

THE EFFECT OF ARM SWING AND ASYMMETRIC WALKING ON GAIT KINETICS IN YOUNG ADULTS

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Abstract

Introduction: Asymmetric gait is prevalent among older adults as well as in people with gait pathologies (e.g., Parkinson's disease, following a stroke) and has been linked to a higher risk of falls. While a certain level of gait asymmetry is present in healthy young adults, the simulation of larger asymmetry in this population provides information about efficient strategies to regulate dynamic stability. Research investigating gait asymmetry has described spatiotemporal and kinematic strategies. However, limited information is available regarding changes in gait kinetics.

Research objectives: This thesis aims to determine the gait control strategies utilized by young adults to regulate simulated gait asymmetry combined with different arm movement amplitude. We hypothesized that the hip joint would be the most affected by different arm swing conditions and asymmetric gait. We also hypothesized that asymmetric gait and active arm swing would lead to increased variability in the lower-limb movements.

Methods: Fifteen healthy young adults (23.4 ± 2.8 years, 7 Females) walked with three arm swing conditions (held, normal, and active) during symmetric and asymmetric walking conditions. The CAREN-extended System (Motek Medical, Amsterdam, NL) was used for data collection. Outcome measures included step length and width mean and variability (meanSD), vertical ground reaction forces, and lower-limb joint moment impulse mean and variability (meanSD) in the sagittal and frontal planes.

Results: When comparing arm swing conditions, the active arm swing led to an increase in step width variability, vertical ground reaction forces, hip and knee variability in the sagittal plane, ankle abduction moment for the fast leg, as well as in knee variability in the frontal plane. As for gait symmetry, the asymmetric condition led to increased step width for the fast and slow legs and increased vertical ground reaction forces for the slow leg. The asymmetric condition further prompted adjustments in the frontal and sagittal planes, particularly at the ankle and hip joints during the braking phase, and in the knee joint during the propulsion phase phases when compared to the symmetric walking condition.

Conclusion: Findings suggest that participants increased hip activity to control the effect of arm motion on the trunk to minimize adaptations in the lower-limb joints. The asymmetry condition demonstrated that participants searched for flexible lower-limb strategies aimed at minimizing bilateral differences between the fast and slow legs during asymmetric walking.

Table of Contents

<i>Abstract</i>	<i>II</i>
<i>Table of Contents</i>	<i>IV</i>
<i>List of Tables and Figures</i>	<i>VI</i>
<i>List of Abbreviations and Definitions</i>	<i>VII</i>
<i>Acknowledgements</i>	<i>VIII</i>
<i>Chapter 1: Introduction</i>	<i>1</i>
<i>Chapter 2: Purpose and Hypotheses</i>	<i>5</i>
2.1 Research Purpose	5
2.2 Independent Variables	5
2.3 Dependent Variables	5
2.3.1 Spatiotemporal	5
2.3.2 Kinetic	5
2.4 Hypotheses	6
<i>Chapter 3: Review of Literature</i>	<i>7</i>
3.1 Human Gait	7
3.2 Spatiotemporal and Kinematics strategies to regulate gait	7
3.3 Kinetics strategies to regulate gait	9
3.4 Arm swing	14
<i>Chapter 4: Methodology</i>	<i>17</i>
4.1 The relationship of this project with previous papers from our group	17
4.2 Participants	18
4.3 Data Collection	18
4.4 Experimental Protocol	19
4.5 Data Analyses	20
4.6 Statistical Analysis	22
<i>Chapter 5: Manuscript</i>	<i>23</i>
<i>Abstract</i>	<i>24</i>
5.1 Introduction	25
5.2 Methodology	27

5.2.1	Participants	27
5.2.2	Data Collection.....	27
5.2.3	Experimental Protocol.....	28
5.2.4	Data Analyses.....	28
5.2.5	Statistical Analysis	29
5.3	Results	30
5.3.1	Spatiotemporal	30
5.3.2	Kinetics	31
5.4	Discussion.....	35
5.4.1	Symmetry	36
5.4.2	Arm swing.....	42
5.4.3	Limitations	44
5.5	Conclusion.....	45
5.6	References	46
Chapter 6: Conclusion		51
6.1	Conclusion.....	51
6.2	Impact.....	51
6.3	Limitations	52
Chapter 7: References		54

List of Tables and Figures

4.1 - Figure 1. Organigram representing the focus of previous studies from our group	18
4.3 - Figure 2. 57-marker set model used for data collection.....	19
4.5 - Figure 3. Anterior-posterior ground reaction forces (N/Kg), and lower-limb joint moment impulse (Nm*Kg) during the stance phase (0-100%).....	21
5.3.1 - Table 1. Spatiotemporal (step length and width) mean (mean (SD)) and meanSD (mean (SD)) for the fast and slow legs and arm swing conditions	29
5.3.2 - Table 2. Vertical GRF (mean (SD)) for the fast and slow legs and arm swing conditions	30
5.3.2 - Table 3. JMI mean (mean (SD)) in sagittal (braking and propulsion phase) and frontal (X) planes for the fast and slow legs ankle, knee, and hip joints.....	31
5.3.2 - Table 4. JMI meanSD (meanSD (SD)) in frontal (X) and sagittal (Y) planes for the fast and slow legs ankle, knee, and hip joints.....	32

List of Abbreviations and Definitions

Dynamic stability	The ability to walk without falling even in the presence of perturbations
ROM	Range of motion
JMI	Joint moment impulse
AP	Anterior-posterior
ML	Mediolateral
VR	Vertical
GRF	Ground reaction forces
BP	Braking phase – First ~60% of the stance phase
PP	Propulsion phase – Last ~40% of the stance phase
COM	Center of mass
CNS	Central nervous system
BOS	Base of support
GC	Gait cycle
BI	Braking impulse – Joint moment impulse during the braking phase
PI	Propulsion impulse – Joint moment impulse during the propulsion phase
DoF	Degree of freedom
HH	Helen Hayes marker set

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Chapter 1: Introduction

Gait dynamic stability is defined by the ability to walk without falling, even in the presence of perturbations. These perturbations can arise from internal factors, such as neurodegenerative conditions, or external sources like a slippery surface or uneven terrain, which could lead to falls if effective strategies are not employed efficiently (Hamacher et al., 2019). Gait control strategies allow for modification of lower-limb joint activity to regulate the gait pattern and dynamic stability. Prior studies have used kinematics or spatiotemporal metrics such as joint angles, joint range of motion (ROM), step length, step time, and step width to characterize gait control (Bailey et al., 2022; Hill & Nantel, 2019; Park et al., 2016; Hinkel-Lipsker & Hahn, 2018; Peterson et al., 2012; Roemmich et al., 2012). However, a recent study by Bailey and colleagues (2022) suggested that relying solely on joint kinematic and spatiotemporal metrics for movement analysis does not provide a comprehensive understanding of gait control strategies. The incorporation of kinetic data into movement analysis offers insights into the magnitude and direction of forces acting on segments and joints, thus contributing to a more comprehensive understanding of gait control strategies (Neptune et al., 2004; Orendurff et al., 2008; Peterson et al., 2012; Roemmich et al., 2012).

To modulate walking speed and maintain postural stability, joint moment impulses (JMI) (i.e., the time integral of the forces applied at a joint in a specific direction) are applied to joints in the lower extremity in the anterior-posterior (AP), mediolateral (ML) and vertical (VR) directions to modify ground reaction forces (GRF) (Peterson et al., 2012; Roemmich et al., 2012). The purpose of JMI is to quantify the joint moment over a defined period. Most studies have overlooked possible kinetic strategies by not accounting for the impulse of kinetic variables. In challenging gait conditions, individuals without gait impairments might primarily modify the timing and intensity of muscle contraction rather than recruiting a different muscle group. Peterson et al. (2012) showed that during challenging gait

conditions, variations in JMI can lead to differences in joint movements, as well as modifications in kinematic and spatiotemporal variables alterations to appropriately maintain gait stability. However, in cases where a normal gait pattern is not possible due to lower-limb asymmetry or atypical arm motion, alternative gait control strategies must be employed to ensure forward progression and maintain dynamic stability. (Bailey et al., 2022; Canton & MacLellan, 2018).

It is possible to determine different gait control strategies used to restore stability during and after motor perturbations while walking asymmetrically (Bruijn et al., 2013; Dietz et al., 1994). Studies assessing asymmetric gait using a split-belt treadmill, where one leg moves at a different speed than the other, have revealed that the timing within each stride cycle can be modified, leading to an extended stance phase time on the slower side. This extension provides additional time to implement necessary adaptations for maintaining stability (Buurke et al., 2019; Bailey et al., 2022; Hof et al., 2008). Furthermore, recent studies looked at the roles of arm swing during gait and determined that compared to normal arm swing, restricted arm swing increases the center of mass (COM) movement in the VR direction and trunk angular movement, both of which contribute to decreased gait stability (Hill & Nantel, 2019; de Graaf et al., 2019). Conversely, actively increasing arm swing decreases the VR COM movement and trunk angular movement, subsequently enhancing gait stability (Hill & Nantel, 2019; Siragy et al., 2020; de Graaf et al., 2019). However, unlike natural arm motion, which is an automatic mechanism (i.e., does not require attention demand (Wu et al., 2016)), actively increasing arm swing amplitude requires attentional demand (Hill & Nantel, 2019; Wu et al., 2016). Consequently, the increased attentional demand during active arm swing could potentially lead to decreased stability and increased variability in the lower-limb motion (McFadyen et al., 2009; Hill & Nantel, 2019; Wu et al., 2016).

Gait stability is regulated by lower-limb stride-to-stride adaptations regardless of arm swing motion. As suggested by Bailey et al. (2022) spatiotemporal variability is likely dictated by variations in

lower-limb joint movements. Various studies examined the effect of arm swing amplitude (Siragy et al., 2020; Hill & Nantel, 2019; Eke-Okoro et al., 2007; Punt et al., 2015) and lower-limb asymmetry (Bailey et al., 2022; McFadyen et al., 2009; Hill & Nantel, 2019) on lower-limb joints variability. While a certain level of motor variability was shown to be preferable for a steady gait pattern, insufficient motor variability leads to reduced gait stability (Bailey et al., 2023). Other studies have supported these findings and have suggested that in healthy young adults, an effective gait control strategy ensures gait stability while allowing a certain degree of variability in gait kinematics (Buurke et al., 2019; Bailey et al., 2022; Hill & Nantel, 2019; Punt et al., 2015). However, none of these studies investigated gait kinetic patterns and variability.

