum model is effective in describing whole-body motion, walking can be further described in spatiotemporal, kinematic, and kinetic measures. Additionally, the metabolic cost of transport can give additional insights to energetic demands of an individual's gait pattern.
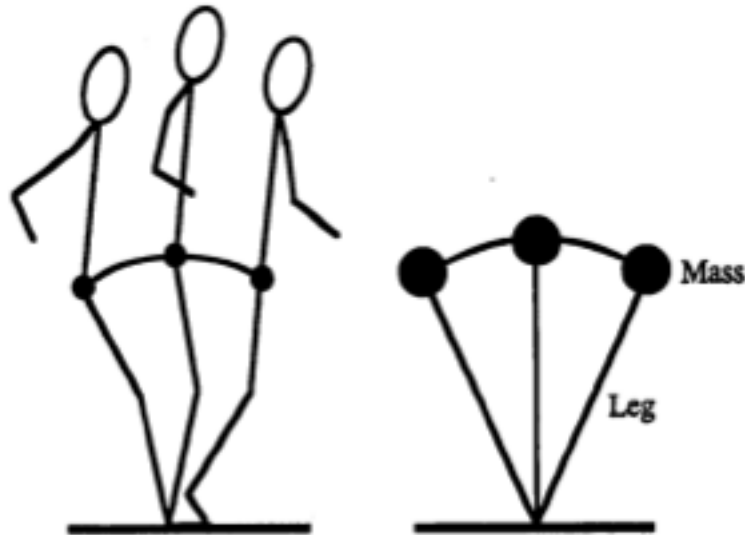
Figure 2.2 Inverted Pendulum of Walking. The center of mass of the body moves as an inverted pendulum during bipedal walking (adapted from Farley & Ferris, 1998).

Spatiotemporal

Walking gait can be characterized by various spatiotemporal parameters or measures: velocity, stride/step frequency or cadence, stride/step length, stance time, and swing time. Gait velocity can be calculated as a ratio of stride length to stride time. Stride frequency and cadence are the number of strides or steps per time, with an average healthy cadence of ~112 steps/min (Kadaba, Ramakrishnan, & Wootten, 1990; Riley, Paolini, Della Croce, Paylo, & Kerrigan, 2007). Stride and step length are the distance between initial contact of one foot to the ipsilateral or contralateral foot contact, respectively. Stance time is measured from initial contact to toe off, where swing time is measured from toe off to initial contact. As walking has no flight phase, it requires a double limb support phase, composing roughly 10% of the time per gait cycle per limb; totaling ~20% (Kadaba et al., 1990). During preferred walking velocities, the gait cycle will consist of ~60% stance phase and ~40% swing phase per limb (Kadaba et al., 1990; Riley et al., 2007).

Spatiotemporal parameters are inherently interrelated, with walking velocity being the most notable; as it is the product of stride length and stride frequency. If walking velocity is maintained, a reduction in stride length will result in an equal increase in stride frequency, or vice versa. When velocity is increased, stride length and stride frequency in turn must increase to the same degree. Velocity is also the ratio of stride length to stride time. However, with an increase in walking velocity, double limb support will be reduced (Dicharry, 2010).

Although spatiotemporal measures can easily be collected during overground walking, treadmill walking allows for the collection of continuous gait cycles. The resulting self-selected velocity of overground walking in healthy young men and women is 1.2–1.6 m/s and has been found to increase in subsequent sessions during overground trials, but not during self-selected treadmill walking (Oberg, Karsznia, & Oberg, 1993; Orendurff et al., 2004; Riley et al., 2007). In turn, previous research has found that the preferred walking velocity on treadmills has been maintained by adapting an increased stride frequency and reduced stride length compare with overground walking (Murray, Spurr, Sepic, Gardner, & Mollinger, 1985). Additionally, stance and swing time have been found to be reduced during treadmill walking without changes to walking velocity (Lee & Hidler, 2008). However, the differences noted between overground and treadmill walking are considered to have a minimal effect on the overall gait patterns (Lee & Hidler, 2008). Although these differences can be noted within participants, the differences fall within expected day-to-day variability, and in turn are negligible in multiple session studies (Riley et al., 2007). Further, treadmill walking allows for a greater amount of gait cycles to be collected for other measures, such as joint level

kinematics and kinetics. Treadmill walking ultimately allows for calculations that require continuous time series to be possible for spatiotemporal parameters and other biomechanical measures.

Kinematics

Motion of the whole-body can be generalized by the movement of the center of mass. As previously mentioned, the center of mass moves in an oscillatory fashion in the vertical and mediolateral planes, with peak vertical height reached during midstance (Farley & Ferris, 1998). The center of mass, as estimated by pelvic displacement, travels 4 cm in the vertical direction for a gait cycle (Inman & Locomotion, 1966). Self-selected walking velocity results in the least amount of mediolateral(~3.3 cm) and greatest vertical (~4.9 cm) displacement of the center of mass with the greatest amount of mediolateral (~7.0 cm) and least vertical (~2.7 cm) at slow velocities of 0.7 m/s (Orendurff et al., 2004). The center of mass velocity, when modeled as an inverted pendulum is similar to resultant velocity; however, center of mass velocity varies as the body brakes and propels forward through stance. In the vertical direction, the center of mass velocity ranges ± 0.2 m/s, about 0 m/s, as the center of mass changes direction throughout the stance phase (Adamczyk & Kuo, 2009).
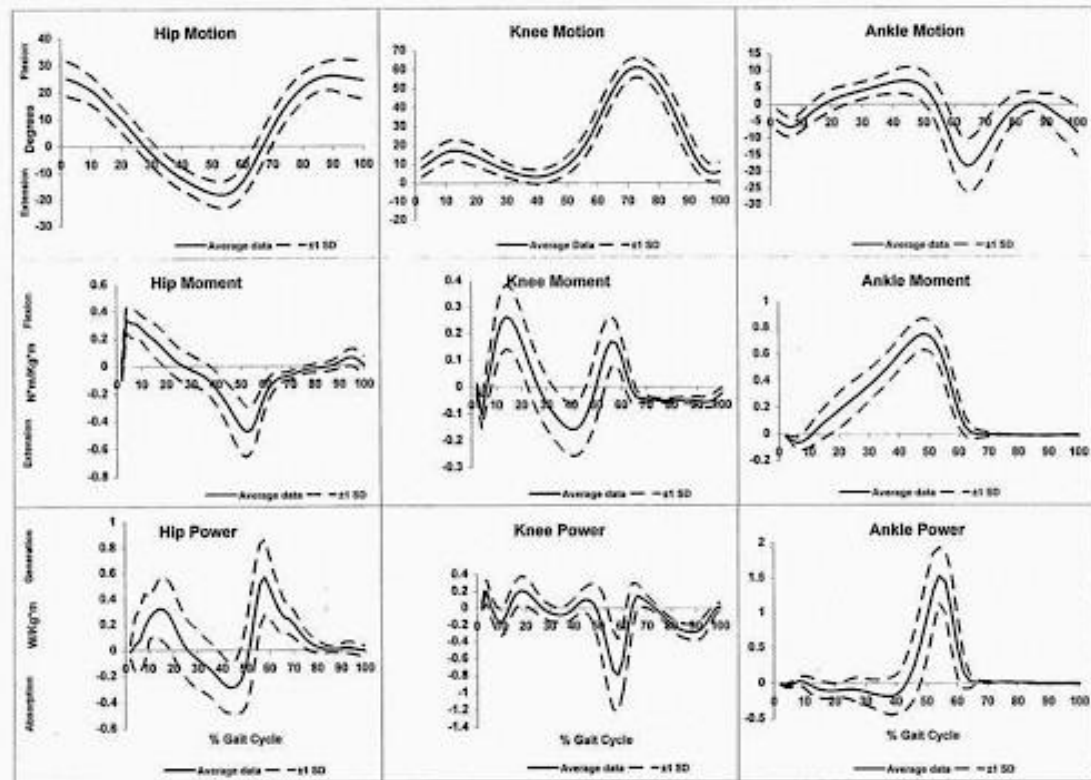
Figure 2.3 Lower Extremity Moment and Power Curves during Walking. (adapted from Pease & Bowyer, 2010).

The changes in center of mass displacement are linked to changes seen in lower extremity joint level kinematics. As seen in Figure 3, from initial contact to toe off the hip moves from flexion to extension in a range of motion 30-50° to peak extension just prior to toe off (Dicharry, 2010; Kadaba et al., 1990; Pease & Bowyer, 2010; Umberger & Martin, 2007). Peak hip flexion occurs in the later portion of the swing phase or just prior to initial contact, as the leg is repositioned in front of the body (Dicharry, 2010; Pease & Bowyer, 2010; Umberger & Martin, 2007). In the frontal plane, peak hip adduction and abduction of 5° occurs in the first 20% of stance and at toe off, respectively (Kadaba et al., 1990). The hip reaches peak angular velocity after toe off as it flexes in the sagittal plane, but has an extension velocity predominantly during stance (Umberger & Martin, 2007).

During stance, the knee flexes and extends slightly as the limb accepts additional weight (~20°) with greater flexion (60°) during swing to allow for toe clearance as the limb progresses forward (Kadaba et al., 1990; Pease & Bowyer, 2010; Umberger & Martin, 2007). The knee has a slight positive flexion velocity at initial contact, with peak flexion and extension velocities occurring during swing to reposition the limb prior to next initial contact (Umberger & Martin, 2007). This extension velocity has been shown to decrease with the addition of a distal mass, and is in turn related to the inertial properties of the limb (Smith, Villa, & Heise, 2013).

The ankle is normally dorsiflexed (~5°) prior to initial contact to allow for heel contact (Kadaba et al., 1990; Pease & Bowyer, 2010; Umberger & Martin, 2007). After foot flat, the ankle returns to a dorsiflexed position, and the limb rotates forward over the foot with peak dorsiflexion (~15°) occurring in late stance (Kadaba et al., 1990; Umberger & Martin, 2007). Just prior to toe off the foot rapidly plantarflexes and continues until just after toe off (~15°), at which peak plantarflexion velocity occurs to return the foot to a dorsiflexed position to allow for toe clearance (Kadaba et al., 1990; Umberger & Martin, 2007). Peak inversion of the ankle occurs as the initial contact, with peak eversion at toe off (Dicharry, 2010). This mobility allows for the limb to adapt to changes in surface and assists with push off from the ground during terminal stance.

Kinetics

Individual limb vertical ground reaction force – time curves produce an "m" shaped curve iconic to normal walking. The two peaks, reaching approximately 1.2x body weight, coincide with peak anteroposterior forces of the horizontal plane (0.25x body weight) as the center of mass reaches peak vertical position during stance phase

(Farley & Ferris, 1998; Inman & Locomotion, 1966). Both peak vertical and horizontal ground reaction force values are positively correlated to gait velocity (Nilsson & Thorstensson, 1989). Besides peak ground reaction force, the vertical curve can also give an indication of how quickly the body experiences force during initial contact. Lastly, the area under the vertical force – time curve represents the impulse experienced during locomotion, largely being effected by the time the limb is in contact with the ground, and is ~500 N•s during slow walking with a reduction to ~300 N•s as velocity increased (Nilsson & Thorstensson, 1989).

Anteroposterior, or horizontal, ground reaction force can give a better indication of forces experienced during the braking and propulsive phases of normal gait. During stable gait, the two peaks should be approximately the same magnitude, with opposite signs for the braking (negative) and propulsive (positive) peaks (Farley & Ferris, 1998). Unlike the single impulse in the vertical force – time curve, the horizontal impulse is again divided into braking and propulsive phases, with both braking and propulsive impulses showing an inverted U-shaped relation to gait velocity (Nilsson & Thorstensson, 1989). These impulses give a greater indication to the amount of propulsive and breaking forces that occurred, and for how long. Further, at higher walking velocities, propulsive impulse became greater than braking impulse and reached a peak value of ~30 N•s (Nilsson & Thorstensson, 1989).

Joint level kinetics can indicate the muscle group and the type of action utilized at a given point during the gait cycle via joint moments and powers (Figure 3) (Pease & Bowyer, 2010). From initial contact to toe off, the hip goes from peak hip extensor moment to peak hip flexor as the limb moves posteriorly. During swing phase, the hip

returns from the peak flexor moment to a hip extensor moment at the ipsilateral initial

contact. Throughout the gait cycle the peak hip power generation occurs during terminal

stance and initial swing phases, with peak hip power absorption occurring during

terminal stance. Knee joint moments alternate between flexor and extensor moment with

the flexor moments occurring during the loading response, midstance, and terminal swing

and the extensor moments occurring during the early stance, terminal stance, and the first

half of swing phase. The peak knee power absorption occurs near toe off. The ankle has a

small dorsiflexor moment during the loading response but predominantly acts as a

plantarflexor moment, with the peak plantarflexor moment occurring prior to the

contralateral heel strike. The peak ankle power generation occurs at terminal stance as the

limb pushes off the ground.

Metabolic Cost

Previous research has unequivocally supported that, when possible, walking

strategies minimizing the energetic cost of walking are utilized (Donelan, Kram, & Kuo,

2002; Holt, Jeng, Ratcliffe, & Hamill, 1995; Neptune, Sasaki, & Kautz, 2008; Ralston,

1958; Zarrugh & Radcliffe, 1978). For example, walking velocities requiring the least

energy to travel a given distance are voluntarily selected (Ralston, 1958). Further, a U-

shaped relationship between metabolic energy expenditure and stride rate is present

during walking, with individuals selecting stride rates that minimize rate of metabolic

energy expenditure (Holt et al., 1995; Zarrugh & Radcliffe, 1978). Similarly, at a fixed

step frequency, an increase in step length, and in turn walking velocity, increases

metabolic cost of walking (Donelan et al., 2002). This phenomenon of attempting to limit

metabolic expenditure during walking may partially be explained by the improved

utilization of elastic energy storage near self-selected walking velocities (Neptune et al., 2008). At self-selected walking velocities, the body is able to maximize the storage of elastic energy in the musculature of the lower extremity, and utilize it leu of actively recruiting musculature to propel the center of mass forward (Neptune et al., 2008).

During walking, there is an energetic cost to supporting body weight, swinging the limbs, and propelling the center of mass of the body forward (Griffin, Roberts, & Kram, 2003). Although swinging the legs forward has been measured to be limited compare with other tasks of human locomotion, the swing phase has been estimated to consume 26% of the energetic demands of walking in comparative studies with guinea fowl (Griffin et al., 2003; Marsh, Ellerby, Carr, Henry, & Buchanan, 2004). In the case of leg swing costs during running, Kram and colleagues found an estimated metabolic cost of approximately 20% (Modica, 2005). The metabolic cost of leg swing during walking has also been estimated to be approximately 20% during walking (Gottschall & Kram, 2003).

Asymmetrical Gait

Historically, lower extremity motion has been assumed to be symmetrical between limbs. Although limbs are generally symmetrical, this assumption was in part to simplify analyses and to limit the amount of time spent on processing data. As technology has improved, biomechanists can more easily make comparisons between limbs, and have noted a level of asymmetry even within healthy populations in both walking and running (Carpes et al., 2010; Herzog et al., 1989; Nasirzade et al., 2017). Beyond normative asymmetry values, asymmetrical gait patterns can develop in various clinical and injured populations (Carpes et al., 2010; Nasirzade et al., 2017). In turn,

symmetry indices are powerful tools to make interlimb comparisons within populations that are inherently asymmetrical. Further, unilateral manipulations to healthy participants can give insights to unique mechanical and energetic deficits that similar asymmetries produce in clinical and injured populations.

Measuring Symmetry

A number of different calculations have been proposed to quantify differences within and between limbs (Carpes et al., 2010; Sadeghi et al., 2000). One of the most widely used is the original symmetry index which proposed that symmetry was defined as a perfect agreement between left and right limbs (Herzog et al., 1989). This symmetry index, and similar variations, provide a percent difference between two discrete measures (Carpes et al., 2010; Robinson et al., 1987). Robinson and colleagues used equation (1) when assessing symmetry of ground reaction forces between the right (X_r) and left (X_l) measures (Robinson et al., 1987).

$$SI_{\%} = \left[\frac{(X_r - X_l)}{\frac{1}{2}(X_r + X_l)} \right] \cdot 100 \tag{1}$$

Others have suggested that perfect agreement may be too strict or an arbitrary approach to determining interlimb differences, and in turn determined asymmetries to be meaningful when statistical differences between measures were noted (Gabbard, 1997). Since Robinson and colleagues (1987) first quantified interlimb asymmetries, asymmetries have been studied in locomotion by utilizing between two and 200 strides (Owings & Grabiner, 2003; Sanderson & Martin, 1997). In spite of the large variability in the number of strides used in researching gait asymmetries, there has been consistent support that gait symmetry cannot be quantified using a single criterion value (Carpes et al., 2010; Herzog et al., 1989; Nasirzade et al., 2017).

In other words, there is not a single symmetry percentage that can be used in quantifying the presence of, or possible concern of, interlimb asymmetries.

Within biomechanical applications some limitations have persisted regardless of calculation utilized. For example, some tasks required 30 meters or approximately 10 strides for gait asymmetries to be measurable; possibly due to gait variabilities (Rumpf et al., 2014). This suggests that to reliably quantify asymmetries during sprinting at least 30 meters, or approximately ten strides, may be required. A similar adaptation phenomenon required two to three minutes of running with altered running techniques before consistent symmetry levels were established (Karamanidis, Arampatzis, & Brüggemann, 2003).

Aside from acclimation, a common challenge of symmetry indices are how influential spurious measures can be in artificially inflating measure magnitudes (Herzog et al., 1989; Zifchock, Davis, Higginson, & Royer, 2008). In turn, these high percentage asymmetries have been found with variables which had absolute magnitudes close to zero (Herzog et al., 1989; Zifchock, Davis, Higginson, & Royer, 2008). For example, measures such as ankle angle at initial contact can be near zero degrees. Any deviation from zero would be relatively small, but is inflated in a symmetry index with magnitudes of up to 13000%, and therefore needs to be interpreted with caution (Herzog et al., 1989). These inflations have been noted particularly in the original symmetry index; however, even adaptations have shown large variation within measures, especially between participants (Kumar et al., 2014; Pappas et al., 2015).

To combat these unique challenges and spurious findings, a significant effort has been taken by the movement science community to develop alternative approaches to

quantifying gait asymmetries. Some of these efforts have focused on limiting the effect of spurious asymmetries by creating composite scores of the traditional symmetry index. For example, the kinematic based Global Gait Asymmetry Index (GGA) has been used to quantify asymmetries during walking as shown in figure 2, where υ are the angular variables, t are the normalized time points, and X is the value for the left or right limb, respectively (Cabral et al., 2016).

$$GGA = \sum_{\upsilon=\upsilon_1}^{\upsilon_{15}} \sqrt{\sum_{t=t_1}^{t_{101}} [\chi_l(t) - \chi_r(t)]^2} \qquad (2)$$

This global index emphasizes whole segment movements, as locomotion requires the entire extremity, in an attempt to limit the effect of single joint asymmetries. The resulting index successfully limits the inflation for some measures (Cabral et al., 2016). Other global indices utilize multiple planes of motion during locomotion (Nigg et al., 2013). This Global Symmetry Index was effective in identifying symmetrical movement patterns in one plane, while identifying asymmetrical movement patterns in another (Nigg et al., 2013). Other, more wide-ranging measures such as the Comprehensive Asymmetry Index has also been proposed, which emphasizes improving "external boundary conditions," such as shoe conditions, that can be overlooked in other measures with implications on various neuromuscular controls (Hoerzer et al., 2015).

Other measures have relied on different statistical techniques to provide insights to interlimb asymmetries. For example, limb loading error scores have been used in addition to a modified symmetry index (Kumar et al., 2014). Intraclass correlation coefficients between limbs have also been used and successfully revealed joint kinematic asymmetries during running by demonstrating low measure reproducibility (Karamanidis et al., 2003). Principal component analysis has also been utilized when comparing

moment curves at each joint (Sadeghi, 2003). This local symmetry was compared with the global function of flexors/extensors. Although global function appeared to be symmetrical when total behavior of the limb was considered, compensations were occurring locally at the joint level (Sadeghi, 2003). In addition to statistical methods, measures of nonlinear dynamics have successfully characterized the interlimb differences within patients with Parkinson's disease. Specifically, cross-fuzzy entropy values of patients with Parkinson's were significantly higher than control subjects, suggesting a greater asymmetry in a patient's gait (Xia et al., 2016).

Symmetry angles have also been an effective way of quantifying interlimb asymmetries by comparing the angle created on a cartesian plane when plotting the two measures (Zifchock, Davis, Higginson, & Royer, 2008). These angles between measures were equally as effective for identifying interlimb differences as the original symmetry index, but not prone to issues associated to non-normalization by providing a standard ±100% scale to compare from (Zifchock, Davis, Higginson, & Royer, 2008). Additional symmetry angle calculations have been effective in quantifying kinetic asymmetries during running (Bredeweg et al., 2013). However these measures still demonstrated a high measure variability common in traditional indices (Bredeweg et al., 2013; Kumar et al., 2014; Pappas et al., 2015).

From traditional symmetry indices, slight alterations have resulted in improving the interlimb comparisons (Carpes et al., 2010). For example Shorter and colleagues created regions of deviation within walking kinematics and compared to a normative cohort (Shorter, Polk, Rosengren, & Hsiao-Wecksler, 2008). During conditions that required prophylactic ankle bracing that created asymmetries in terminal stance and

initial swing, symmetry indices of the whole joint range of motion were identified (Shorter et al., 2008). Other alterations stem from how limbs are defined (left vs right, dominant vs non-dominant), and have a profound influence in studying laterality (Carpes et al., 2010; Sadeghi et al., 2000).

One minor alteration to the original symmetry index that still utilizes perfect agreeance between limbs, but quantifies asymmetry magnitudes regardless of direction, is the absolute symmetry index. The absolute symmetry index is limited to not giving directional insight of which limb is asymmetrical, but does give an improved indication of magnitude regardless of limb (Carpes et al., 2010). For example, the original symmetry index may indicate 0% asymmetry when equal and opposite asymmetries are present, where the Absolute Symmetry Index would indicate the absolute percentage and eliminate any possible negations across multiple strides. In turn, Karamanidis and colleagues successfully utilized this alteration when assessing symmetry of kinematics during running with a number of running techniques (Karamanidis et al., 2003). Although the Absolute Symmetry Index eliminates the specific limitation of possible negations when the direction of asymmetry changes, it is less applicable to populations such as unilateral amputees, where a consistent directional asymmetry is expected for a given measure.

As there are a number of alterations, it is important to note that the original symmetry index proposed by Robinson, Herzog, and Nigg is still an effective tool of measuring interlimb symmetry as it provides clinically meaningful measures and is useful in assessing single joint asymmetries in both isolation or in concert with others (Carpes et al., 2010; Nasirzade et al., 2017; Robinson et al., 1987). The symmetry index is still

widely used in determining asymmetries in the clinical setting, especially with unilateral injuries or rehabilitative goals (Knapik, Bauman, Jones, Harris, & Vaughan, 1991; Louw & Deary, 2014; Nasirzade et al., 2017; Valovich McLeod et al., 2011).

Normative Asymmetry

As symmetry has been explored more extensively over the previous two decades, some normative data within healthy individuals has been revealed. These measures of normative asymmetry have been found to be highly consistent day-to-day, by only varying three to four degrees of range of motion (Wolf, List, Ukelo, Maiwald, & Stacoff, 2009). As walking and running are two distinct movement patterns, trends in asymmetry measures have been provided for both. Regardless of locomotive strategy, similar trends have developed: injury, performance, and laterality to just name a few. A more detailed insight to normative asymmetry in walking and running is provided.

Walking. Support for perfectly symmetrical walking is equivocal (Sadeghi et al., 2000). This lack of asymmetry is supported without identifying differences between dominant and non-dominant limbs (Hamill, Bates, & Knutzen, 1984). This idea of dominance, or laterality, suggests that limbs may serve specific roles (Peters, 1988). Beyond kicking or throwing objects, it is believed that the limbs uniquely contribute to gait mechanics (Sadeghi et al., 2000; Seeley, Umberger, & Shapiro, 2008). Specifically, the non-dominant limb contributes to support, while the dominant limb contributing to propulsion (Seeley et al., 2008). This lack of consensus may partially be due to the high degree of symmetry found in ground reaction forces during walking (Hamill et al., 1984). Links between mechanical variables such as vertical impulse (support) and propulsive impulse, have been shown with an asymmetrical increase in dominant limb impulse with

increased velocities (Seeley et al., 2008). The reported seven percent increase in the dominant limb propulsive impulse at fast velocities provides strong support for laterality but does little to support some form of functional asymmetry below fast velocities (Seeley et al., 2008). Joint power peak generation and absorption appears to be good indicators of a participant's ability to both propel and control balance during gait, respectively (Sadeghi et al., 2000).

Other possible explanations of underlying normative asymmetries during walking may fall within the realm of neuromechanics. One possible relationship is presented between unperturbed walking fractal dynamics and the adaptability of gait. These asymmetrical walking fractal dynamics and limb phase adaptation may represent the locomotor system improving limb interactions to better attenuate external perturbations (Ducharme et al., 2018). One unique external perturbation is walking next to someone at a different gait cadence. Interestingly, gait asymmetry decreased significantly when walking next to someone with a symmetrical gait (Nessler, Gutierrez, Werner, & Punsalan, 2015). Nessler and colleagues described the resulting reduction in asymmetries to stem from an underlying synchronous effect facilitated by proximity.

Even if symmetry in walking gait is equivocally supported, symmetry plays an important factor when assessing functional gait inefficiencies and is in turn a way to measure the effectiveness of a rehabilitation protocol (Nasirzade et al., 2017; Sadeghi et al., 2000). For example, symmetrical joint movements may appear to occur when total behavior of the limb is considered in healthy individuals (Sadeghi, 2003). However, asymmetrical movement patterns were present with compensations occurring at the joint level (Sadeghi, 2003). When considering computer simulations of healthy gait patterns, a

more complex mode with asymmetries may not be beneficial in predicting motion. For example, although asymmetries were detected in walking movement patterns by Ankarali and colleagues (2015), the research group preferentially excluded the complexity the asymmetries added to the model to improve predicting gait dynamics (Ankarali et al., 2015).

Running. As with walking, no single symmetry threshold is expected for biomechanical measures during running; however, a number of adaptations do occur when transitioning to faster velocities. Within a normal range of running velocities no systematic changes in joint level asymmetries are noted with increased velocities (Furlong & Egginton, 2018). As for most lower extremity variables during walking, variability of symmetry is fairly small during running (Pappas et al., 2015). Regardless of measure, these group averages mask the presence of a given individual's asymmetries (Ammann & Wyss, 2015; Pappas et al., 2015). The following are measure specific normative asymmetry values for healthy runners.

