Imposed faster and slower walking speeds influence gait stability differently in Parkinson fallers

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Running Head: Speed-related changes in gait stability

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ABSTRACT

Objective: This cross-sectional study sought to evaluate the effect of imposed faster and slower walking speeds on postural stability in people with Parkinson's disease (PD).

Design: Cross-sectional cohort study

Setting: General community

Participants: 84 PD patients (51 with a falls history; 33 without) and 82 age-matched controls were invited to participate via neurology clinics and pre-existing databases. Of those contacted, 99 did not respond (PD=36; controls=63) and 27 were not interested (PD=18; controls=9). Following screening, a further 10 patients were excluded; 5 had deep brain stimulation surgery and 5 could not accommodate to the treadmill. The remaining 30 patients completed all assessments and were sub-divided in PD fallers (n=10), PD Non-Fallers (n=10) and age-matched controls (n=10) based on falls history.

Protocol: Symptom severity, balance confidence and medical history were established prior to participants walking on a treadmill at 70%, 100% and 130% of their preferred speed.

Main Outcomes: Three-dimensional accelerometers assessed head and trunk accelerations and allowed calculation of harmonic ratios (HRs) and root mean square (RMS) accelerations to assess segment control and movement amplitude.

Results: Head and trunk control was lower for PD Fallers than PD Non-Fallers and Older Adults. Significant interactions indicated head and trunk control increased with speed for PD Non-Fallers and Older Adults, but did not improve at faster speeds for PD Fallers. Vertical head and trunk accelerations increased with walking speed for PD Non-Fallers and Older Adults, while the PD Fallers demonstrated greater anteroposterior RMS accelerations compared with both other groups.

Conclusion: The results suggest that improved gait dynamics do not necessarily represent improved walking stability and this must be respected when rehabilitating gait in PD patients.

Keywords: Gait; Segmental Control; Harmonic Ratio; Parkinson Disease; Falls

Parkinson's disease (PD) is a debilitating neurodegenerative condition that is characterised by motor symptoms that include resting tremor[1], slowness of movement[2], muscle rigidity[2], postural instability[2], and gait disturbances (e.g. freezing of gait (FOG))[1]. Unfortunately, symptoms of postural instability and gait disability are only partially responsive to current pharmacological interventions[3]. In fact, research shows that, even when optimally-medicated, people with PD demonstrate more asymmetric movement patterns[4, 5], walk more slowly[6-9], take shorter strides[6-9] and have less rhythmic acceleration profiles for the head[10] and trunk[11] compared with age-matched controls. The changes in segmental rhythmicity appear to be related, at least in part, to deficits in neuromuscular control[12] and seem to be more prominent in people with PD who prospectively report falls[13, 14]. Given this apparent relationship between postural instability, gait disability and falls in people with PD and the obvious ineptitude of current pharmacological therapies, clinicians and scientists have sought to identify suitable alternatives to manage these symptoms.

Treadmill-based gait retraining that incorporates auditory or visual cues has emerged as a common form of physical therapy and seeks to correct gait impairments in people with PD by increasing their stride length and, ultimately, their walking speed[15]. Importantly, the existing literature concerning gait retraining indicates that this form of therapy succeeds at this goal by helping patients to increase their stride length[16-21], walking speed[17-23] and walking distance[22]. Despite the established benefits of treadmill-based gait retraining for people with PD, the precise relationships between changes in walking speed and walking stability and/or falls risk are far less clear. For example, some prospective research has demonstrated that community-dwelling older adults who walk at slower (<0.6 m/s) or faster (≥1.3 m/s) speeds are at an increased risk of future falls[24]. Similar results were presented in a cross-sectional study involving healthy younger adults, which showed that slower and faster than preferred

63 speeds led to sub-optimal walking stability[25]. However, despite these findings, a series of
64 studies adopting non-linear analyses have suggested that local dynamic stability is
65 significantly improved at slower walking speeds for healthy younger adults[26, 27], older
66 adults[28] and patients with significant peripheral neuropathy[29, 30]. Given these conflicting
67 results, it remains unclear whether the slower walking speeds adopted by people with PD
68 serve to optimise their dynamic stability or contribute to their increased risk of falling. An
69 improved understanding of this relationship would help clinicians to better appreciate how
70 changes to a patient's walking speed might influence their stability and overall risk of falls.
71
72 During dynamic tasks, the maintenance of equilibrium relies upon one's capacity to control
73 the movements of the head and trunk, which represent almost 60% of the body's mass[31, 32].
74 From a functional perspective, the head is considered an important natural frame of reference,
75 as it houses the organs responsible for the visual and vestibular information used in postural
76 control and orientation[33-35]. The trunk is also believed to play a role in maintaining postural
77 stability during locomotion, as it serves to attenuate movement-related forces that project
78 upwards from the feet and threaten to destabilise the head[36, 37]. However, research reporting
79 larger[12-14] and less rhythmic[10, 11] head and trunk movements for people with PD provides
80 evidence to suggest that this population may have an impaired capacity to attenuate these
81 forces. Support for this notion was recently provided in a study that demonstrated people with
82 PD have an impaired capacity to attenuate accelerations from the pelvis and neck to the
83 head[38]. This impairment is likely related to the increased axial rigidity that is evident in
84 people with PD during standing[39] and walking[40], which is seemingly caused by differences in
85 the activation patterns of the paraspinal muscles in this population[12]. While it is widely
86 recognised that the routine use of anti-parkinsonian medication can significantly improve
87 some characteristics of gait[41, 42], it is equally well-documented that the symptoms of axial

rigidity that contribute to postural instability and falls in this population are not well managed with traditional therapies[40, 43]. Given this situation, there appears to be a clear need for research aimed at elucidating whether increasing walking speed in people with PD can be achieved without inadvertently influencing postural stability. As such, it was the purpose of this study to determine whether walking at speeds faster or slower than preferred reduces postural stability for people with PD. Given that slower and faster walking speeds have been linked with a greater risk of falls in older adults[24], it was hypothesised that walking at speeds other than one's preferred walking speed would reduce postural stability and that this relationship would be more pronounced for participants with a history of falling.

