EFFECTS OF ACUTE MUSCLE FATIGUE ON POSTURAL CONTROL IN HEALTHY ADULTS AND PERSONS WITH PARKINSON'S DISEASE

by

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ABSTRACT

Falls are one of the most disabling features of aging and are increasingly common in persons with balance-impairments such as Parkinson's disease (PD). Falls can cause physical injuries such as fractures and/or head injuries leading to functional incapacity, increased risk of nursing home admission, and higher mortality rate. Acute muscle fatigue has been shown to exacerbate fall-correlated end-points such as postural control in healthy young and elderly individuals. The majority of studies investigating these effects, however, have focused on static stance postural control, or tasks that fail to incorporate more functional movements such as those requiring components of anticipatory and reactive postural control. The purpose of this study was to document the effects of acute lower extremity muscle fatigue on anticipatory and reactive postural control in persons with PD and to compare those results to the impact of fatigue on healthy elderly and young populations. Additionally, this investigation sought to gain insight into the chronology for postural control recovery following acute muscle fatigue. This dissertation has yielded a background on acute muscle fatigue, followed by a systematic review of the evidence on the effects of muscle fatigue on anticipatory and reactive postural control in healthy older individuals. The focus of the paper then shifts to components of an experimentally designed cohort study examining the effects of acute muscle fatigue on a centrally initiated movement task and a peripherally directed lean-induced fall in persons with PD and neurologically healthy adults. Results indicated that both anticipatory and reactive postural control are altered following acute muscle fatiguing exercise in neurologically healthy young and older adults. Amelioration of fatigue effects is extended beyond 30 minutes for most measures. Recovery occurs more readily for reactive postural control than anticipatory postural control. No statistically significant results were found from fatigue effects on postural control in the full cohort of persons with PD. However, a supplementary analysis revealed that postural control is altered in persons with PD who exercised beyond a minimal threshold of energy expenditure. More research is needed with larger sample sizes and improved construct validity for muscle fatigue in this cohort. The results of this study should serve to heighten awareness regarding the potential negative effects of acute muscle fatigue, including the possibility of falls in clinical and community based exercise settings for older adults at risk for falls.

This book is dedicated to someone who has played a “behind-the-scenes” role throughout the entirety of this project. To my wife, who has supported and encouraged me for that last 10 years; 9 ½ of which were lived as “students.” My degree of happiness has not suddenly changed now that I’ve reached the professional ranks, because you have been with me all the time. I am fully aware, however, that no one could be happier than you to have this project done! I love you Brigitte.

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CHAPTER 1

INTRODUCTION

Acute Muscle Fatigue

Fatigue and exhaustion have been an area of interest for researchers for more than a century. When applied to muscular exercise, fatigue was seminally referred to as "a failure to maintain the required or expected force"[1] or a failure to "continue working at a given exercise intensity."[2] These early perspectives of muscle fatigue implied a point of exhaustion and suggested that fatigue in working muscles would begin only at a point of task failure. On the contrary, the maximal force generating capacity of muscles begins to decline at the onset of exercise so that fatigue really begins once exercise commences and progressively develops before the muscles fail to perform the task.[3] Consequently, a more appropriate definition of fatigue has evolved into "any exercise-induced reduction in the ability to exert muscle force or power, regardless of whether or not the task can be sustained."[4] A reduction in muscle force has a number of consequences on functional tasks, including the control of posture. Numerous central nervous system changes occur following acute muscle fatigue in healthy young populations, producing clumsiness and diminished precision of motor control (see Table 1). There is also a preponderance of metabolic factors that are known to be altered following localized muscle fatigue, including increases in lactic acid and decreases in pH, which have specifically been shown to reduce postural control in healthy young individuals.[5, 6]

Postural Control

Postural control is defined as the regulation of the body's position in space for the dual purposes of equilibrium and orientation.[7] Postural equilibrium is described as the ability to balance all the forces acting on the body such that it maintains a desired position or moves in a controlled way.[8] Postural orientation involves interpreting various forms of sensory information in order to establish a representation of the body relative to its environment as well as the appropriate positioning of body segments relative to each other and the environment.[8] A narrower definition of postural control, and one that this thesis adopts, is the ability to maintain the projected center of mass (COM) within the actual or anticipated limits of the base of support.[9]

Control of posture is not a steady state but a dynamic interaction between an individual's musculoskeletal and neural control systems and their environment. Musculoskeletal components include such things as soft tissue properties, joint range of motion, spinal flexibility, and biomechanical relationships of linked body segments. Neural components encompass motor processes, sensory mechanisms, and sensory integrative processes. These components form the basis for anticipatory and reactive mechanisms of postural control.[7]

Anticipatory aspects of postural control are *internally* induced processes that prepare sensory and motor systems for postural demands. Anticipatory postural adjustments occur in an "expectant" or feedforward manner prior to action of the prime mover. Examples of anticipatory postural control actions include the initiation of gait[10] and stabilizing the trunk before reaching overhead.[11] Reactive postural control is defined by modifying sensory and motor systems in response to changing tasks and *externally* induced environmental demands. Contrary to anticipatory actions, reactive postural control mechanisms occur in a "compensatory" or feedback manner after the onset of a perturbation. Examples of reactive postural control include responses to slipping or tripping situations induced by sliding force plates,[12] treadmill perturbation,[13] and tether-release models.[14]

Gaps in the Literature

Anticipatory and reactive postural control are utilized daily in dynamic conditions like walking, lifting, and carrying objects. The elucidation of fatigue's influence on anticipatory and reactive postural control is important for safe functional mobility because it is in these contexts where the majority of falls occur in older adults.[15, 16] Falls are one of the most disabling features of aging[17] and are increasingly common in persons with balance impairments such as Parkinson's disease (PD).[18] Falls in these populations can cause physical injuries such as fractures and/or head injuries leading to functional incapacity, increased risk of nursing home admission, and higher mortality rate.[17, 19, 20] Despite the magnitude of published reports examining the epidemiology of falls in these populations, relatively few have examined the contribution of acute muscle fatigue on falls and fall-related endpoints such as postural control. A recent report suggested that there is a significant negative effect of lower extremity and trunk muscle fatigue on balance and functional tasks in older people.[21] When coupled with the known alterations that occur in postural control in healthy young populations, it stands to wonder why no studies have examined the effects of acute muscle fatigue on postural control in persons with known postural instability such as PD. In addition, few studies have examined the effects of fatigue on both anticipatory and reactive postural control tasks in persons with PD and healthy older individuals.

The central premise of this study was that acute bouts of muscle fatigue alter postural control, and when individuals with inherent balance impairment are exposed to muscle fatigue, these individuals may be at an even greater risk of falls and fall-related injury. The primary purpose of this study was to document the acute effects of lower extremity muscle fatigue on anticipatory and reactive postural control in persons with PD and to compare these results to the impact of fatigue on neurologically healthy older and young populations. This dissertation examines these issues in the following aims: first, a systematic review of the effect of acute muscle fatigue on anticipatory and reactive postural control in neurologically healthy older individuals is conducted (see Chapter 2). Second, an investigation into the effect of acute muscle fatigue on anticipatory postural control in persons with PD is examined, and comparisons are made to those results in healthy controls (see Chapter 3). Third, an examination is made into the effect of fatigue on reactive postural control in PD and neurologically healthy controls (see Chapter 4). Each of these chapters contains data regarding the chronology for postural control recovery following acute muscle fatigue in these populations. Insight into each of these aims is clinically relevant in terms of examination of postural control in fatigued and nonfatigued states, acknowledgement of the potential for iatrogenic increases in fall risk as a result of treatment, and for improvements in muscle endurance as a potential target for rehabilitation interventions.