Different studies have examined the roles of the hip, knee, and ankle joints during normal walking (Neptune et al., 2001; Zajac et al., 2002). Others focussed on changes in walking speed when looking at muscle activation and kinetic variables (Peterson et al., 2011; Orendurff et al., 2008). While the existing literature provides a strong understanding of lower-limb joint kinetics during normal walking, there is a significant dearth of knowledge concerning the contribution of lower-limb joint kinetics in governing gait stability under more complex conditions, such as asymmetric gait or modified arm swing motion. To the best of our knowledge, only a few studies have examined lower-limb kinetics during asymmetric gait (Roemmich et al., 2012) or while altered arm swing (Umberger et al., 2008). The study conducted by Roemmich et al. (2012) investigated the lower-limb kinetics during asymmetric gait in healthy young adults, specifically focussing on the AP GRF impulse and AP JMI for the hip, knee, and ankle joints. They found significant interlimb differences in the ankle and hip joints during the braking phase (BP) and propulsion phase (PP) in the sagittal plane. Even though the authors were able to clearly define the kinetics of each of the lower-limb joints in the sagittal plane, they omitted analyses in the frontal plane, despite its critical role for gait dynamic stability. The study of Umberger et al. (2008) explored the effects of restricting arm swing on gait kinematics and kinetics in the sagittal plane. The

kinetic metrics included AP GRF and AP hip, knee, and ankle joint moment and joint power. They found very few kinetic differences between walking with and without arm swing. Their main finding was an increased knee extension moment when walking without arm swing, possibly due to an increased knee flexion posture during the stance phase. Only one transverse plane metric was included in their study (free vertical moment) as they expected arm swing significant differences in that plane of motion. They found increased variability for the external rotation moment, suggesting a possible link to upper-limb movements rather than the lower-limb motion. Only one arm swing condition was included in their protocol and similarly to Roemmich et al. (2012), they did not investigate the frontal plane of motion. A large body of literature has highlighted the critical role of the frontal plane in gait stability (Hinkel-Lipsker & Hahn, 2018; Siragy et al., 2020; Fettrow et al., 2021; Bailey et al., 2022). Thus, investigating kinetic metrics in both the sagittal and frontal planes could offer further insights into gait control strategies implemented to regulate dynamic stability.

The purpose of this study was to determine the gait control strategies used by young healthy adults to regulate simulated gait asymmetry combined with different arm movement amplitude. We hypothesized that the hip joint would be the most affected by different arm swing conditions and asymmetric gait. We further hypothesized that the combination of asymmetric gait and active arm swing would lead to increased variability in the lower limb.

Chapter 2: Purpose and Hypotheses

2.1 Research Purpose

The aim of this study is to examine the lower extremity gait control strategies taking place during asymmetric walking when modifying natural arm swing.

2.2 Independent Variables

- a) Walking conditions: (1) Symmetric, and (2) Asymmetric.
- b) Arm swing condition: (1) Held, (2) Normal, and (3) Active.

2.3 Dependent Variables

2.3.1 Spatiotemporal

- c) Step length mean
- d) Step width mean
- e) Step length meanSD
- f) Step width meanSD

2.3.2 Kinetic

- g) Bilateral vertical ground reaction forces impulses during the braking phase
- h) Bilateral vertical ground reaction forces impulses during the propulsion phase
- i) Bilateral hip, knee, and ankle mediolateral and anterior-posterior joint moments impulses during the braking and propulsive phases.
- j) Bilateral hip, knee, and ankle mediolateral and anterior-posterior joint moments impulses variability (MeanSD).

2.4 Hypotheses

Anterior-posterior and mediolateral joint moment impulse of the:

- a) The hip will be affected by arm swing conditions during symmetric and asymmetric gait conditions.
- b) The knee and ankle will not be affected by arm swing during symmetric and asymmetric gait conditions.

Ground reaction forces impulses will be:

- c) Increased during held arm swing conditions.
- d) Bilaterally different between the fast and the slow leg during asymmetric gait when compared to the symmetry condition.

Active arm swing and asymmetric gait will lead to more variability in the lower limb.

Chapter 3: Review of Literature

3.1 Human Gait

Gait control can be defined by the interaction between two main actions; forward progression, and dynamic stability (Yogev et al., 2007). Therefore, a primary strategy to maintain gait stability when walking in a challenging environment is to reduce gait speed or stop walking completely (Yogev et al., 2007). Gait adaptations are automated actions controlled by the central nervous system (CNS) that alter the lower-limb movement and often occur when coping with challenging daily tasks and situations (Marigold & Patla 2002). These adaptations are essential for optimizing gait efficiency while minimizing the metabolic expenditure associated with walking (Marigold & Patla, 2002; Pieper et al., 2021; Sanchez et al., 2019). When assessing the impact of different conditions on gait activities, measures related to gait velocity, dynamic stability, and control strategies are commonly reported. Therefore, parameters pertaining to spatiotemporal, kinematics, and kinetics of the gait pattern will be reviewed in the following sections.

3.2 Spatiotemporal and Kinematics strategies to regulate gait

The main purpose of gait adaptations and control mechanisms during normal gait is to optimize metabolic efficiency (i.e., reducing metabolic cost) (Sanchez et al., 2019; Umberger. 2008). The efficiency of the walking pattern can be defined based on three characteristics i.e., gait stability; the ability to regulate internal and external perturbations to prevent falls (Hamacher et al., 2018), gait symmetry; when all gait variables from the two sides of the body are similar, independent, and evenly distributed (Wu et al., 2015), and preferred walking speed; the walking speed at which a person naturally chooses to walk (Orendurff et al., 2008). Thus, the most efficient gait can be described as a stable and symmetric gait at a self-selected walking speed (Umberger. 2008).

During normal walking, control strategies are continuously used to adjust gait parameters in order to maintain stability (Orendurff et al., 2008; Roemmich et al., 2012). Gait dynamic stability,

defined as the ability to dynamically redirect the COM within the base of support (BOS) during forward progression, necessitates the ongoing adaptations of spatiotemporal and kinematic variables to accommodate the environment. Spatiotemporal variables include but are not limited to, step length, width, and time and kinematic variables include COM's position and velocity, whole-body angular momentum, and trunk linear and angular velocity. Numerous studies have included these variables when assessing gait kinematic characteristics in normal walking and more challenging conditions such as asymmetric walking. Additionally, the authors presented variables such as the coefficient of variation, the margin of stability, and harmonic ratios (Siragy et al., 2020; Hinkel-Lipsker & Hahn, 2018; Dietz et al., 1994; Allen et al., 2011), which all provide key information on gait variability and stability related to different gait conditions.

Gait kinematics and spatiotemporal parameters are commonly used to measure gait symmetry, one of the three necessary characteristics allowing an efficient gait pattern. When walking in asymmetric conditions, it was shown that participants modified their gait pattern through spatiotemporal parameters such as stride length, swing time (Dietz et al., 1994; Allen et al., 2011), and stance time (McFadyen et al., 2009; Bailey et al., 2022) with the aim to reverting to a symmetric gait pattern. Authors suggested that these changes are made to preserve dynamic stability specifically at the trunk and pelvis (McFadyen et al., 2009; Bailey et al., 2022). Similarly, walking speed is reduced to ensure gait stability and symmetry (Bailey et al., 2022) when one or both variables are not optimal. Conversely, walking speed can be increased once stability and asymmetry are restored to their normal values (Bailey et al., 2022). A split-belt treadmill can be used to simulate gait asymmetry as the belts under each foot can be set to different speeds. Different asymmetry ratios have been used in the literature from a mild ratio of 0.8:1 (Siragy et al. 2020, Hill & Nantel, 2019) up to a larger ratio of 1.5:2 (Hinkel-Lipsker and Hahn, 2018) and 2:1 (Hinkel-Lipsker and Hahn, 2018, Roemmich et al. 2012). Speed protocols also varied with studies using pre-selected walking speeds for each participant (Siragy et al.

2020; Hill & Nantel, 2019, Hinkel-Lipsker and Hahn, 2018) and others using participant-specific walking speeds (Roemmich et al. 2012). However, regardless of the speed and asymmetry ratio, all studies showed increased variability in gait stability during asymmetric gait, this was due to participants' inability to promptly modulate their walking speed in response to the imposed asymmetry (Hill & Nantel, 2019; Siragy et al., 2020; Hinkel-Lipsker & Hahn, 2018; Roemmich et al., 2012).

During the swing phase of a normal gait pattern, alterations in the hip joint ROM affect foot trajectory and contact with the ground (Winter, 1992). During symmetric gait, where both legs share the same walking speed, consistent hip and foot placement are used on the left and the right legs. However, Hinkel-Lipsker and Hahn, (2018) reported that during asymmetric gait conditions, while sagittal plane adaptations were present at the thigh segment for both the slow and the fast limb, these adaptations were different for each leg. For instance, thigh swing velocity increased for the fast limb and decreased for the slow limb, while peak linear velocity and peak linear acceleration decreased for the slow limb. The authors suggest that the thigh adaptations in the slow and fast limb velocity are used to modify the BOS and increase trunk and COM stability (Hinkel-Lipsker & Hahn, 2018). These findings suggest that the hip is partially responsible for postural stability in the frontal plane but may not singularly suffice to maintain dynamic stability. Similar to Fettrow et al. (2021) the authors reported smaller (fine-tuning) shank and foot variations in the sagittal plane for optimal foot placement which would appear to be a strategy used to increase dynamic stability. Authors of both studies propose that the hip and ankle adaptations are used for increasing the COM and dynamic stability, consequently allowing for a more coordinated gait pattern and increased frontal plane dynamic stability. Therefore, two distinct strategies exist to regulate stability in the frontal plane which are controlling for the BOS and the COM.

3.3 Kinetics strategies to regulate gait

Kinetic variables can provide important information regarding the cause of the movements and adaptations during gait. Analysis of GRF provides a comprehensive overview of the biomechanical load

on the lower extremity during the gait cycle (GC). Previous studies divided the GC into different functional phases, i.e., a first loading peak occurring during the double-leg support phase (2-12% of GC) and representing the initial contact of the foot on the ground (weight acceptance and braking impulse (BI)) (Perry & Burnfield, 2010; Winter et al., 1992). During the double-leg support phase, VR GRF traverses from 0 to 100% of body weight (BW) as weight is shifted from one foot to another, leading to the single-leg support phase (12-31% of GC). During the single-leg support phase, VR GRF is reduced below 100% of BW when the center of gravity is transitioning forward and is at its highest point (midstance) to eventually reach the second loading peak during the second double-leg support phase (50-60% of GC). The second loading peak occurs during the pre-swing phase (50-60% of GC), when the opposite foot contacts the ground, and the supporting foot produces a propulsion force (toe-off and propulsion impulse (PI)). The double-leg support phase accounts for the initial 10% of the stance phase, followed by 40% of single stance and a final 10% of double-leg support. Roemmich et al. (2012) demonstrated that gait adaptations predominantly occur during the first and second double-leg support phases (the first and last 10% of the stance phase) which only allows for a short period for adapting gait.

During symmetric walking, the ML GRF impulses are used to maintain dynamic stability (Roemmich et al., 2012), while the AP GRF impulses are used for speed regulation (Perry & Burnfield, 2010). The PI occurs during the pre-swing phase (50-60% of GC) and its purpose is to propel the COM forward while the BI occurring during the initial double-support phase is used to regulate the walking speed by modulating the COM velocity. During asymmetric walking, the analysis of the AP GRF impulses showed asymmetries in propulsive and braking phases between the fast and the slow limb. For the fast leg, the stance phase duration was shorter resulting in greater AP GRF peaks during both propulsive and braking phases (Roemmich et al., 2012). The authors proposed that these differences were caused by unequal weight acceptances due to non-uniform trunk accelerations during asymmetric walking. As for the ML GRF impulses during asymmetric gait, the ankle inversion and eversion moments control the ML

location of the center of pressure under the foot and seem to be a prominent gait stability strategy but are not enough to regulate ML dynamic stability. Therefore, adjustments in spatiotemporal and BOS (step time, step length, and stance width) are essential to maintain dynamic stability during asymmetric gait (Buurke et al., 2019; Rawal & Singer, 2021).