METHODS

Study Population

Between August and November 2014, 84 people clinically-diagnosed with idiopathic PD based on the Parkinson's United Kingdom Brain Bank Criteria[44] were invited to participate via community support groups and neurology clinics. Over the same period, 82 age-matched older adults (Controls) from the Brisbane metropolitan area were contacted via an existing database of individuals who had expressed interest in contributing to research of this nature. Of those contacted, 99 did not respond (PD=36; Controls=63) and 27 were not interested (PD=18; Controls=9). The remaining 30 people with PD and 10 controls were screened and excluded if they had; i) recently undergone surgery; ii) a recurrent history of musculoskeletal injury; iii) an inability to walk without assistance; iv) significant visual (Bailey-Lovie high contrast visual acuity >0.30 logMAR) or cognitive (Addenbrooke's Cognitive Examination score <82 out of 100[45]) impairment; or v) received deep brain stimulation. Following screening, 5 patients were excluded as they had received deep brain stimulation and 5 were excluded as they were unable to accommodate to the treadmill. The remaining participants

reported the number of falls that they had experienced in the past year and these data were used to separate PD Fallers (n=10) from PD Non-Fallers (n=10) and Older Adults (n=10). In all cases, the PD Fallers attributed their falls directly to complications associated with the symptoms and/or treatment of their condition (e.g. freezing of gait; festination, retropulsion; postural instability), rather than to situations that might be considered typical for an otherwise healthy individual. Falls were assessed retrospectively and defined as any unintentional coming to the ground or some lower level not as a result of a major intrinsic event or overwhelming hazard[46].

An a-priori power calculation performed using data presented previously[11] indicated that a sample size of 10 participants per group was sufficient to detect any significant changes in dynamic stability (diff = 0.05, SD = 0.04, Cohen's d = 1.25, Power = 80%, p = 0.05). The experimental protocol was approved by the Human Research Ethics Committee at the Australian Catholic University and, in accordance with the Declaration of Helsinki, all participants gave written informed consent prior to participating in this research.

INSERT TABLE 1 ABOUT HERE

Clinical Assessment

Prior to the gait assessment, details related to each participant's falls history, medical history and current medications were collected via a brief health questionnaire, while balance confidence was assessed using the 6-item Activities-specific Balance Confidence scale[47]. Additionally, an experienced movement disorders researcher completed clinical assessments for the PD participants to establish each patient's symptom severity and quality life. Specifically, symptom severity was assessed using the motor sub-scale of the Unified

Parkinson's Disease Rating Scale (UPDRS III)[48], the Hoehn and Yahr stage score[49] and the Schwab and England Activities of Daily Living (ADL) scale[50]. Additionally, FOG and quality of life were assessed using the Revised Freezing of Gait questionnaire[51] and the 8-item Parkinson's Disease Questionnaire[52], respectively. By calculating the sum of the scores for the items relating to rigidity on the UPDRS III, a global rigidity score was determined using previously-described methods[53]. All procedures were completed while the PD patients were receiving their usual anti-parkinsonian treatment, with 10 PD Fallers (100%) and 9 PD Non-Fallers (90%) being treated with levodopa and/or dopamine agonists (Table 1).

Apparatus

Two wireless 6g microelectromechanical systems (MEMS) tri-axial accelerometers (Noraxon Inc., USA) were positioned over the occipital protuberance of the skull and the spinous process of the 10^{th} thoracic vertebra to measure head and trunk accelerations during treadmill walking. The head accelerometer was attached to a firm-fitting headband, while the trunk accelerometer was firmly affixed to the skin using double-sided tape and Omnifix. Head and trunk accelerations were sampled at 1500 Hz and telemetered wirelessly to a Telemyo DTS receiver connected to a laptop running the MyoResearch XP software (v1.08, Noraxon Inc., USA). Prior to attaching the equipment, a series of static trials were completed while each of the accelerometers' axes were perpendicularly aligned with a horizontal surface to measure gravitational acceleration (1 gravitational unit or 1g) in the absence of movement[54].

The walking trials were completed on a Quasar motorised treadmill (HP Cosmos, DE) that had a moving surface size of 1.70 x 0.65 m (L x W) and an overhead safety frame fitted to facilitate anchoring of the participant safety harness. To ensure that participants were blind to their walking speed and to any changes that were made throughout the testing period, the user

terminal was rotated such that the participants were unable to see the electronic display. Prior to data collection, the validity of the treadmill's belt speed was assessed using a three-dimensional motion analysis system (T-Series cameras with Nexus 1.7; Vicon, UK) and was found to be accurate under both loaded and unloaded conditions at speeds ranging from 0.6 to 2.0 m/s (mean error = ±0.03 m/s).

Data Collection

To ensure that they could safely ambulate on the treadmill, each participant completed a familiarisation period while wearing their own comfortable walking shoes and a safety vest that was attached to the overhead safety frame. Each participant's preferred walking speed was then determined during three independent trials that were each separated by a rest break of no less than 60 seconds. During these trials, the treadmill's speed was systematically increased or decreased in 0.1 m/s increments based on the participant's instruction until they reported that they were walking at a comfortable speed. The average walking speed for these three trials was considered to be representative of the participant's preferred walking speed (100%) and was used to calculate the slower (70%) and faster (130%) walking conditions[26]. Using this information, participants completed a graded walking task that involved walking on the treadmill for 60 seconds at intensities that were equal to 70%, 100% and 130% of their preferred walking speed. To ensure that the acceleration/deceleration phase of each trial did not influence the reported outcomes, each 60-second data collection period did not commence until the treadmill had reached the target velocity and the participants reported having achieved a steady walking pattern. Given people with PD experience greater symptoms of gait impairment[13, 14] and fear of falling[55], the order of walking speeds (Intensity) was progressed from slowest to fastest. Furthermore, to limit the potential influence of fatigue, each walking trial was separated by a mandatory 1-minute rest break.

