The Role of Arm Swing

on Dynamic Stability in People with Parkinson's Disease

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Abstract

Introduction: Idiopathic Parkinson's Disease is a multisystem neurodegenerative disease that is characterized by asymmetric impairment in regions of the midbrain, forebrain, and brainstem. Of the known neurodegenerative diseases, Parkinson's is the second most commonly diagnosed worldwide with a global prevalence expected to reach 9 million individuals by 2030. As fall rates range between 35-68% annually, falling during walking is amongst the primary concerns for this demographic. Interestingly, despite the close association between loss of arm swing (due to Parkinson's Disease) and future falls, evidence to-date has not examined the effect different arm swing conditions have on walking stability during unperturbed and perturbed (cognitive and mechanical) conditions. Dynamic stability research in this demographic is further limited in that evidence examining differences between the least and most affected leg is sparse.

Research Objectives: To examine the differences between natural arm swing (unrestricted) and when arm swing was physical restricted (restricted) in people with Parkinson's Disease. The effect of arm swing was assessed when people with Parkinson's Disease walked in steady-state, dual-task, destabilizing terrains as well as in response to slips. Additionally, this thesis examined differences between the least and most affected sides, during the aforementioned conditions, that stem from the asymmetric progression in Parkinson's Disease.

Methods: Twenty individuals with Parkinson's Disease were recruited for this research. Individuals walked on a CAREN-Extended System with unrestricted (natural) and restricted (absent) arm swing. Arm conditions were combined with steady-state walking, walking while performing a secondary dual-task, walking on minor destabilizing environments (hilly, rocky and mediolateral translational), and in response to slips for the heel-strikes of the perturbed (slipped) leg and recovery (contralateral) leg. The minor destabilizing terrains were assessed separately to

steady-state walking for the arm swing condition resulting in three types of analyses (arms-rocky, arms-rolling hills, and arms-mediolateral). Data were processed in Vicon, Visual 3D, and OpenSim before being exported to Matlab to calculate dynamic stability (Margin of Stability, Harmonic Ratios and Coefficient of Variation), average spatiotemporal parameters, as well as trunk linear and angular velocities. Statistical analyses were conducted in SPSS with a significance level set *a priori* at ($p<0.05$).

Results: During unperturbed walking with the restricted arm swing condition, compared to unrestricted, average trunk angular velocity increased in the transverse plane while instantaneous linear velocity at heel-strike decreased in the sagittal plane. Further, on the least affected leg, the Margin of Stability increased, average step length decreased, and coefficient of variation for step length increased. Contrastingly, step time coefficient of variation increased in the most affected leg. In the presence of the dual-task, average angular velocity in the frontal plane increased, average step time decreased (most affected leg), and step width coefficient of variation increased (bilaterally). Compared to unrestricted arm swing, restricted arm swing reduced average step length (arm-rolling hills) and time (arm-rocky), and increased COV step time (arm-rolling hills). The arm-rolling hills analysis revealed that the most affected leg had a shorter step length than the least affected. The destabilizing surface effects revealed that during the arm-rolling hills and arm-rocky analyses step time decreased, step width increased, and the COV for step time, length and width increased. No main effects occurred for the arm-mediolateral analysis. Additionally, when comparing the arm swing conditions in response to a slip, the restricted arm swing condition, compared to unrestricted, caused a faster step time during the slipped step. Compared to the most affected leg, the least affected had a wider step width during the slipped step. During the recovery step, the least affected leg had a larger anteroposterior Margin of Stability and longer step time

than the most affected.

Conclusion: The findings revealed that when people with Parkinson's Disease walk without arm swing, trunk rotational velocity increases which internally perturbs gait. This destabilization elicited unique responses from dynamic stability metrics that were specific to the terrain encountered. Since Parkinson's Disease primarily affects movement timing, the results suggest that loss of arm swing is particularly perturbing to foot placement timing while changes in spatial foot placement reflect compensation to maintain an existing level of global dynamic stability and symmetry. Additionally, the evidence indicates that the independent behavior of the least and most affected leg respond uniquely to loss of arm swing. However, as people with Parkinson's Disease adjust the least affected leg's foot placement to mirror the contralateral leg, functional interlimb differences may only be revealed when individuals encounter perturbations.

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Formulaic Computations:

Dynamic Stability Measures including:

1) The Margin of Stability (MOS) was computed as:[1–4]

ML-MOS = Lateral heel marker – xCOM

AP-MOS = Heel Marker - xCOM

Where the lateral heel marker was taken as the marker placed on the lateral side of the participant's calcaneus.

$$\boldsymbol{xCOM} = \boldsymbol{COMp} + \left(\frac{\boldsymbol{COMv}}{\omega\boldsymbol{\theta}}\right)$$

COM_p = COM's position, COM_v = COM's velocity.

$$\omega\boldsymbol{\theta} = \frac{\sqrt{\boldsymbol{g}}}{\boldsymbol{l}}$$

In this term, g = 9.81m/s^2 and l is the length of the inverted pendulum determined as the average distance of the right/left lateral heel marker to the COM at heel-strikes.

2) Coefficient of Variation for Spatiotemporal Parameters (step length, time, and width):

(Standard Deviation/Average) x 100

3) Harmonic Ratios (HR-AP, HR-ML and HR-Vert):[2,5–8]

Harmonic Ratios (HR) were calculated on the COM's acceleration taken as the first central difference of the COM's velocity. The HR examines a signal's periodicity by calculating an amplitude ratio of the even and odd harmonics that are obtained through a Fast Fourier Transformation.

HR-AP and HR-VT were calculated as the first 10 even harmonics divided by the first 10 odd harmonics while the HR-ML was the reverse. Higher ratios in each plane indicate that COM accelerations are more in-phase with the stepping action representing greater dynamic stability and gait symmetry.

Average Step Length, Time, and Width:

1) Step Length: Anteroposterior distance (cm) between heel markers at heel-strike
2) Step Time: Time (ms) between successive contacts of each foot
3) Step Width: Mediolateral distance (cm) between lateral heel markers at heel-strike

Trunk Linear (cm/s) and Angular Velocity (degrees/s):

1) Calculated as the first derivative of the COM's position. Average values were calculated throughout the entirety of each gait cycle. Instantaneous velocities calculated at heel-strikes for each foot.

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Chapter 1: **Introduction**

Parkinson's Disease (PD) is the second most common neurodegenerative disease worldwide, , after Alzheimer's Disease, with an increasing incidence among 60+ year old individuals.[9,10] Amongst the primary concerns for people with PD (pwPD) is falling during walking (gait).[11–13] Within this demographic, initial (first) fall rates range between 35-68% during a twelve-month period with the risk of subsequent falls increasing to approximately 70% after the first fall incident.[11–13] These numbers are alarming as falling is closely associated with several severe medical and socioeconomic consequences that negatively impact quality of life.[11,12] While the neurodegenerative processes in pwPD threaten walking stability, fall risk is further exacerbated when individuals encounter destabilizing terrains that mechanically perturb their gait (destabilizing terrains and slippery floors).[5]

Gait related falls arise due to the dynamical nature of this motor skill. Indeed, during the majority of the gait cycle, the center of mass (COM) is outside the base of support (BOS) to achieve forward progression.[14] This is challenging, though, for the neuromuscular system as stability can only be achieved by appropriately placing the swinging leg at upcoming heel-strike. At heel-strike, the COM is temporarily brought back into the BOS.[14] However, even during double-support, when the COM is inside the BOS, stability is challenged as the velocity of the COM is redirected laterally from the unloading to the loading leg.[15] During walking, foot placement in the anteroposterior (AP) and mediolateral (ML) directions is determined by the neuromuscular system predicting the future dynamical state (position and velocity) of the COM at upcoming heel-strike.[14,15] To facilitate this process, the neuromuscular system strives to maintain the COM along a consistent sinusoidal trajectory while it is volitionally displaced inside and outside the BOS (dynamic stability).[14,15] However, the gait impairments in pwPD threatens their ability to effectively fulfill

the conditions of dynamic stability. For instance, in the lower extremity, pwPD ambulate with a shorter stride length, increased stance time, increased spatiotemporal variability, and reduced interlimb coordination relative to aged matched healthy adults.[16–22] Further, in the upper body, pwPD have a reduced trunk velocity profile in the sagittal plane compared to healthy elderly adults.[23] Since the COM is located in the upper body and the lower extremity adapts foot placement to modify the BOS, each is an integral component for maintaining dynamic stability.[5]

Due to the distinct roles and movement patterns of the upper body and lower extremity in gait, as well as their unique anatomical properties (size, geometric shape, inertial properties), the neuromuscular system requires input from multiple domains to achieve dynamic stability.[5,20,24] Indeed, current evidence demonstrates that successful gait integrates information stemming from higher level cortical and subcortical structures, as well as the peripheral sensorimotor network.[5,25,26] Furthermore, Bauby and Kuo demonstrated that gait parameters in the AP direction are controlled by "passive" automated neuromuscular mechanisms while ML parameters require "active" information processing.[24] As a result of this multifaceted control, a comprehensive quantification of dynamic stability requires the use of multiple metrics since each reflects a distinct aspect of neuromuscular control.[5] Although several dynamic stability metrics have been proposed, only relatively few (spatiotemporal variability and harmonic ratios) have been examined in pwPD.[5] Moreover, the literature thus far has yet to implement multiple metrics simultaneously to garner a comprehensive depiction of fall risk in this demographic.

Additionally, this multifaceted neuromuscular control provides a means for compensation should any aspect of the network become impaired.[25] In pwPD, dopamine loss impairs subcortical pathways responsible for gait automaticity and timing.[10,21] To compensate, pwPD recruit higher

level cortical structures to bypass the impaired pathways and direct additional attention on locomotion.[25] However, in ever changing environments where multitasking is commonplace (terrain navigation, reading signs, talking etc.) attention becomes divided between multiple concurrent tasks.[16,25,27] When attention is divided, the resources necessary to compensate for impaired automaticity become strained thereby reducing dynamic stability.[16,25,27] While divided attention is demonstrated to impair lower extremity measures of dynamic stability in pwPD, its effect on additional metrics remains unexamined.

Traditionally, PD fall prevention research and rehabilitation is based on the inverted pendulum model.[28,29] This model proposes that arm swing passively arises from trunk motion, gravity, and inertia and has a negligible contribution to dynamic stability.[29] However, recent evidence demonstrates that an active arm swing component contributes to gait's contralateral arm-leg swing pattern that controls COM angular motion about the vertical axis (transverse plane).[29] Although this improves gait's metabolic efficiency (compared to no arm swing), arm swing's effect on dynamic stability is conflicting.[30–32] Theoretically, by counteracting torques arising from the swinging legs, arm swing may attenuate deviations that would otherwise arise to the COM's trajectory from the stepping leg.[29,32] Contradictory evidence though proposes that walking without arm swing improves dynamic stability by concentrating upper extremity (arms) mass around the trunk, thus causing the upper body to resemble more accurately an inverted pendulum, which increases its inertia in the transverse plane.[29,30] As one's stability level is directly related to their ability to resist perturbations, determining arm swing's effect on dynamic stability holds further implications when ambulating on minor destabilizing terrains (pathways with rocky, rolling hills, and mediolateral translations) that perturb gait.[30] Furthermore, arm movement is demonstrated to facilitate dynamic stability recovery on major destabilizing surfaces (slips) by moving the COM

upward and in the opposite direction of the angular momentum induced by the destabilization.[28,33] However, no literature to-date examines arm swing's effect on dynamic stability in pwPD on unperturbed and destabilizing environments. As pwPD walk with reduced and asymmetric arm swing, which ultimately becomes absent, quantifying arm swing's effect on dynamic stability holds direct relevance for PD fall prevention.[28,29,34–36]

1.1 Research Objectives

The objectives of this thesis are to:

1) Examine the effect of arm swing (restricted and unrestricted arm swing) during visual dual-tasking on multiple measures of dynamic stability in pwPD. A comparison between various dynamic stability measures would reflect which neuromuscular gait control (as reflected by dynamic stability metrics) mechanisms are affected by the disease and how pwPD attempt to compensate for this impairment via more intact supraspinal pathways. This objective examined how pwPD ambulate in scenarios that simulate attentionally perturbations they may encounter in everyday environments.
2) Determine the role of arm swing (restricted and unrestricted) on dynamic stability in pwPD when exposed to minor destabilizing environmental terrains (rocky, hilly and mediolateral translational). Further, this objective aimed to elucidate the differences in foot placement behavior of the least and most affected legs in pwPD when walking on mechanical perturbations. Therefore, this objective examined how pwPD ambulate in scenarios that simulate the various mechanical terrains that exist in everyday environments.
3) Quantify the effect of arm motion (restricted and unrestricted) on recovery strategies from major destabilizing environments (slips) in pwPD.

1.2 Thesis Impacts

1) An increase in average trunk angular velocity about the vertical axis was found when pwPD walk with restricted arm swing. This increased velocity about the vertical subsequently acts as an internal destabilization that perturbs pwPD's gait. However, dynamic stability metrics responded uniquely and reflecting both compensation and motor impairment.
2) Findings demonstrated that during a visual word searching dual-task, pwPD walk with, in the frontal plane, an increased trunk average angular velocity and instantaneous velocity variability. These findings indicate that dual-tasking has a strong effect on ML postural control in pwPD. Interestingly, bilateral ML foot placement adjustment occurred which allowed individuals to maintain their ML dynamic stability as no changes occurred in the Margin of Stability nor Harmonic Ratios. This suggests that therapeutic protocols should facilitate ML foot placement adaptation to retain ML stability in pwPD. Further, these results are particularly beneficial as pwPD are more susceptible to loss of stability in this direction compared to healthy aged-matched adults.[37,38]
3) Evidence was provided that as absent arm swing arises in pwPD, internal timing of foot placement is disrupted as step timing variability increases on both unperturbed and destabilizing terrains. This indicates that the internal timing of foot placement is perturbed not only by the continued neurodegeneration of the disease, but also mechanically by the increases in trunk rotational velocity in the transverse plane due to loss of arm swing. Thus, therapeutic protocols should consider restoring the contralateral arm-leg swing pattern in these individuals to facilitate rhythmic foot placement. However, careful consideration should be given to the modality of restoring arm swing since increasing arm swing amplitude is demonstrated to increase spatiotemporal variability in health young adults.[2,32]

4) Results from this thesis demonstrated that the unrestricted arm swing in pwPD was not more effective than the restricted arm swing condition to restore dynamic stability following a slip. This finding was the first to examine the effect of arm swing in restoring dynamic stability in pwPD after a major perturbation. As such, it provided a basis for following research to examine arm swing differences between these individuals and healthy elderly adults.
5) This thesis demonstrated that the asymmetric neurodegeneration in PD affects how the least and most affected legs respond to the destabilization caused by absent arm swing. However, while the independent behavior of each leg is different, larger external perturbations are necessary to elicit functional interlimb differences. This is likely due to the least affected leg adapting its foot placement to match the most affected side (to reduce gait asymmetry). As such, external perturbations are necessary in order to overcome this compensatory strategy in order to examine interlimb differences during therapies.

Chapter 2: **Literature Review**

2.1 Parkinson's Disease Epidemiology

Parkinson's Disease (PD) is the second most common neurodegenerative worldwide, after Alzheimer disease, and predominately affects individuals over the age of 60.[39–41] In Canada, between 2013-2014, PD had a 0.4% prevalence with approximately 84,000 individuals over the age of 40 diagnosed with the disease.[42] Further, the incidence of PD in Canada (2013-2014) was 55.1 per 100,000.[42] Additionally, in 2009, the incidence for the disease ranged between 108-257 per 100,000 individuals and was reported to affect 1.2 million individuals in Europe.[43] However, in the face of an aging population within developed countries, the estimates for registered PD cases are expected to increase drastically. Indeed, the American Parkinson's Association reports that

nearly 10 million individuals currently live with the disease worldwide, a value that already exceeds previous predictions of 9 million individuals by 2030.[44,45] Reports within North America (USA and Canada) showcased that in 2010 the overall prevalence of PD was 572 per 100,000 individuals aged 45 years or older.[46] Similar data have been reported by European countries such as Germany (713/100,00), the Netherlands (1400/100,000), France (328/100,000), Italy (257/100,000) and Spain (1280/100,000) have reported the substantially increasing prevalence of the disease within the last 10 years alone.[43,47–49]

As the prevalence of PD is predicted to rise, there is an increased need to address the challenges faced by these individuals. One of the primary concerns that affects pwPD and their caregivers is the increased risk of falling.[11–13,35] Within this demographic, falling during walking is a debilitating problem that leads to musculoskeletal injuries, reduced mobility, loss of independence, increased morbidity rate, poor quality of life, and increased nursing home admission.[12,43] Epidemiological evidence reports that fall rates in pwPD range between 35-68% during a 12-month period.[11,12] Further, approximately 70% of all pwPD who sustained a fall in the previous year reported suffering one or more falls in the immediately following year.[13] To assess the etiology of falls, epidemiological studies classify falls as resulting from either external (environmental) or internal (neurological) factors.[12] While the former poses a substantial threat to walking stability by creating challenges to foot placement or causing a mechanical destabilization to act on the individual, it does not account for the higher fall risk in pwPD compared to aged-matched controls.[12] To control for environmental variables, Bloem et al. compared fall rates in pwPD with healthy elderly adults of the same household.[12] The authors reported that PD participants in their study still had a nine-fold increase in sustaining recurrent falls compared to controls.[12] This evidence indicates that a strong association exists between the neurodegeneration

of PD and loss of stability while walking.

2.2 Parkinson's Disease Pathophysiology

2.2.1 The Dopaminergic System

Parkinson's disease is a progressive neurodegenerative disease that affects the function of the Basal Ganglia (BG), a circuit composed of several midbrain structures involved with movement modulation.[39,40,50] These structures consist of the Striatum (putamen and caudate nucleus), the Globus Pallidus Internus (GPi) and Externus (GPe), the Substantia Nigra pars compacta (SNc) and pars reticulata (SNr), and the Subthalamus.[39] The current working model suggests that the BG acts as a filtering loop whereby multiple motor signals are sent from the cortex to the BG, which then projects and relays the signals to the Thalamus before returning to the cortex.[39,40,50] Although the exact functions of the BG within the loop remain elusive, research indicates that the BG exerts a tonic inhibition on the Thalamus, which can either be reduced or enhanced depending on the excitation level of the BG's output.[39,40,50] Reducing thalamic inhibition facilitates the motor signal before it returns to the cortex, while enhancing thalamic inhibition impedes the signal.[39,40]

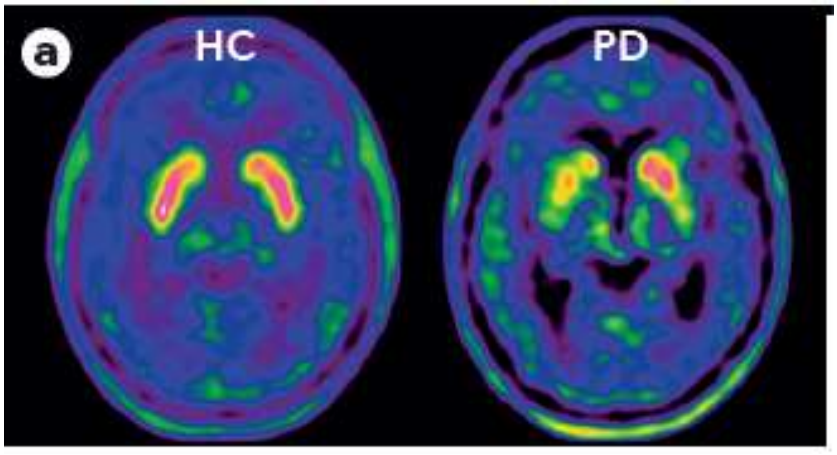

Figure 2.2.1-1 F-DOPA-PET Image of Dopaminergic System's in Healthy and Asymmetric Neurodegeneration in Parkinsonian Brain[39]

The BG's excitation level begins when cortical signals are received by the Striatum (BG's primary input structure) triggering two processes known as the Direct and Indirect Pathways **(Figure 2.2.1-2)**.[39,40] In the Direct Pathway, striatal synapses project directly with the BG's output structures (GPi and SNr) inhibiting their output to the Thalamus.[39,40,50] In the Indirect Pathway, striatal synapses project onto the GPe causing an inhibitory effect on this structure.[39,40] This in turn reduces the inhibition the GPe exerts onto the Subthalamus and the BG's output structure thereby increasing their activity.[39,40] As the Subthalamus directly synapses and excites the BG's output structures, disinhibition from the GPe, which arises from the Indirect Pathway, ultimately increases the overall BG's tonic inhibition of the Thalamus.[39,40,50] The final result, therefore, is a reduction in overall motor signals sent from the Thalamus back to the Motor Cortex.[39,40] Although classically the Direct and Indirect Pathways are both neuroanatomically and functionally separated, recent evidence indicates that both pathways are in fact interconnected and coordinate with one another in influencing the BG's overall output level.[40,50] Additionally, research indicates that a third pathway (Hyperdirect Pathway, **Figure 2.2.1-3**) exists wherein cortical projections bypass the Striatum and provide direct excitatory input to the Subthalamus.[39] The final result of the Hyperdirect Pathway is that it increases overall BG output onto the Thalamus by providing further excitation to the BG's output structures.[39] Poewe et al. suggested that in healthy individuals this may serve as a reinforcer to the Indirect Pathway whereby their activity prevents the execution of incorrect premature responses.[39] The authors further discussed that the "breaking" activity of both pathways allows for additional time for the cortex to select the appropriate motor response.[39]

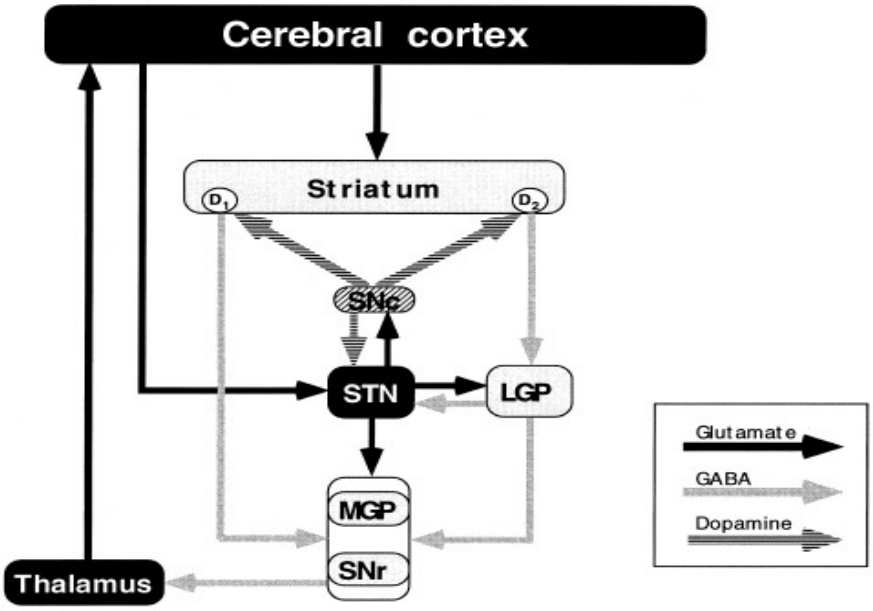

Figure 2.2.1-2: Schematic of Basal Ganglia Circuitry. Direct Pathway follows D1 receptors on Striatum projecting to Medial Globus Pallidus (MGP) and Substantia Nigra Reticulata (SNr). Indirect Pathway follows D2 receptors on Striatum projecting to Lateral Globus Pallidus (LGP) then to MGP and SNr. Hyperdirect Pathway projects from the Cerebral Cortex to the Subthamalus Nucleus (STN) then to MGP and SNr.[40]

According to the BG model, the proper and balanced functioning of these pathways is contingent upon Dopamine released from the SNc that projects onto the Striatum and Subthalamus.[39,40,50] Although the exact causes (environmental, genetic, etc.) are inconclusive, PD results in dopaminergic neuron loss that begins in the SNc and propagates further into additional structures of the BG as the disease progresses.[39] This Dopamine loss creates a cascading effect in the subsequent release of additional neurotransmitters that are vital for not only the BG's pathways but also for additional cortical structures.[39,40,50] The current working model proposes that dopaminergic loss in the SNc causes a shift in the dopamine dependent firing rate of medium spiny

neurons located in the Striatum.[39,40] These medium spiny neurons are responsible for the subsequent release of gamma-amino-butyric acid (GABA) into the direct and indirect pathways.[39,40] In the direct pathway, loss of dopamine from the SNc reduces the release of GABA from the Striatum to the GPi and SNr.[40] As GABA has an inhibitory effect on its recipient structures, the reduction of GABA causes disinhibition of the Striatum on the BG's output structures.[40] Similarly, the loss of GABA within the indirect pathway reduces the tonic inhibition the GPe exerts on the subthalamus as well as the GPi and SNr.[40] Due to the disinhibition from the GPe, there is an additional release of glutamate from the subthalamus which has an excitatory effect on the BG's output structures.[40] The culminating result of the reduced inhibition from the direct and indirect pathways on the GPi and SNr, along with the increased excitation from the subthalamus, increases the tonic inhibition the BG exerts on the Thalamus.[40]

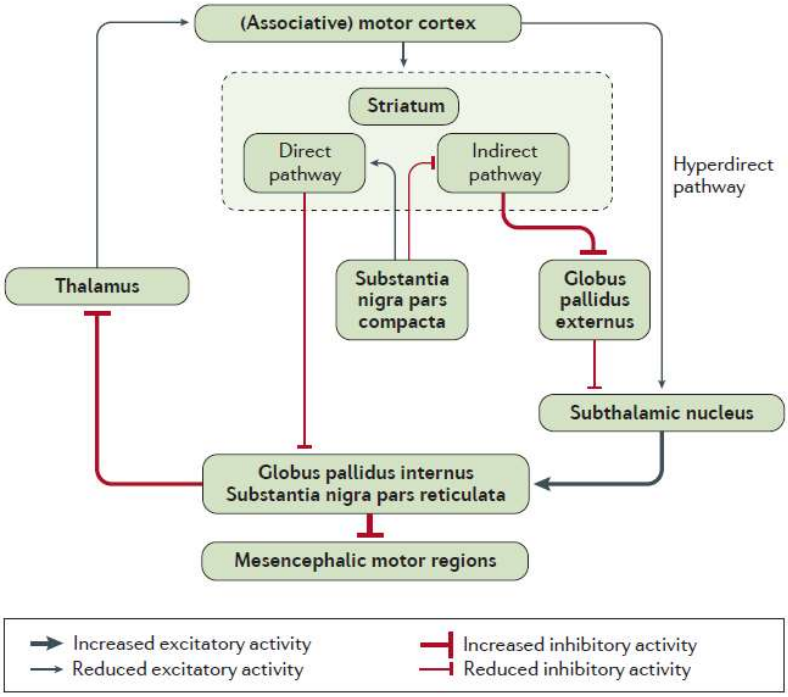

Figure 2.2.1-3: Schematic of Updated Basal Ganglia Circuitry[39]

While this classical model provides a working understanding of PD's pathophysiology, it has been acknowledged as oversimplifying the complexity of the BG and the perturbing effects of dopamine loss.[39,40] Indeed, it is important to note that structures within the BG share reciprocal synaptic projections with one another that modulates the release of dopamine, GABA, and glutamate.[39,40] This raises the possibility that the initial loss of dopaminergic neurons in the SNc may not strictly cause a linearly cascading effect as depicted by the model. Additionally, recent evidence demonstrates that neurodegeneration in PD affects additional systems other than the dopamine dependent structures of the BG.[51,52]