Joint moments are critical in modifying spatiotemporal variables and BOS during gait. Modifications of joint moments alter the trajectory during the swing phase and the leg position during the stance phase, thereby impacting metrics like step length and width (Bailey et al., 2022). During normal gait, the hip AP moments have been shown to be critical for the regulation of postural stability followed by AP knee and ankle moments and foot placement (Peterson et al., 2012). ML regulation strategies are minimal during normal gait due to the absence of significant GC perturbations (Peterson et al., 2012). However, Peterson et al. (2012) suggested that joint moment modifications during gait in healthy individuals are mostly used to optimize gait efficiency and regulate walking speed. Neptune et al. (2004) reported that the eccentric knee extensor moment and the concentric hip extensor moments provide body support during the early stance phase. These moments serve to decelerate knee flexion motion and walking speed (BI). Furthermore, by decelerating the knee flexion moment during the early stance phase, the knee and hip extensors provide body weight acceptance during the downward motion of the COM. Neptune et al. (2004) also proposed that changes in walking speed are regulated by increasing or decreasing the BI rather than the PI since in healthy adults, the hip extensors and flexors moments are involved in the BI but not in the PI. It is only during the mid-stance phase (12-31% of the GC) that a concentric knee flexion moment and an eccentric hip extension moment accelerate the trunk forward to allow forward progression (Neptune et al., 2004). The eccentric knee flexion moment, immediately followed by a concentric knee flexion moment seems to be efficient to redirect the COM downward and forward motion to an upward and forward motion. This redistribution of segmental power aids in the transformation of energy to facilitate efficient forward progression (Neptune et al.,

2004). Ankle activity was shown to have a far more important contribution to the trunk forward progression compared to hip and knee activity (Peterson et al., 2012; Neptune et al., 2004). Peterson et al. (2012) suggested that during normal walking a slight eccentric ankle plantar flexion moment is generated just before heel contact to reduce impact forces, but the ankle plantar flexion moment is not related to the BI. In contrast, the concentric ankle plantar flexion moment is generated to help raise the heel from the ground just before the toe-off, to provide forward propulsion (Peterson et al., 2012; Neptune et al., 2004). Similar to the knee joint activity, the transition from the eccentric to the concentric ankle plantar flexor moment during the single-leg phase may be an efficient mechanism to assure body forward progression by using the stored energy and elastic muscle properties (Neptune et al., 2004). Supporting these insights, Orendurff et al. (2008) proposed that the emergence of the concentric ankle plantar flexion moment during the propulsive phase correlates with increased walking speed rather than being a mechanism aimed at regulating walking speed. These nuanced adaptations in joint moments showcase the intricate mechanisms through which the body efficiently manages gait dynamics and stability, offering valuable insights into the interplay between joint kinetics and the mechanics of forward progression.

During normal gait, the roles of joint moments are not only limited to speed regulation and forward propulsion. They are also related to the ability to regulate symmetry and the position and speed of the COM (Sun et al., 2018; Ardestani et al., 2016; Allen et al., 2011; Pandy et al., 2010). Furthermore, as suggested by Ardestani et al. (2016), joint moments serve as a predictive indicator of gait control strategies when the walking pattern is regulated by modifying step length and not cadence. The authors proposed that increasing or decreasing step length has important AP and ML joint moment modifications. Joint moments are valuable to describe the amplitude and direction of a force applied to a specific joint. However, opposite to the impulse, the joint moment does not provide information regarding the duration of the applied force. The total duration of the joint moment can significantly

change the joints' behaviour. Roemmich et al. (2012) studied JMI during asymmetric gait conditions and reported a smaller ankle JMI and a larger knee and hip JMI for the braking phase and an increased knee and hip JMI during the propulsion phase when comparing the slow limb to the fast limb. The authors suggested that these joint moment modifications, specifically the increased hip activity during the propulsion phase, reflect potential gait control strategies to compensate for the asymmetric gait pattern. The increased AP hip activity during the asymmetric walking condition suggests that the ML hip activity should also be increased as the gluteus maximus is one of the main contributors to the hip AP movement and is partially responsible for the ML hip movement (Neptune et al., 2004).

While the study by Roemmich and collaborators (2012) allowed us to better understand gait strategies used during asymmetric walking, they solely investigated gait patterns in the AP direction. Therefore, they might have overlooked some key kinetic strategies used during split-belt walking since stability regulation in the ML direction is critical during gait (Fetrow et al., 2021; Buurke et al., 2019; Rawal & Singer, 2021; Vlutters et al., 2016; Hof et al., 2008; Sun et al., 2018; Telhan et al., 2010; Simpson et al., 2020). Previous studies looked at ML adaptations during asymmetric gait and found that individuals adopted a shorter single-leg support phase time (reduce COM lateral oscillation) and a wider step pattern (ML COP modifications) on the fast leg compared to the slow leg (Buurke et al., 2019; Rawal & Singer, 2021; Vlutters et al., 2016; Hof et al., 2008; Sun et al., 2018; Telhan et al., 2010; Simpson et al., 2020). While these studies suggested that ML kinetic changes in the ankle, knee, and hip joints should occur during asymmetric gait, none of them measured kinetic variables for quantifying ML gait control strategies. The absence of comprehensive kinetic analyses in the ML direction represents a notable gap in the understanding of how individuals regulate stability during asymmetric gait. Therefore, investigating ML kinetic variables, alongside the AP adaptations, would offer a more comprehensive understanding of the complex control mechanisms employed by the human body during challenging gait conditions.

3.4 Arm swing

Arm swing is a crucial aspect of bipedal gait, serving multiple functions (Meyns et coll., 2013; Umberger, 2008). It plays a role in reducing body angular momentum, VR GRF, joint moments, and improves gait efficiency and dynamic stability (Buurke et al., 2019; Vlutters et al., 2016; de Graaf et al., 2019). A study by Siragy and collaborators, (2020) assessed the effect of different arm swings (normal, held, and active) on postural control, trunk kinematics, and dynamic balance during symmetric and asymmetric walking. They found that both held and active arm swings increased trunk variability which in turn, prompted adaptations in foot placement to maintain the COM within the BOS. These adaptations in foot placement (e.g., increased step length and width) are controlled by the neuromuscular system for optimal final foot placement at heel strike (Umberger, 2008). Moreover, results by Siragy and collaborators (2020) highlight the central role of active dynamic gait strategies when modifying arm motion.

Umberger (2008) proposed that kinetic variables are beneficial to understand the influence of arm swings on the lower extremity during gait. The authors investigated the effect of suppressing arm swing on gait kinetics and found that in the sagittal plane, the knee joint flexion and extension moment and the free vertical moment had the largest significant changes. They found that participants had smaller knee flexion moments and greater knee extension moments during the stance phase, which was in line with the participants' tendency to maintain greater knee flexion posture when arm motion was restricted. This suggests that participants adjusted their COM to a lower position to assure postural stability. However, the authors did not provide an explanation of the importance of knee joint moment alterations during the stance phase.

Regarding the free vertical moment, Umberger. (2008) found a larger external rotation moment in the mid-to-late stance phase in the restricted arm swing condition. The increased external rotation moment of the free vertical moment is necessary to decrease the transverse plane rotation of the

shoulder girdle. However, while the authors reported variances at the knee joint in the sagittal plane of motion when the arm motion was restricted, they focussed solely on the sagittal plane. Consequently, the study did not comprehensively elucidate the adaptations occurring in this scenario, particularly in the ML directions. Additionally, Umberger (2008) controlled both stride frequency and stride length across all conditions, potentially influencing some gait adaptations. A study by Eke-Okoro and collaborators (2007), reported that restricted arm swing leads to faster and shorter steps. Furthermore, Punt et al. (2015) found that greater ML dynamic stability during active arm swing motion is not related to step width or step time adaptations but rather is a result of a reduction of the whole-body angular momentum which is controlled by the hip muscles. The reduction of the whole-body angular momentum allows for greater ML dynamic stability as it results in lower corrective transverse plane hip joint moments in the stance leg. The hip muscles are responsible for the transverse and ML plane movements. Given the anatomy of the hip musculature, as its contribution decreases in the transverse plane in the active arm swing condition, its contribution increases in the ML plane, resulting in greater ML dynamic gait stability (Punt et al., 2015; Neptune et al., 2004).

However, regulating AP and ML gait stability in gait becomes more challenging during asymmetric gait. During asymmetric walking, inter-limb and intra-limb coordination, and lower-limb muscle activation coordination (Bailey et al., 2022) are regulated by automated gait control mechanisms. Nonetheless, maintaining dynamic balance during the loading and unloading of asymmetric gait limbs demands an increased level of conscious attention (McFadyen et al., 2009). McFadyen and collaborators (2009) proposed that when asymmetric gait is performed simultaneously with an active arm movement, it cannot be considered a “Dual Task” because of the level of automated processes taking place to regulate asymmetric gait. However, recent studies demonstrated that combining gait asymmetry and altering arm swing (held and active) leads to alterations in spatiotemporal, kinematic, and kinetic aspects of the gait pattern (Bailey et al., 2022; Siragy et al., 2020;

Hill & Nantel, 2019; Grabiner & Troy, 2005). Grabiner and Troy, (2005) found that while performing an attention-demanding task combined with increased arm swing, young adults adopted a more cautious gait pattern (i.e., using larger step width, shorter step length and shorter step time). However, in this study, the larger step width was also combined with a reduction in its variability. The authors proposed that decreased step width variability was associated with decreased ML trunk motion, indicating an increase in dynamic stability. Bailey et al. (2022) suggested that the stability of the trunk and pelvis during asymmetric walking might be due to the high variability seen in the BOS. The variability in the BOS is driven by the lower limb joint motion, to increase the control over the trunk and pelvis during this challenging walking task. During the active arm swing trials, Bailey et al. (2022) found plane-specific stride-to-stride adaptations for the ankle and the hip joints to maintain dynamic stability. They found decreased stability in the hip abduction (frontal plane) and ankle dorsiflexion joint angles as well as a decreased regularity in ankle dorsiflexion (sagittal plane). Although joint-specific changes are mostly observed in joints with larger kinematic redundancies (hip and ankle), changes in the knee joint also occur. Since the kinetic variables govern the movements of lower-limb joint and modifications in the BOS, this indicates that kinetic patterns would be responsible for preserving dynamic stability when arm swing amplitude is increased. Finally, Umberger. (2008) reported different magnitudes of the free vertical moment at the foot and knee joint moment and power during restricted arm swing and asymmetric gait conditions. Building on this, Bailey et al. (2022) suggested that arm swing exerts a more substantial influence on lower-extremity kinetics than kinematics during gait. As suggested by previous findings related to the lower-limb anatomy musculature (Punt et al., 2005) and plane-specific modifications (Bailey et al., 2022), further studying of AP and ML gait kinetic variables may be critical to better understand gait control strategies used for regulating asymmetric gait during held and active arm swing motion.

Chapter 4: Methodology

4.1 The relationship of this project with previous papers from our group

This study was part of a bigger project on interlimb coordination during human locomotion (figure 1).

Previous studies of our group looked at two populations: healthy young adults and people with

Parkinson's Disease, during multiple gait conditions. The people with Parkinson's disease group were

studied for two main gait conditions; 1) Attentional perturbations (Siragy et al., 2020) and 2) larger

perturbations (Siragy et al., 2021; Siragy & Nantel, 2018). On the other hand, the healthy young adults'

group was studied for; 1) the larger perturbation conditions which included three sub-conditions; incline

and decline walking and running (MacDonald et al., 2021), slip and trip perturbations (Gholizadeh et al.,

2019; Gholizadeh et al., 2020), and walking on irregular surfaces (Mezher et al., 2021) and 2) asymmetric

gait and arm swing modification. The asymmetric gait and arm swing modifications studies were

separated into two main categories, 1) kinetics and 2) kinematics and spatiotemporal, according to the

gait metrics used in their study. Previous studies of our group looked at the effect of arm swing

amplitude and asymmetric gait conditions on gait stability (Hill & Nantel, 2019) and on motor and trunk

variability (Bailey et al., 2022; Siragy et al., 2020) while looking at gait spatiotemporal and kinematic

metrics. This study constitutes an original contribution as it focuses on the gait kinetic metrics with the

main goal to determine different strategies used by healthy young adults to maintain stability during

asymmetric gait conditions and arm swing amplitude modifications.