Data Analysis

Following data collection, the raw three-dimensional head and trunk accelerations were transformed to a horizontal-vertical orthogonal coordinate system using an extrapolation of simple trigonometry[36]. In short, transformation of the accelerations was required to correct for tilt in the AP and ML directions, such that the accelerometer's vertical axis was realigned with the gravity vector (i.e. global vertical axis)[56]. The transformation algorithm achieved this by assuming that the head and trunk accelerometers were rotated (i.e. r(theta1, theta2)) and that this angle was constant throughout the trial. This assumption was guided by previous research, which reported that the orientation of the upper body changes minimally during gait[57, 58] and, hence would only influence gait-related accelerations to a small degree[36, 59]. During pilot testing, the performance of the transformation process was assessed by comparing the transformed accelerations from the Noraxon system with data simultaneously collected using XSens inertial measurement units (IMUs). Data from the IMUs were rotated using the device's internal gyroscope and comparison of the anteroposterior (AP), mediolateral (ML) and vertical (VT) acceleration profiles from the two systems returned correlation coefficients of 0.8 or greater for all three axes. Following transformation, the timing of individual foot contacts was identified via the recurring peaks in the vertical trunk acceleration profile[11, 60, 61] and used to crop each trial to a length that included 10 left and 10 right gait cycles (i.e. 20 gait cycles total). The cropped data were then low-pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 30 Hz[59, 62].

To examine changes in the rhythmicity of AP, ML and VT head and trunk accelerations at the different walking speeds, the harmonic ratio (HR) was calculated by firstly dividing the continuous data series into individual gait cycles (i.e. 20 per trial). Data for each gait cycle

were then converted to the frequency domain using the Fast Fourier Transformation, which allowed the harmonics of the signal's fundamental frequency (i.e. stride frequency[63]) to be identified[64]. As each gait cycle is comprised of two steps, the AP and VT acceleration profiles of a healthy individual are typically characterised by two comparable peaks[25]. As these peaks repeat in multiples of two, the frequency spectra of AP and VT accelerations are dominated by the even harmonics (i.e. 2, 4), which represent the in-phase component of these signals. In contrast, ML accelerations are characterised by two opposing peaks; 1 corresponding with a weight shift to the left leg and 1 corresponding with a weight shift to the right leg. This unique characteristic of the ML acceleration profile means that the odd harmonics (i.e. 1, 3) dominate this component and, hence represent the in-phase component of this signal. Using the first 20 harmonics for each gait cycle (i.e. 10 in-phase; 10 out-of-phase), the AP, ML and VT harmonic ratios were calculated for the head and trunk by dividing the sum of the in-phase harmonics by the sum of the out-of-phase harmonics[64]. Given this calculation, larger HRs represent a greater proportion of in-phase accelerations relative to out-of-phase accelerations, which is indicative of greater movement rhythmicity and poorer segmental control[64, 65].

To provide insight into the amplitude of head and trunk accelerations during the walking task, the root mean square (RMS) amplitude of the time-series data for the AP, ML and VT accelerations was also calculated[66]. In addition to the three-dimensional HRs and RMS accelerations, the timings of each individual foot contact were used to calculate a number of spatiotemporal characteristics. Specifically, cadence (steps/min) was assessed by determining the number of steps taken by each participant during the 60-second trial, while stride timing variability (ms) was derived by calculating the standard deviation of the time taken by the participant to complete each of the 20 gait cycles (i.e. stride time)[67, 68]. Lastly, given that

238 walking speed is a composite measure representing stride length (i.e. distance) divided by
239 stride time, stride length was calculated by multiplying walking speed (m/s) by stride time.
240 These outcome measures were selected as they have been extensively used to assess walking
241 in people with PD[11, 65, 69] and have been previously shown to discriminate retrospective fallers
242 from non-fallers in this population[10]. All processing of the raw head and trunk accelerations
243 was performed using a custom Matlab program (R2015b, The MathWorks, USA).
244
245 *Statistical Analysis*
246 A one-way analysis of variance (ANOVA) was used to compare the groups for differences in
247 demographics, falls history, fear of falling, quality of life and symptom severity. When a
248 significant main effect was identified, the Tukey's Honestly Significant Difference (HSD)
249 post-hoc test was used to determine where the statistically significant differences existed.
250 When the assumptions of ANOVA were violated, the non-parametric Kruskal-Wallis Test
251 was used to compare the groups, while the degree of association between categorical
252 variables was assessed using the chi-square (χ^2) test.
253
254 To determine mean differences between the PD Fallers, PD Non-Fallers and Older Adults for
255 the accelerometer-based measures of gait rhythmicity and segmental motion, linear mixed
256 model (LMM) analyses with one repeated (Intensity, 3 levels) and one fixed (group, 3 levels)
257 factor were used. As gait speed and stride time variability both influence segmental
258 accelerations[10], both were entered as covariates for the analysis of HRs and RMS
259 accelerations. Furthermore, to determine whether differences in disease duration, symptom
260 severity and/or medication use accounted for any differences in HRs or RMS accelerations, a
261 series of sub-analyses were conducted for the PD Fallers and Non-Fallers, with these clinical
262 scores also entered as covariates. Where significant main effects or interactions were

identified, Tukey's Least Significant Difference post-hoc tests were used to conduct pairwise comparisons between the groups. All statistical procedures were conducted using SPSS v.22 and the level of significance was set at $p < 0.05$.

RESULTS

Demographics and Clinical Assessments

PD Fallers, PD Non-Fallers and Older Adults did not differ significantly with respect to age, gender distribution, height or mass, but PD Fallers had increased rigidity, poorer quality of life and greater symptom severity than patients in the PD Non-Faller group. PD Fallers also tended to report poorer balance confidence than the other participants (p=0.08) and to be taking larger daily doses of levodopa than PD Non-Fallers (p=0.06); however, these trends did not achieve statistical significance. Similarly, the PD Faller and Non-Faller groups were not different with respect to disease duration or the proportion of patients prescribed dopamine agonists, catechol-o-methyl transferase (COMT) inhibitors, monoamine oxidase inhibitors (MAOIs) and/or benzodiazepines (Table 1).