2.2.2 The Cholinergic System

An additional region affected by the widespread neurodegeneration of PD is the cholinergic system.[51] The cholinergic system is composed of the nucleus basalis of Meynert (nbM), the pedunculopontine nucleus (PPN), and cholinergic interneurons of the striatum, all of which provide acetylcholine to different regions within the brain.[51] Specifically, the nbM is located in the basal forebrain and supplies the cerebral cortex as well as the thalamus with cholinergic input.[51] Alternatively, the PPN is located within the caudal mesopontine tegmentum and provides acetylcholine to the thalamus, striatum, cerebellum, and brain stem while sharing reciprocal synapses with the substantia nigra, subthalamus, and GPi.[51] Finally, the cholinergic interneurons of the striatum intrinsically produce acetylcholine for local uptake within this structure.[51] Unlike the role of dopamine in the BG's pathways, the neuroanatomical function of acetylcholine remains poorly understood.[51]

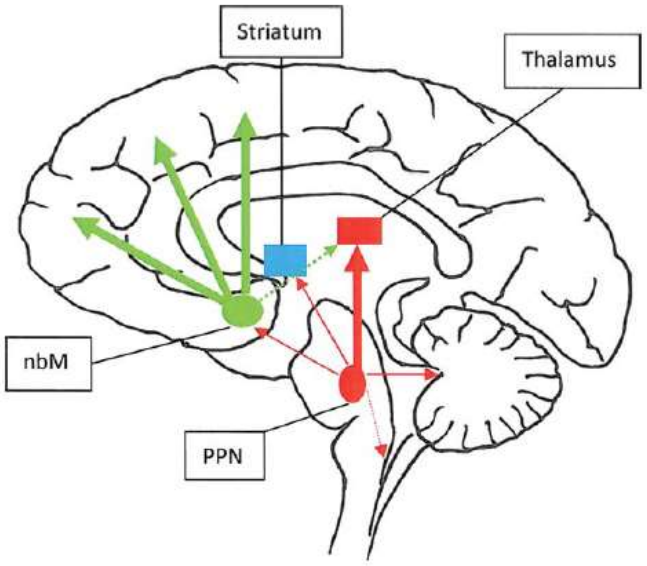

Figure 2.2.2-1: Schematic of Cholinergic System[51]

Within the cholinergic system, PD predominately causes neurodegeneration within the nbM and the PPN which is theorized to cause several non-motor (dementia, cognitive and attention impairment) and motor (posture, gait instability and freezing of gait) symptoms in pwPD.[51] While the neuroanatomical basis underlying non-motor symptoms in PD likely stems from a combination of impaired dopaminergic, cholinergic, and noradrenergic systems; the loss of acetylcholine in the nbM is closely associated with reduced attentional functioning.[51] Alternatively, cholinergic degeneration in the PPN has been implicated in gait and posture impairment since pwPD with impaired gait and balance exhibit increased degeneration in this region relative to pwPD without balance impairment.[51] Further, the PPN is suggested to be a major nuclei of the mesencephalic locomotor region; an area in the midbrain that modulates postural tone, induces stepping, and running.[53,54]

The evidence from neuroimaging, pharmacological, and deep brain stimulation studies of both the dopaminergic and cholinergic systems demonstrates that, in contrast to the traditional

view, PD is in fact a multisystem neurodegenerative disease.[39,40,51,53,54] As a result, the culmination of degeneration in these systems causes PD's cardinal symptoms (bradykinesia, rigidity, and tremor) as well as a multitude of motor impairments.[39] Amongst these impairments is the disrupted gait pattern and impaired dynamic stability commonly observed in pwPD.[5,10,39,40] Interestingly, the neurodegeneration occurs asymmetrical in the early and moderate stages of the disease, before becoming more widespread, and is a hallmark feature that characterizes idiopathic PD diagnoses from other Parkinsonian type syndromes.[55] Subsequently, the asymmetric degeneration causes motor symptoms to be expressed to a greater degree on one side of the body (most affected) than the other (least affected).[55,56]

2.3 Parkinson's Disease Sub-types and Clinical Heterogeneity

In PD, the multi-system neurodegeneration that occurs results in a heterogeneity of motor and non-motor symptoms[57–63]. The expression of different symptomatology within this demographic has raised the notion that different phenotypes of PD exist within the larger affected population. In turn, each phenotype provides indirect evidence for the different cortical and subcortical regions affected by the disease as well as acts an indicator for how widespread the neurodegeneration is. In regard to motor symptoms, pwPD are classified typically into the two main phenotypes of tremor dominance or postural instability and gait disorders (PIGD).[59,61–63] On the other hand, non-motor phenotypes include early versus late onset, benign versus malignant, and several further symptom driven phenotypes (mild cognitive impairment, loss of olfactory capabilities, anxiety, sleep disorders) that occur during the "pre-motor" phase (absence of a clear expression of motor impairment).[57,58,60] Of these non-motor symptoms, mild cognitive impairment is associated with reduced health-related quality of life, functional impairment, and an increased risk of developing Parkinson's Disease Dementia.[57,58,64,65] Mild cognitive impairment is defined

in the general population as a state of reduced cognitive function, based on performance differences on neuropsychological tests, stemming from unknown origins.[57] As mild-cognitive impairment and motor symptoms are relevant to this thesis, only these phenotypes will be discussed henceforth.

By definition, pwPD in the PIGD subtype have greater impairments to balance and gait as well as tend to express bradykinesia (slowness of movement) compared to tremor dominant individuals.[59,61–63] In contrast, tremor dominant individuals are predominately affected by essential tremor during rest and when engaging in motor activities. One of the main differences underlying the neurodegenerative processes in PIGD versus tremor dominant individuals is that the former is associated with a more widespread cortical atrophy, particularly in the frontal and pre-frontal regions.[59,61–63] Indeed, using cluster analyses on a database of 242 PD donors, Selikhova et al. found that non-tremor dominant individuals, compared to tremor dominant, had increased Lewy body scores (a marker for graded dementia) in higher level cortical regions beyond the brainstem.[61] Further, compared to the tremor dominant group, non-tremor dominant pwPD were found to have earlier signs of cognitive impairment, after disease onset, as well as more prevalent posture and gait impairments.[61] However, as this was a retrospective analysis of a dataset posthumous, differences in gait were not elaborated on.[61] In contrast, Herman et al. examined gait differences between these two phenotypes and found that PIGD individuals walked slower, had shorter strides, increased spatiotemporal variability, and impairment to their COM's sinusoidal acceleration.[63] However, the authors discussed that pwPD often have overlapping symptomology and the classifications used in their study may have limited clinical applicability.[63] Indeed, while PIGD individuals express more drastic gait impairments and have a greater risk of falling compared to tremor dominant, PD's neurodegeneration as a whole results in a multitude of gait impairments

that increases their fall risk compared to healthy elderly adults.

2.4 Parkinsonian Gait

Compared to healthy age-matched adults, pwPD ambulate with a reduced velocity, shorter stride length, increased double support percentage time, increased spatiotemporal asymmetry, and reduced bilateral coordination.[5,16] Additionally, Sofuwa et al. highlighted the reduced magnitude of kinetic profile of pwPD, compared to age matched adults, at the hip, knee and ankle joints.[66] Specifically, compared to controls, pwPD had reduced ankle power generation in the preswing phase, reduced knee power absorption in the late stance phase, and reduced hip power generation as well as absorption in the stance phase.[66] Of these impairments, this study further stated that the most notable difference to controls was the reduced ankle power generation of the PD group.[66] However, changes in these parameters are not closely associated with the increased fall likelihood in PD.[67,68] but rather directly affect the above mentioned spatial-temporal gait parameters.

Parkinsonian gait is characterized by an increase in spatiotemporal variability that is not readily discernable through observation alone, during the disease's initial stages, but becomes more apparent as the disease progresses.[16,21,69] This increased variability has been attributed to the difficulty of pwPD to independently generate rhythmic motor outputs that arise from the BG's disrupted internal cueing.[21] The increased variability in gait's spatiotemporal parameters (stride length and time) has been defined as a marker for prospectively identifying fall risk in pwPD and is considered a hallmark feature of parkinsonian gait. [10,19,70]

In addition to increased variability, dynamic stability in PD is further threatened due to the paroxysmal phenomenon known as Freezing of Gait (FOG).[71,72] FOG is defined as transient episodes when individuals are unable to effectively engage in the stepping process and is

commonly reported as the perception of one's feet being "glued to the ground".[70,73,74] Although the underlying pathophysiology is still unknown, FOG has been linked to impaired visuospatial perception, cognitive loading, and difficulty during gait initiation, turning and steady-state walking.[19,71–76] Furthermore, freezers (pwPD with FOG) exhibit increased spatiotemporal asymmetry and variability, as well as reduced interlimb coordination when compared to non-freezers, even during optimally medicated states.[69,70,77,78] In a literature review, Bloem et al. suggested that FOG may be due to a misfiring of oscillators that cause the legs to move too rapidly for effective stepping.[69] During an examination of vertical ground reaction force data in a stepping in-place task, Nantel et al. demonstrated that freezing episodes were characterized by force profiles that were rapid, irregular and reduced in amplitude.[77] As a result, freezers were unable to effectively perform the stepping action.[77] Although freezers are reported to have a higher likelihood of falling, the altered gait pattern stemming from PD increases the fall risk in both freezers and non-freezers.[12,13,53,54,69] However, dynamic stability research in PD is limited as current evidence has only quantified a select few dynamic stability metrics.

2.5 Neuromuscular Control of Gait and Dual-Tasking in PD

Traditionally, steady-state gait has been classified as an automated motor skill that requires limited higher-level cognitive resources. However, in their seminal article, Bauby and Kuo demonstrated that gait is actually a multifaceted motor skill that requires the integration of both passive and active neuromuscular mechanisms for effective stepping.[24] Current evidence further substantiates the finding that the neuromuscular control of gait arises from both subcortical and supraspinal structures that exert, respectively, automatic and conscious attentional control during walking.[5,25] Specifically, parameters occurring in the AP direction, along with step timing, are suggested to be automated by subcortical structures while parameters in the ML direction require

supraspinal processing for effective execution.[5,24,25] Such a complex neuromuscular network is necessary not only due to the upper and lower extremities having distinct functional roles in each movement plane during walking, but also to provide a means for compensation should any aspect of the network become impaired.[5,25] In PD, the progressive dopamine loss in the BG impedes subcortical networks responsible for gait automaticity and timing.[25,69,70] Therefore, pwPD attempt to compensate by recruiting higher-level supraspinal structures to bypass the impaired subcortical network and direct attention to consciously control otherwise automated parameters.[25,27,79] To examine the extent to which attention regulates gait in PD, dual-task paradigms are implemented wherein participants are required to walk while simultaneously performing an attentionally demanding secondary mental task.[25] The rationale behind these paradigms is that if gait parameters become further impaired during the dual-task then supraspinal structures were required to direct attentional resources for their execution.[25]

Current evidence demonstrates that compared to walking alone (as a single-task), dual-task walking leads to a reduction in their gait velocity, stride length, and lower extremity coordination with an increase in their double and single support time, spatiotemporal variability (stride length, time and swing time), lower extremity and arm swing asymmetry in pwPD.[16,17,19,20,22,25] Although multiple theories such as the capacity-sharing, bottleneck and multiple resources have been proposed to explain dual-tasking effects, these theories do not fully account for the diversity of results in dual-tasking paradigms.[25] Alternatively, previous research proposed that task prioritization may be a more holistic explanation for the multitude of findings.[16,25,79] Indeed, compared to healthy age-matched adults, pwPD have been reported to naturally prioritize performance on the secondary cognitive task over gait performance.[25] Hausdorff et al. suggested that the inability for pwPD to correctly prioritize tasks may be associated with more widespread

neurodegeneration in supraspinal structures.[16] This is particularly concerning as gait occurs in continuously changing environments where dual-tasking occurs on a daily basis.[80] This holds direct implications for fall risk in pwPD, as they would prioritize performance of the secondary mental task over their dynamic stability. However, the current evidence is limited in that only spatiotemporal variability has been examined as a measure of dynamic stability during dual-tasking.[27,79] It is critical to examine additional dynamic stability measures during dual-tasking in PD to determine the degree each one is affected by the disease's and the extent to which pwPD compensate for its regulation through attentional mechanisms. This is particularly crucial as a more widespread neurodegeneration affecting supraspinal structures would affect pwPD's ability to circumvent impaired automatic networks and may only be reflected by specific dynamic stability measures.[5]

2.6 Quantification of Dynamic Stability

In classical biomechanics, when the COM is further separated from the BOS an individual is considered more unstable.[81,82] However, Pai and Patton demonstrated that when the COM is outside the BOS, balance conditions can be achieved if the COM's velocity is directed toward the BOS, such as during the single-support phase, when the COM is transferred medially and anteriorly for heel-strike preparation.[81] Therefore, previous research proposed several alternatives to both clinical tests and the COM-BOS separation method to quantify dynamic stability.[5,10] However, walking is a complex skill that requires multiple aspects of neuromuscular control for successful ambulation.[5,25] In a systematic literature review, Siragy and Nantel demonstrated that each dynamic stability method reflects a specific aspect of neuromuscular control.[5] As such, the authors concluded that implementing a single quantification method is insufficient to reflect walking stability in its entirety.[5] Measures such as the Coefficient of Variation (CoV), Harmonic

Ratios (HR) and the Margin of Stability (MOS) are common dynamic stability metrics used in steady-state and perturbation (slips and destabilizing terrains) paradigms.[5]

2.6.1 Spatiotemporal Variability

Spatiotemporal variability is an overarching method that defines stability in the lower extremity by quantifying the magnitude of the variability in step time, length and width (termed the Coefficient of Variation).[5] However, previous research indicates that the variability in each of these parameters reflects distinct motor control aspects for lower extremity stability.[5,24,68,83] For instance, Hausdorff et al. suggested that step time variability reflects the internal timing mechanisms that automatically control walking's temporal consistency.[68,83] Current evidence further demonstrates that step length and width reflect their own unique motor control processes.[5,24,84] Bauby and Kuo demonstrated that lower extremity anteroposterior placement is governed by automatic passive mechanisms while mediolateral placement is actively determined.[24] As the neuromuscular system determines foot placement based on its ability to predict the COM's future position and velocity, large variability in the spatiotemporal parameters of pwPD indicates the neuromuscular system's inability to consistently predict and control erratic COM movement.[5,14,85] Indeed, current evidence demonstrates that pwPD walk with increased step time and length variability that is attributed to disrupted BG control of anteroposterior foot placement and timing.[5,21,70] Furthermore, Rochester et al. demonstrated that de novo pwPD have an increased step width variability than healthy age matched adults suggesting that the disease's neurodegeneration affects the proper functioning of higher level cortical centers actively selecting appropriate mediolateral foot placement.[86]

However, despite variability being a strong predictor of fall risk in pwPD, some amount of variability is necessary to facilitate appropriate environmental adaptation.[5,87] In pwPD, the ability

to adapt to the environment may be partially preserved, in the least affected hemisphere, due to the neurodegeneration in the BG occurring asymmetrically.[56,88,89] This asymmetric neurodegeneration subsequently impairs mobility on one side of the body to a greater extent (most affected) than the other (least affected).[89,90] Although still emerging, initial evidence demonstrates that pwPD are capable of adapting their least affected leg to compensate for mobility impairment on the most affected side.[56,87] Indeed, Cantu et al. found increased variability in muscle activity of the least affected leg compared to the most affected leg during a stepping in place task.[87] Furthermore, in a pilot study, Ricciardi et al. demonstrated that physical therapy interventions were more effective when they targeted the less affected leg.[56] Therefore, it is plausible that a threshold for variability exists in pwPD whereby values within a certain range indicate adaptation while values outside the range indicate fall risk.[5,87] Additional research is required though to further elucidate the role of variability in gait for pwPD.

2.6.2 Harmonic Ratios

Deriving from harmonic theory, Harmonic Ratios (HR) define dynamic stability in terms of an acceleration signal's periodicity.[5–8,91–93] This method is typically applied to trunk acceleration data as controlling the trunk's COM is critical to avoid falling.[92] Harmonic Ratios examines an acceleration signal's periodicity by calculating a ratio of the amplitudes of the even and odd harmonics.[5–8] In the anteroposterior and vertical directions, the acceleration signals are considered stable if their Harmonic Ratio repeat in even-numbered multiples.[5–7] This is due to the assumption that the trunk's acceleration period is biphasic (gait cycle consisting of two steps) thereby repeating in even multiples to be "in phase" with stepping.[5–7] However, if the Harmonic Ratios repeat in odd-numbered multiples then they are considered unstable as the signal's biphasic nature is "out of phase" with stepping.[5–7] Unlike the anteroposterior and vertical directions, trunk acceleration in

the mediolateral direction is monophasic and considered stable if the HR repeats in odd-numbered multiples.[5–8] The discrepancy is due to the trunk shifting only once, in the mediolateral direction, during weight transfer in the double support phase.[5]

Theoretically if the Harmonic Ratios are "in phase", reflected by larger HR values, then it indicates that the trunk progresses between stance limbs along a consistent sinusoidal trajectory.[6–8,91,92] According to previous research, AP trunk stability is controlled by automatic passive mechanisms as the COM's forward and downward momentum initiates stepping with limited guiding input from subcortical structures and the somatosensory system to catch the falling COM.[14,24,85] However, ML trunk stability is suggested to be predominately controlled by active supraspinal input that determines lateral lower limb placement.[5,24] Thus, HRs that are "out of phase" would reflect an impairment in either the active or passive mechanisms controlling the COM. As PD causes neurodegeneration in supraspinal and subcortical structures, individuals have compromised HRs compared to healthy adults.[6,94] However, current evidence is conflicting in which plane pwPD have impaired HR.[5,6,94] As such, future research should examine HRs in PD to determine in which plane pwPD are impaired in controlling the trajectory of their COM.

2.6.3 Margin of Stability

The traditional method for assessing dynamic stability (the distance between the COM-COP) is limited in that it does not take into account that balance can be achieved if the COM's velocity is directed toward the BOS even if the COM is located outside of it.[82,95] Therefore, to overcome this limitation, Hof proposed the Extrapolated Center of Mass (**Figure 2.7.3-1)** concept (xCOM) to quantify the COM's position and velocity as a single metric.[82,95] Stability is then determined by taking the distance between the xCOM and the edge of the BOS, which is then termed the Margin of Stability (MOS).[82,95] Hof suggested that an individual is more stable when

the MOS' value is greater as the xCOM is further from the BOS' boundaries and that negative values reflect stability loss since the xCOM lies outside the BOS.[82] The author further discussed the interpretation that the term is equal to the impulse required to disturb balance which either initiates stepping or results in a fall.[82]

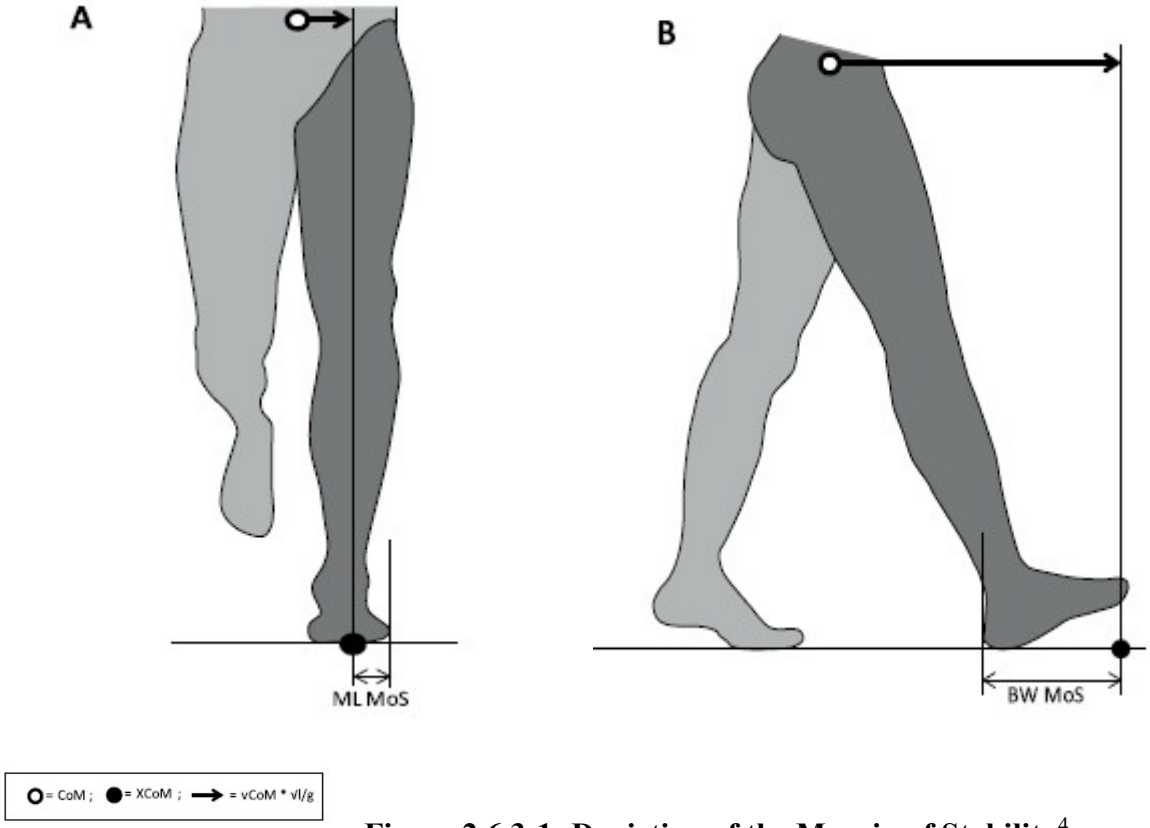

Figure 2.6.3-1: Depiction of the Margin of Stability[4]

Although the xCOM-BOS distance is an intuitive concept, experimental evidence appears to only support the MOS' current stability definition in the ML direction during steady-state walking.[5,96,97] Indeed, previous research demonstrates that during steady-state walking, individuals adapt their step width to their current environment to maintain their xCOM inside the mediolateral boundaries of their BOS.[98] However, during steady-state walking, MOS quantified in the anteroposterior direction consistently demonstrates that the xCOM lies outside the anteroposterior

boundaries of their BOS, resulting in a negative value.[96,97,99,100] Based on Hof's definition, this would indicate that individuals are constantly in a state of instability.[82] Furthermore, the evidence creates an additional paradox in the anteroposterior direction as elderly adults (fallers and non-fallers) are reported to have a shorter distance, a negative value with reduced magnitude, compared to healthy young adults.[5] This would suggest that these individuals have greater dynamic stability despite this demographic having a heightened fall risk during walking.[5] Therefore, Lugade et al. proposed an alternative interpretation whereby a MOS with an increased magnitude, albeit still negative, in the anteroposterior direction is indicative of enhanced dynamic stability as individuals are able to handle more dynamical states of the COM away from their BOS.[97] This interpretation may be substantiated by considering the differences between young and elderly adults on two aspects. Firstly, the increased anteroposterior MOS in young adults may be representative of the higher gait velocity compared to elderly adults. Thereby implying that the MOS in this direction is a plausible indicator of an individual's forward progression.[95] Secondly, an anteroposterior MOS that is closer to zero, which would represent that the xCOM is outside the BOS but at a reduced distance, may be representative of a perceived threat to dynamic stability.[5] During perturbed walking, both young and elderly adults adopt a "cautious gait" strategy and alter trunk kinematics in response to potentially arising mechanical perturbations.[101–104] The outcomes of this strategy are that individuals shift the COM's position in the opposite direction of where they perceive the effect of the mechanical perturbation while simultaneously reducing its velocity.[5] Siragy and Nantel suggested that these are adaptive responses to potential environmental perturbations that can facilitate kinematic recovery responses at perturbation onset and reduce the distance the xCOM has to overcome to return to the inside of the BOS.[5] Thus, the reduced anteroposterior MOS in elderly adults may be additionally representative of their commonly observed "cautious gait"

during steady-state walking.[67,105]

Interestingly, there is currently a lack of evidence examining the MOS in pwPD despite the high fall risk in this demographic.[5] Examining the MOS in pwPD during steady-state and perturbed walking would provide critical information regarding perceived stability threats in this demographic as well as their ability to control their xCOM. This is particularly beneficial in the ML direction as stability loss in this direction is closely associated with fall risk.[15,106] As current PD dynamic stability evidence is relatively sparse regarding how the disease affects stability in this direction, quantifying the MOS would provide insight into the degree active information processing is affected. Finally, the MOS is a common dynamic stability metric when examining the effect and adaptations to mechanical perturbations.[5] Thus, its use in perturbation paradigms for pwPD would provide essential insight into the neuromuscular control of the xCOM in various environments in this demographic.

2.7 Arm Swing's Role in Dynamic Balance Stability

Traditionally, the human body is modeled as an inverted pendulum with the upper body (head, arms, and trunk) accounting for two-thirds an individual's mass.[14,15,85] This model is founded on the notion that the head, arms and trunk are a single nearly rigid body with arm swing arising only as a passive result of trunk movement, gravity, and inertia.[29] From this model researchers developed numerous equilibrium equations that became the foundation for current dynamic stability research.[14,15,15,33,107]

However, recent evidence demonstrates that shoulder joint muscular activity and moments contribute to arm swing during walking that cannot be accounted for by gravity and joint accelerations alone.[28,29,108,109] During walking, a 1:1 contralateral arm-leg swing pattern occurs

during preferred gait speeds.[29] In a literature review, Meyns et al. reported that when the arms swing freely, there is a reduction in metabolic energy costs by up to 8% during walking relative to walking without arm swing.[29] This is due to the swinging arm creating a torque that directly opposes the angular moment of both the swinging limb and rotating trunk within the transverse plane.[29] These mechanisms, in turn, regulate COM angular momentum in the transverse plane from becoming excessive thereby reducing energy costs.[29] However, regulating COM angular momentum in the horizontal plane theoretically has additional direct implications for dynamic stability. As the COM is located at the trunk level, reducing trunk rotation within this plane may contribute to maintaining the COM along its nominal sinusoidal trajectory.[2] Arm swing may further enhance dynamic stability as the contralateral arm-leg coupling would regulate COM angular momentum in the transverse plane thereby reducing torques acting about the stance limb.[110] Additionally, prior research demonstrated a strong neural coupling between the arms and legs during gait, which was suggested to facilitate rhythmic leg motion and help regulate lower extremity variability.[28,33,111–114]

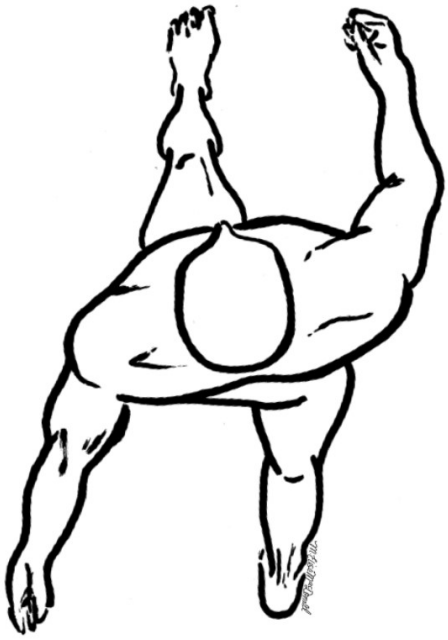

Figure 2.7-1 Contralateral Arm-Leg Swing (original fig. Mary-Elise MacDonald)

While this theoretical framework suggests a beneficial role of arm swing in enhancing dynamic stability, experimental results remain inconclusive.[30,115] Indeed, Wu et al. examined differences between active arm swing (arm swing with increased amplitude) and natural arm swing and found that active arm swing enhanced the trunk's mediolateral local dynamic stability – measured as the logarithmic rate of divergence in a system via the Lyapunov Exponent in the mediolateral direction -- but had no additional effects on dynamic stability in the AP or VT directions.[31] Similarly, Hu et al. examined differences between restricted, natural and active arm swing on local dynamic stability and found that active arm swing, compared to the other two conditions, improved stability in all three cardinal directions.[116] Contrastingly, however, Brujin et al. found no differences on dynamic stability during steady-state walking between absent and normal arm swing conditions.[30] The authors suggested that maintaining the arms closer to the trunk concentrates upper extremity mass into a single body, which increases upper body inertia thereby

improving trunk resistance to changes in motion.[30]

Interestingly, Hill and Nantel demonstrated conflicting, yet reconcilable, effects of arm swing on dynamic stability.[32] The authors examined the effects of active (increased arm swing amplitude), normal, and absent arm swing on dynamic stability and coordination in healthy young adults.[32] In regard to local dynamic stability of the trunk, active arm swing increased stability compared to normal and absent arm swing, however, a simultaneous increase in spatiotemporal variability occurred for the same arm swing comparisons.[32] These directionality changes may be reconciled by the fundamental differences in the anatomical and neuromuscular systems they are classifying.[5,32] Further, the authors discussed the possibility that improvements in trunk local dynamic stability arose either from an increased ability of the arms to regulate trunk angular momentum or that the increased variability reflected foot placement adjustment which has a subsequent stabilizing effect on the trunk.[32] Interestingly, Siragy et al. reported that trunk velocity values were more variable and HR values decreased when healthy young adults walked with active arm swing, compared to normal and absent.[2] The authors also found an accompanying increase in spatiotemporal variability (active arm swing compared to normal and absent) which they proposed reflected foot placement adjustment that acted to maintain an already existing level of global dynamic stability (reflected by no changes in the ML MOS) to avert a fall.[2] Examined together, the findings by Hill and Nantel as well as Siragy et al., raise the notion that dynamic stability is retained when arm swing increases through effective foot placement adjustment rather than as an attenuation of angular momentum via increased arm swing amplitude.[2,32]

Indeed, the increased arm swing amplitude causes the COM to move more variably and along a less sinusoidal trajectory (reduced HR) compared to natural and absent arm swing. Thereby reflecting reduced postural control of the upper body. However, healthy young adults appear to

compensate for this perturbed posture by adjusting foot placement to attenuate the rate (local dynamic stability) at which the trunk's variability causes unintended trajectories of the COM to occur.[2,32] Further, the compensatory foot placement adjustment allowed individuals to maintain their global dynamic stability by keeping their xCOM at a safe distance from the edges of their BOS.[2] Further support for the effective contribution of foot placement over increased arm swing stems from the reduced 1:1 contralateral arm-leg coordination found in both studies.[2,32] Due to this reduced coordination, the effective counterbalancing of torques between the contralateral arm and leg becomes impaired which would cause excessive movement at the trunk.[2,32] As such, in these protocols the main mechanism of the arms in controlling COM angular momentum was rendered more ineffective despite the increased arm swing amplitude. However, as a result of the breakdown of arm-leg coordination, it remains inconclusive if increasing arm swing while maintaining the 1:1 arm-leg swing ratio holds additional benefits over natural and absent arm swing for dynamic stability. What remains from the current findings though is that actively increasing arm swing increases spatiotemporal variability.[2,32] While this plausibly reflects foot placement adjustment in young adults, increased spatiotemporal variability is a strong predictor of falls in elderly adults and pwPD.[5,67,68] Thus, potential improvements stemming from increasing arm swing amplitudes may be outweighed by the increased risk of falling in elderly individuals and those with PD.