Spatiotemporal measures on average appear to have the least amount of asymmetries when compared with kinematic and kinetic measures during running, with an average of less than 10% compared with less than 40% (Frayne, 2014). For example, ground contact time asymmetries of ~3% have been noted during a 5 kilometer competition (Ammann & Wyss, 2015). Further, these ground contact times did not change over the course of the competition, suggesting no acute effect due to fatigue was present (Ammann & Wyss, 2015). A similar magnitude of ground contact asymmetry was noted during sprints between 400 and 1000 meters, again without the appearance of fatigue affecting symmetry (Gilgen-Ammann et al., 2017).

In running, like walking, kinetic profiles were originally reported to be highly symmetrical (Hamill et al., 1984). However, since Hamill and colleagues first reported symmetrical kinetic profiles, a number of studies have reported a wide range of asymmetries in healthy individuals (Girard, Brocherie, Morin, & Millet, 2017; Herzog et al., 1989; Karamanidis et al., 2003; Munro, Miller, & Fuglevand, 1987; Pappas et al., 2015; Rumpf et al., 2014). For example, kinetic measures are 15-20% asymmetrical have been observed in youth runners (Rumpf et al., 2014). These observed asymmetries in kinetic profiles during running tend to range from 4-28% for various measures, however spurious measures can exceed 13000% (Herzog et al., 1989; Karamanidis et al., 2003; Munro et al., 1987). These spurious measures are generally unexpected as previous research has reported a limited amount of variability in kinetic measures (Hamill et al., 1984; Pappas et al., 2015). No unique trends are present in higher velocities, with a consistent asymmetry ranging from 12-13% for horizontal forces and vertical forces under 5% for sprinting, and 4-28% asymmetry for a number of kinetic measures in elite runners (Girard et al., 2017; Karamanidis et al., 2003). Although it took at least 30 meters for asymmetries to present during sprinting, the onset of fatigue during running does not appear to alter kinetic asymmetries (Girard et al., 2017; Rumpf et al., 2014).

Similar to kinetic measures, kinematic measures range from 3-54% asymmetry depending on the variable of interest (Karamanidis et al., 2003). Specific to fast running, kinematics asymmetries were limited to under 10% (Girard et al., 2017). Further, these asymmetries during fast running were not affected by the onset of fatigue (Girard et al., 2017). Kinematic symmetry has also been correlated to anatomical symmetry, where the highly trained runner shows the highest levels of symmetry (Seminati et al., 2013).

Seminati and colleagues also found no correlations between kinematic variables and metabolic cost of transport, regardless of training. Further, the anatomically asymmetrical runners ran more asymmetrically at the same metabolic cost; raising questions to what degree of symmetry is blankly appropriate for all (Seminati et al., 2013).

A number of running specific measures, such as foot strike patterns, shoe conditions, and leg stiffness have been explored. For example, previous research has shown 6% of runners exhibit an asymmetrical interlimb foot strike pattern (one forefoot, one rearfoot), with a decline in asymmetry from the 10 km to 32 km mark of a marathon (Larson et al., 2011). This is one of few measures to show the possible influence of fatigue in interlimb mechanics (Brown, Zifchock, & Hillstrom, 2014; Girard et al., 2017). The overall fatigue state during a prolonged running protocol has also been demonstrated in some kinematic measures (Ali, Hiangl, Gerald, & Balasekaran, 2016). Joint level gait asymmetries have also been reduced when running in shoes compared with running barefoot (Hoerzer et al., 2015). These findings were supported by previous research showing interlimb asymmetries in rearfoot control being attenuated when wearing shoes (Vagenas & Hoshizaki, 1992). Leg stiffness has been reported symmetrical in healthy individuals, with asymmetries between limb dominance only present in peak vertical force and flight time (Pappas et al., 2015). In addition traditionally running specific measures, limb dominance did not relate to kinematic or kinetic asymmetries, with no changes after a fatiguing run (Brown et al., 2014).

Pathological Asymmetry

Although asymmetries, of various normative magnitudes are present during walking and running, in general, asymmetries are viewed as detrimental to locomotion.

These interlimb abnormalities have been associated with injury, and in turn a return to interlimb symmetry has been a common rehabilitative goal (Nasirzade et al., 2017). However, not all pathological asymmetries in gait stem from unilateral musculoskeletal injuries with a number stemming from anatomical and neurological origins.

A large portion of research on interlimb symmetry is focused on better understanding the underlying etiology of musculoskeletal injuries, and to determine if asymmetrical mechanics may contribute. However, a number of measures and pathologies have fallen short in providing definitive links. For example, no significant difference was noted in runners who had previously sustained stress fractures (Zifchock et al., 2006). Other findings on running related overuse injuries found similar asymmetry levels between injury and uninjured groups for all variables (Zifchock, Davis, Higginson, McCaw, et al., 2008). These findings support that injury risk may be related to bilateral risk factors, not necessarily asymmetries (Zifchock, Davis, Higginson, McCaw, et al., 2008). Forczek & Staszkiewicz (2012). Note that asymmetries during walking have sometimes been considered to indicate the presence of pathology (De Stefano, Burridge, Yule, & Allen, 2004). However, others have found asymmetries in spatiotemporal parameters (Forczek & Staszkiewicz, 2012). Lastly, the high levels of natural asymmetry in running kinetics was believed to be predictive of side of running related overuse injury, as one limb would be exposed to additional tissue loading (Bredeweg et al., 2013). However, those with a history of injuries compared with noninjured presented with no difference in kinetic asymmetries between the injured and uninjured limb (Bredeweg et al., 2013).

Other research has more clearly established a link between mechanical asymmetries and injury. Within the running population, symmetry commonly contributed to various running related overuse injuries, such as ITBS (Louw & Deary, 2014). Specific measures such as kinetic asymmetries have also been identified as a risk factor for bone health (Mizrahi, Verbitsky, & Isakov, 2000). Other interlimb biomechanical factors associated with injury were hip internal rotation ROM and peak tibial accel, where both were elevated on the side with a history of injury (Zifchock, Davis, Higginson, McCaw, et al., 2008). Therefore, asymmetries may influence the side injured instead of causing the injury (Zifchock et al., 2006).

Although there are equivocal results linking injury rates to asymmetrical movement patterns, it may be that pertinent asymmetries only occur during certain tasks. For example, ground contact asymmetries of athletes with a history of injury were higher than those without a history, with detections only at high intensity sprinting when measured over 400 meters (Gilgen-Ammann et al., 2017). It appears that asymmetries may also display differently between shod and unshod running. Injured runners with the highest loading rate asymmetry displayed lower asymmetries when running barefoot (Tenforde et al., 2018). However, this additional input of running barefoot only improved loading symmetry when habitual loading rates were already highly asymmetric (Tenforde et al., 2018). Gait asymmetries due to equipment, such as shoes, may be an additional risk for the onset of pain or injury (Vincent et al., 2014). Specifically, asymmetrical wear patterns on the soles of shoes may be an indicator of asymmetrical movement patterns (Vincent et al., 2014).

As some asymmetrical patterns have been related to the onset of injury, rehabilitative goals for unilateral injury usually include returning to a symmetrical capacity of the limbs. So much so, that the National Athletic Trainers' Association official position on preventing and rehabilitating overuse injuries in pediatric athletes includes the monitoring of movement pattern symmetry (Valovich McLeod et al., 2011). This is especially pertinent as kinetic asymmetries at the magnitude of 15-20% have been observed in youth runners, with some measures eclipsing normative levels in adult runners (Rumpf et al., 2014). Specific injuries have additionally received special attention, especially at the knee where the same limb injury history is related to hamstring injury recurrence and possibly performance (Croisier, Forthomme, Namurois, Vanderthommen, & Crielaard, 2002).

For college and adult aged athletes, gait symmetry remains to be a primary rehabilitative goal (Nasirzade et al., 2017). Symmetry is especially a common clinical goal with post-surgical populations (Diop et al., 2004; Hesse et al., 2003; Hodt-Billington, Helbostad, Vervaat, Rognsvåg, & Moe-Nilssen, 2011; Patterson, Nadkarni, Black, & McIlroy, 2012). For example, after anterior cruciate ligament reconstruction, clinicians set a goal of returning to activity within 10% ground reaction force asymmetry during drop-landing and commonly require no gross gait asymmetries during running (Myer, Paterno, Ford, Quatman, & Hewett, 2006). Additionally, higher trends of future injury have been established for with collegiate athletes that had ~15% asymmetrical strength and flexibility in knee flexors / hip extensors (Knapik et al., 1991).

With an interest in maintaining symmetrical movement patterns, different clinical tools have been developed to quantify global asymmetries. Particularly, the Functional

Movement Screen (FMS) is a set of 7 movement patterns aimed to determine if an individual is at an increased risk of injury/re-injury. However, the FMS has shown mixed results with control scores not differing between those injured and those not, with only the bilateral lunge associated with injury (Warren et al., 2014). In closing, the functional control of dynamic tasks, such as a lunge or single legged squat can be useful in detecting asymmetrical movement patterns (Valovich McLeod et al., 2011; Vincent et al., 2014). However, other movements involving arm and trunk movement should not be overlooked; with asymmetries being identified and ultimately corrected as they possess the possibility of affecting the development of a musculoskeletal injury (Nasirzade et al., 2017; Valovich McLeod et al., 2011; Vincent et al., 2014).

Although there is an extensive amount of interest in sport related pathologies, asymmetries are present in a number of other clinical populations. For example, it is estimated that at least 40% of the population possess some level of leg length discrepancy; with these discrepancies as a possible cause of musculoskeletal problems (Gurney, 2002). Not only may the length discrepancy be a possible source of musculoskeletal injury, there are trends to injure the shorter limb (Subotnick, 1981).

Further, individuals that suffer a stroke, post-recovery are faced with increased spatial and temporal asymmetries (Yen et al., 2015). Rehabilitative protocols of increasing swing resistance and assistance have been utilized to test energetic costs, with resistive locomotor retraining, not necessarily assistive interventions, deemed effective (Yen et al., 2015).

A uniquely asymmetrical population are those with unilateral amputations. The number of persons with amputations has grown with an increased rate of type-2 diabetes,

advancements in lifesaving medical treatments, and improved protective military gear;
with 42% of war related amputations at the transtibial (TTA) level (Krueger et al., 2012).
Following surgery, persons with unilateral TTA are fitted with light-weight prosthetic
limbs creating an inertial asymmetry with the intact limb. These inertial asymmetries are
thought to contribute to the asymmetrical gait patterns during prosthetic use (Mena et al.,
1981). These asymmetrical movement patterns have been shown to precipitate
throughout the gait cycle, with specific interlimb temporal deviations including shorter
contact times and longer swing times on the prosthetic side compared to the intact limb
(Adamczyk & Kuo, 2015; Czerniecki et al., 1994; Sanderson & Martin, 1996). TTAs also
exhibit increased metabolic costs during walking and running compared with non-
amputees (Mengelkoch et al., 2014). In both walking and running unilateral TTAs have
an increased metabolic cost (~20-30%) compared with those without amputations in spite
of self-selecting slower walking speeds (Gailey et al., 1994; Mengelkoch et al., 2014;
Waters & Mulroy, 1999).

Persons with unilateral amputations – mechanics. Persons with unilateral
amputations will rely on movement patterns that are uniquely asymmetrical to allow them
to locomote. This includes altered mechanics to overcome the limited capacity of the
prosthesis to store and generate force. These additional issues of symmetry may stem
from reduced proprioception and physiological loading limits in the intact limb (Nolan,
2008). In turn, limb symmetry has commonly been used as a primary variable when
examining persons with an amputation (Wanamaker et al., 2017). For example, marked
asymmetry in joint moments of amputees has been noted without uniformly increased
moments at increased velocities (Sanderson & Martin, 1997). Other studies have noted

the joint level impulses of the affected limb were reduced by ~35% and braking forces reduced by ~30% compared with the intact limb (Prince, Allard, Therrien, & Mcfadyen, 1992). Further, work at the estimated ankle joint of the affected limb is reduced during propulsion and results in compensations at the knee and hip (Adamczyk & Kuo, 2015).

Step lengths and widths of persons with amputations are also altered compared with able-bodied individuals (Arellano, McDermott, Kram, & Grabowski, 2015; Hak et al., 2014; Roerdink, Roeles, van der Pas, Bosboom, & Beek, 2012). For example, changes in step length may develop as a functional compensation, where shorter step lengths of the intact leg (5%) are adapted to increase dynamic stability (Hak et al., 2014). These asymmetrical step lengths may vary in magnitude between individuals but appear to be consistent within individuals (Roerdink et al., 2012). However, the variability of mediolateral foot placement increased systematically with increased velocity (Arellano et al., 2015). These increases in mediolateral foot placement were more variant and more asymmetrical for unilateral amputees (Arellano et al., 2015). Ultimately, these interlimb asymmetries may be unavoidable (Adamczyk & Kuo, 2015).

Although the keel and prosthetic limb are not as dynamic as an intact limb, they do possess the capacity to be tuned to improve functionality and performance. One basic adjustment is to match the mass and moment of inertia of the prosthetic limb to remove the systematic asymmetry. However, this matching of mass and moment of inertia by adding mass results in a greater level of asymmetry in spatiotemporal parameters, defeating the intended purpose (Mattes et al., 2000). Besides the prosthetic limb, keel stiffness can be changed to hopefully improve the overall limb mechanics. By increasing keel stiffness categories, limb symmetry was improved in transtibial amputees (Nolan,

2008). Contrarily, a more flexible keel has also been found to improve asymmetries (Prince et al., 1992). In bilateral amputees, reducing keel stiffness did reduce the metabolic cost of walking (Beck, Taboga, & Grabowski, 2017b). Ultimately, asymmetries may be present regardless of keel in unilateral amputees (Prince et al., 1992). Further improvements to interlimb symmetry may stem from changing the load line of the keel to a more posterior position, allowing for an increase in plantar flexion angle (Nolan, 2008).

Interlimb inertial differences are to be expected in persons with unilateral amputations as the center of mass location, mass difference, and moment of inertia are altered in the prosthetic versus the intact limb. Specifically, a passive prosthetic used for those with a TTA average 35% less mass, with the center of mass 35% closer to the knee, and a 65% smaller moment of inertia in the sagittal plane (Lin-Chan, Nielsen, Yack, Hsu, & Shurr, 2003). When replicating interlimb inertial differences with a unilaterally added mass, kinetic adaptations occurred while joint level kinematics remained similar (Smith & Martin, 2007). It may be that joint level mechanics are maintained to preserve spatiotemporal parameters to some extent. Although adaptations to the new limb inertial properties can occur in less than five minutes, with some suggesting as few as 40 strides, there is an increased muscle activation during swing phase to help control the increased inertia of the limb (Noble & Prentice, 2006; Smith & Martin, 2007, 2011; Smith, Villa, et al., 2013). Attempts have been made to reduce inertial asymmetry by matching interlimb moments of inertia between the affected and intact limb, however this resulted in an increased metabolic demand during walking (Mattes et al., 2000).

Just as in able-bodied locomotion, persons with unilateral amputations adapt unique strategies for walking and running. However, even elite level athletes, in the form of Paralympians, face additional challenges that able-bodied athletes do not. Namely, transfemoral Paralympians run with more asymmetries when compared with walking (Burkett, Smeathers, & Barker, 2003). This proliferation of an asymmetrical gait negatively effects their performance so much so that these elite level athletes, in spite of the finely tuned prosthetics, have a reduced running velocity when compared with able-bodied athletes (Morrien, Taylor, & Hettinga, 2016). This reduction in velocity expands beyond running and effects a number of athletic events, even those that are adapted to sitting (Morrien et al., 2016).

Persons with unilateral amputations – energetics. One challenge that persons with amputations face is the increase energetic demand to locomote, with a 20% increase in energetic demands previously reported in those with lower extremity amputations (Gailey et al., 1994; Waters & Mulroy, 1999). Gait adjustments, such as reduced walking cadence, even result in an increase in energy expenditure in persons with amputations (Rowe et al., 2014). Previous research has attempted to explain the underlying cause of these increased energetic demands that persons with lower extremity amputations face. Within healthy gait, no differences are noted between joints to contributing to net positive power and energetics (Farris & Sawicki, 2012). As the prosthetic is not as effective as propelling the body forward at toe off, the increase in overall work to compensate for the lack of push off could contribute to the increase in metabolic costs (Houdijk, Pollmann, Groenewold, Wiggerts, & Polomski, 2009). These compensations are present to

overcome the lack of ankle production in those with transtibial amputations (Houdijk et al., 2009).

Ultimately a number of sources contribute to the increased metabolic costs in amputees, of which a reduced push off and an increase in intact limb collision work during step-to-step transition, as well as altered timing, elastic energetic storage, co-contraction, and poor energy transfers (Czerniecki & Morgenroth, 2017). Amputation level and etiology also may play an important role in determining metabolic rates, with higher metabolic demands for those with amputations due to a traumatic event (Czerniecki & Morgenroth, 2017). Further, there is a positive correlation between asymmetries and metabolic cost of walking; suggesting the resulting asymmetrical patterns alone contribute to the already increased metabolic demands (Ellis et al., 2013; Finley, Bastian, & Gottschall, 2013).

A number of interventions and gait alterations have been attempted to reduce the increased metabolic demand of locomotion. As walking cadence correlated with energy expenditure symmetrical auditory cues have been introduced, however they resulted in no improvement of the overall symmetry in amputees (Rowe et al., 2014). Alterations to prosthetic stiffness, height, and subsequent symmetry of stride kinematics were all made without a positive effect on metabolic costs of running (Beck, Taboga, & Grabowski, 2017a). However, Beck and colleagues did identify that the peak vertical ground reaction force correlated with the metabolic cost of running (Beck et al., 2017a). Other manipulations have included adding mass to the prosthetic shank to attempt to remove the mass and moment of inertia asymmetry between the affected and intact limb. These attempts have largely failed in reducing energetic demands, resulting in a greater energy

expenditure of approximately 6-7% (Mattes et al., 2000). Specifically, the addition of the mass increased energetic demands by ~5% for each kg of mass added (Mattes et al., 2000). Further, adding 100% of prosthetic mass difference to the prosthesis increased metabolic costs by 5-7% (Smith & Martin, 2013). The estimated difference between intact limb and prosthetic limb masses was ~2.35 kg, and ranged from 1.75-2.68 kg (Smith & Martin, 2013). With the addition of unilateral mass, kinematic and temporal symmetry did not improve regardless of the added mass locations, ultimately resulting in a 3-7% increase in stance and swing time asymmetries (Smith & Martin, 2013). Other studies have shown no significant increase in energetic costs of adding mass with above or below knee amputees (Czerniecki et al., 1994; Lin-Chan et al., 2003). Finally, smaller loads of 0.85 kg or 1.34 kg near shank center of mass resulted in no significant effect on the metabolic cost of walking (Czerniecki et al., 1994).

Gait Manipulation

To better understand the mechanical and energetic challenges that persons with unilateral amputations face, researchers have attempted to replicate the mechanical asymmetries in healthy individuals. These studies have usually resulted in the addition of mass to a single the distal limb to replicate the magnitude of asymmetrical property differences between the limbs. This addition of mass at or distal to the center of mass has successfully increased metabolic costs (Smith & Martin, 2013). Other studies have attempted to replicate the asymmetrical movement patterns by manipulating the spatiotemporal parameters unilaterally with an audible metronome(Beck et al., 2018). When manipulations are not feasible, computer models have been utilized. For example, a model estimating instantaneous energy consumption was used to determine the cost of

each gait phase (Umberger, 2010). The leg swing represented 29% of total muscular cost, while double and single limb support accounted 27% and 44%, respectively (Umberger, 2010). This model further demonstrated that an increased stride rate resulted in a greater metabolic cost of walking during double limb support and lower cost during swing, without changing costs associated with the single limb support (Umberger, 2010).

A number of studies have added 2 kg distally to replicate the magnitude of difference between the prosthetic and residual limb (Mattes et al., 2000; Smith & Martin, 2007; Smith, Villa, et al., 2013). In matching mass and moment of inertia of prosthetic to intact limb, Mattes and colleagues determined that the additional mass was detrimental to symmetry and energetics, with distal loading deemed an effective manipulation to replicated unilateral transtibial gait (Mattes et al., 2000). These mass manipulations resulted in changes to a number of spatiotemporal, joint kinematic and joint kinetic measures. For example, the loaded limb has exhibited greater peak sagittal plane moments at knee and hip during the swing phase (Smith & Martin, 2007). Additionally, sagittal plane motion during the swing phase revealed increased angular impulses at the hip and knee of the loaded limb (Smith, Villa, et al., 2013).

However, the addition of 2 kg unilaterally does require time to adapt. For example, approximately 40 and 50 strides are required to fully adjust joint moments and joint kinematics, respectively (Noble & Prentice, 2006; Smith, Villa, et al., 2013). After adjusting to the perturbation, it took 20, 70, and 70 strides to return to perturbation kinematics at the ankle, knee, and hip, respectively (Noble & Prentice, 2006). The removal of the mass presented as a greater disruption than the addition of mass, with both representing a recalibration of the internal limb representation (Noble & Prentice, 2006).

This change in internal limb representation included: first the recalibration of mechanical parameters, and second, the actions to maintain the integrity of locomotor objectives such as propulsion (Noble & Prentice, 2006). Smith and colleagues also added 2 kg to better understand how the body recalibrates over a week long period (Smith & Martin, 2007). This longer term adaptation resulted in net joint moments alterations at the knee and hip, with acute adaptations, such as spatiotemporal parameters, complete within 5 minutes (Smith & Martin, 2007). Although kinetic adaptions appeared to occur due to the altered inertial properties and mass, joint level kinematics remained similar (Smith & Martin, 2007). The stability of joint kinematics with altered joint moments has since been supported by additional literature (Smith, Villa, et al., 2013). Altered spatiotemporal parameters presented as an increase in stance time of the unloaded limb and an increase in swing time for the loaded limb, further supporting previous findings (Skinner & Barrack, 1990; Smith & Martin, 2007).

Within the limb, coordination between joints is maintained by requiring the majority of gait adaptations to occur between the limbs (Haddad et al., 2006). For example, no changes in spatiotemporal patterns within the limb were revealed during walking with an asymmetrical load but were present when comparing between limbs (Haddad et al., 2006). Haddad and colleagues proposed that this finding further supports the hypothesis that the dominant limb provides propulsive force with nondominant limb providing support for body weight (Hirokawa, 1989; Sadeghi, Allard, & Duhaime, 1997). Other studies have examined intersegmental moments with unilateral limb loading during walking (Smith, Royer, & Martin, 2013). The increased moments occurred throughout the swing phase, without resulting in a change of the unloaded limbs (Smith, Royer, et

al., 2013). These altered inertial properties of the manipulated limb not only affected the amount of muscular effort required to swing the leg, but also changed the interlimb interactions (Smith, Royer, et al., 2013). This additional burden of musculature increased as mass was added due to the greater need to counteract the increased intersegmental interactions (Smith, Royer, et al., 2013). This ultimately presented as an increase in gravitational moments and a subsequent reduction in muscle moments, in which the body allowed for motion not produced by the muscles to occur in an attempt to minimize energetic costs (Smith, Royer, et al., 2013).

Masses other than 2 kg have also been added to the distal shank (2.3 kg – 4.6 kg), consistently resulting in altered spatiotemporal parameters and ground reaction forces and a consistent 4-6% increase in stance phase asymmetries (Muratagic, Ramakrishnan, & Reed, 2017). Smaller loads of 0.85 kg and 1.34 kg have also been added to healthy individuals resulting in an increase of the metabolic cost of walking by ~4% (Czerniecki et al., 1994). Other alterations have included altering the inertial properties of the limbs while not increasing gravitational forces (De Witt, Hagan, & Cromwell, 2008). Ground reaction forces altered by changes to limb inertia, with unique adaptations during walking and running (De Witt et al., 2008). Namely, peak vertical ground reaction force and loading rates increased with greater inertial manipulations during walking compared with decreased vertical ground reaction forces and loading rates in running (De Witt et al., 2008). Additionally, stride time increased during walking (De Witt et al., 2008). In spite of a number of methodologies unilaterally added masses, gait asymmetries have been shown to be reduced significantly when walking next to someone with a symmetrical

gait, suggesting external factors may play a role in the presence of asymmetrical gait patterns (Nessler et al., 2015).

Much less frequently utilized than the addition of mass unilaterally, spatiotemporal manipulations can alter gait patterns without the addition of mass in healthy individuals. Currently, two studies have used an audible metronome to enforce locomotion with temporal manipulations. Ellis and colleagues explored the metabolic demands associated with highly asymmetrical walking patterns. Walking asymmetrically required more metabolic power than the preferred symmetrical gait (Ellis et al., 2013). Further, the positive mechanical power production increased in parallel with metabolic power. These adaptations were made during the double support phase of asymmetrical walking (Ellis et al., 2013). Specifically, asymmetrical walking to an asymmetrical metronome resulted in an increase in power absorption and an increase in power production during the single support phase (Ellis et al., 2013). The moderate asymmetrical conditions resulted in a ~20-30% increase in metabolic demands. Similar trends were presented by Beck and colleagues, where running was manipulated asymmetrically by the same audible metronome. This asymmetrical temporal manipulation resulted in symmetrical stance times but asymmetrical aerials times (Beck et al., 2018). The asymmetrical gait pattern was achieved by an increase in vertical ground reaction forces and resulted in an increase in metabolic cost of running (Beck et al., 2018).

CHAPTER III

METHODOLOGY

General Methodology

The purpose of this series of studies was to determine the number of strides required to consistently quantify asymmetries during locomotion and if asymmetrical perturbations would result in an increase in metabolic costs during walking. The initial two studies were designed to better understand how many strides of walking and running were required to obtain a stable mean of symmetry indexes on lower extremity variables of interest. These initial studies aimed to provide a better understanding of which biomechanical variables to explore and the number of strides to provide consistent results. The final study was designed to determine the metabolic demand of asymmetrical locomotion by altering inertial and temporal properties unilaterally. This was achieved by determining the temporal changes of gait with a unilaterally added mass and replicating the asymmetrical temporal properties without the added mass by using an audible metronome that when walked to resulted in the same asymmetrical pattern of when the mass was present. The series of conditions systematically controlled for the effect of inertial and temporal manipulations on the metabolic cost of walking.