Walking Assessment

PD Fallers walked significantly slower and took significantly shorter strides, but did not differ from the PD Non-Fallers or Older Adults with respect to cadence and stride time variability. Significant main effects for Intensity indicated that stride length and cadence systematically increased from the 70% to 100% to 130% conditions, while stride time variability systematically decreased as walking speed increased (Figure 1). With respect to head and trunk rhythmicity, significant main effects for Intensity indicated that harmonic ratios were significantly reduced (poorer) during the 70% trials compared with the 100% and 130% conditions. Furthermore, ML head and trunk rhythmicity was significantly improved

when participants walked at the 130% walking speed compared with their preferred walking speed (100%). Significant main effects for Group were reported for the ML and VT axes of head and the AP, ML and VT axes of the trunk. Post-hoc analyses revealed that PD Non-Fallers recorded significantly lower head (ML, VT) and trunk (AP, ML, VT) rhythmicity than the Older Adults (Figure 2). Similarly, PD Fallers had significantly lower head (ML, VT) and trunk (AP, ML, VT) harmonic ratios than PD Non-Fallers and Older Adults and sub-analysis of the PD Fallers and Non-Fallers suggested that these findings were not attributable to differences in disease duration, symptom severity and/or daily levodopa equivalent dose.

INSERT FIGURE 1 ABOUT HERE

In addition to these main effects, significant Group*Intensity interactions were reported for AP and VT harmonic ratios for the head and AP, ML and VT harmonic ratios for the trunk. Further examination of these interactions showed that the speed-related changes in head and trunk rhythmicity for PD Fallers were significantly different to those observed for PD Non-Fallers and Older Adults. Specifically, head AP and VT harmonic ratios for the PD Non-Fallers and Older Adults significantly increased as walking speed increased. An improvement in AP and VT head rhythmicity between the 70% and 100% walking speeds was also evident for the PD Fallers, but AP head rhythmicity was unchanged between the 100% and 130% conditions, while VT head rhythmicity declined at the faster speed. Similarly, AP, ML and VT trunk harmonic ratios remained unchanged or improved as walking speed increased for the PD Non-Fallers and Older Adults, while both AP and VT trunk harmonic ratios were significant reduced for the PD Fallers during the 130% walking trial, compared with the 100% condition (Table 2).

INSERT FIGURES 2 AND 3 ABOUT HERE

The RMS accelerations demonstrated that PD Fallers had significantly greater AP head accelerations than PD Non-Fallers and Older Adults, but were not dissimilar with respect to any other component of head or trunk acceleration. The sub-analyses conducted for the two PD groups indicated that the larger RMS head accelerations (AP) recorded for the PD Fallers were largely explained by differences in disease duration, symptom severity and/or levodopa daily equivalent doses. Significant main effects for Intensity suggested that AP and ML head accelerations and ML trunk accelerations were significantly greater during the 70% condition relative to the 100% and 130% walking trials (Figure 3). In contrast, VT RMS accelerations for the head and trunk were significantly greater during the 130% condition compared with the 70% and 100% conditions. Significant Group*Intensity interactions for VT head and trunk accelerations indicated that VT acceleration amplitudes were consistent for the PD Fallers across the walking speeds, but were significantly increased at the fastest speed for PD Non-Fallers and Older Adults. Furthermore, the significant Group*Intensity interaction for AP RMS accelerations indicated that PD fallers had significantly greater head accelerations at the slowest walking speed compared with the 100% and 130% conditions (Table 2).

INSERT TABLE 2 ABOUT HERE.

DISCUSSION

The results of this cross-sectional study only partially supported our hypothesis that walking at speeds slower and faster than preferred would correspond with poorer head and trunk rhythmicities. As hypothesised, poorer stability was observed for all participant groups at walking speeds that were slower than preferred, but as walking speed increased, head and

trunk rhythmicity generally improved as well. These findings are in contrast to previous research involving healthy younger adults, which showed that pelvic and, to a lesser extent, head rhythmicities were optimal when participants walked at their preferred speed, but declined at faster and slower speeds[25]. Similarly, the results of a longitudinal study indicated that the risk of falling was significantly greater in older adults who walked slower (<0.6 m/s) or faster (≥1.3 m/s)[24], suggesting that stability may be optimised at specific movement speeds. The disparity between the results of the current study and those presented in this earlier research may be explained by differences in the coordination and variability of segmental motion during treadmill and overground walking. For example, research shows that individuals exhibit reduced variability in their stride-to-stride gait patterns and joint kinematics during treadmill walking compared with overground gait[70, 71]. Such differences are argued to be due to the relatively fewer task constraints imposed by overground walking, which ultimately gives individuals a greater number of performance options that are equally appropriate for achieving the desired outcome[71, 72]. Interestingly, the results of this study also showed that stride timing variability systematically decreased from the slowest to the fastest walking speed, while separate research examining overground walking in younger adults reported increased stride time variability at speeds slower and faster than preferred[73]. Considering that the harmonic ratio provides a measure of the in-phase to out-of-phase segmental accelerations, it is possible that the improved stability demonstrated by the participants at the faster speed was reflective of the less variable walking patterns recorded for these individuals during this condition.