Table 1. Sensory and Motor Alterations Affecting Postural Control Due to Muscle Fatigue

Description	*Reference*
Reduced conduction velocity of afferent inputs	Broman et al., 1985[22]
Reduced conduction velocity of motor output	Broman et al., 1985[22]
Proprioception deficits, including decreased perception of body position and direction of movement of body segments	Lundin et al., 1993[23]
Relevance of myotatic proprioceptive afferents is degraded due to ▪ Altered sensitivity of types Ia and II afferent fibers ▪ Decreased activation of γ-motoneurons ▪ Regression of discharge frequency of sensorial fibers of muscle spindles	Madigan et al., 2006[24]
Increased latency of EMG activity of fatigued muscles	Mello et al., 2007[25]
Hoffman (*H*) reflex amplitude decreased	Laudani et al., 2009[26]

CHAPTER 2

EFFECTS OF ACUTE MUSCLE FATIGUE ON ANTICIPATORY AND REACTIVE POSTURAL CONTROL IN OLDER INDIVIDUALS: A SYSTEMATIC REVIEW OF THE EVIDENCE

Introduction

Muscular fatigue has been defined as any exercise-induced reduction in the ability to exert muscle force or power[3] and is known to modify the neuromuscular system leading to impaired muscle performance. In addition to decrements in muscular contractile ability, muscle fatigue modifies both the peripheral proprioceptive system and the central processing of sensory inputs,[27] producing clumsiness and diminished precision of motor control.[28] Extensive studies of both general and local exercises producing acute muscle fatigue have been shown to contribute to altering the effectiveness of sensory inputs and motor output of postural control (see Table 1). There is also a preponderance of metabolic factors that are altered following localized muscle fatigue,[29] which may have an effect on postural control. These changes are not benign in terms of postural control as it has been shown that fatiguing lower extremity muscles by performing repetitive dynamic contractions induces changes in postural steadiness[30] and increases postural sway during quiet stance.[30–32]

Although fatigue degrades postural control during static stance, relatively few studies have examined the effects of fatigue during more dynamic postural tasks. A recent review highlighted 7 articles in healthy older individuals, which suggested that some components of postural control were significantly diminished immediately following muscle fatigue.[21] However, despite stating an emphasis on "functional tasks," this review included several articles investigating static stance postural stability. Recently there has been a shift away from static postural stability testing toward testing dynamic postural control as it may be more functional[33] and serve to uncover underlying sensorimotor control issues in at-risk populations.[34] Furthermore, static stance postural stability testing neglects an important discussion of the particular biomechanical outcomes utilized during daily functional tasks, specifically anticipatory and reactive aspects of postural control (see Figure 1). Anticipatory aspects of postural control are processed internally when individuals prepare sensory and motor systems for postural demands. Anticipatory postural adjustments occur in an "expectant" or feedforward manner prior to action of the prime mover. Examples of anticipatory postural control actions include transitions to single limb stance[35] and rise-to-toes tasks,[36] as well as the initiation of gait[10] and functional reach tests.[35]

Reactive postural control is defined by modifying sensory and motor systems in response to changing tasks and externally induced environmental demands. Contrary to anticipatory actions, reactive postural control mechanisms occur in a "compensatory" or feedback manner in response to some external perturbation. Models of reactive postural control research paradigms include sliding force plates,[12] treadmill perturbation training,[13] vibratory platforms,[37] and trigger-release load cell devices.[14]

Anticipatory and reactive postural control are utilized daily in dynamic conditions like walking, lifting, and carrying objects, and poor postural control increases the risk of injurious falls in older persons during these daily activities.[15] Furthermore, it has been reported that the majority of falls in older adults occur in the context of extrinsic factors, requiring reactive postural control appropriations.[16] Additionally, the chances of sustaining a fall are particularly high during slipping or tripping situations in fatigued conditions[38, 39] as may be present at the end of a day.[40, 41] Despite the apparent negative effect of fatigue on postural control, there is a paucity of information for clinicians regarding how acute muscle fatigue impacts anticipatory and reactive aspects of postural control. To address these gaps in the literature, the purpose of this paper was to systematically review how anticipatory and reactive postural control are affected by acute bouts of muscle fatigue in healthy older individuals. Such information is important in that it may influence clinical fall risk examinations and postexercise treatment precautions for patients at risk of falls, as well as provide insight into a potential target for therapeutic intervention.

Methods

Search Methodology

The goal of this systematic review was to capture studies in international medical journals, published in the English language through June 2013, which examined the effects of acute muscular fatigue on postural control outcomes during anticipatory and reactive control tasks in persons over 50 years of age. To generate the list of articles, an extensive search of the following research literature databases was conducted: Cumulative Index to Nursing and Allied Health Literature (CINAHL), MEDLINE, PubMed, SPORTDiscus, and AgeLine. The key words *fatigue, muscle, posture, postural control,* and *postural stability* were used to conduct the searches. In addition, literature was identified by bibliographic review from included studies. Initial screening of search results was performed by one author (E.P.) using titles and abstracts.

A study was included if it met the following criteria: (1) a controlled clinical trial methodology was used (meeting definitions for levels I, II, and III evidence according to the Methodology to Develop Systematic Reviews of Treatment Interventions developed by the American Academy for Cerebral Palsy and Developmental Medicine [AACPDM] [2008 version, revision 1.2]; (2) the target population were healthy individuals over the age of 50; (3) the independent variable was acute skeletal muscle fatigue of the lower extremities or trunk muscles (except diaphragm or pelvic floor muscles); (4) the outcomes included dynamic anticipatory and reactive postural control assessments; and (5) the article was available in English.

Articles excluded were review papers, methodological or descriptive papers, and articles on postural control in bilateral stance static conditions (e.g., postural sway). Articles using participants diagnosed with musculoskeletal, neurologic, or other diseases, studies performed in animals or *in vitro*, or any articles examining the effect of resistance training protocols (i.e., > a single exercise session) were also excluded. Secondary review of articles in question was made by another author (L.D.) and inclusion or exclusion decisions were made as a result of consensus decisions from L.D. and E.P. Based on these criteria, a list of final citations was generated, and full text articles were procured for full article review. Figure 2 illustrates the process of the search strategy using the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) flow sheet guidelines.

Full Article Review: Level of Evidence, Quality Assessment, and Data Extraction

Two authors (E.P., H.G.), using standardized methods, independently extracted the data from each article selected for full review. The data extraction forms included general study information (manuscript title, authors, publication year, journal), study characteristics (sample data, groups, outcome measures), and results. Study quality assessments were also performed independently by E.P. and H.G. using the AACPDM guidelines. Any discrepancies in data extraction or quality assessment were resolved by reference to the original article and discussion between both researchers.[42] If there were questions and it was possible, the original investigators were asked for additional data or clarification of methods. If the first two authors reached no consensus, a third reviewer (L.D.) made the final judgment.