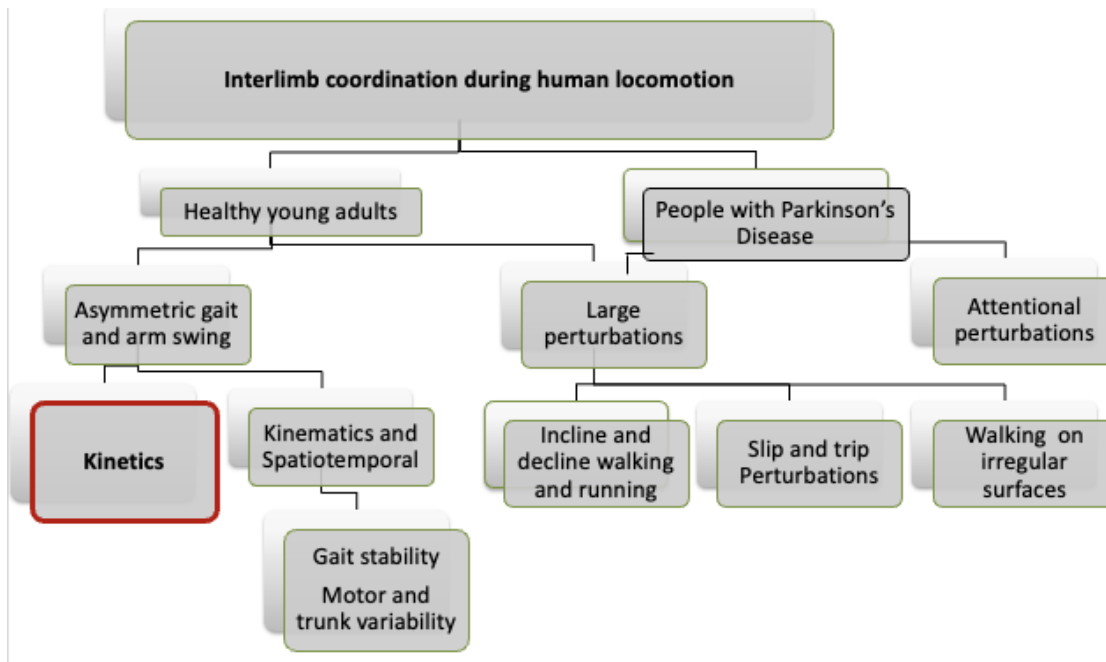


Figure 1. Organigram represents the focus of previous studies of our group, including the studied population and gait conditions. Indicated in the red frame is the focus of this study.

4.1 Participants

All participants were healthy young adults with high or medium physical activity levels. Fifteen participants (8 males and 7 females, age 23.4 ± 2.8 years, height 170.2 ± 8.1 cm, and weight 72.3 ± 13.5 kg) were recruited from the Ottawa-Outaouais community for this study. Participants were excluded if they presented any discomfort using a virtual reality system as well as musculoskeletal injuries that could have interfered with gait. This study was approved by local ethics and scientific committees and all participants freely signed informed written consent.

4.2 Data Collection

Participants were asked to wear a form-fitting terry cloth motion-capture suit to securely holds the reflective markers in place. Before data collection, a static calibration trial was performed to define the relationship between markers and a digitized trial was done to define joint centers for the shoulder, elbows, ankle, and knee (Wilken et al, 2012). The computer-assisted rehabilitation environment (CAREN-

Extended System, Motek Medical, Amsterdam, NL) was used to complete 3D motion capture using a virtual park scenario (Lemaire et al., 2012). The CAREN-Extended system combines a six-degree-of-freedom motion platform with an embedded, instrumented split-belt treadmill, 12 camera Vicon motion capture system using a 57-marker set (figure 2) (6 degrees of freedom (DoF) and Helen Hayes (HH) marker set) to track full-body kinematics (Wilken et al., 2012), 180-degree projector screen, and safety harness. Kinematic data were collected at 100 Hz and GRF at 1000 Hz. The 6DoF marker set was used to identify hip joint centers and track segment motion with separate clusters (Wilken et al., 2012; Camomilla et al., 2006; Schwartz et al., 2005; Collins et al., 2009). The HH marker set was used to track the motion of each segment by placing individual markers on anatomical landmarks (Wilken et al., 2012; Davis et al., 1991; Collins et al., 2009).

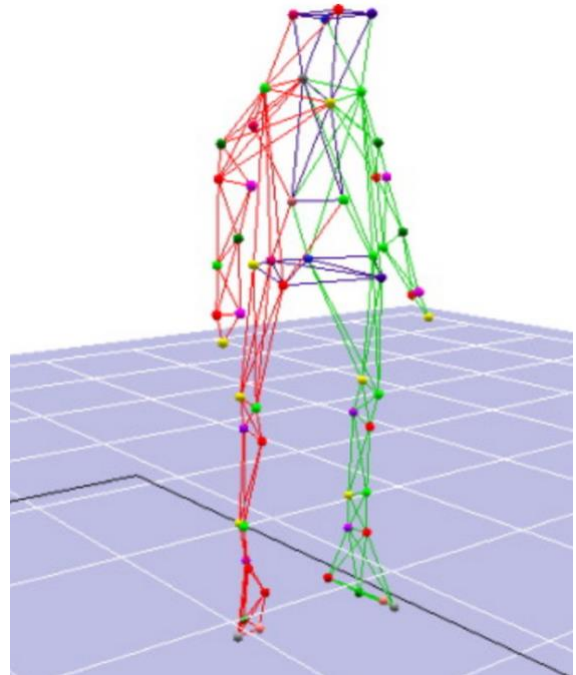


Figure 2. 57-marker set used for data collection.

4.3 Experimental Protocol

Participants were asked to walk on a split-belt treadmill (Bertec Corp., Columbus, OH). For symmetric walking, speed was set at 1.2 m/s and for asymmetric walking, walking speed was set at 1.2m/s and 0.96 m/s (1: 0.8 ratio), for the left and right leg respectively. For the sake of clarity, we will refer to the left and right legs respectively as the fast and slow legs when reporting on the asymmetric walking conditions. For symmetric and asymmetric walking conditions, participants were asked to execute three different arm swing conditions which included Held, Normal and Active. For the Held arms condition, the participants' arms were held along the body side with a harness. During the normal arms condition, the participants were asked to walk with a natural arm motion. For the third arm swing

condition, the active arm condition, the participants were instructed to bring their arm horizontally (wrist at shoulder height) at peak arm swing. A total of only 6 trials were used for data analyses and participants were allowed to rest as needed between trials to minimize fatigue. At all times, participants wore a safety harness attached to an overhead structure. All trials were performed for a total duration of three minutes.

4.4 Data Analyses

Data was imported into Visual3D (C-motion, Germantown, MD) and filtered at 10 Hz for kinematics and 20 Hz for kinetics using a 4th order zero-lag low-pass Butterworth filter (Winter, 2009). Heel strikes and toe offs gait events were calculated using an algorithm based on the heel marker velocity and vertical position (local extrema) (Roerdink et al., 2008) and used to determine the stance phase. The first 25 seconds of the trials were ignored to give time for the participant to reach a steady-state and constant walking speed.

The Following variables were calculated using Matlab (R2020b, MathWorks Inc., Natick, MA, USA):

Spatiotemporal

- a) Step length mean
- b) Step width mean
- c) Step length meanSD
- d) Step width meanSD

Kinetic

- e) Vertical ground reaction forces impulses during the braking phase
- f) Vertical ground reaction forces impulses during the propulsion phase
- g) Frontal plane joint moment impulses (hip, knee, and ankle) during the braking and propulsion phases.

- h) Sagittal plane joint moment impulses (hip, knee, and ankle) during the braking and propulsion phases.
- i) Frontal and sagittal plane (hip, knee, and ankle) joint moments impulses variability (MeanSD).

The BI and PI were determined as the negative (dark grey) and positive (light grey) values of the AP GRF, respectively (figure 3). The JMI in the frontal plane was calculated from 0-100% of the stance phase. For each frontal and sagittal plane metric, the magnitude of variability (mean standard deviation - meanSD) was calculated for each stride. Continuous series were normalized to 101 points (0-100%), finding the standard deviation between strides at each normalized point, and then taking the mean value (Bailey et al., 2022).

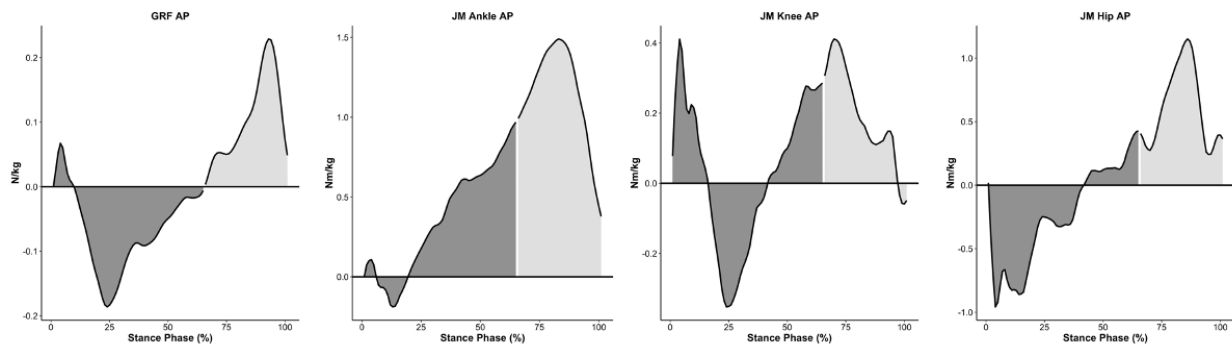


Figure 3. Anterior-posterior GRF (N/Kg), and lower-limb JMI (Nm*Kg) during the stance phase (0-100%). The filled dark grey area represents the BI and the light grey area represents the PI and is determined when the AP GRF crosses the X-axis approximately at 65% of the stance phase.

4.5 Statistical Analysis

A power analysis performed using G*power (Erdfelder & Buchner, 1996) indicated that with our sample size of 15 participants, the smallest powered detectable effect (effect size) is $f = 0.27$ when statistical power is set at 80%, at the 0.05 level of significance. Therefore, our study is well-powered to observe a medium effect size.

Using SPSS (v27, IBM, Armonk, NY, USA) a three (arm motion: Held, Normal, Active) by two (symmetry: Symmetric, Asymmetric) repeated measures ANOVA was performed to quantify the effects of arm motion on lower-limb spatiotemporal and kinetics variables. A three by two repeated measures ANOVA was performed for each of the following dependent variables. 1) Spatiotemporal gait parameters: step length mean and meanSD, step width mean and meanSD. 2) Kinetics: VR GRF, JMI (frontal and sagittal plane) and meanSD.

The significance level was set at $\alpha = 0.05$. Outliers, defined as 3 times the interquartile range away from the median, were identified for each variable using boxplots in SPSS and removed from the analysis. Sphericity was inspected and Greenhouse-Geiser p correction was reported if Mauchly's test of sphericity was violated. The assumption of normality was tested with the Shapiro-Wilks test and all post-hoc tests were performed using a Bonferroni correction.