Despite the results tending to suggest that increased walking speeds lead to improved head and trunk stability in older adults and people with PD, the post-hoc analyses indicated that head and trunk accelerations either remained unchanged or decreased at the faster walking

speed for PD Fallers. Considering this finding with the overall deficits in head and trunk control and the increased AP head accelerations that were evident for the PD Fallers, it seems that these individuals may have a reduced capacity to control these larger segments, which would directly impact their postural stability. These results are in agreement with previous research showing that people with PD have significantly greater AP and ML head accelerations than healthy younger and older adults, which are likely to influence their capacity to recover from a perturbation[74]. Collectively, these finding suggest that while some patients (e.g. PD Non-Fallers) may have the capacity to adapt to the changing demands of a task, patients who have a history of falls and typically walk at slower preferred speeds may not. A possible explanation as to why the PD Fallers demonstrated different patterns of head and trunk control at the faster walking speed might be found in the higher global rigidity scores reported for these patients at baseline. According to previous research, the rigidity of the axial system (e.g. trunk, pelvis, neck) significantly increases at faster walking speeds for people with PD[40]. Given the axial skeleton essentially serves as a biological shock absorber to minimise the effects of movement-related forces on the visual and vestibular systems[33-36], an increase in the rigidity of this system would likely influence its capacity to perform this role. As such, the higher prevalence of rigidity evident in the PD fallers may have made these individuals more susceptible to speed-related changes in axial rigidity and account for a plateau or decline in head and trunk stability during the faster walking trials. Nevertheless, the significant decline in some aspects of dynamic stability at the faster walking speed suggests that the assessment of gait during fast-paced walking may be more suitable for identifying people with PD who are at an increased risk of falling[75]. Furthermore, it seems that if therapists are not monitoring changes in postural stability during gait retraining programs, it is possible that improvements in gait dynamics may come at the cost of an increased falls risk for some patients.

Study Limitations

There are a number of methodological factors that should be considered when reviewing our results, as they have the potential to limit our capacity to directly compare our findings with previous research. First, we elected to conduct our assessments on a motorised treadmill to strictly control changes in walking speed and to ensure the safety of the participants. However, previous research has shown that treadmill walking is not a perfect analogue for overground walking, as it generally returns different values for some spatiotemporal characteristics[76, 77], gait variability[71, 77] and joint kinetics[76, 78]. Second, the use of tri-axial accelerometers to assess head and trunk rhythmicity during the walking trials limited our capacity to objectively evaluate other factors that may potentially have influenced gait stability (e.g. arm swing, base of support). Although there is a growing body of evidence to suggest that the size of one's base of support is not significantly influenced by their walking speed[79-81], research has consistently reported a relationship between arm swing and walking speed in healthy younger[82] and older adults[83]. While it remains unclear whether arm swing directly influences walking stability[84] or whether it serves to recover a stable walking pattern following a perturbation[85], it is important to acknowledge that differences in arm swing between the groups may have potentially impacted the reported outcomes. Future research should seek to determine the specific role(s) of arm swing in stabilising the gait patterns of people with PD and evaluate whether imposed faster and slower walking speeds influence walking stability in a similar way during overground walking in this population. Despite the shortcomings of this methodological approach, our findings are likely to be of significant clinical relevance, as physical therapists are often restricted to using treadmills for gait retraining due to space limitations and the need to minimise patient risk in the clinical setting. Furthermore, if we consider that those patients who are most likely to be referred to physical

therapists for gait retraining are those who present with significant gait disability that limits their walking speed, then these findings have obvious implications for current practice.

CONCLUSIONS

While systematic evidence indicates that gait retraining can improve stride length[16-21], walking speed[17-23] and walking distance[22] in people with PD, the results of this study suggest that these changes may lead to an increased risk of future falls for some patients if postural stability is not targeted. As such, we recommend that gait retraining should not be implemented as a stand-alone therapy for high-risk PD patients, but rather should be coupled with other physical therapy that seeks to address any underlying balance impairments that may be present for an individual.