The AACPDM tool rates the level of evidence on a 5-category scale founded on Sackett's levels of evidence and the National Health Service Research and Development Centre for Evidence Based Medicine (CEBM Oxford, England) (level 1 = systematic review, level 5 = expert opinion case study). In addition, it quantifies study quality by awarding 1 point for each of the following internal and external validity study characteristics: (1) well-defined inclusion and exclusion criteria, (2) intervention adequately described and adherence to intervention, (3) measures used were valid and reliable, (4) outcome assessor was blinded, (5) authors conducted tests of and reported statistical power, (6) dropouts were reported and were less than 20%, and (7) appropriate methods for controlling confounding variables were used. A score of 3 or less was considered to be low quality, a score of 4 or 5 was considered to be moderate quality, and a score of 6 or greater was considered to reflect a high-quality trial.

Results

General Aspects

A total of 334 citations were found, with 152 from SPORTDiscus, 75 from PubMed, 53 from CINHAL, 51 from MEDLINE, and 3 from AgeLine. Titles were scanned for evidence of a skeletal muscle fatigue intervention with postural control outcomes in older adults. Relevant articles were recorded and duplicates were removed, leaving 294 studies. After screening titles an additional 197 articles were excluded, and the abstracts of the remaining 97 articles were then reviewed with attention to the exclusion criteria, leaving a total of seven studies. These articles were then subjected to a full-text review (see Figure 2).

Study Design and Quality

Six out of seven articles were 2-group, prospective cohort studies, with healthy young subjects acting as controls on the main effect of age. These six articles were classified as level III on the AACPDM level of evidence scale. One study lacked a control group and was classified as level IV evidence.[35] This resulted in its exclusion from further synthesis leaving a total of six articles for inclusion and qualitative analysis. Two of the six articles were assessed a weak study quality rating,[37, 43] whereas the remaining four articles were considered to provide moderate individual study quality.[44–47]

Study Samples

All older participants were considered healthy, community dwelling adults, with a mean age of 67.5 years, range 62.2–71.7 years. The healthy young control subjects had a mean age of 24.1 years and a range of 19.4–32.0 years. All studies, except for two, described the older subjects as having no history of falls within the past year, and those two reports described their subjects as being “physically active”[47] or “active in sports.”[43] Two studies employed solely female subjects,[37, 44] two utilized a strict male cohort,[43, 47] and two remaining articles were published examining both sexes.[45, 46]

Fatigue Protocols

All fatigue protocols focused on lower extremity muscles with the exception of two articles, which fatigued the lumbar extensors in addition to the ankle plantar flexors.[45, 46] Specific muscles and/or muscle groups as well as muscle contraction types (concentric, isometric, etc.) used during fatiguing protocols can be seen in Tables 2 and 3.

Each study reported the time interval between postfatiguing exercise and initiation of posttesting differently. The time latency between the end of the fatigue bout and the initiation of postfatigue postural control testing ranged from “immediately after” exercise[37, 44, 47] to 3 and 4 minutes.[45, 46] A singular study examined post exercise recovery at four timepoints up to 20 minutes.[37]

In three of the six studies, the endpoint for fatigue protocols was based on a load related to the participants’ maximum voluntary contraction (MVC).[45–47] These endpoints differed for each study, ranging from 50–70% of patients’ MVC. One study determined the exercising endpoint by a failure to complete the task.[37] Two other articles used available active range of motion (AROM) as the benchmark for fatigue. For example, Bellew et al. defined fatigue as when subjects “failed to reach 50% AROM of their exercises” (or also a failure to keep pace with a metronome),[44] whereas Mademli et al. declared that subjects were fatigued when they could not lift the given weight through "the whole range of motion."[43]

Postural Control Paradigms

Five of the six articles reviewed utilized reactive postural control paradigms. The remaining study used an anticipatory postural control design.

Reactive Postural Control Paradigms

Two of the studies[45, 46] utilized a swinging pendulum to apply externally driven perturbations. The design provoked the largest possible perturbation that could be withstood without inducing a stepping response. Adlerton and Moritz also examined recovery from perturbation without taking a step by using vibration-induced center of pressure oscillations.[37] Two other studies allowed stepping responses but utilized either a treadmill-induced perturbation or a tether-release induced perturbation.[43, 47]

Three out of five studies investigating the effects of fatigue using external perturbations found that postural control was diminished in older individuals after acute muscle fatiguing exercise relative to pre-fatigue.[37, 45, 47] Davidson et al. found that changes in the center of mass (COM) trajectory were consistent with a localized muscle fatigue-induced decrement in the ability to recover from perturbations without stepping (COM peak displacement $p < 0.001$).[45] Likewise, Adlerton and Moritz reported an immediate but short-lasting effect of fatiguing exercise on vibration-induced center of pressure (COP) oscillations via increased COP displacement in single-limb stance ($p = 0.03$).[37] Using alternating treadmill speeds, Granacher et al. reported that acute ankle fatigue decreased functional reflex activity of the tibialis anterior ($p < 0.001$) and increased antagonist muscle co-activity ($p = 0.03$), which impacted the older individuals' ability to compensate for gait perturbations.[47]

Anticipatory Postural Control Paradigms

One of the six articles reviewed investigated the effects of fatigue on anticipatory postural control tasks.[44] This article examined the anticipatory aspect of postural stability by having subjects voluntarily initiate movement from bipedal stance into single limb stance. Accordingly, the Lower Extremity Reach Test (LERT) and a single limb balance test were employed. The LERT is a lower-extremity analog of the Functional Reach Test and has been previously described by Bellew et al.[48]

Bellew et al. investigated postural control after fatigue to musculature responsible for frontal plane stability (hip abductor muscles).[44] The authors reported no significant differences in prefatigue and postfatigue performance on the study outcomes despite reports that the subjects used considerably altered movement strategies following fatigue.

Biomechanical Postural Control Task Outcomes

The postural control task outcomes can be broadly categorized into 3 biomechanical classes: temporal measures, spatial measures, and endpoints focused on lower-extremity joint kinetics (Table 4). Four of the six[37, 43, 44, 46] articles utilizing temporal outcome assessments failed to approach statistical significance in their measures. Several of these studies noted deteriorations following muscle fatigue including slowing of reaction time,[43] shorter time to complete the postural control task,[44] and decreases in COP average angular velocity,[37] though these did not reach statistical significance. Just two of five articles employing spatial measures reported statistically significant effects of fatigue on spatial postural control outcomes; specifically increases in peak center of mass[45] and COP displacements were reported.[37] Only one article employed the use of lower-extremity kinetic measures to explore the effect of fatigue on postural control, reporting statistically significant declines in the support limb knee extension moment and vertical ground reaction forces until touchdown by the stepping limb after a fall.[43]

Statistical Analysis Considerations

The inclusion of relevant statistical design details varied between studies. Two studies provided an adjustment of the level of significance as a control for type I statistical error risk.[37, 44] For outcome measures where no statistical differences between pre- and postfatigue existed, none of the studies reported post hoc power calculations to provide estimates of type II statistical error risk. Two[45, 47] of the six studies provided post hoc effect sizes. Additionally, no studies included an a priori sample size estimate based on previous studies. In terms of reliability, two authors[44, 47] reported on tester or instrument reliability of their outcome measures. Intention-to-treat analyses and blinding of evaluators were not reported in any of the studies.