Participants

Ten active persons were recruited from the student population at the University of Northern Colorado as well as members of the surrounding community. Inclusion of participants were determined from a pre-participation modified physical activity

readiness questionnaire and based on the following criteria: 18-30 years old, free of any

existing neuromuscular or skeletal injury or condition that may prevent them from

completing all tasks, injury free in the trunk and lower extremity within the last six

months, and average 150 minutes of moderate or 75 minutes of vigorous physical activity

a week.

The Institutional Review Board at the University of Northern Colorado provided

oversight for the study upon approval. Along with the pre-participation questionnaire,

participants were presented with an informed consent document, procedures were

verbally explained, and written consent was obtained with a copy of the informed consent

offered to the participant.

Data Collection

All studies required the use of an instrumented treadmill (AMTI, Watertown,

MA) and motion capture system (VICON, Englewood, CO) and were collected over two

visits to the Biomechanics Lab at the University of Northern Colorado. A detailed history

of physical activity habits, demographic information, and other health metrics (such as

age, height, weight, leg dominance, and leg lengths) were verbally collected from the

participants during the first session. For both sessions, participants were asked to change

into form-fitting clothing for data collection purposes. Retroreflective markers were

placed over bony landmarks on their trunk and lower and upper extremities to allow for

the participants movements to be captured. A 10-camera motion capture system (100 Hz)

were used to capture motion data (VICON, Englewood, CO). All conditions were

collected on a tandem-belt instrumented treadmill (AMTI, Watertown, MA), allowing for

ground reaction forces to be collected for consecutive steps of walking or running (2000

Hz). These conditions each required participants to either run for eight minutes at 2.5 m·s-1 or walk at 1.5 m·s-1 for eight minutes on the treadmill.

During study three, participants were also asked to wear a mask designed to cover the mouth and nose that allowed for the collection of exhaled air. Gas exchange and metabolic cost were measured via indirect calorimetry with a TrueOne 2400 metabolic cart (Parvo Medics, Sandy, UT). Before collecting condition trials, participants stood quietly for five minutes and separately walked for three-minutes to establish a baseline metabolic rate and allowed for acclimation to walking with a metabolic mask on the treadmill, respectively.

Data Analysis

Rate of oxygen consumption and carbon dioxide production during quiet stance and the four conditions were averaged over the last two minutes. The average metabolic rates were then used to calculate the metabolic power. Net normalized metabolic power were calculated by subtracting the quiet stance metabolic power from each condition's average metabolic power and dividing by the participant's mass. Average respiratory exchange ratio (RER) was used to characterize the metabolic findings.

All biomechanical variables were calculated for individual limbs over the last two minutes of each condition. Marker trajectories along with ground reaction forces were filtered using a lowpass 4th order, zero lag, Butterworth filter at 6 and 10 Hz, respectively. An individual Visual 3D model was used to calculate spatiotemporal, joint kinematic, and joint kinetic measures.

Study One Specific Methodology

Participants

Ten individuals who report running as their primary form of physical activity were recruited to participate in the study. All were free of lower extremity injury for at least six months prior to data collections and reported no history of surgery on the lower extremity or trunk. Participants completed all conditions in their own shoes, all of which were traditional, non-minimalist, footwear. The Institutional Review Board at the University of Northern Colorado provided oversight for the study. Along with the pre-participation questionnaire, participants were presented with an informed consent document, procedures were verbally explained, and written consent was obtained with a copy of the informed consent offered to the participant.

Data Collection

After providing informed written consent, participants answered a modified physical activity questionnaire and survey to determine participant eligibility and provide demographic details. Form fitting clothes was provided to allow retroreflective markers to be placed over bony landmarks of the pelvis, thigh, shank, and foot of both extremities. Segment position data was collected using clusters of reflective markers, with the joint axis identified with additional markers during a standing calibration trial. Motion (100 Hz) and ground reaction force (2000 Hz) data was captured using a ten-camera motion analysis system (VICON, Englewood, CO) and a tandem-belt instrumented treadmill (AMTI, Watertown, MA), respectively. Participants completed two sessions of running at 2.5 m·s$_{-1}$, in which each session consisted of running for nine minutes. Within each

session, two 60 second trials were collected at the 6-and 8-minute marks. The first 75 consecutive strides of each 60 second trial was analyzed.

Gaps in reflective marker trajectories of segments that were less than 10 frames long were filled as rigid bodies in Nexus 2.6 (VICON, Englewood, CO). Gap-filled motion and ground reaction force data was exported to Visual 3D (C-Motion, Germantown, MD) where data was filtered with a lowpass 4th order, zero lag, Butterworth filter at 10 Hz for ground reaction force data and 6 Hz for motion data. Spatiotemporal, joint kinematic, and joint kinetic variables were calculated, for hip, knee, and ankle joint angles, velocities, moments, and powers. Discrete measures of variable maximums and minimums throughout a stride for each limb, as well as joint angles at initial contact, was determined.

Data Analysis

Interlimb symmetry was calculated in MATLAB (MathWorks, Natick, MA) between discrete variables of the left and right limbs within a single stride, for each stride, resulting in 75 individual symmetry values for each variable of each trial. Symmetry values were calculated using two previously described methods. Robinson and colleagues used equation (1) when assessing symmetry of ground reaction forces between the right (X_r) and left (X_l) measures (Robinson et al., 1987). Although this symmetry index (SI) is widely used and provides a reference of directionality, it may underestimate the average magnitude of asymmetry when disregarding direction, by cancelling out limb asymmetries when present in both limbs.

$$\text{SI}_\% = \left[\frac{(X_r - X_l)}{\frac{1}{2}(X_r + X_l)} \right] \cdot 100 \tag{1}$$

In turn, Karamanidis and colleagues used equation (2) when assessing symmetry of kinematics during running (Karamanidis et al., 2003). This absolute symmetry index (ASI) provided an absolute magnitude of asymmetry without giving insight to the directionality of the difference between limbs. Perfect symmetry in both equations was represented by zero. Mean symmetry for each running trial was determined by computing an average for all 75 strides.

$$\text{ASI}_\% = \left[\frac{|(X_r - X_l)|}{\frac{1}{2}(X_r + X_l)} \right] \cdot 100 \tag{2}$$

A stable mean symmetry (SMS) was determined by taking a sequential average of each individual trial and noting when the sequential average remained within a ±2 SD range about the mean symmetry value of the trial. A ±2 SD range was used so that a SMS would reached for the majority of data points within a normal bell curve. Trials were removed prior to determining a threshold if individual strides within a trial greatly differed from subsequent strides.

The average number of strides required to achieve a SMS was used to determine the size of bin used to compare groups of strides within a trial. Separate bins, of equal size, were created to group the first, middle, last, and random strides. Bins of the first, middle, and last strides were consecutive, with the bin of random strides selected using a random number generator within MATLAB. These bins could then be compared to a bin containing all 75 strides, an average for the trial.

Statistical Analysis

A series of 2-factor (method x bin) analyses of variance (ANOVA) ($p < 0.05$) were used to determine if differences between SI and ASI calculations and between random, first, middle, last, and all strides bins symmetries were statistically significant.

The method factor was treated as a between subject factor, while the bin factor was treated as a repeated measure. A Bonferroni adjustment was utilized in post hoc pairwise comparisons. Since between-trial comparisons were not examined, each trial was treated as individual subjects. Statistical analyses were completed using SPSS 24 (SPSS Inc., IBM, Chicago, IL).

Study Two Specific Methodology

Participants

Ten individuals who met the minimum ACSM guidelines of weekly physical activity were recruited to participate in the study. All were free of lower extremity injuries for at least six months prior to data collections and reported no history of surgery on the lower extremity or trunk. Participants completed all conditions in their own shoes, all of which should be traditional, non-minimalist, footwear. The Institutional Review Board at the University of Northern Colorado provided oversight for the study. Along with the pre-participation questionnaire, participants were presented with an informed consent document, procedures were verbally explained, and written consent was obtained with a copy of the informed consent offered to the participant.

Data Collection

After providing informed written consent, participants answered a modified physical activity questionnaire and survey to determine participant eligibility and provide demographic details. Form fitting clothes were provided to allow retroreflective markers to be placed over bony landmarks of the pelvis, thigh, shank, and foot of both extremities. Segment position data was collected using clusters of reflective markers, with the joint axis identified with additional markers during a standing calibration trial. Motion (100

Hz) data was captured using a ten-camera motion analysis system (VICON, Englewood, CO). Participants completed two sessions of walking at 1.5 m·s-1, in which each session consisted of walking for nine minutes. Within each session, two 60 second trials were collected at the 6-and 8-minute marks. The first 50 consecutive strides of each 60 second trial were analyzed.

Gaps in reflective marker trajectories of segments that were less than 10 frames long were filled as rigid bodies in Nexus 2.7 (VICON, Englewood, CO). Gap-filled motion data were exported to Visual 3D (C-Motion, Germantown, MD) where data were filtered with a lowpass 4th order, zero lag, Butterworth filter at 6 Hz. Spatiotemporal and joint kinematic variables were calculated, for hip, knee, and ankle joint angles and velocities. Discrete measures of variable maximums and minimums throughout a stride for each limb were determined.

Data Analysis

Interlimb symmetry was calculated in MATLAB (MathWorks, Natick, MA) between discrete variables of the left and right limbs within a single stride, for each stride, resulting in 50 individual symmetry values for each variable of each trial. Symmetry values were calculated using a symmetry index (SI). Robinson and colleagues originally used equation (1) when assessing symmetry of ground reaction forces between the right (X_r) and left (X_l) measures (Robinson et al., 1987). This SI is widely used and provides a reference of directionality. Perfect symmetry in both equations was represented by zero. Mean symmetry for each walking trial was determined by computing an average for all 50 strides.

$$SI_\% = \left[\frac{(X_r - X_l)}{\frac{1}{2}(X_r + X_l)} \right] \cdot 100 \qquad (1)$$

A stable mean symmetry (SMS) was determined by taking a sequential average of each individual trial and noting when the sequential average remained within a ±2 SD range about the mean symmetry value of the trial. A ±2 SD range was used so that a SMS would be reached for the majority of data points within a normal bell curve. Trials were removed prior to determining a threshold if individual strides within a trial greatly differed from subsequent strides.

Subsets of consecutive strides were created within each trial to compare symmetry indices. Subset sizes were chosen to reflect the minimum number of strides collected when studying gait kinematics (three and five) and the number of strides to achieve a SMS. Additional subsets were created using random strides within a trial to determine if strides needed to be consecutive.

Statistical Analysis

Since between-trial comparisons were not examined, we treated each trial (four per participant) as individual subjects (n = 40). A series of analyses of variance (ANOVA) ($p < 0.05$) with repeated measures were used to determine whether same sized subsets differ within a trial, with each trial being treated as an individual participant. Additional ANOVAs with repeated measures were used to determine if differences in the number of strides within a subset would result in significant differences. Statistical analyses were completed using R (version 3.4.1).

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Study Three Specific Methodology

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Participants

Ten active persons were recruited from the student population at the University of Northern Colorado, as well as members of the surrounding community. Participants

successful inclusion in the study was determined from a pre-participation, modified

physical activity readiness, questionnaire and based on the following criteria: 18-30 years

old, free of any existing neuromuscular or skeletal injury or condition that may prevent

them from completing all tasks, injury free in the trunk and lower extremity within the

last six months, and average at least 150 minutes of moderate or 75 minutes of vigorous

physical activity a week.

The Institutional Review Board at the University of Northern Colorado provided

oversight for the study. Along with the pre-participation questionnaire, participants were

presented with an informed consent document, procedures were verbally explained, and

written consent was obtained with a copy of the informed consent offered to the

participant.

Data Collection

Participants changed into form-fitting clothing for data collection purposes and

were provided a pair of Brooks Launch 5 athletic shoes to complete all conditions in

(Brooks Running Seattle, WA). A detailed history of lower extremity overuse injuries,

demographic information, and other health metrics (such as age, height, weight, leg

dominance, and leg lengths) were collected from the participants. Retroreflective markers

were placed over bony landmarks on their trunk and lower and upper extremities to allow

for the participants movements to be captured. Additional light-weight thermoplastic

plates with retroreflective marker clusters were placed over upper and lower extremities.

A 10-camera motion capture system (100 Hz) was used to capture motion data (VICON,

Englewood, CO).

Participants were also asked to wear a mask designed to cover the mouth and nose that allows for the collection of exhaled air. Gas exchange and metabolic cost were measured via indirect calorimetry with a TrueOne 2400 metabolic cart (Parvo Medics, Sandy, UT).

Participants first stood quietly for five minutes to provide the metabolic cost of standing and then walked at 1.5 m·s-1 for three minutes to acclimate to the laboratory treadmill. All conditions were collected on a tandem-belt instrumented treadmill (AMTI, Watertown, MA), allowing for ground reaction forces to be collected for consecutive steps of walking (2000 Hz). Conditions required participants to walk for eight minutes at 1.5 m·s-1 on the treadmill while wearing the metabolic mask.

Data collections were divided in half with the first two conditions providing a baseline to symmetrical and asymmetrical walking and the average swing percent symmetry and strides per minute for walking with and without a unilaterally added 2kg mass to a distal shank (Table 3.1). The second half of the data collection included two manipulation conditions where participants were asked to complete: manipulated symmetrical walking with a 2-kg mass added to the distal shank of the dominant limb to a symmetrical metronome and manipulated asymmetrical walking to an asymmetrical metronome without the unilateral mass. The audible metronome utilized the preferred percent asymmetry and the average strides per minute from the two baseline walking conditions. The participants were also provided with real-time biofeedback of the goal swing times for the given conditions. Biofeedback was displayed in front of the participants with a stacked bar graph of goal swing symmetries overlaid with superimposed bar graphs of the actual swing times for each limb. Swing times were

calculated by streaming ground reaction forces from the instrumented treadmill into a

MATLAB script (MathWorks, Natick, MA).

Table 3.1

Summary of Proposed Conditions for Study Three

Conditions	Baseline	Manipulation
1	Preferred SF, no mass added	Symmetrical C1 SF, mass added
2	Preferred SF, mass added	Asymmetrical C2 SF, no mass added

Data Analysis

Rate of oxygen consumption and carbon dioxide production during quiet stance

and the four conditions were averaged over the last two minutes of each condition. The

average metabolic rates were then be used to calculate the metabolic cost of standing and

walking, respectively. Net normalized metabolic cost of walking was calculated by

subtracting the metabolic cost of quiet stance from each condition's average metabolic

cost of walking and dividing by the participant's mass. Average respiratory exchange

ratio (RER) was also be used to characterize the metabolic findings and to ensure

participants were in a steady-state effort.

The final 90 seconds of walking were biomechanically analyzed, with each stride

being analyzed separately. A low-pass, 4_{th} order, zero lag Butterworth digital filter was

used on marker trajectories ($F_c = 6$ Hz). All calculations utilized participant specific

models created in Visual3D (C-motion, Germantown, MD). Spatiotemporal parameter

calculations included step length, step frequency, step time, and swing time. A symmetry

index was calculated for each spatiotemporal parameter (Robinson et al., 1987).

Statistical Analysis

Differences between baseline and manipulation conditions were compared as percent differences to ensure participants were accurately completing the two manipulation conditions. A repeated measure analysis of variance (ANOVA) with an $\alpha =$ 0.05 was performed in SPSS 24 (SPSS Inc., IBM, Chicago, IL) to determine if the metabolic cost of walking differed between conditions. A two factor (load x swing time) ANOVA with an $\alpha = 0.05$ was performed on net metabolic cost of walking data with interactions and pairwise comparisons to further understand where differences between factors occurred. A final two factor (load x manipulation) ANOVA with an $\alpha = 0.05$ was performed to determine if an interaction occurred between load and manipulation (baseline versus manipulation conditions), and whether forcing gait symmetry away from natural adaptations to unilateral load resulted in a greater net metabolic cost of walking. Spatiotemporal measures were also compared with a series of repeated measure ANOVAs to further explain conditional adaptations.

CHAPTER IV

STUDY ONE: MINIMUM NUMBER OF STRIDES TO
DETERMINE STABLE INTERLIMB SYMMETRY
INDEX DURING RUNNING

Introduction

Interlimb symmetry is commonly used to describe the quality of running gait,

with asymmetries suggesting pathology or a reduction in performance (Bredeweg et al.,

2013; Gilgen-Ammann et al., 2017; Seminati et al., 2013; Zifchock et al., 2006).

Research on runners with a history of injury has demonstrated significant asymmetries in

joint angles and ground contact time in the affected limb (Ciacci, Di Michele, Fantozzi,

Merni, & Mokha, 2013; Ellis et al., 2013). Additionally, asymmetries in step time have

been shown to significantly increase the metabolic cost of walking and running (Beck et

al., 2018; Gilgen-Ammann et al., 2017). There are a number of methods to calculate

symmetry between limbs (Carpes et al., 2010). One method, originally proposed by

Robinson and colleagues (1987), is the ratio of differences between limbs and limb

average (Robinson et al., 1987). This calculation is effective in characterizing the

direction of limb asymmetry (Herzog et al., 1989). When the magnitude regardless of

direction of the asymmetry is desired, the ratio of absolute differences between limbs and

limb average has been used (Karamanidis et al., 2003; Pappas et al., 2015). Although

both calculations are widely used, the results may differ as limb measurements that are

equal but opposite, are effectively cancelled out when using the symmetry index method.

It has been established that using one criterion value to assess gait symmetry is not appropriate across all biomechanical variables (Robinson et al., 1987). This may be partially due to the large variability of calculated symmetry between strides (Herzog et al., 1989; Zifchock et al., 2006). Previous researchers analyzing running average 5-10 strides, with discontinuous overground running required for some lab setups (Beck et al., 2018; Gilgen-Ammann et al., 2017; Herzog et al., 1989; Pappas et al., 2015; Zifchock et al., 2006). With asymmetries being of particular interest in some clinical populations, it is essential to not collect an excessive number of strides. However, a limited number of strides may not provide for an accurate measure of interlimb symmetry. In turn, the number of strides required to have a stable mean symmetry index needs to be better understood.

Previous research has explored the minimum numbers of strides required to achieve a stable mean for running mechanics for instance, the seminal work of Bates, Osternig, Sawhill, and James (1983) established that four strides were necessary for ground reaction force data. However, some variables analyzed required all five strides, suggesting that more strides may have been required to achieve a stable mean (Bates et al., 1983). More recent research has utilized sequential averaging and intraclass correlation coefficients to determine the number of trials required for a stable mean (Gittoes & Moore, 2016; James, Herman, Dufek, & Bates, 2007; Moore & Gittoes, 2015). These studies analyzed 10 to 20 strides and quantified the mean as stable when the trial averages fell within .25 SD of the total trial mean. Bates' original findings were partially supported by the more recent work of Gittoes & Moore (2016) and James et al. (2007) who reported an average of 8-12 trials, or 4-6 strides were required to reach a

stable mean (Gittoes & Moore, 2016; James et al., 2007). Further, Moore & Gittoes established that the intraclass correlation coefficients were more liberal and required fewer trials than sequential averaging (Moore & Gittoes, 2015). With primarily ground reaction force data and limited kinematic variables reported in previous literature, the number of strides required to establish a stable mean for other biomechanical variables, such as spatiotemporal parameters, joint kinetics, and joint kinematics, may be different.

The purpose of this study was to determine 1) the number of strides required to determine a stable mean of lower extremity joint kinetic, joint kinematic and spatiotemporal symmetry indices, 2) if symmetry values differ between continuous and discontinuous data points, and 3) if the calculation used to measure symmetry would result in any differences in the number of strides required for a stable mean.

Methodology

Participants

Ten individuals (F = 6, 1.73 ± 0.12 m, 66.0 ± 12.0 kg, 25 ± 3 years, 326 ± 178 min/week of activity) who reported running as their primary form of physical activity participated in the study. Eight of the ten participants exhibited rearfoot strike patterns during running, while two varied between rearfoot and mid or forefoot strike patterns throughout the data collections. All were free of lower extremity injury for at least 6 months prior to data collections and reported no history of surgery on the lower extremities or trunk. Participants completed all conditions in their own shoes, all of which were traditional, non-minimalist, footwear. One participant did use a custom-orthotic during data collections but reported habitually wearing the orthotic during

running. The University's Institutional Review Board approved the protocol, with each participant providing informed written consent prior to collecting data.

Data Collection and Processing

After providing informed written consent, participants answered a modified physical activity questionnaire and survey to determine participant eligibility and provided demographic details. Form fitting clothes were provided to allow retroreflective markers to be placed over bony landmarks of the pelvis, thigh, shank, and foot of both extremities. Segment position data were collected using clusters of reflective markers, with the joint axis identified with additional markers during a standing calibration trial. Motion (100 Hz) and ground reaction force (2000 Hz) data were captured using a ten-camera motion analysis system (VICON, Englewood, CO) and a tandem-belt instrumented treadmill (AMTI, Watertown, MA), respectively. Participants completed two sessions of running at 2.5 m·s-1, in which each session consisted of running for nine minutes. Within each session, two 60 second trials were collected at the 6- and 8-minute marks. The first 75 consecutive strides of each 60 second trial were analyzed.

Gaps less than 10 frames in length in reflective marker trajectories of segments were filled as rigid bodies in Nexus 2.6 (VICON, Englewood, CO). Gap-filled motion and ground reaction force data were exported to Visual 3D (C-Motion, Germantown, MD) where data were filtered with a lowpass 4th order, zero lag, Butterworth filter with cutoff frequencies at 10 Hz for ground reaction force data and 6 Hz for motion data. Spatiotemporal, joint kinematic, and joint kinetic variables were calculated, for hip, knee, and ankle joint angles, velocities, moments, and powers.

Discrete measures of variable maximums and minimums throughout a stride for each limb, as well as joint angles at initial contact, were determined.

Symmetry Calculations

Interlimb symmetry was calculated in MATLAB (MathWorks, Natick, MA) between discrete variables of the left and right limbs within a single stride, for each stride, resulting in 75 individual symmetry values for each variable of each trial. Symmetry values were calculated using two previously described methods. Robinson and colleagues used equation (1) when assessing symmetry of ground reaction forces between the right (X_r) and left (X_l) measures. (Robinson et al., 1987) Although this symmetry index (SI) is widely used and provides a reference of directionality, it may under estimate the average magnitude of asymmetry when disregarding direction, by cancelling out equal but opposite limb asymmetries.

$$SI_\% = \left[\frac{(X_r - X_l)}{\frac{1}{2}(X_r + X_l)} \right] \cdot 100 \qquad (1)$$

In turn, Karamanidis and colleagues (2003) used equation (2) when assessing symmetry of kinematics during running.(Karamanidis et al., 2003) This absolute symmetry index (ASI) provides an absolute magnitude of asymmetry without giving insight to the directionality of the difference between limbs. Perfect symmetry in both equations is represented by zero. Mean symmetry for each running trial was determined by computing an average for all 75 strides, individually (Figure 3.1A).

$$ASI_\% = \left[\frac{|(X_r - X_l)|}{\frac{1}{2}(X_r + X_l)} \right] \cdot 100 \qquad (2)$$

Determining Threshold and
Comparisons Within Trials

A stable mean symmetry (SMS) was determined by taking a sequential average of each individual trial and noting when the sequential average remained within a ±2 SD range about the mean symmetry value of the trial (Figure 3.1B). A ±2 SD range was used so that a SMS was reached for the majority of data points. Trials were removed prior to determining a threshold if individual strides within a trial greatly differed from subsequent strides, with no more than ten trials per measure.

The average number of strides required to achieve a SMS was used to determine the size of bin used to compare groups of strides within a trial. Separate bins, of equal size, were created to group the first, middle, last, and random strides. Bins of the first, middle, and last strides were consecutive, with the bin of random strides selected using a random number generator within MATLAB. These bins could then be compared to a bin containing all 75 strides, an average for the trial.

Statistical Analysis

A series of two factor (method x bin) analysis of variances (ANOVA) ($p < 0.05$) were used to determine if differences between SI and ASI calculations and between random, first, middle, last, and all strides bins were statistically different using SPSS 24 (SPSS Inc., IBM, Chicago, IL). The method factor was treated as a between subject factor, while the bin factor was treated as a repeated measure. A Bonferroni adjustment was utilized in post hoc pairwise comparisons. Since between-trial comparisons were not examined, we treated each trial as individual subjects.
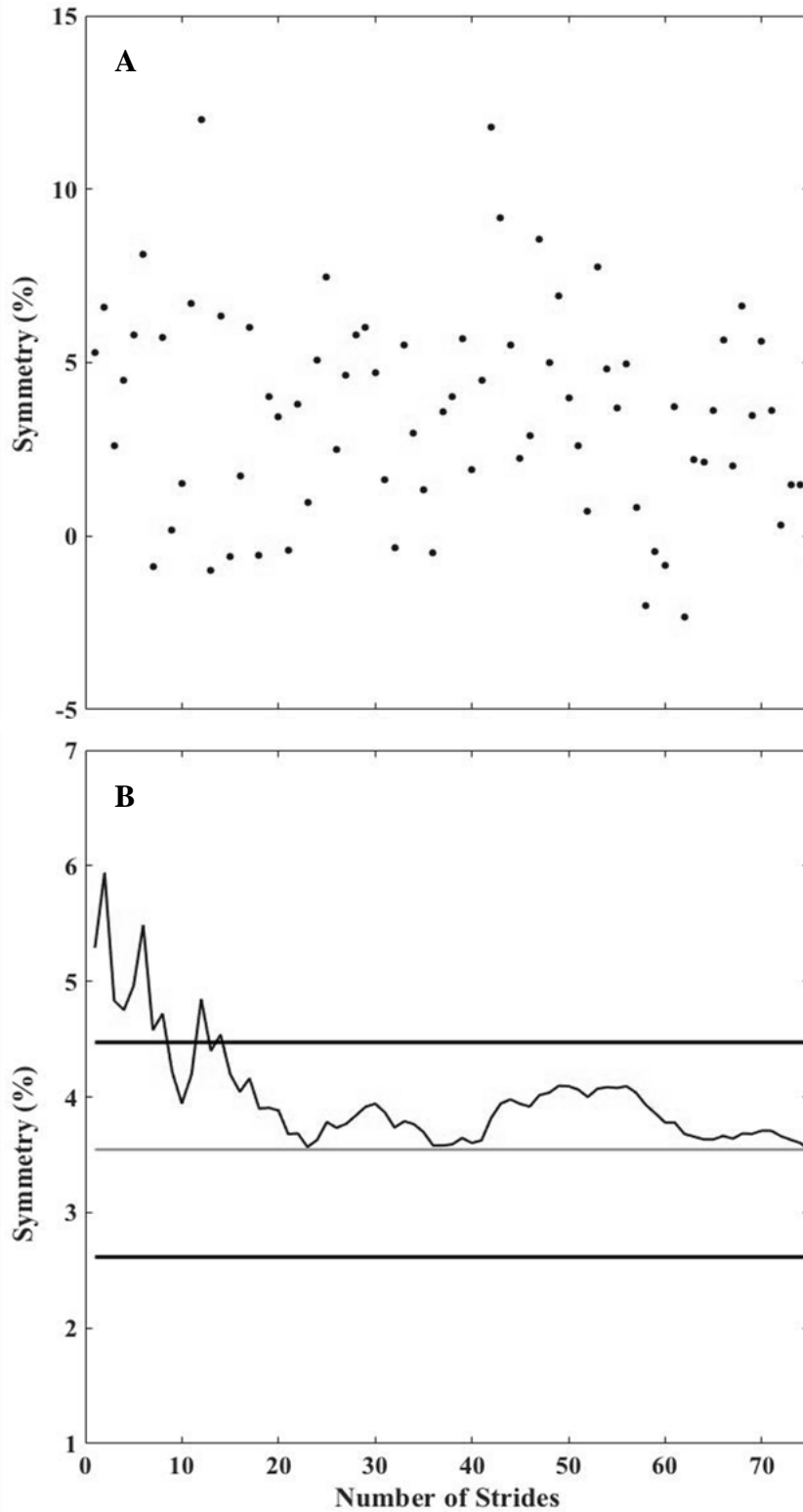
Figure 4.1 Measuring Symmetry and Determining Stable Thresholds. Individual stride symmetries were measured (A) with a sequential average measured within ± 2 SD (B).