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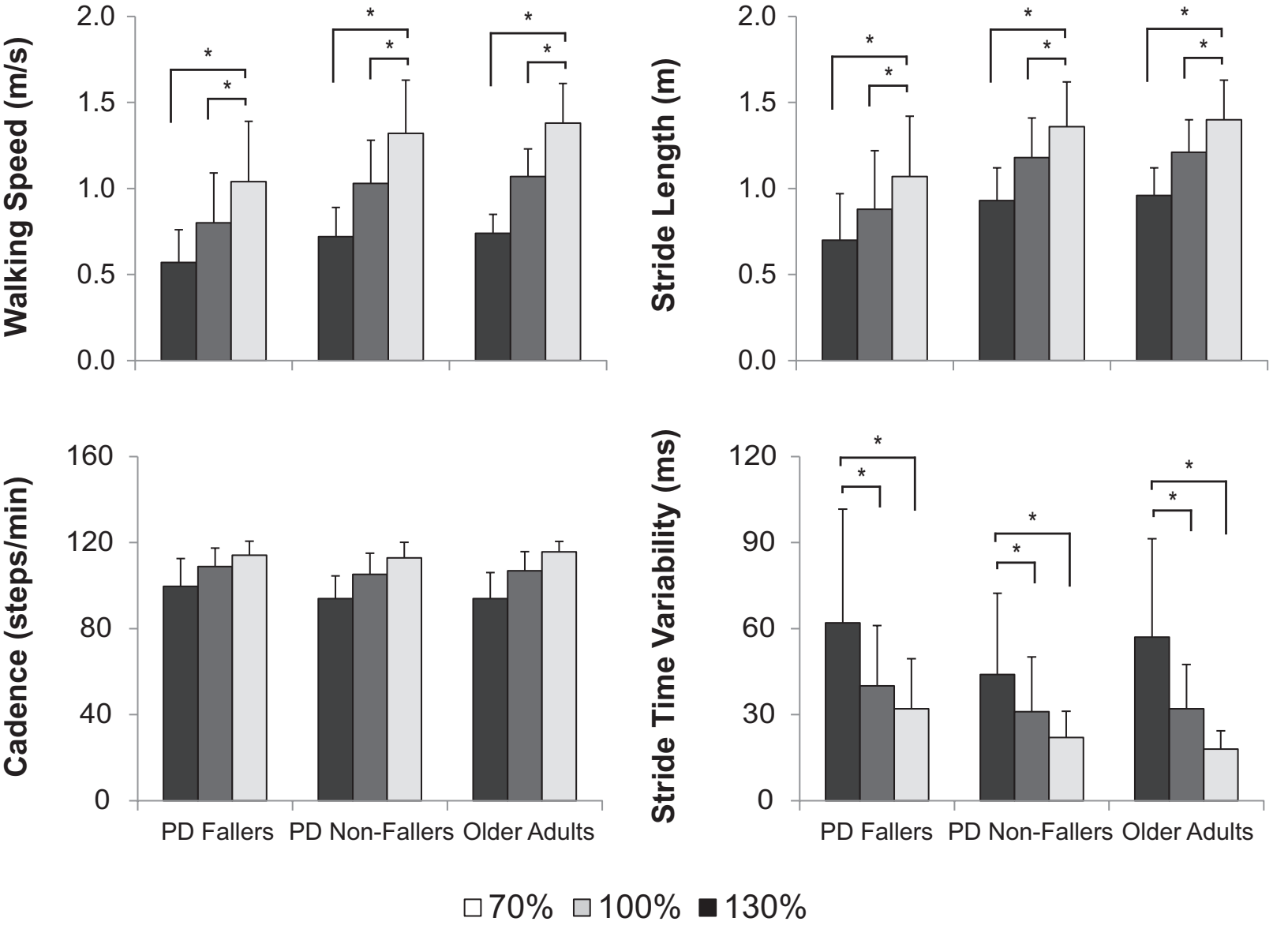
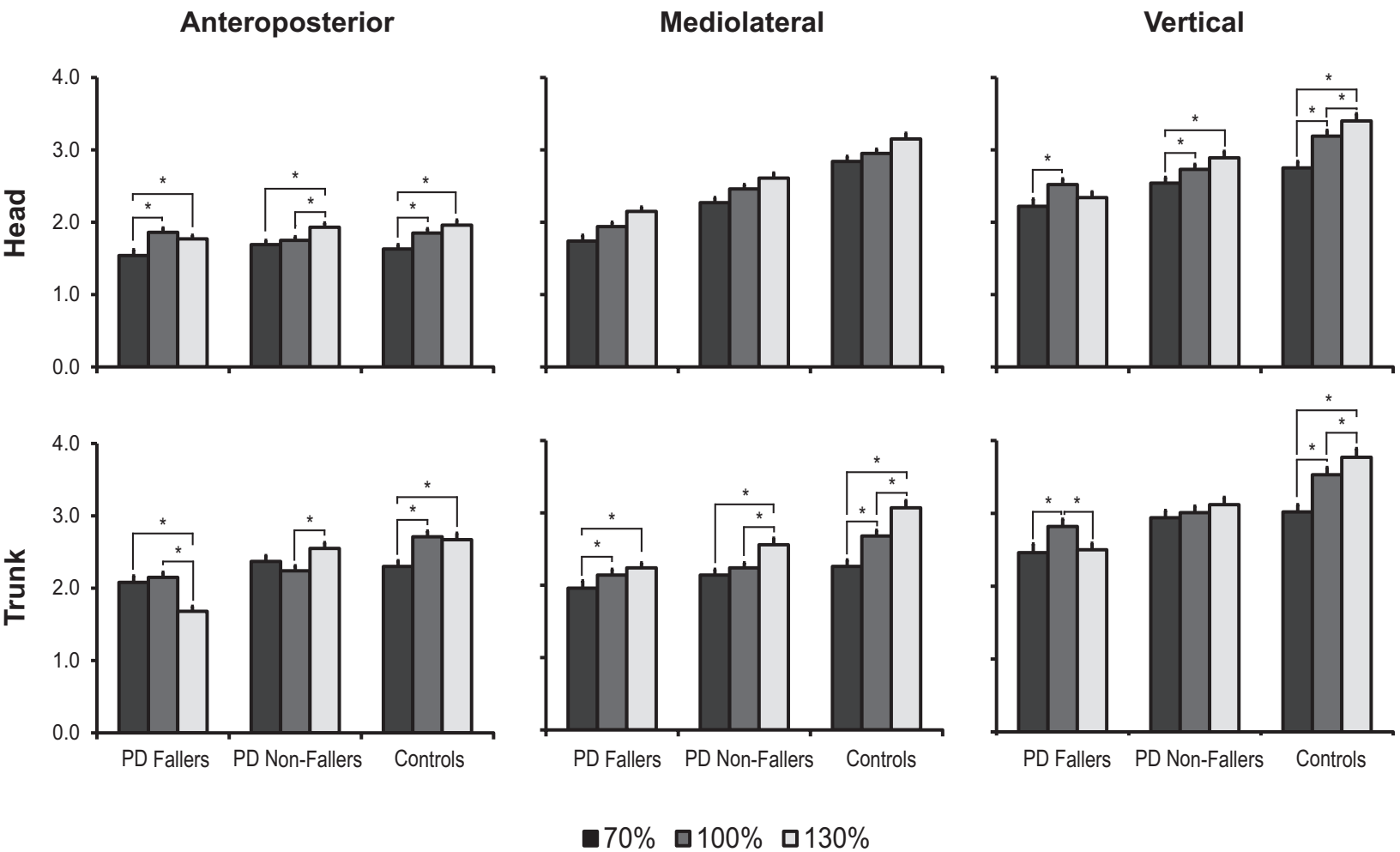
FIGURE LEGENDS

Fig. 1: Mean (+1 SD) walking speeds, stride lengths, cadences and stride time variability for the PD fallers, PD Non-Fallers and age-matched Older Adults while walking on the treadmill at 70%, 100% and 130% of their preferred walking speed.

Fig. 2: Estimated Marginal Means (EMM) and standard errors (SE) for the head and trunk harmonic ratios (adjusted for walking speed and stride time variability) for the PD fallers, PD Non-Fallers and Older Adults while walking on the treadmill at 70%, 100% and 130% of their preferred walking speed. **Note:** Larger harmonic ratios depict a greater proportion of in-phase relative to out-of-phase accelerations and, hence represent more stable gait patterns.

Fig. 3: Estimated Marginal Means (EMM) and standard errors (SE) for head and trunk RMS accelerations (adjusted for walking speed and stride time variability) for the PD fallers, PD Non-Fallers and age-matched Older Adults while walking on the treadmill at 70%, 100% and 130% of their preferred walking speed.