Discussion

Accidental or environment-related falls are the most frequently cited cause of falling in older individuals, accounting for 30–50% of cases. The second most common cause is postural instability and/or gait problems.[17] When muscle fatigue is added to these inherent fall risks, older individuals may become increasingly susceptible to falls.[38, 39, 41] This systematic review provides insight into the effects of lower limb and trunk muscle fatigue on reactive and anticipatory postural control in older individuals. In order to expand upon a previous narrative review,[21] this study utilized systematic methodology to consolidate biomechanical data from multiple studies utilizing dynamic postural control tasks (no static stance). Although the study methodologies varied considerably (sample sizes, fatigue protocols, and outcome measures), the composite results appear to indicate that fatigue induces postural control deficits in older individuals during tasks requiring reactive postural control (externally induced destabilizing conditions). Because of a lack of studies examining anticipatory postural control outcomes, the effects of fatigue on this type of postural adjustment remains unclear. These results are important to clinical fall risk examinations, post exercise precautions, and to identify potential targets for therapeutic intervention.

Clinical Examination of Fatigue Related Declines in Postural Control

The majority of research reviewed here coupled with studies reporting the alteration of the effectiveness of sensory inputs and motor output of postural control strongly suggests that fatigue has a measurable clinical effect on stability and potentially on fall risk. Despite this evidence, the authors are not aware of any clinical guidelines that suggest both pre- and postfatigue examination of postural control. In addition, this review emphasizes that the effects of fatigue extend beyond increases in postural sway during static stance. Such results point to the need to conduct post fatigue postural control examinations using reactive postural control tasks. Further research is needed to understand the effects of fatigue on anticipatory postural control tasks.

Postexercise Postural Control Precautions

There is no question that one of the goals of exercise in older patients is to improve function and reduce fall risk. Unfortunately, little thought is given to the potential for fatigue-induced iatrogenic falls. The research designs used in the reviewed studies all examined the acute effects of fatigue on postural control outcomes in older individuals. With the exception of one study, no regard for the time course for recovery of prefatigue levels of postural control was found. Adlerton and Moritz[37] reported that the amplitude of COP displacement increased immediately after fatigue in the sagittal plane, but returned to baseline within 5 minutes and remained there at two other timepoints up to 15 minutes. Unfortunately, there is little evidence-based guidance beyond this regarding the recovery time after fatigue for postural control measures. Reports in healthy young individuals have indicated that postural control returns to baseline in as little as 75 seconds[49] or as long as 20 minutes[31] after acute bouts of localized muscle fatigue. Hakkinen[50] found that localized muscle fatigue recovery was significantly shorter in an older group of females (70 y/o) compared to 2 younger female groups (30 & 50 y/o), but this measure was based purely on a decrease in maximal force production and did not take into account measures of postural control. In order to prevent the inadvertent increase in the risk of postfatigue iatrogenic falls, additional research is needed to examine appropriate recovery periods after localized muscle fatiguing exercises for older individuals.

A Potential Target for Fall Risk Intervention

The degradation of postural control by acute muscle fatigue would appear to reveal a potential target for intervention. If exercise programs were explicitly designed to make lower extremity muscles more fatigue resistant, the participant might derive postural control benefits. To date, several chronic muscle endurance-training studies have been employed using an amalgam of postural control outcomes.[51–55] However, these studies have employed clinical balance correlates like static stance posture, gait speed, the Berg balance test, the Dynamic Gait Index, and others, which fail to incorporate measures of reactive postural control. Although multidimensional fall risk assessment and exercise interventions have shown promise in reducing falls,[56] these interventions are generally composites of neuromuscular reeducation and lower extremity muscle strength and endurance activities. Because of this, the differential benefits of muscle endurance training versus coordination training are unclear. Controlled trials are needed to examine the efficacy of training regimens on muscle fatigue induced instability.

Experimental Design Considerations

The heterogeneity in methodologies used to induce and to measure fatigue as well as the poorly controlled threats to internal validity (small sample sizes, consistent lack of control groups) may have influenced the observed results. In addition, in several studies, there was a lack of specificity of the muscles fatigued relative to the postural control task.

The acute muscle fatigue induced in these studies can be categorized into two methodologies. One method centered on subjects' MVC and the other focused on the ability to perform repetitions of exercises within an available AROM. Two of the three articles that induced fatigue via measurements of MVC produced statistically significant reductions in measurements of postural control.[45, 47] Meanwhile, both of the articles that induced fatigue via an AROM index failed to produce significant changes.[43, 44] In the future, fatigue-inducing protocols should be based more rigorously on objective measurements of muscle force, such as MVC, than on less direct measures of force production like available AROM.

The various biomechanical postural control task outcomes employed in these studies included temporal, spatial, and lower-extremity kinetic measures. Despite significant alterations to postural control occurring across the three broad categories, the lack of a unanimous approach with clear sensitivity to postural control changes makes it difficult to suggest a particular biomechanical methodology for future investigations.

The relevance of the dependent measure to the fatigue task may have also influenced the results of the reviewed studies. Although previous research has reported older individuals to be more fatigable than young during velocity-dependent power tasks,[57] none of the dependent measures in the reviewed studies examined such tasks. In order to develop a more clear understanding of the effects of acute muscle fatigue on postural control, future research should examine a variety of postural control tasks including but not limited to rapid force production of reactive or anticipatory tasks.

Study Limitations

The study quality of articles included in this review does not meet the highest standards for quality as provided by standard systematic review guidelines and therefore, the results should be interpreted cautiously. Certainly, additional controlled trials examining the acute effects of muscle fatigue on anticipatory and reactive postural control in healthy older individuals is needed. Another limitation to this study may be the inclusion of English-language publications only, causing potentially valuable data to have been overlooked.

Conclusions and Directions for Future Research

Using systematic review methodology, this paper has demonstrated that there is a negative effect of acute muscle fatigue on postural control in older individuals. This fatigue-induced decline in postural control is apparent in dynamic conditions of reactive postural control. Collectively, this evidence points to the potential for exercise induced iatrogenic increases in fall risk. Such results have implications in the examination and management of fall risk and may be even more important in populations with pre-existing fall risk factors. Meanwhile, more work is needed to define the effect of fatigue on healthy older individuals in anticipatory postural control conditions. Future research is also needed to examine the clinical merit of pre- and postfatigue postural control examinations, the need for dissipation of fatigue effects after exercise bouts, and for the improvement of muscle endurance as a target for postural control interventions in persons with increased fall risk.