Despite evidence from healthy young adults indicating no differences in dynamic stability and postural control when walking with normal and absent arm swing, no evidence exists examining these conditions in pwPD. In PD, one of the main motor symptoms that is expressed relatively early after diagnosis is a reduced and asymmetric arm swing which becomes completely absent in the disease's later stages.[29,35,36,117] Further, loss of arm swing (either unilaterally or bilaterally) is demonstrated to be a strong predictor for future falls in pwPD.[35] As such, the strong

association between loss of arm swing and future falls indicate that arm swing may play a functional role in the dynamic stability of these individuals.[35,117] To this end, it may suggest that lack of differences between absent and normal arm swing (in healthy young adults) does not occur from an effective increase in upper body inertia (during absent arm swing), rather, adaptation in either trunk musculature or foot placement act as compensatory mechanisms.[2,30,32] However, in pwPD, the disease's neurodegeneration impairs their ability to adapt, thus, impeding mechanisms necessary for compensating for absent arm swing to maintain dynamic stability.[38,38,118,119] Due to the limitations in the current evidence, there is a need to further elucidate on the potential influence arm swing has on dynamic stability in pwPD. As current fall prevention and rehabilitation paradigms in pwPD are based on research that utilizes the inverted pendulum model there is a need to examine the potential contribution arm swing has on dynamic stability in this demographic.[28,29] Therefore, examining the effect of absent and normal arm swing in pwPD has direct implications for advancing fall prevention protocols.

2.8 Gait Adaptations on Destabilizing Terrains

When examining the effect of arm swing on dynamic stability, it is important to consider walking conditions beyond that of steady-state or unperturbed gait. The underlying rationale is that unperturbed steady-state walking is not entirely representative of the different environmental terrains that individuals encounter in their everyday lives.[80] Therefore, to reduce fall risk, it is critical to understand how pwPD control dynamic stability on various destabilizing terrains and to elucidate whether arm swing plays a functional role in these conditions. Recently, advancements in virtual reality and instrumented platform technology has provided a safe means to study the effect of destabilizing surfaces on gait.[1,2,32,120,121] Within this paradigm, destabilizing surfaces such as rocky, rolling hills, and mediolateral translations are common conditions used to simulate real-

world surfaces that perturb one's gait.[122,123] Indeed, rocky surfaces simulate walking on rocky surfaces, rolling hills emulate a forest trial, and the mediolateral translational simulates walking on a train or a bus.[122]

Evidence thus far demonstrates that healthy young adults and individuals with lower limb amputations adapt their gait pattern in a manner that is specific to the terrain encountered.[1,120,122,123] These terrain specific responses are a result of the distinct destabilizing effects these terrains have on the COM's trajectory.[122–124] Indeed, McAndrew et al. demonstrated that decreases in trunk stability corresponded to the acting direction of the perturbation.[124] For instance, reductions in AP trunk stability were only elicited during terrains that destabilized individuals in the AP direction, with mirroring results for ML parameters.[124] Similarly, in an earlier study, McAndrew discussed that in order to challenge stability, terrain destabilizations should be applied in the direction in which change is desired such as during therapeutic interventions.[120] Interestingly, Sturk et al. found no changes in the ML-MOS (mediolateral MOS) in healthy young adults when they compared unperturbed walking to rocky, rolling hills, and mediolateral translational surfaces.[122] One possible explanation for the lack of differences is that healthy adults adapt their ML foot placement to maintain an already existing level of global dynamic stability.[95,96,98] However, evidence from static postural perturbations indicates that the neurodegeneration in pwPD impairs their ability to appropriately adapt to perturbations.[26,38,118,119] Thus, appropriate gait adaptation to destabilizing terrains may equally be impaired in these individuals. However, the literature to-date has yet to examine how pwPD ambulate on surfaces that emulate real-world conditions. Further, as arm swing affects COM movement, elucidating how loss of arm swing impacts dynamic stability while on destabilizing surfaces is paramount for PD fall risk assessment.[2,29,32,110]

2.9 Arm Swing's role in restoring Dynamic Stability after Slips

Current evidence on mechanical perturbations demonstrate that the neuromuscular system employs active recovery strategies to return the perturbed COM to a state of dynamic stability.[5,28,33] Initial evidence demonstrates that arm swing may be part of a larger strategy to recover stability as rapid bilateral bursts in the deltoid muscles have been reported in response to mechanical perturbations.[28,33,115] This burst of deltoid muscular activity has been observed to coincide with an arm elevation strategy that participants employed after slip onset.[28,115] Researchers suggested that this strategy serves to dissipate backward momentum induced by a slip by moving the COM in the opposite forward direction.[28,33]

However, current arm swing research have only been able to partially quantify the role of arm swing in dynamic balance recovery strategies. This is due to the majority of research either examining solely upper body muscular activity, limitations in arm swing conditions, and absence of a dynamic stability measures. As such, it remains inconclusive how arm swing contributes to recovery of perturbed dynamic stability. Indeed, Marigold and Misiaszek stated that no conclusive evidence exists to indicate if arm movement after a perturbation is part of a coordinated dynamic stability recovery strategy or if arm movement is evoked simply by the mechanical linkage of the musculoskeletal system.[28] Furthermore, there is currently a lack of evidence examining arm swing's role to recover dynamic stability after a slip in pwPD. As static postural perturbation demonstrates that pwPD are more susceptible to falls in the backward direction (the same destabilizing direction of slips), examining arm swing's effect on slip recovery holds strong potential for PD fall prevention interventions.[125] Additionally, evidence to-date has yet to examine the effect of slips in general on the dynamic stability of pwPD.

Chapter 3: **Methodology**

3.1 Research Design

This research was conducted within a larger study "*Early Detection of Postural Impairments and Gait Disturbances phenotype in individuals with Parkinson's disease: new strategies to detect and slow the progression of PIGD"* funded by the Ministry of Economic Development and Innovation (ON, Canada). A within-subjects repeated measures design was used to examine the effect of arm swing and dual-tasking during steady-state walking. Further, this study examined the effect of arm swing as well as differences between the least and most affected leg in pwPD during minor and major destabilizing terrains.

3.2 Independent Variables:

Research Objective 1 Dual-tasking and Arm Swing in Steady-State Walking (2x2):

Participants walked with and without a word-searching task to examine dual-tasking effects. These conditions were combined with two types of arm swing (1) unrestricted and (2) restricted. Therefore, Research Objective 1 consists of a single 2x2 analysis.

Research Objective 2 Arm Swing and Minor Perturbations (2x2x2):

Two arm swing conditions were used to quantify arm swing's effect to maintain and recover dynamic stability during minor mechanical perturbations. Further, differences between the least (1) and most affected (2) legs were examined. Arm swing conditions included: (1) unrestricted and (2) restricted. Steady-state walking (1) was compared separately to each of the following (2) minor environmental terrains: rocky, rolling-hills, and mediolateral translations. As such, Research Objective 2 consisted of three separate 2x2x2 analyses (arm x terrain x leg).

Research Objective 3 Arm Swing and Major Perturbations (Slips) (2x2):

Research Objective 3 included two separate 2x2 analyses. In the first analysis, the effect of arm swing (restricted and unrestricted) and leg (least and most affected) was examined on the dynamic stability of the slipped leg. The slipped leg was defined as the leg which was perturbed upon heel-strike. The second analysis examined the same arm swing and leg conditions on the recovery leg after a slip. The recovery leg was defined as the limb contralateral to the slipped leg that performed the first heel-strike following the slip.

3.3 Participants

A convenience sample of twenty (13 males and 7 females) people with Parkinson's Disease, aged between 48-79 years old (average age 63.8 ± 9.0 years, mass 72.3kg ± 19.4, height 172.8cm ± 8.0) were recruited from the Ottawa-Gatineau community. All participants had received a diagnosis by their neurologist for idiopathic Parkinson's Disease prior to recruitment in this study. Since two individuals had severe dyskinesia and one individual had missing data, only 17 participants (11 males and 6 females, 64.8±7.4 years) were included in the final analysis. Within the original 20 participants sample, one fall occurred during data collection. However, this participant was not included in the final analyses as this individual had missing data. The remaining 17 participants had an average disease duration of 8.0 ± 5.1 years (post-diagnosis). Participants motor symptom severity was assessed with the original Unified Parkinson's Disease Rating Scale Motor Examination[1] (11.0 ± 6.0) and were between I-III on the Hoehn & Yahr scale.[132,133] Further, seven participants reported experiencing freezing of gait based on the original Freezing of Gait

[1] https://www.movementdisorders.org/MDS/MDS-Rating-Scales/MDS-Unified-Parkinsons-Disease-Rating-Scale-MDS-UPDRS.htm)

Questionnaire **(Appendix A)**.[134] Individuals with freezing of gait were not excluded as these participants had a comparable UPDRS and H&Y rating as non-freezers. Further, no freezing episodes occurred during data collection. Participants were requested to take their prescribed medication thirty minutes prior to arrival for data collection in order to be tested on their optimally medicated state.

Exclusion criteria included any physical discomfort using a virtual reality system, any injuries and/or orthopedic surgeries that interfered with gait, walking only with a walking aid, and additional illnesses other than Parkinson's Disease. Further, participants were verbally screened for Parkinsonian dementia and cognitive impairment. All participants provided written informed consent and the study was approved by the Ottawa Health Science Network Research Ethics Board and the University of Ottawa Research Ethics Board. The study was conducted in accordance with the Tri-Council Policy statement; Ethical Conduct for Research Involving Humans; The International Conference on Harmonization- Good Clinical practice: Consolidated Guideline; and the provisions of the Personal Health Information Protection Act 2004.

3.4 Data Collection

All data collection per participant for this thesis occurred during a single session. At the beginning of the session, participants were assessed with Unified Parkinson's Disease Rating Scale III (UPDRS-III) and Hoehn and Yahr (H&Y) to determine stage of disease progression and motor impairment. Further, to determine symptom laterality, the UPDRS-III scores along with participants' self-reporting of symptom laterality were used. The UPDRS-III as well as H&Y scales were conducted by a single researcher (Julie Nantel). Participants wore a fitted full-body

motion capture suit and shoes they were comfortable to exercise in. A single researcher collected anthropometric measures of height and weight and placed reflective markers on the participants' bony landmarks using a 57 full-body marker set as depicted in **Figure 3.4-1**.[135] After marker setup, participants were given 5 minutes as a familiarization period to get accustomed to treadmill walking. No motion capture data were collected during this time. At the end of the familiarization period, the final preferred walking speed for participants was recorded in order to set all subsequent trials (where motion capture data were collected) to this speed. Participants wore a safety harness at all times during gait trials.

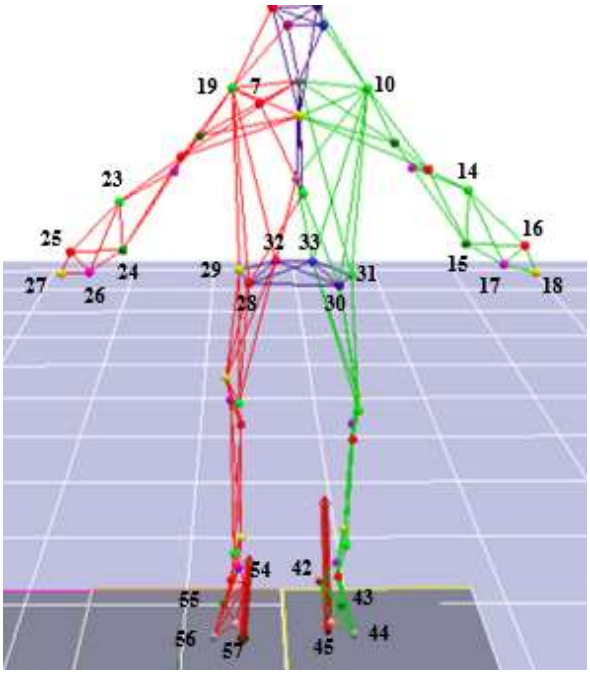

Figure 3.4-1: 57 Full-Body Marker Set[135]

	Marker Label	Description		Marker Label	Description
1			34	LTH1	Left Thigh Plate (anterior, superior)
2			35	LTH2	Left Thigh Plate (anterior, inferior)
3			36	LTH3	Left Thigh Plate (posterior, inferior)
4			37	LTH4	Left Thigh Plate (posterior, superior)
5	C7	7th Cervical Vertebra	38	LSK1	Left Shank Plate (anterior, superior)
6	T8	8th Thoracic Vertebra	39	LSK2	Left Shank Plate (anterior, inferior)
7			40	LSK3	Left Shank Plate (posterior, inferior)
8	STRN	Sternal Notch	41	LSK4	Left Shank Plate (posterior, superior)
9	XYPH	Xyphoid Process	42	LHEE	Left heel
10			43	LLHL	Left lateral heel
11			44	L5MT	Left 5th Metatarsal Head
12			45	LTOE	Left 2nd Metatarsal Head
13			46	RTH1	Right Thigh Plate (anterior, superior)
14			47	RTH2	Right Thigh Plate (anterior, inferior)
15			48	RTH3	Right Thigh Plate (posterior, inferior)
16			49	RTH4	Right Thigh Plate (posterior, superior)
17			50	RSK1	Right Shank Plate (anterior, superior)
18			51	RSK2	Right Shank Plate (anterior, inferior)
19			52	RSK3	Right Shank Plate (posterior, inferior)
20			53	RSK4	Right Shank Plate (posterior, superior)
21			54	RHEE	Right heel
22			55	RLHL	Right lateral heel
23			56	R5MT	Right 5th Metatarsal Head
24			57	RTOE	Right 2nd Metatarsal Head
25					
26					
27					
28	RASI	Right Anterior Superior Iliac Spine			
29	RASI2	Right Iliac Crest			
30	LASI	Left Anterior Superior Iliac Spine			
31	LASI2	Left Iliac Crest			
32	RPSI	Right Posterior Superior Iliac Spine			
33	LPSI	Left Posterior Superior Iliac Spine			

Table 3.4-1: Description of 57 Full Body Marker Set

To examine the effect of arm swing on people with Parkinson's Disease two arm swing conditions were used for all Research Objectives. An "unrestricted arm swing" condition was used to determine a participant baseline and was each participant's natural arm swing motion. To examine the effect of absent arm swing on dynamic stability in pwPD, a "restricted arm swing" condition was used whereby participants placed their arms inside of the worn harness. More specifically, participants were asked to place their arms along their torso and thighs and insert their hands, wrists and lower forearms inside the safety harness. This allowed to physically restrict their

arm motion.

The data collection for this thesis occurred in a block randomized manner that progressed in the following order Steady-State Single-Task **(Research Objective 1, block 1)**, Steady-State Dual-Task **(Research Objective 1, block 2)**, Slips **(Research Objective 3, block 3)**, and Destabilizing Terrains **(Research Objective 2, block 4)**. All trials within each of these blocks were entirely randomized thereby minimizing trial effects in this thesis.

Research Objective 1: Unperturbed (steady-state) trials were as follows. Participants walked at their preferred walking pace on a split belt treadmill on the CAREN system (Motek Medical, Amsterdam, NL). Unperturbed single-task trials lasted 3 minutes with no perturbations and were conducted under each arm condition (unrestricted and restricted arm swing) for a total of 2 trials (block 1). These trials composed the first randomized block of trials. This was followed by unperturbed dual-task trials (block 2) that lasted 2 minutes each and consisted of the same arm swing conditions as the single-task trials, however, participants now walked while performing a simultaneous dual-task. These dual-task trials composed the second block. All single and dual-task trials were randomized within their respective blocks. The dual-task consisted of twelve words that randomly appeared on either the right or left side of a screen in the participants' visual field. To ensure instruction consistency between participants and avoid creating excessive awareness, minimal information was given regarding the motor and attentional tasks. Minimal instructions were provided in order to avoid influencing the participants' task prioritization and to avoid priming participants. Prior to each trial in block 1 and 2, participants were informed whether to walk with their natural arm swing (unrestricted arm swing) or to insert their arms inside the safety harness they wore (restricted arm swing). Further, at the beginning of block 2, participants were provided with the general information that words would randomly appear at different portions of

the screen and that they should call the words out as they appeared on the screen. Words were provided in participants' native or preferred language (French or English) and appeared at varying visual angles varying from 20 to 70 degrees for a duration of three seconds with a 2-4 second pause before the subsequent word's appearance. The length of the pause was randomly determined but bound to that duration per trial. This task was selected as it resembled conditions that individuals would likely encounter in their daily lives, specifically, scanning the environment for relevant cues. The dual-task lasted a total of 1 minute and 20 seconds and took place within the overall 2-minute trial duration with an onset only occurring after the first 25 seconds of the trial (to allow participants to reach a steady-state pace). Upon dual-task completion, participants walked for the trial's remaining 15 seconds. The word-searching task had a duration of 1 minute and 20 seconds in order to ensure that the words appeared during the steady-state portion of the trials and not during the starting of the trial (first 25 seconds) nor stopping of the trials (remaining 15 seconds).

Research Objective 2: During the minor perturbation trials, participants walked at their preferred walking pace in a virtual park environment within the CAREN system which simulated 200 meters of walking. There were two trials that occurred for this Research Objective (walking with unrestricted arm swing and restricted arm swing), which composed block 4. Trial order was randomized within this Research Objective. During these trials, participants first walked for 40 meters of unperturbed walking before encountering minor perturbations. Minor perturbations within the park terrain environment included a "rolling-hills terrain" (anteroposterior rotation of both a sum of sine waves), "rocky terrain" (pseudorandom perturbation in 3 different axes), and a mediolateral translation. Each terrain simulated a specific environmental condition. Specifically, the rolling-hills simulated walking along a forest path, the rocky terrain simulated walking along a rocky path, and mediolateral terrain simulating walking within a train or a bus.[122] The

perturbation magnitude of the terrains are located in **Table 2**. All terrains lasted for a simulated 20 meters of walking, within the overall 200m length of the trial, and were separated from one another by 40 meters of unperturbed walking. Therefore, participants encountered all three minor perturbations during both trials. The order of the perturbations within each of the trials was determined by the CAREN system operator in a pseudorandom manner. Trials in this Research Objective were conducted without a dual-task. Participants received no explicit instructions and were only informed if the upcoming trial was an unrestricted or restricted arm swing condition.

Research Objective 3: To examine the effect of recovery responses to slips, participants completed 4 perturbed walking trials. These trials included two arm swing conditions (unrestricted and restricted) that lasted 2 minutes each and were completed without a dual-task (block 3). All trials included one slip that was induced on either the least or most affected leg. Slips were caused by backward acceleration of the treadmill belt at $1.7 m/s^2$ for 0.75 seconds and were automatically triggered, at heel-strike, but only once the leg that was about to be slipped had passed the contralateral stance limb in the forward direction.[131] Immediately following the slip, the belt was decelerated at the same rate and duration to return participants to symmetrical walking at their preferring walking speed. Displacement from the slip was 1.8m. The onset of slips within each trial was determined in a pseudorandom manner manually by the CAREN system operator. All trials within this Research Objective were completely randomized. Slip onset did not occur during the first 25 seconds of the trials in order to allow participants to reach their preferred walking speed. Participants received no explicit instructions and were only informed if the upcoming trial was an unrestricted or restricted arm swing condition. Participants were informed at the beginning of this block that mechanical perturbations may now occur randomly during the remainder of the data collection session.

Participants were tested on the Computer Assisted Rehabilitation Environment "CAREN" (CAREN-Extended, Motek Medical, Amsterdam, Netherlands) in the Ottawa Hospital Rehabilitation Centre. The CAREN system combines a 6 degrees of freedom platform with an integrated split belt treadmill (Beretc Corp, Columbus, OH**)** with force plates under each track sampling at 1000Hz, a 180^{o} screen for 3D virtual world projection, and a 12 camera Vicon motion capture system (Vicon 2.6, Oxford, UK) sampling at 100Hz. Furthermore, the CAREN system recorded perturbation initiation at 300Hz. A six degrees of freedom marker set was used for motion tracking. The set consists of three markers on the platform to track platform motion and a 57-marker set to track full body kinematics. A static standing trial was obtained and afterward dynamic tasks were collected.[135]

3.5 Analysis and Data Reduction

Data collected from the static trial was used to create a whole-body kinematic model using Visual 3D (C-Motion, Germantown, MD, USA) and OpenSim. Visual 3D was used to create the kinematic model for inverse kinematics in Research Objectives 1 and 2. The Visual 3D kinematic model was a 13-segment (feet, shanks, thighs, pelvis, trunk, lower arms, upper arms, head) anthropometric model constructed from a linked set of rigid segments. Anthropometric data were taken from Hanavan (1964).[136] All joints in Visual 3D were modeled with 6 degrees of freedom. For Research Objective 3, an OpenSim 12-segment anthropometric model (feet, shanks, thighs, pelvis, trunk, lower arms, upper arms) was used that had 29 total degrees of freedom.[137] The OpenSim model (*Full Body Running Model)* did not include an articulated head segment. Each lower extremity of this model has 5 degrees of freedom, with the hip modeled as a ball-and-socket joint with 3 degrees of freedom. The knee joint is a custom model with only 1 degree of freedom while the lumbar motion is modeled as a ball-and-socket joint with 3 degrees of freedom. Further,

the arms have 5 degrees of freedom, the shoulder with 3 degrees of freedom, and the elbow as well as forearm with 1 degree of freedom. All kinematic data was passed through a fourth-order, zero-lag low pass Butterworth filtered at 10Hz (Article 1) and 12Hz (Article 2&3). Although raw trajectory markers for gait have a power-spectrum within 8Hz, due to the custom setup at the Ottawa Hospital Rehab Centre, it was found that a 12Hz cut-off frequency was more suitable for minimizing the data residuals. Therefore, the cut-off frequency was raised from 10 to 12Hz for Articles 2 and 3. Custom Matlab scripts (Mathworks, Natick, MA**)** were used to analyze all processed data for dynamic stability metrics, trunk kinematics, as well as spatiotemporal averages. To ensure that differences in arm swing range of motion occurred as a result of our experimental conditions (unrestricted and restricted arm swing), shoulder joint range of motions in the sagittal plane were calculated for all components of this thesis. Shoulder joint angle was calculated as the angle between the humerus and the torso in the Sagittal Plane.

3.6 Statistical Analysis

Statistical analyses included two-way repeated-measures within-subjects ANOVAs to examine **Research Objectives 1** and **3**. A three-way repeated-measures ANOVA was used to assess **Research Objective 2**. All dependent variables were imported into SPSS (Version 24, IBM Corporation, Armonk, NY) for statistical analyses. The average of trials was used to compare the variables across conditions. Main effects were assessed with an alpha level set *a prioi* at 0.05 and post-hoc with a Sidak-Bonferroni correction was used to analyze all main effects and interactions. Statistical Analyses were conducted on the following dependent variables: Margin of Stability, Harmonic Ratios, and Coefficient of Variation (Spatiotemporal Variability) all of which assess dynamic stability. Further, trunk linear and angular velocities along with spatiotemporal averages were assessed.

Chapter 4: **Limitations**

When considering the findings of this thesis, several limitations should be considered. Evidence comparing overground and treadmill walking demonstrates significant differences between the two conditions.[138–141] Indeed, compared to overground walking, individuals walk with a reduced gait speed and reduced spatiotemporal variability on treadmills.[140] In pwPD, the reduction in spatiotemporal variability is caused by the treadmill belt acting as an external pacemaker that facilitates rhythmic foot placement.[138] Additionally, participants in this thesis were tested on their optimally medicated state which affects their foot placement and reduces spatiotemporal variability.[73] Further, participants were in the mild to moderate stages of the disease. Finally, differences between freezers and non-freezers were not examined. As freezers demonstrate greater postural instability and increased spatiotemporal variability than non-freezers, the arm swing conditions in this research may have a distinct effect on their dynamic stability.[53,54,76]

Chapter 5: **Articles**

5.1 Article 1:

Background Summary:

- Parkinson's Disease is a neurodegenerative disease that progresses asymmetrically between the two hemispheres.
    - This asymmetric progression causes symptoms to be expressed to a greater extent on one side of the body (most affected) than the other (least affected).
    - However, despite this known asymmetry, examination of symptom laterality in gait in people with Parkinson's Disease is relatively sparse.
- Additionally, neurodegeneration within the Basal Ganglia impairs gait automaticity and timing in this demographic.
    - As a means to compensate for this impairment, people with Parkinson's Disease recruit the more intact higher level cortical regions to direct conscious attentional control on their locomotion.
    - However, in everyday environments, this attentional control strategy breaks down as attention is split between environmental cues and controlling their mechanics while walking.
    - Interestingly, despite the known asymmetric neurodegeneration of Parkinson's Disease, gait research examining dual-tasking effects have yet to examine differences between the least and most affected sides.
- Reduced and ultimately absent arm swing is a primary motor symptom of Parkinson's Disease.

- Current evidence demonstrates that the reduction of arm swing throughout the course of the disease is a strong predictor of falls in this population.
    - Despite this close association though, the effect of arm swing on Dynamic Stability in people with Parkinson's Disease remains unexamined.

