Mediolateral stability during gait in people with Parkinson's disease

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Mediolateral stability during gait in people with Parkinson’s disease

by

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For my family.
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ABSTRACT

Gait control is a clinical problem in people with Parkinson’s disease (PD). Gait variability leading to instability is commonly measured using spatio-temporal variables like step length, step time, step width and cadence. Another measurement that provides information about directional instability is harmonic ratios (HRs). The purpose of this study was to examine the relationship between step width and mediolateral stability using HRs in people with Parkinson’s disease (n = 19) and age matched controls (n = 19). The participants walked at their preferred pace and then with a wider step width and narrower step width. The results showed that the PD group exhibited lower HRs compared to controls in preferred gait and a narrower step width. As expected, HRs were lower for both groups when walking with a narrow step width compared to preferred gait, but counter to expectations, the decrease was similar between groups. Overall, these data indicate that step width directly influences ML-HRs, and that decreased ML dynamic balance with PD severity may be related to a reduction in step width. The information gathered in this study may help in improving intervention strategies for gait instability.
CHAPTER 1.

INTRODUCTION

Control of gait is an important clinical problem in people with Parkinson’s disease (PD). Parkinson’s disease gait is characterized by short and shuffling steps with an increase in gait variability (Schaafsmaa, et al., 2003; McIntosh, Brown, Rice, & Thaut, 1997). Studies show that this variability leads to increased gait instability and a predisposition to falls while walking (Guimaraes & Isaacs, 1980; Woolley, Czaja, & Drury, 1997; Van Swearingen, Paschal, Bonino, & Yang, 1996).

Gait variability leading to instability is commonly measured using spatio-temporal variables like step length, step time, step width, and cadence. Studies have characterized changes in these parameters as an indication of gait adaptation to enhance stability (Menz, Lord & Fitzpatrick, 2003; Oberg, Karsztnia & Oberg, 1993).

People with this controlled mode of gait exhibit a slow speed of walking and walk with a wider base of support, presumably to improve stability. However these changes in gait could be from fear of falling (Maki, 1997; Ashburn, Stack, Pickering, & Ward, 2001; Bloem, Hausdorff, Visser, & Giladi, 2004) and not instability. Thus finding a more sensitive tool that can identify changes in gait instability is needed. A method of detecting more subtle changes in gait is measuring trunk accelerations while walking (Yack & Berger, 1993; Henriksen, Lund, Moe-Nilssen, Bliddal, & Danneskoid-Samsoe, 2004), which could help in an early detection of gait instability.

Not only are trunk accelerometers more sensitive but they can be used to determine direction-specific instability. Few studies have examined anterior-posterior (AP), vertical (V), and mediolateral (ML) instability in PD. Research evidence shows that maintenance of stability
in the ML direction requires active control and hence is complex as compared to AP stability, which requires passive control (O’Connor & Kuo, 2009). ML stability can be measured using harmonic ratios (HRs), which measure trunk movements. Studies have shown that HRs in the ML direction are low in people with a predisposition to falls (Yack and Berger, 1993) and are low in people with PD compared to neurologically healthy age-matched controls (Lowry, Smiley-Oyen, Carrel & Kerr, 2009). In fact, ML direction was the only direction correlated with disease severity.

The purpose of this study was to examine the relationship between step width and ML stability using HRs in people with Parkinson’s disease and age matched controls. This relationship was examined by: 1) measuring step width, step width variability, and HRs during preferred gait in PD and age-matched controls; and 2) manipulating step width (wide or narrow base of support) and measuring differences in HRs, with a focus on the ML-HR, in PD and the controls.

It was hypothesized that people with PD would exhibit wider step width and greater step width variability, and lower HRs in all three directions during preferred walking compared to age-matched healthy controls. It was also hypothesized that walking with a wider step would increase ML-HRs compared to preferred walking for all participants. Finally, it was hypothesized that narrowing step width would lower ML-HRs for all participants compared to preferred, but the change would be greater in people with PD compared to age-matched controls. We also hypothesized that AP- and V-HRs will be lower when step width is adjusted away from preferred. It is hoped that information gained from this study will help to identify individuals susceptible to falls and to contribute information to make intervention strategies for gait instability more effective (Zetterberg, Elmeron, & Anderson, 1984).
CHAPTER 2.
REVIEW OF LITERATURE

Normal gait mechanics

Bipedal locomotion is designed for efficiency and to save on expenditure of mechanical energy while walking (Grasso, Zago, & Lacquaniti, 2000). It requires maintenance of equilibrium during progression with an ability to adapt to changes in the environment (Woollacott & Tang, 1997). Maintenance of this equilibrium is challenging for the postural control system because for a considerable period of the walking cycle the body is supported on a single limb leading to instability in older adults (Winter, 1995; Cali & Kiel, 1995; Norton, Campbell, Lee-Joe, Robinson & Butler, 1997).

The normal gait is a rhythmic cycle with alternating propulsive and retropulsive motions of the lower extremities (Winter, 1979). While the beginning of gait has a clear demarcation in young adults (YA) and healthy older adults (OA), the ending pattern of the cycle is comparatively less well defined. Therefore in order to analyze gait accurately and efficiently the gait cycle is divided into different phases. One gait cycle, referred to as a stride, is recorded as the point of initial contact to the point at which the same extremity contacts the ground again. Each stride is comprised of two steps, and each step is comprised of stance and a swing phase. The stance phase forms about 60% of the gait cycle and is recorded as the point of contact of the heel of the extremity (heel strike) to the point until the toe leaves the ground (toe-off). The swing phase forms 40% of the gait cycle and is the point at which the toe of the extremity leaves the ground until the point at which the heel of the same extremity comes in contact with the ground again. A period of double limb support, which forms 20% of the gait cycle, occurs between the phases of walking. This phase is more stable because both lower extremities are in contact with
the ground. The percentage of time spent in double support varies and might be increased especially with age or in people with balance disorders (Murray, 1967).

Gait is evaluated based on temporal and spatial parameters. Typical temporal variables include stance time, swing time, stride and step time, cadence and speed. It is expected that with age the temporal parameters show more variability or decline in values when compared to YA (Menz et al., 2003). The spatial parameters help in analysis of balance in the OA. It is expected that OA typically walk at a slower speed with shorter step length and a wider base of support as compensatory mechanism. This controlled mode of walking helps in maintaining balance (Menz, et al., 2003; Oberg, Karsznia & Oberg, 1993).

While many studies have addressed stability while standing, very few studies have evaluated dynamic balance while walking (Menz, Lord & Fitzpatrick, 2003). Maintenance of dynamic stability is a complex skill requiring integration of neurophysiological and biomechanical variables (Lord & Sturnieks, 2005; Sturnieks, St George & Lord, 2008). The stability during walking is more challenging due to a constant shift of center of mass and about 70% of the falls occur during locomotion (Cali & Kiel, 1995; Norton et al., 1997). The balance tests done in standing, hence can only moderately predict walking ability and there is a need for methods that analyze and predict the susceptibility to falls when there is a shift in dynamic stability (Menz et al., 2003).

**Gait and Aging**

Few studies have addressed why OA reduce the walking speed, thus adapting a controlled mode of walking. Older adults tend to walk slowly and with greater stride variability (Berg, Alessio, Mills & Tong, 1997). The postural strategies adopted by OA suggest that the gait changes are compensatory mechanisms to control the degrees of freedom (Maki, 1997). Maki
(1997) examined compensatory strategies in OA and found walking slowly with a wider base of support could be perceived by OA as an attempt to stabilize the posture and prevent a fall. However, in spite of the decrease in speed of walking, the persistence of falls indicates the influence of other variables such as age related biomechanical alterations and postural inefficiency in maintenance of balance (Menz et al., 2003).

Further evidence can be found in studies examining controlled walking. Dean and Alexander (2007) and Bauby and Kuo (2000) found the wider base of support exhibited by OA to be a compensatory strategy requiring active control of mediolateral (ML) balance via foot placement as compared to maintenance of AP balance, which requires passive control. The findings of these studies will be further explained later.

Kerrigen, Todd, Della, Lipsitz and Collins (1998) assessed the biomechanical changes in OA while walking. The authors tested 31 healthy older (aged 65-84) and young adults (aged 18-36). The participants walked barefoot at their comfortable speed across a 30-foot walkway. In addition to walking at their preferred speed the participants were asked to walk faster across the walkway. They found a decrease in some kinetic and kinematic differences when the speed of walking was increased. However biomechanical differences like reduced hip extension, increased anterior pelvic tilt, and reduced ankle movements persisted in older adults even after controlling for speed. The decreased hip extension in older adults is related to an increased anterior pelvic tilt due to postural changes as a result of age. The authors concluded that these findings would be exaggerated in older adults with a predisposition to falling, as falling is associated with activities requiring hip extension and ankle movements such as walking. Based on the findings of this study it could be hypothesized that OA with exaggerated changes in these
biomechanical parameters would exhibit a controlled mode of walking to prevent loss of balance and falling.

This controlled mode of walking exhibited in OA could be compared with that of YA. Menz et al. (2003) studied postural control in YA and OA while walking on various even and challenging surfaces. The authors recruited 30 healthy YA and OA and evaluated the response of the postural control system to different walking conditions. The participants walked on even and uneven surfaces and the acceleration patterns of the head and the pelvis while walking were recorded. The authors found that while the YA maintained the speed of walking and walked with longer stride length, their cadence varied and was slower on irregular surfaces. This strategy is in contrast to that of OA who controlled the speed of walking to avoid loss of balance (Menz et al., 2003).