Results

A SMS for all variables was achieved with an average of 16.4 ± 6.3 strides; however, the required number of strides varied significantly between variables (Table 4.1). Overall average SMS includes variables, such as ankle angle at initial contact and minimum knee angle, which required a greater number of strides to achieve a stable measure compared to other measures. The greater number of strides to achieve a SMS was due to measures being approximately zero, and in turn the relatively small perturbations away from zero resulted in a large amount of calculated asymmetry. For variables that average values are near zero, on average 27.5 ± 5.3 strides were required to reach a SMS per trial. On average a SMS was met in 14.1 ± 3.2 strides when excluding variables near zero. Five measures were affected by this phenomenon and were subsequently removed from analysis; in turn, 15 strides were used per bin to determine if calculation or bin of strides analyzed would alter findings.

There was a significant difference between SI and ASI calculations for all variables ($p < 0.05$), however there was no significant difference ($p > 0.05$) between bins of 15 strides (Table 4.2) when comparing the first, middle, and last strides. Further, there was no significant difference between bins of consecutive or random strides, and therefore no difference between continuous and discontinuous data collections are expected. Lastly, no significant interaction between calculation or bin factor were noted for all comparisons.

Table 4.1

Average number of strides to Reach a Stable Mean Symmetry Index.

Variables		SI	ASI
Spatiotemporal Parameters	Step Length	10	12
	Step Frequency	9	8
	Stance Time	9	10
	Step Time	9	8
	Swing Time	11	12
	STP Average	**10**	**10**
Joint Angles	Ankle Angle at IC	23	24
	Ankle Angle Maximum	17	15
	Ankle Angle Minimum	14	12
	Knee Angle at IC	24	27
	Knee Angle Maximum	14	14
	Knee Angle Minimum	38	37
	Hip Angle at IC	19	18
	Hip Angle Maximum	14	16
	Hip Angle Minimum	28	27
	JA Average	**21**	**21**
Joint Velocities	Ankle Velocity Maximum	9	14
	Ankle Velocity Minimum	12	12
	Knee Velocity Maximum	13	14
	Knee Velocity Minimum	13	17
	Hip Velocity Maximum	11	12
	Hip Velocity Minimum	15	15
	JV Average	**12**	**14**
Joint Moments	Ankle Peak Plantarflexion Moment	11	14
	Knee Peak Extension Moment	20	20
	Hip Peak Extension Moment	10	15
	JM Average	**14**	**16**
Joint Powers	Ankle Peak Power Absorption	10	14
	Ankle Peak Power Generation	20	20
	Knee Peak Power Absorption	15	19
	Knee Peak Power Generation	11	13
	Hip Peak Power Absorption	25	23
	Hip Peak Power Generation	13	15
	JP Average	**16**	**17**

Table 4.2

Mean Symmetry Index for 15 Strides.

	Variables	First 15 Strides	Middle 15 Strides	Last 15 Strides	Random 15 Strides	All 75 strides
Spatiotemporal Parameters	Step Length	0.5 ± 1.8	0.5 ± 1.7	0.6 ± 2.0	0.6 ± 1.8	0.6 ± 1.6
	Step Frequency	-0.1 ± 1.4	-0.1 ± 1.7	-0.2 ± 1.9	-0.4 ± 1.6	-0.3 ± 1.4
	Stance Time	-0.5 ± 2.2	-0.5 ± 2.3	-0.4 ± 2.5	-0.4 ± 2.3	-0.6 ± 2.3
	Step Time	0.1 ± 1.4	0.1 ± 1.7	0.2 ± 1.9	0.3 ± 1.7	0.3 ± 1.4
	Swing Time	0.3 ± 1.4	0.3 ± 1.6	0.3 ± 1.6	0.3 ± 1.6	0.4 ± 1.5
Joint Angles	Ankle Angle at IC	-0.8 ± 57.3	-1.4 ± 57.3	-0.6 ± 59.6	-1.4 ± 56.7	-0.7 ± 56.6
	Ankle Angle Maximum	5.8 ± 11.4	6.5 ± 10.8	6.4 ± 11.3	6.3 ± 11.2	6.1 ± 11.2
	Ankle Angle Minimum	-5.3 ± 12.2	-5.5 ± 11.9	-6.4 ± 11.6	-4.7 ± 11.3	-5.7 ± 11.0
	Knee Angle at IC	6.1 ± 33.8	7.2 ± 32.4	5.3 ± 29.2	7.7 ± 31.7	6.4 ± 30.8
	Knee Angle Maximum	-0.6 ± 5.7	-0.6 ± 5.6	-0.9 ± 5.9	-0.7 ± 5.7	-0.7 ± 5.7
	Knee Angle Minimum	23.2 ± 46.2	25.0 ± 45.0	23.7 ± 45.2	24.1 ± 45.3	23.1 ± 44.1
	Hip Angle at IC	-1.6 ± 9.2	-1.4 ± 9.4	-1.7 ± 9.5	-1.6 ± 9.6	-1.6 ± 9.2
	Hip Angle Maximum	-4.2 ± 11.2	-4.1 ± 11.2	-4.4 ± 11.8	-4.3 ± 11.4	-4.2 ± 11.3
	Hip Angle Minimum	1.5 ± 27.5	-0.1 ± 27.5	-0.7 ± 28.0	-0.9 ± 29.0	-0.2 ± 28.1
Joint Velocities	Ankle Velocity Maximum	-0.1 ± 5.9	0.3 ± 5.6	0.0 ± 5.6	-0.3 ± 5.6	-0.1 ± 5.5
	Ankle Velocity Minimum	3.1 ± 10.5	3.1 ± 9.1	2.7 ± 10.0	2.7 ± 9.2	2.9 ± 9.1
	Knee Velocity Maximum	-1.6 ± 7.0	-1.6 ± 6.8	-1.3 ± 7.0	-1.2 ± 7.1	-1.4 ± 6.9
	Knee Velocity Minimum	0.4 ± 6.5	0.2 ± 6.6	0.0 ± 6.6	0.3 ± 6.8	0.2 ± 6.5
	Hip Velocity Maximum	0.4 ± 5.4	0.7 ± 5.9	0.5 ± 6.3	0.5 ± 5.4	0.6 ± 5.5
	Hip Velocity Minimum	-0.5 ± 9.4	-1.0 ± 9.4	-0.8 ± 8.8	-1.4 ± 9.4	-0.9 ± 8.8
Joint Moments	Ankle Peak Plantarflexor Moment	-3.5 ± 4.7	-3.8 ± 4.8	-3.8 ± 4.8	-3.6 ± 4.8	-3.8 ± 4.6
	Knee Peak Extensor Moment	2.9 ± 11.6	4.0 ± 11.6	4.4 ± 11.3	3.1 ± 11.7	3.7 ± 11.2
	Hip Peak Extensor Moment	-8.2 ± 10.7	-7.8 ± 10.4	-8.6 ± 12.4	-7.8 ± 10.7	-8.4 ± 10.7
Joint Powers	Ankle Peak Power Absoption	1.8 ± 10.1	1.6 ± 11.0	1.5 ± 10.2	1.6 ± 10.9	1.6 ± 10.4
	Ankle Peak Power Generation	-0.6 ± 7.3	-0.5 ± 7.3	-0.9 ± 7.8	-0.6 ± 7.0	-0.8 ± 7.2
	Knee Peak Power Absoption	-1.5 ± 10.1	-1.2 ± 10.9	-0.2 ± 12.0	-1.5 ± 10.9	-1.1 ± 10.1
	Knee Peak Power Generation	2.1 ± 20.4	3.1 ± 19.1	1.5 ± 21.0	2.3 ± 19.4	1.9 ± 19.5
	Hip Peak Power Absoption	2.6 ± 13.7	3.6 ± 14.3	1.6 ± 12.8	3.9 ± 14.7	2.7 ± 13.6
	Hip Peak Power Generation	-7.1 ± 16.1	-7.4 ± 15.5	-7.4 ± 16.7	-6.8 ± 16.4	-7.1 ± 15.5

Note. Highlighted measures removed from stable symmetry measure. Negative values indicate a greater value for the left limb. No significant differences were found for all comparisons.

Discussion

The purpose of this study was to establish the minimum number of strides required to achieve a SMS of lower extremity discrete spatiotemporal, joint kinetic and joint kinematic variables. Additional aims were to determine if SMS were different when collected continuously versus discontinuously or at different times within a data collection, and to identify if two common symmetry calculations (ASI and SI) required different numbers of strides to achieve a SMS. It was determined that an average of 15 strides were required to establish a SMS, and there were no differences between the first, middle, or last strides, regardless of being collected consecutively or nonconsecutively. Furthermore, the two methods of calculating symmetry tested resulted in different symmetry values without requiring a different number of strides to achieve a SMS.

It is important to note that not all discrete measures reached a SMS in the same number of strides (Table 4.1), with some measures such as joint angles requiring a greater number of strides. This discrepancy between variables may partially be due the greater variability in the discrete measures are during normal gait (Bredeweg et al., 2013). Further differences may stem from the calculations used to measure symmetry (Pappas et al., 2015). As variables near zero, such as ankle angle at initial contact, smaller changes away from zero have a greater effect on symmetry indices compared to the same magnitude of change from a value not near zero. This limitation to symmetry indices has been previously noted, and is presently exemplified by the minimum knee angle (Carpes et al., 2010; Herzog et al., 1989). Minimum knee angle reached near zero values during the gait cycle and required nearly 40 strides to establish a SMS. As researchers apply the current findings, added emphasis may be placed on joint moments and powers thresholds.

As forces cause motion, changes to joint moments and powers can cause alterations in the joint kinematic and spatiotemporal parameter values.

Based on previous research, it was expected that the two symmetry indices would result in different symmetry values (Carpes et al., 2010). The two measures of symmetry (ASI and SI) compared were both found to require a similar number of strides to obtain a stable mean symmetry index. This implies that different data analysis techniques will not require unique alterations to data collection methods. Furthermore, with no differences in SMS values found between randomly selected and consecutive strides the same level of confidence can be placed in symmetry findings regardless of studies utilizing over ground or treadmill running.

As interlimb symmetry has been used as both a performance measure and a rehabilitative goal, understanding how to best quantify symmetry is important when establishing clinical and research protocols (Beck et al., 2018; Bredeweg et al., 2013; Gilgen-Ammann et al., 2017; Nasirzade et al., 2017). Although the current findings suggest collecting more strides than common, the findings do not conclude that dozens or hundreds of strides are required to confidently measure joint level interlimb symmetry. This is especially important in populations that may need to limit activity due to increased injury risk or those returning to sport (Bredeweg et al., 2013; Gilgen-Ammann et al., 2017; Tenforde et al., 2018; Zifchock et al., 2006). In turn, clinicians and researchers alike can be confident in the observed asymmetries when collecting 15 strides. Additionally, these scientists do not need to be concerned with measuring interlimb symmetry only during consecutive strides or in a certain time point within a gait analysis.

There are a number of ways the present study is limited in scope. In particular the current study only addresses lower extremity measures, with upper extremity measures possessing the chance of requiring a different number of strides to reach a SMS. This study is also limited to comparing the averages of discrete values and did not explore the number of strides to achieve a SMS for the variability of measures. Further, the current study is limited to exploring symmetry in a healthy and relatively symmetrical cohort. Future studies may aim to better understand how many strides are required for upper extremity measures, measures of variability, and inherently asymmetrical populations.

Conclusion

An average of 15 strides are required to achieve a stable mean symmetry index of the lower extremity spatiotemporal, joint kinematic, and joint kinetic variables. The timing within a data collection, and the method of collecting either continuous or discontinuous strides does not affect the number of strides required to achieve a stable mean symmetry index. Although the symmetry index and absolute symmetry index calculated different percent differences between limbs, there was no difference in the number of strides to achieve a stable mean between the two measures. When examining interlimb symmetry, a greater emphasis should be placed on collecting a sufficient number of strides to achieve a stable mean than emphasizing strides be collected continuously or at a specific time point within the data collection.

CHAPTER V

STUDY TWO: INFLUENCE OF NUMBER OF STRIDES
ANALZYED ON MEAN KINEMATIC SYMMETRY
INDICIES DURING WALKING

Introduction

Interlimb symmetry is a common goal for clinicians as they work with those with

unilateral deficiencies (Louw & Deary, 2014; Nasirzade et al., 2017; Wanamaker et al.,

2017). When gait asymmetries are difficult to quantify in real time, kinematic gait

analyses are a powerful tool (Baker, Esquenazi, Benedetti, & Desloovere, 2016). In turn,

clinicians and researchers are more frequently relying on symmetry indices from gait

analyses to describe the interlimb differences (Nasirzade et al., 2017).

One common measure of interlimb symmetry is the ratio of differences between

limbs (Robinson et al., 1987). This symmetry index (SI) provides insight into the

magnitude and direction of the asymmetry for each measure and has been applied to

active and clinical populations (Carpes et al., 2010; Smith & Martin, 2007). The SI

requires discrete measures for each limb during a gait cycle and can be compared over

multiple strides to give a better indication of an individual's average symmetry.

In previous studies the number of strides utilized to describe interlimb symmetry

has been limited by technology and the computing time required to process positional

data. More recently, with improvements in processing, as many as ~350 strides have been

used in studying walking mechanics (Owings & Grabiner, 2003). Although more strides

are able to be analyzed, researchers and clinicians can still be limited in the number of

strides analyzed due to the physical abilities of participants with unilateral deficiencies (Adamczyk & Kuo, 2015; Mattes et al., 2000; Sadeghi et al., 2000; Sanderson & Martin, 1997; Smith, Royer, et al., 2013). With the need to find a balance between accurately quantifying symmetry and the need to limit the number of strides asked of a participant, a better understanding of the number of strides needed to achieve a stable mean symmetry index (SMS).

Methodologies used in gait analyses can also vary greatly, in particular treadmill protocols allow for consecutive strides to be analyzed, while overground protocols inherently result in discontinuous strides being analyzed. During walking, adaptations to gait asymmetries are acutely addressed by strides following an asymmetrical perturbation (Kozlowska et al., 2017; Sadeghi et al., 2000). In turn, the analysis of discontinuous strides may result in different findings as this acute response is not necessarily observed. Therefore, a better understanding is required of how continuous and discontinuous strides effects the results of a SI analysis.

Although a greater number of strides may be required to achieve a SMS, previous research has commonly used three to five strides (Carpes et al., 2010; Nasirzade et al., 2017; Sadeghi et al., 2000). It is important to better understand whether means from three or five strides will differ from a mean derived with a greater number of strides; i.e. the number of strides to achieve a SMS. This will allow for a greater rationalization of how many strides should be collected in a gait analysis focused on kinematic symmetry.

This study aims to establish the average number of strides to achieve a SMS for lower extremity kinematics. Further, the purpose of this study is to understand if the order or number of strides will result in a different symmetry index value. Lastly, the study

aims to understand if there is a difference between the calculated interlimb symmetry of the first three strides, first five strides, and the first number of strides to achieve a SMS.

Methodology

Participants

Ten individuals (F = 6, 1.73 ± 0.12 m, 66.0 ± 12.0 kg, 25 ± 3 years, 326 ± 178 min/week of activity) participated in the study, with all free of lower extremity injury for at least six months prior to data collections and reported no history of surgery on the lower extremity or trunk. Participants completed all conditions in their own shoes, all of which were traditional, non-minimalist, footwear. Along with the pre-participation questionnaire, participants were presented with an informed consent document, procedures were verbally explained, and written consent was obtained; with a copy of the informed consent offered to the participant. The Institutional Review Board at the University of Northern Colorado approved this study.

Data Collection

After providing informed written consent, participants answered a modified physical activity questionnaire and survey to determine participant eligibility and provide demographic details. Form fitting clothes were provided to allow retroreflective markers to be placed over bony landmarks of the pelvis, thigh, shank, and foot of both extremities. Segment position data were collected using clusters of reflective markers, with the joint axis identified with additional markers during a standing calibration trial. Motion (100 Hz) data was captured using a ten-camera motion analysis system (VICON, Englewood, CO). Participants completed two sessions of walking at 1.5 m·s-1, in which each session consisted of walking for nine minutes.

Within each session, two 60 second trials were collected at the 6-and 8-minute marks. The first 50 consecutive strides of each 60 second trial was used for analysis.

Gap-filled motion data were exported to Visual 3D (C-Motion, Germantown, MD) where data were filtered with a lowpass 4th order, zero lag, Butterworth filter with a cutoff frequency of 6 Hz. Spatiotemporal and joint kinematic variables were calculated for hip, knee, and ankle joint angles and velocities, with discrete measures determined for variable maximums and minimums throughout a stride.

Data Analysis

Interlimb symmetry was calculated in MATLAB (MathWorks, Natick, MA) between discrete variables of the left and right limbs within a single stride, resulting in 50 individual symmetry values for each variable of each trial. Symmetry values were calculated using a symmetry index (SI). Robinson and colleagues used equation (1) when assessing symmetry of ground reaction forces between the right (X_r) and left (X_l) measures (Robinson et al., 1987). Perfect symmetry is represented by zero. Mean symmetry for each walking trial will be determined by computing an average for all 50 strides.

$$SI_\% = \left[\frac{(X_r - X_l)}{\frac{1}{2}(X_r + X_l)} \right] \cdot 100 \tag{1}$$

A SMS was determined by taking a sequential average of each individual trial and noting when the sequential average remained within a ±2 SD range about the mean symmetry value of the entire trial. A ±2 SD range was used so that a SMS would be reached for the majority of data points within a normal bell curve. Individual trials were compared to identify whether any trial possessed a spurious stride of asymmetry; none were noted.

Subsets of consecutive strides were created within each trial to compare symmetry indices. Subset sizes were chosen to reflect the minimum number of strides collected when studying gait kinematics (three and five) and the number of strides to achieve a SMS. Additional subsets, of the same sizes mentioned, were created using random strides within a trial to determine if differences would be found within a trial with inconsecutive strides.

Statistical Analysis

A series of analysis of variance (ANOVA) ($p < 0.05$) with repeated measures were used to determine whether same sized subsets differed within a trial, with each trial being treated as an individual participant. Since between-trial comparisons will not be examined, each trial was treated as individual subjects (Zucker, Ruthazer, & Schmid, 2010). Additional ANOVAs with repeated measures were used to determine if differences in the number of strides within a subset would result in significant differences. All statistical analyses were completed using R (version 3.4.1).

<div align="center">

Results

</div>

SMS was achieved within an average of 8.3 ± 0.8 strides for all variables, ranging from seven to ten strides. Therefore, eight strides were used as the additional subset when comparing groups of strides. Few differences were noted between subsets of strides within trials, with no discernable trends suggesting that consecutive strides were required (Table 5.1). The measures with statistical differences are presented in Table 5.2 to demonstrate the relatively small differences between subsets. Further, no differences were found between the averages of the first three, five, and eight strides for all variables ($p > 0.05$) (Table 5.3).

Table 5.1

P values of Comparisons between Stride Subsets.

	Subsets of 3	Subsets of 5	Subsets of 8	Random Subsets of 3	Random Subsets of 5	Random Subsets of 8
Ankle Angle Max	0.77	0.51	0.50	0.07	0.40	**0.02***
Ankle Angle Min	0.18	0.38	0.69	0.60	0.49	0.17
Ankle Velocity Max	0.17	0.52	0.92	0.06	**0.04***	0.83
Ankle Velocity Min	**0.01***	**0.01***	0.31	0.16	0.61	**0.04***
Hip Angle Max	0.97	0.72	0.42	**0.01***	0.11	0.16
Hip Angle Min	0.16	0.05	0.37	0.61	0.59	0.89
Hip Velocity Max	0.17	0.34	0.19	0.68	0.18	0.51
Hip Velocity Min	0.83	0.86	0.94	0.54	0.77	0.98
Knee Angle Max	0.52	0.20	0.94	0.51	0.86	0.98
Knee Angle Min	0.92	0.58	0.75	0.86	0.40	0.27
Knee Velocity Max	0.60	0.75	0.41	0.51	0.46	0.49
Knee Velocity Min	0.24	0.72	0.95	0.84	0.29	0.27
Stance Time	0.41	0.36	0.39	0.27	0.12	0.89
Step Length	0.81	0.75	0.21	0.78	0.93	0.87
Step Time	0.30	0.86	0.06	0.63	0.45	0.57
Swing Time	0.65	0.65	0.53	0.59	0.19	0.76

Note. Comparisons were made between three, five, and eight subsets either consecutively or randomly assigned.

* Indicates a significant difference ($p < 0.05$) between one of the subsets within a trial.

Table 5.2

Comparisons of Measures with Significantly Different Subset Symmetries.

	Random Subsets of 8 Ankle Angle Max	Random Subsets of 8 Ankle Velocity Min	Random Subsets of 5 Ankle Velocity Max	Subsets of 5 Ankle Velocity Min	Random Subsets of 3 Hip Angle Max	Subsets of 3 Ankle Velocity Min
Subset 1	5.7 ± 22.1	7.3 ± 16.8	4.5 ± 8.9	2.0 ± 16.5	-1.5 ± 8.6	1.4 ± 18.2
Subset 2	5.9 ± 23.9	7.2 ± 14.2	4.7 ± 8.8	4.7 ± 18.9	-2.1 ± 9.7	5.1 ± 15.0
Subset 3	7.9 ± 22.3	6.3 ± 15.4	5.4 ± 9.2	3.8 ± 16.9	-1.1 ± 9.5	5.0 ± 18.1
Subset 4	7.5 ± 22.1	4.8 ± 18.7	4.2 ± 9.5	5.2 ± 18.1	-1.6 ± 9.5	5.0 ± 18.5
Subset 5	8.5 ± 22.4	4.2 ± 17.7	4.0 ± 9.1	6.7 ± 15.1	-1.3 ± 9.3	3.8 ± 16.9
Subset 6	9.9 ± 23.3	4.4 ± 17.1	4.2 ± 9.6	8.1 ± 15.9	-1.9 ± 9.9	6.1 ± 16.1
Subset 7			3.4 ± 9.9	6.2 ± 15.3	-1.2 ± 9.8	5.9 ± 17.6
Subset 8			4.1 ± 8.7	4.9 ± 15.4	-1.6 ± 9.1	8.2 ± 16.6
Subset 9			3.8 ± 9.5	7.0 ± 17.8	-1.3 ± 9.2	4.7 ± 17.3
Subset 10			3.8 ± 9.2	6.9 ± 14.4	-1.2 ± 9.4	8.1 ± 15.9
Subset 11					-1.4 ± 9.1	6.4 ± 15.7
Subset 12					-1.1 ± 9.3	7.8 ± 14.6
Subset 13					-1.4 ± 9.1	6.4 ± 15.6
Subset 14					-0.9 ± 9.5	6.8 ± 17.4
Subset 15					-1.1 ± 9.2	7.0 ± 17.8
Subset 16					-0.8 ± 9.7	4.6 ± 16.9
Measure Avg	7.6 ± 22.3	5.7 ± 16.3	4.2 ± 9.1	5.6 ± 16.2	-1.3 ± 9.3	5.8 ± 16.6

Table 5.3

P value and Symmetry of First Subsets of Strides.

	p value	First 3 Strides	First 5 Strides	First 8 Strides
Ankle Angle Max	0.18	9.7 ± 23.1	8.1 ± 23.2	7.5 ± 23.2
Ankle Angle Min	0.30	-0.2 ± 18.9	0.4 ± 18.4	1.5 ± 20.9
Ankle Velocity Max	0.90	3.5 ± 9.7	3.7 ± 10.0	3.6 ± 8.8
Ankle Velocity Min	0.07	1.4 ± 18.2	2.0 ± 16.5	5.0 ± 16.3
Hip Angle Max	0.37	-0.7 ± 9.2	-1.2 ± 9.7	-1.1 ± 9.2
Hip Angle Min	0.81	7.0 ± 14.1	5.9 ± 14.3	7.2 ± 15.8
Hip Velocity Max	0.99	1.5 ± 6.5	1.1 ± 6.8	1.5 ± 6.5
Hip Velocity Min	0.44	1.5 ± 11.5	0.1 ± 10.7	0.7 ± 10.7
Knee Angle Max	0.73	0.1 ± 5.5	-0.1 ± 5.3	0.2 ± 4.9
Knee Angle Min	0.50	-13.9 ± 35.8	-14.7 ± 38.4	-15.6 ± 36.7
Knee Velocity Max	0.79	-1.1 ± 5.6	-1.9 ± 6.2	-1.3 ± 5.7
Knee Velocity Min	0.86	1.9 ± 5.1	1.4 ± 5.2	1.8 ± 5.0
Stance Time	0.33	0.5 ± 1.4	0.5 ± 1.2	0.7 ± 1.0
Step Length	0.47	0.2 ± 2.6	0.3 ± 2.7	0.1 ± 2.8
Step Time	0.24	-0.8 ± 1.6	-0.8 ± 1.7	-1.2 ± 1.8
Swing Time	0.96	-0.7 ± 2.2	-0.9 ± 2.1	-0.8 ± 1.6

Note. Negative values represent a greater value on the left limb measure.