Summary of Findings:

- Restricted Arm Swing increases trunk rotational velocity in the transverse plane
    - This subsequently acted as an internal destabilization to gait in this sample of individuals with mild to moderate stage Parkinson's Disease.
- In response, participants attempted to execute a compensation strategy, but were only effective in doing so on their least affected side:
    - Specifically, at heel-strike of the least affected leg, participants had a larger mediolateral MOS during the restricted arm swing condition.
        - This, subsequently, increased their Global Dynamic Stability as their Extrapolated COM was moved further away from the lateral edges of their BOS.
    - Additionally, participants reduced their average step length of their least affected leg during the restricted arm swing condition.
        - By reducing their step length, participants executed a strategy where they attempted to attenuate the increased trunk rotational amplitudes in the transverse plane.
- In contrast, the most affected side appeared ineffective in executing this compensation strategy. Instead, it appeared that step timing control in this limb was particularly sensitive to the loss of arm swing.

    - This impairment to step timing control was evidenced by the increase in step time variability that occurred during the restricted arm swing condition.
        - As one of the primary symptoms of Parkinson's Disease is impaired and variable movement timing, the increased step time variability is potentially indicative of this neuromuscular impairment.

- In regard to the dual-task, participants had reduced postural control in the frontal plane during the word searching task.
    - However, participants compensated for their impaired postural control by bilaterally adjusting their mediolateral foot placement to maintain their existing level of Dynamic Stability.

Absent Arm Swing and Dual-Tasking Decreases Trunk Postural Control and Dynamic Balance in People with Parkinson's Disease

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Tables: 2

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Abstract

Introduction: Falling during walking is a common occurrence in people with Parkinson's Disease and is closely associated with severe social and medical consequences. Recent evidence demonstrates that arm swing affects dynamic balance in healthy young adults, however, it remains unexamined what its effect is in people with Parkinson's Disease, particularly when combined with a secondary dual-task.

Methods: Twenty people with Parkinson's Disease (63.8±9.0 years) walked with two arm swing conditions (absent and normal) with and without a secondary dual-task. Data were collected on a split-belt treadmill CAREN Extended-System (Motek Medical, Amsterdam, NL). Average and standard deviations for trunk linear and angular velocity were calculated along with their instantaneous values (during foot-strikes) in all three axes. Averages and Coefficient of Variations for step length, time and width, Margin of Stability and Harmonic Ratios were also calculated.

Results: Compared to normal arm swing, absent arm swing reduced the least affected leg's average step length and increased its step length Coefficient of Variation while increasing step time Coefficient of Variation in the most affected leg. Further, absent arm swing reduced trunk anteroposterior instantaneous angular velocity (least affected leg) and reduced anteroposterior instantaneous linear velocity (bilaterally). For the vertical axis, absent arm increased the trunk's average angular velocity but reduced its instantaneous linear velocity and angular velocity standard deviation (least affected leg). Additionally, the Margin of Stability increased when the arms were absent (least affected leg). Alternatively, dual-tasking reduced average step time (most affected leg) and increased the step width Coefficient of Variation (bilaterally). Additionally, dual-tasking increased the mediolateral average angular velocity, instantaneous linear velocity standard deviation (bilaterally), and instantaneous angular velocity standard deviation (least affected leg).

For the vertical axis, dual-tasking increased average linear and angular velocity standard deviation as well as instantaneous angular velocity standard deviation (bilaterally).

Conclusion: Findings suggest that participants attempted to control extraneous trunk movement (due to absent arm swing) through compensatory responses in both lower and upper extremities. However, participants appeared to predominately compensate on their least affected side. Contrastingly, modifying mediolateral foot placement appeared to be the main means of maintaining walking stability while dual-tasking.

Introduction

Parkinson's Disease (PD) is the second most common neurodegenerative disease worldwide and is caused by progressive neurodegeneration within the Basal Ganglia of dopaminergic neurons[39,40] In addition to causing PD's cardinal symptoms (bradykinesia, rigidity and tremor), the Basal Ganglia's impaired function disrupts the gait pattern in people with PD (pwPD)[39]. This is particularly concerning as falling during walking is a debilitating threat that is closely associated with reduced autonomy and quality of life, hip fractures, and morbidity[12,13].

Current evidence demonstrates that gait impairments in the lower extremity for pwPD include shorter stride length, increased stance time, increased spatiotemporal variability, and reduced interlimb coordination compared to healthy elderly adults[5-11]. Although previous work has examined how PD affects postural control during static conditions and gait initiation, the quantification of postural control in pwPD during steady-state walking is not as prevalent[12-17]. However, initial evidence demonstrates that pwPD walk with reduced peak trunk velocity in the frontal and sagittal planes in respect to age matched adults[23]. As the Center of Mass (COM) is located in the upper body and the lower extremity adapts foot placement to modify the base of support, each is an integral component for maintaining dynamic balance (defined as maintaining the COM within a moving base of support)[5,107].

Traditionally, dynamic balance and rehabilitation research in pwPD are based on the inverted pendulum model which proposes that arm swing passively arises from trunk motion, gravity and inertia[28,142]. However, EMG and inverse dynamic evidence demonstrates that an active arm swing component assists in controlling trunk angular motion about the vertical axis[22]. While this improves gait's metabolic efficiency, research is conflicting whether arm swing affects

dynamic balance and trunk postural control[22-24]. Indeed, it is proposed that dynamic balance is either enhanced by arm swing facilitating a stable COM trajectory or by absent arm swing concentrating upper body mass thereby increasing inertia[30,143]. Furthermore, it remains unexamined how arm swing affects trunk postural control and dynamic balance in pwPD. As reduced arm swing occurs in pwPD and intensifies as the disease progresses until completely absent, determining arm swings' effects holds direct implications for fall prevention in pwPD[36].

However, the comprehensive quantification of dynamic balance requires multiple metrics as each reflects a distinct aspect of gait's neuromuscular control [5]. Indeed, neuromuscular control of gait requires input from both supraspinal and subcortical structures as well as the integration of feedback from a complex peripheral sensorimotor network[19,26,27]. This multilayered neuromuscular control is necessary as the upper and lower extremities have distinct movement patterns while walking [5,14,85]. Bauby and Kuo were the first to demonstrate that gait parameters in the anteroposterior (AP) direction were controlled by "passive" automated neuromuscular control while mediolateral (ML) parameters required "active" information processing[144]. These mechanisms serve to not only control stability locally of an anatomical segment, but also work in a coordinated manner to maintain global stability to avert a fall. As each dynamic balance metric is dependent on the trajectory and physical properties of the anatomical segment quantified, the direction of motion, time point in the gait cycle examined, and formulaic computation, no single metric is capable of quantifying dynamic balance in its entirety [5,145]. Thus, multiple metrics should be used simultaneously to quantify information that may be left unexamined by the use of any single metric.

Additionally, this multifaceted neuromuscular control provides a means for compensation should any aspect of the network become impaired[19,26]. For instance, in pwPD, dopamine loss

impairs subcortical pathways responsible for gait automaticity and timing[8,19,32,33]. Thus, to compensate, pwPD recruit higher supraspinal structures to bypass the impaired subcortical pathways and direct additional attention for effective ambulation[5,26]. However, in ever changing environments where multitasking is commonplace (terrain navigation, reading signs, talking etc.) attention becomes divided between multiple concurrent tasks [80]. When attention is divided between simultaneous tasks, the resources necessary to compensate for impaired gait automaticity become strained thereby impairing effective locomotion[26]. While divided attention is demonstrated to impair lower extremity measures of dynamic balance in pwPD, its effect on additional dynamic balance metrics remains to be examined, particularly in the presence of varying arm swing conditions[26].

Therefore, our study's purpose was to examine normal and absent arm swing's effect on linear and angular trunk velocities as well as lower and upper body measures of dynamic balance with and without a dual-task in pwPD. We hypothesize that absent arm swing and dual-tasking will decrease instantaneous trunk linear and angular velocities as well as variability but increase these parameters' average values. Further, both arm swing and dual-tasking will elicit unique responses from dynamic balance measures. Additionally, we predict that absent arm swing while dual-tasking will be more destabilizing than all other conditions. Alternatively, walking with normal arm swing and dual tasking will only be more destabilizing compared to walking with normal arm swing without the dual-task.

Methods

Participants

Twenty pwPD (13 males; 7 females), aged 48 - 79 years old (63.8 ± 9.0 years) were

recruited from the Ottawa-Gatineau region. Participants were assessed with the original Unified Parkinson's Disease Rating Scale Motor examination (11.0± 6.0) and were between I-III on the Hoehn & Yahr scale. Average disease duration (8.0±5.1 years) and age at onset (56.8±9.6 years) data were collected. Further, seven participants reported freezing of gait based on the Freezing of Gait Questionnaire. However, due to two participants presenting severe dyskinesia and one participant having an incomplete session, only 17 participants were used in the analyses. As pwPD who develop dyskinesia are characterized by excessive movement dyscontrol, we excluded these participants from further analyses for our sample to be more representative of PD. Participants were tested on their optimally medicated state. An *a priori* power analysis revealed that 12 participants were adequate to achieve power at $\beta = 0.8$. Prior to data collection, volunteers were excluded if they reported any physical discomfort using a virtual reality system, reported any injuries and/or orthopedic surgeries that could interfere with gait, could walk only with the use of a walking aid, and had any additional illnesses other than PD. All participants provided written informed consent and the study was approved by local ethics and scientific committees.

Procedures

Participants walked with two arm swings (absent and normal) during single (ST) and dual-task (DT) conditions for a total of four trials. ST trials lasted three minutes each, while DT trials were two minutes each. To assure safety and prevent falls, participants wore a safety harness attached to an overhead structure at all times. Arm conditions were randomized per block with all ST trials occurring before DT trials. The DT consisted of a word searching task with twelve familiar words randomly appearing in the participants' visual field. Words appeared one at a time on both left and right sides of a screen in front of participants and varied between 20-70 degrees. Each word was shown for 3 seconds with a 2-4 second pause between subsequent words.

Participants verbally called out each word as they appeared. This task was chosen as people with Parkinson's Disease rely heavily on their vision to compensate for the impairment caused to sensory feedback systems (proprioception) from the disease's neurodegeneration. During the absent arm swing trials, participants inserted their arms inside the safety harness, which effectively prevented arm motion. Participants were allowed to rest whenever necessary to minimize fatigue.

3D motion analysis was completed with the CAREN-Extended System (Motek Medical, Amsterdam NL) using a virtual park terrain. This system combines a six degree-of-freedom motion platform with embedded dual-belt treadmill, 12 Camera Vicon motion capture system, 180-degree projector screen, and safety harness **Figure 5.1-1**. The dual-belt was synchronized so that both belts were symmetrically set at the participants preferred walking speed. Three markers placed in the periphery of the treadmill were used to track platform motion, and a 57-marker set for tracking full body kinematics [123,145]. Kinematic data were collected at 100Hz and ground reaction forces (GRF) at 1000Hz.

Kinematic and Kinetic Analyses

Markers and GRF data were processed in Vicon Nexus (Nexus 2.6, Oxford, UK), while 3D kinematics and kinetics were calculated in Visual 3D. A 4^{th} order, low-pass Butterworth filter with a 10Hz cut-off frequency was used to filter marker data. To remove start-up effects, the first 25 seconds were removed before data analysis. Data were analyzed by custom Matlab scripts (Mathworks, Natick, MA) to calculate average trunk linear and angular velocities (average LV and AV) and variability (LV-SD and AV-SD) throughout the gait cycles as well as instantaneous velocities (instantaneous LV and AV) and velocity variability (instantaneous LV-SD and AV-SD) at heel-strike. As the lower extremity is accepting the trunk's weight over a relatively small support

at heel-strike, quantifying instantaneous velocities provides potential subtleties overshadowed by average values [14]. Additionally, average spatiotemporal parameters (step time, length and width) and dynamic balance measures including the Coefficient of Variation (COV), Margin of Stability (MOS), and Harmonic Ratios (HR) were quantified. The COV was calculated as:

$$\boldsymbol{COV = \left(\frac{SD}{average}\right) x100}$$

Where SD is the standard deviation for the spatiotemporal parameter. The MOS was calculated bilaterally at both heel strike's and defined as the distance of the Extrapolated Center of Mass (xCOM) to the right/left lateral heel marker.

$$\boldsymbol{MOS = Lateral\ Heel\ Marker - xCOM}$$

The formula for xCOM was:

$$\boldsymbol{xCOM = COMp + \left(\frac{COMv}{\omega\theta}\right)}$$

Where COM_p = COM's position, COM_v = COM's velocity. ω_Θ was calculated as:

$$\boldsymbol{\omega\theta = \sqrt{g/l}}$$

In this term, g = 9.81m/s^2 and l is the length of the inverted pendulum determined as the average distance of the right/left lateral heel marker to the COM at heel-strikes. Visual 3D was used to calculate the COM's position and velocity. The MOS was only calculated in the ML direction as this metric is only valid in this direction during steady-state walking [96,145].

HRs were calculated on the COM's acceleration defined as the first central difference of the COM's velocity. The HR examines a signal's periodicity by calculating a ratio of the amplitudes of the even and odd harmonics obtained through a Fast Fourier Analysis. The HR for the AP (HR-AP) and VT (HR-VT) directions were calculated as the first 10 even harmonics divided by the first 10 odd harmonics while the HR-ML calculation was the inverse, higher values

in all cases indicated greater dynamic balance [5,6,8,94,145]. All discrete metrics were quantified at heel-strikes for least and most affected legs.

Statistical Analyses

Data were analyzed using SPSS 23.0 and $p<0.05$ was considered statistically significant. The normality of variables was verified using Shapiro-Wilks test and a two-way repeated measure ANOVA was performed to find the effect of arm swing, dual-tasking and potential interactions. If statistical significance was achieved with the ANOVA ($p<0.05$) then pairwise comparisons with a Sidak-Bonferroni adjustment for multiple comparisons were used for post-hoc analyses. After the correction, findings were only deemed statistically significant when ($p<0.026$).

Results

Dynamic balance (Harmonic Ratios, Margin of Stability, and Coefficient of Variation) measures are reported in **Table 5.1-1a** while Spatiotemporal averages are reported in **Table 5.1-1b**. Average trunk linear and angular velocities as well as variabilities are reported in **Table 5.1-2a** while their instantaneous values at heel-strike are in **Table 5.1-2b**.

Arm Swing

An arm-task interaction occurred for COV step length in the least affected leg where the single-task normal arm swing had less variability than the single-task absent arm swing ($F(1,16)=6.08$, $p = 0.025$, $\eta_p^2= 0.275$). A further interaction existed for the least affected leg's COV step time where single-task normal arm swing had less variability than dual-task absent arm swing ($F(1,16)=5.21$, $p = 0.037$, $\eta_p^2= 0.245$). An arm-task interaction also occurred for instantaneous trunk LV at heel-strike of the most affected leg ($p=0.031$) but post-hoc tests were non-significant.

ANOVA results revealed that absent arm swing elicited the following responses in the least affected leg: a reduced average step length ($F(1,16)=7.06$, $p = 0.017$, $\eta_p^2 = 0.306$), increased COV step length ($F(1,16) = 7.48$, $p = 0.015$, $\eta_p^2 = 0.319$). Further, absent arm swing increased the most affected leg's step time COV ($F(1,16)= 5.90$, $p = 0.027$, $\eta_p^2= 0.269$). In the AP direction, absent arm swing reduced the trunk's instantaneous LV at heel-strike of the least ($F(1,16)= 11.314$, $p = 0.004$, $\eta_p^2= 0.414$) and most ($F(1,16)= 7.217$, $p = 0.016$, $\eta_p^2= 0.311$) affected legs. Additionally, the trunk's instantaneous AV decreased without arm swing at heel-strike of the least affected leg ($F(1,16)= 9.161$, $p=0.008$, $\eta_p^2=0.364$). Alternatively, in the ML direction, absent arm swing increased the MOS ($F(1,16)=7.00$, $p = 0.018$, $\eta_p^2= 0.304$) at heel-strike for the least affected leg. Along the vertical axis, absent arm swing reduced the trunk's instantaneous LV at heel-strike of the least affected leg ($F(1,16)=13.831$, $p = 0.002$, $\eta_p^2= 0.464$). Further, absent arm swing increased average trunk AV ($F(1,16)= 8.37$, $p = 0.011$, $\eta_p^2= 0.343$), a trend for reduced the instantaneous trunk AV-SD at heel-strikes of the least ($F(1,16)=4.45$, $p = 0.051$, $\eta_p^2= 0.218$) and most ($F(1,16)= 8.740$, $p = 0.025$, $\eta_p^2= 0.276$) affected legs.

Dual-Task

For dual-tasking main effects, walking with the secondary task reduced the least affected leg's average step time ($F(1,16)= 6.117$, $p = 0.025$, $\eta_p^2= 0.277$). Dual-task walking also increased the step width COV for the least ($F(1,16) = 9.252$, $p = 0.008$, $\eta_p^2= 0.366$) and most ($F(1,16)=10.321$, $p = 0.005$, $\eta_p^2= 0.392$) affected legs. In the ML direction, dual-tasking increased the trunk's average AV ($F(1,16)= 4.793$, $p = 0.044$, $\eta_p^2= 0.230$), AV-SD ($F(1,16)=6.748$, $p = 0.016$, $\eta_p^2= 0.297$), and LV-SD ($F(1,16)= 15.947$, $p = 0.001$, $\eta_p^2= 0.499$). Further, DT increased the trunk's instantaneous LV-SD at the heel-strikes for the least ($F(1,16)=9.335$, $p = 0.008$, $\eta_p^2= 0.368$) and most ($F(1,16)=12.550$, $p = 0.003$, $\eta_p^2= 0.440$) affected legs as well as the trunk's instantaneous

AV-SD for the least affected leg ($F(1,16)=5.90$, $p = 0.027$, $\eta_p^2= 0.269$). Furthermore, the trunk's vertical LV-SD ($F(1,16)= 6.555$, $p = 0.021$, $\eta_p^2= 0.291$) and AV-SD ($F(1,16)= 12.746$, $p = 0.003$, $\eta_p^2= 0.443$) increased throughout the entire gait cycle while dual-tasking. Additionally, dual-tasking increased the trunk's vertical instantaneous AV-SD at heel-strike of the least ($F(1,16)= 7.28$, $p = 0.016$, $\eta_p^2= 0.313$) and most ($F(1,16) = 9.078$, $p = 0.008$, $\eta_p^2= 0.362$) affected leg.

Discussion

Main Findings

This study examined how absent and normal arm swing affects dynamic balance and postural control while walking with and without a DT in pwPD. Our results supported our hypothesis that each dynamic balance measure would respond uniquely to absent arm swing and DT. Indeed, absent arm swing increased the MOS and step length COV for the least affected leg as well as increased step time COV in the most affected leg. However, absent arm swing did not affect the HR. Alternatively, DT increased step width COV for both legs but did not affect the MOS or HR values. Further, our hypothesis on trunk postural control was only partially supported. Indeed, absent arm swing only increased average AV about the vertical axis. However, as predicted, absent arm swing decreased instantaneous LV and AV in the AP direction as well as decreased instantaneous AV-SD about the vertical axis. Alternatively, DT increased average AV only in the ML direction as well as increased average and instantaneous variabilities only in the ML direction and vertical axis.

Arm Swing

Our hypothesis that absent arm swing would increase trunk velocity was only partially supported. Indeed, only the average trunk AV about the vertical axis increased when arm swing

was removed. This was expected as the 1:1 arm-leg swing ratio controls the trunk's angular motion about the vertical axis by equalizing the torques acting on the COM (22). However, as hypothesized, instantaneous AP-LV and VT-LV and AP-AV decreased during absent arm swing; suggesting that our participants adopted a compensatory trunk stiffening strategy. Previous research demonstrates that pwPD adopt this strategy to control excessive trunk movement during dynamic movements such as walking [23]. These changes in instantaneous velocities at heel-strike, particularly in the AP direction, may be due to the importance this time point has in the gait cycle as it marks the beginning of interlimb trunk transfer [14,85]. Our participants may have reduced the instantaneous AP-LV to reduce the likelihood of a forward balance loss or the need for a reactive recovery step while the weight of the trunk is transferred between limbs. Additionally, at heel-strike, the trunk flexes forward under the force of gravity but is attenuated by an opposing force from lower back and hip musculature [14,85]. Therefore, the reduced instantaneous AP-AV suggests that our participants adopted a stiffening strategy to maintain a more erect posture to attenuate gravity's perturbing effect.

While a trunk stiffening strategy accounts for our AP and VT findings, it does not account for the lack of changes in the ML direction. This disparity likely arose due to the different postural strategies observed for trunk AP and ML movement in pwPD[40]. Additionally, Jehu and Nantel discussed that while a stiffening strategy enhances postural control in the short term, it could also lead to an inability to efficiently implement adaptive responses to internal and external perturbations [23]. Thus, our participants may not have restricted ML mobility to appropriately adapt to the increased trunk rotation during absent arm swing. Preserving ML mobility would also account for the increased MOS when the arms were absent. By increasing the xCOM distance to the base of support's edge, our participants enhanced their ability to mitigate global balance loss

in the ML direction despite the faster rotating trunk [147,148].

Interestingly, increases in our participants' MOS only occurred at heel strike of the least affected side. In PD, asymmetric neurodegeneration compromises mobility in one leg to a greater degree than the other causing an asymmetric gait[43-47]. In an examination of asymmetric walking, Buurke et al. demonstrated that the MOS is larger in the faster leg than the slower in healthy adults[149]. The authors discussed that this disparity arose either passively from the faster leg's reduced stance time (which increases the MOS) or actively as a compensatory response[149]. In our study, no change in step time occurred during the absent arm condition suggesting active adaptation. Therefore, our MOS findings likely resulted from the asymmetric neurodegeneration impairing the most affected leg's adaptation to absent arm swing. While this strategy is intended to enhance global balance, it potentially could be maladaptive since the xCOM is now shifted closer towards the most affected leg. Should an external perturbation occur on the most affected leg, the xCOM would now have to overcome a greater distance to be in a position where the contralateral leg could act as a base of support. Similarly, asymmetric neurodegeneration would account for the reduction in only the least affected leg's step length. Huang et al. demonstrated that spinal rotational amplitudes decrease when healthy adults reduce their step length[49]. Thus, our participants plausibly adapted their step length to attenuate the increased trunk rotation during absent arm swing. This may be a mechanism to decrease and partially counteract the torques acting on the COM's vertical angular motion, a task normally accomplished through contralateral arm-leg swing[22]. However, lack of findings in the most affected leg suggests that this side was impaired in its ability to execute a similar response.

Additionally, adaptation differences between legs to the absent arm swing may account for our COV findings. Although large spatiotemporal variability is a strong predictor of falls in

pwPD, previous research suggests that a certain amount of variability is necessary for lower extremity adaptation[5,150]. However, the COV only quantifies variability's magnitude and is therefore incapable of parsing adaptive responses from neuromuscular impairments[19,33]. The increases in step length COV for the least but not the most affected leg may partially be due to the aforementioned step length adaptation. Alternatively, the most affected leg's increased step time COV may indicate motor impairment in that leg's rhythmic temporal sequence. However, additional research is necessary to elucidate variability differences in lower limb motor adaptability and instability in pwPD. To this end, our interaction (least affected leg's: COV step length being lower during ST normal arm swing than ST absent arm swing and its COV step time being lower during ST normal arm swing than DT absent arm swing) results plausibly reflect the differences in variability that could arise from adaptation or from neuromuscular impairment. For instance, the interaction that occurred for the least affected leg's COV step length between ST conditions supports the notion that the changes in this parameter were an adaptive mechanism to partially counteract the increased trunk rotation. However, the COV step time interaction in the least affected leg is plausibly more indicative of neuromuscular impairment as the DT absent arm swing is arguably the most destabilizing condition. Further, as our participants were in the moderate stage of disease progression, more variable step timing may only have been elicited when the two destabilizing conditions (DT and absent arm swing) were combined. Differences in step time COV as a reflection of neuromuscular impairment holds with evidence demonstrating impairment to the internal timing of movement in pwPD[5,26,51]. Unexpectedly, no HR differences were observed. HRs quantify dynamic balance by examining the COM's periodicity and reflect (a)symmetric leg motion[6,8]. As pwPD walk with increased spatiotemporal asymmetry, our findings suggest that absent arm swing did not affect our participants' gait asymmetry level [153].

While our arm swing results demonstrate the contribution of nominal arm swing to dynamic balance and postural control in pwPD, which becomes asymmetric and absent with disease progression, consciously increasing arm swing may also elicit changes in gait parameters[25,52]. Indeed, empirical evidence demonstrated that increasing arm swing beyond nominal levels deteriorates dynamic balance and postural control in healthy young adults [145]. This may be due to an internal focus of attention elicited when individuals are made to consciously modify their own mechanics and, similarly, a dual-tasking effect as individuals would direct their attention to the secondary task of moving their arms. As pwPD have a reduced DT ability due the disease's neurodegenerative nature, this demographic may have additional difficulty in consciously modifying and maintaining arm motion[5,26]. Thus, future work and clinicians should consider examining interventions that are more implicit and do not tax attentional resources to restore the contralateral arm-leg swing pattern.

Dual-Task

Our hypothesis that DT would increase trunk velocity and variability was supported as DT increased variability in the mediolateral and vertical axes. However, the additional increase in the ML average AV suggests that DT was more threatening to our participants' postural control in this direction. During walking, Mackinnon and Winter discussed that the neuromuscular system mitigates gravity's destabilizing torque in this direction through a counterbalancing torque generated by hip and trunk musculature[15]. This careful balancing of torques maintains an individual's erect posture and leveled visual field[15]. In healthy adults, precisely counterbalancing gravity's perturbing effects on the trunk is partially accomplished through proprioception [26]. However, previous research indicates that this system is impaired relatively early in pwPD which reduces their kinesthetic sense[26,155,156]. Therefore, pwPD compensate by relying on visuomotor

control which remains largely spared from the disease's neurodegeneration as it is conveyed primarily through neuronal afferents to the cerebellum[26,157]. However, visuomotor feedback requires attentional resources from higher level cortical structures for active information processing [144]. As such, our participants' increased frontal plane AV as well as AV-SD and LV-SD during DT plausibly arose as attention was divided between postural control and the visual word searching task. Interestingly, participants' instantaneous AV-SD in the ML direction only increased for the least affected leg. This may be due to the asymmetric neurodegeneration limiting mobility and potential adaptive responses that could occur at heel-strike on the most affected side.

Additionally, as trunk movement determines lower extremity foot placement, the increases in our ML trunk variability during DT accounts for the simultaneous increase in step width COV for both legs [14,15,85]. Although large changes in step width COV indicate reduced dynamic balance, a certain amount of variability is necessary for foot placement adaptation to maintain ML balance [5,84,150]. Therefore, when our step width COV results are examined alongside the MOS findings, our results suggest a more adaptive response whereby our participants correctly predicted the trunk's irregular ML trajectory and appropriately adapted foot placement to preserve their existing global dynamic balance[145]. However, our findings conflict with previous reports demonstrating no changes in step width COV in pwPD while DT[150]. This may be due to conflicting evidence demonstrating both increases and decreases in pwPD's postural sway compared to age matched adults. During walking, the magnitude of trunk sway directly influences ML foot placement, therefore, multiple responses to DT would elicit various responses in step width COV.

Various DT responses further accounts for our lack of findings in step time and length COV as well as HRs. Indeed, Yogev et al. discussed that DT responses in pwPD are dependent on task complexity and the stage of disease progression[26]. In our study, the simple word searching

task may not have challenged our participants' attentional resources to a degree that caused a breakdown in their rhythmic spatiotemporal control nor their level of gait asymmetry. Interestingly, our participants' lower extremity response to DT was somewhat unexpected as step time was reduced in the least affected leg. This conflicts with evidence demonstrating that pwPD display a more "cautious gait" strategy when DT which includes an increase in step timing[5,6,26]. However, previous research demonstrates that spatiotemporal asymmetry increases in pwPD during DT[9,26]. In our study, therefore, the least affected leg's reduced step time plausibly arose due to the DT progressively increasing our participants' gait asymmetry. In summary, our DT results indicate that clinicians should consider therapeutic programs that facilitates appropriate foot placement to adapt the base of support in pwPD to maintain global dynamic balance. Further, clinicians should carefully consider the complexity of the DT used in the therapy to be appropriate to the stage of PD progression and the phenotype of PD presented. This is particularly crucial as the amount of attentional resources available to control posture will vary which impacts pwPD ability to maintain postural control in the ML direction during multitasking.