Kang and Dingwell (2008) studied the effects of age and walking speed on gait variability. The authors hypothesized that a decrease in the walking speed increases gait variability. They tested 18 (72 ±6) older and 17 young (23 ± 3) adults on a treadmill walking at a speed 80% to 120% of their preferred speed. The authors treated speed as a confounding factor and hence controlled it to identify other age related causes of variability. They recorded variability of spatio-temporal gait measures, lower extremity joint angles, and trunk motions with bilateral isometric leg strengths and passive range of motion. The authors found that the preferred speed of walking was similar in both age groups. In both groups gait variability was dependent on speed for stride time, frontal hip and knee motions, knee internal /external rotations and trunk motions. However older adults exhibited more variability for trunk motions and stride time irrespective of changes in speed. The authors reasoned that this variability was a possible result of age related biomechanical factors such as poor leg strength and flexibility rather than
slowing of speed. It could be concluded from the studies that the biomechanical alterations leading to a controlled mode of walking increases gait instability in OA (Guimaraes & Isaacs, 1980; Woolley et al., 1997; Van Swearingen et al., 1996).

Drawing evidence from the studies it could be concluded that the impaired balance is due to a systemic deterioration and degeneration of the gait mechanics in OA. The OA tend to walk with a shorter step length and an increase in double limb support time, which reduces the overall speed during walking. This strategy, though offering more control over balance, does not always prevent a fall (Menz et al., 2003). As adaptation to a conservative walking pattern does not decrease the risk of falling in OA, it is necessary to evaluate dynamic balance for accurate analysis of gait and clinical prevention of falls.

Research shows gait variables like gait speed and step length could be referred to as indicators of fear of falling (Maki, 1997). Step width, a compensation for instability, along with step time and gait variability, could be a reference for direct prediction of risk of fall (Gabell & Nayak, 1984; Hausdorff, Rios & Edelberg, 2001).

Studies have found that while biomechanical alterations, such as decrease in speed, influence gait instability, there are other age-related factors that have a profound influence on gait such as an increase in the noise in the system while processing sensory information leading to susceptibility to falling laterally (Dean & Alexander, 2007). Postural efficiency decreases with an increase in the noise and is an important cause of falls in the elderly (Lord & Sturrieks, 2005). The processing interference causing the noise (Dean & Alexander, 2007) is explained later in the review.

The OA compensate for age-related impaired physiological functions like impaired visual acuity, depth perception, vibration sense, ankle dorsiflexion, quadriceps strength, and increase in
postural sway by walking slowly. These adaptations add to the instability. These spatio-temporal adaptations to control balance however provide only an indirect measure of stability and thus give a limited insight to the maintenance of balance (Menz et al., 2003). Hence there is a need for other measurements that gives reliable insight into examination of balance and dynamic stability. A reliable measure is the use of trunk accelerations to examine gait instability (Kavanagh & Menz, 2008; Moe-Nilssen, 1998; Henriksen et al., 2004). This measure is addressed later in the review.

**Directional Instability**

Walking studies have found that the instability is direction-specific in the AP and ML directions. Hilliard et al. (2008) found that ML stability is an important parameter in recovery of dynamic balance in the community dwelling OA. O’Connor and Kuo (2009) examined directional dependent control mechanisms. The study was based on the theory that self-stabilizing aspects of gait required little or no central nervous system control. The authors hypothesized AP movements were passively stable from step to step, but that ML balance was more challenging and required motor control via active foot placement. They stated that if this hypothesis were true then humans would rely less on integrated sensory feedback while maintaining balance in AP direction as compared to the ML direction. The researchers recruited ten healthy participants and applied AP and ML perturbations in the visual field while measuring the foot placements during treadmill walking. The authors found that there was a significant increase in step variability during ML perturbations. The direction sensitivity changed and the balance in AP direction became more sensitive when the participants were given vibrations during quiet standing. The authors further tested the directional sensitivity in tandem stance (heel to toe). It was found that dynamic balance with a narrow base of support further increased ML
sensitivity. It could be concluded from this study that while stability in AP direction is passively maintained ML stability in the lateral directions requires active control strategies. This requirement makes ML stability more complex to maintain.

In addition to an increase in postural sway while standing there is evidence that older adults exhibit greater lateral instability during walking. Dean and Alexander (2007) studied the age related effect of lateral stability on gait. The researchers suggested that walking is unstable laterally in older adults because of an age-related decrease in sensory and motor functions, resulting in increased noise. They hypothesized that these age-related deficits could lead to an increase in step width variability due to control of foot placement in response to this increased noise. Poorer control of lateral stability could influence step width variability and larger step widths might be seen as compensation to reduce lateral instability as seen in older adults. The authors found that the energetic costs of walking when lateral stability was increased (via a belt around the waist attached to lateral external supports) were similar between the old and young adults. Thus, when lateral stability was controlled, differences between old and young adults were reduced.

In another study Bauby and Kuo (2000) measured variability of foot placement during gait and tested the control of lateral balance during dynamic instability. The simple dynamical model used by the authors had a passive gait with an exception of active control during a single unstable movement mainly laterally. This instability was controlled by adjusting the lateral foot placement. The model predicted variability in foot placement when there was an uncertainty in the active feedback loop. A similar situation along with loss of sensory feedback by closing of eyes showed a larger increase in lateral variability. This principle when applied to the human subjects showed the need for significant active control to maintain lateral stability while walking.
Oates, Patla, Frank, and Greig (2005) evaluated the changes in AP and ML directions in YA while walking on irregular surfaces. The authors found that while walking on slippery surfaces YA compensate for lack of sufficient braking force by transferring the AP kinetic energy to ML kinetic energy. This compensation helps YA maintain their balance on irregular surface. As the ML stability in OA is compromised they find it difficult to maintain balance on an irregular surface. It could be concluded from these studies that directional stability especially in ML direction provides a significant insight into gait instability. However in spite of the significance there is limited research addressing ML stability.

**Trunk Accelerometry**

The ability to regulate balance can be examined using trunk accelerations (Henriksen, Lund, Moe-Nilssen, Bliddal, & Danneskoid-Samsoe, 2004). The upper body forms a large proportion of the body mass and it is challenging for the central nervous system to control upright trunk posture while maintaining dynamic stability. This maintenance of an upright posture is even more challenging for the body as it encounters constant perturbations as the foot contacts the ground. The upper body tends to rotate forward causing rapid stride to stride trunk accelerations (Kavanagh, 2009). Trunk accelerations have been found to be an important source of maintaining ML balance along with foot placements and contributions from the ankle (Aminian et al., 1999). When evaluating dynamic stability, an interesting observation in OA is reduction in magnitude of accelerations experienced by the head and the pelvis when walking. This is a compensation strategy to maintain balance with impaired physiological functions and reduced strength of the lower extremity (Lord & Sturnieks, 2005).

Grossman, Leigh, Abel, Lanska, and Thurston (1988) studied locomotion in healthy OA and found that a rhythmic activity such as walking leads to corresponding rhythmic oscillations
of the trunk and head in both the sagittal and frontal direction. This arrangement is not only thought to control movement trajectory but is also essential for sensory feedback to maintain balance. As analysis of balance and gait stability is difficult, especially while standing, an alternative could be using the trunk accelerations and gait variables (maximum step length and stride variability) as an efficient predictor and a tool for detection of balance impairments in older adults (Lindemann et al., 2008). Trunk accelerations are an alternative to the traditional use of spatio-temporal parameters like step length, foot angle and step and stance times for assessing the gait stability. This method has been used since the 1960’s for analysis of gait in amputees and while walking with crutches. Accelerometry is now used for evaluating effects of aging and different walking surfaces on gait thus measuring the walking stability.

In a review, Kavanagh and Menz (2008) weighed the benefits of using accelerometry for quantification of movement patterns while walking. Accelerometers can measure three-dimensional (3D) accelerations directly, eliminating errors associated with differences in displacement and velocity. They are economical as compared to other expensive laboratory equipment. The small size of accelerometers allows the participants to walk unrestricted and being portable it is not confined to testing within the laboratory settings. A variety of accelerometer designs offer various dynamic ranges and levels of sensitivity. The ability to capture a number of gait cycles is also a strength of use of accelerometers (Kavanagh, Morrison, James & Barrett, 2006).

The use of accelerometers for gait analysis is a reliable measure (Moe-Nilssen, 1998; Henriksen et al., 2004). The examiners assessed the reliability of collecting acceleration data from two testing sessions conducted a day apart. Both the studies employed similar methods that included placing a triaxial accelerometer over the L3 spinous process and collecting data at
various walking speeds. The authors found that the reliability was high when the participants walked on a level surface. The studies also found that while walking on an even surface the intraclass correlation coefficients were higher for AP (0.88), followed by ML (0.78) and V (0.77) directions. This reliability was affected only slightly when the participants walked on an uneven surface, with a slight decrease in correlations in AP direction and an increase in ML and V directions. The authors concluded that a single accelerometer attached to the lower trunk is a reliable technique of directly measuring trunk accelerations in healthy individuals without mobility impairments.

Kavanagh et al. (2006) tested the reliability of segmental accelerations measured using a wireless gait analysis system. The authors examined the reliability of accelerometry as well as the inter- and intra-examiner (placement and replacement) reliability. The 3D accelerations of upper and lower body were measured during self selected slow, preferred, and fast walking speeds. Eight young adult males were tested in two sessions where accelerometers were positioned on the head, neck, lower trunk and right shank. While the accelerometer placement on the shank was found to be most reliable, placements over the other areas was found to have a high reliability, too. The results showed that stride-stride acceleration reliability was not significantly different from inter- and intra-examiner reliability, further reinforcing the conclusion that errors associated with reapplication or placement of accelerometers by same or different examiners was minimal.