Table 2. Summary of Literature-Search Results, Anticipatory Postural Control Tasks

Study		
	Bellew et al., 2009[44]	
Population		
Aged Population	20 healthy female adults	71.65 ± 7.2 yrs
Fatigue Protocol		
Comparison Group	20 healthy young females	23.0 ± 1.5 yrs
Muscle Group	Unilateral hip abductors	
Contraction type	Concentric	
Intensity	Ankle weight at 3% of body weight	
Sets & rate or speed	Established via metronome at 25 lifts/min	
Total reps and duration	Not specified	
Endpoint (force/velocity drop)	1. Failure to reach 50% available AROM* or 2. lose sync with metronome or both for 3 consecutive repetitions	
Outcome		
Functional task	mFRT**	LERT*^
Main fatigue effect	No	No

*AROM: Active Range of Motion **mFRT: Modified Functional Reach Test
*^LERT: Lower-Extremity Reach Test

Table 3. Summary of Literature-Search Results, Reactive Postural Control Tasks

	Population		Fatigue Protocol						Perturbation	Outcome	
STUDY	Aged population	Comparison group	Muscle groups	Contraction type/ Intervention	Intensity	Sets & rate or speed	Total reps and duration	Endpoint (force/ velocity drop)	Mechanism	Functional task/ equivalent	Main fatigue effect
Davidson et al., 2009[45]	16 older adults 62.2 ± 5.1 yrs 8 M, 8 F	16 younger adults 19.4 ± 1.4 yrs 8 M, 8 F	Ankle plantar flexors Bilateral: Lumbar extensors	Concentric	45% of MVC*	1 set of 23 reps/min Rate not specified	~322 reps (~14 min)	~70% MVC	Ballistic pendulum administered in sagittal plane while eyes closed	Maximum forward trunk perturbation withstood without stepping	Yes

Table 3. Continued

Davidson et al., 2011[46]	16 older adults 62.2 ± 5.1 yrs 8 M, 8 F	16 younger adults 19.4 ± 1.4 yrs 8 M, 8 F	Ankle plantar flexors Bilateral: Lumbar extensors	Concentric /Eccentric	40% of MVC	Continuous reps with sporadic MVCs (# reps adjusted each min so that MVC dropped 30% over 14 mins)	~14 min	70% of MVC	Ballistic pendulum administered in sagittal plane while eyes closed	Neural control of dynamic postural stability/ neural controller gains and time-delay in the sensory feedback, delay margins	No
Granacher et al., 2010[47]	14 older males 67.2 ± 3.7 yrs	14 younger males 27.0 ± 3.1 yrs	Ankle plantar and dorsiflexors	Concentric	Peak torque	60deg/s ("continuous repetitions")	not indicated	50% Fmax**	Decelerating treadmill (drop from 3.5km/hr to 0.6km/hr in 0.4s)	Recovery from treadmill perturbations while walking	Yes

Table 3. Continued

Mademli et al., 2008[43]	11 older males 65 ± 3 yrs	11 younger males 32 ± 7 yrs	Bilateral knee extensors/ flexors	Concentric/ Eccentric	25% of MVC	3 sets at a rate of 1 Hz via metronome (2 sec extension, 2 sec flexion).	not indicated	To exhaustion (when subjects could not lift given weight through full ROM*^)	Unexpected release from forward lean using load cell (33% of body weight)	Single step forward	No
Adlerton and Moritz, 2001[37]	15 older females 53.9 ± 2.6 yrs.	15 younger females 21.7 ± 1.8	Ankle plantar flexors	Concentric	NA	Followed beat of metronome	varied per individual	To exhaustion (when subjects could no longer raise the heel)	Vibration (80 Hz with a peak-to-peak amplitude of 0.4 mm)	Center of pressure amplitude alterations in single leg stance	Yes

*MVC: Maximal Voluntary Contraction **Fmax: Maximal Torque *^ROM: Range of Motion

Table 4. Summary of Biomechanical Postural Control Task Outcomes

Overall Task Outcomes					
Temporal Measures		Spatial Measures		Lower Extremity Kinetic Measures	
Average COP* velocity	Adlerton and Moritz, 2001[37]	COM** kinematics	Mademli et al., 2008[43]	Hip, knee, ankle joint moments	Mademli et al., 2008[43]
EMG*^ latencies	Granacher et al., 2009[47]	Force plate COP* amplitude projections / changes	Adlerton and Moritz, 2001;[37] Davidson et al., 2009[45]		
Reaction time	Mademli et al., 2008[43]			Vertical ground reaction forces	Mademli et al., 2008[43]
Average time in single limb stance	Bellew et al., 2009[44]	Distance reached in single limb stance	Bellew et al., 2009[44]		
COP & COM** average velocities	Davidson et al., 2009[45]	Margin of stability	Mademli et al., 2008[43]		
Joint angular velocity	Davidson et al., 2011[46]				