Chapter 5: Manuscript

The Effect of Arm Swing Amplitude and Asymmetric Walking on Gait Kinetics in Young Adults

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Abstract

The aim of this study was to determine the gait control strategies employed by young adults to regulate simulated gait asymmetry in combination with different amplitudes of arm movement.

Fifteen healthy young adults (23.4 ± 2.8 years, 7 Females) walked with three arm swing conditions (held, normal, and active) during two symmetry walking conditions (symmetric and asymmetric). The CAREN-extended System (Motek Medical, Amsterdam, NL) was used for data collection. Outcome measures included the following spatiotemporal and kinetic gait parameters; step length and width mean and variability (meanSD), vertical ground reaction forces, lower-limb joint moment impulse mean and variability (meanSD) in the sagittal and frontal planes.

When comparing arm swing conditions, the active arm swing motion had the most effects on lower-limb joint kinetics in the frontal and sagittal planes compared to both held and normal arm conditions. Active arm swing led to an increase in step width variability, increased vertical ground reaction forces, increased hip and knee variability in the sagittal plane, increased ankle abduction moment for the fast leg, as well as in knee variability in the frontal plane. As for gait symmetry, the asymmetric condition led to increased step width for the fast and slow legs and increased vertical ground reaction forces for the slow leg. The asymmetric condition also prompted adjustments in frontal and sagittal planes, primarily involving the ankle and hip joints during the braking phase and in the knee joint during the propulsion phase phases in contrast to the symmetric walking condition. Findings suggest that participants increased hip activity to control the effect of arm motion on the trunk, aiming to minimize adaptations in the lower-limb joints. The asymmetry condition demonstrated that participants searched for flexible lower-limb strategies to minimize bilateral differences between the fast and slow legs during asymmetric walking.

Keywords: Dynamic Stability, Gait, Arm Swing, Symmetry, Joint Moment Impulse, Ground Reaction Forces.

5.1 Introduction

Gait dynamic stability is defined by the ability to walk without falling even with the presence of perturbations. These perturbations can be internal (e.g., due to a neurodegenerative condition) or external perturbations from the environment (e.g., a slippery surface or an uneven walking surface) which can lead to a fall if adequate strategies are not efficiently implemented (Peterson et al., 2012; Bailey et al., 2022; Canton & MacLellan, 2018). Gait control strategies allow modification of lower-limb joint activity to regulate the gait pattern and dynamic stability. While previous studies showed increased step width, decreased step length, joint angles and ROM modifications it was suggested that movement analyses perform solely using joint kinematics do not allow for a full understanding of gait control strategies (Bailey et al., 2022; Hill & Nantel, 2019; Park et al., 2016; Hinkel-Lipsker & Hahn, 2018; Peterson et al., 2012; Roemmich et al., 2012). Incorporating kinetic data in movement analyses provides a mean to quantify the magnitude and direction of forces acting on the limbs. These forces are derived from the interaction between the foot and ground, offering a comprehensive insight into the strategies used to control gait stability (Neptune et al., 2004; Orendurff et al., 2008; Peterson et al., 2012; Roemmich et al., 2012). A common kinetic metric used in the literature is the joint moment impulse (JMI) (i.e., the time integral of the forces applied at a joint in a specific direction), which is applied to the joints of the lower extremity in the anterior-posterior (AP), mediolateral (ML) and vertical (VR) directions to modify ground reaction forces (GRF) (Peterson et al., 2012; Roemmich et al., 2012).

Specific joint behaviour at the hip, knee, and ankle plays a crucial role in facilitating forward progression and ensuring stability during normal and asymmetric gaits (Bailey et al., 2022; Canton & MacLellan, 2018; Bruijn et al., 2013; Dietz et al., 1994). Studies assessed the role of the hip, knee, and ankle joints during normal walking (Neptune et al., 2001; Zajac et al., 2002). Others focussed on walking speed alterations when looking at muscle activation and kinetic variables (Peterson et al., 2011; Orendurff et al., 2008). These studies and others demonstrated that the ankle plantar flexion and knee

extension moments are responsible for forward progression and that the ankle plantar flexor, knee extension and hip extension moments support overall body stability (Peterson et al., 2011; Orendurff et al., 2008; Neptune et al., 2004). They also demonstrated that the hip is the joint mainly responsible for maintaining the head, arms, and trunk (HAT segment) level and regulating vertical accelerations of the pelvis and trunk. While the existing literature provides a strong understanding of lower-limb joint kinetics when walking in normal conditions, there is limited knowledge of the role of lower-limb joints kinetics in the control for gait stability during more challenging gait conditions (e.g., asymmetric gait conditions and modified arm swing motion). To the best of our knowledge, only a handful of studies looked at lower-limb kinetics during asymmetric gait (Roemmich et al., 2012) or while altering natural arm swing motion (Umberger et al., 2008). Roemmich et al. (2012) reported increased ankle moment impulses during the braking and propulsion phases and lower knee moment impulses during the propulsive phase for the fast leg. The authors also reported differences in the lower limb mechanics during asymmetric gait compared to normal walking. In a simulation study, Umberger et al. (2008) focussed on the effect of suppressing natural arm motion and reported similar lower limb kinetics; including joint angles, GRF, joint moments, and free vertical moment compared to walking with natural arm movements. However, this study only included sagittal planes and one variable in the transverse plane of motion which does not provide a comprehensive evaluation of lower limb kinetics.

Furthermore, studies looked at the effect of arm swing amplitude (Siragy et al., 2020; Hill & Nantel, 2019; Eke-Okoro et al., 2007; Punt et al., 2015) and lower-limb asymmetry (Bailey et al., 2022; McFadyen et al., 2009; Hill & Nantel, 2019) on lower-limb joints variability. They demonstrated that while a certain amount of motor variability seems to be preferable for a steady gait pattern, too much or too little motor variability both results in decreased gait stability (Bailey et al., 2023). The literature proposed that in healthy young adults, an efficient gait control strategy regulates gait stability while allowing a certain degree of variability in gait kinematics (Buurke et al., 2019; Bailey et al., 2022; Hill &

Nantel, 2019; Punt et al., 2015). While this information is important to understand how gait is regulated in normal and challenging environments, providing information on gait kinetics would allow for a more holistic characterization of gait strategies.

Therefore, the purpose of this study was to examine the effect of arm swing amplitude and lower-limb asymmetry on step length and width, GRFs, and lower-limb joint kinetics (JMI) and their variability across the frontal and sagittal planes. We hypothesized that the hip joint would be the most affected by different arm swing conditions and asymmetric gait. We also hypothesized that asymmetric gait and active arm swing would lead to increased variability in the lower-limb parameters.

5.2 Methodology

5.2.1 Participants

All participants were healthy young adults with high or medium physical activity levels. Fifteen participants (8 males and 7 females, aged 23.4 ± 2.8 years, 170.2 ± 8.1 cm tall, and weighing 72.3 ± 13.5 kg) were recruited from the Ottawa community for this study. Participants were excluded if they presented any discomfort using a virtual reality system as well as musculoskeletal injuries that could have interfered with gait. This study was approved by local ethics and scientific committees and all participants freely signed informed written consent.

5.2.2 Data Collection

The computer-assisted rehabilitation environment (CAREN-Extended System, Motek Medical, Amsterdam, NL) was used to complete 3D motion capture using a virtual park scenario (Lemaire et al., 2012). The CAREN-Extended system combines a six-degree of freedom motion platform with an embedded, instrumented split-belt treadmill, 12 camera Vicon motion capture system using a 57-marker set (6 DoF and HH marker set) to track full-body kinematics (Wilken et al., 2012), 180-degree projector screen, and safety harness. Kinematic data were collected at 100 Hz and GRF at 1000 Hz.

5.2.3 Experimental Protocol

Participants were asked to walk on a split-belt treadmill (Bertec Corp., Columbus, OH). For symmetric walking, speed was set at 1.2 m/s and for asymmetric walking, walking speed was set at 1.2m/s and 0.96 m/s for the left and right leg (1: 0.8 ratio), respectively. For symmetric and asymmetric walking conditions, participants were asked to execute three different arm swing conditions which included 1; Held – participants' arms were held along the body side with a harness, 2; Normal – where participants were asked to walk with a natural arm motion, and 3; Active - where participants were instructed to bring their arm horizontally (wrist at shoulder height) at peak arm swing. A total of only 6 trials for a duration of 3 minutes were used for data analyses.

5.2.4 Data Analyses

Data was imported into Visual3D (C-motion, Germantown, MD) and filtered at 10 Hz for kinematics and 20 Hz for kinetics using a 4th order zero-lag low-pass Butterworth filter (Winter, 2009). Heel strikes and toe offs gait events were calculated using an algorithm based on the heel marker velocity and VR position (local extrema) (Roerdink et al., 2008) and used to determine the stance phase. The first 25 seconds of the trials were ignored to give time for the participant to reach a steady-state and constant walking speed. Using Matlab (R2020b, MathWorks Inc., Natick, MA, USA), VR GRF, spatiotemporal (step length and step width), and JMI (frontal and sagittal plane for ankle, knee, and hip) means were calculated during the BI and PI of the stance phase. The BI and PI were determined as the negative and positive values of the AP GRF, respectively. The JMI in the frontal plane was calculated from 0-100% of the stance phase. For each frontal and sagittal plane metric, the magnitude of variability (mean standard deviation - meanSD) was calculated for each stride. Continuous series were normalized to 101 points (0-100%), finding the standard deviation between strides at each normalized point, and then taking the mean value (Bailey et al., 2022).

5.2.5 Statistical Analysis

Using SPSS (v27, IBM, Armonk, NY, USA) a three (arm motion: Held, Normal, Active) by two (symmetry: Symmetric, Asymmetric) repeated measures ANOVA was performed to quantify the effects of arm motion on lower-limb spatiotemporal and kinetics variables. This was performed for each of the following variables: VR GRF, step length mean and meanSD, step width mean and meanSD, JMI (frontal and sagittal plane) and meanSD. The significance level was set at $\alpha = 0.05$. Outliers, defined as 3 times the interquartile range away from the median, were identified for each variable using boxplots in SPSS and removed from the analysis. Sphericity was inspected and Greenhouse-Geiser p correction was reported if Mauchly's test of sphericity was violated. The assumption of normality was tested with the Shapiro-Wilks test and all post-hoc tests were performed using a Bonferroni correction.

5.3 Results

5.3.1 Spatiotemporal

Table 1. Spatiotemporal (step length and width) mean (mean (SD)) and meanSD (mean (SD)) for the fast and slow legs. A total of six gait conditions are divided into two symmetry conditions (Symmetric vs Asymmetric) with three arm swing conditions (Normal, Held, and Active).