Discussion

The purpose of this study was to establish the minimum number of strides required to achieve a SMS of lower extremity joint kinematics and spatiotemporal variables. Additional aims were to determine if measures would be affected by analyzing consecutive or nonconsecutive strides, and if any significant differences would be noted between the first three, five, and the average number of strides to achieve a SMS. On average, a lower extremity measure achieved a SMS with eight strides of data, and no differences were noted between consecutive and nonconsecutive strides. Further, no differences were noted between the first three, five, and eight strides for all measures.

The magnitudes of asymmetry in lower extremity joint kinematics and spatiotemporal measures are consistent with previous findings (Diop et al., 2004; Hesse et al., 2003; Nasirzade et al., 2017; Nessler et al., 2015; Nolan, 2008). The relatively higher asymmetries in knee angle minimum is also consistent with previous findings where measures near zero have been found to result in a greater calculated asymmetry (Herzog et al., 1989). Although not all previous findings utilized eight strides, the present findings suggest that no significantly different findings would be expected if they had. This supports that a stable measure is not necessary to achieve an accurate measure, however a measure based off of fewer strides will be at an increased risk of being effected by a stride with spurious asymmetry, as previously noted (Herzog et al., 1989).

The lack of systematic differences between measures with consecutive and random subsets of strides has multiple implications. First, there should not be a difference in measured symmetry between data collections that analyze continuous strides via treadmill, and those analyzing discontinuous strides via overground walking. Although the present study did not utilize overground walking, in turn there is a chance that differences are present in SI values between treadmill and overground walking, these differences should not stem from calculating the interlimb symmetry (Lee & Hidler, 2008). Secondly, it would be reasonable to expect intermittent breaks between trials will not affect the calculated SI. That being said, previous research has suggested that gait symmetry can vary day to day by upwards of four degrees of range of motion (Wolf et al., 2009). This suggests that SI can be calculated more reliably within a session but has a greater chance to differ if calculated from intersession strides. Lastly, either acute adjustments to asymmetrical steps did not occur in the healthy cohort, or the SI was

robust enough not to be altered by a given step. Previous research has noted that asymmetrical steps result in subsequent step adjustments and that the body possesses the capacity to acutely compensate (Kozlowska et al., 2017). As Kozlowska and colleagues recruited a similar cohort and used similar methodologies, it is more likely that the acute adjustments were occurring. However, these adjustments were not drastic enough to result in a difference between the smaller subsets of strides, the subsets most vulnerable to spurious strides.

As interlimb symmetry is a common clinical goal for returning to sport in athletic populations that have suffered a unilateral musculoskeletal injury and a long-term goal of individuals with more systemic unilateral deficiencies; being able to confidently establish a symmetry index is vital to clinicians and researchers alike. The current findings do suggest that a more stable measure is achieved with eight strides of walking, regardless of analyzing continuous strides or not. Compared with the average number of strides to achieve a SMS for running variables, walking required relatively fewer strides. This may partially be explained by walking not including a flight phase, a phase that which may make running more variant relative to walking. The SMS found during walking should provide the sport scientist with the most confident and consistent findings as the mean should not fall outside of a two standard deviation window. However, these findings also suggest that if needed, we should not expect to find a statistically different result if fewer strides are analyzed. This should allow for the flexibility to collect fewer strides when needed, depending on the individual's needs.

Although the application of the present findings is aimed at assisting researchers that work with clinical populations, the application may be limited due to the healthy, relatively symmetrical, cohort recruited. In particular, populations such as unilateral amputees commonly present with more variant gait patterns, with some asymmetries possibly unavoidable (Adamczyk & Kuo, 2015; Arellano et al., 2015). This greater variability may increase the number of strides required to achieve a SMS, and in turn may result in significantly different SI than found in smaller subsets of strides. Future studies should aim to address a more population specific stride threshold when attempting to calculate SI during walking.

Conclusion

An average of eight strides are required to achieve a stable mean symmetry index of the lower extremity joint kinematics and spatiotemporal measures. No trend was determined to indicate a difference between findings of continuous and random strides; suggesting no difference would be found between symmetry indices during overground and treadmill walking for most measures. Although a stable symmetry index was achieved after eight strides for most measures, no difference was noted between the first three, five, and eight strides for all measures. This suggests that although a measure may be more likely to be affected by a single stride, there should be no difference between means with less strides analyzed in a healthy population.

CHAPTER VI

STUDY THREE: INFLUENCE OF ASYMMETRY ON
THE METABOLIC COSTS OF WALKING

Introduction

The Amputee Coalition of America estimates two million people are living with a

major amputation; expecting the population to double by 2050. Specifically, the number

of persons with transtibial amputations (TTA) is growing with an increased rate of distal

limb amputations via complications of type-2 diabetes, advancements in lifesaving

medical treatments, and improved protective military gear; with 42% of combat related

amputations occurring at the TTA level (Belatti & Phisitkul, 2013; Epstein et al., 2010;

Krueger et al., 2012). Although the hip and knee joints are largely preserved on the

affected limb with a unilateral TTA, the loss of the ankle and associated musculature

results in a number of mechanical challenges that the individual must overcome to

successfully locomote (Czerniecki & Morgenroth, 2017; Hak et al., 2014; Mattes et al.,

2000; Nolan, 2008; Wanamaker et al., 2017; Warren et al., 2014).

Following surgery, a person with a unilateral TTA is fitted with a light-weight

prosthetic limb, resulting in an interlimb mechanical asymmetry. The prosthesis differs

from the intact limb not only in the reduced capacity to propel during terminal stance but

includes a reduction in mass and altered inertial properties (Czerniecki et al., 1994;

Gitter, Czerniecki, & DeGroot, 1991; Mattes et al., 2000). In spite of advancements in

prosthetic design, the mechanical asymmetries are thought to contribute to the

asymmetrical gait patterns during prosthetic use (Czerniecki & Morgenroth, 2017; Mena

et al., 1981; Nolan, 2008). Such interlimb temporal deviations include shorter contact times and longer swing times of the affected limb compared with the intact limb (Adamczyk & Kuo, 2015; Czerniecki et al., 1994; Sanderson & Martin, 1996).

Individuals with a TTA also exhibit increased metabolic costs during walking compared with those without an amputation (Mengelkoch et al., 2014). When walking, individuals with a unilateral TTA have an increased oxygen consumption (~20-30%) at self-selected slower speeds compared with healthy controls (Gailey et al., 1994; Mengelkoch et al., 2014; Waters & Mulroy, 1999). This increase in the metabolic cost of walking may partially explain why the majority of persons with amputations do not engage in a sufficient amount of physical activity to avoid comorbidities of a sedentary lifestyle (Langford, Dillon, Granger, & Barr, 2019).

Post amputation, clinicians work to restore a symmetrical gait pattern in an attempt to limit the perception of a pathological gait, facilitate physical activity, and prevent chronic complications of asymmetrical loading of the intact limb (Cutti, Verni, Migliore, Amoresano, & Raggi, 2018; Handžic & Reed, 2015; Highsmith et al., 2016). In spite of rehabilitation and gait retraining, the inherent mechanical interlimb asymmetry may hinder the ability to walk with symmetrical spatiotemporal parameters (Adamczyk & Kuo, 2009; Hof, van Bockel, Schoppen, & Postema, 2007; Winter & Sienko, 1988). These mechanical and gait asymmetries provide the foundation of a metabolic penalty unique to persons with unilateral amputations.

Although other health factors, such as psychosocial, may be positively affected by training a symmetrical gait pattern, the energetic consequences of forcing symmetry of an asymmetrical system is not fully understood (Adamczyk & Kuo, 2009; Cutti et al., 2018;

Hof et al., 2007; Mattes et al., 2000; Winter & Sienko, 1988). Previous research has utilized the addition of a unilateral load to replicate the magnitude of mechanical asymmetry between a prosthetic and intact limb for a healthy participant (Mattes et al., 2000; Smith & Martin, 2007; Smith, Villa, et al., 2013). Although location and magnitude of load effects the mechanics and the metabolic cost of walking (Browning, Modica, Kram, & Goswami, 2007; Royer & Martin, 2005), a 2 kg mass at the ankle has been widely used as it generates a similar magnitude of metabolic demand in able-bodied persons as those with a TTA (Mattes et al., 2000; Noble & Prentice, 2006; Smith & Martin, 2007; Smith, Villa, et al., 2013). The estimated difference between intact limb and prosthetic limb masses is approximately 2.3 kg, and ranged from 1.8-2.7 kg (Smith & Martin, 2013).

Other research has replicated asymmetrical gait in able-bodied individuals via an audible metronome that produces an asymmetrical beat (Beck et al., 2018; Ellis et al., 2013). Without physically loading an individual limb, this method of unilateral perturbation resulted in asymmetrical temporal parameters (Ellis et al., 2013). Specifically, the manipulated gait resulted in altered swing and stance times where participants adapted an increased swing time and a contralateral reduction in stance time. These adaptations are similar to previous findings on persons with a unilateral load and a person with a unilateral TTA (Ellis et al., 2013; Mattes et al., 2000; Sanderson & Martin, 1997; Smith & Martin, 2007). This temporal asymmetry increased metabolic demand by approximately 20-30% during walking (Ellis et al., 2013).

In either case of persons with unilateral TTA or able-bodied individuals asymmetrically manipulated, it is not understood how much the mechanical and spatiotemporal asymmetries contribute to the metabolic cost of asymmetrical walking. More simply, it is not clear whether unilateral load or asymmetrical swing times would result in a greater energetic penalty during walking. Therefore, the purpose of this study was to determine if walking with a unilaterally added mass and the associated asymmetrical swing time will individually increase the metabolic cost of walking, and to determine whether the factor of symmetry or added load had a greater effect on the metabolic cost of walking.

Methodology

Participants

Ten active persons (F = 6, 1.73 ± 0.09 m, 66.5 ± 14.3 kg, 27 ± 2 years, 275 ± 143 min/week of activity) were recruited from the student population at the University of Northern Colorado. Participants successful inclusion in the study was determined from a pre-participation, modified physical activity readiness questionnaire, and based on the following criteria: 18-30 years old, free of any existing neuromuscular or skeletal injury or condition that may prevent them from completing all tasks, injury free in the trunk and lower extremity within the last six months, and average at least 150 minutes of moderate or 75 minutes of vigorous physical activity a week. The Institutional Review Board at the University of Northern Colorado approved this study. Along with the pre-participation questionnaire, participants were presented with an informed consent document, procedures were verbally explained, and written consent was obtained with a copy of the informed consent offered to the participant.

Data Collection

  Baseline and manipulation conditions. The data collection was structured to allow for two baseline conditions and two subsequent manipulation conditions to be collected in one session. The baseline conditions were designed to represent normal walking in symmetrically unweighted and unilaterally weighted walking. The first baseline condition, No Load Symmetrical (NLS), was normal unperturbed walking where participants walked with approximately 0% swing time asymmetry, as expected in a healthy population. The second baseline condition, Load Asymmetrical (LA), was normal perturbed walking where participants walked with a unilaterally added 2kg mass and developed approximately 6% swing time asymmetry, similar to a population with a unilateral transtibial amputation (Sanderson & Martin, 1997). The added mass was located at the distal shank of the dominant limb. These two baseline conditions provided the percent offset for the following two manipulation conditions.

  Manipulation conditions were designed to replicate gait manipulations where symmetry or asymmetry were forced and do not represent the natural gait patterns used in the baseline conditions. The first manipulation condition, Load Symmetrical (LS), required the participant to walk with a unilaterally added mass while maintaining an approximately 0% asymmetrical swing time. The LS condition utilized the normal swing time symmetry from baseline condition NLS. The second manipulation condition, No Load Asymmetrical (NLA), required the participant to walk without a unilaterally added mass but to adapt the swing time asymmetry measured during the baseline condition LS. These manipulations were achieved using an audible metronome and visual biofeedback explained in more detail in the following section.

Conditions were pseudo-randomized by counterbalancing within baseline and manipulation conditions across participants. However, as manipulation conditions were dependent on baseline conditions, the four conditions were not completely counterbalanced as baseline conditions were always collected prior to manipulation conditions.

Overview of data collection. Participants changed into form-fitting clothing for data collection purposes and were provided a pair of Brooks Launch 5 athletic shoes to complete all conditions in (Brooks Running, Seattle, WA). Demographic information and other health metrics (such as age, height, weight, leg dominance, and leg lengths) were collected from the participants. Dominant leg was determined by asking participants which leg they would use to kick a ball for distance. Leg length was measured from the Anterior Superior Iliac Spine to the ipsilateral Medial Malleolus. Individual retroreflective 14 mm markers were placed over bony landmarks on their trunk and lower and upper extremities to allow for the participants movements to be captured. Additional light-weight thermoplastic plates with clusters of retroreflective 14 mm marker were placed over upper and lower extremities to track segment motion. A 10-camera motion capture system (200 Hz) was used to capture motion data (VICON, Englewood, CO). Participants were also asked to wear a mask designed to cover the mouth and nose that allowed for the collection of expired air. Gas exchange and metabolic cost were measured via indirect calorimetry with a TrueOne 2400 metabolic cart (Parvo Medics, Sandy, UT). A heart rate monitor was also used to capture the average heart rate during each condition.

Participants first stood quietly for five minutes to provide a baseline energetic demand of standing and then walked at 1.5 m·s-1 for three minutes acclimating to walking on the laboratory treadmill and to the auditory metronome and visual feedback used to manipulate symmetry during the two manipulation conditions. All conditions were collected on a tandem-belt instrumented treadmill (AMTI, Watertown, MA), allowing for ground reaction forces to be collected for consecutive steps of walking (2000 Hz). Each of the four walking conditions included eight minutes of walking at 1.5 m·s-1 while wearing the metabolic mask. The average stride rate, symmetry of swing times, along with the absolute left, right, and average swing time were calculated by streaming ground reaction forces from the instrumented treadmill into a MATLAB script (MathWorks, Natick, MA).

As explained previously, the data collection included two baseline (NLS, LA) and two manipulation (NLA, LS) conditions. Both manipulation conditions utilized an audible metronome and visual biofeedback to assist participants in obtaining the preferred asymmetrical or symmetrical walking pattern. Participants were asked to walk to the audible metronome, where initial contact of each foot coincided with a crisp beep from a portable speaker located on the front of the treadmill. This metronome utilized the average stride rate and symmetry of swing times, previously measured during baseline conditions, within MATLAB to generate the desired symmetrical or asymmetrical beat pattern. For NLA, the metronome could be offset by the desired percent of asymmetry for the individual based on the percent swing time asymmetry during the baseline LA condition. The metronome percent symmetry offset for the LS condition was set to the NLS baseline condition symmetry.

The visual biofeedback was calculated from the same real-time streaming MATLAB script mentioned previously and provided an update to the participant every five seconds. The biofeedback was displayed on a computer monitor positioned at eye level in front of the treadmill where two stacked bar graphs displayed the goal symmetry of swing times for the given condition with superimposed bars of the actual swing times, where longer graphs represented increased swing times. Participants also received verbal coaching throughout the conditions, from the research team, to reduce the amount of error away from the desired swing time symmetry.

Data Analysis

Rate of oxygen consumption and carbon dioxide production during quiet stance and the four conditions were averaged over the last two minutes of each condition. The average metabolic rates were then used to calculate the metabolic cost of standing and walking, respectively. Metabolic cost was calculated by dividing the rate of the oxygen consumption per meter divided by the participant's mass (O_2 ml·m$_{-1}$·kg$_{-1}$). Net normalized metabolic cost of walking was calculated by subtracting the metabolic cost of quiet stance from each condition's average metabolic cost of walking. Average heart rate and respiratory exchange ratio (RER) were also used to characterize the metabolic findings and to ensure participants were in a steady-state.

Walking motion data were analyzed with a low-pass, 4_{th} order, zero lag Butterworth digital filter used to smooth marker trajectories ($F_c = 6$ Hz). All calculations utilized participant specific models created in Visual3D (C-motion, Germantown, MD), with spatiotemporal parameter calculations including step length, stance time, and swing time based off of gait events within Visual3D. Gait events were determined directly from

ground reaction force data. Step by step measures were exported and an interlimb

symmetry index calculated for each spatiotemporal parameter (Robinson et al., 1987).

The average of the first 75 strides, from the last 90 seconds of each condition, were used

for analysis.

Statistical Analysis

Error between swing times of paired baseline and manipulation conditions (NLS

vs LS and LA vs NLA) were compared as percent differences to ensure participants were

accurately completing the two manipulation conditions. A repeated measure analysis of

variance (ANOVA) with an $\alpha = 0.05$ was performed in SPSS 24 (SPSS Inc., IBM,

Chicago, IL) to determine if the metabolic cost of walking differed between conditions. A

two factor (load x swing time) ANOVA with an $\alpha = 0.05$ was performed on net metabolic

cost of walking data with interactions and pairwise comparisons to further understand

where differences between factors occurred. A final two factor (load x manipulation)

ANOVA with an $\alpha = 0.05$ was performed to determine if an interaction occurred between

load and manipulation (baseline versus manipulation conditions), and whether forcing

gait symmetry away from natural adaptations to unilateral load resulted in a greater net

metabolic cost of walking. Spatiotemporal measures were also compared with a series of

repeated measure ANOVAs to further explain conditional adaptations.

<div align="center">

Results

</div>

All participants (n = 10) were determined to be right leg dominant with an

average leg length difference of 0.4 ± 0.3 cm, with no participant exceeding a 1.0 cm

interlimb discrepancy. On average, it had been 6 ± 5 hours since participants had a meal

prior to coming into the biomechanics lab. Participants averaged less than 2% error in

matching the goal swing time symmetry established during baseline conditions during the

subsequent manipulation conditions (Figure 6.1).
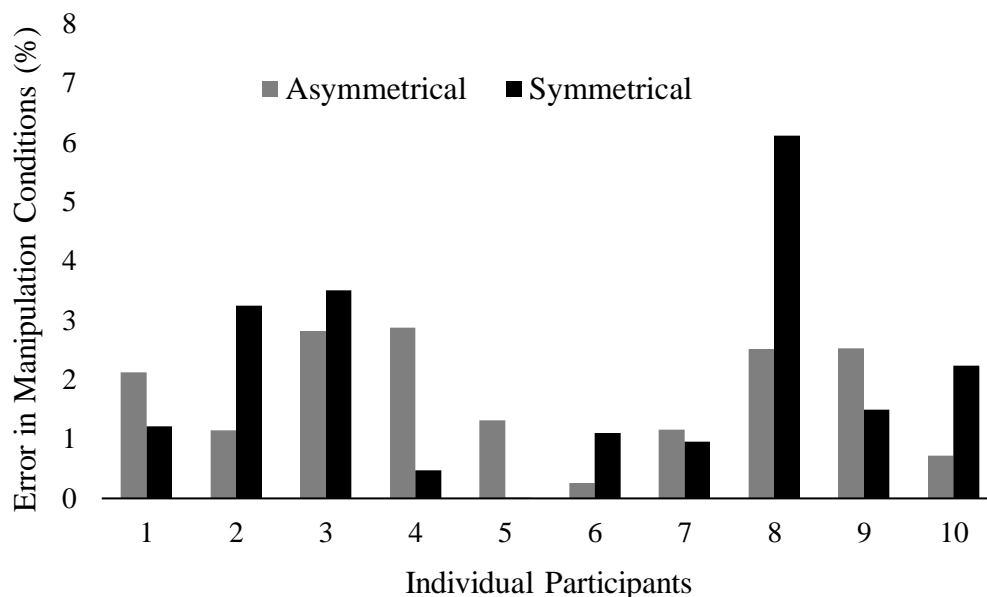
Figure 6.1 Error in Manipulation Conditions. Individual participant error in achieving the same swing time symmetry during the manipulation conditions as compared with the baseline conditions.

In general, the findings reveal a significant increase in the net metabolic cost of

walking for all conditions compared with the baseline normal walking (NLS) data. This

confirms the hypothesis that walking with a unilateral mass and the associated

asymmetrical swing time will individually increase the metabolic cost of walking.

Further, there was an interaction between load and swing time symmetries where

metabolic costs increased when forcing non-normal swing time symmetry, e.g. a greater

cost during NLA compared with NLS and a greater cost during LS compared with LA.

This is supported by manipulation conditions resulting in a significantly greater

metabolic cost than baseline conditions. In turn, one factor of load or swing time

symmetry may not have a greater effect on the metabolic cost of walking, but rather a

manipulation from the natural gait pattern symmetry increases the metabolic costs of walking. Specific statistical comparisons are provided below.

A repeated measure ANOVA of the four conditional net metabolic cost of walking revealed significant differences, with pairwise comparisons showing a significant difference ($p < 0.05$) between NLS baseline condition and all other conditions (Figure 6.2). The two factor (load x swing time) repeated measure ANOVA revealed a significant interaction between load and swing time factors ($p < 0.01$) with a significant difference in the net metabolic cost of walking load factor ($p = 0.003$) and not the swing time symmetry factor ($p = 0.521$). An additional two factor (load x manipulation) repeated measure ANOVA revealed no interaction between load and manipulation but indicated a significant main effect for manipulation ($p < 0.01$) (Figure 6.3).

A series of repeated measure ANOVAs on spatiotemporal parameters revealed significant differences between conditions for swing time, stance time, and step length symmetries (Figure 6.4). Pairwise comparisons for swing time and stance time symmetry revealed a significant difference ($p < 0.05$) between NLS and LS vs NLA and LA conditions but not within symmetrical or asymmetrical swing and stance time conditions. However, only LA vs NLS reported a significant difference ($p = 0.02$) in step length symmetry due to the high amount of variability within conditions (Figure 6.4).
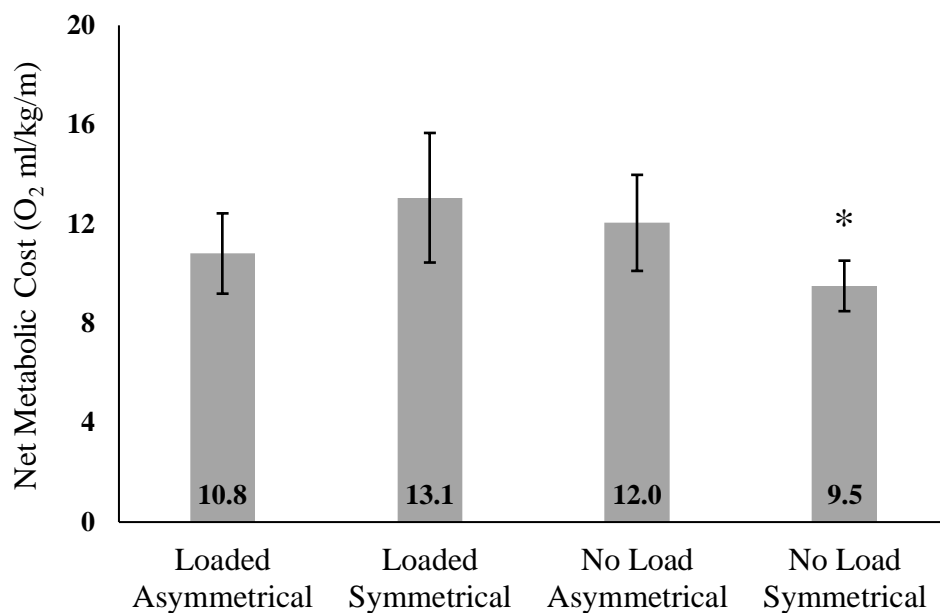
Figure 6.2 Metabolic Cost of Walking by Condition. Asterisk denotes a significant difference on the net metabolic cost of walking between the No Load Symmetrical condition from all other conditions.
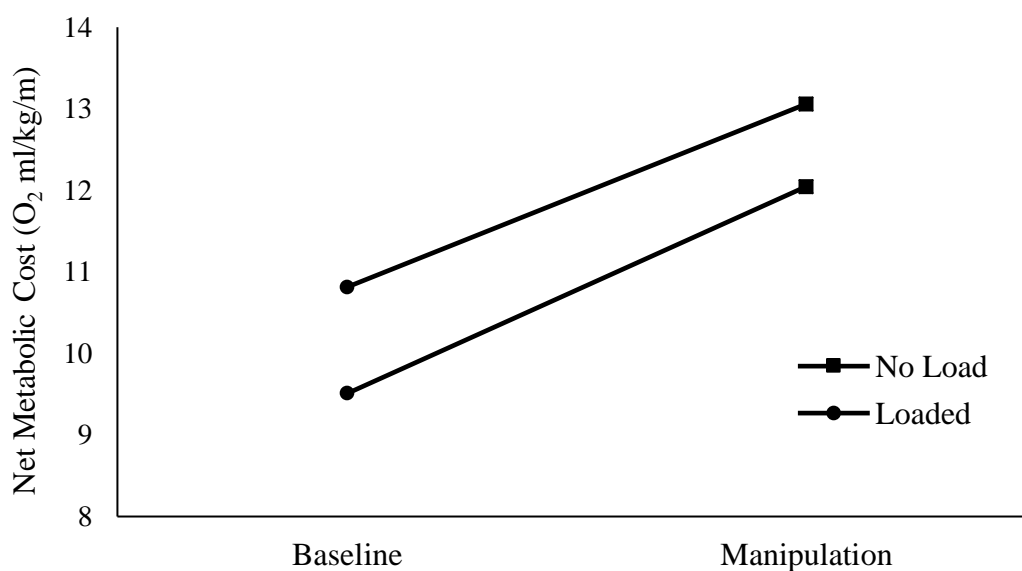
Figure 6.3 Main Effect of Manipulating Swing Time Symmetry. No interaction was noted between factors of load and manipulation, however both main effects were found to be significant ($p < 0.05$).
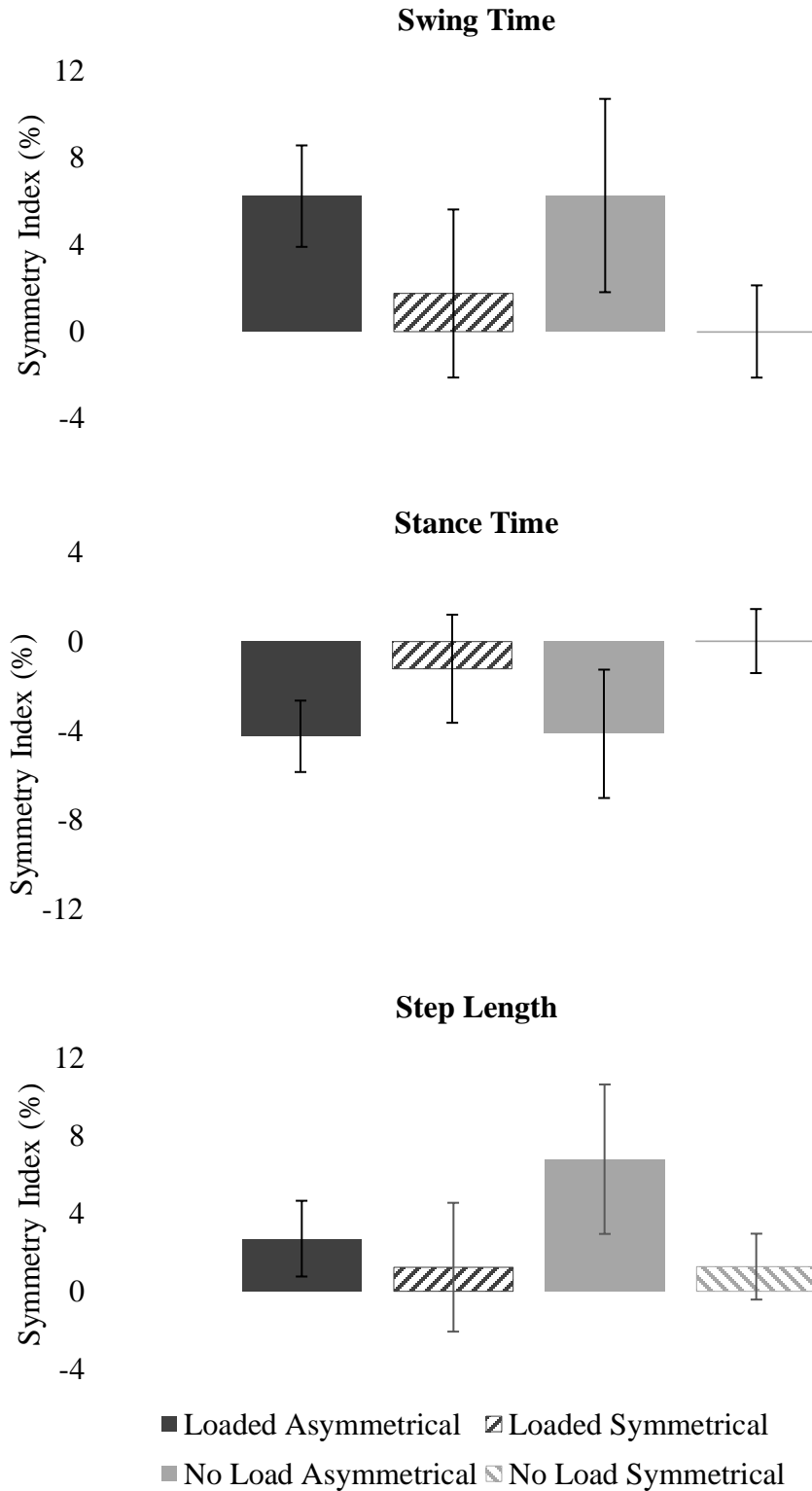
Figure 6.4 Conditional Spatiotemporal Comparisons. Comparisons were made for each condition with Asymmetrical (solid) and Symmetrical (striped) swing times. Conditions are grouped by Loaded (dark grey) and No Load (light grey) conditions.