Limitations

When considering our findings, it is important to consider that our participants were tested in the optimally medicated state. Empirical evidence on L-Dopa demonstrates that ML postural sway (displacement and velocity) deteriorates when pwPD are in their "ON" medicated state compared to "OFF"[40]. Further, our COV parameters would be impacted by the medication state as medication reduces spatiotemporal variability in pwPD[32, 60]. Additionally, differences between freezers and non-freezers were not examined which may yield insightful nuances into the responses elicited in each group during absent arm swing[26,47,61]. As neurodegeneration is suggested to be more widespread in freezers than non-freezers, the ability to execute adaptive responses to

the DT and absent arm swing may be unique to each group[26]. Finally, our results are limited in that no aged matched control was used to compare the effects of arm swing and dual-tasking. Future research should consider comparing pwPD to an aged-matched control group as both groups may respond and compensate differently to absent arm swing and dual-tasking conditions.

Conclusion

To conclude, removing arm swing in pwPD increased average trunk AV about the vertical axis and elicited compensatory responses in both the upper and lower extremity. These responses arose to compensate for the absent 1:1 arm-leg swing ratio that controls trunk angular motion about the vertical axis. Indeed, the trunk stiffening strategy in our participants was an upper body response to plausibly avert forward balance loss and maintain a more erect posture at heel-strike when the arms were removed. Additionally, the alterations in the least affected leg (increased MOS, increased step length variability, decreased average step length) suggest that pwPD attempt to enhance global dynamic balance and attenuate the increased trunk rotation as compensation for absent arm swing. However, PD's asymmetric neurodegeneration appears to impede the ability to bilaterally execute these strategies in response to internal perturbations such as the increased trunk rotation. Alternatively, the absent arm swing condition plausibly disrupted the most affected leg's temporal sequence as indicated by the increased step time variability. Our findings, therefore, point to the effective role arm swing plays in maintaining dynamic balance in pwPD. As arm swing amplitude is reduced and asymmetric in pwPD, clinicians should consider programs that aim to restore gait's contralateral arm-leg swing ratio to enhance dynamic balance and postural control. However, caution must be executed in the therapeutic method used to restore nominal arm-leg swing motion as consciously increasing arm swing while walking is demonstrated to negatively impact gait mechanics. Finally, therapeutic programs should incorporate strategies that maintains

trunk postural control, particularly in the mediolateral direction, and facilitate lower extremity adaptation during DT. As multitasking is commonplace in everyday situations, strategies aimed at training postural control during DT would be beneficial in fall prevention for pwPD. Indeed, our results demonstrate that DT reduces pwPD ability to maintain postural control in the ML direction as indicated by the increase in our participants average AV as well as AV-SD and LV-SD. Although postural control is an integral component that contributes to dynamic balance, our participants were able to avert a balance loss by adapting their ML foot placement to keep their xCOM within their base of support thereby demonstrating that adaptation is partially preserved in pwPD.

Author Contributions

Julie Nantel conceptualized and organized the research project; **2)** Data Analysis: A. First Analysis (Tarique Siragy), B. Review and Critique (JN) , **3)** Statistical Analysis: A. Design (TS and JN), B. Execution (TS), C. Review and Critique (JN); **3)** Manuscript: A. Writing of the first draft (TS), B. Review and Critique (JN).

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Conflict of Interest

We declare no conflict of interest.

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Table 5.1-1a. Dynamic Balance measures for arm (absent and normal) and task (single and dual) conditions: Coefficient of Variation for step length, width and time, Harmonic Ratios in all 3 axes, and mediolateral Margin of Stability. Coefficient of Variation and Margin of stability were quantified at foot-strike of the least and most affected legs. P-values from the two-way repeated measures ANOVA are reported for arm swing main effects.

		Single-Task		Dual-Task		
		Normal	Absent	Normal	Absent	P-Value
Harmonic Ratios	AP	3.07±1.03	2.93±0.88	3.03±0.87	3.16±0.84	.956
	ML	5.16±0.94	5.30±1.25	5.07±1.28	5.13±1.31	.640
	Vert	3.97±1.87	3.53±1.71	3.54±1.23	3.63±1.29	.135
Margin of Stability (cm)		11.40±1.2	11.79±1.4	11.39±1.49	11.73±1.69	.018†
		11.09±1.87	11.35±1.88	11.04±1.97	11.24±2.12	. 117
COV Step Length		3.69±1.62	4.73±2.36	4.19±1.95	4.15±1.64	.015†°
		3.96±1.19	4.54±2.29	4.40±1.89	4.43±1.39	.269
COV Step Time		2.47±0.54	3.09±1.04	3.06±1.03	3.00±0.68	.152°
		2.61±0.58	3.33±1.37	2.89±0.89	2.94±0.73	.027†
COV Step Width		8.45±3.16	8.34±3.91	10.19±3.92	10.00±4.39	.690*
		8.04±2.00	8.47±3.14	10.13±3.85	9.70±3.89	.985*

† Arm Swing Main Effects at $p<0.05$
* Dual-task Main Effects at $p<0.05$
° Arm-Task Interaction at $p<0.05$

Table 5.1-1b. Averages for step length, time, and width for arm (absent and normal) and task (single and dual) conditions at foot contact for least (white) and most (gray) affected legs. P-values from the two-way repeated measures ANOVA are reported for arm swing main effects.

	Single-Task		Dual-Task		
	Normal	Absent	Normal	Absent	P-value
Average Length (cm)	50.30±6.78	48.11±6.89	49.73±6.82	49.25±6.14	.017†
	48.96±6.76	46.93±6.85	47.82±6.31	47.86±6.11	.057
Average Time (ms)	554±54	554±61	542±56	541±52	.861*
	538±39	536±45	536±41	530±44	.155
Average Width (cm)	18.99±3.96	19.56±4.36	18.91±4.52	19.46±4.73	.071
	18.94±3.98	19.31±4.23	18.99±4.47	19.36±4.71	.206

† Arm Swing Main Effects at $p<0.05$

* Dual-task Main Effects at $p<0.05$

Table 5.1-2a. Trunk average linear and angular velocities and variabilities in all 3 axes for arm (absent and normal) and task (single and dual) conditions. Values were rounded to the nearest whole number. P-values from the two-way repeated measures ANOVA are reported for arm swing main effects.

		Single-Task		Dual-Task		
		Normal	Absent	Normal	Absent	P-Value
Trunk linear velocity (cm/s) ($x10^{-2}$)	AP	22±3	13±20	14±26	18±24	.407
	ML	15±3	14±8	13±10	17±11	.420
	Vert	-15±4	-14±5	-15±4	-15±5	.178
Trunk linear velocity variability (cm/s) ($x10^{2}$)	AP	312±90	327±127	377±181	333±106	.302
	ML	201±53	199±66	251±64	254±69	.923*
	Vert	31±10	32±12	38±18	35±12	.482*
Trunk angular velocity (°/s) ($x10^{-2}$)	AP	12±10	11±11	13±9	13±11	.797
	ML	4±14	5±12	6±13	7±15	.271*
	Vert	6±8	12±8	7±8	10±11	.011†
Trunk angular velocity variability (°/s) ($x10^{-2}$)	AP	92±24	95±25	105±40	99±35	.614
	ML	80±22	78±18	94±30	91±22	.421*
	Vert	158±51	153±50	226±67	219±66	.321*

† Arm Swing Main Effects at $p<0.05$
* Dual-task Main Effects at $p<0.05$

Table 5.1-2b. Instantaneous Trunk linear and angular velocities and variabilities in all 3 axes for arm (absent and normal) and task (single and dual) conditions at heel-strike of least (white area) and most affected (gray area) legs. P-values from the two-way repeated measures ANOVA are reported for arm swing main effects.

		Single-Task		Dual-Task		
		Normal	Absent	Normal	Absent	P-Value
Inst. Trunk linear Velocity (cm/s)	AP	4.7±1.8	3.2±1.6	4.5±2.6	3.7±1.9	.004†
		4.0±2.9	3.0±2.1	3.8±2.9	3.1±2.4	.016†
	ML	-1.7±15.4	-1.3±15.3	-1.3±15.3	-0.8±15.3	.062
		1.3±15.8	1.2±15.5	1.0±16.0	1.0±15.8	.852
	Vert	-17.0±4.0	-15.4±5.0	-16.7±10.0	-15.7±5.0	.002†
		-15.3±3.9	-13.7±4.2	-14.8±4.0	-15.2±4.7	.069
Inst. Trunk linear velocity Variability (cm/s)	AP	3.5±0.9	3.6±1.1	3.8±1.4	3.5±1.0	.568
		3.4±1.0	3.5±1.1	3.8±1.4	3.6±1.2	.829
	ML	2.7±0.5	2.5±0.7	3.1±0.8	3.1±0.7	.491*
		2.4±0.5	2.5±0.6	2.9±0.7	2.9±0.6	.228*
	Vert	16.8±0.6	18.2±0.7	18.6±0.8	19.5±0.6	.191
		18.4±0.6	18.4±0.6	18.4±0.6	19.3±0.6	.544
Inst. Trunk Angular Velocity (°/s)	AP	6.1±5.2	3.9±6.5	6.0±6.2	4.5±7.4	.008†
		6.6±4.5	4.3±4.4	7.1±4.8	4.6±5.6	.005†
	ML	-0.2±6.0	-0.7±7.7	-0.9±7.8	-0.9±7.9	.844
		0.5±4.7	0.8±5.7	2.1±5.4	1.4±6.9	.782
	Vert	-2.9±10.5	-3.1±17.6	-3.1±11.3	-3.9±18.2	.828
		1.4±11.4	4.2±14.6	3.1±14.0	3.5±17.0	.336
Inst. Trunk Angular Velocity Variability (°/s)	AP	3.8±1.2	3.8±1.0	4.1±1.0	4.0±1.3	.506
		3.5±1.1	3.7±1.0	3.7±1.0	4.0±1.1	.131
	ML	2.7±1.0	2.5±0.5	2.8±1.0	2.9±0.6	.620*
		2.7±1.0	2.6±0.6	2.7±0.8	2.9±1.1	.523
	Vert	5.8±2.0	5.1±2.2	6.5±2.3	5.9±1.9	.051*
		5.7±1.6	4.7±1.2	6.1±2.1	5.7±1.6	.025†*

† Arm Swing Main Effects at $p<0.05$

* Dual-task Main Effects at $p<0.05$

Supplementary Material

Table 1: Average range of motion (degrees) in the Sagittal Plane based on shoulder angle for absent and normal arm swing conditions. Data were combined for the least and most affected sides (per condition) as no statistical differences existed between them. Data were analyzed with a within-subjects repeated measures ANOVA.

	Single-Task		Dual-Task		
	Normal	Absent	Normal	Absent	P-value
Range of Motion(°)*	20.49±12.22	3.98±4.49	18.52±13.60	5.68±6.14	< .001

* Arm Main Effects at $p<0.05$

Table 2: Average Accuracy for Performance on the 12 Words-Word Searching Task for Normal and Absent Arm Swing.

	Normal	Absent
Accuracy	11.05±1.47	11.5±0.89

Figure 5.1-1. Experimental Setup for the CAREN Extended virtual environment

5.2 Article 2:

Background Summary:

- In people with Parkinson's Disease, reduced and ultimately absent arm swing is a primary motor symptom the develops throughout the course of the disease.
    - Interestingly, reduced arm swing amplitude has been demonstrated to be a strong predictor of falls in this demographic.
- Initial evidence demonstrated that when people with Parkinson's Disease walk with restricted arm swing, there is an increase in trunk rotational velocity in the transverse plane during steady-state conditions (**Article 1**).
    - This increase in trunk rotational velocity has been proposed to act as an internal destabilization to the gait pattern in this population.
    - However, the effect of arm swing on dynamic stability in people with Parkinson's Disease is unexamined on destabilizing terrains.
- Examining the effect of the loss of arm swing on different destabilizing terrains is critical for assessing fall risk as multiple terrains are encountered in everyday environments.
    - Destabilizing terrains such as rocky (rocky surface), rolling-hills (forest trail), and mediolateral-translational (bus or train) are common paradigms used to simulate real-world surfaces.
    - Therefore, it is paramount to examine the effect of arm swing on dynamic stability in people with Parkinson's Disease as they ambulate on these terrains.
- Additionally, it remains unexamined whether the asymmetric neurodegeneration of Parkinson's Disease affects foot placement differently, between the least and most affected legs, when walking on these destabilizing surfaces.

Summary of Findings:

- Evidence from **Article 1** demonstrated that when people with Parkinson's Disease walk with restricted arm swing, during steady-state conditions, trunk rotational velocity in the transverse plane increased.
    - This in turn acted as an internal destabilization to the participants' gait.
- Interestingly, various effects of the restricted arm swing condition were observed on dynamic stability and foot placement metrics on the destabilizing terrains in this study.
    - These distinct differences in the metrics suggest that the internal destabilization to gait, that arises from the restricted arm swing condition, depends on the terrain people with Parkinson's Disease encounter.
- However, the increased step time variability (rolling hills) and reduced average step time (rocky) suggest that step timing control appears particularly sensitive to the loss of arm swing in people with Parkinson's Disease.
    - These findings on restricted arm swing are in line with those observed in **Article 1**. Specifically, an increased step time variability would reflect neuromuscular impairment to step timing control in these individuals.
    - Further, a faster step time (reduced average step time) reduces the amount of time for determining optimal foot placement.
- Findings from the destabilizing terrains indicate that the participants were partially capable of bilateral foot placement adjustment to maintain their existing level of global dynamic stability.
    - This was demonstrated by the increased spatiotemporal variability that occurred on the rolling hills and rocky terrains without any changes in the Margin of Stability.

        - On destabilizing terrains, no changes in variability would in fact be an incorrect strategy as it would indicate that foot placement is not being adjusted to new terrains encountered.
    - Furthermore, support for foot placement adjustment in this sample is demonstrated by the participants' faster (rolling hills) and wider (rocky) steps.
        - Indeed, faster steps allowed participants to pass over the destabilizing surface quicker while wider steps increased their BOS.
        - Both responses have been have been reported for healthy adults when encountering these destabilizing terrains.
- Finally, results suggest that mechanical perturbations may be required to elicit underlying differences in foot placement that exist as a result of the disease's asymmetry.
    - Indeed, in people with Parkinson's Disease, the least affected leg may adapt its foot placement to match the contralateral (most affected) to foster symmetrical walking.
    - Therefore, to overcome this compensation, mechanical perturbations may be required in order to reveal functional differences in foot placement.

Restricted Arm Swing in People with Parkinson's Disease Decreases Step Length and Time on Destabilizing Surfaces

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Abstract

Introduction: Fall rates in people with Parkinson's Disease range between 35-68% with the majority of falls occurring while walking. Initial evidence suggests that when walking without arm swing, people with Parkinson's Disease adapt their stepping foot placement as a means to preserve dynamic stability. However, it remains unexamined what arm swing's effect has on dynamic stability when walking on destabilizing surfaces.

Methods: Twenty people with Parkinson's Disease (63.8 ± 9.0 years) walked with restricted and unrestricted arm swing on unperturbed, rocky, rolling-hills, and mediolateral translational surfaces. Data were collected on a split-belt treadmill CAREN Extended-System (Motek Medical, Amsterdam, NL). Bilateral averages and coefficient of variations for step time, length, and width; and mediolateral margin of stability were calculated.

Results: Results were examined in three separate analyses that included arm conditions during each of the destabilizing surfaces compared to unperturbed walking (arm-rolling hills, arm-rocky, and arm-mediolateral). Compared to unrestricted arm swing, restricted arm swing reduced average step length (arm-rolling hills) and time (arm-rocky), and increased COV step time (arm-rolling hills). The arm-rolling hills analysis revealed that the most affected leg had a shorter step length than the least affected. The destabilizing surface effects revealed that during the arm-rolling hills and arm-rocky analyses step time decreased, step width increased, and the COV for step time, length and width increased. No main effects occurred for the arm-mediolateral analysis.

Conclusion: Results indicate that foot placement in response to restricted arm swing, in people with Parkinson's Disease, depends on the encountered destabilizing surface. The arm-rolling hills analysis revealed that participants appropriately reduced step length as compensation to their restricted arm swing. However, the arm-rocky analysis revealed that individuals prioritized

forward progression over dynamic stability as they decreased average step time. Additionally, the increased spatiotemporal variability in response to the rocky and rolling hills conditions indicate partial foot placement adaptation to maintain an already existing level of global dynamic stability as no changes in the Margin of Stability occurred. Adaptation is further corroborated by the decreased step time and increased step width. These responses reflect attempts to pass the destabilizing terrains faster while increasing their base of support.

Introduction

Fall rates in people with Parkinson's Disease range between 35-68% during a twelve-month period, with the majority of falls occurring during walking [11–13]. These numbers are alarming as falling is closely associated with severe medical and socioeconomic consequences which hold lasting impacts on quality of life [11,12].

The dynamical nature of gait increases fall risk. Indeed, during the swing phase, the center of mass (COM) is outside the base of support (BOS) [14]. Thus, the neuromuscular system can only achieve stability by accurately predicting the future position of the COM in order to correctly place the foot at ground contact [14]. Additionally, even during double-support when the COM is within the BOS, stability is challenged as the velocity of the COM is redirected laterally from the unloading to the loading leg [15]. As the stepping foot's placement in both the anteroposterior and mediolateral directions is determined by the COM's trajectory; the neuromuscular system strives to maintain a sinusoidal trajectory while it is volitionally displaced inside and outside the BOS (dynamic stability) [14,15]. Disruptions to this trajectory would either require corrective stepping or result in a fall [14,15]. For instance, in people with Parkinson's Disease, impairments to postural control and gait threatens their ability to maintain the COM along a stable trajectory thereby heightening their fall risk [5,18,19,21,23]. Further, their balance in the mediolateral direction is particularly affected as posturography studies demonstrate that people with Parkinson's Disease have greater trunk sway in this direction than healthy elderly adults [163,164]. This is particularly concerning as impaired mediolateral balance is predictive of falls, closely associated with hip fractures, and increased mortality rates [165,166].

Classically, gait paradigms in people with Parkinson's Disease are based on the inverted pendulum model [28,142]. This model proposes that arm swing has only a minimal impact on COM

motion since the head, arms and trunk are considered a single nearly-rigid body [142]. While arm swing's effect on dynamic stability in healthy adults demonstrates no differences between constrained and unconstrained arm swing, initial evidence suggests that walking without arm swing during unperturbed conditions in people with Parkinson's Disease has a detrimental effect on their dynamic stability [2,30,32,142,167]. Siragy and Nantel found that when arm swing was experimentally constrained, people with Parkinson's Disease increased the distance of their COM's dynamical state (position and velocity) to the edges of their BOS compared to unconstrained arm swing conditions [167]. The authors suggested that this was a compensatory mechanism to mitigate destabilization caused by an observed increase in trunk angular velocity during their constrained arm swing trials [167].

Thus, arm swing may hold additional implications for dynamic stability in people with Parkinson's Disease when ambulating on challenging surfaces that mechanically perturb gait. Destabilizing surfaces such as rocky, rolling hills, and mediolateral translations are common terrains used in virtual reality gait paradigms to simulate everyday real-world surfaces that destabilizes the COM's trajectory during walking [122,123]. Indeed, rocky terrains simulate walking on rocky surfaces, rolling hills simulate a forest trail, and the mediolateral translational simulates walking on a train or a bus [122]. When walking on these terrains, healthy young adults adapt their gait specifically to each surface as each terrain has distinct destabilizing effects on the COM's trajectory due to perturbation type and movement direction [120,122–124]. Determining how individuals with Parkinson's disease ambulate on destabilizing surfaces that emulate real-world terrains will provide a greater depiction of how these individuals walk in their daily lives. Further, as arm swing becomes reduced and ultimately absent with disease progression, determining its effect on dynamic stability in people with Parkinson's Disease while on these surfaces holds direct

implications for fall prevention [36].

Therefore, the purpose of this research is to examine unrestricted and restricted arm swing's effect on dynamic stability in people with Parkinson's Disease when walking on destabilizing surfaces (rocky, rolling hills, and mediolateral translational). We hypothesize that restricted arm swing will be more unstable, compared to unrestricted, for all terrains and that each terrain will be more unstable than steady-state walking. Additionally, we hypothesize an interaction where participants will have a larger step width during restricted arm swing on the destabilizing surface compared to unrestricted arm swing on the same surface. However, no step width difference will occur between unrestricted and restricted arm swing during unperturbed walking conditions.

Methods

Participants

A convenience sample of twenty people with Parkinson's Disease (13 males; 7 females), aged 48 - 79 years old (63.8 ± 9.0 years) were recruited from the Ottawa-Gatineau community. However, as two participants presented severe dyskinesia and two had missing data, only 16 participants were included in the final analysis. Participants were assessed with the original Unified Parkinson's Disease Rating Scale Motor examination (11.0± 6.0) and were between I-III on the Hoehn & Yahr scale [132,168]. Additionally, motor asymmetry to determine the least and most affected sides was determined with the UPDRS and by participants self-reporting laterality. Average disease duration (8.0 ± 5.1 years) and age at onset (56.8 ± 9.6 years) data were collected. Further, seven participants reported freezing of gait based on the Freezing of Gait Questionnaire. Participants were tested on their optimally medicated state. Prior to data collection, volunteers

were excluded if they reported any physical discomfort using a virtual reality system, reported any injuries and/or orthopedic surgeries that could interfere with gait, could walk only with the use of a walking aid, and had any additional illnesses other than Parkinson's Disease. All participants provided written informed consent and the study was approved by both the Hospital and University ethics and scientific committees.

Procedures

Data collection was conducted as part of a larger protocol that examined the effects of unrestricted and restricted arm swing conditions in people with Parkinson's Disease during unperturbed and perturbed walking conditions [167]. This article examines the differences between restricted and unrestricted arm swing conditions during the unperturbed and destabilizing surface trials within this protocol. Participants walked with two arm conditions (restricted and unrestricted) on unperturbed and destabilizing surfaces. Restricted arm swing trials were conducted by participants inserting their arms into their safety harness. This safety harness was worn at all times and attached to an overhead structure to ensure participant safety. The two unperturbed surface trials, one per arm swing condition, lasted 3 minutes each and included steady-state treadmill walking. During these trials, participants walked through a virtual park environment that had its optical flow synchronized to the treadmill. For the destabilizing surface trials, participants completed two walking trials, one per arm swing condition. During the destabilizing surfaces trials, participants also walked through a virtual park environment, which included 3 different destabilizing surfaces (rocky, rolling hills and mediolateral translational) [123]. The rocky terrain was simulated with a pseudorandom perturbation in 3 different axes (vertical, pitch, and roll), rolling hills with an anteroposterior rotational perturbation, and the mediolateral translational was a lateral platform translation [123]. All terrain magnitudes and specifications are depicted in **Table 1**. Terrain

specifications were replicated from previously published protocols examining gait on destabilizing surfaces in individuals with lower limb amputation [123]. The optical flow during destabilizing surface trials was tied to the treadmill. Specifically, the virtual park environment displayed a park terrain where participants walked along a virtual trail. The trail lasted a simulated 200 meters in total and was divided into flat unperturbed segments and the three destabilizing surfaces. Within the 200 meters, participants encountered all three destabilizing surfaces that were presented in a pseudorandom order. Each surface lasted 20 meters and were separated from each other by 40 meters of flat unperturbed walking. When participants encountered one of the destabilizing surfaces, the projection for the park trail changed to mirror the surface condition. For instance, the trail for the rocky surface segment was a visual simulation of rocks appearing on the simulated pathway, the rolling-hills segment was a series of small incline and decline slopes in the pathway, and the mediolateral translation was a projection of a jagged pathway. No visual perturbations nor distortions to the participants' optical field occurred. For all trials, participants walked at preferred walking speeds with both belts running symmetrically. Trial order was randomized per block (unperturbed and destabilizing surfaces). Participants were provided with rest time, both between trials and within the larger protocol, for as long as they desired in order to minimize fatigue.

3D motion analysis was completed using the CAREN-Extended System (Motek Medical, Amsterdam NL) using the virtual park environment setting. This system combines a six degrees-of-freedom motion platform with embedded dual-belt treadmill, 12 camera Vicon motion capture system, 180-degree projector screen for virtual world projection, and a safety harness. Three markers placed in the periphery of the treadmill tracked platform motion along with a 57-marker set for tracking full body kinematics [2,32,123,167].

Data Analysis

Markers and ground reaction forces (GRF) were processed in Vicon Nexus (Nexus 2.6, Oxford, UK), while 3D kinematics and kinetics were calculated in Visual 3D. A 4th-order, low-pass, Butterworth filter with a 12Hz cut-off frequency was used to filter marker data. To remove start-up effects, the first 25 seconds were removed from trials before data analysis. As each destabilizing surface had an average of 20 steps, 20 consecutive steps were taken at random from the unperturbed trials for all analyses. Data were further analyzed in custom Matlab scripts (Mathworks, Natick, MA) to calculate average spatiotemporal parameters (step time, length and width) and dynamic stability measures including the Coefficient of Variation (COV) and Margin of Stability (MOS) using previously reported methods [2,82,95,121,167].

The COV was calculated (standard deviation/average x 100) for step time, length and width. The MOS was calculated bilaterally at both heel strike's and defined as the distance of the Extrapolated Center of Mass (xCOM) to the right/left lateral heel marker.

$$\boldsymbol{MOS = Lateral\ Heel\ Marker - xCOM}$$

The formula for xCOM was:

$$\boldsymbol{xCOM = COMp + (\frac{COMv}{\omega\theta})}$$

Where COM_p = COM's position, COM_v = COM's velocity. ω_Θ was calculated as:

$$\boldsymbol{\omega\theta = \sqrt{g/l}}$$

In this term, g = 9.81m/s^2 and l is the length of the inverted pendulum determined as the average distance of the right/left lateral heel marker to the COM at heel-strikes. Visual 3D was used to calculate the COM's position and velocity. The MOS was only calculated in the mediolateral direction as this metric is only valid in this direction during walking [96]. Visual 3D was used to calculate the COM's position and velocity.

Statistical Analysis

Data were analyzed in SPSS version 26 and $p<0.05$ was considered statistically significant. Normality of variables was verified using Shapiro-Wilks tests. Three separate three-way repeated measures ANOVA were performed to determine the effect of arm swing (restricted and unrestricted), surface (unperturbed and destabilizing surfaces), leg (most and least affected). Walking speed between trials were assessed with a paired samples t-test. If statistical significance was found for preferred walking speed, a General Linear Model with speed as a covariate was performed. Pairwise comparisons with Sidak-Bonferroni corrections were used for post-hoc analyses.

Results

Descriptive statistics for dynamic stability measures (COV and MOS) are reported in **Table 5.2-2a** (Arm-Rolling Hills), **2b** (Arm-Rocky), and **2c** (Arm-ML Translational). Averages for step length, time and width are reported in **Figures 5.2-1, 2, and 3,** respectively. Results are reported bilaterally for least and most affected legs. Average walking speeds for each condition are included in **Table 5.2-3**. Since no difference between arm swing conditions within each terrain were found ($p>0.05$) the two conditions for each terrain were averaged for assessing differences between unperturbed walking and each destabilizing surface.