Spatiotemporal (cm)		Symmetric walking			Asymmetric walking		
		Normal	Held	Active	Normal	Held	Active
Step length mean	Fast	85.38 (16.51)	84.43 (20.23)	84.99 (19.56)	84.34 (20.82)	83.24 (22.53)	92.54 (21.37)
	Slow	85.48 (16.50)	84.52 (20.22)	85.03 (19.62)	84.41 (20.87)	83.28 (22.48)	92.50 (21.37)
Step length meanSD	Fast	13.45 (3.04)	15.11 (5.17)	13.12 (3.04)	14.18 (3.99)	14.31 (3.99)	14.45 (3.49)
	Slow	13.26 (3.05)	14.86 (5.23)	12.74 (3.00)	13.84 (4.16)	14.16 (4.02)	14.14 (3.38)
Step width mean	Fast ^b	18.37 (3.61)	17.62 (3.19)	18.86 (3.68)	19.95 (4.06)	19.88 (3.34)	20.59 (4.01)
	Slow ^b	18.38 (3.60)	17.62 (3.18)	18.85 (3.70)	19.95 (4.07)	19.89 (3.34)	20.58 (4.00)
Step width meanSD	Fast ^a	1.91 (0.55)	1.96 (0.60)	2.61 (0.83)	1.97 (0.52)	1.89 (0.56)	2.26 (0.57)
	Slow ^a	2.10 (0.48)	2.16 (0.76)	2.96 (1.01)	2.21 (0.54)	2.04 (0.64)	2.63 (0.77)

^a Arm swing significant difference ($p < 0.05$)

^b Symmetry significant difference ($p < 0.05$)

Spatiotemporal main effects were found for arm swing and symmetry conditions. Active arm swing increased the average fast leg step width standard deviation (SD) compared to normal ($F(2,18) = 15.360$, $p = 0.003$) and held ($F(2,18) = 15.360$, $p = 0.005$) arm swing conditions. Active arm swing also increased the average slow leg step width standard deviation (SD) ($F(2,18) = 6.898$, $p = 0.032$) compared to the normal arm swing condition.

Asymmetric walking increased the average step width for the fast ($F(1,9) = 38.509$, $p < 0.001$) and slow ($F(1,9) = 38.430$, $p < 0.001$) legs compared to the symmetric walking condition.

5.3.2 Kinetics

5.3.2.1 Ground reaction forces

Table 2. Vertical ground reaction forces (mean (SD)) of six gait conditions divided into two symmetry conditions (Symmetric vs Asymmetric) with three arm swing conditions (Normal, Held, and Active).

GRF (N)		Symmetric walking			Asymmetric walking		
		Normal	Held	Active	Normal	Held	Active
Vertical	Fast ^a	550.406 (31.93)	554.08 (34.79)	616.38 (37.30)	554.012 (36.85)	540.95 (35.72)	606.74 (51.68)
	Slow ^{a b}	546.41 (29.99)	547.70 (38.20)	615.21 (42.19)	620.33 (48.62)	603.48 (47.22)	672.76 (74.69)
^a Arm swing significant difference		(p < 0.05)					
^b Symmetry significant difference		(p < 0.05)					

Several GRF main effects were found for arm swing and symmetry conditions. Active arm swing increased the fast leg VR GRF compared to normal ($F(2,18) = 17.569$, $p = 0.003$) and held ($F(2,18) = 17.569$, $p = 0.009$) arm swing conditions. Active arm swing also increased the slow leg VR GRF ($F(2,18) = 9.117$, $p = 0.014$) compared to the normal arm swing condition. No significant differences were found between the normal and held condition.

The treadmill asymmetric condition increased the slow leg VR GRF ($F(1,9) = 32.503$, $p < 0.001$) compared to the symmetric walking condition. No significant differences were found between the asymmetric and the symmetric walking conditions for the fast leg VR GRF.

5.3.2.2 Joint moment impulse

Table 3. JMI mean (mean (SD)) in sagittal (BP and PP) and frontal (X) planes for the fast and slow legs ankle, knee, and hip joints. A total of six gait conditions are divided into two symmetry conditions (Symmetric vs Asymmetric) with three arm swing conditions (Normal, Held, and Active).

Joint moment impulse (Nm*kg)		Symmetric walking			Asymmetric walking		
		Normal	Held	Active	Normal	Held	Active
Fast, ankle	BP ^b	13.14 (7.66)	13.97 (7.74)	13.83 (10.55)	17.83 (9.51)	16.45 (7.58)	17.50 (11.12)
	PP	39.62 (12.79)	39.78 (15.10)	36.99 (19.75)	39.05 (16.53)	43.43 (12.22)	40.12 (17.06)
	X ^{a b}	12.91 (11.66)	12.15 (12.46)	11.82 (12.19)	13.21 (12.70)	13.46 (11.94)	15.07 (9.91)
Fast, knee	BP ^b	-1.94 (10.13)	-3.04 (9.77)	-2.49 (10.23)	-1.94 (10.22)	-2.78 (9.58)	-0.68 (10.79)
	PP	7.32 (4.95)	9.72 (5.13)	10.19 (4.76)	9.93 (4.14)	9.69 (5.28)	8.94 (6.34)
	X	28.86 (12.86)	25.83 (13.67)	30.43 (13.64)	27.95 (12.31)	20.93 (13.69)	28.99 (13.99)
Fast, hip	BP ^{a b}	-6.21 (5.89)	-6.32 (6.77)	-9.28 (7.72)	-2.20 (5.59)	-4.62 (5.17)	-6.81 (6.43)
	PP	18.09 (6.81)	16.35 (5.22)	19.10 (8.62)	17.59 (6.99)	17.49 (5.57)	19.34 (8.53)
	X ^{a b}	32.69 (10.63)	26.67 (16.76)	35.58 (15.38)	30.04 (16.89)	24.17 (11.49)	28.63 (17.30)
Slow, ankle	BP ^b	18.52 (5.87)	17.59 (9.20)	14.78 (10.33)	12.46 (6.51)	11.56 (5.31)	14.16 (11.11)
	PP	47.11 (3.2)	47.71 (3.13)	44.10 (11.70)	43.76 (5.48)	43.07 (4.96)	56.09 (5.71)
	X ^b	8.94 (5.77)	8.15 (5.49)	11.12 (7.84)	10.39 (7.91)	9.10 (4.95)	10.75 (7.59)
Slow, knee	BP	0.83 (10.37)	0.53 (10.10)	1.57 (9.05)	-0.46 (11.58)	-2.30 (7.48)	-0.55 (14.30)
	PP ^b	10.56 (5.33)	10.48 (5.45)	10.41 (6.64)	8.65 (5.44)	8.60 (4.67)	10.39 (5.87)
	X	20.50 (10.54)	18.92 (10.41)	23.03 (12.43)	21.80 (11.57)	17.99 (10.47)	23.37 (11.96)
Slow, hip	BP	-8.79 (7.39)	-8.08 (7.91)	-10.26 (9.98)	-10.35 (5.74)	-10.25 (6.38)	-13.10 (8.28)
	PP	18.65 (4.26)	16.27 (5.48)	16.47 (6.89)	17.43 (6.69)	15.90 (6.10)	19.01 (6.58)
	X	25.05 (12.32)	21.53 (11.23)	23.68 (15.90)	23.44 (10.62)	25.44 (11.13)	27.38 (12.49)

BP : Braking phase during the stance phase (AP)

PP : Propulsive phase during the stance phase (AP)

X: Mediolateral direction from 0-100% SP

^a Arm swing significant difference (p < 0.05)

^b Symmetry significant difference (p < 0.05)

Table 4. JMI meanSD (meanSD (SD)) in frontal (X) and sagittal (Y) planes for the fast and slow legs ankle, knee, and hip joints. A total of six gait conditions are divided into two symmetry conditions (Symmetric vs Asymmetric) with three arm swing conditions (Normal, Held, and Active).

Joint moment		Symmetric walking			Asymmetric walking		
impulse	(Nm*kg)	Normal	Held	Active	Normal	Held	Active
Fast, ankle	X	0.032 (0.008)	0.038 (0.022)	0.039 (0.024)	0.033 (0.020)	0.047 (0.037)	0.052 (0.045)
	Y	0.092 (0.049)	0.113 (0.096)	0.108 (0.081)	0.081 (0.045)	0.155 (0.096)	0.128 (0.099)
Fast, knee	X ^a	0.053 (0.017)	0.060 (0.029)	0.065 (0.034)	0.048 (0.017)	0.059 (0.019)	0.072 (0.042)
	Y ^a	0.065 (0.022)	0.069 (0.030)	0.072 (0.036)	0.063 (0.023)	0.078 (0.025)	0.089 (0.038)
Fast, hip	X	0.062 (0.016)	0.073 (0.061)	0.078 (0.030)	0.059 (0.043)	0.089 (0.079)	0.089 (0.081)
	Y ^{a b}	0.083 (0.020)	0.091 (0.029)	0.108 (0.031)	0.086 (0.017)	0.107 (0.030)	0.125 (0.047)
Slow, ankle	X ^b	0.041 (0.009)	0.043 (0.015)	0.059 (0.042)	0.042 (0.008)	0.040 (0.009)	0.049 (0.013)
	Y	0.094 (0.044)	0.107 (0.061)	0.159 (0.113)	0.099 (0.017)	0.100 (0.022)	0.115 (0.026)
Slow, knee	X	0.050 (0.014)	0.051 (0.018)	0.067 (0.029)	0.049 (0.007)	0.049 (0.009)	0.061 (0.012)
	Y	0.074 (0.020)	0.077 (0.032)	0.098 (0.048)	0.089 (0.023)	0.086 (0.025)	0.097 (0.026)
Slow, hip	X ^a	0.061 (0.022)	0.068 (0.040)	0.081 (0.036)	0.061 (0.021)	0.061 (0.020)	0.075 (0.025)
	Y	0.089 (0.021)	0.093 (0.030)	0.124 (0.041)	0.119 (0.055)	0.117 (0.045)	0.124 (0.041)

X: ML

Y: AP

^a Arm swing significant difference (p < 0.05)

^b Symmetry significant difference (p < 0.05)

5.3.2.3 Significant interaction effect

Only one significant arm*symmetry interaction was found in this study. During the asymmetric walking condition compared to the symmetric walking condition, while walking with normal arm swing, the slow leg ankle propulsive JMI was significantly greater ($F(2,18) = 4.409, p < 0.001$).

5.3.2.4 Sagittal plane

The main effects in the sagittal plane included an increased hip propulsion JMI for the slow leg during the held arm swing condition ($F(2,18) = 4.812, p = 0.018$) compared to normal arm swing and for the fast hip braking impulse, however, the post-hoc analyses were not significant (active-normal, ($F(2,18) = 4.428, p = 0.115$) & active-held, ($F(2,18) = 4.428, p = 0.275$)). Further, active arm swing increased the hip and knee impulse variability. The fast hip JMI meanSD ($F(2,18) = 10.135, p = 0.007$) and fast knee moment impulse meanSD ($F(2,18) = 3.804, p = 0.035$) were significantly greater during the active arm condition when compared to the normal arm swing condition.

Asymmetric gait led to increased fast leg ankle braking moment impulse ($F(1,9) = 21.491, p = 0.001$) and slow leg knee propulsive moment impulse ($F(1,9) = 9.593, p = 0.013$). However, asymmetric gait decreased the braking moment impulse for the slow ankle ($F(1,9) = 6.465, p = 0.032$) and fast hip ($F(1,9) = 19.659, p = 0.02$). Additionally, the fast leg hip JMI ($F(1,9) = 5.394, p = 0.045$) increased in variability during the asymmetric gait condition.

5.3.2.5 Frontal plane

In the frontal plane, the main effects were the following. The active arm condition increased the ankle abduction JMI ($F(2,18) = 5.712, p = 0.045$) for the fast leg when compared to the held arm swing condition and also increased variability for knee JMI ($F(2,18) = 6.155, p = 0.004$) of the fast leg when compared to the normal arm motion condition. Variability main effects were found for the slow knee (active-normal, ($F(2,18) = 5.250, p = 0.135$) & active-held, ($F(2,18) = 5.250, p = 0.144$)) and hip (active-

normal, ($F(2,18) = 3.869, p = 0.088$) & active-held, ($F(2,18) = 3.869, p = 0.434$)), however, the post-hoc analyses were not significant.