Kavanagh (2009) provides further evidence of the reliability of trunk accelerometry. He examined the effect of speed of walking on the lower trunk motion of healthy YA while they walked on a straight line (five trials) at self selected slow, preferred, and fast walking speeds. Lower trunk accelerations were measured in the AP, V and ML directions using a triaxial
accelerometer. Stride to stride acceleration amplitude, regularity and repeatability were examined with root mean square (RMS) accelerations. The results showed the RMS acceleration increased with gait speed in all directions. During slow walking ML and V accelerations showed less regularity and repeatability. The stride to stride acceleration regularity and repeatability however did not differ between the preferred and fast walking speed conditions. The researchers concluded that walking at speeds slower than preferred lowers trunk acceleration in the frontal plane. The features of trunk acceleration do not seem to change during preferred and fast walking conditions. It could be concluded from this study that OA with controlled or slower speed of walking would possibly demonstrate lower accelerations in the frontal plane.

Kavanagh, Barrett, and Morrison (2004) assessed the difference in upper body accelerations in eight young (23 ± 4) and older adults (73 ± 3) while walking on a 20 m walkway at a preferred speed using a triaxial accelerometer. The authors assessed stride, stance and swing durations, cadence, gait velocity, step length, and 3D head and trunk accelerations. They found that there was an increased variability in trunk accelerations in the AP direction in the OA compared to YA. The authors concluded that OA exhibit different patterns of upper body motion as an attempt to increase dynamic stability while walking.

Moe-Nilssen and Helbostad (2005) measured acceleration at the level of the lumbar spine and reported higher average accelerations in people with balance impairments. During walking, stabilization of the head is essential for optimizing the visual apparatus and maintaining the stability of the head is one of the primary functions of the body’s postural control system while walking. Hence, acceleration patterns of the head and pelvis may help in developing a more concrete model of walking stability and could be used as an indicator of whole body balance when walking with a manipulated step width gait.
Harmonic Ratios

Harmonic ratios (HRs) provide another measure of gait stability that are based on frequency analysis of trunk accelerations (Yack & Berger, 1993). Supporting evidence can be found in a study by Yack and Berger (1993) in which they assessed the ability of trunk acceleration measures to distinguish between walking patterns of the elderly with or without stability problems and compared these results to young adults. They found that the older adults with balance problems exhibited a decrease in smooth trunk movements when compared to young and healthy OA, indicating that individuals with stability problems have a greater difficulty controlling their trunk movements. They also found that within stride variability of HRs differentiated between the OA with and without stability problems, thus reflecting disturbances in forces applied to maintain trunk stability.

Menz et al. (2003) further studied acceleration patterns of head and pelvis in YA and OA in three orthogonal directions (V, AP and ML). The authors used HRs of the head and the pelvis as an indicator of stability. A stable rhythmic gait pattern was expected to have acceleration pattern that repeat in multiples of two within any given stride (two steps). Acceleration patterns that deviated were considered as being out of phase and a potential indicator of gait instability.

Older adults have a less control over trunk displacement as compared to YA (Winter, Patla, Frank & Walt, 1990). People with a high risk of falling exhibit smaller AP, ML and V-HR while walking (Menz et al., 2003). The HRs decrease significantly in the V direction while walking on irregular surfaces (Menz et al., 2003). Based on these findings it could be hypothesized that the OA with balance problems would exhibit lower HR in the three directions.
Parkinson’s Disease and Gait

Parkinson's disease (PD) negatively affects movement including gait. The gait is typically characterized by short shuffling steps, greater stride variability, shorter step length, episodic freezing, and an increased susceptibility to fall while turning (Ashburn et al., 2001). PD is an extrapyramidal, progressive, degenerative motor disorder usually affecting OA, resulting from a decrease in the neurotransmitter dopamine (DA) secreted in the substantia nigra in the basal ganglia.

This dysfunction of the basal ganglia leads to symptoms that present as motor disturbances. This disturbance in the complex basal ganglia circuitry leads to problems with timing and scaling of movements manifesting as difficulty in movement initiation (akinesia), generalized slowing of movement (bradykinesia), as well as poor posture, and gait instability (for a review see Bartels & Leenders, 2009). Due to the basal ganglia involvement the ability to internally generate or represent a movement is also diminished in PD. It has been found that sensory cues help in improving movement (Nieuwboer, Feys, Weerdt, & Dom, 1997).

Another problem that affects movement in PD is a generalized decrease in movement amplitude clinically referred to as hypometria. As a result of this the movements of PD patients become smaller. An example of this is micrographia, where the stroke size in handwriting becomes smaller (Broderick, Van Gemmert, Shill, & Stelmach, 2009).

In the review by Bartels and Leenders (2009), the authors explained the influence of the basal ganglia on the tone of muscles and the integral role in maintenance of posture and equilibrium. This control is exerted by facilitation of the desired motion while inhibiting the undesired ones through the basal ganglia pathways. The network of circuits is essential for movement preparation and execution in form of feedforward and feedback mechanisms. Involvement of the basal ganglia in PD thus leads to a marked disturbance in equilibrium. This
instability is eventually reflected, in the walking patterns of PD patients in form of loss of balance, co-ordination, and gait abnormalities. Due to the nature of the PD, at an advanced stage it is very difficult to alter gait patterns in these patients, because along with further degeneration, the patients tend to adapt to a gait of their own that allows maximum stability and balance control. PD patients at a later stage have an inclination towards a festinating gait, which is characterized by short, shuffling steps and an increased step velocity.

PD symptoms like rigidity, bradykinesia, and akinesia lead to instability during upright posture, with an increased risk of falling during ambulation (van Wegen, van Emmerik, Wagenaar, & Ellis, 2001). This adaptation results in altered gait rhythmicity stemming from the stride variability. Research shows the stride variability in PD increases with walking speeds lower than the optimal value (Schaafsma et al., 2003; Thaut et al., 1996; Rochester et al., 2005). Additional factors further increasing the gait variability are slowing of movements (bradykinesia) and a slower speed of walking.

Toledo et al. (2005) found a relation between speed of walking and gait variability in PD. The authors recorded stride and swing variability in 36 patients with PD and 30 healthy adults who walked on a treadmill at four different speeds. The walking speeds were manipulated based on the preferred walking. The results showed a decrease in stride length and average swing time with increased stride variability in PD. Due to the PD pathology, the ability to maintain a steady gait rhythm and a stable walking pattern with minimum stride to stride changes is difficult (Schaafsma et al., 2003). The authors reasoned the slowing of gait observed could be due to a fear of falling causing self imposed restrictions while walking (Ashburn et al., 2001; Bloem, Hausdorff, Visser, & Giladi, 2004).
Plotnik, Giladi, and Hausdorff (2008) demonstrated that PD patients have a difficulty in stride-stride adjustments. This difficulty increases in PD patients demonstrating freezing of gait (FOG) while walking. The authors compared the gait of PD patients with a history of FOG to PD patients without FOG. The participants were given the FOG questionnaire to self report the history of FOG (Giladi et al., 2000). A visual analog scale was used for subjective ranking of motor state. The participants ranked their present motor states on a scale of 0 (worst state) to 10 (best state). The testing was done in both “on” and “off” mobile stages. The patients got up from a chair and walked a total distance of 80m. The walking protocol included turning and returning back to a seated position. The results showed the participants with a history of FOG had a greater impaired ability to regulate stride to stride variations with an inability to control the cadence when compared to the PD patients without FOG. The study suggested patients with arrhythmic walking patterns show a higher predisposition to FOG. Research has shown that arrhythmic walking patterns are also due to impaired postural strategies to maintain balance while walking.

Few studies have investigated inefficient postural strategies in PD. Mesure, Azulay, Pouget, and Amblard (1999) compared the postural strategies adopted by PD patients while walking to those of the age-matched control group. They examined head and trunk stabilization methods in the sagittal and frontal direction while walking at a preferred pace on flat ground. The methods of segmental stabilization were determined by correlating angular movements of body segments. The researchers concluded that the PD patients walked with a shorter step length, greater step width and a slower gait velocity than the controls. While there was no significant difference in angular dispersion of the body segments, the PD patients did adopt a strategy of an “en bloc” functioning of the head-shoulder unit to achieve head stabilization on shoulder while walking. The authors explained this strategy as a possible attempt to control degrees of freedom
and maintain stability while walking. They proposed that in PD, limiting head angular oscillations could be a compensatory approach to maintain walking stability in the lateral direction by reducing lateral trunk oscillations.

In order to address the sensory-motor challenges in PD, many authors have suggested external cueing strategies to improve balance and stability. Studies have shown that PD gait can be improved by external cueing, but few studies have addressed the underlying mechanism of gait disturbances in PD. Hanakawa et al. (1999) investigated and compared gait induced cortical activity between ten age matched controls and ten PD patients. The participants walked at a preferred stride length on a treadmill at a predetermined speed of 13m/min for 5 minutes. To control the effect of visual inputs, participants were asked to look ahead towards a white wall. It was observed that the PD patients walked with a typical hypokinetic gait with a higher cadence and shorter stride length as compared to the controls. In the PD patients an under activity was seen in the medial frontal area, right precuneus, and left cerebellar hemisphere. In contrast an over activity was observed in the left temporal cortex, right insula, left cingulate cortex, and the cerebellar vermis. The reduced activity in the medial frontal lobe explains the abnormality of motor performance in PD. The vermis controls the posture and balance in healthy individuals and an over activity in the vermis is suggestive of an attempt to control the loss of lateral gravity shift while walking to maintain gait stability.