Walking Speed (m/s)
2.0
1.5
1.0
0.5
0.0
Stride Length (m)
2.0
1.5
1.0
0.5
0.0
Cadence (steps/min)
160
120
80
40
0
Stride Time Variability (ms)
120
90
60
30
0
PD Fallers
PD Non-Fallers
Older Adults
PD Fallers
PD Non-Fallers
Older Adults
□70% ■100% ■130%


Anteroposterior
Mediolateral
Vertical
Head
Trunk
4.0
3.0
2.0
1.0
0.0
PD Fallers
PD Non-Fallers
Controls
70%
100%
130%

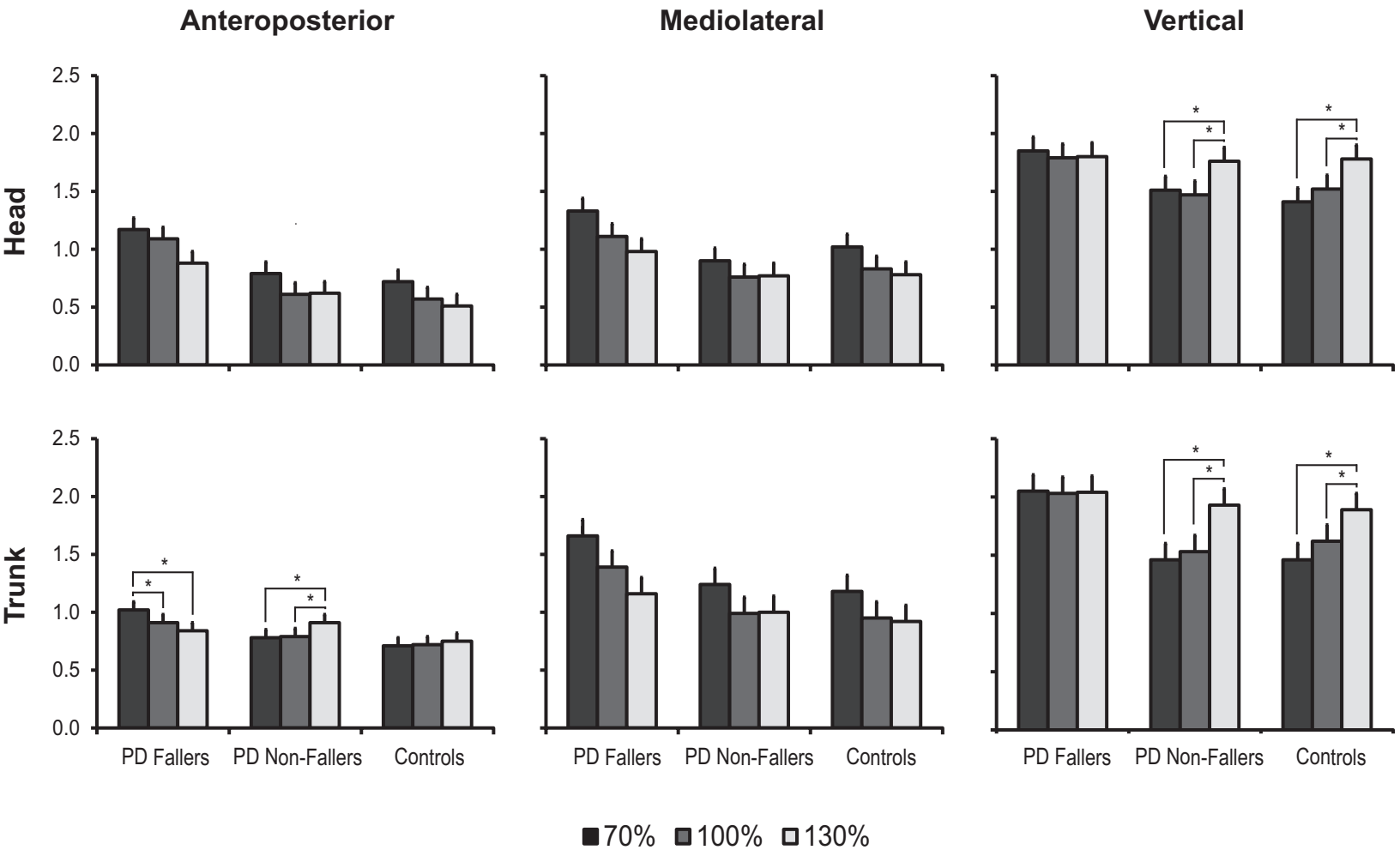

Anteroposterior
Mediolateral
Vertical
Head
Trunk
2.5
2.0
1.5
1.0
0.5
0.0
*
PD Fallers
PD Non-Fallers
Controls
70%
100%
130%

Table 1: Demographic data and disease-specific scores for the participants with PD Fallers, PD Non-Fallers and the age-matched Older Adults. Data represent the mean (standard error of the mean (SEM)) values or absolute numbers and percentages. Test 1 = one-way ANOVA; Test 2 = Kruskal-Wallis Test; Test 3 = χ^2 test.

	PD Fallers	PD Non-Fallers	Older Adults	Test	Sig
Demographics					
Age (Years)	69.3 (2.2)	66.5 (2.5)	68.6 (2.8)	1	ns
Gender (Male)	6 (60.0%)	6 (60.0%)	6 (60.0%)	3	ns
Height (cm)	165.7 (3.5)	168.5 (3.8)	168.7 (2.7)	1	ns
Mass (kg)	65.9 (6.2)	67.9 (3.8)	65.9 (3.1)	1	ns
Falls History and Fear of Falls					
Activities-Specific Balance Confidence	59.3 (8.9)	78.7 (4.7)	82.3 (7.0)	2	ns
Previous Falls (12 months)	9.5 (4.8)	0.0 (0.0)	0.4 (0.2)	2	a, b
Quality of Life					
8-item Parkinson's Disease Questionnaire	28.8 (4.9)	14.4 (2.1)		2	a
Neurological Exam					
Disease Duration (years)	7.0 (1.7)	4.6 (0.6)		2	ns
Levodopa (mg/day)	810.8 (147.8)	451.6 (102.9)		1	ns
Dopamine Agonists	2 (20.0%)	2 (20.0%)		3	ns
Catechol-O-Methyl Transferase Inhibitors	2 (20.0%)	2 (20.0%)		3	ns
Monoamine Oxidase Inhibitors	6 (60.0%)	3 (30.0%)		3	ns
Benzodiazepine	0 (0.0%)	0 (0.0%)		3	ns
No Medication	0 (0.0%)	1 (10.0%)		3	ns
UPDRS III	22.6 (1.9)	13.1 (2.1)		1	a
Hoehn & Yahr Stage Score	2.2 (0.2)	1.4 (0.2)		2	a
Schwab & England ADL Scale	77.0 (2.4)	89.5 (2.0)		1	a
Revised Freezing of Gait Score	10.8 (3.2)	2.1 (2.1)		2	a