Rolling Hills

An arm swing main effect occurred whereby restricted arm swing reduced step length ($F(1,15) = 5.86$, $p = 0.029$, $\eta_p^2 = .281$), increased COV step time ($F(1,15) = 5.98$, $p = 0.027$, $\eta_p^2 = 0.285$), and increased Step Time Standard Deviation (SD) ($F(1,15) = 5.00$, $p = 0.041$, $\eta_p^2 = .250$). The ANOVA further revealed a terrain main effect where the rolling hills surface, compared to

unperturbed walking, had a reduced average step time ($F(1,15) = 6.11$, $p = 0.026$, $\eta_p^2 = .289$), increased COV step length ($F(1,15) = 55.62$, $p < 0.001$, $\eta_p^2 = 0.788$), increased Step Length SD ($F(1,15) = 65.98$, $p < 0.001$, $\eta_p^2 = 0.815$), increased COV step time ($F(1,15) = 57.58$, $p < .001$, $\eta_p^2 = .793$), increased Step Time SD ($F(1,15) = 57.89$, $p < .001$, $\eta_p^2 = .794$), increased COV step width ($F(1,15) = 21.49$, $p < 0.001$, $\eta_p^2 = .589$) and increased Step Width SD ($F(1,15) = 29.10$, $p < 0.001$, $\eta_p^2 = .660$). Additionally, a leg main effect was observed where the most affected leg had a reduced step length compared to the least affected ($F(1,15) = 5.42$, $p = 0.034$, $\eta_p^2 = .265$). No further main effects or interactions were found.

Rocky

An interaction for COV step time occurred for arm, leg, and surface ($F(1,15) = 5.29$, $p = 0.036$, $\eta_p^2 = .261$), however, the post-hoc revealed no significant differences. The ANOVA demonstrate an arm swing main effect in that restricted arm swing had a reduced average step time ($F(1,15) = 6.712$, $p = 0.020$, $\eta_p^2 = .309$). Further, the ANOVA revealed a surface main effect in that the rocky surface, compared to unperturbed walking, had a reduced average step time ($F(1,15) = 14.00$, $p = 0.002$, $\eta_p^2 = 0.483$), increased average step width ($F(1,15) = 12.24$, $p = 0.003$, $\eta_p^2 = .449$, increased COV Step Length ($F(1,15) = 52.27$, $p < 0.001$, $\eta_p^2 = 0.777$), increased Step Length SD ($F(1,15) = 92.21$, $p < 0.001$, $\eta_p^2 = .860$), increased COV step time ($F(1,15) = 88.56$, $p < 0.001$, $\eta_p^2 = 0.856$), increased Step Time SD ($F(1,15) = 83.89$, $p < 0.001$, $\eta_p^2 = .848$), increased COV step width $F(1,15) = 16.38$, $p < 0.001$, $\eta_p^2 = 0.522$) and an increased Step Width SD ($F(1,15) = 61.99$, $p < 0.001$, $\eta_p^2 = .805$). No additional main effects or interactions were found.

Mediolateral Translational

As speed was statistically significant between unperturbed walking trials and the

mediolateral translational surface ($t(1,15)= -2.445$, $p = 0.028$), the General Linear Model with speed as a covariate was used for this analysis. This analysis revealed no main effects for arm or terrain ($p>0.05$). Additionally, a leg main effect occurred for average step width ($F(1,15) = 4.97$, $p = 0.045$, $\eta_p^2 = 0.274$), however, the post-hoc comparison was non-significant.

Discussion

This study examined how arm swing (restricted and unrestricted) and destabilizing surfaces (rocky, rolling hills, and mediolateral translational) affected dynamic stability in people with Parkinson's Disease. The results support our hypothesis that restricted arm swing was more destabilizing than unrestricted arm swing and that the rocky and rolling hills surfaces were more destabilizing than unperturbed treadmill walking. However, both dynamic stability measures (COV and MOS) responded uniquely to both arm and surface conditions. For instance, COV metrics increased during both arm and surface conditions yet no changes occurred in the MOS. Interestingly, a leg main effect was observed during the arms-rolling hills analysis where step length was shorter in the most affected leg compared to the least affected. Unexpectedly, no difference occurred between the mediolateral translational surface and unperturbed walking.

Arm Swing

Our results demonstrated that restricted arm swing reduced average step length (arm-rolling hills) and time (arm-rocky) while increasing variability for step time (arm-rolling hills) compared to the unrestricted arm swing condition. Previous research demonstrates that when people with Parkinson's Disease walk with restricted arm swing, compared to unrestricted, trunk angular velocity about the vertical axis increases [167]. During walking, a nominal 1:1 contralateral arm-leg swing controls trunk angular motion about the vertical axis by equalizing torques acting

on the COM [142]. However, when arm swing is restricted, the torques arising from the leg act on the COM unattenuated. This in turn causes the trunk to rotate faster as no counterbalancing torque from the arms are present to mitigate the torques from the legs. In people with Parkinson's Disease, the increased trunk rotation consequently causes the COM's average angular velocity about the vertical axis to increase [167]. This increase in angular velocity subsequently acts as an internal perturbation as it increases spatiotemporal variability in these individuals [167]. As foot placement at heel-strike is based on the COM's predicted trajectory, the increased variability indicates that the COM is moving along a more variable trajectory. In response to this increased velocity, people with Parkinson's Disease adopt a trunk stiffening strategy and adapt the magnitude of their foot placement [167]. Both of which are suggested to be compensatory responses to reduce forward loss of balance during heel-strike when the COM begins to transfer from the loading to the unloading leg [167].

For instance, reductions in step length, during restricted compared to unrestricted arm swing conditions, are suggested to attenuate simultaneous increases in trunk angular velocity about the vertical axis, in lieu of the contralateral arm-leg swing pattern [167]. Indeed, in an examination of step length amplitudes, Huang et al. demonstrated that reductions in step length were accompanied by simultaneous reductions in spinal rotation amplitudes [169]. As such, the reduced step length in our study appears to be a strategy to control internal destabilization arising from excessive trunk rotation when arm swing is restricted. Contrastingly, the increased COV step time in the arm-rolling hills analysis could be more indicative of motor impairment than adaptation. Indeed, in people with Parkinson's Disease, neurodegeneration in the Basal Ganglia impairs rhythmic internal movement timing [10]. Similar findings were reported by Siragy and Nantel in an assessment of restricted arm swing and unperturbed walking in this demographic [167]. The authors

proposed that the increase in trunk angular velocity about the vertical axis, during the restricted arm swing condition, disrupts the rhythmic temporal sequence of foot placement [167]. This holds direct implications for clinicians as COV step time is a strong predictor of falls in this demographic [5,83]. Specifically, as arm swing becomes completely absent as the disease progresses, the internal timing of foot placement during walking is disrupted by both the increases in trunk rotation as well as due to the continued neurodegeneration of the disease. Based on this evidence, future research should consider examining the effect of mechanically restoring the contralateral arm-leg swing pattern in people with Parkinson's Disease as an intervention to facilitate rhythmic gait.

Interestingly, the arms-rolling hills analysis further revealed that the most affected leg had a reduced average step length compared to the least affected leg. In Parkinson's Disease, one of the primary symptoms of the disease is the reduced movement amplitude (hypokinesia) that occurs when individuals are performing a motor task [10,39,40,170]. In gait, one of the main manifestations of hypokinesia is the reduced step length exhibited by this demographic compared to healthy elderly adults [10,170]. However, in the early to moderate stages of the disease, the neurodegeneration is often asymmetric, thereby causing the symptomatology to be expressed more in one limb than the other [55,171]. Thus, step length in people with Parkinson's Disease is not only reduced in amplitude but is also asymmetric compared to healthy elderly adults [18,19]. However, it is interesting to note that the step length interlimb differences in our study only occurred during the arms-rolling hills analysis. This is likely due to the distinct perturbing effects that different destabilizing surfaces have on gait [120,122–124]. In our study, the main destabilization that arose from the rolling hills surface acted in the anteroposterior direction. As such, differences between limbs in anteroposterior foot placement may have only been elicited when a mechanical perturbation acts in the same direction.

Similarly, the current findings also suggest that the effect of restricted arm swing on our

participants varied based on the specific terrain encountered. Indeed, unlike the results from the arm-rolling hills analysis, the arm-rocky analysis revealed a reduction in average step time for the restricted arm swing condition. This finding was unexpected as individuals typically increase step time when dynamic stability is threatened as part of the "cautious gait" strategy [172]. However, during unperturbed walking, people with Parkinson's Disease increase their cadence, relative to healthy aged matched adults, as a means to maintain forward progression [170]. In our study, the faster step time during the arm-rocky analysis may have resulted from our participants incorrectly prioritizing their forward progression over postural stability. This would be in line with previous research demonstrating incorrect task prioritization during walking in this demographic [151].

Destabilizing Surfaces

In line with our hypothesis, spatiotemporal variability increased, compared to unperturbed walking, when our participants walked on the rocky and rolling hills destabilizing surfaces. However, in contrast to our hypothesis, no differences were found when participants walked on the mediolateral surface compared to unperturbed walking.

Increased spatiotemporal variability was in line with our hypothesis as the destabilizing surfaces mechanically perturbed our participants' gait. However, the COV only measures the magnitude of variability. And while it is a strong predictor of falls in this demographic, it is unable to parse out differences in adaptation versus motor impairment [5,83,167,173]. Indeed, McAndrew et al. discussed that increases in variability can reflect either correct or incorrect foot placement adaptation in response to destabilizing surfaces [120]. The authors further went on to discuss that no increases in variability, in response to destabilizing surfaces, would reflect an incorrect response as the neuromuscular system is not correcting foot placement to adapt the base of support at heel-

strike to the encountered surface [120]. By adapting foot placement, individuals would be accounting for their COM's destabilized trajectory to ensure a stable transfer of the COM between the legs [2,5,10,83]. As our sample had a moderate disease progression, it is unlikely that all of the increases in variability were due to incorrect foot placement responses. Rather, increases in spatiotemporal variability may also reflect our participants correctly predicting the destabilization of their COM and appropriately adapting. This would account for lack of findings in the mediolateral-MOS during all the destabilizing surfaces as our participants modified their foot placement to maintain their already existing level of global dynamic stability [2,98]. The ability to adapt to destabilizing surfaces, may be linked to partially intact proprioception in people with Parkinson's Disease [26]. Indeed, proprioceptive evidence demonstrates that these individuals are still capable of recognizing changes in limb movement and position, but require larger displacement magnitudes, compared to healthy aged matched adults, to do so [26]. Thus, it is plausible that the destabilizing surfaces occurred at a magnitude large enough where participants could perceive the destabilization and, at least partially, adapt their foot placement. Further support for this theory is demonstrated by the decreased bilateral average step time, on both the rolling hills and rocky surfaces, as well as the increased average step width on the rocky surface. Current evidence demonstrates that both healthy adults and individuals with lower limb amputation increase walking speed, which would reduce step time, to step off destabilizing surfaces faster and increase step width to widen their mediolateral BOS when traversing destabilizing surfaces [122,123]. Therefore, by our participants executing the correct responses, it indicates that adaptation to destabilizing surfaces is at least partially intact in people with Parkinson's Disease with moderate disease progression. Alternatively, a partially impaired proprioceptive sense may account for the lack of differences between the mediolateral surface and unperturbed walking. Out of all three

destabilizing surfaces, the mediolateral surface is the most similar to unperturbed treadmill walking and arguably the least destabilizing condition as the platform only laterally translates. Thus, despite humans being inherently unstable in the mediolateral direction, a larger destabilizing amplitude may have been necessary in order to elicit adaptive stepping responses [15,120,144,174].

Our results hold an important implication for clinicians and researchers to consider. Specifically, therapies should carefully consider terrain type and amplitude when fostering gait adaptation. Particular consideration should be given to the role of variability in reflecting motor adaptation or impairment. Furthermore, clinicians and researchers should carefully consider stage of disease progression when interpreting the COV for fall risk assessment in this demographic.

Limitations

Several limitations should be considered when interpreting our results. For instance, participants were tested on their optimally medicated state which is demonstrated to affect spatiotemporal variability. Moreover, differences between freezers and non-freezers were not examined. As freezers demonstrate greater postural instability than non-freezers, absent arm swing may have a distinct effect on their dynamic stability. Future research should also consider examining differences between people with Parkinson's Disease and healthy aged matched controls to parse out differences due to age and those that arise due to Parkinson's Disease. Finally, due to our limited sample size, we may have lacked the statistical power to detect all possible interactions of destabilizing surfaces, arm swing conditions and leg asymmetry.

Conclusion

The current findings on arm swing in all the different analyses suggest that responses to absent arm swing in people with Parkinson's Disease vary depending on the specific terrain

encountered. This is expected as evidence examining gait on destabilizing surfaces indicates that compensatory responses are unique to each type of terrain. This in turn would affect how people with Parkinson's Disease respond to absent arm swing on these surfaces. Therefore, rehabilitation therapies should provide diverse environment types that reflect real-world terrains over focused repetitive tasks to foster adaptation to absent arm swing. Further, clinicians should consider interventions that strive to restore the contralateral arm-leg swing pattern. However, based on results in young healthy adults considering which modality to use for restoring arm swing is crucial as explicitly directing individuals to increase arm swing taxes attentional resources and may hold negative consequences on gait in individuals with Parkinson's disease [2]. Finally, our results from destabilizing surfaces demonstrate that spatiotemporal variability increases in people with Parkinson's Disease when walking on destabilizing surfaces. While increased variability is a strong predictor of falls, no difference in the MOS suggests that our participants adapted lower limb placement to preserve their already existing global dynamic stability. Adaptation in people with Parkinson's Disease is further corroborated by the reduced average step time and increased step width during the rolling hills and rocky surfaces as these findings mirror responses seen in healthy young adults.

Author Contribution

Julie Nantel conceptualized and organized the research project; **2)** Data Analysis: A. Main Analysis (Tarique Siragy), B. Secondary analysis (Mary-Elise Macdonald) B. Review and Critique (JN, MEM) , **3)** Statistical Analysis: A. Design (TS and JN), B. Execution (TS), C. Review and Critique (JN, MEM); **4)** Manuscript: A. Writing of the first draft (TS), B. Review and Critique (JN, MEM).

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Conflict of Interest

We declare no conflict of interests.

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Table 5.2-1: Destabilizing Surfaces descriptions within the CAREN-Extended System.

Terrain	Description
Rocky	The CAREN Rumble module causes the platform to oscillate simultaneously in three directions. There was a maximum range of ±2 cm at 0.6Hz vertically, ±1 degree at 1Hz pitch, and ±1 degree at 1.2Hz roll [123].
Rolling-Hills	In the AP direction, the platform oscillates based on a sum of four sines at 0.16, 0.21, 0.24 and 0.49Hz. The maximum range was ±3 degrees based on an amplitude scaling of $A_w = 0.01$ [123].
Mediolateral Translational	The platform mediolaterally oscillates based on a sum of four sines at 0.16, 0.21, 0.24 and 0.49Hz. The maximum range was ±4 cm based on amplitude scaling of $A_w = 0.015$ [123].

Table 5.2-2a. Dynamic Stability Measures for arm (unrestricted and restricted) and terrain (steady-state and rolling hills) conditions: Margin of Stability and Coefficient of Variation for Step Time, Length and Width along with spatiotemporal averages. Values are reported for the least and most affected legs. P-values from the three-way repeated measures ANOVA are reported for arm swing main effects.

		Unrestricted		Restricted		
	Leg	Steady-State	Rolling Hills	Steady-State	Rolling Hills	P-value
Margin of Stability (cm)	Least	11.30±1.1	11.46±1.5	11.80±1.6	11.39±1.9	.086
	Most	10.96±1.8	11.02±2.4	11.36±2.0	11.62±2.4	
COV Step Length*	Least	3.02±1.1	7.09±2.1	4.41±2.7	7.61±2.2	.169
	Most	3.79±1.4	7.04±3.2	4.18±2.2	7.01±1.8	
SD Step Length (cm)*	Least	1.5±0.4	3.6±1.2	2.0±1.2	3.8±1.2	.295
	Most	1.8±0.5	3.4±1.7	2.0±1.0	3.3±0.8	
COV Step Time*†	Least	2.56±0.5	5.71±2.4	2.86±1.5	6.45±2.3	.027
	Most	2.37±0.6	6.12±2.0	3.26±2.0	7.00±2.4	
SD Step Time (ms)*†	Least	14±3	30±12	16±8	33±9	.041
	Most	13±3	33±11	18±12	36±11	
COV Step Width*	Least	8.75±3.2	11.16±3.5	7.21±3.4	10.41±4.7	.156
	Most	7.91±2.7	10.79±3.9	7.31±2.9	10.59±4.1	
SD Step Width (cm)*	Least	1.6±0.6	2.1±0.6	1.3±0.5	2.0±0.7	.171
	Most	1.4±0.5	2.0±0.6	1.3±0.3	2.1±1.0	

† Arm Swing Main Effects at $p<0.05$

* Terrain Main Effects at $p<0.05$

Table 5.2-2b: Dynamic Stability Measures for arm (unrestricted and restricted) and terrain (steady-state and rocky) conditions: Margin of Stability and Coefficient of Variation and Standard Deviation for Step Time, Length and Width. Values are reported for the least and most affected legs. P-values from the three-way repeated measures ANOVA are reported for arm swing main effects.

		Unrestricted		Restricted		
	Leg	Steady-State	Rocky	Steady-State	Rocky	P-value
Margin of Stability (cm)	Least	11.30±1.1	11.79±1.5	11.80±1.6	11.66±1.2	.127
	Most	10.96±1.8	10.64±2.4	11.36±2.0	11.22±1.9	
COV Step Length*	Least	3.02±1.1	7.59±2.8	4.41±2.7	7.63±3.4	.613
	Most	3.79±1.4	8.15±3.0	4.18±2.2	7.44±2.0	
SD Step Length (cm)*	Least	1.5±0.4	3.6±1.1	2.0±1.2	3.6±1.4	.687
	Most	1.8±0.5	3.8±1.2	2.0±1.0	3.5±0.8	
COV Step Time*	Least	2.56±0.51	6.26±2.0	2.86±1.5	6.80±1.8	.457
	Most	2.37±0.56	6.82±2.4	3.26±2.0	6.19±1.9	
SD Step Time (ms)*	Least	14±3	33±10	16±8	34±8	.625
	Most	13±3	36±12	18±12	32±10	
COV Step Width*	Least	8.75±3.2	15.63±9.9	7.21±3.4	13.73±5.5	.095
	Most	7.91±2.7	14.93±8.5	7.31±2.9	14.25±7.3	
SD Step Width (cm)*	Least	1.6±0.6	2.9±0.9	1.3±0.5	2.8±0.8	.207
	Most	1.4±0.5	2.9±1.0	1.3±0.3	2.8±0.8	

† Arm Swing Main Effects at $p<0.05$

* Terrain Main Effects at $p<0.05$

Table 5.2-2c: Dynamic Stability Measures for arm (unrestricted and restricted) and terrain (steady-state and mediolateral translational) conditions: Margin of Stability and Coefficient of Variation for Step Time, Length and Width along with spatiotemporal averages. Values are reported for the least and most affected legs. P-values from the three-way repeated measures ANOVA are reported for arm swing main effects.

		Unrestricted		Restricted		
	Leg	Steady-State	Mediolateral	Steady-State	Mediolateral	P-value
Margin of Stability (cm)	Least	11.23±1.2	11.03±1.5	11.78±1.7	11.36±1.8	.442
	Most	10.83±1.8	10.81±2.2	11.20±2.0	11.05±2.1	
COV Step Length	Least	3.02±1.1	4.46±1.6	4.41±2.7	5.11±1.9	.492
	Most	3.79±1.4	4.82±1.5	4.18±2.2	5.26±1.7	
SD Step Length (cm)	Least	1.5±0.4	2.3±0.9	2.0±1.2	2.6±0.9	.451
	Most	1.8±0.5	2.4±0.7	2.0±1.0	2.6±0.9	
COV Step Time	Least	2.56±0.51	3.94±2.2	2.86±1.5	3.78±2.5	.928
	Most	2.37±0.56	4.05±1.6	3.26±2.0	4.29±2.0	
SD Step Time (ms)	Least	14±3	21±11	16±8	20±13	.992
	Most	13±3	22±8	18±12	23±9	
COV Step Width	Least	8.75±3.2	16.27±7.5	7.21±3.4	15.06±8.4	.638
	Most	7.91±2.7	16.29±6.4	7.31±2.9	15.97±9.7	
SD Step Width (cm)	Least	1.6±0.6	2.9±1.0	1.3±0.5	2.7±1.0	.809
	Most	1.4±0.5	2.9±0.8	1.3±0.3	2.9±1.3	

† Arm Swing Main Effects at $p<0.05$
* Terrain Main Effects at $p<0.05$

Table 5.2-3: Average walking speed and standard deviation for each walking trial for arm swing conditions and destabilizing surfaces.

Condition	Average Walking Speed (m/s)
Unperturbed unrestricted arm swing	0.99 ± 0.16
Unperturbed restricted arm swing	0.96 ± 0.17
Rolling Hills unrestricted arm swing	1.03 ± 0.13
Rolling Hills restricted arm swing	1.03 ± 0.14
Rocky unrestricted arm swing	1.04 ± 0.14
Rocky restricted arm swing	1.03 ± 0.14
Mediolateral Translational unrestricted arm swing	1.04 ± 0.14
Mediolateral Translational restricted arm swing	1.05 ± 0.14

Figure 5.2-1. Average step length with respect to terrain conditions for most (MA) and least affected (LA) leg in both restricted and unrestricted arm swing conditions. † represents a significantly larger value in the unrestricted arm swing condition than restricted within the same leg and terrain condition $p < 0.05$. * represents a significantly larger value in LA than MA within the same terrain at $p < .05$.

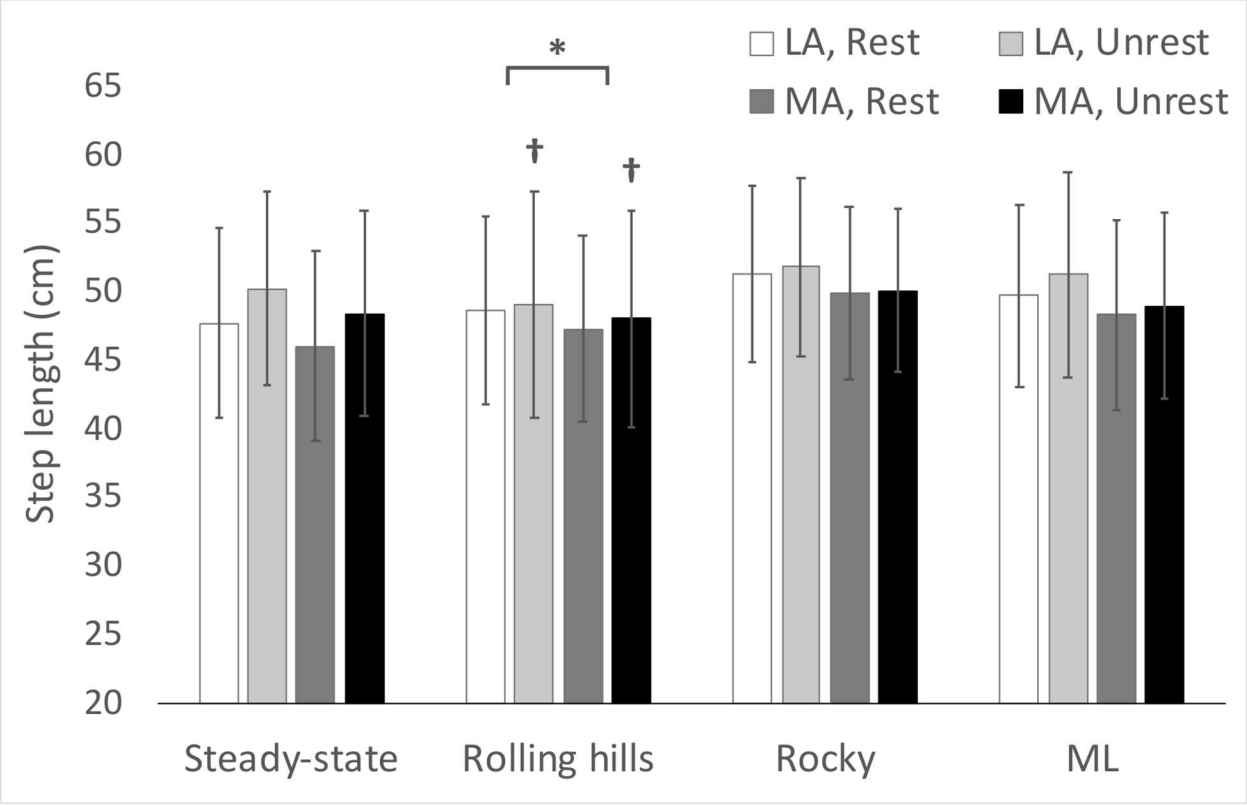


Figure 5.2-2. Average step time with respect to terrain conditions for most (MA) and least affected (LA) leg in both restricted and unrestricted arm swing conditions. † represents significantly larger value in the unrestricted arm swing condition than restricted within the same leg and terrain condition $p < 0.05$. ‡‡‡ represents a significant difference between terrain conditions at $p < .05$.