*COP: Center of Pressure **COM: Center of Mass *^EMG: Electromyography

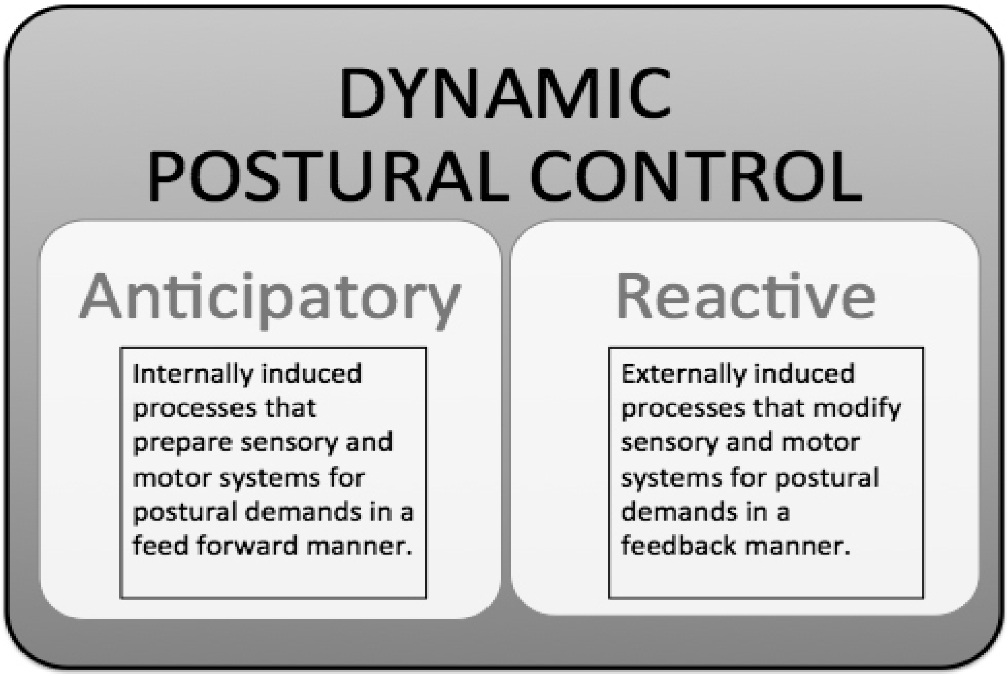


Figure 1. Diagram of Dynamic Postural Control Subtypes

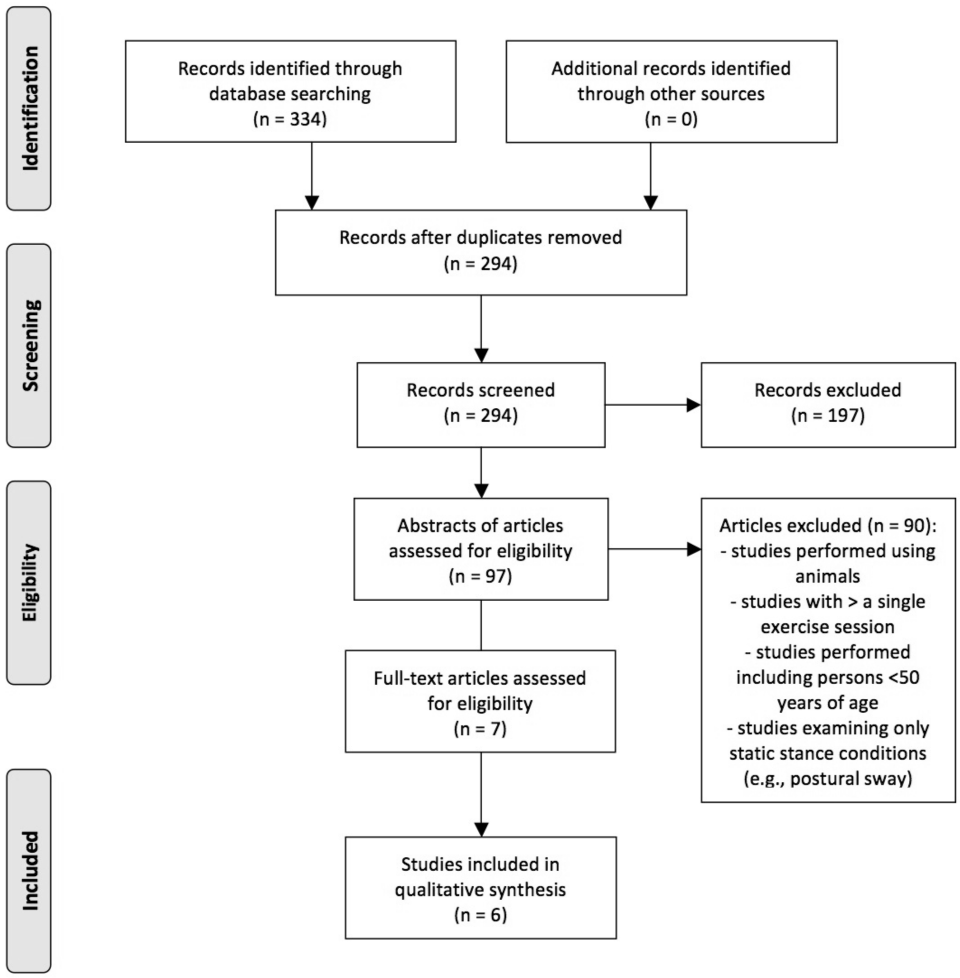


Figure 2. Literature Search Strategy, According to the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) Guidelines

CHAPTER 3

EFFECTS OF ACUTE MUSCLE FATIGUE ON ANTICIPATORY POSTURAL CONTROL IN PERSONS WITH PARKINSON'S DISEASE AND HEALTHY ADULTS

Introduction

It is easy to overlook the difficulties required to perform more-or-less automatic behaviors in everyday tasks. The many tasks involved in rising from a seated position, stepping over a raised curb, or even reaching overhead during bipedal stance place heavy demands on the systems that control posture and balance. Each of the above mentioned tasks require the use of a balance measure known as anticipatory postural control. In order to understand anticipatory postural control in the individual, one must understand what it takes to achieve postural control and examine the effect that task demands play on the individual.

When an individual voluntarily reaches overhead to retrieve an item from a cupboard, for example, their postural equilibrium becomes challenged. First, the changes in the upper extremity limb orientation lead to a change in the projection of the center of mass. Second, transmission of forces and torques from the upper extremity limb through the body's linked segments causes transient forces at other joints. However, because the central nervous system "expects" such perturbations, it induces anticipatory postural actions in a feedforward manner based on predictions of expected postural perturbations.[58, 59]

Many factors affect the initiation of anticipatory postural adjustments such as the magnitude and direction of the internal perturbation,[60] body stability,[61] and body configuration.[62] Fear of falling has also been reported to influence anticipatory postural control.[63] However, information regarding other factors that may affect the anticipatory control of posture under specific physiologic conditions is scarce. Specifically, little is known about how acute muscle fatigue influences anticipatory postural control.

The negative consequences of fatigue on anticipatory postural control have been reported in brevity in young and older individuals at the tissue-specific level as well as during task-specific demands. In healthy young individuals, early onset of activation was demonstrated in the semitendinosus muscle following isometric knee flexor fatigue.[64] Similarly, early onset of erector spinae muscle activation was seen in response to self-initiated body perturbation after dead-lift exercise to exhaustion.[65] In healthy older individuals, performance on clinical balance measures requiring feedforward control of posture (Modified Functional Reach Test, Lower-extremity Reach Test) declined after acute fatiguing exercises around the knee and ankle,[35] but not for the hip abductors.[44]

Although these studies have provided valuable insight into the effect of muscle fatigue on anticipatory postural control in healthy populations, their impact is limited due to the general paucity of investigations. Furthermore, to the author's knowledge, there is no study that has investigated the effect of muscle fatigue on the feedforward control of posture in persons with inherent postural control impairment, such as Parkinson's disease (PD). As individuals with PD are becoming increasingly advised by medical providers to seek balance and strengthening interventions clinically,[66–68] investigations into the effects of acute muscle fatigue on postural control in rehabilitation settings are needed in this cohort.

The primary focus of this study was to investigate the effect of acute muscle fatigue on anticipatory postural control in persons with PD and to compare those outcomes to a cohort of neurologically healthy adults. This study also attempted to examine the chronology of acute muscle fatigue specific to anticipatory postural control recovery. Such a design was intended to shed light on the effect that acute muscle fatigue may have on persons with known postural instability and provide insight into these fatigue effects across the lifespan. Because persons with PD are known to exhibit smaller than normal postural stability margins,[69] it was hypothesized that the negative consequences of fatigue would be magnified in this cohort, in comparison with the expected decrements in postural control in the neurologically healthy individuals.

Methods

Study Sample

Twenty-six participants were recruited from the local community including nine individuals with PD, eight healthy older adults (HO), and nine healthy young adults (HY). Sample sizes were estimated using an effect size of 0.81 (large effect) for time-to-peak center of pressure (COP) displacement following postural perturbations in previously published data[45] and estimate tables[70] based on an α level = 0.05 and 90% power. Participants were screened for self-reported musculoskeletal disorders or neurologic impairments beyond PD that could affect postural stability. Inclusion criteria for both PD and HO groups required that individuals be older than 50 years of age, and between 18–35 years of age for the HY group. In addition, all persons were required to have the ability to flex their knees greater than 90 degrees and stand on one leg for greater than 5 seconds without assistance. Hoehn and Yahr stages 1.5–3.0 were also used as inclusion criteria for persons with PD. Exclusion criteria for persons with PD included any previous surgical management of PD (pallidotomy, DBS) or motor fluctuations and/or dyskinesias uncontrolled by medications. All persons were exempted from the investigation if they participated in vigorous exercise 24 hours prior to initiating the study. This study was reviewed by the University of Utah Institutional Review Board, and participants provided informed consent prior to participation.