ns: No significant differences between groups; **a:** PD Fallers significantly different to PD Non-Fallers; **b:** PD Fallers significantly different to Older Adults; **c:** PD Non-Fallers significantly different to Older Adults

Table 2: Estimated Marginal Means (EMM) and standard errors (SE) for the head and trunk harmonic ratios and RMS accelerations (adjusted for walking speed and stride time variability) for the PD fallers, PD Non-Fallers and Older Adults while walking on the treadmill at 70%, 100% and 130% of their preferred walking speed. **Note:** Larger harmonic ratios depict a greater proportion of in-phase relative to out-of-phase accelerations and, hence represent more rhythmic gait patterns.

		70% Preferred Walking Speed			**100% Preferred Walking Speed**			**130% Preferred Walking Speed**			
		PD Fallers EMM (SE)	**PD Non-Fallers** EMM (SE)	**Older Adults** EMM (SE)	**PD Fallers** EMM (SE)	**PD Non-Fallers** EMM (SE)	**Older Adults** EMM (SE)	**PD Fallers** EMM (SE)	**PD Non-Fallers** EMM (SE)	**Older Adults** EMM (SE)	**Sig**
Harmonic Ratios											
Head	Anteroposterior	1.54 (0.08)	1.69 (0.06)	1.63 (0.06)	1.86 (0.06)	1.75 (0.05)	1.85 (0.06)	1.77 (0.05)	1.93 (0.06)	1.96 (0.07)	**¥, §, †**
	Mediolateral	1.74 (0.08)	2.27 (0.07)	2.84 (0.07)	1.94 (0.06)	2.46 (0.06)	2.95 (0.06)	2.15 (0.06)	2.61 (0.07)	3.15 (0.08)	**a, b, c, ¥, Ŧ, §**
	Vertical	2.22 (0.10)	2.54 (0.08)	2.75 (0.09)	2.52 (0.08)	2.73 (0.07)	3.19 (0.08)	2.34 (0.08)	2.89 (0.09)	3.40 (0.10)	**a, b, c, ¥, §, †**
Trunk	Anteroposterior	2.08 (0.09)	2.37 (0.08)	2.30 (0.08)	2.15 (0.07)	2.24 (0.07)	2.71 (0.08)	1.68 (0.07)	2.55 (0.08)	2.67 (0.09)	**a, b, c, ¥, †**
	Mediolateral	1.96 (0.10)	2.14 (0.08)	2.26 (0.09)	2.14 (0.08)	2.24 (0.07)	2.68 (0.08)	2.24 (0.07)	2.56 (0.09)	3.07 (0.10)	**a, b, c, ¥, Ŧ, §, †**
	Vertical	2.46 (0.12)	2.94 (0.10)	3.02 (0.10)	2.82 (0.10)	3.01 (0.09)	3.53 (0.10)	2.50 (0.09)	3.12 (0.10)	3.77 (0.12)	**a, b, c, ¥, §, †**
RMS Acceleration (m/s^2)											
Head	Anteroposterior	1.17 (0.10)	0.79 (0.10)	0.72 (0.10)	1.09 (0.10)	0.61 (0.10)	0.57 (0.10)	0.88 (0.10)	0.62 (0.10)	0.51 (0.10)	**a, b, ¥, §**
	Mediolateral	1.33 (0.11)	0.90 (0.11)	1.02 (0.11)	1.11 (0.11)	0.76 (0.11)	0.83 (0.11)	0.98 (0.11)	0.77 (0.11)	0.78 (0.11)	**¥, §**
	Vertical	1.85 (0.12)	1.51 (0.12)	1.41 (0.12)	1.79 (0.12)	1.47 (0.12)	1.52 (0.12)	1.80 (0.12)	1.76 (0.12)	1.78 (0.12)	**Ŧ, §, †**
Trunk	Anteroposterior	1.02 (0.07)	0.78 (0.07)	0.71 (0.07)	0.91 (0.07)	0.79 (0.07)	0.72 (0.07)	0.84 (0.07)	0.91 (0.07)	0.75 (0.07)	**†**
	Mediolateral	1.66 (0.14)	1.24 (0.14)	1.18 (0.14)	1.39 (0.14)	0.99 (0.14)	0.95 (0.14)	1.16 (0.14)	1.00 (0.14)	0.92 (0.14)	**¥, §**
	Vertical	2.05 (0.14)	1.46 (0.14)	1.46 (0.14)	2.03 (0.14)	1.53 (0.14)	1.62 (0.14)	2.04 (0.14)	1.93 (0.14)	1.89 (0.14)	**Ŧ, §, †**

ns: No significant differences between groups; **a:** PD Fallers significantly different to PD Non-Fallers; **b:** PD Fallers significantly different to Older Adults; **c:** PD Non-Fallers significantly different to Older Adults; **¥** 70% significantly different to 100%; **Ŧ** 100% significantly different to 130%; **§** 70% significantly different to 130%; † significant Group*Speed interaction.

Highlights

- Parkinson’s patients with a falls history had poorer rhythmicity at all gait speeds
- Improvements in walking speed do not necessarily imply improved postural stability
- Combining gait retraining with other therapies may benefit high-risk Parkinson’s patients