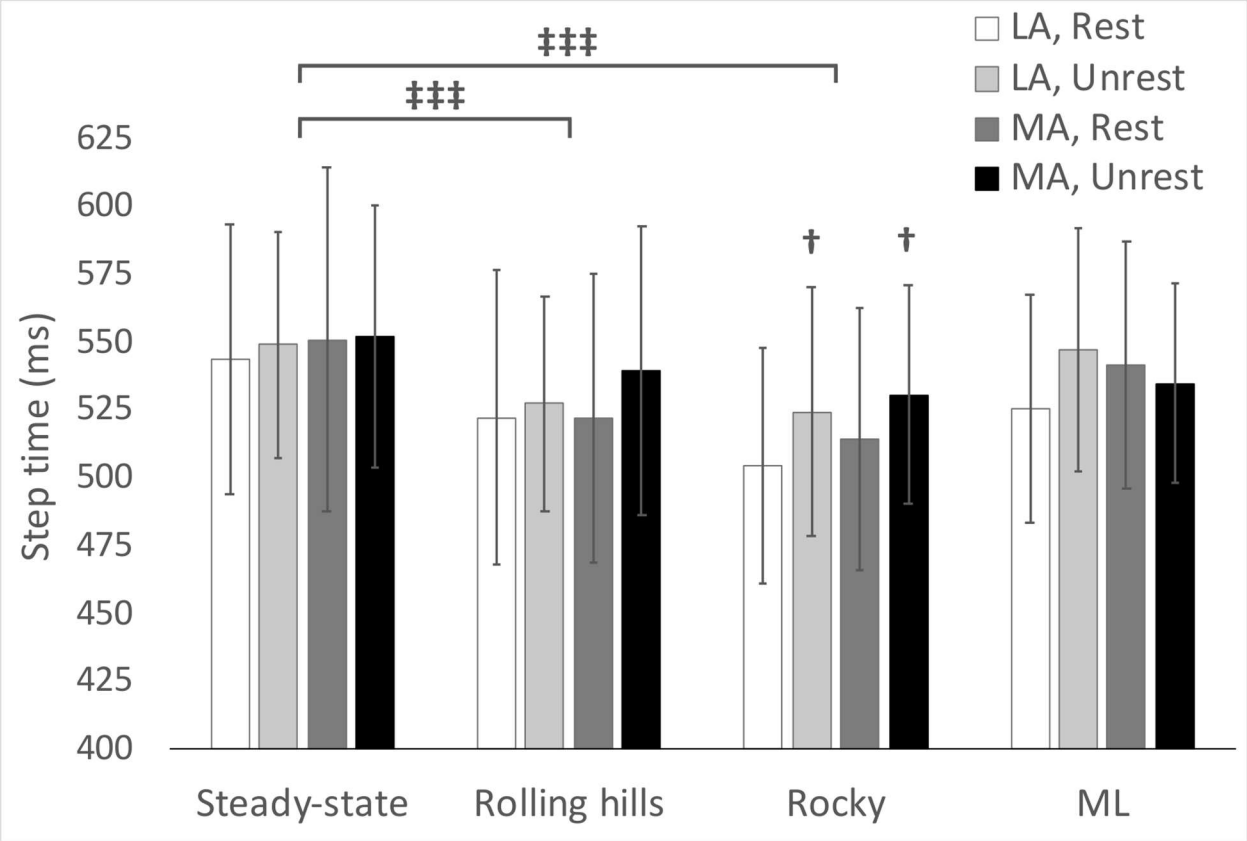
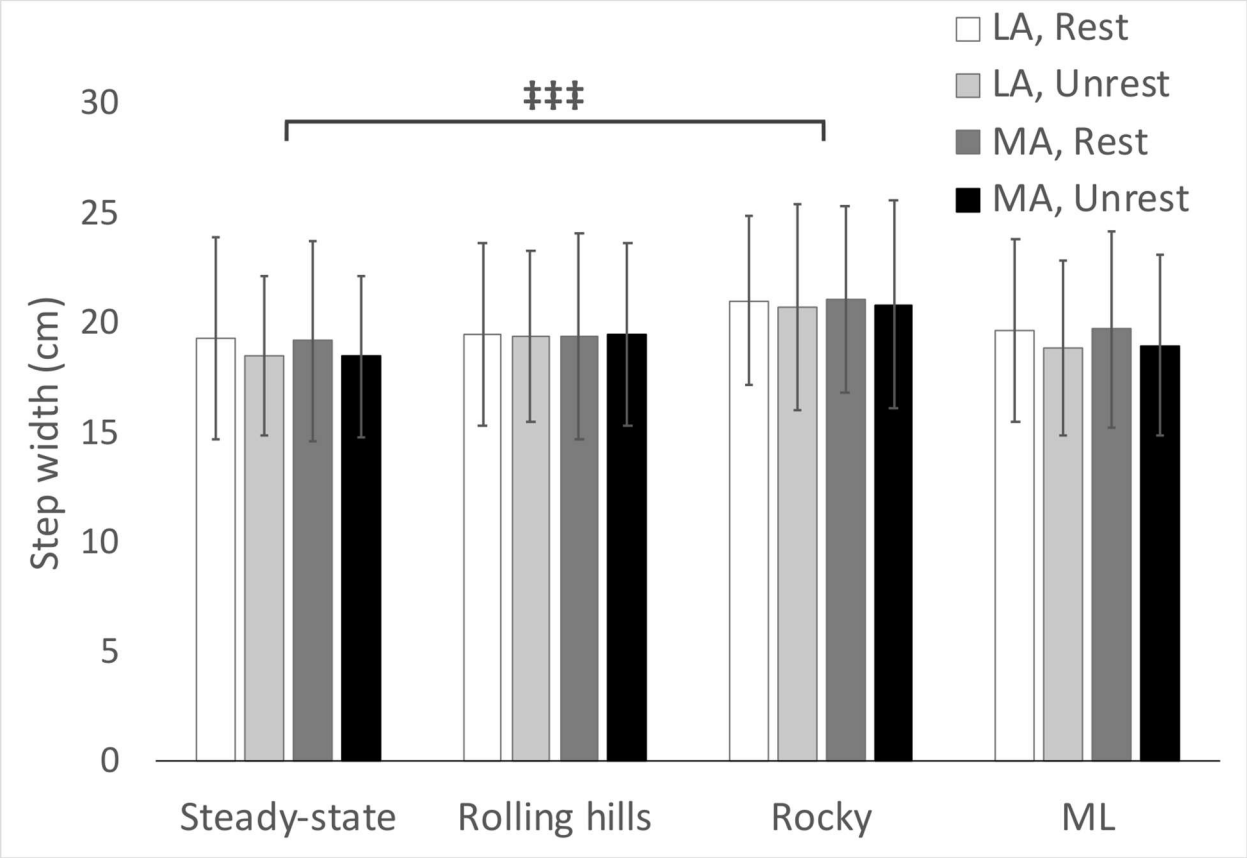


Figure 5.2-3. Average step width with respect to terrain conditions for most (MA) and least affected (LA) leg in both restricted and unrestricted arm swing conditions. ‡‡‡ represents a significant difference between terrain conditions at $p < .05$.



Supplementary Material

Table 1: Average range of motion (degrees) for restricted and unrestricted arm swing during each terrain condition. Data were combined for the least and most affected sides (per condition) as no statistical differences existed between them. Data were analyzed with a within-subjects repeated measures ANOVA.

	Terrain	Unrestricted	Restricted	P-value
Range of Motion°*	Steady-State*	20.59±12.3	3.94±4.7	<.001
	Rolling Hills*	21.33±13.0	3.39±1.4	
	Rocky*	21.62±11.1	3.54±1.5	
	Mediolateral*	21.70±11.2	3.43±1.6	

* Arm ROM differences at $p<0.05$

5.3 Article 3

Background Summary:

- Research from static postural perturbations demonstrate that people with Parkinson's Disease are particularly susceptible to backward stability loss.
    - This may increase their risk of falling when encountering slips which act in the same backward direction.
    - This is concerning as slips are one of the primary contributors to falling in individuals over the age of 65.
- However, while gait impairments that contribute to falls in people with Parkinson's Disease are well documented during steady-state conditions, evidence during perturbed walking conditions is non-existent.
- As part of the whole-body response to slips, individuals engage the arm elevation strategy. to move the Center of Mass in the forward and opposite direction to the backward displacement induced by the slip.
- However, the role of arm motion in response to slips in people with Parkinson's Disease is currently lacking.
    - As such, it remains unclear how arm motion contributes to stability in people with Parkinson's Disease when encountering slips.
- Additionally, one of the characteristic aspects of idiopathic Parkinson's Disease is the asymmetric neurodegeneration that affects mobility in one limb (most affected) to a greater extent than the other (least affected).
    - Evidence from steady-state walking proposed that the least affected leg is more capable of adjusting foot placement at ground contact than the most affected.

- Thus, the ability to execute recovery responses to slips, may equally be asymmetrically impaired in people with Parkinson's Disease.
- The role of both of these limbs in response to slips in people with Parkinson's Disease, however, has not been determined.

Summary of Findings:

- Unexpectedly, no differences occurred between the unrestricted and restricted arm swing conditions on neither dynamic stability metrics nor trunk kinematics (position and velocity).
  - Lack of differences between arm conditions raised the possibility that the unrestricted arm swing condition was not more effective, than the restricted, in re-establishing dynamic stability after a slip.
- This finding was surprising as the Arm Elevation Strategy is an integral component in restoring stability following a slip in both healthy young and elderly adults.
- Research on static postural perturbations found that in people with Parkinson's Disease, the arm recovery responses had a more variable trajectory and were engaged for a shorter time span compared to health aged-matched adults.
  - Further, the recovery arm responses in people with Parkinson's Disease were particularly ineffective and abnormal during perturbations in the backward direction (same acting direction as slips).
  - This may account for the lack of differences between the unrestricted and restricted arm swing conditions in our study.
- However, in line with the findings observed in the first two studies of this thesis (**Article 1 & 2**), the restricted arm swing condition appeared to be perturbing to step timing control in this demographic.

  - Indeed, in this study (**Article 3**), the restricted arm swing condition caused a reduction in average step time for the slipped leg.
  - This faster step, during the restricted arm swing condition, would provide less time for the neuromuscular system to determine optimal foot placement for the following (contralateral) recovery step.

- In contrast to the arm swing results, more differences in foot placement and dynamic stability metrics were observed between the least and most affected legs.
  - Specifically, the least affected leg had both a longer step time and larger anteroposterior MOS compared to the most affected side.
    - The longer step time is a beneficial response to slips as it provides more time to determine optimal foot placement for subsequent steps.
    - Additionally, the larger Anteroposterior MOS would reduce the risk of backward stability loss by moving the Extrapolated COM in the forward and opposite direction of the slip.
    - Therefore, the findings indicate that, due to the milder symptoms on the least affected leg, this limb is more effective in executing appropriate slip recovery responses.

Recovery of Dynamic Stability during Slips Unaffected by Arm Swing in People with Parkinson's Disease

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Abstract

The arm elevation strategy assists in recovering stability during slips in healthy young and elderly individuals. However, in people with Parkinson's Disease, one of the main motor symptoms affecting the upper limbs is reduced arm swing which intensifies throughout the course of the disease before becoming absent. This holds direct implications for these individuals when encountering slips as the arm elevation strategy is an integral component in the interlimb slip response to restore stability. Arm swing's effect in recovering from slips in people with Parkinson's Disease though remains unexamined. Twenty people with Parkinson's Disease (63.8 ± 9.0 years) walked with restricted and unrestricted arm swing conditions on a dual-belt treadmill where slips were induced on the least and most affected sides. Data were collected on the CAREN Extended System (Motek Medical, Amsterdam, NL). The Margin of Stability, linear and angular trunk velocities, as well as step length, time, and width were calculated. Data were examined during the slipped step and recovery step. The restricted arm swing condition, compared to unrestricted, caused a faster step time during the slipped step. Compared to the most affected leg, the least affected had a wider step width during the slipped step. During the recovery step, the least affected leg had a larger anteroposterior Margin of Stability and longer step time than the most affected. No differences between our arm swing conditions suggests that the normal arm swing in our participants was not more effective at restoring stability after an induced slip compared to when their arm motion was restricted. This may be due to the arm elevation strategy being ineffective in counteracting the slip's backward destabilization in these individuals. Differences between the legs revealed that our participants were asymmetrically impaired in their slip recovery response.

1 Introduction

One of the primary concerns for people with Parkinson's Disease is the increased risk of falling during walking [1–5] . Research from static postural perturbations demonstrate that people with Parkinson's Disease are particularly susceptible to backward stability loss, which may increase their risk of falling when encountering slips [6]. This is concerning as slips are one of the primary contributors to falling in individuals over the age of 65 [7–9]. However, while gait impairments that contribute to falls in people with Parkinson's Disease are well documented during unperturbed conditions, evidence during perturbed walking conditions is sparse [10–15] .

When first encountering an unexpected slip, healthy young and elderly adults rely on rapid reactive recovery responses to mitigate the destabilizing effects of the perturbation [16–21]. For instance, when the perturbed leg contacts the slipping surface, the knee and ankle joints flex to lower the Center of Mass (COM) to the ground [17,18]. However, while this flexing response is a strategy to increase stability, extension occurs in the ipsilateral hip due to the hamstring contraction which flexes the knee [18]. This subsequently causes trunk extension and causes the COM to further fall posteriorly (backward) [18]. Thus, to dissipate the COM's posterior motion, the arms flex forward and upward in an arm elevation strategy to move the COM in the opposing anterior direction [16–18]. Meanwhile, in the unperturbed (recovery) limb, the hip and knee extend to bring the leg that is entering the swing phase rapidly to the ground to increase the base of support (BOS) [16]. After slip exposure, individuals walk with a "cautious gait" strategy where step time and width are increased, and step length is reduced [20,22,23]. This is accompanied by a shift in the dynamical state (position and velocity) of their COM anteriorly and away from the mediolateral edges of their base of support at heel-strike [10,18,20].

However, in people with Parkinson's Disease, the multi-system neurodegeneration that occurs, impedes the ability to execute appropriate responses to sudden environmental changes [6]. For instance, evidence from static postural perturbations demonstrates that people with Parkinson's Disease execute responses that are slower, reduced in amplitude, and have an incorrect directionality with respect to where the perturbation is acting [6]. Additionally, one of the characteristic aspects of idiopathic Parkinson's Disease is the asymmetric neurodegeneration that affects mobility in one limb (most affected) to a greater extent than the other (least affected) [24,25]. Evidence from steady-state walking proposed that the least affected leg is more capable of adjusting foot placement at ground contact than the most affected [15]. Thus, the ability to execute recovery responses to slips, may equally be asymmetrically impaired in people with Parkinson's Disease. The role of both of these limbs in response to slips in people with Parkinson's Disease, however, has not been determined.

Additionally, the role of arm swing in response to slips in this demographic is currently lacking. In people with Parkinson's Disease, one of the main motor symptoms affecting the upper limbs is reduced arm swing which intensifies throughout the course of the disease before becoming completely and bilaterally absent [26,27]. This holds direct implications for these individuals when encountering slips as the arm elevation strategy is an integral component in the interlimb slip response to restore stability [16–19]. However, previous research on restricted arm swing on healthy adults suggests that when the arms are restricted, the upper extremity's mass is concentrated around the trunk [28]. This subsequently is suggested to increase trunk inertia which in turn increases resistance to changes in COM motion induced by a perturbation [28]. As such, it remains unclear how arm swing contributes to stability in people with Parkinson's Disease during slips.

Therefore, this study's purpose examines differences between the least and most affected legs to slips in people with Parkinson's Disease with unrestricted and restricted arm swing conditions. We hypothesize that the least affected leg, compared to the most affected, will display a "cautious gait" strategy and increase the distance of the COM's dynamical state to the edges of the BOS in the mediolateral and anteroposterior directions. We further predict that trunk angular velocities, at heel-strike, will be larger during the unrestricted arm swing condition (the regular arm motion in people with Parkinson's Disease) compared to the restricted condition. Finally, we hypothesize an interaction where the distance between the COM's dynamical state to the BOS will be larger at heel-strike of the least affected leg during the unrestricted arm swing condition compared to the heel-strike of the most affected leg when paired with restricted arm swing.

2 Methods

2.1 Participants

A convenience sample of twenty people with Parkinson's Disease (13 males and 7 females), ages 48-79 years (63.8 ± 9.0 years) were recruited from the Ottawa-Gatineau community. Since two individuals had severe dyskinesia and one individual had missing data, only 17 participants were included in the final analysis. Participants were assessed with the original Unified Parkinson's Disease Rating Scale Motor Examination (11.0± 6.0) and were between I-III on the Hoehn & Yahr scale [29]. Further, seven participants reported freezing of gait based on the original Freezing of Gait Questionnaire. Participants were tested on their optimally medicated state. Exclusion criteria included any physical discomfort using a virtual reality system, any injuries and/or orthopedic surgeries that interfered with gait, walking only with a walking aid, and additional illnesses other than Parkinson's Disease. All participants provided written informed

consent and the study was approved by the Ottawa Health Science Network Research Ethics Board and the University of Ottawa Research Ethics Board. The study was conducted in accordance with the Tri-Council Policy statement; Ethical Conduct for Research Involving Humans; The International Conference on Harmonization- Good Clinical practice: Consolidated Guideline; and the provisions of the Personal Health Information Protection Act 2004.

2.2 Procedure

A familiarization period was provided for participants prior to data collection to determine their preferred walking speed on the treadmill. Experimental setup is depicted in **Figure 1**. Participants walked at their preferred speed on a dual-belt treadmill with two arm swing conditions (restricted and unrestricted). During the restricted arm swing trials, participants inserted their arms inside the safety harness, which effectively prevented arm motion. The unrestricted arm swing trials were participants' normal arm swing movements. Arm swing conditions were paired with a slip that was induced on either the least or the most affected side (one slip per trial). This resulted in a total of four trials per participant with each trial lasting 2 minutes. Slips were caused by accelerating the left/right treadmill belt at $1.7 m/s^2$ for 0.75 seconds before decelerating at the same rate for 0.75 seconds to return to the participants preferred walking speeds [30]. Slips were automatically triggered, at heel-strike, after the participants' heels crossed each other during the swing phase in the anteroposterior direction [30]. Twenty-five seconds were provided at the start of each trial for participants to reach steady-state walking before data collection began. To minimize slip anticipation, trials were randomized using a random number generator in Excel 2016 (Microsoft, Seattle, WA, USA) and slip onset within each trial was manually determined in a pseudorandom manner by the CAREN system operator. To ensure safety, and prevent falls,

participants wore a safety harness attached to an overhead structure at all times. Participants were encouraged to rest whenever necessary to minimize fatigue.

Data collection was completed using the CAREN-Extended System (Motek Medical, Amsterdam NL) in a virtual park terrain environment. This system combines a six degrees of freedom motion platform with embedded dual-belt instrumented treadmill, 12 camera Vicon Motion Capture system, 180-degree projector screen, and a safety harness. A 57-marker set was used for tracking full body kinematics [15,31,32]. Kinematic data were collected at 100Hz.

2.3 Data Analysis

Markers were synchronized and processed in Vicon Nexus (nexus 2.6, Oxford, UK), while 3D kinematics were calculated in OpenSim with a full-body model [33]. A fourth-order, low-pass Butterworth filter with a 12Hz cutoff frequency was used to filter marker data. Data were analyzed by custom Matlab scripts (MathWorks, Natick, MA) to calculate step time, length, and width, as well as instantaneous linear and angular velocities at heel-strike for both feet. Linear velocities were calculated as absolute values. Additionally, the Margin of Stability (MOS) was calculated at bilateral heel-strikes as a measure of dynamic stability in the anteroposterior and mediolateral directions [34-36]. In the anteroposterior direction, this was calculated as the distance of the extrapolated COM to the right/left heel maker and in the mediolateral direction to the right/left lateral heel marker at heel-strike [15,31,36]. Data were categorized by least and most affected legs which were then divided into slipped and recovery limbs. The slipped limb was defined as the leg which was perturbed upon heel-strike, while the recovery limb was defined as the contralateral limb that performed the first heel-strike following the slip.

2.4 Statistical Analysis

Data were analyzed using SPSS 23.0 (SPSS, IBM, Chicago IL), and $p < 0.05$ was considered statistically significant. An alpha = 0.05 was set a prioi for statistical significance. The normality of variables was verified using Shapiro-Wilk's test. A two-way repeated measures ANOVA for arm (unrestricted and restricted) and leg (least and most affected leg) was performed to find main effects and interactions for the slipped limb. Additionally, an identical but separate two-way repeated measures ANOVA assessed main effects and interactions for the recovery limb. If statistical significance was achieved, then pairwise comparisons with a Sidak-Bonferroni adjustment for multiple comparisons were used for post-hoc analyses.

3 Results

Slipped limb data for the MOS and spatiotemporal variables are in **Table 1** while linear and angular velocities are in **Table 2**. Recovery limb data for the MOS and spatiotemporal data are in **Table 3** while velocity data are in **Table 4**. Arm range of motion for both arm swing conditions are presented in **Table 5**. Average walking speeds for conditions are included in the **Supplementary Material**. No differences existed between the legs ($p > 0.05$) for walking speed. Thus, both sides were averaged before assessing walking speeds between both arm swing conditions with a paired t-test. Results from the t-test revealed no differences in walking speed between the unrestricted and restricted arm swing conditions ($p > 0.05$).

Table 5.3-1: Slipped Leg average Step Time, Length and Width with Anteroposterior and Mediolateral Margin of Stability during unrestricted (unrest.) and restricted (rest.) arm swing and for the least and most affected legs.

				P-value	
	Arm	Least Affected	Most Affected	Leg	Arm
AP Margin of Stability(cm)	Unrest.	-20.31±7.12	-20.23±4.66	.503	.336
	Rest.	-18.57±7.00	-20.26±7.06		
ML Margin of Stability(cm)	Unrest.	11.57±3.05	11.00±2.27	.252	.372
	Rest.	11.73±1.69	11.54±2.39		
Step Time(ms)*	Unrest.	436±87	435±44	.691	.006
	Rest.	412±66	426±51		
Step Length(cm)	Unrest.	50.51±11.24	50.58±8.34	.547	.755
	Rest.	49.21±10.53	51.10±8.83		
Step Width(cm)†	Unrest.	26.79±4.73	25.75±5.04	.005	.391
	Rest.	27.65±4.30	26.00±4.46		

† Leg Main Effects at $p<0.05$

* Arm Main Effects at $p<0.05$

Table 5.3-2: Slipped Leg trunk instantaneous linear and angular velocities in all 3 axes for arm (unrestricted and restricted) and leg (least affected and most affected). Values were rounded to the nearest whole number. P-values from the two-way repeated measures ANOVA are reported for leg main effects.

		Unrestricted Arm		Restricted Arm		
		Least Affected	Most Affected	Least Affected	Most Affected	P-Value
Linear Velocity (cm/s) ($x10^{-2}$)	AP	16±6	16±6	18±5	16±6	.15
	ML	4±4	4±3	6±5	4±4	.13
	Vert	18±7	16±4	15±7	16±5	.59
Angular Velocity (°/s) ($x10^{-2}$)	AP	-11±10	-9±8	-17±21	-12±9	.30
	ML	-1±17	-1±10	-1±14	-1±11	.84
	Vert	3±19	-1±10	3±18	0±19	.61

† Leg Main Effects at $p<0.05$
* Arm Main Effects at $p<0.05$

Table 5.3-3: Recovery Leg average Step Time, Length and Width with Anteroposterior and Mediolateral Margin of Stability during unrestricted (unrest.) and restricted (rest.) arm swing and for the least and most affected legs.

				P-value	
	Arm	Least Affected	Most Affected	Leg	Arm
AP Margin of Stability(cm)†	Unrest.	-12.82±0.71	-10.46±0.92	.007	.272
	Rest.	-14.15±0.75	-11.36±0.75		
ML Margin of Stability(cm)	Unrest.	10.46±2.61	10.58±3.08	.416	.063
	Rest.	11.18±2.31	11.80±3.16		
Step Time(ms)†	Unrest.	425±111	399±109	.019	.492
	Rest.	437±115	372±97		
Step Length(cm)	Unrest.	56.43±8.90	50.94±11.39	.064	.898
	Rest.	55.56±11.29	52.22±9.98		
Step Width(cm)	Unrest.	22.93±3.84	23.81±5.45	.266	.190
	Rest.	23.92±3.51	24.91±5.49		

† Leg Main Effects at $p<0.05$
* Arm Main Effects at $p<0.05$

Table 5.3-4: Recovery Leg trunk instantaneous linear and angular velocities in all 3 axes for arm (restricted and unrestricted) and leg (least affected and most affected). Values were rounded to the nearest whole number. P-values from the two-way repeated measures ANOVA are reported for leg main effects.

		Unrestricted Arm		Restricted Arm		
		Least Affected	Most Affected	Least Affected	Most Affected	P-Value
Linear Velocity (cm/s) ($x10^{-2}$)	AP	16±7	17±9	16±6	17±4	.55
	ML	12±8	8±7	9±7	10±8	.29
	Vert	29±10	26±15	26±12	24±19	.52
Angular Velocity (°/s) ($x10^{-2}$)	AP	-30±24	-25±26	-30±25	-29±23	.30
	ML	-12±23	-6±22	-1±28	-3±28	.84
	Vert	-11±50	4±60	-1±50	-2±61	.61

† Leg Main Effects at $p<0.05$
* Arm Main Effects at $p<0.05$

Table 5.3-5: Average range of motion (degrees) for unrestricted and restricted arm swing conditions for shoulder joint angle in the sagittal plane.

	Unrestricted	Restricted	P-Value
Range of Motion°*	22.6±11.8	5.0±2.3	< .001

* Arm Main Effects at $p<0.05$

Figure 1: Experimental Setup for the CAREN system virtual environment.

3.1 Arm Swing

As no statistical significance ($p > 0.05$) in sagittal plane range of motion (ROM) existed between the least and most affected arms, the two arms were averaged for further analysis. The paired samples t-test revealed that the restricted arm swing condition had a reduced arm ROM than the unrestricted arm swing condition ($t(16) = -6.13$, $p < 0.001$). The ANOVA revealed that participants had a faster step time ($F(1,16)$, $p = 0.006$, $\eta_p^2 = 0.389$) during the restricted arm swing condition compared to unrestricted arm swing during heel-strike of the slipped limb. No further arm swing main effects or interactions were found.

3.2 Least and Most Affected Legs

Our results demonstrated that when the least affected leg was the slipped limb, it had a wider step than the most affected ($F(1,16) = 10.788$, $p = 0.005$, $\eta_p^2 = 0.403$). Further, when the least affected leg was the recovery limb, it had a larger anteroposterior MOS ($F(1,16) = 9.654$, $p = 0.007$, $\eta_p^2 = 0.376$), a longer step time ($F(1,16)= 6.825$, $p = 0.019$, $\eta_p^2 = 0.299$), and had a trend for a longer step length ($F(1,16) = 3.955$, $p = 0.064$, $\eta_p^2 = 0.198$) than the most affected. No additional main effects or interactions were found.

4. Discussion

This study examined the effect of unrestricted and restricted arm swing conditions during treadmill slips on the least and most affected legs in people with Parkinson's Disease. Our results did not support our hypothesis that the unrestricted arm swing condition would improve dynamic stability compared to the restricted arm swing condition. Rather, no differences existed between our arm swing conditions for the MOS nor for velocities. This suggests that, even when available, participants did not effectively engage an arm elevation strategy to recover dynamic stability after the slip. However, our participants had a faster step time on the leg undergoing the slip during the restricted arm swing condition. Aligned with our hypothesis, foot placement and slip recovery responses were asymmetric in our participants. Indeed, our participants had a larger step width on their least affected leg compared to their most affected for the slipped limb. Further, during the recovery step, our participants reduced the risk of backward stability loss on their least affected leg by increasing this leg's anteroposterior MOS and step time.

4.1 Arm Swing

Contrary to our hypothesis, there were no differences between the unrestricted and restricted arm swing conditions neither for the MOS nor linear and angular velocities. This occurred despite differences in arm ROM between our arm swing conditions. These findings were unexpected due to the functional role arm swing has in the recovery strategy to slips [16–18]. Marigold et al. [16,17] demonstrated that in response to a slip, healthy individuals execute a rapid arm elevation response where the arms are flexed forward and upward. The authors discussed that the arms flex specifically in these directions to move the COM further anteriorly and in the opposite direction of the trunk's initial backward displacement caused by the slip [16]. An anterior shift in the COM's position would theoretically be reflected as an increase in the anteroposterior

MOS as it measures the distance of the COM's dynamical state to the BOS [10,36]. This in turn would reflect that the arm elevation strategy is effective in reducing dynamic stability loss in the backward direction [16–18]. However, when our MOS results are examined alongside our linear and angular velocities, the lack of findings indicate that our participants' arm movements did not influence their trunk kinematics. One potential explanation for this is that our participants may have been ineffective in executing a targeted arm elevation response to the slip. Carpenter et al. [6] demonstrated that the arm responses in people with Parkinson's Disease during platform perturbations were more variable in their trajectory and had a shorter response time compared to controls. Further, the arms returned to a position near their trunk relatively quickly in people with Parkinson's Disease [6]. The authors further found that the recovery arm responses in people with Parkinson's Disease were particularly ineffective and abnormal when perturbations were induced in the backward direction (the same destabilizing direction for slips) [6]. This could explain the lack of findings in our study between our arm swing conditions for the slipped and recovery limb analyses. However, further research is needed to discriminate the impact of age and Parkinson's disease on arm strategies in response to slips. Carpenter et al. [6] found that the arm response to perturbations was ineffective despite the deltoid muscles activating earlier in people with Parkinson's Disease than controls. Therefore, future research should examine the EMG profile of the deltoid muscles between these two groups.

Interestingly, the leg undergoing the slip had a faster step during the restricted arm swing condition compared to the unrestricted. This response could reflect destabilization from the restricted arm swing condition in our participants as it may have impeded their movement timing [15]. Indeed, a faster step reduces the amount of time the contralateral leg (recovery limb) has to determine appropriate foot placement. Previous research suggests that walking with restricted arm

swing disrupts the rhythmic temporal sequence of foot placement in this demographic [15]. One of the primary symptoms of Parkinson's Disease is impaired internal movement timing [37]. Therefore, our finding supports the notion that when arm swing becomes completely absent (due to the disease), foot placement timing is further disrupted [15]. As such, disruption to temporal foot placement arises both mechanically from the absent arm swing and neurologically from the Basal Ganglia's neurodegeneration [15].

4.2 Least and Most Affected Leg

As hypothesized, our analysis revealed differences between the least and most affected leg. This potentially reflects the asymmetric neurodegeneration in people with Parkinson's Disease. Indeed, our participants demonstrated asymmetric impairment to mediolateral foot placement as they had a wider step on their least affected leg, compared to the most affected, when the least affected leg was undergoing the slip. When stability is threatened, both healthy individuals and people with Parkinson's Disease widen their steps to maintain the COM within the lateral boundaries of the base of support [15,38]. However, during walking, the neuromuscular system determines foot placement by predicting the COM's future position at the upcoming heel-strike [39-41]. In our study, slip occurrence was randomized between the legs and slip onset within each trial, was established in a pseudorandom manner. This minimized the possibility for our participants to predict the timing and laterality of the perturbation. Thus, the least affected leg's larger step width (at slip onset) reflects an already existing step asymmetry prior to the perturbation. Since our study did not examine differences between the first and subsequent slips, step width differences potentially arose as participants attempted to widen their BOS after initial slip exposure. However, the results indicate that our participants were more effective in adjusting the BOS mediolaterally with their least affected leg.