Instrumentation

Whole body kinetic and kinematic data were collected during all postural control assessments using a Vicon 10-camera motion analysis system (Vicon Motion Systems; Oxford, UK). Kinematic data were collected at a 250 Hz frame rate by tracking movement of subjects instrumented with reflective markers based on a standardized gait analysis marker set (Plug-In-Gait marker set; Vicon Motion Systems; Oxford, UK). Kinetic data were measured using 2 AMTI OR6 series force platform systems (AMTI; Watertown, MA) at a rate of 1000 Hz. Kinematic and kinetic data were filtered using a fourth order, low-pass, zero-phaseshift Butterworth filter at 6 and 20 Hz for trajectory and analog data, respectively.

Postural Control Assessments

The anticipatory postural control task was conducted using the Lower-extremity Reach Test (LERT), which is an assessment tool that incorporates dynamic control of single-limb balance and is considered a lower-extremity analog of the Modified

Functional Reach Test.[48] Subjects stood on one force platform on their self-reported dominant leg and were instructed to reach the other limb as far as possible while balancing on their dominant leg in single limb stance (Appendix A). The reaching tasks were performed in anterior and posterior directions. Subjects were required to maintain their maximum reach position for a 3-second count without using the reaching limb for weight bearing or stability. Any trial that did not last 3 seconds or which included the use of the reaching limb for support was excluded. Subjects were permitted to use any body motion while reaching including knee flexion on the stance limb, extension of the arms for balance, or trunk extension and rotation. Five consecutive trials were performed in each of the forward and backward reach directions, with data analysis being performed on the first three complete trials for each direction. Complete trials were those trials that were completed successfully and were free from instrumentation issues, such as loss of view of markers. Anticipatory postural control outcome measures included: reach length normalized by height, reach velocity, peak anticipatory postural adjustment (APA), peak center of pressure (COP) displacement toward stance limb, center of mass (COM) minimum, center of pressure-center of mass (COP-COM) difference, COP variability, reach foot variability, time-to-peak COP displacement, and joint angular displacements. The *peak APA* was gathered during double limb support and was defined as the peak COP displacement in the mediolateral direction toward the reaching limb and was used to quantify the magnitude of the anticipatory postural adjustment, which created the propulsive force for the transition of the COM to a position over the initial stance limb. The *peak COP displacement toward the stance limb* represented the COP control necessary for transferring the load from the reaching limb to the supporting limb for the maintenance of postural stability. The *COM minimum* represented the lowest point of the body COM in the superior-inferior direction during single limb stance. The *COP-COM difference* examined the difference between the COP position at its peak displacement in the mediolateral direction and the COM at the concomitant timepoint. The COP is considered to be a significant controller of body kinematics.[71] The COM is derived from the mass of all body segments. When considered together, the COP-COM construct provides a unique ability to capture the interplay between momentum generation and dynamic stability.[71, 72] A large COP-COM difference is indicative of robust postural control. With changes in body position during the reach task the distance between COP and COM increases. Because the individuals in this study were required to stay upright for 3 seconds for successful completion of the task, effective postural control was required to limit excessive straying of the COM outside of the functional base. Meanwhile, a small COP-COM difference may represent impaired postural control, or at least a conservative approach to postural tasks, in that the performer does not feel stable enough to allow separation of the COP and COM. *COP variability* under the stance limb during the reaching task was intended to measure lower extremity ground reaction force control and was quantified using the coefficient of variation (COV). *Foot variability* was utilized to measure the smoothness of the trajectory of the reaching limb during the LERT and was quantified using COV. More detailed operational definitions for the outcome measures can be seen in Appendix A, along with explanations of how each of the variables was calculated.

Fatigue Protocol

The principal muscle groups fatigued in this study were those required for the control of center of mass stability in posturally dynamic positions, namely the quadriceps and hip extensors. Lower extremity resistance exercise was performed on a motorized, isokinetic ergometer (EccentronTM, BTE Technologies, Inc., Hanover, MD) that appears to be like a normal seated ergometer (Appendix B). Participants resisted a motorized foot pedal that moved toward them at a self-selected pace between 20–40 rpm and experienced eccentric muscle contractions about the knee and hip extensors. Eccentric muscle contractions were utilized because they are capable of producing 2–3 times greater force than can be produced either isometrically or concentrically.[73, 74] In addition, because eccentric exercise requires a much lower energetic cost and has reduced cardiovascular activation compared to traditional concentric resistance exercise,[75–78] it minimizes the effects of cardiovascular contributions to fatigue while maximizing the effects of local muscular fatigue. Fatigue was determined by real-time biofeedback of a 30% drop in individual participant's maximal voluntary contraction (MVC), which has been shown to induce a deterioration in postural control following localized muscle fatigue.[79] This was accomplished in the present study by asking participants to "resist the pedals as hard as you can for 10 seconds" prior to beginning the fatiguing bout of exercise. Biofeedback was provided on a computer monitor with the average of four maximal effort pedal strokes being represented by a horizontal line. A second line below the first indicated a 30% decline in their baseline peak torque, and subjects were required to perform the exercise until they dropped below the 30% level for four consecutive pedal strokes. An image of the computer screen participants viewed during exercise can be seen in Appendix B.

Procedure

Postural control assessments were performed before (T0) and immediately after (T1) muscle fatiguing exercise as well as after 15 minutes (T15) and 30 minutes (T30) after exercise (Figure 3). Lower limb dominance was determined by asking which leg the subject would use to kick a ball. Prior to participants entering the Motion Capture Laboratory, the motion capture cameras and force plates were calibrated. After calibration, each participant was asked to wear form-fitting clothing and instrumented with light reflective markers over bony prominences using a modified plug in gait marker set (Plug-In-Gait marker set; Vicon Motion Systems; Oxford, UK). In addition, for all LERT trials, a fall restraint tether was attached from a trunk harness to a ceiling support to prevent any falls during the task. Prior to baseline testing, participants were exposed to 1–3 trial sessions of the LERT in order to become familiar with the testing protocols, and to overcome the fear of falling. After performing the baseline (T0) assessments, markers on the posterior aspect of the trunk and pelvis were removed, and masking tape was applied to indicate their location. This allowed for the seated fatiguing exercise to be performed without the threat of losing markers and to ensure accurate re-application of joint specific markers. Immediately (<2 mins) after the fatiguing exercise the markers were re-applied, and the immediate post (T1) postural control assessments were performed.

Statistical Analyses

Prior to statistical analysis, descriptive statistics were generated for each outcome variable. In the case where heavy-tailed outliers existed, an assessment was first made as to their validity, followed by the application of a 10% winsorization.[80] Separate paired *t*-tests were used to determine within group differences in pre-post fatigue outcome measures. In order to examine between group differences, change scores were calculated on pre-post differences for all outcome measures. Due to small sample sizes and unequally sized groups, a nonparametric Kruskal–Wallis ANOVA on the change scores was performed. Significant main effects were further examined using Mann–Whitney U post hoc pair-wise comparisons. Effects were considered statistically significant when $p \leq 0.05$. The classification of effect sizes for between-group comparisons on change scores were calculated using partial eta-squared. The above analyses were performed using Statistical Package for Social Sciences (SPSS) version 21.0. Effect sizes for within group differences were calculated in Microsoft Excel (version 14.3.7) using Cohen's d.