Discussion

The addition of a unilateral 2 kg mass resulted in a 6% asymmetrical swing time during the LA condition. Previous research of persons with unilateral TTA revealed a similar 4% swing time interlimb asymmetry (Sanderson & Martin, 1997). Swing time asymmetry is especially important as the swing phase is likely when the increased metabolic demand occurs due to the added mass (Smith, Royer, et al., 2013). Further, the percent differences of metabolic costs between conditions suggest that our perturbations induced a similar metabolic penalty as walking with a unilateral TTA. The present spatiotemporal absolute values are also similar in magnitude and direction as previously reported in unilateral load manipulations to able-bodied individuals (Table 6.1) (Smith & Martin, 2007).

Participants were able to match the desired step times from the baseline conditions during the manipulation conditions with limited error. Specifically, participants on average were able to match the step time symmetry within 2% error during manipulation conditions to the paired baseline conditions. Participants achieved $1.8 \pm 0.9\%$ error and $2.0 \pm 1.8\%$ error for NLS vs LS and LA vs NLA conditions, respectively. With the added biofeedback to the participants and streaming swing time symmetry percentages, the research team was able to coach participants on how to adjust in real-time.

Table 6.1

Spatiotemporal Absolute Values.

	Loaded Asymmetrical		Loaded Symmetrical		No Load Asymmetrical		No Load Symmetrical	
	Unloaded Leg	Loaded Leg	Unloaded Leg	Loaded Leg	Unloaded Leg	Loaded Leg	Unloaded Leg	Loaded Leg
Stance Time (s)	0.65 ± 0.01	0.62 ± 0.01	0.63 ± 0.01	0.62 ± 0.01	0.64 ± 0.02	0.62 ± 0.01	0.63 ± 0.01	0.63 ± 0.01
Swing Time (s)	0.42 ± 0.01	0.45 ± 0.01	0.42 ± 0.01	0.43 ± 0.01	0.41 ± 0.01	0.44 ± 0.02	0.42 ± 0.01	0.42 ± 0.01
Step Length(m)	0.79 ± 0.01	0.81 ± 0.01	0.78 ± 0.02	0.79 ± 0.02	0.77 ± 0.03	0.82 ± 0.02	0.78 ± 0.01	0.79 ± 0.01

Note. Mean $\pm$ SD for the loaded and unloaded leg for each condition, with no significant differences noted between all measures.

Previous research utilizing an audible metronome to generate an asymmetrical gait pattern during running noted the difficult time participants had matching small percent asymmetries (Beck et al., 2018). Presently, it is believed that the combination of verbal cues from the research team, the audible metronome, and the visual biofeedback facilitated the participant to choose the stimulus best for them. Some participants did note a preference of one stimulus over the others and choose to focus on the preferred and depended less on the other two for adjustments during the data collection.

The hypothesis that both LA and NLA conditions would result in an increase in the net metabolic cost of walking compared with NLS was supported by the current results. This supports previous research that found similar metabolic penalties of these asymmetrical perturbations (Ellis et al., 2013; Mattes et al., 2000). Pairwise comparisons of each condition did not reveal significant differences between LA, NLA, and LS conditions ($p > 0.05$). However, the LS condition resulted in the greatest net metabolic cost of walking and trended towards a significant difference from the LA condition ($p = 0.072$) with a Cohen's d effect size of $es = 0.46$ and an 18.8% difference between conditions. Comparisons between the NLS and LS conditions and the NLS and NLA conditions presented with medium sized Cohen's d effect size of $es = 0.67$ and $es = 0.64$ and a 31.5% and 23.6% difference between conditions, respectively. These approximately 20-30% differences between conditions mirrored the percent difference of metabolic demands previously found between able-bodied individuals and persons with unilateral TTA (Mengelkoch et al., 2014).

These findings also suggest that the limb mass difference may induce a greater metabolic penalty relative to the metabolic penalty associated with asymmetrical swing times as the percent difference in metabolic costs between NLS and LS conditions was greater than the percent difference between NLS and NLA conditions.

The second purpose of the study was to determine whether the factor of load or swing time symmetry had a greater effect on the metabolic cost of walking. There was an interaction between load and swing time symmetries where metabolic costs increased when forcing non-normal swing time symmetry. A greater metabolic cost of walking occurred during NLA walking compared with baseline NLS walking and a greater cost during LS walking compared with baseline LA walking. The present results show an intricate relationship between the two factors of load and temporal asymmetries. That is, when mechanically symmetrical (No Load), the least metabolically demanding gait pattern is symmetrical. When mechanically asymmetrical (Loaded), forcing a symmetrical walking pattern will result in a metabolic penalty.

The comparison of baseline and manipulation conditions clarified this interaction by revealing a significant main effect of forcing temporal symmetry or asymmetry where the two manipulation conditions resulted in a significantly greater metabolic cost than baseline conditions. In turn, one factor of load or swing time symmetry may not have a greater effect on the metabolic cost of walking. Rather, manipulating the natural gait symmetry results in a metabolic penalty when walking.

The present results suggest that attempting to maintain a symmetrical gait pattern, with altered limb mass, is metabolically detrimental. Specifically, when mechanically asymmetrical, attempts to overcome the resulting swing time asymmetries coincide with

a metabolic penalty. As symmetrical walking has presently been shown to induce a metabolic penalty with interlimb mass differences, clinicians that work to rehabilitate persons with unilateral TTA should consider placing less emphasis on returning to a symmetrical gait pattern. As walking is required for may activities of daily living and is an important aspect with regards to quality of life, any metabolic penalty that deters an individual from being physically active should be avoided if possible. Although attempts should be made to overcome some of the unique imbalances noted in the population via balance training and improved socket to residual limb interactions, research has suggested that some gait asymmetries may be unavoidable (Adamczyk & Kuo, 2015; Highsmith et al., 2016). In part, attempting to overcome these unavoidable mechanical asymmetries via symmetrical walking may exacerbate the metabolic penalty presently described. Although a metabolic penalty may be attributed to symmetrical gait patterns, clinicians and patients may still choose to pursue symmetry during gait retraining to avoid the perception of a pathological gait and prevent chronic complications of asymmetrical loading of the intact limb (Cutti et al., 2018; Handžic & Reed, 2015; Highsmith et al., 2016). A decision to force temporal symmetry on persons with interlimb mechanical asymmetries should first weigh the costs and benefits before committing to a symmetrical gait retraining rehabilitation protocol.

Comparisons of spatiotemporal parameters revealed similar swing and stance times between NLS vs LS and NLA vs LA conditions, suggesting participants replicated swing time symmetries in a consistent manner. However, only LA vs NLS reported a significant difference ($p = 0.02$) in step length symmetry. Although other conditional comparisons of step length symmetry had a similar mean difference between conditions,

variance also appeared to be greater. This increased variability during LS and NLA conditions suggests that participants were less consistent with adapting interlimb step length symmetry as they completed the manipulation conditions. Temporal symmetry may have been maintained by sacrificing a consistent spatial symmetry during manipulation conditions.

While fewer strides have previously been used when quantifying spatiotemporal symmetry, presently more strides were used to demonstrate that participants were consistently matching desired symmetry throughout the final 90 seconds of the trial when metabolic data were analyzed (Nasirzade et al., 2017; Nessler et al., 2015). Although conditions resulted in spatiotemporal parameters varying on average $\pm$ 2.25% (Table 6.1), symmetry indices generally were more variant during LS, LA, and NLA conditions than the NLS normal walking condition (Fig 6.4). As interlimb adjustments occur acutely via subsequent steps (Kozlowska et al., 2017), it may be that adjustments are more frequent during perturbed walking. In turn, symmetry indices of spatiotemporal parameters during perturbed walking required a greater number of strides. This variability in symmetry indices further supports analyzing the final 90 seconds of spatiotemporal data.

Although the cohort averaged ~2% error in matching manipulation condition swing time symmetry, individual error varied (Figure 6.1). For example, participant 08 had an increased error of 6% in matching LS swing time symmetry to the NLS baseline condition. Although the participant was more symmetrical during the LS than the LA condition, they were less accurate than any other participants. The inability to overcome the unilateral load and walk symmetrically during the LS manipulation condition was likely due to the relative perturbation to the participant's body mass. The 2 kg mass was

4.3% of the total body mass for participant 08 compared with the group average of 3%. Further, the LS manipulation condition was the last condition for participant 08. The added perturbation relative to body mass and being the last condition completed during the data collection could have contributed to the onset of fatigue and the inability to more accurately match the intended swing time symmetry. Of note, no other participant showed signs of fatigue during perturbation that would have resulted in a decreased ability to match the desired swing time symmetry.

Even with care taken to use a mass similar to the magnitude of mass loss from a TTA and feedback to improve participant accuracy to conditional demands, the study is limited in scope as participants were all able-bodied. Factors such as loss of proprioception and altered musculature that persons with a TTA are uniquely challenged with are not easily replicated in able-bodied participants and may affect the metabolic cost of walking. Future research should focus on utilizing similar methodologies used presently to replicate the effect of interlimb and temporal symmetries on persons with unilateral TTA.

Conclusion

On average participants were able to replicate the magnitude and direction of asymmetrical walking as persons with a unilateral transtibial amputation within ~2% when walking with a unilaterally added mass or to an asymmetrical metronome. Both unilateral mass and asymmetrical metronome resulted in an increase in the net metabolic cost of walking relative to unperturbed walking. Further, the factor of manipulating the natural temporal gait patterns that arise from the presence of a unilaterally added mass significantly increased the net metabolic cost of walking. In turn, the attempt to overcome

the resulting temporal asymmetry from an interlimb mass asymmetry resulted in a metabolic penalty. To avoid increasing the metabolic demand that persons with a unilateral transtibial amputation are faced with during walking, clinicians should avoid forcing a symmetrical gait pattern.

CHAPTER VII

DISCUSSIONS AND GENERAL CONCLUSION

Study One Findings

The quality of running mechanics is often characterized by interlimb symmetry, during which treadmill and overground running will result in either consecutive or inconsecutive strides. The present study aimed to determine the minimum number of strides required to establish a stable mean symmetry index (SMS) of discrete joint-level measures and to determine if differences occurred between consecutive and inconsecutive strides within trials. A sequential average was used to determine how many strides were required for a SMS. Multiple two factor ANOVAs were used to determine if differences between bins of strides and symmetry calculations were significantly different. A SMS was achieved on average in 16.4 ± 6.3 strides, however this included measures that were highly variant due to SMS calculated between near zero values. In turn, bins of 15 strides were used for comparisons. There were no significant differences ($p > 0.05$) found between continuous and discontinuous data or order of strides within the discontinuous data. Although there were significant differences between symmetry calculation values ($p < 0.05$), there was no significant difference between the numbers of strides required for stable symmetry for either symmetry index presently utilized.

Although previous research examining running symmetry rarely exceeded 15 strides, or explicitly noted the number of strides analyzed, similar asymmetries were found in the current study as previously noted (Beck et al., 2018; Gilgen-Ammann et al.,

2017). Despite the majority of previous research not analyzing more than 15 strides, a number of studies achieved similar results to the present findings (Beck et al., 2018; Gilgen-Ammann et al., 2017; Hamill et al., 1984; Herzog et al., 1989; Pappas et al., 2015; Zifchock et al., 2006). As these findings with fewer strides are in agreeance with the stable means presently calculated, it may be that a stable mean is not required to achieve an accurate measure. However, research that analyze a limited number of strides may be more susceptible to the overall findings being skewed by spurious asymmetries, as previously noted (Herzog et al., 1989). In addition to spurious asymmetries caused by calculations, the body compensates for asymmetrical movement patterns of a given step with acute adjustment of the subsequent step. A non-stable mean symmetry index could misrepresent the average asymmetry if one of these short-term compensations occur (Kozlowska et al., 2017). A SMS in turn provides confidence when both methodological considerations and mechanical adaptations can result in changes to a symmetry index.

It is important to note that not all discrete measures reached a SMS in the same number of strides, with some measures such as joint kinetics and joint angles requiring a greater number of strides. This discretion between variables may partially be due to how variant the discrete measures are during normal gait (Bredeweg et al., 2013). Further differences may stem from the calculations used to measure symmetry (Pappas et al., 2015). For example, variables with values near zero will result in smaller changes away from zero having a greater effect on symmetry indices compared to the same magnitude of change from a value not near zero; sagittal plane ankle angle at initial contact.

This limitation to symmetry indices has been previously noted, and is presently exemplified by the minimum knee angle (Carpes et al., 2010; Herzog et al., 1989). Minimum knee angle reached near zero values during the gait cycle and required nearly 40 strides to establish a SMS.

Based on previous research, it was expected that the two symmetry indices would result in different symmetry values (Carpes et al., 2010). Absolute Symmetry Index (ASI) and the traditional Symmetry Index (SI) were both found to require a similar number of strides to obtain a SMS. This implies that different data analysis techniques will not require unique alterations to data collection methods. Furthermore, with no differences in SMS values found between randomly selected and consecutive strides the same level of confidence can be placed in symmetry findings regardless of studies utilizing over ground or treadmill running.

As interlimb symmetry has been used as both a performance measure and a rehabilitative goal, understanding how to best quantify symmetry is important when establishing clinical and research protocols (Beck et al., 2018; Bredeweg et al., 2013; Gilgen-Ammann et al., 2017; Nasirzade et al., 2017). Although the current findings suggest collecting more strides than common, the findings do not conclude that dozens or hundreds of strides are required to confidently measure joint level interlimb mean symmetries. This is especially important in populations that may need to limit activity due to increased injury risk or those returning to sport (Bredeweg et al., 2013; Gilgen-Ammann et al., 2017; Tenforde et al., 2018; Zifchock et al., 2006). In turn, clinicians and researchers alike can be confident in the observed asymmetries when collecting 15 strides of running.

Additionally, these sport scientists do not need to be concerned with measuring interlimb symmetry only during consecutive strides or in a certain time point within a gait analysis.

There are a number of ways the present study is limited in scope. In particular the current study only addresses lower extremity measures, with upper extremity measures possessing the chance of requiring a different number of strides to reach a SMS. This study is also limited to comparing the averages of discrete values and did not explore the number of strides to achieve a stable variability symmetry index of the measures. Further, the current study is limited to exploring symmetry in a healthy and relatively symmetrical cohort.

Study Two Findings

Measuring interlimb symmetry can be a powerful tool for researchers and clinicians that work with populations possessing unilateral deficiencies. However, gait analyses can become difficult for participants to complete if easily fatigued or multiple conditions and trials are collected. In turn, it is important to understand how many strides can consistently represent asymmetries present during walking, and if these strides should be collected consecutively. As few as three to five strides have been utilized when studying gait kinematics and spatiotemporal parameters during walking, it is also important to understand if the stable mean will differ from three and five stride means. A sequential average was used to determine that eight strides on average is required to achieve a stable mean symmetry index. A repeated measure ANOVA on lower extremity joint kinematics and spatiotemporal parameters revealed no systematic difference between subsets of three, five, and eight strides, regardless of being calculated from

consecutive or inconsecutive strides. Further, no differences were noted between the first three, five, and eight strides of symmetry indices for all measures (p > 0.05).

The magnitudes of asymmetry in lower extremity joint kinematics and spatiotemporal measures presently found are consistent with previous findings (Diop et al., 2004; Hesse et al., 2003; Nasirzade et al., 2017; Nessler et al., 2015; Nolan, 2008). The relatively higher asymmetries in knee angle minimum is also consistent with previous findings where measures near zero have been found to result in a greater calculated asymmetry (Herzog et al., 1989). Although not all previous findings utilized eight strides, the present findings suggest that no significantly different findings would be expected if they had. This supports findings from Study One that a stable measure is not necessary to achieve an accurate measure, however a measure based off of fewer strides will still be at an increased risk of being effected by a stride with spurious asymmetry, as previously noted (Herzog et al., 1989).

The lack of systematic differences between measures with consecutive and inconsecutive subsets of strides has multiple implications. First, there should not be a difference in measured symmetry between data collections that analyze continuous strides via treadmill, and those analyzing discontinuous strides via overground walking. Although the present study did not utilize overground walking, data were analyzed in a way to replicate the random strides analysis. In turn, differences present in SI values between treadmill and overground walking should not stem from calculating the interlimb symmetry (Lee & Hidler, 2008). Secondly, it would be reasonable to expect intermittent breaks between trials to not affect the calculated SI. That being said prolonged breaks may, as previous research has suggested that gait symmetry can vary

day-to-day by upwards of four degrees of range of motion (Wolf et al., 2009). This suggests that SI can be calculated from intrasession strides but may be less reliable if calculated from intersession strides. Lastly, either acute adjustments to asymmetrical steps did not occur in the healthy cohort, or the SI was robust enough not to be altered by a given step. Previous research has noted that asymmetrical steps result subsequent step adjustments and that the body possesses the capacity to acutely compensate (Kozlowska et al., 2017). As Kozlowska and colleagues recruited a similar cohort and used similar methodologies, it is more likely that the acute adjustments were occurring. In turn, these adjustments were not drastic enough to result in a difference between the smaller subsets of strides; the subsets most vulnerable to spurious strides.

As interlimb symmetry is a common clinical goal for returning to sport in athletic populations that have suffered a unilateral musculoskeletal injury and a long-term goal of individuals with more systemic unilateral deficiencies; being able to confidently establish a symmetry index is vital to clinicians and researchers, alike. The current findings do suggest that a more stable measure is achieved with eight strides of walking, regardless of analyzing continuous strides or not. This SMS should provide the sport scientist with the most confident and consistent findings as the mean should not fall outside of a two standard deviation window. However, these findings also suggest that if needed, we should not expect to find a statistically different result if fewer strides are analyzed. This should allow for the flexibility to collect fewer strides when needed, depending on the participant's or researcher's needs.

Although the present findings are aimed at assisting researchers that work with clinical populations, the application may be limited due to the healthy, relatively

symmetrical, cohort recruited. In particular, populations such as unilateral amputees commonly present with more variant gait patterns, with some asymmetries may possibly be unavoidable (Adamczyk & Kuo, 2015; Arellano et al., 2015). The greater variability may increase the number of strides required to achieve a SMS, and in turn may result in significantly different SI than found in smaller subsets of strides.

Study Three Findings

Persons with a unilateral amputation are faced with a unique mechanical, and in turn temporal, asymmetry that negatively effects their ability to ambulate by inciting a greater metabolic demand. These asymmetries can be replicated in able-bodied persons by adding a unilateral mass at the ankle or an asymmetrical audible metronome, respectively. The present study aimed to determine if walking with a unilateral mass and the associated asymmetrical swing time would individually increase the metabolic cost of walking. Second, the purpose of this study was to determine if the factor of symmetry or added load had a greater effect on the metabolic cost of walking.

In general, the findings reveal a significant increase in the net metabolic cost of walking for all conditions compared with the baseline normal walking. This confirms the hypothesis that walking with a unilateral mass and the associated asymmetrical swing time will individually increase the metabolic cost of walking. Further, there was an interaction between load and swing time symmetries where metabolic costs increased when forcing non-normal swing time symmetry, e.g. a greater cost during NLA compared with NLS and a greater cost during LS compared with LA. This is supported by manipulation conditions resulting in a significantly greater cost than baseline conditions. In turn, one factor of load or swing time symmetry does not have a greater

effect on the metabolic cost of walking, but that it is a manipulation from the natural gait pattern symmetry that increases the metabolic costs of walking.

The addition of a unilateral 2 kg mass resulted in a 6% asymmetrical swing time during the LA and NLA conditions. Previous research of persons with unilateral TTA revealed a similar 4% swing time interlimb asymmetry (Sanderson & Martin, 1997). Swing time asymmetry is especially important as the swing phase is likely when the increased metabolic demand occurs (Smith, Royer, et al., 2013). The similarly asymmetrical swing times, along with the metabolic percent difference findings suggest that our perturbations induced a similar metabolic penalty as a unilateral transtibial amputation. The present spatiotemporal absolute values are also similar in magnitude and direction as previously reported in unilateral load manipulations to able-bodied individuals (Table 6.1) (Smith & Martin, 2007).

Further, participants were able to match the desired step times with limited error. Specifically, participants on average were able to match the step time symmetry within 2% error during manipulation conditions to the paired baseline conditions. Participants achieved $1.8 \pm 0.9\%$ error and $2.0 \pm 1.8\%$ error for NLS vs LS and LA vs NLA conditions, respectively. With the added biofeedback to the participants and streaming swing time symmetry percentages, the research team was able to coach participants on how to adjust in real-time. Previous research utilizing an audible metronome to generate an asymmetrical gait pattern during running noted the difficult time participants had matching small percent asymmetries (Beck et al., 2018). Presently, it is believed that the combination of verbal cues from the research team, the audible metronome, and the visual biofeedback facilitated the participant to choose the stimulus best for them. Some

participants did note a preference of one stimulus over the others and choose to focus on the preferred and depend less on the other two for adjustments during the condition.

The hypothesis that both LA and NLA conditions would result in an increase in the net metabolic cost of walking compared with NLS was supported by the current results. This supports previous research that found similar metabolic penalties of these asymmetrical perturbations (Ellis et al., 2013; Mattes et al., 2000). Pairwise comparisons of each condition did not reveal significant differences between LA, NLA, and LS conditions ($p > 0.05$). However, the LS condition resulted in the greatest net metabolic cost of walking and trended towards a significant difference from the LA condition ($p = 0.072$) with a Cohen's d effect size of $es = 0.46$ and an 18.8% difference between conditions. Comparisons between the NLS and LS conditions and the NLS and NLA conditions presented with medium sized Cohen's d effect size of $es = 0.67$ and $es = 0.64$ and a 31.5% and 23.6% difference between conditions, respectively. These ~20 - 30% differences between conditions mirrored the percent difference of metabolic demands previously found between able-bodied individuals and persons with unilateral TTA (Mengelkoch et al., 2014). These findings also suggest that the limb mass difference may induce a greater metabolic penalty relative to the metabolic penalty associated with asymmetrical swing times.

The second purpose of the study was to determine whether the factor of load or swing time symmetry had a greater effect on the metabolic cost of walking. There was an interaction between load and swing time symmetries where metabolic costs increased when forcing non-normal swing time symmetry. A greater metabolic cost of walking occurred during No Load Asymmetrical walking compared with baseline NLS walking

and a greater cost during Loaded Symmetrical walking compared with baseline LA walking. The present results show an intricate relationship between the two factors of load and temporal asymmetries. That is, when mechanically symmetrical (No Load), the least metabolically demanding gait pattern is symmetrical. When mechanically asymmetrical (Loaded), forcing a symmetrical walking pattern will result in a metabolic penalty.

The comparison of baseline and manipulation conditions clarified this interaction by revealing a significant main effect of temporal manipulation where manipulation conditions resulted in a significantly greater metabolic cost than baseline conditions. In turn, one factor of load or swing time symmetry does may not have a greater effect on the metabolic cost of walking. Rather, manipulating the natural gait symmetry results in a metabolic penalty when walking.