In addition, asymmetric neurodegeneration would account for the differences between the legs in attenuating backward stability loss. Indeed, the longer step time and larger anteroposterior MOS when the least affected leg was the recovery limb would reflect appropriate and asymmetric slip responses. A longer step time provides more time for stable foot placement of the contralateral leg before the COM is transferred between limbs. Additionally, the larger anteroposterior MOS reduces the risk of backward stability loss by moving the COM's dynamical state further anteriorly and in the opposite direction to the slip [10,20,21]. As step time and anteroposterior COM movement are regulated passively by subcortical structures, the differences between the legs may reflect the asymmetric neurodegeneration of dopamine within the Basal Ganglia [10,24,25]. These interlimb differences hold important implications for clinicians. Specifically, clinicians should consider the efficacy of targeting the least affected leg to promote gait adaptation and stability in slip recovery responses. This would be in line with initial evidence from Ricciardi et al. [42], who demonstrated in a pilot study that physical therapy targeting the least affected side improved more in the UPDRS-III and Tinetti scale sub-scores than standard therapies targeting both legs.

4.3 Limitations

Several limitations should be considered in the context of our study. For instance, participants were tested on their optimally medicated state which affects foot placement [37]. Additionally, differences between freezers and non-freezers were not examined. As freezers demonstrate greater postural instability than non-freezers, restricted arm swing may have a distinct effect on their dynamic stability [43,44]. Future research should also consider examining differences between people with Parkinson's Disease and healthy aged-matched controls to parse out differences due to age and those that arise due to Parkinson's Disease. In addition, it has been

shown that after perturbation exposure, healthy adults adopt a more cautious gait i.e. with shorter, wider, and slower steps as a means to enhance their dynamic stability to prevent potential future perturbations [10,22]. It is possible that our participants used a similar strategy when walking after initial slip exposure. Therefore, this should be taken into account when interpreting the findings of this article. Similarly, individuals with a higher Hoehn & Yahr score may respond differently to slips than those with a lower score as these individuals have further reductions in their postural control. Research on perturbation recovery responses in people with Parkinson's Disease would also benefit from an examination of the harness load data, an aspect unexamined in our study, to assess balance loss versus recovery. Further, future studies should examine differences in the recovery response between the first slip and following slips since differences may occur in the proactive and reactive recovery responses in restoring dynamic stability. Finally, this study did not examine differences between the dominant and non-dominant leg which may elicit further differences in response to slip recovery.

5. Conclusion

No differences between our arm swing conditions suggests that the normal arm swing in our participants was not more effective at restoring stability after an induced slip compared to when their arm motion was restricted. Lack of differences plausibly arose from our participants ineffectively implementing the arm elevation strategy to move their COM in the opposite direction from the slip's backward displacement despite differences in arm ROM between the two arm swing conditions. To determine if the arms follow an uncoordinated trajectory to counteract slip destabilization in people with Parkinson's Disease, future research should compare differences between these individuals and healthy elderly adults. Contrastingly, to our arm swing results, the

increased anteroposterior MOS, increased step time and width in the least affected leg, compared to the most, suggest that our participants were asymmetrically impaired in executing their slip recovery responses. Differences between the legs may reflect the asymmetric neurodegeneration that occurs in this demographic. Since mobility is more intact in the least affected leg, our participants adjusted foot placement of this limb to mitigate the destabilizing effects of the slip to recover stability. As such, clinicians should consider therapies that facilitate adaptive responses in the least affected leg as mobility may be too impaired in the most affected leg for effective perturbation recovery.

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Summary of Main Findings

Arm Swing:

Despite the close association between the loss of arm swing in pwPD and future falls, research to-date has yet to examine its effect on dynamic stability during steady-state walking as well as when exposed to different types of perturbations. As such, this thesis addressed this gap in the literature by demonstrating that the loss of arm swing (restricted arm swing condition) increases trunk rotational velocity in the transverse plane for this demographic (**Article 1**). Subsequently, the increase in rotational velocity in the transverse plane acted as an internal destabilization to the participants' gait. But due to the multifaceted nature of dynamic stability, each metric responded uniquely to the restricted arm swing condition. The destabilizing effect that the loss of arm swing has though on gait in pwPD depends on the terrain that individuals encounter (**Article 2**). Interestingly, despite the differences between arm swing conditions during the first two studies, the unrestricted arm swing condition was not more effective, than the restricted arm swing condition, in restoring dynamic stability following a slip (**Article 3**). This finding raises the possibility that the arm elevation strategy in pwPD is potentially ineffective. Altogether, the common result amongst all three components of this thesis (**Article 1-3**) was that the restricted arm swing condition, and its internal destabilization to gait, was particularly perturbing to step timing control in the participants.

Symptom Laterality:

In addition to the arm swing findings, this thesis addressed the sparse literature on the effects of Parkinson's Disease asymmetric progression on gait. Indeed, in all three studies (**Articles 1-3**), differences between the least and most affected legs were observed that stemmed from the disease's asymmetry. Specifically, it was found that the loss of arm swing results in

independent changes to stepping behavior for each limb (**Article 1**). Indeed, the independent changes in stepping behavior for the least affected leg, during restricted arm swing, were consistent with a compensation. This strategy appeared to be an attempt by participants to attenuate the destabilization caused by the restricted arm swing condition (**Article 1**). Interestingly, while unilateral stepping behavior occurred between the arm conditions, bilateral changes to mediolateral foot placement occurred during the dual-task (**Article 1**). This dual-task finding indicates that pwPD, in the mild to moderate stages, may still be capable of appropriate bilateral foot placement adjustment when upper body postural control is threatened. In turn, these adjustments to foot placement allowed participants in this thesis to maintain their pre-existing level of dynamic stability. Additionally, evidence from this thesis demonstrates that pwPD are capable of foot placement adjustment not only in cognitively challenging environments (**Article 1**), but also when encountering challenging destabilizing terrains (**Article 2**).

However, simultaneously, destabilizing terrains may reveal underlying differences in foot placement that stem from the disease's asymmetric progression (**Article 2 & 3**). This is due to the fact that the least affected leg is suggested to adjust its foot placement, to match the most affected side, in order to foster symmetrical walking. Thus, perturbations may be necessary to overcome this compensation and reveal underlying differences in foot placement. In turn, these differences reveal that due to the milder symptom progression on the least affected side, this leg has a more intact ability to execute appropriate dynamic stability recovery responses (**Article 3**).

Chapter 6: **General Discussion**

Despite loss of arm swing being closely associated with future falls for pwPD, fall research to-date has not specifically quantified its effect on dynamic stability.[35] Thus, to quantify the effect

of arm swing on dynamic stability in pwPD, this thesis examined differences between restricted and unrestricted arm swing conditions during unperturbed and perturbed walking. These findings are beneficial to clinicians and researchers, who aim to reduce fall rates, by presenting which aspects of neuromuscular control are particularly sensitive to loss of arm swing. Specifically, this thesis demonstrates that fall risk is exacerbated in these individuals due to loss of arm swing acting as an internal perturbation that disrupts control of step timing. Additionally, the presented findings raise the notion that fall risk can be partially reduced, given the appropriate modality, by therapeutic protocols facilitating mechanisms that control spatial awareness in pwPD. Indeed, the collective evidence demonstrates that pwPD, with mild to moderate disease progression, are capable of readjusting their spatial mechanics to compensate for loss of arm swing. Further, the presented evidence underscores that additional research on arm swing in pwPD is necessary before guidelines for future therapies can be established to restore the contralateral arm-leg swing pattern. This is particularly true for therapies aiming at facilitating whole-body recovery responses to major perturbations as the presented evidence demonstrates that natural arm motion is ineffective in slip recovery responses in this demographic.

In addition to the arm swing findings, this thesis is relevant to clinicians and future research as it examines the differences in stepping behavior in the least and most affected legs. Interestingly, despite the established neurodegenerative asymmetry in PD, research examining differences between the legs is relatively sparse. This information is crucial to future therapies as interventions targeting the least affected leg may be more effective than standard protocols that examine both sides.[56] As such, this thesis showcases that PD's asymmetric neurodegeneration impacts how individuals maintain dynamic stability in cognitively demanding scenarios (dual-tasking) as well as to mechanical perturbations that arise both internally (loss of arm swing) and externally

(destabilizing environments). Altogether, this thesis advanced the state of research for this demographic by quantifying fall risk in the context of the disease's neurodegenerative asymmetry and on terrains encountered in everyday environments. The following sections will discuss the exact impact of the findings in relation to the original objectives of this thesis, which were all achieved.

6.1 Arm Swing

In accordance with our hypothesis, walking without arm swing during unperturbed conditions increased average trunk angular velocity about the vertical axis. This finding was expected as trunk angular velocity is controlled by a 1:1 contralateral arm-leg swing pattern.[29] In contrast to findings on absent and normal arm swing in healthy adults, the increased angular velocity acted as an internal perturbation that destabilized gait in pwPD.[2,30,32] Specifically, absent arm swing was particularly perturbing to gait's temporal control. This was evidenced by the distinct responses from each dynamic stability metric as well as the changes to average spatiotemporal values in response to absent arm swing (**Objective 1-3)**. By examining these metrics alongside one another, this thesis was capable of parsing out mechanisms of the neuromuscular system that were compensatory and those that reflected motor impairment.

Indeed, the increased step time variability (**Article 1** and **2**) and reduced average step time (**Article 1** - **3**) indicate that internal timing of foot placement is particularly sensitive to the loss of arm swing in pwPD. While increases in the COV can reflect either adaptation or motor impairment, the COV of step time is a strong predictor of falls in this demographic.[5] Further, in a series of papers, Hausdorff et al. demonstrated that, unlike healthy young and older adults, step timing variability in pwPD does not exhibit long-range correlations thereby indicating that each step is

more independent and unrelated to previous steps.[83,175] The authors proposed that since impaired internal movement timing is a central feature of PD, the absence of long-range correlations may reflect pwPD continuously restarting the motor process that controls stepping instead of building off of the lower extremity's previous dynamical states.[83,175] Therefore, the increased COV step time in this thesis likely reflects additional impairment to control of step timing, that arose from the increased trunk rotational velocity when walking without arm swing, rather than appropriate foot placement adaptation. Perturbing effects to step timing control is further reflected in the reduced average step time found in this thesis (**Article 1-3**). Indeed, a reduced step time provides less overall time for the neuromuscular system to determine optimal foot placement to support the COM at upcoming heel-strike. A response not only counterproductive to reducing fall risk, but precarious for pwPD as individuals require additional time for movement execution due to bradykinesia.[55,176,177] Disruptions to internal timing of foot placement would, in part, explain the strong association between loss of arm swing and future falls.[35] Specifically, as arm swing becomes completely absent with disease progression, the internal timing of foot placement during walking is disrupted by both the increases in trunk rotational velocity as well as the continued neurodegeneration of the disease.

Further evidence supporting that the loss of arm swing is particularly perturbing to internal movement timing in pwPD comes from the directional change in spatial foot placement parameters as well as the MOS and HR. For instance, reductions in step length (**Article 1** and **2**) and increases in the ML-MOS (**Article 1**) are appropriate spatial adaptive responses to mechanical destabilization in healthy young and elderly adults.[5] Indeed, these strategies, respectively, reduce spinal rotational amplitudes and increase the distance of the COM's dynamical state from the edges of the BOS.[5,169] Further, and somewhat unexpectedly, the HR did not differ between arm swing

conditions. As the HR measures gait (a)symmetry and is based on the sinusoidal trajectory of the COM, upper and lower extremity spatial adjustments to the restricted arm swing condition may have been sufficient for maintaining the participants' already existing COM trajectory. Therefore, in contrast to step timing parameters, the findings here indicate that pwPD are still capable, at least partially, of spatially adjusting their upper and lower extremities to mitigate perturbations to their COM's trajectory from arm swing loss. Spatial foot placement adaptation agrees with evidence from split-belt gait asymmetry protocols in pwPD.[88] In a systematic literature review, Seuthe et al. compiled evidence highlighting that pwPD are able to adapt their step length to the asymmetric belt speeds in a manner similar to controls.[88] However, while the ability to spatially adapt foot placement is retained in pwPD, the literature also demonstrates that pwPD require additional time to switch to the appropriate step length pattern during asymmetric treadmill walking compared to controls.[88] Slower adaptation to new environments and perturbations likely stems from an impaired ability to readily execute movements that are environmentally specific and relevant.[118] This impairment would exacerbate fall risk in pwPD during more instantaneous discrete perturbations as individuals are incapable of readily switching to an appropriate motor response that mitigates the encountered destabilization.[118] This would account for the lack of differences between unrestricted and restricted arm swing during the slip analysis in this thesis.

Unlike the continuous perturbations during the destabilizing terrains (**Article 2**), the induced slips (**Article 3**) were fast discrete perturbations. As such, participants would have to engage the arm elevation strategy relatively quickly to move the COM in the opposite (forward) direction of the slip's backward destabilization.[28,107] Dimitrova et al. discussed that in healthy adults, the tonic dopamine released in the BG is responsible for selecting an appropriate muscle activation pattern to quickly switch to goal directed motor responses that are specific to the

environment encountered.[38] However, in PD, loss of dopamine in the BG impairs its ability to filter out muscle activation patterns for movements that are not desired nor specific to the environment.[38] Indeed, despite muscular activity occurring earlier in pwPD, compared to controls during static standing perturbations, pwPD exhibit earlier and prolonged activation of antagonist muscles.[38,125] In regard to this thesis, lack of arm swing differences during the slip condition potentially arose due to the inability of participants activating the correct muscle patterns responsible for the arm elevation strategy. However, as this thesis did not examine EMG activity, future research should examine differences in shoulder joint EMG activity between pwPD and healthy aged-matched adults to determine potential differences in muscle recruitment exist.

Overall, the arm swing findings within this thesis holds several important implications for future researchers and clinicians. First, the strong association between arm swing loss and future falls in this demographic may be due to the increased trunk rotational velocity mechanically perturbing internal timing of foot placement. Secondly, additional research is required to determine the arm muscle activation patterns in pwPD, compared to healthy aged-matched adults, during unperturbed and perturbed walking. In doing so, a clearer understanding of the neuromuscular impairments that leads to both general arm swing loss and ineffective arm swing responses, in regard to perturbations, would be acquired. This in turn would facilitate the development of protocols that aim to restore the nominal 1:1 arm-leg swing pattern and strategies for engaging an effective arm elevation response to slips. However, with the current state of arm swing research, clinicians should instead consider protocols that foster rhythmic and controlled stepping. Such an approach would have more efficacy than increasing arm swing as this may further increase spatiotemporal variability in pwPD thereby exacerbating their fall risk.

6.2 Least and Most Affected Legs

One of the overarching contributions this thesis provides to dynamic stability research in pwPD is examining potential differences between the least and most affected legs. Indeed, despite the known asymmetry of dopaminergic neurodegeneration, the majority of research thus far did not examine interlimb differences. This may be due to the classical notion that there is no correlation between the UPDRS asymmetry scores and stepping asymmetry. However, the UPDRS scores asymmetry predominantly by questions on upper body functionality and provides comparatively few questions on the lower extremity.[132] To overcome this issue, this thesis defined the least and most affected legs by both the UPDRS III (motor section) and participant self-reporting motor symptoms dominance. In doing so, the presented research provided critical insight regarding how asymmetric neurodegeneration impacted how pwPD responded to loss of arm swing (internal perturbation), destabilizing terrains (external perturbation), and dual-tasking (cognitive perturbation).

When participants walked without arm swing during the unperturbed terrain **(Article & Objective 1)**, the least affected leg exhibited characteristics of a compensation response while the most affected reflected motor impairment. Specifically, the increased MOS and reduced step length only occurred in the least affected leg in the restricted, compared to unrestricted, arm swing condition. The ability to spatially adjust the least affected leg may be a result of the asymmetric neurodegeneration impairing integration of proprioceptive feedback to respond to the increased trunk rotational velocity on the most affected side. In a literature review, Konczak et al. highlighted evidence that motoneurons in the subtahalamus nucleus and global pallidus internus respond to joint specific movement changes in healthy individuals.[26] However, in pwPD, neurodegeneration leads to either hyper- or hypoactivity in the BG thereby broadening the proprioception receptive

field of these neurons.[26] The loss of limb specificity ultimately impairs the BG's ability to filter out relevant motor information to the motor cortex when selecting an appropriate motor skill.[26] In turn, the authors proposed that this impairment leads to either constant facilitation or braking of the BG's efferent targets.[26] In regard to this thesis, asymmetric neurodegeneration may have impaired participants' ability to effectively integrate proprioceptive feedback in order to engage correct foot placement adjustments of the most affected leg to compensate for restricted arm swing.

Interestingly, the independent behavior of each leg further corroborates the suggestion that loss of arm swing is particularly perturbing to internal movement timing in pwPD. Support for this perspective stems from the increased step time variability in the most affected leg and the increased step time variability interaction in the least affected (dual-tasking with restricted arm swing compared to single-task walking with unrestricted arm swing, **Article 1**). Regarding the most affected leg, the rhythmic step timing would be more sensitive to perturbations as the automatic internal timing functionality of the BG is further deteriorated in this cortical hemisphere.[55,89] To compensate for impaired internal timing, pwPD recruit higher level cortical structures (frontal lobe and prefrontal cortex) to direct additional attentional resources to control their mechanics.[25] Since neurodegeneration on the most affected side is more advanced, relative to the least affected side, extra attentional resources would theoretically be required to control step timing in this limb. As a result of this higher cognitive load, the most affected side may have a relatively lower threshold for handling disruptions to its temporal control that arise from perturbations. During the restricted arm swing condition, additional attentional resources would be required to adjust foot placement to mitigate the increased trunk rotational velocity. Allocating additional attention (due to restricted arm swing compensation) on top of the already elevated cognitive resources required to control stepping of the most affected leg may have strained attentional resource capacity in

participants.

In contrast to the most affected side, neurodegeneration in the least affected side has a relatively milder progression.[89] Therefore, this potentially requires less compensation from higher level cortical structures to regulate step timing. The observed interaction, therefore, suggests that control of step timing in the least affected leg is potentially more robust to perturbations as a certain amount of automaticity is intact on this side. Thus, both cognitive and mechanical perturbations may be required to sufficiently strain and reveal the extent attentional resources control this limb's step timing. Interestingly, the dual-task alone did not increase temporal variability in either limb. Yogev et al. discussed that both the difficulty and the type of dual-task determine which aspects of gait, as well as the magnitude of change, are altered in both healthy elderly adults and pwPD.[25] In this thesis, the dual-task was a simple word searching task used to emulate secondary tasks one may encounter outside the laboratory (ex: walking while glancing at street signs). Due to the relatively simplistic nature, and as this type of secondary task is common in everyday life, participants may have been rather adept at allocating resources between bilateral step timing and the dual-task. Indeed, studies on dual-task interventions demonstrate that destabilization from a cognitive secondary task can be mitigated with sufficient task exposure.[178]

However, while task difficulty and potential prior exposure account for the aforementioned findings, the increased ML trunk linear variability indicate that trunk postural control is particularly susceptible to cognitive perturbations. Unlike the AP direction, movement in the ML direction is dictated by active information processing that stems from higher level cortical structures.[24] For healthy young and elderly adults (specifically non-fallers) this is relatively unproblematic as these demographics have a large reserve of attentional resources and automatic control of gait parameters (ex: step length and time) is still preserved.[25] Contrastingly, the reduced

gait automaticity in pwPD requires individuals to utilize attentional resources to control aspects of their gait that would otherwise be automated.[25] As such, ML trunk postural control appeared to be especially sensitive to dual-tasking effects as attentional resources typically used for ML control are strained between the word searching task and controlling additional (normally automated) aspects of gait. Interestingly, participants were able to account for their reduced postural control by bilaterally adjusting their ML foot placement (increased step width variability) to maintain their already existing global dynamic stability (no changes in MOS) **(Article 1)**. Although ML foot placement is also mediated by active information processing, foot placement adjustment may have occurred due to the subthalamus nucleus preferentially activating the lower extremity, over the upper, as proposed by Baron et al.[117]

While the independent behavior of each leg may be distinct, functional differences between the legs may require external perturbations to be revealed **(Articles 2 & 3)**. Ricciardi et al. discussed that pwPD may adapt their least affected leg's foot placement, to match the most affected side, in order to foster symmetrical walking.[56] Therefore, to overcome this compensation, external perturbations may be necessary to reveal underlying interlimb differences. This would account for differences between the legs in step length **(Article 2 & 3)** and AP-MOS **(Article 3)** found within this thesis during the minor and major destabilizing terrains that did not occur during unperturbed walking **(Articles 2)**. Further, the presented evidence suggests that perturbations acting in the same direction as the gait parameter of interest are more effective in eliciting interlimb differences than perturbations in other directions. Indeed, differences between the legs only occurred during the rolling-hills terrain **(Article 2)** and the slips **(Article 3).** As both of these perturbations acted in the AP direction, their destabilizing effects on gait would be reflected more in step length and the AP-MOS.

However, an increased average step width **(Article 3)** of the least affected leg still occurred despite the slips acting in the AP direction. The differences in step width may be accounted for by both how the slip was induced as well as the neuromuscular control of ML foot placement. Slips were induced by a transitory asymmetry in the treadmill belts (accelerating the belt under the slipping leg). In response to asymmetric walking, McFadyen et al. proposed that individuals consciously direct attention to their lower extremity to modify foot placement.[179] In pwPD, modifying segmental movement is largely dependent on visuospatial feedback as it is relayed to the cerebellum over a pathway relatively spared from the disease.[26] Since visuospatial feedback is an integral component for determining ML foot placement, attentional resources may have been strained when participants attempted to adjust their gait to the treadmill.[24] As the least affected side is more capable of foot placement adjustment, participants were thus only capable of increasing step width on this side.

The findings on the least and most affected leg in this thesis overcomes a long-standing gap in the biomechanical assessment of gait in pwPD. Indeed, the presented evidence indicates that the independent stepping behavior of each leg is different in pwPD in response to both loss of arm swing and dual-tasking. When considering the biomechanical parameters of each, the changes in the spatial parameters of the least affected side may mirror the milder neurodegeneration in this hemisphere. Opposingly, the changes in temporal parameters in the most affected side suggest that the more severe neurodegeneration process on this side exacerbates the destabilizing effects from perturbations. However, since the least affected side is capable of foot placement adjustment to reduce gait asymmetry, functional differences between the legs requires external perturbations in order to be revealed. Altogether this evidence indicates that clinicians and researchers should give deference to the diseases asymmetric neurodegeneration when examining gait and fall risk.

Specifically, clinicians should consider facilitating the capability of the least affected leg to adjust foot placement. In a similar fashion, future research should examine the extent to which both mechanical and cognitive perturbations elicit differences between the legs.

Chapter 7: **Conclusion**

The purpose of this thesis was to examine the effect of unrestricted and restricted arm swing on the dynamic stability of pwPD during steady-state walking and on mechanically perturbing terrains. This thesis demonstrated that when pwPD walk without arm swing, trunk rotational velocity increases since COM motion is no longer regulated by the contralateral arm-leg swing pattern. Subsequently, this increase in trunk rotational velocity acts as an internal perturbation that destabilizes gait in these individuals. Interestingly, the cumulative evidence within this thesis demonstrates that, in pwPD, each aspect of neuromuscular control responds uniquely to the destabilization from the increased trunk velocity. This was substantiated by the differences found in dynamic stability and spatiotemporal metrics. Indeed, dynamic stability metrics that quantify spatial mechanics reflected appropriate attempts by pwPD to adjust their postural control to mitigate the risk of falling. Further, evidence from step length and width metrics corroborate the notion that pwPD in the mild to moderate disease stages retain the ability to spatially readjust their mechanics as participants modified their spatial foot placement to compensate for the restricted arm swing condition. In contrast, dynamic stability metrics that quantify balance on temporal mechanics, along with reductions in average step time, indicate that the increased trunk rotational velocity during the restricted arm swing condition was particularly perturbing to rhythmic timing of foot placement in pwPD.

In addition to the findings on arm swing, this thesis showcases the differences between the least and most affected leg in pwPD in response to mechanical perturbations. Specifically, when

walking without arm swing, the asymmetric neurodegeneration in PD causes independent changes in the least and most affected leg with the former reflecting compensation while the latter reflects motor impairment. However, as the least affected leg is suggested to adapt its foot placement to match the contralateral side, functional differences between the limbs in pwPD may only be elicited when these individuals encounter large destabilizations. Further, observable differences between the limbs, in response to mechanical perturbations, primarily occur in the direction the destabilizing effect acts in.

Chapter 8: **Contributions**

These findings hold several important applications to the examination of fall risk in this demographic. First, the evidence from spatial metrics raise the notion that pwPD in the early to moderate stages retain the capability of adjusting their spatial mechanics to the environment. Subsequently, the presented evidence indicates that the main strategy pwPD implement to compensate for arm swing loss is a reduction in step length. Therefore, clinicians should consider therapies that foster reductions in step length in order to mitigate increases in trunk rotational velocity when pwPD lose their arm swing during the course of the disease. Secondly, this thesis highlights the fact that the close association between loss of arm swing and future falls is a result of impaired temporal control of foot placement. Indeed, the overarching pattern from the Research Objectives within this thesis demonstrates that loss of arm swing causes, in addition to the disease's neurodegenerative effects, further mechanical disruptions to rhythmic timing of foot placement in pwPD.

A holistic examination of this thesis, therefore, points to the importance for PD therapies to facilitate the "cautious gait" strategy (reducing step length while increasing step width and time). This strategy holds several advantages over attempts to restore the contralateral arm-leg swing

pattern through increases in arm swing amplitude. Specifically, this strategy may reduce the fall risk caused by arm swing loss by attenuating increases in trunk rotation amplitudes (reduced step length), widening the ML base of support to ensure a more stable COM transfer between the limbs during double-support (increased step width), and provide additional time to determine appropriate foot placement (increased step time). In contrast, therapies increasing arm swing amplitude may be too cognitively challenging for pwPD and may in fact increase fall risk due to potential increases in spatiotemporal variability. However, additional research is necessary in order to examine the efficacy of modalities that implicitly aim to restore the contralateral arm-leg swing pattern as this remains unexamined. Finally, therapeutic protocols should consider potentiating the ability of the least affected leg to adapt to multiple scenarios as mobility is more preserved in this limb.

Chapter 9: **References**

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Appendix A: Original Freezing of Gait Questionnaire

1. Freezing of Gait Questionnaire (FOGQ)

1.1. During your worst state—Do you walk:

0 Normally
1 Almost normally—somewhat slow
2 Slow but fully independent
3 Need assistance or walking aid
4 Unable to walk

1.2. Are your gait difficulties affecting your daily activities and independence?

0 Not at all
1 Mildly
2 Moderately
3 Severely
4 Unable to walk

1.3. Do you feel that your feet get glued to the floor while walking, making a turn or when trying to initiate walking (freezing)?

0 Never
1 Very rarely—about once a month
2 Rarely—about once a week
3 Often—about once a day
4 Always—whenever walking

1.4. How long is your longest freezing episode?

0 Never happened
1 1–2 s
2 3–10 s
3 11–30 s
4 Unable to walk for more than 30 s

1.5. How long is your typical start hesitation episode (freezing when initiating the first step)?

0 None
1 Takes longer than 1 s to start walking
2 Takes longer than 3 s to start walking
3 Takes longer than 10 s to start walking
4 Takes longer than 30 s to start walking

1.6. How long is your typical turning hesitation: (freezing when turning)

0 None
1 Resume turning in 1–2 s

2 Resume turning in 3–10 s
3 Resume turning in 11–30 s
4 Unable to resume turning for more than 30 s