In a review on functional neuroimaging of gait, Bakker, Verstapper, Bloem, and Toni (2007) discussed the cortical activation patterns during locomotion in healthy adults and patients with PD. The authors focused on imaging during walking as well as during gait initiation and imagery of gait. The review mentioned a study by Fukuyama et al. (1997) in which they mapped the cerebral activity during walking. The study was among the first of its kind to record
increased activation in supplementary motor area (SMA), sensorimotor areas, striatum, vermis, and the visual cortex in healthy participants. This finding is in contrast to the study by Hanakawa et al., (1999), where the authors demonstrated an under activity in the medial frontal lobe and an over activity in the vermis in PD patients when they walked on the treadmill at a predetermined speed. This over activity could be explained as a compensatory attempt to control lateral stability while walking. The review also addressed the effect of external cueing on gait initiation. According to a study by Yazawa et al. (1997) there is an increase in the activity in the medial frontal cortex during initiation of gait following an external cueing. As a decrease in the activity in the medial frontal cortex has been noted in the PD (Hanakawa et al., 1999), this finding supports the use of external cueing to improve motor efficiency in FOG.

Brown and Marsden (1988) and Buytenhuijis et al. (1994) found that PD patients depend extensively on explicit cues for prompting an appropriate response. This dependence is in contrast to the healthy individuals who depend on internally generated movements or implicit cues (Spaendonck, Berger, Horstink, Borm, & Cools, 1995). Studies support that PD patients perform as well as the controls on problem solving tasks when presented with cues. However when the cues are withdrawn their problem solving capacity deteriorates drastically (Barbeau, 1974). There continues to be a debate on the nature and specificity of the cue that could help the PD patients.

**Directional Instability in PD**

Studies have found that PD patients have inefficient postural strategies and demonstrate directional instability. Supporting evidence can be found in a study by van Wegen et al. (2001) in which the authors examined boundary related postural control while leaning during quiet standing by asking participants to sway forwards or backwards without bending at the hips. The
geometric stability boundary was defined by the feet along with position and variability of the center of pressure. They measured the change in distribution of center of pressure of the foot during the sway. The results showed that there was an overall increase in variability of the center of pressure in the OA and PD groups. However when compared to the healthy age-matched OA, the PD group showed more variability in ML center of pressure. The boundary relevant center of pressure measures thus confirmed a change in control strategies and the presence of lateral instability in PD.

King and Horak (2008) also found that people with PD have greater lateral instability. They used a movable platform with lateral translations and observed three postural strategies: lateral side-step, crossover step, and no step. Latency to step following perturbations, step length, step velocity, and anticipatory postural adjustments were recorded. Results indicated that the two groups used similar lateral stepping strategies, although the PD group’s responses to the perturbation and recovery were delayed. The authors concluded that the PD group showed a lack of anticipatory weight shift as well as bradykinesia (slowness), thus contributing to more postural instability and a susceptibility to falls.

Mitchell, Collins, De Luca, Burrows, and Lipsitz (1995) also found an increased lateral instability in people with PD. They showed an increase in ML sway during quiet standing in PD as compared to age matched controls. They thought this could be a compensatory mechanism for AP instability based on previous research in which sway patterns in PD displayed an increase in ML excursion as AP movement of center of pressure increased (Archer, Winter, & Prince, 1994.)

Horak, Dimitrova, and Nutt (2005) studied response to direction-specific perturbations and the role of stance width instability to evaluate the directional prevalence of falls in PD. The researchers manipulated the base of support (narrow or wide stance width) and recorded postural
response to surface translations in eight directions. The stability value for reference was quantified as the difference between the peak center of pressure and the peak center of mass in response to the perturbations. While the control group maintained a consistent stability margin, the PD participants had smaller than normal stability margins in all directions, especially for the backward sway in both stance widths and lateral sway in the narrow stance width. The excessive displacement of the body center of mass was attributed to the lack of trunk flexibility for the lateral sway and restricted knee flexion for the backward sway. There was a relation between the stability margins and scores on the Unified Parkinson’s Disease Rating Scale; stability margins decreased with increased severity of the disease. The authors concluded that the PD patients have direction-specific instability due to the increased inflexibility and inability to modify the postural response to situations displacing the center of mass due to a change in direction. Together, these studies emphasize the lack of postural stability and the presence of balance impairment as a result of altered mechanics in people with PD. This direction-specific instability can be measured reliably using HRs.

**Harmonic Ratios as a Measurement of Gait Instability**

Yack and Berger (1993) encouraged the study of walking for a dynamic global measure of stability rather than standing balance. They reasoned that challenges to stability occur when a person is moving and hence measuring gait parameters during walking would be a more appropriate form of assessing stability. However, a complex pattern such as walking is difficult to quantify using a simple objective criterion. Measuring the spatio-temporal parameters of gait that are easy to quantify are not necessarily sensitive to detect subtle gait changes. At the same time observed changes in gait provide a qualitative measurement but do not quantify the
changes. These problems in measuring stability while walking could be addressed with more sophisticated measures for evaluation of gait. One such method is use of HRs.

People with PD may be less stable in the ML direction compared to age-matched controls. Lowry et al. (2009) examined the differences in HRs between PD and healthy OA using a triaxial accelerometer to measure trunk accelerations. They also examined the relation between HRs and stride parameters. The authors, using standardized HRs and spatio-temporal parameters to control for the influence of gait velocity, found that the PD group showed a lowered HR in all directions especially AP and ML. In addition, greater variability in stride time was related to lower AP-HRs, while the severity of PD was correlated with lower ML-HRs.

Trunk accelerations using HRs are thus an efficient method of assessing balance control during walking and have been used as an evaluative parameter to distinguish between the healthy and frail OAs more effectively than temporal parameters (Moe-Nilssen & Helbostad, 2005). Similarly, it is thought that HRs may be more sensitive to gait deviations early in PD compared to temporal parameters (Lowry et al., 2009). Supporting evidence shows that PD patients demonstrate reduced and more variable trunk accelerations, backing the sensitivity and efficiency of HRs as a measure for assessing stability (van Emmerik, Wagenaar, Winogrodzka, & Wolters, 1999).

**Purpose**

The purpose of this study was to examine the relationship between step width and ML stability using HRs in PD. This relationship was examined by:

1) measuring step width, step width variability, and HRs during preferred gait in PD and age-matched controls;

2) manipulating step width (wide or narrow base of support) and measuring differences in
HRs, with a focus on the ML-HR, in PD and the controls.

It was hypothesized that people with PD will exhibit wider step width and greater step width variability, and will exhibit lower HRs in all three directions during preferred walking compared to age-matched healthy controls. It was hypothesized that walking with a wider step would increase ML-HRs compared to preferred walking for all participants, and walking with a narrow step width would lower ML-HRs for all participants compared to preferred, but the change would be greater in people with PD compared to age-matched controls. Finally, it was hypothesized that AP- and V-HRs would be lower for all participants when step width was manipulated.
CHAPTER 3.

METHODOLOGY

Participants

Nineteen PD patients and 19 age-matched controls between the ages of 49 and 79 participated in the study. The exclusionary criteria for all participants included: 1) neurological damage or disease other than PD; 2) use of assistive devices while walking indoors; 3) presence of dyskinesia during gait; 4) moderate to severe dementia (score on Mini-Mental State Examination < 24) or depression (score on Geriatric Depression scale ≤ 19); 5) recent cardiac event, significant cardiovascular disease such as congestive heart failure, or high blood pressure not controlled by medication; and 6) musculoskeletal impairment such as pain during walking, recent trauma, or presence of any spinal disorder that affected accelerometer placement. The PD patients were on medication and were tested during their mobile or ‘on’ phase.

Paper pencil tests were administered to characterize the samples. The Activities-specific Balance Confidence Scale (ABC), Geriatric Depression Measure (GDM) and Mini Mental State Examination (MMSE) questionnaire were administered to all participants and gait portions of the Unified Parkinson’s Disease Rating Scale (UPDRS) were administered to characterize the severity of PD. The statistical analysis of the ABC scores revealed that the participants were fairly confident about their ability to perform daily activities. Even though the PD group did show slightly lower scores these were not statistically different from that of the controls. Similarly while the participants were not depressed, the PD group did show slightly higher but not statistically different GDM scores (the GDM score of one participant from the PD group was imputed by the group mean to compensate for the missing data). The MMSE showed that participants were did not have dementia. (See Table 2 for the means and standard deviations
PD patients with prominent tremors, rigidity, and severe balance and gait impairments were excluded from the study. The study followed the IRB guidelines and participants signed a consent form before beginning the study.

**Apparatus**

Footfalls were measured using the GAIT Rite® (GAITRite Gold, CIR Systems, Inc.) electronic walkway. It records temporal and spatial gait parameters such as step length, step width, step velocity, stride width, and cadence (8.2m long; active area 61cm x 732cm; separation between sensors 1.27cm; sampling rate 80Hz). To measure accelerations, a triaxial accelerometer (Crossbow CXLO2LF3, range ±2g) recorded trunk accelerations in x, y, z directions (anterior, posterior and vertical). The accelerometer was mounted on a belt around the subject’s waist at the level of L2-L3 vertebrae. A data logger (Crossbow AD2000 Ready DAQ) with a sampling frequency of 200Hz was placed in a small backpack worn by the subject.