Presently, our results suggest that attempting to maintain a symmetrical gait pattern, with altered mass is metabolically detrimental. Specifically, when mechanically asymmetrical, attempts to overcome the resulting swing time asymmetries coincide with a metabolic penalty. As symmetrical walking has presently been shown to induce a metabolic penalty with interlimb mass differences, clinicians that work to rehabilitate persons with unilateral TTA should consider placing less emphasis on returning to a symmetrical gait pattern. As walking is required for may activities of daily living and is an important aspect with regards to quality of life, any metabolic penalty that deters an individual from being physically active should be avoided if possible. Although attempts should be made to overcome some of the unique imbalances noted in the population via balance training and improved socket to residual limb interactions, research has

suggested that some gait asymmetries may be unavoidable (Adamczyk & Kuo, 2015; Highsmith et al., 2016). In part, overcoming these unavoidable mechanical asymmetries via symmetrical walking may exacerbate the metabolic penalty presently described. Although a metabolic penalty may be attributed to symmetrical gait patterns, clinicians and patients may still choose to pursue symmetry during gait retraining to avoid the perception of a pathological gait and prevent chronic complications of asymmetrical loading of the intact limb (Cutti et al., 2018; Handžic & Reed, 2015; Highsmith et al., 2016). A decision to force gait symmetry with interlimb asymmetries should weigh the costs and benefits before committing to a symmetrical gait retraining rehabilitation protocol.

Comparisons of spatiotemporal parameters revealed similar swing and stance times between NLS vs LS and NLA vs LA conditions, suggesting participants replicated swing time symmetries in a consistent manner. However, only LA vs NLS reported a significant difference ($p = 0.02$) in step length symmetry. Although other conditional comparisons of step length symmetry had a similar mean difference between conditions, variance also appeared to be greater. This increased variability during LS and NLA conditions suggests that participants were less consistent with adapting interlimb step length symmetry as they completed the manipulation conditions. Temporal symmetry may have been maintained by sacrificing a consistent spatial symmetry during manipulation conditions.

While fewer strides have previously been used when quantifying spatiotemporal parameter symmetry, presently more strides were used to ensure that participants were consistently matching desired symmetry throughout the final 90 seconds of the trial when

metabolic data were analyzed (Nasirzade et al., 2017; Nessler et al., 2015). Albeit all conditions resulted in spatiotemporal parameters varying less than ± 0.03 meters and seconds (Table 6.1), symmetry indices generally were more variant during conditions that participants were perturbed in some way (Fig 6.4). As interlimb adjustments occur acutely via subsequent steps (Kozlowska et al., 2017), it may be that symmetry indices of spatiotemporal parameters during perturbed walking could require a greater number of strides. This variability in symmetry indices further supports analyzing the final 90 seconds of spatiotemporal data.

Although the cohort averaged ~2% error in matching manipulation condition swing time symmetry, individual error varied (Figure 6.1). For example, participant 08 had an increased error of 6% in matching LS swing time symmetry to the NLS baseline condition. Although the participant was more symmetrical during the LS than the LA condition, they were less accurate than any other participant. The inability to overcome the unilateral load and walk symmetrically during the LS manipulation condition was likely due to the relative perturbation to the participants body mass. The 2 kg mass was 4.3% of the total body mass for participant 08 compared with the group average of 3%. Further, the LS manipulation condition was the last condition for participant 08. The added perturbation relative to body mass and being the last condition could have contributed to the onset of fatigue and the inability to more accurately match the intended swing time symmetry. Of note, no other participant showed signs of fatigue during perturbation that would have resulted in a decreased ability to match the desired swing time symmetry.

Even with care taken to use a mass similar to the magnitude of mass loss from a TTA and feedback to improve participant accuracy to conditional demands, the study is limited in scope as participants were all able-bodied. Factors such as loss of proprioception and altered musculature that persons with a TTA are uniquely challenged with are not easily replicated in able-bodied participants and may affect the metabolic cost of walking. Future research should focus on utilizing similar methodologies used presently to replicate the effect of interlimb and temporal symmetries on persons with unilateral TTA.

Conclusion

An average of 15 strides are required to achieve a stable mean symmetry index of the lower extremity spatiotemporal, joint kinematic, and joint kinetic variables when running. The timing within a data collection, and the method of collecting either continuous or discontinuous strides does not affect the number of strides required to achieve a stable mean symmetry index when running. Although the symmetry index and absolute symmetry index calculated different percent differences between limbs, there was no difference in the number of strides to achieve a stable mean between the two measures. When examining interlimb symmetry, a greater emphasis should be placed on collecting a sufficient number of strides to achieve a stable mean than emphasizing strides be collected continuously or at a specific time point within the data collection.

An average of eight walking strides are required to achieve a stable mean symmetry index of the lower extremity joint kinematics and spatiotemporal measures. No trend was determined to indicate a difference between findings of continuous and random strides; suggesting no difference should be found between symmetry indices collect with

overground and treadmill walking for most measures. Although a stable symmetry index was achieved after eight strides for most measures, no difference was noted between the first three, five, and eight strides for all measures. This suggests that although a measure may be more likely to be affected by a single stride, there should be no difference between means with less strides analyzed in a healthy population when walking.

Lastly, on average participants were able to replicate the magnitude and direction of asymmetrical walking as persons with a unilateral transtibial amputation within ~2% when walking with a unilaterally added mass or to an asymmetrical metronome. Both unilateral mass and asymmetrical metronome resulted in an increase in the net metabolic cost of walking relative to unperturbed walking. Further, the factor of manipulating the natural temporal gait patterns that arise from the presence of a unilaterally added mass significantly increased the net metabolic cost of walking. In turn, the attempt to overcome the resulting temporal asymmetry from an interlimb mass asymmetry resulted in a metabolic penalty. To avoid increasing the metabolic demand that persons with a unilateral transtibial amputation are faced with during walking, clinicians should avoid forcing a symmetrical gait pattern.

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APPENDIX A

INSTITUTIONAL REVIEW BOARD APPROVAL

UNIVERSITY OF NORTHERN COLORADO

Institutional Review Board

DATE: August 10, 2018

TO: Shane Murphy
FROM: University of Northern Colorado (UNCO) IRB

PROJECT TITLE: [1301508-1] Biomechanical factors driving increased Metabolic Costs during Locomotion via Asymmetrical Inertial and Temporal Manipulations
SUBMISSION TYPE: New Project

ACTION: APPROVED
APPROVAL DATE: August 10, 2018
EXPIRATION DATE: August 9, 2019
REVIEW TYPE: Expedited Review

Thank you for your submission of New Project materials for this project. The University of Northern Colorado (UNCO) IRB has APPROVED your submission. All research must be conducted in accordance with this approved submission.

This submission has received Expedited Review based on applicable federal regulations.

Please remember that informed consent is a process beginning with a description of the project and insurance of participant understanding. Informed consent must continue throughout the project via a dialogue between the researcher and research participant. Federal regulations require that each participant receives a copy of the consent document.

Please note that any revision to previously approved materials must be approved by this committee prior to initiation. Please use the appropriate revision forms for this procedure.

All UNANTICIPATED PROBLEMS involving risks to subjects or others and SERIOUS and UNEXPECTED adverse events must be reported promptly to this office.

All NON-COMPLIANCE issues or COMPLAINTS regarding this project must be reported promptly to this office.

Based on the risks, this project requires continuing review by this committee on an annual basis. Please use the appropriate forms for this procedure. Your documentation for continuing review must be received with sufficient time for review and continued approval before the expiration date of August 9, 2019.

Please note that all research records must be retained for a minimum of three years after the completion of the project.

If you have any questions, please contact Nicole Morse at 970-351-1910 or nicole.morse@unco.edu. Please include your project title and reference number in all correspondence with this committee.

APPENDIX B

STUDY ONE INTRACLASS CORRELATION COEFFICIENTS

Table B.1

*ICC Values for Intrasession and Intersession Reliability.*

Variable description	ICC (95% CI)	
	Intrasession	Intersession
Ankle Angle at Initial Contact (sagittal)	0.995 (0.988-0.998)	**0.746 (0.344-0.900)**
Ankle Angle Range of Motion (sagittal)	0.988 (0.971-0.995)	0.954 (0.887-0.982)
Ankle Angle Range of Motion (frontal)	0.989 (0.972-0.996)	0.784 (0.448-0.915)
Ankle Moment Peak (sagittal)	0.999 (0.997-1.000)	0.861 (0.651-0.945)
Ankle Moment Peak (frontal)	0.998 (0.995-0.999)	0.756 (0.391-0.903)
Ankle Power Peak Generation (sagittal)	0.999 (0.997-0.999)	0.915 (0.784-0.966)
Ankle Power Peak Absorption (frontal)	0.997 (0.992-0.999)	0.874 (0.686-0.950)
Knee Angle at Initial Contact (sagittal)	0.996 (0.980-0.999)	**0.647 (0.093-0.861)**
Knee Angle Range of Motion (sagittal)	0.990 (0.891-0.997)	0.947 (0.864-0.979)
Knee Angle Range of Motion (frontal)	0.992 (0.979-0.997)	**0.378 (-0.562-0.753)**
Knee Moment Peak (sagittal)	0.990 (0.975-0.996)	0.874 (0.686-0.950)
Knee Moment Peak (frontal)	0.998 (0.994-0.999)	0.917 (0.778-0.968)
Knee Power Peak Generation (sagittal)	0.987 (0.958-0.995)	0.906 (0.761-0.963)
Knee Power Peak Absorption (sagittal)	0.991 (0.977-0.996)	0.863 (0.650-0.946)
Hip Angle at Initial Contact (sagittal)	1.000 (0.999-1.000)	0.969 (0.922-0.988)
Hip Angle Range of Motion (sagittal)	0.995 (0.981-0.998)	0.970 (0.924-0.988)
Hip Angle Range of Motion (frontal)	0.997 (0.989-0.999)	0.991 (0.979-0.997)
Hip Moment Peak (sagittal)	0.980 (0.950-0.992)	0.888 (0.714-0.956)
Hip Moment Peak (frontal)	0.996 (0.991-0.999)	0.871 (0.672-0.949)
Hip Power Peak Generation (sagittal)	0.985 (0.954-0.994)	0.950 (0.872-0.980)
Hip Power Peak Absorption (sagittal)	0.985 (0.962-0.994)	0.956 (0.890-0.983)
Average ICC values	0.993	0.858

Note. Bold values indicate an ICC value less than 0.75.

APPENDIX C

STUDY THREE INDIVIDUAL METABOLIC COST PLOTS

Below are a series of graphs for individual participant metabolic data across all conditions (Figure C.1). Of note, most participants follow a similar trend. Absolute spatiotemporal data for all 75 and first 8 strides are provided (Table C.1 and C.2).
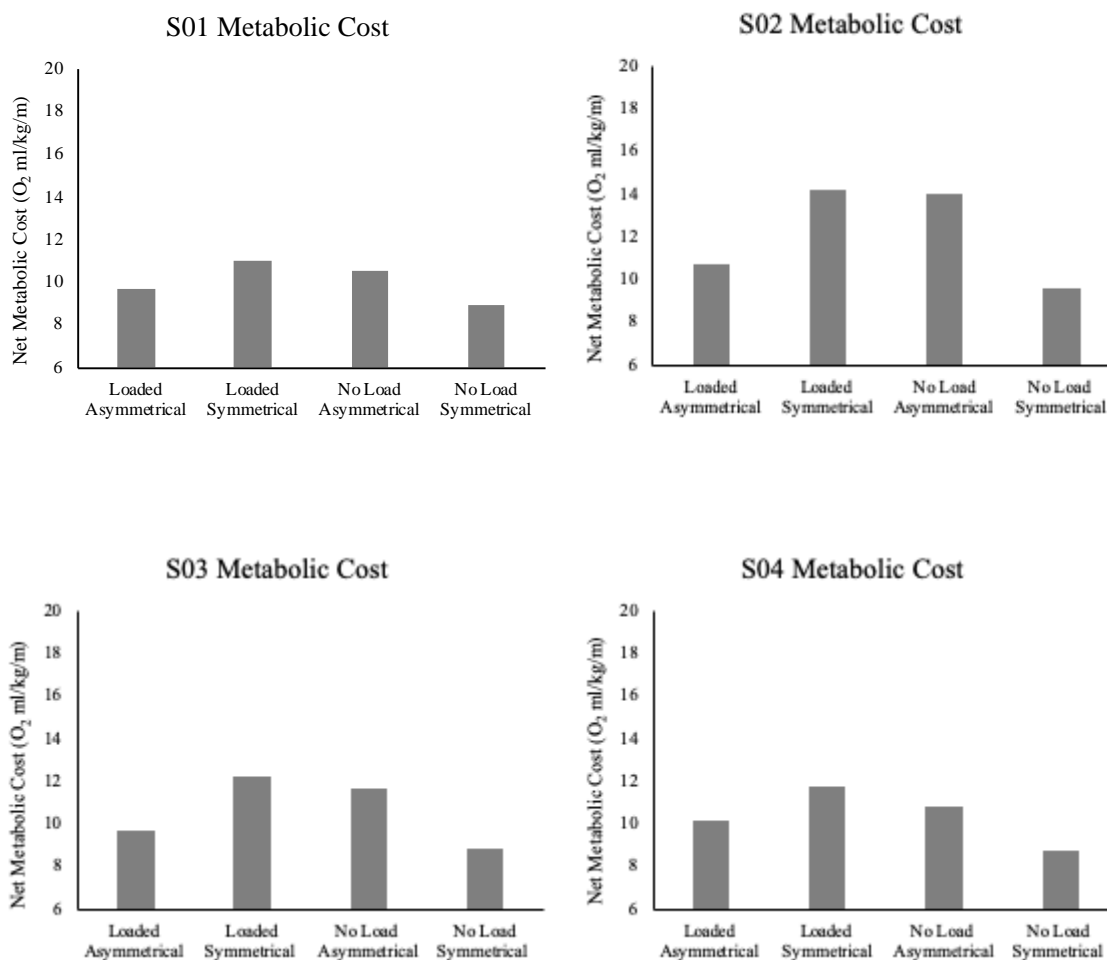
Figure C.1 Individual Metabolic Costs of Walking. A series of individual participant net metabolic costs of walking for all four conditions.
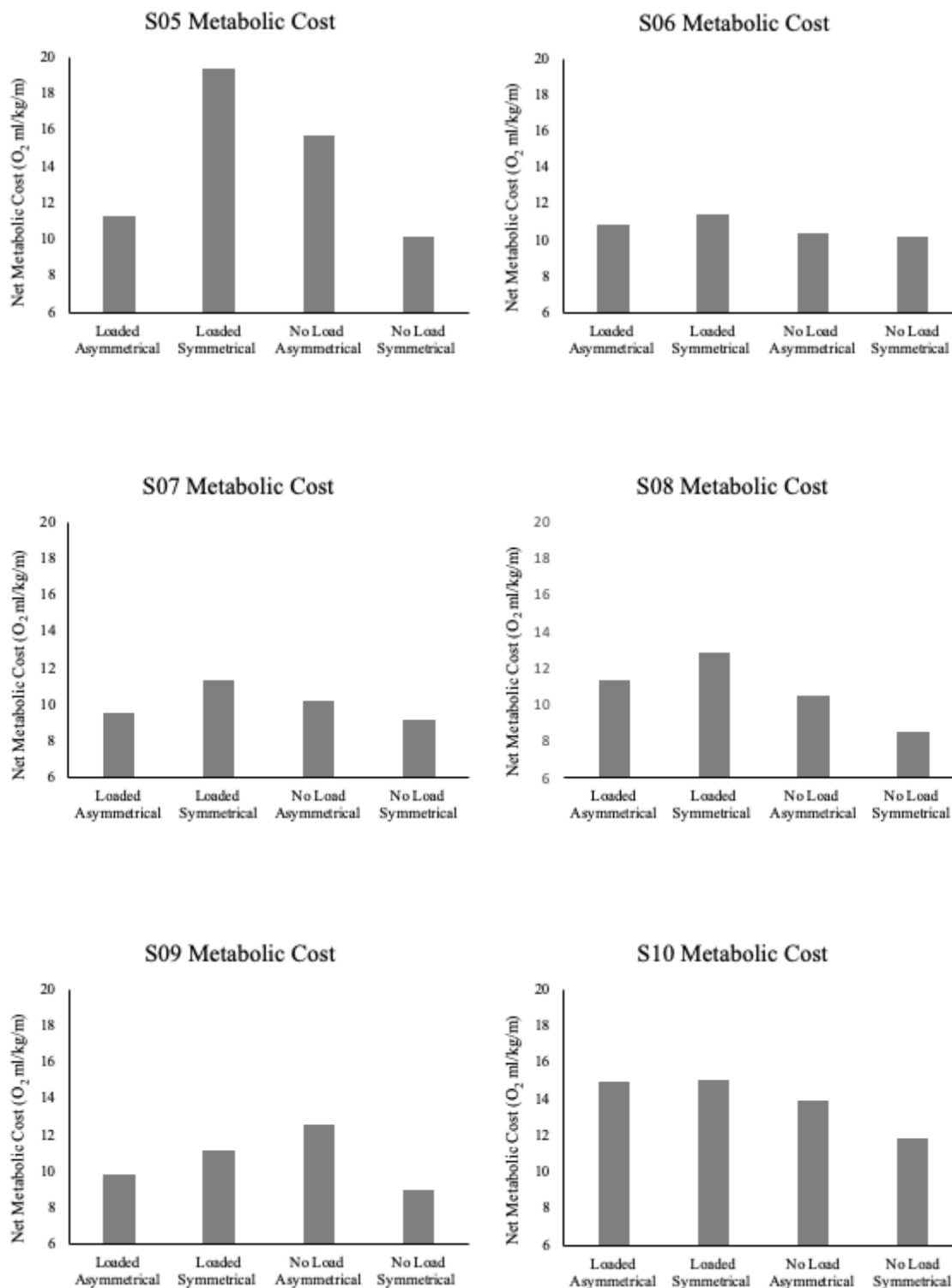
Figure C.1 (Continued) Individual Metabolic Costs of Walking. A series of individual participant net metabolic costs of walking for all four conditions.

Tabel C.1

Absolute Spatiotemporal Conditional Values for the First 8 Strides.

	Load Asymmetrical		Load Symmetrical		No Load Asymmetrical		No Load Symmetrical	
	Unloaded Leg	Loaded Leg	Unloaded Leg	Loaded Leg	Unloaded Leg	Loaded Leg	Unloaded Leg	Loaded Leg
Stance Time (s)	0.65 ± 0.01	0.62 ± 0.01	0.62 ± 0.01	0.61 ± 0.01	0.64 ± 0.01	0.62 ± 0.01	0.63 ± 0.01	0.63 ± 0.01
Swing Time (s)	0.42 ± 0.01	0.45 ± 0.01	0.42 ± 0.01	0.43 ± 0.01	0.41 ± 0.01	0.44 ± 0.01	0.42 ± 0.01	0.42 ± 0.01
Step Length (m)	0.79 ± 0.01	0.81 ± 0.01	0.78 ± 0.02	0.78 ± 0.02	0.76 ± 0.03	0.82 ± 0.02	0.78 ± 0.01	0.79 ± 0.01

Tabel C.2

Absolute Spatiotemporal Conditional Values for All Strides.

	Load Asymmetrical		Load Symmetrical		No Load Asymmetrical		No Load Symmetrical	
	Unloaded Leg	Loaded Leg	Unloaded Leg	Loaded Leg	Unloaded Leg	Loaded Leg	Unloaded Leg	Loaded Leg
Stance Time (s)	0.65 ± 0.01	0.62 ± 0.01	0.63 ± 0.01	0.62 ± 0.01	0.64 ± 0.02	0.62 ± 0.01	0.63 ± 0.01	0.63 ± 0.01
Swing Time (s)	0.42 ± 0.01	0.45 ± 0.01	0.42 ± 0.01	0.43 ± 0.01	0.41 ± 0.01	0.44 ± 0.02	0.42 ± 0.01	0.42 ± 0.01
Step Length (m)	0.79 ± 0.01	0.81 ± 0.01	0.78 ± 0.02	0.79 ± 0.02	0.77 ± 0.03	0.82 ± 0.02	0.78 ± 0.01	0.79 ± 0.01

Although the study two supports using 8 strides for spatiotemporal measures, study three manipulation conditions appear to increase the variability of the symmetry indices. This resulted in a reduced average error in matching baseline symmetry for all strides compared with the first 8 strides (Figure C2 and C.3).
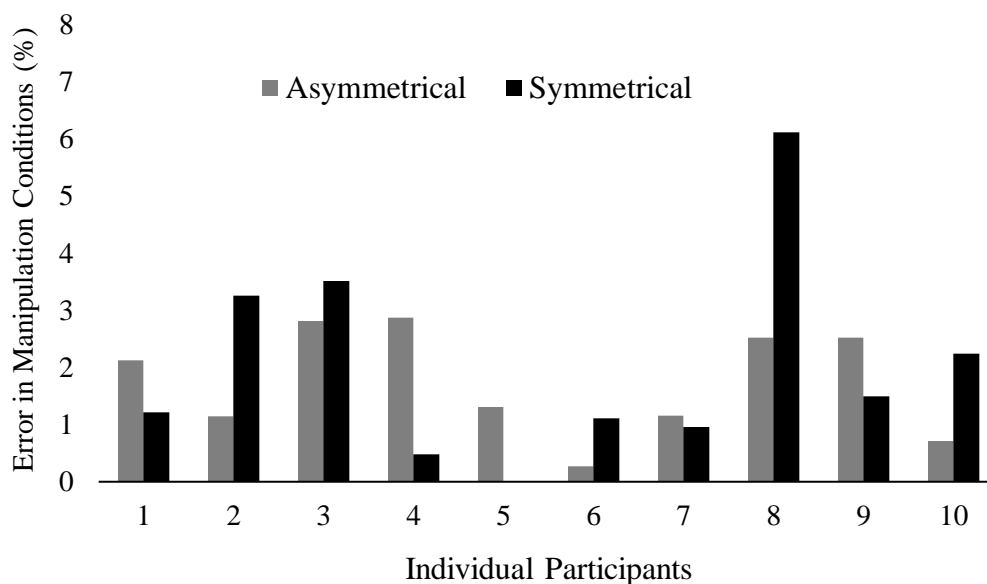

Figure C.2 Error in Manipulation Conditions for All Strides. Individual participant error in achieving the same swing time symmetry during the manipulation conditions.
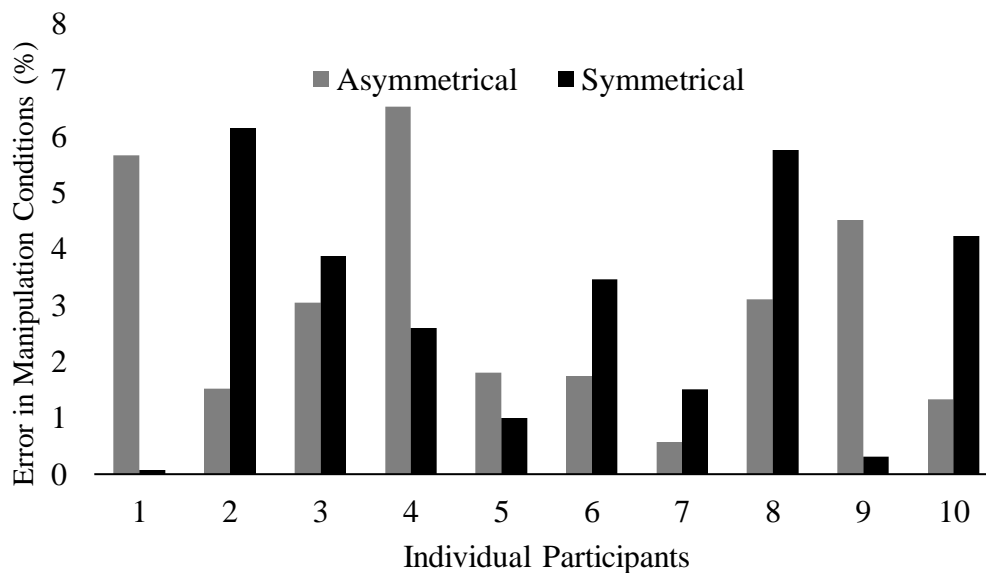
Figure C.3 Error in Manipulation Conditions for First 8 Strides. Individual participant error in achieving the same swing time symmetry during the manipulation conditions.

APPENDIX D

STUDY THREE MATLAB SCRIPTS

The following are copies of Matlab scripts used to complete the collection of study three. The Treadmill_biofeedback_main_v3.m is a script used in conjunction with Vicon Nexus and an AMTI tandem belt instrumented treadmill to retrieve, calculate, and display symmetry indices in real-time. Calculations are completed via the second script LiveLocomotionForceCalculations.m and are presented in bar graph form in the end of the main script. Previous versions did not remove spurious symmetry calculations from display or use a try statement to prevent the script from crashing.

The third and fourth scripts are contingent upon two files not included below; Metronome_SI.vi and Tick.wav. Although these two files are not in Matlab script form, the Metronome_Main.m and CurrentasymMetronome.m are dependent upon them to work properly. The Metronome_Main.m takes user inputs to apply to the CurrentasymMetronome.m script and generates a sound file with the desired symmetry. The Metronome_Main script allows for a testing period prior to committing to the audio file.