Finally, during the asymmetric walking condition, an increased abduction JMI was observed for the fast ankle ($F(1,9) = 6.366, p = 0.033$) while a decreased abduction JMI was observed for the slow ankle ($F(1,9) = 11.267, p = 0.008$). To continue, no significant differences were found for the knee joints while comparing the symmetry conditions. However, the asymmetric walking condition increased the hip abduction JMI ($F(1,9) = 6.051, p = 0.036$) for the slow leg when compared to the symmetric walking condition.

5.4 Discussion

In this study, we examined how arm swing amplitude and lower-limb asymmetry affect gait stability and variability while using several kinetic metrics. When comparing arm swing conditions, the active arm swing motion had the most effects on lower-limb joint kinetics in the frontal and sagittal planes compared to both held and normal arm conditions. Regarding gait symmetry, the asymmetrical condition led to frontal and sagittal plane adjustments, mostly at the ankle and hip joints, during the braking and propulsion phases, in comparison to the symmetric walking condition. Our results partially confirm our first hypothesis as arm swing and symmetry conditions led to adaptations at the hip. However, we did not expect several adaptations at the ankle joint. Our second hypothesis is also partially supported as we found different GRF impulses for the slow leg when comparing different symmetry conditions. However, contrary to our hypothesis, it was the active arm swing and not the held arm condition that resulted in differences in GRF impulses. Finally, both arms swing and symmetry conditions led to higher variability in our measurements, thus confirming our hypothesis.

5.4.1 Symmetry

5.4.1.1 Braking moment

A previous study demonstrated that asymmetric walking causes larger linear and angular velocity in the frontal plane, which was interpreted by the authors as larger trunk instability (Siragy et al., 2020). They found that participants adopted a “cautious gait”, adjusting their gait pattern in both the frontal and sagittal planes simultaneously (Siragy et al., 2020). This is similar to our results that showed BOS, hip, and trunk modifications during the stance phase's braking moment. Consistent with the existing literature, we found that in the frontal plane, participants modify their BOS by increasing step width (Bailey et al., 2022) and altering trunk kinematics (Hill & Nantel, 2019) with a significant increase in ML hip JMI. We also observed an increased ankle abductor moment, which might serve to stabilize the ankle joint due to the increased step width in our asymmetric walking condition.

Hinkle-Lipkser and Hahn. (2018) reported that in asymmetric gait, the whole-body COM progresses at a higher velocity during the stance phase of the fast leg compared to the slow leg. Given that the slow leg experiences a prolonged stance phase in contrast to the faster leg during asymmetric gait, less time can be dedicated to controlling the COM velocity in the shorter stance of the fast leg. Consequently, this results in higher COM control during the stance phase of the slow leg (McFadyen et al., 2009; Bailey et al., 2022). Therefore, we suggest that during the stance phase of the slow leg, hip frontal plane adaptations must be implemented to decelerate, redirect, and reduce the variability of the COM. A higher COM variability is observed for the slow leg, possibly due to reduced control from the fast leg.

In recent studies, it was suggested that the CNS provides predictive adjustments for the hip frontal plane of motion for the fast limb, while changes seen at the hip level in the slow limb reflect a passive feedback control mechanism. This mechanism responds to the step-to-step variability of the COM velocity at the end of the stance phase of the fast leg (Hinkle-Lipkser et al., 2018, Fettle et al.,

2021). Different studies proposed that the asymmetry in gait kinetic and kinematic variables (Hinkle-Lipkser and Hahn, 2018; Fettrow et al., 2019) and the margin of stability (Buurke et al., 2019; Fettrow et al., 2021) between the fast and the slow leg during asymmetric gait emerges due to a lateral shift of the COM away from the fast-moving limb during the walking condition. The COM lateral shift is necessary for allowing spatiotemporal asymmetries between the fast and slow leg. By doing so, different gravitational forces are applied to the slow and the fast leg, requiring different strategies to uphold postural stability during the walking task. Fettrow et al. (2021) explained that gravitational forces acting on the COM are defined by the COM lateral distance from the COP and are proportionate to the VR acceleration force. The longer the stance time, the more time the gravitational force pulls on the COM. For the fast leg, the COM is positioned towards the opposite leg, resulting in a shorter stance phase time, which means that gravitational forces have less time to act on the limb. Therefore, an increase in the lateral distance from the COM and the CoP is necessary to compensate for the shorter stance time. The opposite is observed for the slow leg where the COM is positioned closer to the COP. This proximity creates a shorter lever arm, necessitating a longer stance time to create an equal opposite gravitational acceleration force. This compensatory mechanism allows for bilateral equal total acceleration of gravitational forces acting on the pelvis during asymmetric gait and contributes to frontal plane stability (Fettrow et al., 2019). The increased stance time for the slow leg explains the increased hip abduction JMI for the slow leg and highlights the importance of strong hip muscles for gait stability during asymmetric gait. Fettrow et al., (2019) proposed that the COM lateral shift during asymmetric gait functions as a control strategy, enabling young adults to adopt a more symmetric gait pattern in the frontal plane with minimal alterations in lower-limbs kinematics and kinetics. However, the literature proposes that older adults and individuals with gait impairments are unable to employ the same gait control strategy as young adults, mainly due to diminished lower extremity muscle strength and reduced recruitment of distal lower extremity muscles (Eng & Tang, 2007; Skinner et al., 2019). The investigation

of COM lateral shift in young adults is used in this study as a baseline comparison for other studies involving populations such as older adults and/or with gait impairments to enhance our comprehension of the strategies used by those populations in regulating ML stability during asymmetric gait.

Regarding the sagittal plane, previous studies (McFadyen et al., 2009; Bailey et al., 2022; Peterson et al., 2011) found differences in spatiotemporal metrics (step time, length, width, and frequency) between the faster and slower limb during asymmetric walking. These studies suggested that these changes are specific to each side and are made to preserve dynamic stability, particularly at the level of the trunk and pelvis. The authors expected asymmetries in the measured gait metrics because of the asymmetric walking speed of the two legs. They explained that the differences observed between the fast and the slow leg throughout the entirety of the trial serve as strategies employed to regulate dynamic stability. Bailey et al. (2022) mentioned that multiple yet simultaneous adaptations in the lower extremity joints are present during the stance phase and the swing phase to maintain stability. Similarly, during asymmetric walking conditions, Hinkel-Lipsker and Hahn, (2018) showed that the hip is partially responsible for sagittal plane stability during the stance phase. However, the hip adaptations during the stance phase in this direction may not suffice to maintain dynamic stability. Indeed, the authors also reported adaptations in the distal segment (shank and foot) occurring for optimal foot placement during the swing phase of the contralateral leg. Consistent with the literature reporting on the JMI during the stance phase, we showed significantly different BI for the hip and ankles during asymmetric walking when compared to symmetric walking. The decreased braking impulse seen at the hip in the faster leg can likely be attributed to the slower walking speed of the opposite (slow) leg. This adjustment could potentially serve as a mechanism to minimize the asymmetry between the faster and slower legs, as proposed by Orendurff et al. (2008). During the early stance phase, both the hip and ankle joints help to control for stability and forward progression, with the hip joint extension moment being the main

contributor. The contribution of the ankle joint extension moment during the early stance phase is to regulate the BI but its effect is lesser than the hip.

As the stance phase progresses, different control strategies are required to maintain stability. Towards the end of the stance phase, the opposite is observed, the ankle's extension moment contribution becomes highly important, and the hip extension joint moment contribution diminishes. During the early stance phase, the ankle musculature is responsible for generating the appropriate limb propulsion moment (Hinkle-Lipsker and Hahn, 2018; Winter et al., 2009). Furthermore, our findings in the ankle JMI are in line with the literature as the ankle moment adaptation was shown to be the most compelling strategy for regulating stability (Orendurff et al., 2008). In our study, we observed significant differences in JMI in the sagittal (braking moment) plane for the fast (increased) and slow (decreased) ankles during asymmetric walking, as compared to normal walking.

According to Orendurff et al. (2008) the increased ankle moment impulse (increase in plantar flexion moment) for the fast leg and the decreased ankle moment (decrease in plantar flexion moment) for the slow leg during the braking phase observed in our study, are strategies used to consistently regulate the COM's position relative to the COP and therefore regulate walking speed. In Orendurff and collaborators' study in 2008, the treadmill speed was constant (1.0m/s) and then increased (1.0m/s to 1.4m/s) or decreased (1.4m/s to 1.0m/s), depending on the testing condition. They found that participants adapted their walking speed within the initial two steps using an ankle strategy. However, our study maintained constant treadmill speed for the fast (1.2 m/s) and slow (0.96 m/s) belts throughout the trial. This resulted in the presence of adaptations in the sagittal plane ankle joint moment during the entire duration of the trial. In agreement with the literature, we observed the importance of hip activity during the braking phase (Winter et al., 2009). We also demonstrated the importance of ankle activity during the braking phase even if its contribution is less significant (Winter et

al., 2009). Therefore, we suggest that a multi-joint strategy is favourable to controlling the BI for sagittal and frontal plane adaptations during asymmetric gait.

5.4.1.2 Propulsion moment

Similar to other studies assessing asymmetric gait, no significant differences were found for the propulsion phase moment at the hip or the ankle. This suggests that participants regulate the walking speed with the BI rather than the PI. However, a significant increase in the slow-leg knee flexion moment (slow side only) during the propulsion phase was found in our study when comparing the asymmetric condition to the symmetric condition. Two explanations can be considered. First, Neptune et al. (2001) and Peterson et al. (2011) proposed that the fast-leg gastrocnemius' contraction is increased during the propulsion phase to increase the plantar flexor moment. The increased plantar flexor moment is necessary during the asymmetric condition to help swing initiation of the fast leg and position the ankle joint anteriorly just before the beginning of the swing phase. The increased plantar flexion moment also helps reduce the recruitment of the bigger muscle group at the thigh, thereby maximizing gait efficiency. The plantarflexed position of the foot during the pre-swing phase increases the ankle joint's VR position, creating an increased flexion moment at the knee. Peterson et al. (2011) also proposed a possible explanation for the absence of knee flexion moment for the fast leg. In contrast to Neptune's study, these authors focussed on the behaviour of the fast leg. In agreement with the existing literature, they suggest that the rectus femoris is the main muscle responsible for knee stability during the stance phase. Given that the rectus femoris muscle is responsible for knee extension, an increased contraction would lead to higher knee joint extensor moment and leg stiffness.

We observed different behaviours of the fast and slow legs' knee joints during the propulsive phase of asymmetric gait when compared to symmetric gait. Based on the previous literature and our results, we suggest that the slow leg knee joint main functions are to redistribute forces from the ankle to the hip and to facilitate forward progression during gait. We suggest that during asymmetric gait, the

fast leg knee joint main function is to regulate leg stability and, consistent with the existing literature, stabilize the upper body. To better comprehend knee muscle activity and its impact on kinetics during asymmetric walking, future work including electromyography would be essential in supporting and refining this theory.

5.4.1.3 Proximal vs distal strategy

In healthy young adults, a distal (ankle) strategy is mainly used to control for AP and ML trunk and COM stability. When the distal end of the lower limb is compromised or muscle recruitment of the gastrocnemius is decreased, there is a distal-to-proximal redistribution of the forces. This new strategy predominantly implicates the hip muscles to control trunk and COM stability (Pieper et al., 2021; Fernandez et al., 2019). The distal-to-proximal redistribution of the forces is a strategy used by older adults and people with gait pathologies such as Parkinson's disease and Stroke patients. Previous studies demonstrated that these two populations have decreased hip muscle strength (Eng & Tang, 2007; Skinner et al., 2019), resulting in a reduced capacity to laterally shift the COM towards the slow leg, which was observed in healthy young adults (Fettrow et al., 2021).