**Tasks and Conditions**

Participants were asked to walk approximately 9m with the GaitRite® centered in the walkway (there were at least two strides before and after the GaitRite®). There were four walking conditions: preferred pace, preferred pace with metronome, wide step width with metronome and narrow step width with metronome. The rate of the metronome was the same pace as the average of the trials of their preferred walking rate and was used to encourage the participants to maintain their speed under step width manipulations. They were instructed to match their pace to the beats of the metronome. Participants completed five trials of each condition. The first condition was always preferred, followed by preferred walking with a metronome. Then wide and narrow walking conditions were counter-balanced across participants.
Procedures

After reading and signing the consent form, participants’ anterior superior iliac spine width (ASIS) and leg length were measured. No significant differences were found within the ASIS and leg length measurements of the groups. The wide width walking condition was equal to their anterior-superior iliac spine (ASIS), and the narrow width was ½ of their ASIS. After measurements were taken, they were also asked to balance on each leg as long as possible up to 45 seconds while looking at a circle that was placed at eye level. They completed two trials on each leg. The ability to balance was further tested by asking the participants to stand in a Tandem (heel to toe) stance for 45 seconds.

Participants then completed the four walking conditions. After preferred walking, continuous taped lines were placed on the walkway as guidelines for the participants. For the wide condition, the distance between the lines was equal to their ASIS and they were asked to walk either outside the lines. For the narrow condition, the distance between the lines was ½ ASIS and they were asked to walk inside the lines. They were told not to aim their footfalls, but to use the tape as a guideline to either widen or narrow their step while walking. After each condition and between trials the placement of the accelerometer was checked and if required leveled, to maintain accuracy of the collected data.

Data Processing and Statistical Analyses

Primary dependent variables were HRs in each direction based on acceleration data. The AP and V accelerations during walking exhibit two major acceleration peaks per stride. As each stride has two steps, it could be said that the accelerations are biphasic for a step. Frequency decomposition through Fourier analysis yields dominance to the even harmonics, where the even harmonics represent in-phase components and the odd harmonics represent out-of-phase
components of gait. It was expected that the out-of-phase components would be minimized in a healthy gait. Anterior-posterior and V-HRs are calculated by dividing even harmonics by odd harmonics. A higher HR suggests smoother and more stable gait (Yack & Berger, 1993; Lowry et al., 2009).

Accelerations in ML plane are monophasic exhibiting one acceleration peak per stride, leading to a dominance of odd harmonics. The odd harmonics are in-phase while the even harmonics are out-of-phase. Therefore, the ML-HR is calculated from the ratio of odd harmonics divided by even harmonics, and again, higher indicative of smoother gait.

Each trial of acceleration data was processed using custom software (Visual Basic software incorporating National Instruments Measurement Studio™ 6.0 libraries). Using this software, each trial was examined for accurate stride identification. The first two strides and last two strides for each trial were dropped to reduce acceleration and deceleration effects. Harmonic ratios for the remaining strides were calculated for each stride, then averaged across strides within one trial, and then averaged across trials. The first trial was dropped, as we considered this practice. Thus, averaged data reflected four trials.

We also measured velocity, cadence, step time, step time variability, step length, step length variability, step width, step width variability, and percent time in double support based on footfalls on the GAIT Rite®. Velocity (m/s) was based on time to traverse the middle 9 m of the walkway and was calculated by averaging performance across trials 2 through 5. Percent time in double support was the sum of initial double support added to terminal double support and was also calculated by averaging performance across trials 2 through 5. Step time was the time elapsed from first contact of one foot to first contact of the opposite foot. Step length was measured as line of progression, from the heel center of the current footprint to the heel center of
the previous footprint on the opposite side. Step width was measured as midline midpoint of the
current footprint to the midline midpoint of the previous footprint on the opposite foot. Cadence
was calculated as number of steps per minute. Variability of performance was based on the
within subject standard deviations of the means for each variable.

The means and standard deviations of these step variables were calculated by averaging
across steps for all four trials. As each trial had at least four steps on the GaitRite (or two
strides), a minimum of 16 steps (the range being 16-20) were analyzed for each condition.

To test the first hypothesis, a one way Analysis of Variance (ANOVA) was conducted to
examine if there were group differences in the dependent variables. To test hypotheses two and
three a 2 (group) x 3 (condition) Repeated Measures ANOVA was conducted for each variable.
Support for hypothesis two (comparing wide to preferred step width) would be a main effect for
group. Support for hypothesis 3 (comparing narrow to preferred step width) would be an
interaction.
CHAPTER 4.

RESULTS

Preferred Gait

Spatio-temporal variables. The means and standard for all spatio-temporal variables are presented in Table 3. It was expected that the PD group’s preferred velocity would be slower and step length shorter than age-matched controls. Mean values were consistent with these expectations, but one-way ANOVAs revealed a nonsignificant group difference in velocity and only a trend for step length (p = .065). Step length variability was higher for the PD group but not significant. Step time was shorter and percent time in double support was higher, but neither was statistically significant. The only spatio-temporal variable that was significantly different between groups was step time variability, F(1,36) = 6.65, p = .014, with the PD group exhibiting greater variability. It was hypothesized that step width and step width variability would be greater in the PD group, but a one-way ANOVA revealed only a trend for the PD group to walk with a narrower (not wider) step width (p = .066). Step width variability was greater in the PD group, but was not significant.
**FIGURE 1.** Average step time (ST) variability during preferred walking (PF). Error bars are standard deviations.

**FIGURE 2.** Average step time (ST) during preferred walking (PF). Error bars are standard deviations.

**Harmonic Ratios.** Based on previous research, we hypothesized that the PD group would exhibit lower HRs in all three directions. The mean values were consistent with these hypotheses, however only AP direction was significant, $F(1,36) = 5.30, p = .027$. There was also
a weak trend in the V direction (p = .104). (See Table 3 for HR means and standard deviations.)

FIGURE 3. Average mediolateral harmonic ratios (ML-HR) during preferred walking (PF).

Error bars are standard deviations.

FIGURE 4. Average anterior-posterior harmonic ratios (AP-HR) during preferred walking (PF).

Error bars are standard deviations.
FIGURE 5. Average vertical harmonic ratios (V-HR) during preferred walking (PF). Error bars are standard deviations.

Experimental Conditions

Spatio-temporal variables. To determine if participants complied with the experimental instructions we conducted a repeated measures ANOVA across three conditions (preferred pace with metronome, wide with metronome, and narrow with metronome). The Group x Condition ANOVA with repeated measures on the second factor revealed a Condition main effect, $F(2,72) = 60.64$, $p < .001$. Post-hoc tests indicated that wide was significantly wider than preferred, and there was a trend for narrow to be more narrow than preferred ($p = .06$). There was no Group x Condition interaction indicating that both groups equally complied. There was also a Group main effect, $F(1,36) = 5.08$, $p = .03$, with the PD group exhibiting a more narrow step width across conditions. There were no significant effects for step width variability. (See Table 3 for means and standard deviations.)

There was a trend for the PD group to walk more slowly than the control group ($p = .098$). There was a Condition main effect, $F(2, 72) = 3.807$, $p = .027$, but post hoc tests indicated
only a trend for wide to be slower than preferred metronome (p = .052). There was no Group x Condition interaction. Both groups exhibited greater double support cycle time for the wide condition, but this increase was not significant. There were also no signification differences for cadence.

There was a significant interaction for step time, $F(2, 72) = 3.23, p = .045$, with both groups walking slower following step width manipulation. A one-way ANOVA was conducted to compare step time between groups in the narrow condition (the condition in which the difference was the greatest between groups) and it revealed that there was a trend for the groups to be different (p = .06). For step length, there was a Group main effect, $F(1, 36) = 4.995, p = .032$, with the PD group exhibiting shorter steps, but no interaction. There was also a Condition main effect for step length, $F(2,72) = 4.37, p = .016$, with a shorter step length in the wider condition compared to the preferred metronome (p = .024), and a trend for step length to be shorter in the narrow condition (p = .092).

**FIGURE 6.** Average step time (ST) following step width manipulation. Error bars are standard deviations.
There was a Condition main effect for step time variability, $F(2,72) = 7.294$, $p < .001$, with variability greater in the wide condition ($p < .001$) and a trend in the narrow condition ($p = .051$). There was also a Group main effect, with the PD group exhibiting greater step time variability, $F(1,36) = 6.36$, $p = .016$. There was a Condition main effect for step length variability, $F(2,72) = 9.83$, $p < .001$, with greater variability in the wide ($p < .001$) and narrow ($p = .03$) conditions. There were no significant interactions for any measure of variability.

**FIGURE 7.** Average step time (ST) variability following step width manipulation. Error bars are standard deviations.

**Harmonic Ratios.** Group x Condition ANOVA’s revealed a significant main effect for Condition in the ML-HR, $F(2, 72) = 25.31$, $p < .001$, with a higher HR in the wide condition and a lower HR in the narrow condition. Post-hoc comparisons indicated that only the decrease in the narrow condition was significant ($p < .001$). There was also a Group main effect, $F(1, 36) = 4.65$, $p = .038$, with the PD group exhibiting a lower ML-HR. (See Table 3 for means and SDs.)
FIGURE 8. Average mediolateral harmonic ratios (ML-HR) following step width manipulation. Error bars are standard deviations.