Treadmill_biofeedback_main_v3.m

```matlab
%% Treadmill_biofeedback_main_v3
% SPM 5/10/2019
% Run with AMTI tandem belt to determine live spatiotemporal symmetry
% Dependent: LiveLocomotionForceCalculations.m

%% Participant Information

% Reset
clear
clc
close all

% Provide Saving Filename
SubNum = input('Subject Numner (for example, 01):  ','s');
SubCond = input('Input current condition code. (NWS,AWA,AWS,NWA): ','s');
Filename = strcat('D:\Vicon
Databases\Research\Asym_Locomotion\Data_Collections\S',SubNum,'\S',SubNum,'_',SubCond,'_TrialSymmetry.xlsx');
% Participant Weight
subN = input('Provide Subject Weight (N): ');

% Determines if graph provided
manip_cond = input('Is feedback required? [y/n]:  ','s');
if manip_cond == 'y'
    swing_time_input = input('What is the average swing time during unloaded conditions?:  ');
    asym_pct = (input('What is the goal asymmetry percentage? (i.e., -5 = 5% to the Left):  ')/100);
    LimbLoad = input('Which limb has been loaded? (L=0 R=1):  ');
else
    asym_pct = 0;
end
%% Treadmill communication

% Add path with Vicon / MATLAB integration functions
addpath('C:\Program Files (x86)\Vicon\DataStream SDK\Win64\MATLAB')

% Program options
TransmitMulticast = false;
EnableHapticFeedbackTest = false;

% A dialog to stop the loop
MessageBox = msgbox( 'Stop DataStream Client', 'Vicon DataStream SDK' );
```

```matlab
% Load the SDK
fprintf( 'Loading SDK...' );
Client.LoadViconDataStreamSDK();
fprintf( 'done\n' );

% Program options
HostName = 'localhost:801';

% Make a new Vicon client in the MATLAB workspace
MyClient = Client();

% Connect to a server
fprintf( 'Connecting to %s ...', HostName );
while ~MyClient.IsConnected().Connected
 % Direct connection
 MyClient.Connect( HostName );
 fprintf( '.' );
end
fprintf( '\n' );

% Enable device data type
MyClient.EnableDeviceData();

% Set the streaming mode
MyClient.SetStreamMode( StreamMode.ClientPullPreFetch );

% Set the global up axis (Not sure if this is "correct" but gives correct
% vertical GRF(1) and ML COP(2)
MyClient.SetAxisMapping( Direction.Right, ...
                Direction.Forward, ...
                Direction.Up );

Output_GetAxisMapping = MyClient.GetAxisMapping();

% Discover the version number
Output_GetVersion = MyClient.GetVersion();

%% Inputs for calculations
% Force plate settings
threshold = -50; % Force in Newtons

% General loop settings and counters
Counter = 1;
row_index = 0;
FP_back = zeros(5000,3);
```

```
FP_front = zeros(5000,3);

% Figure settings
f = figure;
f.Units = 'inches';
f.Position = [40.0000 0.4167 20.0000 15.0313];
f.WindowState = 'maximized';
%% Data Streaming

% Loop until the message box is dismissed
while ishandle( MessageBox )
 % Get a frame
 MyClient.GetFrame().Result.Value ~= Result.Success;

 % Progresses after both plates sampled
 row_index = row_index + 1;
 %Collects both plates 2=Front 1=Back
  for ForcePlateIndex = 1:2 %ForcePlateCount
      Output_GetGlobalForceVector = MyClient.GetGlobalForceVector(
ForcePlateIndex, 1 );
      Output_GetGlobalCentreOfPressure =
MyClient.GetGlobalCentreOfPressure( ForcePlateIndex, 1 );
      ref = clock;
      % Saving Loop of data streaming
      % Back Plate Loop
      if ForcePlateIndex == 1
        if Output_GetGlobalForceVector.ForceVector(3) < threshold
          % Fz
          FP_back(row_index, 1) =
Output_GetGlobalForceVector.ForceVector(3);
          % COPx
          FP_back(row_index, 2) =
Output_GetGlobalCentreOfPressure.CentreOfPressure(2);
          % Time stamp (s)
          FP_back(row_index, 3) = (ref(5)*60)+ref(6);
        elseif Output_GetGlobalForceVector.ForceVector(3) >= threshold
          FP_back(row_index, 1) = 0;
          FP_back(row_index, 2) = 0;
          FP_back(row_index, 3) = (ref(5)*60)+ref(6);
        end
      % Front Plate Loop
      elseif ForcePlateIndex == 2
        if Output_GetGlobalForceVector.ForceVector(3) < threshold
          % Fz
          FP_front(row_index, 1) =
Output_GetGlobalForceVector.ForceVector(3);
```

```
                    % COPx
                    FP_front(row_index, 2) =
Output_GetGlobalCentreOfPressure.CentreOfPressure(2);
                    % Time stamp (s)
                    FP_front(row_index, 3) = (ref(5)*60)+ref(6);
                elseif Output_GetGlobalForceVector.ForceVector(3) >= threshold
                    FP_front(row_index, 1) = 0;
                    FP_front(row_index, 2) = 0;
                    FP_front(row_index, 3) = (ref(5)*60)+ref(6);
                end
            end
        end
     %% Symmetry Calculation and reset database
        MaxFP_front = -1*(min(FP_front(:,1)));
        if FP_front(row_index,3)-FP_front(1,3) >= 5 && MaxFP_front > .5*subN
            % Run Calculations
            try

[AVG_StepSymmetry,AVG_StrideSymmetry,AVG_SwingSymmetry,AVG_StanceSym
metry,StridesPerMin,Avg_LeftStepTime,Avg_RightStepTime,Avg_LeftStrideTime,Avg
_RightStrideTime,Avg_LeftStanceTime,Avg_RightStanceTime,Avg_LeftSwingTime,Av
g_RightSwingTime] = LiveLocomotionForceCalculations(subN,FP_front,FP_back);
                %Catch for spurious symmetry and remove
                if AVG_SwingSymmetry >25 || AVG_SwingSymmetry < -25 ||
((Avg_LeftSwingTime+Avg_RightSwingTime)/2) < 0.2 ||
((Avg_LeftSwingTime+Avg_RightSwingTime)/2) > 0.8
                    Trial_Symmetry(Counter,1:13) = NaN;
                    disp('Error Occured')
                else
                    Trial_Symmetry(Counter,1) = StridesPerMin;
                    Trial_Symmetry(Counter,2) = AVG_StepSymmetry;
                    Trial_Symmetry(Counter,3) = AVG_StrideSymmetry;
                    Trial_Symmetry(Counter,4) = AVG_SwingSymmetry;
                    Trial_Symmetry(Counter,5) = AVG_StanceSymmetry;
                    Trial_Symmetry(Counter,6) = Avg_LeftStepTime;
                    Trial_Symmetry(Counter,7) = Avg_RightStepTime;
                    Trial_Symmetry(Counter,8) = Avg_LeftStrideTime;
                    Trial_Symmetry(Counter,9) = Avg_RightStrideTime;
                    Trial_Symmetry(Counter,10) = Avg_LeftSwingTime;
                    Trial_Symmetry(Counter,11) = Avg_RightSwingTime;
                    Trial_Symmetry(Counter,12) = Avg_LeftStanceTime;
                    Trial_Symmetry(Counter,13) = Avg_RightStanceTime;

                    fprintf('Percent Swing Symmetry:  %2.1f%%\n',
AVG_SwingSymmetry);
```

```
                fprintf('Average Swing time:  %1.3f\n',
((Avg_LeftSwingTime+Avg_RightSwingTime)/2));
                fprintf('Strides Per Minute:  %2.1f\n', StridesPerMin);
                % Add Goal
                if manip_cond == 'y'
                    LongSwingGoal =
(swing_time_input)+(swing_time_input*(asym_pct/2));
                    ShortSwingGoal = (swing_time_input)-
(swing_time_input*(asym_pct/2));
                    if LimbLoad == 0
                        LongSwing = Avg_LeftSwingTime;
                        ShortSwing = Avg_RightSwingTime;
                    elseif LimbLoad == 1
                        ShortSwing = Avg_LeftSwingTime;
                        LongSwing = Avg_RightSwingTime;
                    else
                        Disp('Error in goal setting.');
                    end
                    UpperLim=LongSwingGoal*1.2;
                    LowerLim=ShortSwingGoal*.8;
                    if LimbLoad == 0
                        figure(f)
                        ylim([LowerLim UpperLim])
                        bar(0,LongSwingGoal,'g','stacked')
                        hold on
                        errorbar(0,LongSwingGoal,(LongSwingGoal*.02),'-
k','CapSize',250,'LineWidth',2.5)
                        bar(1,ShortSwingGoal,'g','stacked')
                        errorbar(1,ShortSwingGoal,(ShortSwingGoal*.02),'-
k','CapSize',250,'LineWidth',2.5)
                        bar(0,LongSwing,'r','BarWidth',.7,'FaceAlpha',0.75)
                        bar(1,ShortSwing,'r','BarWidth',.7,'FaceAlpha',0.75)
                        hold off
                        drawnow
                    elseif LimbLoad == 1
                        figure(f)
                        ylim([LowerLim UpperLim])
                        bar(1,LongSwingGoal,'g','stacked')
                        hold on
                        errorbar(1,LongSwingGoal,(LongSwingGoal*.02),'-
k','CapSize',250,'LineWidth',2.5)
                        bar(0,ShortSwingGoal,'g','stacked')
                        errorbar(0,ShortSwingGoal,(ShortSwingGoal*.02),'-
k','CapSize',250,'LineWidth',2.5)
                        bar(1,LongSwing,'r','BarWidth',.7,'FaceAlpha',0.75)
                        bar(0,ShortSwing,'r','BarWidth',.7,'FaceAlpha',0.75)
```

```matlab
                    hold off
                    drawnow
                end
            else
            end
        end
    catch err
        disp('Error Occured')
        Trial_Symmetry(Counter,1:13) = NaN;
    end
    % Reset database
    row_index = 0;
    clear FP_back FP_front
    FP_back = zeros(5000,3);
    FP_front = zeros(5000,3);
    Counter = Counter+1;

    elseif FP_front(row_index,3)-FP_front(1,3) >= 5 && MaxFP_front < .5*subN
        row_index = 0;
        clear FP_back FP_front
        FP_back = zeros(5000,3);
        FP_front = zeros(5000,3);
        fprintf('Resetting database \n');
    end
end
%% End Display and save
Trial_AVG_StridesPerMin = nanmean(Trial_Symmetry(:,1));
Trial_AVG_StepSymmetry = nanmean(Trial_Symmetry(:,2));
Trial_AVG_SwingSymmetry = nanmean(Trial_Symmetry(:,4));
Trial_AVG_LeftSwingTime = nanmean(Trial_Symmetry(:,10));
Trial_AVG_RightSwingTime = nanmean(Trial_Symmetry(:,11));
Trial_AVG_SwingTime =
((Trial_AVG_LeftSwingTime+Trial_AVG_RightSwingTime)/2);
fprintf(' Average Strides Per Minute: %2.2f\n', Trial_AVG_StridesPerMin);
fprintf(' Average Percent Swing Symmetry: %2.2f%%\n',
Trial_AVG_SwingSymmetry);
fprintf(' Left Swing Time: %1.3f\n', Trial_AVG_LeftSwingTime);
fprintf(' Right Swing Time: %1.3f\n', Trial_AVG_RightSwingTime);
fprintf(' Average Swing Time: %1.3f\n', Trial_AVG_SwingTime);
xlswrite(Filename,Trial_Symmetry);
```

LiveLocomotionForceCalculations.m

```
function
[AVG_StepSymmetry,AVG_StrideSymmetry,AVG_SwingSymmetry,AVG_StanceSym
metry,StridesPerMin,Avg_LeftStepTime,Avg_RightStepTime,Avg_LeftStrideTime,Avg
_RightStrideTime,Avg_LeftStanceTime,Avg_RightStanceTime,Avg_LeftSwingTime,Av
g_RightSwingTime] = LiveLocomotionForceCalculations(subN,FP_front,FP_back)
    %Completes symmetry calculation based off of force data
    %   Requires inputs to be three columns: 1)Fz 2) COPx 3)time stamp in sec
    %   Runs in conjunction with 'Treadmill_biofeedback_main_v1.m'
    %   SPM 4/23/2019

    %% Establish data
    Fz1 = FP_front(:,1);
    CoPx1 = FP_front(:,2);
    TimeStamp1 = FP_front(:,3);

    Fz2 = FP_back(:,1);
    CoPx2 = FP_back(:,2);
    TimeStamp2 = FP_back(:,3);

    minpkht_cutoff = .50*subN;
    % Set up filter where fc/.5*Samplerate
    [b,a] = butter(2,(50/1000),'low');

    % Flip force to GRF
    Fz1 = Fz1*-1;
    Fz2 = Fz2*-1;
    Fz1 = filter(b,a,Fz1);
    Fz2 = filter(b,a,Fz2);
    CoPx1 = filter(b,a,CoPx1);
    CoPx2 = filter(b,a,CoPx2);

    % Remove noise from signal
    for i = 1:length(Fz1)
       if Fz1(i,1) < 10
          Fz1(i,1) = 0;
       else
          Fz1(i,1) = Fz1(i,1);
       end
    end

    for i = 1:length(Fz2)
       if Fz2(i,1) < 10
          Fz2(i,1) = 0;
```

```
      else
          Fz2(i,1) = Fz2(i,1);
      end
end

%% Standardize Data
[~,firstzeroFP1] = min(Fz1);
for i = 1:firstzeroFP1-1
    Fz1(i,1) = NaN;
    Fz2(i,1) = NaN;
    CoPx1(i,1) = NaN;
    CoPx2(i,1) = NaN;
end

Fz1(any(isnan(Fz1),2),:) = [];
Fz2(any(isnan(Fz2),2),:) = [];
CoPx1(any(isnan(CoPx1),2),:) = [];
CoPx2(any(isnan(CoPx2),2),:) = [];

[~,firstzeroFP2] = min(Fz2);
for i = 1:firstzeroFP2-1
    Fz2(i,1) = 0;
end

%% Removes small gaps between impact peak and full curve
for j = 2:length(Fz1)-40
    if Fz1(j,1) < 50 && Fz1(j+40,1) >= 50 && Fz1(j-1,1) >=50
        Fz1(j,1) = 50;
    elseif Fz1(j,1) >= 50
        Fz1(j,1) = Fz1(j,1);
    else
        Fz1(j,1) = 0;
    end
end

for j = 2:length(Fz2)-40
    if Fz2(j,1) < 50 && Fz2(j+40,1) >= 50 && Fz2(j-1,1) >=50
        Fz2(j,1) = 50;
    elseif Fz2(j,1) >= 50
        Fz2(j,1) = Fz2(j,1);
    else
        Fz2(j,1) = 0;
    end
end

%% Determine L(0) vs R(1) foot
```

```
[~,locs1] = findpeaks(Fz1,'MinPeakProminence',minpkht_cutoff);
for d = 1:length(locs1)
   CoPx1_peakGRF(d,1) = CoPx1(locs1(d,1),1);
end

CoP1_mean = mean(CoPx1_peakGRF);

for e = 1:length(CoPx1_peakGRF)
   if CoPx1_peakGRF(e,1) < CoP1_mean
     CoPx1_peakGRF(e,2) = 0;
   elseif CoPx1_peakGRF(e,1) > CoP1_mean
      CoPx1_peakGRF(e,2) = 1;
   end
end
[~,locs2] = findpeaks(Fz2,'MinPeakProminence',minpkht_cutoff);
for d = 1:length(locs2)
   CoPx2_peakGRF(d,1) = CoPx2(locs2(d,1),1);
end

CoP2_mean = mean(CoPx2_peakGRF);

for e = 1:length(CoPx2_peakGRF)
   if CoPx2_peakGRF(e,1) < CoP2_mean
     CoPx2_peakGRF(e,2) = 0;
   elseif CoPx2_peakGRF(e,1) > CoP2_mean
      CoPx2_peakGRF(e,2) = 1;
   end
end
%% Define Gait Events

diff_Fz2 = diff(Fz2 == 0);
OFFs = find(diff_Fz2 == 1);
diff_Fz1 = diff(Fz1 == 0);
ONs = find(diff_Fz1 == -1)+1;

% Assign L or R on to gait events
if OFFs(1,1) < locs2(1,1)
   OFFs(1,1) = NaN;
else
   OFFs(1,1) = OFFs(1,1);
end

OFFs(any(isnan(OFFs),2),:) = [];
% removing OFFs that are too close caused by noise
for i=2:length(OFFs)
```

```
    if OFFs(i)-OFFs(i-1) < 200
       OFFs(i) = NaN;
    else
       OFFs(i) = OFFs(i);
    end
end
OFFs(any(isnan(OFFs),2),:) = [];

for i = 1:length(locs1)-1
   if locs1(i,1) > ONs(i,1)
      ONs(i,2) = CoPx1_peakGRF(i,2);
   elseif locs1(i,1) < ONs(i,1) && locs1(i+1,1) > ONs(i,1)
      ONs(i,2) = CoPx1_peakGRF(i+1,2);
   else
      ONs(i,2) = 3;
   end
end
if length(OFFs) <= length(locs2)
   for i = 1:length(OFFs)-1
      if locs2(i,1) < OFFs(i,1)
         OFFs(i,2) = CoPx2_peakGRF(i,2);
      else
         OFFs(i,2) = CoPx2_peakGRF(i+1,2);
      end
   end
else
   for i = 1:length(locs2)
      if locs2(i,1) < OFFs(i,1)
         OFFs(i,2) = CoPx2_peakGRF(i,2);
      else
         OFFs(i,2) = CoPx2_peakGRF(i+1,2);
      end
   end
end
if length(OFFs) < length(ONs)
   ONs(length(ONs),1) = NaN;
else
   ONs(length(ONs),1) = ONs(length(ONs),1);
end
ONs(any(isnan(ONs),2),:) = [];

% pair gait events

if ONs(1,1) < OFFs(1,1) && ONs(1,2) == OFFs(1,2)
   for i = 1:length(ONs)-1
      %Frame of ON
```

```
                GaitEvents(i,1) = ONs(i,1);
                %Frame of OFF
                GaitEvents(i,2) = OFFs(i,1);
                %Foot where L(0) and R(1)
                GaitEvents(i,3) = ONs(i,2);
            end
        else
            for i = 1:length(ONs)-1
                GaitEvents(i,1) = ONs(i,1);
                GaitEvents(i,2) = OFFs(i+1,1);
                GaitEvents(i,3) = ONs(i,2);
            end
        end

        %remove extra event on end
        for i = 1:length(GaitEvents)-1
            if GaitEvents(i,3) == GaitEvents(i+1,3)
                GaitEvents(i+1,1) = NaN;
                GaitEvents(i+1,2) = NaN;
                GaitEvents(i+1,3) = NaN;
            else
                GaitEvents(i+1,3) = GaitEvents(i+1,3);
            end
        end

        GaitEvents(any(isnan(GaitEvents),2),:) = [];

        %% Spatiotemporal Calculations
        % Step time (On contralateral On)
        for i = 1:length(GaitEvents)-1
            if GaitEvents(i,3) ~= GaitEvents(i+1,3)
                GaitEvents(i,4) = (TimeStamp1(GaitEvents(i+1,1),1)-
TimeStamp1(GaitEvents(i,1),1));
            else
                GaitEvents(i,4) = 0;
            end
        end

        for i = 1:length(GaitEvents)
            if GaitEvents(i,4) == 0
                GaitEvents(i,4) = NaN;
            else
                GaitEvents(i,4) = GaitEvents(i,4);
            end
        end
```

```matlab
        % Stride time (On to ipsilateral On)
        GaitEvents(1:length(GaitEvents),5) = 0;
        for i =1:length(GaitEvents)-2
           if GaitEvents(i,3) == GaitEvents(i+2,3)
              GaitEvents(i,5) = (TimeStamp1(GaitEvents(i+2,1),1)-
TimeStamp1(GaitEvents(i,1),1));
           else
              GaitEvents(i,5) = 0;
           end
        end

        for i = 1:length(GaitEvents)
           if GaitEvents(i,5) == 0
              GaitEvents(i,5) = NaN;
           else
              GaitEvents(i,5) = GaitEvents(i,5);
           end
        end

        StridesPerMin = 60/nanmean(GaitEvents(:,5));

        %Swing time (Off to On)
        GaitEvents(1:length(GaitEvents),6) = 0;
        for i = 1:length(GaitEvents)-2
           if GaitEvents(i,3) == GaitEvents(i+2,3)
              GaitEvents(i,6) = (TimeStamp1(GaitEvents(i+2,1),1)-
TimeStamp2(GaitEvents(i,2),1));
           else
              GaitEvents(i,6) = 0;
           end
        end

        for i = 1:length(GaitEvents)
           if GaitEvents(i,6) == 0
              GaitEvents(i,6) = NaN;
           else
              GaitEvents(i,6) = GaitEvents(i,6);
           end
        end

        % Stance time (On to ipsilateral Off)
        GaitEvents(1:length(GaitEvents),7) = 0;
        for i = 1:length(GaitEvents)
           if isnan(GaitEvents(i,6))
              GaitEvents(i,7) = NaN;
           else
```

```
        GaitEvents(i,7) = (TimeStamp2(GaitEvents(i,2),1)-
TimeStamp1(GaitEvents(i,1),1));
        end
    end

    % Divide Events
    for i = 1:length(GaitEvents)
        if GaitEvents(i,3) == 0 && ~isnan(GaitEvents(i,5))
            Left_GE(i,1) = GaitEvents(i,4);
            Left_GE(i,2) = GaitEvents(i,5);
            Left_GE(i,3) = GaitEvents(i,6);
            Left_GE(i,4) = GaitEvents(i,7);
        elseif GaitEvents(i,3) == 1 && ~isnan(GaitEvents(i,5))
            Right_GE(i,1) = GaitEvents(i,4);
            Right_GE(i,2) = GaitEvents(i,5);
            Right_GE(i,3) = GaitEvents(i,6);
            Right_GE(i,4) = GaitEvents(i,7);
        else
            Left_GE(i,1:4) = NaN;
            Right_GE(i,1:4) = NaN;
        end
    end

    %Remove zeros
    for i=1:length(Left_GE)
        if Left_GE(i,1) == 0
            Left_GE(i,1:4) = NaN;
        else
            Left_GE(i,1:4) = Left_GE(i,1:4);
        end
    end
    for i=1:length(Right_GE)
        if Right_GE(i,1) == 0
            Right_GE(i,1:4) = NaN;
        else
            Right_GE(i,1:4) = Right_GE(i,1:4);
        end
    end
    % Averages and Symmetry
    Avg_LeftStepTime = nanmean(Left_GE(:,1));
    Avg_RightStepTime = nanmean(Right_GE(:,1));
    Avg_LeftStrideTime = nanmean(Left_GE(:,2));
    Avg_RightStrideTime = nanmean(Right_GE(:,2));
    Avg_LeftSwingTime = nanmean(Left_GE(:,3));
    Avg_RightSwingTime = nanmean(Right_GE(:,3));
    Avg_LeftStanceTime = nanmean(Left_GE(:,4));
```

```
        Avg_RightStanceTime = nanmean(Right_GE(:,4));
        AVG_StepSymmetry = 100*((Avg_RightStepTime -
Avg_LeftStepTime)/((Avg_RightStepTime + Avg_LeftStepTime)/2));
        AVG_StrideSymmetry = 100*((Avg_RightStrideTime -
Avg_LeftStrideTime)/((Avg_RightStrideTime + Avg_LeftStrideTime)/2));
        AVG_StanceSymmetry = 100*((Avg_RightStanceTime -
Avg_LeftStanceTime)/((Avg_RightStanceTime + Avg_LeftStanceTime)/2));
        AVG_SwingSymmetry = 100*((Avg_RightSwingTime -
Avg_LeftSwingTime)/((Avg_RightSwingTime + Avg_LeftSwingTime)/2));
```

Metronome_Main.m

```
%% Metronome_Main
% SPM 4/19/19
% Provides inputs for metronome function
% Dependents: CurrentasymMetronome.m, Tick.wav, Metronome_SI.vi
% Inputs Required: stride_time, offset_proportion, trial_length

%% User Inputs
% Offset Percentage
Goal_offset = input('Percent offset (for example, 5% = .05):  ');
offset_proportion = 0.5-(Goal_offset*.5);
% Stride Time
StridePerMin = input('Strides per Minutes (#):  ');
stride_time = 60/StridePerMin;
% Verify settings
disp(Goal_offset);

disp(StridePerMin);
Verify = input('Verify inputs [y/n]:  ','s');
if Verify == 'y'
   disp('Metronome will play for 30 seconds. Wait until metronome has stopped
to respond to next question.')
      % Test settings for 30 sec
      trial_length = 30;
      CurrentasymMetronome(stride_time,offset_proportion,trial_length)
      % Run Metronome
      Test = input('Does metronome sound correct? [y/n]: ','s');
      if Test == 'y'
         disp('Metronome will play for 8 minutes.');
         % Actual Trial Length set to 8 minutes by default
         trial_length = 8*60;
         CurrentasymMetronome(stride_time,offset_proportion,trial_length)
      else
         disp('Please start over.');
      end
else
   disp('Please start over.');
end
```

CurrentasymMetronome.m

```
%% Asymmetrical Metronome
% Chad Healy
%
% Inputs:
% stride_time (seconds)
% offset_proportion (value from 0 - 1)
% trial length (seconds)
%
%
% While this constructs a perfectly timed metronome,
% you also cannot stop it once it starts, because it
% is stuck in the "sound" function of MATLAB, and
% thus cannot be broken with a simple ctrl+c

% Check Inputs
if nargin == 0
    disp('Using Default Settings.')
    % Set Frequency/Time Settings
    stride_time = 60/65;%time needed to take a stride (use 60 sec / strides per min)
    offset_proportion = 0.475;% proportion of stride offset (0.5 = symmetry) 0.4
would be one step at 40% of stride time and one step at 60% of stride time
    trial_length =1*60;% How long the metronome plays for
elseif offset_proportion < 0 || offset_proportion > 1
    error('Offset must be a value from 0 to 1.')
elseif stride_time <= 0
    error('Stride time must be a positive number.')
elseif trial_length <=0
    error('Trial length must be a positive number.')
elseif trial_length < stride_time
    error('Trial length must be greater than stride time.')
end

% Find offset_time - This print's out our definition of the symmetry index
% (R-L)/(0.5*(R+L))
UL=(1/(stride_time));
AL=(UL*2*(offset_proportion));
check=AL/UL;
offset_proportion2 = ((UL-AL)/ (0.5*(UL+AL)));
fprintf('the Asym is %2.5f \n' ,offset_proportion2)

offset_time = offset_proportion*stride_time;
```

```
% 0% Asym = 0.500
% 2.5% Asym = 0.4875
% 5% Asym = 0.475
% 7% Asym = 0.466
% 10% Asym = 0.45
% 14% Asym = 0.4345
% 21% Asym = 0.4049
% 28% Asym = 0.392


% Load in Sound
[Ybeep,FSbeep] = audioread('Tick.wav');
% There's a ton of lag in the beginning and end of this wav file
soundstart = find(Ybeep(:,1)~=0,1,'first');
soundend = find(Ybeep(:,1)~=0,1,'last')+1;
Y = Ybeep(soundstart:soundend,:);

% % Check if offset is shorter than the sound
% % Note: A limitation similar to this will eventually be needed
% %      but for now, this is too harsh, so it is commented out
% if offset_time*FSbeep < size(Y,1)
%     error(['Offset time is too short compared to sound length.',...
%         ' Choose a shorter sound or longer offset time.'])
% end

%Construct Giant WAV File...
Ylong = zeros(floor(stride_time*FSbeep),size(Y,2));
Ylong(1:size(Y,1),:) = Y;
Ylong(floor(offset_time*FSbeep)+1:floor(offset_time*FSbeep)+size(Y,1),:) = Y;
Ytrial = zeros(floor(trial_length/stride_time),size(Y,2));
for ii = 1:floor(trial_length/stride_time)
   Ytrial(1+(ii-1)*size(Ylong,1):ii*size(Ylong,1),:) = Ylong;
end

%Play Sound
sound(Ytrial,FSbeep)
```