The different gait modifications embraced by older adults and people with gait pathologies, compared to healthy young adults, seem to be effective for regulating gait stability but have negative impacts on gait variability. The greater recruitment of hip muscles by older adults to regulate gait has been shown to lead to increased gait variability (Fernandez et al., 2019). This increase in gait variability in older adults could originate from the recruitment of the hip muscles which are situated farther from the point of contact with the ground compared to the knee and ankle muscles (Fernandez et al., 2019). Therefore, the increased distance of the force of action of the hip muscles to the foot might lead to a more variable trajectory of the lower extremity and final foot placement in both frontal and sagittal planes during the swing phase. However, specific lower-limb exercise programs aiming to increase lower-limb muscle strength and mobility have been proven effective in increasing executive control

effectiveness, increased lower-leg mobility, increasing gait stability, and consequently reducing risks of falls in older adults and people with gait pathologies (Zhao et al., 2022; Pizzigalli et al., 2011; Persch et al., 2009).

5.4.2 Arm swing

Recent studies assessing arm swing motion demonstrated that natural arm swing amplitude is dependent on walking speed and that whole-body adaptations are present when natural arm motion is altered (Hill & Nantel, 2019; Bailey et al., 2023). Specifically, studies demonstrated that modifying upper-limb ROM during gait has distinctive impacts on the trunk and the BOS (Hill & Nantel, 2019; Bailey et al., 2023; Siragy et al., 2020). It was shown that active arm swing increased the trunk's angular momentum, thus increasing its resistance to change (Hill & Nantel, 2019). Consequently, an increase in lower-limb variability would not impact the trunk's local dynamic stability in frontal and sagittal planes. The absence of disparities in VR GRF between the left and right legs suggests an absence of COM lateral shift, confirming the trunk's resistance to change in the frontal plane during all our arm swing conditions. However, another study from our group suggested the possibility that a more dynamic and variable nature of the BOS could be responsible for trunk stability when arm swing amplitude is increased, due to its ability to control the COM's position and velocity (Siragy et al., 2020). The results of the present study support this notion, as we found increased step width variability during our active arm swing condition when compared to the normal and held arm conditions. Therefore, consistent with the literature, we suggest that whole-body adaptations are required to maintain dynamic stability when arm swing amplitude is actively increased.

Further, Bailey et al. (2023) proposed that young adults continuously seek for flexible strategies to optimize gait stability by either directing more conscious attention to the upper limb or the lower limb. They reported that young adults alternate between greater whole-body motion adjustments and fewer trunk motion adjustments, depending on conditions of walking (asymmetric or symmetric) and

arm swing (active, normal, held). When a participant prioritized the arm motion task, the trunk's local dynamic stability was preserved, yet variability in the lower limb and BOS increased (Bailey et al., 2023). Our results support this as we found increased frontal plane knee JMI, sagittal plane knee and hip JMI variability and increased step width variability. We also found an increased frontal plane ankle abduction JMI during the active arm swing condition when compared to the held condition, suggesting that participants must adapt lower-limb joint movements to compensate for the increased BOS variability in order to preserve trunk dynamic stability.

Our group found that in healthy young adults, the absence of arm swing during gait has little to no effect on trunk and lower-limb kinematics (Bailey et al., 2022; Hill & Nantel, 2019; Siragy et al., 2020). Based on the results presented in their study, the authors proposed that restricted arm swing could have a greater effect on gait kinetics than kinematics and suggested that future studies investigate kinetic metrics to confirm that theory. Indeed, Siragy et al. (2020) suggested that the lack of differences in trunk kinematics might be due to the increased compensation from the trunk and hip musculature. Our results confirmed this hypothesis. We found decreased VR GRF for the fast leg and an increase in the slow leg hip extension JMI during the propulsion phase for the held arm condition when compared to the normal arm condition. These results suggest that the increased hip propulsion moment in the slow leg may be a compensation mechanism to increase the forward momentum of the COM to levels observed during natural arm swing movement. These findings not only support our first hypothesis that the absence of arm swing affects the hip JMI, but also confirm that the absence of arm swing has more effect on lower-limb kinetic than kinematics, as previously suggested by the literature.

Additionally, a previous study observed an increased rotational moment at the trunk and shoulder girdle when arm swing is absent (Umberger, 2008). This suggests that additional upper-limb regulation strategies are required to stabilize for trunk angular movements to maintain dynamic stability. Our findings and the recent literature suggest that young adults seek simultaneous strategies

for both upper-extremity and lower-extremity during gait. This approach is driven by the objective of optimizing gait efficiency, COM stability, and overall gait smoothness (Bailey et al. 2023; Hinkel-Lipsker & Hahn, 2019; Siragy et al., 2020; Umberger, 2008). However, older adults and people with gait pathologies are not able to adapt their gait patterns as efficiently as healthy young adults. Finally, while our study did not aim to investigate the trunk and upper-limb adjustments, our protocol did allow to collect these aspects. Therefore, future studies incorporating upper-limb kinetics and muscle activation measurements could offer valuable insight into the interplay between upper-limb and lower-limb kinetics, as well as their influence on the BOS in various populations – ranging from young adults to older adults and individuals with gait pathologies.

5.4.3 Limitations

In our study, the held arm swing condition was conducted by securing the participants' arms at their sides using a harness. This approach differs from other studies with similar conditions, where participants were either instructed to refrain from moving their arms or to cross the arms on their chest while walking. When the participants were asked not to move their arms, natural shoulder muscle activation was altered to prevent arm motion. Conversely, when the arms are crossed on the participants' chests, the COM is positioned higher and could alter the trunk behaviour. Therefore, varying arm positions may influence upper-body behaviour during gait, making it difficult to directly compare held arm swing condition results with previous literature. Second, the use of a treadmill was necessary for our study to simulate the asymmetric gait pattern. When walking on a treadmill, the belt is moving at a set speed and the participants simply need to match the belt speed to walk naturally, while overground walking demands to push against the ground to ensure the forward propulsion of the leg and the COM. Therefore, walking on a treadmill differs from overground walking and does not allow us to directly compare our asymmetry condition results with an asymmetric gait pattern observed during overground walking in individuals with gait pathologies. A third limitation is that our study was not

powered to compare the left and right leg during asymmetric gait. Although we found side-specific differences for the fast and slow legs within the asymmetry condition when comparing them to the same leg in the symmetry condition, it would be interesting to compare fast and slow leg adaptations during asymmetric gait. Finally, our study included healthy young adults with moderate-to-high physical activity levels which could lead to more efficient gait pattern adaptations compared to populations with lower physical activity levels.

5.5 Conclusion

Our results demonstrate that healthy young adults, when walking on a split-belt treadmill to simulate asymmetric gait, regulate ML dynamic stability by modifying the lateral position of the COM. We also demonstrated that during the braking phase of asymmetric gait, the hip and ankle joints are critical to maintain dynamic stability in the AP direction. Meanwhile, the knee joint primarily facilitates force redistribution towards the hip and ankle joints. Our results are in accordance with the existing literature proposing that lower-limb force redistribution is required to regulate forward progression, particularly during the propulsion phase. Our results regarding arm swing amplitude showed that actively increasing arm swing leads to more variability in the lower limb kinetics and the BOS and that arm swing restriction leads to increased compensation from the hip to assure forward progression. Overall, this study provides further knowledge on lower-limb gait control strategies employed by young adults to regulate asymmetric gait and arm swing amplitude to remain stable during challenging gait conditions. Future studies using similar protocols should consider investigating populations with gait impairments, including older adults and people with Parkinson's disease. By doing so, we can gain deeper insights into the impact of both healthy and pathological aging on gait control strategies. We suggest that these findings could potentially provide valuable insight into treadmill rehabilitation protocols for these two populations.

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Chapter 6: Conclusion

6.1 Conclusion

The findings of this thesis allow for a better understanding of how healthy young adults modify their gait patterns in the sagittal and frontal planes to maintain stability during asymmetric walking. Notably, asymmetric walking resulted in sagittal (braking impulse) and frontal plane ankle and hip joint moment impulse modifications for the slow and fast legs. Additionally, we observed an increased knee propulsive impulse in the slow leg aiding in propelling the slower limb forward and minimizing asymmetry between the two legs. Overall, this study provides further knowledge on lower-limb gait control strategies employed by young adults to regulate asymmetric gait and arm swing amplitude to remain stable during challenging gait conditions. Future studies using similar protocols should consider investigating populations with gait impairments, including older adults and people with Parkinson's disease. By doing so, we can gain deeper insights into the impact of both healthy and pathological aging on gait control strategies.

6.2 Impact

The findings of this study have significantly contributed to the knowledge of gait control strategies employed by healthy young adults when modulating arm swing amplitude in both symmetric and asymmetric walking conditions. First, healthy young adults predominantly adjust the magnitude of the ankle braking impulse (increased for the slow leg and decreased for the fast leg) to effectively regulate forward progression and ensure dynamic stability. While the hip and knee joints are also implicated in forward progression and dynamic stability regulations, they are not the primary contributors to this population. Our findings imply that the timing of the ankle, knee, and hip joint actions varies across the gait cycle. The utilization of distinct joint actions and their varied timing to control forward progression and dynamic stability suggest that young adults can quickly adapt complex

control strategies during challenging gait conditions. Thus, future studies should focus on investigating young adults with challenging working environments (e.g., construction workers) to compare if gait control strategies differ between different populations of similar age. Second, we used a similar protocol and similar sagittal plane kinetic metrics as previous studies, enabling us to establish a meaningful comparison with the existing literature. In our analysis, we also added frontal plane kinetic metrics that were not included in other studies and allowed us to deepen our knowledge of ML gait control strategies among young adults. This knowledge offers valuable insights into how young adults control dynamic stability in the frontal plane and can serve as a baseline comparison in future studies involving various populations. Last, the asymmetry ratio used in our protocol (0.8:1) is similar to the asymmetry ratio observed in people with Parkinson's disease. Additional research on the Parkinson's population could define gait control strategies used by that population. Comparing their results with the results reported in this study could potentially identify divergent adaptation strategies between the two populations. That knowledge would be useful in the development of a new gait rehabilitation protocol using a split-belt treadmill for the Parkinson's population.

6.3 Limitations

In our study, the held arm swing condition was conducted by securing the participants' arms at their sides using a harness. This approach differs from other studies with similar conditions, where participants were either instructed to refrain from moving their arms or to cross the arms on their chest while walking. When the participants were asked not to move their arms, natural shoulder muscle activation was altered to prevent arm motion. Conversely, when the arms are crossed on the participants' chests, the COM is positioned higher and could alter the trunk behaviour. Therefore, varying arm positions may influence upper-body behaviour during gait, making it difficult to directly compare held arm swing condition results with previous literature. Second, the use of a treadmill was

necessary for our study to simulate the asymmetric gait pattern. When walking on a treadmill, the belt is moving at a set speed and the participants simply need to match the belt speed to walk naturally, while overground walking demands propulsion against the ground for forward movement. Therefore, walking on a treadmill differs from overground walking and does not allow us to directly compare our asymmetry condition results with an asymmetric gait pattern observed during overground walking in individuals with gait pathologies. A third limitation is that our study was not powered to compare the left and right leg during asymmetric gait. Although we found side-specific differences for the fast and slow legs within the asymmetry condition when comparing them to the same leg in the symmetry condition, it would be interesting to compare fast and slow leg adaptations during asymmetric gait. Finally, our study included healthy young adults with moderate to high physical activity levels which could lead to more efficient gait pattern adaptations compared to populations with lower physical activity levels.

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