In the AP direction there was a Condition main effect, $F(2, 72) = 55.62, p < .001$, with the HR lower in wide and narrow compared to preferred metronome. Post-hoc tests revealed that both were significant ($p < .001$). There was also a Group main effect, $F(1, 36) = 6.58, p = .015$, with the PD group lower. There was no interaction. In the V direction there was a Condition main effect, $F(2, 72) = 13.21, p < .001$, with HRs lower in wide and narrow compared to preferred metronome ($p < .001$). The PD group exhibited lower HRs in the V direction, but this difference did not reach significance.
FIGURE 9. Average anterior-posterior harmonic ratios (AP-HR) following step width manipulation. Error bars are standard deviations.

FIGURE 10. Average vertical harmonic ratios (V-HR) following step width manipulation. Error bars are standard deviations.
### TABLE 1. Severity of disease and list of medications for people with Parkinson’s disease.

<table>
<thead>
<tr>
<th>PD</th>
<th>Age</th>
<th>Gender</th>
<th>Unified Parkinson’s Disease Rating Scale for Gait (0-16)</th>
<th>Medication*</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>65</td>
<td>F</td>
<td>2</td>
<td>L/C, Ra, RHCL</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>70</td>
<td>M</td>
<td>1</td>
<td>Ra, AHCL, RHCL, L-dopa</td>
<td>Rigid when walking; problems with coordination (when meds not working)</td>
</tr>
<tr>
<td>3</td>
<td>62</td>
<td>M</td>
<td>0</td>
<td>C/L/E, SHCL, RHCL</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>49</td>
<td>F</td>
<td>2</td>
<td>C/L/E, RHCL, AHCL, Zo, Ra</td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>66</td>
<td>M</td>
<td>2</td>
<td>L/C, AHCL</td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>60</td>
<td>M</td>
<td>0</td>
<td>L/C</td>
<td>Problems with coordination; uses equipment for walking</td>
</tr>
<tr>
<td>7</td>
<td>60</td>
<td>F</td>
<td>0</td>
<td>Ra, Tri, AHCL, C/L/E</td>
<td>Hand tremors</td>
</tr>
<tr>
<td>8</td>
<td>79</td>
<td>M</td>
<td>3</td>
<td>AHCL, AB, SHCL, PP</td>
<td></td>
</tr>
<tr>
<td>9</td>
<td>56</td>
<td>F</td>
<td>3</td>
<td>PP, AHCL, C/L/E, EO</td>
<td>Stiffness and rigidity in hands</td>
</tr>
<tr>
<td>10</td>
<td>62</td>
<td>M</td>
<td>3</td>
<td>C/L/E, PP, Ra</td>
<td>Out of breath when walking</td>
</tr>
<tr>
<td>11</td>
<td>69</td>
<td>F</td>
<td>2</td>
<td>L/C, Ra</td>
<td>Problems with fine motor movements; dizziness at times</td>
</tr>
<tr>
<td>12</td>
<td>53</td>
<td>F</td>
<td>2</td>
<td>RHCL, Ra</td>
<td></td>
</tr>
<tr>
<td>13</td>
<td>78</td>
<td>M</td>
<td>3</td>
<td>L/C</td>
<td>Uses cane outdoors; problems walking up and down stairs; macular degeneration</td>
</tr>
<tr>
<td>14</td>
<td>52</td>
<td>F</td>
<td>1</td>
<td>Ra</td>
<td></td>
</tr>
<tr>
<td>15</td>
<td>75</td>
<td>M</td>
<td>3</td>
<td>L/C</td>
<td>Problems walking up and down stairs and with coordination; spinal fusion (L4-L5)</td>
</tr>
<tr>
<td>16</td>
<td>53</td>
<td>F</td>
<td>3</td>
<td></td>
<td></td>
</tr>
<tr>
<td>17</td>
<td>74</td>
<td>M</td>
<td>5</td>
<td>L/C</td>
<td>Problems walking up and down stairs and with coordination</td>
</tr>
<tr>
<td>18</td>
<td>68</td>
<td>F</td>
<td>2</td>
<td></td>
<td>Morning stiffness</td>
</tr>
<tr>
<td>19</td>
<td>72</td>
<td>F</td>
<td>2</td>
<td>L/C</td>
<td>Problems with coordination</td>
</tr>
</tbody>
</table>

*Generic names for medication: Levodopa with Carbidopa (L/C); Rasagiline (Ra); Ropinirole hydrochloride(RHCL); Amantidine hydrochloride (AHCL); Carbidopa/levodopa/entacapone(C/L/E); Zonisamide(ZO); Trihexphenidyl(Tri); Amlodipine Besylate(AB); Selegiline HCL(SHCL); Pramipexole(PP); Escitalopram Oxalate(EO); Levodopa(L-dopa).
TABLE 2. Performance characteristics for each group.

Tests include longest time in balance in single leg stance (SLS) and tandem stance, Geriatric Depression Measure (GDM), Mini Mental State Examination (MMSE), and Activities-specific Balance Confidence Scale (ABC). Mean is reported with standard deviation (SD) in parentheses and group range below.

<table>
<thead>
<tr>
<th></th>
<th>PD</th>
<th>Controls</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Age</strong></td>
<td>64.4 (9.13)</td>
<td>65.2 (8.05)</td>
</tr>
<tr>
<td></td>
<td>49-79</td>
<td>51-77</td>
</tr>
<tr>
<td><strong>ASIS Width</strong></td>
<td>27.70 (1.83)</td>
<td>27.30 (1.89)</td>
</tr>
<tr>
<td>(cm)</td>
<td>25.5-32</td>
<td>24.5-31.5</td>
</tr>
<tr>
<td><strong>Leg length</strong></td>
<td>91.05 (4.31)</td>
<td>91.26 (4.30)</td>
</tr>
<tr>
<td>(cm)</td>
<td>82.5-99.5</td>
<td>82-99.5</td>
</tr>
<tr>
<td><strong>Balance</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(max of 45 sec.)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>SLS (sec)</td>
<td>20 (15)</td>
<td>28 (17.11)</td>
</tr>
<tr>
<td></td>
<td>3-45</td>
<td>3-45</td>
</tr>
<tr>
<td>Tandem (sec)</td>
<td>34 (15)</td>
<td>43 (4.53)</td>
</tr>
<tr>
<td></td>
<td>5-45</td>
<td>30-45</td>
</tr>
<tr>
<td><strong>GDM</strong></td>
<td>6 (4)</td>
<td>2 (2.17)</td>
</tr>
<tr>
<td>0-9 normal</td>
<td>1-15</td>
<td>1-7</td>
</tr>
<tr>
<td>10-19 mild</td>
<td></td>
<td></td>
</tr>
<tr>
<td>≥20 severe</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>MMSE</strong> (≤24 dementia)</td>
<td>29.16 (1.17)</td>
<td>29.53 (.96)</td>
</tr>
<tr>
<td>≤ 9 severe</td>
<td>26-30</td>
<td>27-30</td>
</tr>
<tr>
<td>10-20 moderate</td>
<td></td>
<td></td>
</tr>
<tr>
<td>21-24 mild</td>
<td></td>
<td></td>
</tr>
<tr>
<td><strong>ABC</strong></td>
<td>85 (10)</td>
<td>97 (2.95)</td>
</tr>
<tr>
<td>0 = no confidence</td>
<td>59-97</td>
<td>93-100</td>
</tr>
<tr>
<td>50 = somewhat confident</td>
<td></td>
<td></td>
</tr>
<tr>
<td>100 = confident</td>
<td></td>
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</tbody>
</table>
TABLE 3. Spatio-temporal variables in PD and controls with mean (and SD).

<table>
<thead>
<tr>
<th>Variable</th>
<th>PF</th>
<th>PM</th>
<th>WD</th>
<th>NW</th>
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<tbody>
<tr>
<td></td>
<td>PD</td>
<td>Control</td>
<td>PD</td>
<td>Control</td>
</tr>
<tr>
<td>Velocity (m/s)</td>
<td>1.35 (.21)</td>
<td>1.44 (.16)</td>
<td>1.39 (.21)</td>
<td>1.49 (.15)</td>
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<tr>
<td>Cadence (steps/min)</td>
<td>117 (7.51)</td>
<td>116 (5.89)</td>
<td>119 (8.37)</td>
<td>118 (6.03)</td>
</tr>
<tr>
<td>% Time Double Support</td>
<td>24 (3.23)</td>
<td>23 (2.55)</td>
<td>25 (5.23)</td>
<td>22 (2.61)</td>
</tr>
<tr>
<td>Step Time (ms)</td>
<td>515 (31)</td>
<td>516 (23)</td>
<td>505 (34)</td>
<td>508 (26)</td>
</tr>
<tr>
<td>Step Time SD</td>
<td>19.3 (6.8)</td>
<td>14.8 (3.3)</td>
<td>19.2 (5.8)</td>
<td>14.7 (4.8)</td>
</tr>
<tr>
<td>Step Length (cm)</td>
<td>69.3 (8.7)</td>
<td>74.2 (7.1)</td>
<td>70.3 (8.0)</td>
<td>76.0 (6.8)</td>
</tr>
<tr>
<td>Step Length SD</td>
<td>2.21 (0.67)</td>
<td>2.03 (0.82)</td>
<td>2.68 (1.15)</td>
<td>2.11 (0.66)</td>
</tr>
<tr>
<td>Step Width (cm)</td>
<td>7.0 (0.8)</td>
<td>7.5 (0.7)</td>
<td>7.1 (0.8)</td>
<td>7.6 (0.8)</td>
</tr>
<tr>
<td>Step Width SD</td>
<td>.23 (0.80)</td>
<td>.20 (0.07)</td>
<td>.27 (.12)</td>
<td>.39 (.46)</td>
</tr>
<tr>
<td>ML HR</td>
<td>2.32 (0.68)</td>
<td>2.60 (0.60)</td>
<td>2.41 (0.71)</td>
<td>2.90 (0.56)</td>
</tr>
<tr>
<td>AP HR</td>
<td>3.21 (1.17)</td>
<td>3.98 (0.87)</td>
<td>3.34 (1.14)</td>
<td>4.07 (0.75)</td>
</tr>
<tr>
<td>V HR</td>
<td>3.04 (0.84)</td>
<td>3.46 (0.70)</td>
<td>3.23 (1.05)</td>
<td>3.63 (0.66)</td>
</tr>
</tbody>
</table>

* Abbreviations: Harmonic Ratios (HR); Mediolateral (ML); Anterior-posterior (AP); Vertical (V);
CHAPTER 5.

DISCUSSION

The purpose of this study was to examine the relationship between step width and ML stability using HRs in people with Parkinson’s disease and age matched controls. This relationship was examined by: 1) measuring step width, step width variability, and HRs during preferred gait in PD and age-matched controls; and 2) manipulating step width (wide or narrow base of support) and measuring differences in HRs, with a focus on the ML-HR, in PD and the controls.

HRs represent trunk accelerations and have been found to be more sensitive to gait changes especially in populations walking at a different speed (Moe-Nilsen & Helbostad, 2005). It was hypothesized that while walking at a preferred gait the PD group would exhibit wider step width, greater step width variability, and lower HRs compared to the controls. We found that the PD group actually walked with a narrower step width with no difference in step width variability. We also found that the PD group did exhibit lower HRs in all three directions, with AP-HRs reaching statistical significance and a trend in the V direction. This finding is consistent with that of Lowry et al. (2009).

An explanation for the lack of statistically lower ML-HRs could be that ML balance is controlled by active foot placement (Bauby & Kuo, 2000). It has been found that movement in the AP direction is different from that of ML direction while walking. ML balance has been found to be unstable and the control is based on feedback-driven lateral foot placement (Donelan, Shipman, Kram & Kuo, 2004). The ‘passive’ control of balance in the AP direction is at a lower level and is based on sensory information from limbs. In contrast to this, ML stability requires an ‘active’ control from the higher centers, essential for ML stabilization (Bauby &
Kuo, 2000).

Lowry, Carrel, McIlrath & Smiley-Oyen (2010) examined the effect of verbal and cognitive cueing strategies on improving gait stability in PD. They found that verbal and cognitive cueing strategies improved balance only in AP and V directions. The authors also found that balance in ML direction was not enhanced. This finding further emphasizes the need for an active control to maintain ML stability. Since the PD group was not in an advanced stage of disease, they could possibly adjust their gait to maintain the stability in ML direction. This is consistent with Lowry et al. (2009) in which severity of disease was found to be related to ML-HRs.

Interestingly, it was found that the PD group walked with a narrower step width in preferred, counter to the hypothesis. This may be the result of hypometria where there is a decrease in the movement amplitude. Supporting evidence can be found in studies that show the force-production characteristics in PD are different from the healthy population. Wing (1998), found that force is modulated at a lower rate in PD. Likewise Stelmach and Worricham (1988) found that initiation as well as development of force is slower in PD. This leads to smaller and slower movements that could manifest in gait as walking with a narrower step width. As hypometria increases with progression of the disease, the lowering of ML-HRs with a greater disease severity could be due to continued decrease in step width while walking. More research is needed to examine step width later in the disease and/or while participants are off their PD medications.

As expected, both groups responded similarly to widening step width, with HRs decreasing in AP and V directions. It is interesting that ML-HRs were maintained (actually slightly increased in the PD group). In the narrow step condition, HRs were lower in all three
directions for both groups, but counter to our expectations, the decrease was not greater in the PD group. In addition, as expected, the PD group exhibited a longer step time and shorter step length in the narrow condition. The results of ML-HRs with step width manipulations indicate that step width contributes to ML dynamic stability. This finding is consistent with the findings of Lowry et al. (2010) where the authors found that, while cueing strategies helped in improving spatio-temporal parameters, it did not enhance the biomechanical models like control of center of mass over a change in the base of support. This control mechanism is crucial for ML stability. The authors suggested a further examination of effect of cueing strategies on step width, a gait parameter influencing active control in the ML direction.

The only measure of variability that differentiated the groups in any of the four conditions was step time variability. An increase in step time variability has been associated with gait variability, which is associated with loss of balance and falls (Richardson, Thies & Ashton-Miller, 2008). This result is consistent with Lowry et al. (2009). The authors found an increase in stride time variability in the PD group and in age matched older adults exhibiting poor balance. They also found that increased stride time variability was associated with lower AP-HR.

Increased temporal gait irregularity, such as an increase in step time or stride time variability, increases gait variability leading to falls (Richardson et al., 2008; Brach, Studenski, Perera, VanSwearingen & Newman, 2007). This increase in variability has been repeatedly found in populations exhibiting slower speed of walking also termed as ‘cautious’ gait (Herman, Giladi, Gurevich & Hausdorff, 2005). The slower speed of walking increases gait variability and indicates an increase in the double support time percent in a gait cycle, which can also be used to directly evaluate control of balance (Gabell & Nayak, 1984). The finding of increased step time variability is consistent with previous research.
There were several limitations with this study. We had a small sample size, wide range in age, and limited range of disease severity, and all the patients were on medication. However, it is important to note that even with these limitations, we found a significant difference in the AP-HR. Another possible limitation in generalizing these results was the use of metronome auditory cues. Based on previous research (Nieuwboer et al., 1997; Thaut et al., 1996), the cues may have improved gait parameters for the PD group. Statistical comparison of preferred gait to metronome gait showed no significant differences in the PD group, thus indicating our cue had minimal effect. In fact, our purpose was to maintain their velocity during step width manipulation whereas other studies used auditory cueing increase gait speed (Thaut et al., 1996). Thus, we think our results can be generalized to walking without auditory cues.

Cools, Berger, Buytenhuijs, Horstink, and Spaendoek (1993) described cues as stimuli associated with behavior executed based on past experiences. As explained by Nieuwboer et al. (1997), external cueing strategies bypass the basal ganglia circuits that are essential for internal initiation of movement and use alternate cortical routes. This bypassing helps improve mobility by acting as a cued retrieval of a motor program. Researchers have also examined the effect of transverse lines for cueing and have found that it significantly improves PD gait (Wang, Wai, Weng, Yu & Wang, 2008). An interesting observation was that while transverse lines improved gait efficiency parallel lines did not enhance gait to the same degree. Thaut et al. (1996) found further support for external cueing through use of rhythmic auditory stimulation, which improved gait parameters like velocity, stride length and step cadence. The authors used a study design with three different music tempos ‘normal’ (pretest cadence), ‘quick’ (5% to 10% faster), and ‘fast’ (an additional 5% to 10% faster) to increase the normal gait velocity of PD patients. The experimental group was trained to walk at a new faster tempo each week. The researchers
took care not to exceed the tempo to more than 130 steps/min. The rate of increase was based on the participant’s ability to match the tempo. The experimental group also walked on a flat and over an inclined surface. The authors found that the gait training improved gait in PD.

In conclusion, this study supports the position that people with PD, even when mildly affected and on medication, show lower HRs, indicating poorer dynamic balance. Another finding of note is that the PD group walked with a narrower step width even in preferred. We conclude that narrowing step width directly affects ML-HRs, with a narrower step width greatly decreasing ML-HRs. Our data are consistent with the position that lower ML-HRs in people with greater PD severity is likely due to continued decrease in step width. Further research is needed to determine this possibility.
Title of Study: Step Width and Walking Stability in Older Adults and Parkinson's Disease

Investigators: Ann Smiley-Oyen, PhD; Kristin Lowry, PT, MS; Sudeshna Chatterjee, PT; Aaron Shoop, BS; Katherine Swanson; Darcy Kruger, Jenna Graven; Stephanie Kirk

This is a research study. Please take your time in deciding if you would like to participate. Please feel free to ask questions at any time.

INTRODUCTION
The purpose of this study is to learn about how different types of walking affect balance control in young adults, older adults, and adults with Parkinson’s disease. You are being invited to participate in this study because you are either a college-age young adult, a healthy older adult, or have been diagnosed with Parkinson’s disease.

DESCRIPTION OF PROCEDURES
If you agree to participate in this study, your participation will involve one session, and that session will last approximately 1 to 1.5 hours. During the study you may expect the following study procedures to be followed. You will be asked to walk over ground: 1) at your preferred speed; 2) at fast and slow speeds; 3) at your preferred speed while counting backwards; 4) with a wider and more narrow stance; and 5) while being cued to take larger steps. While you walk, a trunk accelerometer will record movements of your body. The trunk accelerometer is mounted to a gait belt, which will be secured around your waist as you walk. In addition, you will wear a lightweight backpack that houses a data logger. You will complete 6 walking trials in each condition listed above. You will walk no more than 30 feet in each walking trial. In addition, your height, leg length, and weight will be recorded. You will be given the opportunity to rest as needed during all testing sessions. In addition, you will be asked to complete assessments of functional balance, complete health survey, balance confidence questionnaires, and a general test of general cognitive function. You may skip any question you do not wish to answer or that makes you feel uncomfortable.

RISKS
While participating in this study you may experience the following risks: There is some risk of a loss of balance while walking and during the balance testing. At all times an experimenter will walk or stand closely behind you to guard you.

BENEFITS
If you decide to participate in this study there will be no direct benefit to you. It is hoped that the information gained in this study will benefit society by better understanding how balance control during walking changes with age, and under what conditions older adults may be more
vulnerable to falls. If you choose to have your data retained in the Department of Kinesiology Gait Database, then a written report will be sent to you regarding characteristics of your gait.

COSTS AND COMPENSATION
You will not have any costs from participating in this study. You will not be compensated for participating in this study.

PARTICIPANT RIGHTS
Your participation in this study is completely voluntary and you may refuse to participate or leave the study at any time. If you decide to not participate in the study or leave the study early, it will not result in any penalty or loss of benefits to which you are otherwise entitled.

CONFIDENTIALITY
Records identifying participants will be kept confidential to the extent permitted by applicable laws and regulations and will not be made publicly available. However, federal government regulatory agencies and the Institutional Review Board (a committee that reviews and approves human subject research studies) may inspect and/or copy your records for quality assurance and data analysis. These records may contain private information.

To ensure confidentiality to the extent permitted by law, the following measures will be taken: each subject will be assigned a unique code and this code will be used on forms and in data files. The data will be kept in the locked research lab and on a computer that will be accessible only to people working on the project. If the results are published, your identity will remain confidential.

QUESTIONS OR PROBLEMS
You are encouraged to ask questions at any time during this study. For further information about the study contact Dr. Ann Smiley-Oyen at 294 – 8261. If you have any questions about the rights of research subjects or research-related injury, please contact the IRB Administrator, (515) 294-4566, IRB@iastate.edu, or Director, (515) 294-3115, Office of Research Assurances, Iowa State University, Ames, Iowa 50011.

***************************************************************************

PARTICIPANT SIGNATURE
Your signature indicates that you voluntarily agree to participate in this study, that the study has been explained to you, that you have been given the time to read the document and that your questions have been satisfactorily answered. You will receive a copy of the signed and dated written informed consent prior to your participation in the study.

Subject’s Name (printed)  

(Subject’s Signature)  (Date)
INVESTIGATOR STATEMENT
I certify that the participant has been given adequate time to read and learn about the study and all of their questions have been answered. It is my opinion that the participant understands the purpose, risks, benefits and the procedures that will be followed in this study and has voluntarily agreed to participate.

(Signature of Person Obtaining Informed Consent)   (Date)
APPENDIX B
DATA SHEET

Datasheet Step Width

Fall ‘08/Spring ‘09

Subject ID: ________ Database ID ________ Day & Date: ________________

Consent Form:____________________ Database Consent Form:______________

Health Questionnaire/DOB: ________________

ASIS ________ cm (communicate so tape marks can be started on the GaitRite)

Leg Length: R _______cm L ________cm

Height ________cm Weight ___________lbs. (use biomechanics scale)

**SLS**\_max 45 sec
(comfortable standing position, focus on circle, arms folded, lift leg to level of other ankle, not touching other leg; trial ends when arms leave position, lifted leg braces or touches floor, trunk moves more than 45 degrees, practice once on each leg).

R (standing on) L(standing on)

Trial 1 _______________sec ______________sec

Trial 2 _______________sec ______________sec

**TANDEM**\_max 45 sec
(subject will stand heel to toe, may choose either foot to put forward, subject can be supported while getting into position, time begins after researcher lets go of subject)

Trial 1 _______________sec

Trial 2 _______________sec
MINI MENTAL__________  (OA’s and PD’s only)

UPDRS__________  (PD’s only)

MARK THIS PARTICIPANT’S ORDERS AND MARK THE DATA SHEET ACCORDINGLY

ORDER 1:  NARROW – WIDE
A:  FAST, SLOW, DUAL
ORDER 2:  WIDE - NARROW
B:  SLOW, DUAL, FAST
C:  DUAL, FAST, SLOW
D:  FAST, DUAL, SLOW
E:  SLOW, FAST, DUAL
F:  DUAL, SLOW, FAST

Preferred = walk at your usual, comfortable pace until you reach the tape. Then

<table>
<thead>
<tr>
<th>Condition</th>
<th>Trial #</th>
<th>Sequence #</th>
<th>Comments</th>
</tr>
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<tbody>
<tr>
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<tr>
<td>On/Off</td>
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</tr>
</tbody>
</table>

Preferred w/ metronome = We’ve set the beat of the metronome to your walking pace. This time, match your steps with the metronome. The metronome will stop, continue walking until you reach the tape.
<table>
<thead>
<tr>
<th>Condition</th>
<th>Trial #</th>
<th>Sequence #</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>Metronome</td>
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<td>PM</td>
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<td>5</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

(HAVE OLDER PARTICIPANTS SIT DOWN)

# 50% narrow

Narrow Step Width = ASIS /by 2 /2

Narrow Stance = The metronome is set at the same pace. Like before, match your steps with the metronome but walk inside the lines.

<table>
<thead>
<tr>
<th>Condition</th>
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<th>Sequence #</th>
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<td></td>
<td>4</td>
<td></td>
<td></td>
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<tr>
<td></td>
<td>5</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

(HAVE OLDER PARTICIPANTS SIT DOWN)

# Wide Step Width = ASIS /by 2

Wide Step Width = ASIS /by 2
Wide Stance = The metronome is set at the same pace. Like before, match your steps with the metronome but walk outside the lines.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Trial #</th>
<th>Sequence #</th>
<th>Comments</th>
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<td></td>
<td>3</td>
<td></td>
<td></td>
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<td></td>
<td>4</td>
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<td></td>
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<tr>
<td></td>
<td>On/Off</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

DOWNLOAD CROSSBOW DATA (SAVE AS SWID#_Date)

*TURN OFF METRONOME*

REST/ ABC__________ GDS__________

PD PATIENTS ONLY
Big Step = Take bigger steps than you normally would. (NO METRONOME OR TAPE)
<table>
<thead>
<tr>
<th>Condition</th>
<th>Trial #</th>
<th>Sequence #</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
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<td>Big Steps</td>
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<td></td>
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<tr>
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<td></td>
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<td></td>
<td>5</td>
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<tr>
<td>On/Off</td>
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</tbody>
</table>

Fast Pace = walk as fast as possible without running or taking risks

<table>
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<th>Condition</th>
<th>Trial #</th>
<th>Sequence #</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
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<td></td>
<td>5</td>
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<tr>
<td>On/Off</td>
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</tr>
</tbody>
</table>

Slow Pace = walk slower than preferred, as if you have ample time to get somewhere

<table>
<thead>
<tr>
<th>Condition</th>
<th>Trial #</th>
<th>Sequence #</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>Slow Pace</td>
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<tr>
<td></td>
<td>2</td>
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<tr>
<td>On/Off</td>
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<td></td>
</tr>
</tbody>
</table>
### Dual Task

Dual Task = the experimenter will present you with a number, this is your cue to begin walking. Count backwards, by ones, from the presented number until you reach the endline.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Trial #</th>
<th>Sequence #</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>Slow Pace</td>
<td>1</td>
<td></td>
<td></td>
</tr>
<tr>
<td>SP</td>
<td>2</td>
<td></td>
<td></td>
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<tr>
<td></td>
<td>3</td>
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<tr>
<td>On/Off</td>
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</tbody>
</table>

TAKE OFF ACCEL
DOWNLOAD CROSSBOW DATA (SAVE AS DBID#_Date)
APPENDIX C
HEALTH SURVEY

Health information

1) Do you wear glasses or contact lenses? Y / N If so, how much (e.g., for reading only?) If so, do you wear bifocals; any problems with table top work? Any difficulties related to vision?

2) Any problems related to eye movement?

3) Any problems with hearing? Y / N (give details)

4) Is it difficult for you to look at a computer screen because difficulty straightening your neck?

5) Do you have on/off fluctuations from your medication?

6) Do you experience involuntary movements (dyskinesias)? If so, describe.

7) Any problems with use of hands or arms, such as mobility or coordination? Y / N

8) Any problems walking? Y / N (e.g., up and down stairs). Need assistance or use any equipment for walking? Y / N How often?

9) Have you fallen in the last 6 months? If so, describe the situation. Do you fall frequently?

10) How much exercise do you get in a typical week (what activities and how often)?

11) Any medications that impact movement?: (For those with PD, have them bring their PD meds with them when they come to be tested so we can get names, dosage and frequency.

12) Have you ever had any of the following: (Give brief details, e.g., when the condition was diagnosed, severity, etc.)

stroke

head injury

neurological illnesses (other than PD)

Diabetes
Arthritis

Artificial joints or prostheses

Do you ever suffer from vertigo or dizziness?
Do you have any other problems that affect your movement or general health?
APPENDIX D

EXCERPTS FROM UNIFIED PARKINSON’S DISEASE RATING SCALE (UPDRS)
(Score 0-16)

Questions asked to the participants -

II Activities of Daily Living

Falling-Unrelated to Freezing
0-none
1-rare falls
2-occasional, less than one per day
3-average of once per day
4->1 per day

Freezing When Walking
0-normal
1-rare, may have start hesitation
2-occasional falls from freezing
3-frequent freezing, occasional falls
4-frequent falls from freezing

Walking
0-normal
1-mild difficulty, day drag legs or decrease arm swing
2-moderate difficultly requires no assist
3-severe disturbance requires assistance
4-cannot walk at all even with assist

Activities observed by the examiner -

III Motor Exam

Gait
0-normal
1-walks slowly, may shuffle with short steps, no festination or propulsion
2-walks with difficulty, little or no assistance, some festination, short steps or propulsion
3-severe disturbance, frequent assistance
4-cannot walk
REFERENCE LIST


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