A subsequent analysis was performed in order to assess the potential for the commitment of type I errors in the preceding statistical methodology. Separate 3x4 (group x time) ANOVAs with repeated measures on the time factor were used to determine between-group, within-group, and interaction effects for all postural control variables of the LERT tests. If the assumption of sphericity was failed in specific variables, a Greenhouse–Geiser correction was used. Post hoc analyses were performed using the Bonferroni correction to identify differences in the time factor (T0, T1, T15, T30). Between-group differences were analyzed with Tukey's HSD post hoc comparisons, or the Games–Howell assessment when equal variances were not assumed.

Effect sizes were calculated using partial eta-squared. The above statistical analyses were performed using SPSS (version 21.0). Post hoc statistical power was estimated using G*Power[81] (version 3.1.3).

Results

Table 5 describes the anthropometric characteristics of each group. Statistical between-group differences were found for age in PD versus HY ($p = 0.02$), PD versus HO ($p < 0.001$), and HO versus HY groups ($p < 0.001$). The groups did not differ on height, weight, or BMI factors. Descriptive characteristics of total work performed and total exercise time for each group can be seen in Table 6. These data demonstrate that HY and HO cohorts generally produced more muscular work than the persons with PD.

Less Conservative Analysis

Anterior LERT

Within-Group Analysis

Several measures of postural control were affected by acute muscle fatigue with statistically significant changes occurring in the healthy older and healthy younger groups (Table 7). Two spatial-based measures were significantly affected by acute muscle fatigue in the HO group, including a 25% increase in mediolateral (M/L) COP variability ($p = 0.016$) (Figure 4) and a 160% decrease in COP-COM difference toward the stance limb ($p = 0.006$). In addition, the HO group experienced two significant changes in kinematic outcomes, specifically a 33% increase in support limb hip angular displacement ($p = 0.048$) and a 34% increase in support limb knee angular displacement ($p = 0.035$) (Figure 5).

One spatial-based measure was significantly affected by muscle fatigue in the HY group, namely a 14% increase in antero-posterior (A/P) COP variability (p = 0.007) (Figure 6), and one temporal alteration was noted, a 28% decrease in time-to-peak COP displacement toward the stance limb (p = 0.026).

No statistically significant changes were seen in the PD group.

Between-Group Analysis

Muscle fatigue induced a significant between-group effect on change scores in A/P COP variability (p = 0.029) and COP-COM difference toward the stance limb (p = 0.038) (Figure 7). Pairwise comparisons revealed a decrease in A/P COP variability in the PD group after fatigue while it increased in the HY group (p = 0.004). Additionally, COP-COM difference toward the stance limb was increased in the PD group after fatigue, whereas it decreased in the HO group (p = 0.006).

Posterior LERT

Within-Group Analysis

Acute muscle fatigue resulted in significant alterations to two spatial-based measures of postural control in HO individuals, including a 27% increase in M/L COP variability (p = 0.018) under the stance limb (Figure 4) and a 30% increase in M/L foot variability (p = 0.03) of the reaching limb (Table 8). Additionally, two significant kinematic changes were noted in the HO group, including a 36% increase in support limb knee angular displacement (p = 0.018) (Figure 5) and a 21% increase in support limb ankle angular displacement (p = 0.043).

In the HY group, two statistically significant spatial-based measures were affected by acute muscle fatigue, including a 4.3% increase in peak COP displacement toward the stance limb (p = 0.032) and a 31% decrease in COP-COM difference toward the stepping limb (p = 0.016).

No statistically significant changes occurred in the PD group.

Between-Group Analysis

Significant between-group effects were found for change scores following acute muscle fatigue in normalized reach length (p = 0.049) and knee angular displacement of the support limb (p = 0.031) (Figure 7). Pairwise comparisons revealed that reach length was decreased in the PD and HY groups while it increased in the HO group due to fatigue (p = 0.036). Additionally, knee angular displacement of the support limb was decreased in the PD group versus a large increase in the HO group after fatigue (p = 0.008).

Recovery from fatigue. None of the measures of postural control that were altered by acute muscle fatigue returned to baseline within the 30-minute postfatigue window (Table 9).

More Conservative Analysis

A statistically significant interaction effect was found between group and time for COM minimum (F = 2.77, p = 0.042, $\eta_p^2 = .201$) (Table 10). HY subjects lowered their COM further than PD and HO participants at all timepoints. However, in the presence of fatigue the HY cohort restricted the lowering of the COM, while the PD and HO groups increased the lowering of the COM (T0 Mdiff = 9.3 cm, p = 0.00 and T0 Mdiff = 9.5 cm, p = 0.00). The lowering of the COM returned toward baseline by increasing at each subsequent timepoint for the HY group, while the return toward baseline for the PD and HO groups required a decrease in the lowering of the COM (T15, Mdiff = 10.3 cm, p =

0.00; T30, Mdiff = 11.1 cm, p = 0.001 and T15, Mdiff = 10.4 cm, p = 0.001; T30, Mdiff = 11.3 cm, p = 0.001).

Statistically significant main effects of group were seen for A/P COPV (F=4.90, p=0.01, η_p^2=.471), COM minimum (F = 14.71, p = 0.00, η_p^2 = .572) as well as hip (F = 5.94, p = 0.09, η_p^2 = .351), knee (F = 39.65, p = 0.00, η_p^2 = .783), and ankle (F = 43.08, p = 0.00, η_p^2 = .797) angular displacements of the supporting limb during the anterior LERT task. Post hoc comparisons revealed that A/P COPV was significantly larger in the HY compared to the PD (Mdiff = 6.7%, p = 0.001) and HO groups (Mdiff = 5.0%, p = 0.012). The lowest point of COM was also greater in HY participants compared to PD (Mdiff = 10.4 cm, p = 0.00) and HO individuals (Mdiff = 10.6 cm, p = 0.00). Additionally, post hoc comparisons for joint kinematics revealed that HY individuals allowed for more range of motion in the support limb at the hip, knee, and ankle joints than PD subjects (Mdiff = 37.3 deg, p = 0.038; Mdiff = 101.0 deg, p = 0.001; Mdiff = 39.7 deg, p = 0.000) and HO participants (Mdiff = 36.2 deg, p = 0.041; Mdiff = 91.7 deg, p = 0.001; Mdiff = 36.4 deg, p = 0.000). No between group differences were found for the PD and HO groups.

Main group effects were also seen in the posterior LERT task for normalized step length (F = 4.90, p = 0.01, η_p^2 = .308), COM minimum (F = 20.93, p = 0.00, η_p^2 = .656), and hip (F = 15.16, p = 0.00, η_p^2 = .580), knee (F = 46.81, p = 0.00, η_p^2 = .810), and ankle (F = 21.78, p = 0.00, η_p^2 = .665) angular displacements of the supporting limb. Post hoc comparisons revealed that step length was larger in the HY group compared to the PD cohort (Mdiff = 18.9 cm, p = 0.027). The lowest point of COM was greater in HY participants than PD (Mdiff = 14.4 cm, p = 0.00) and HO individuals (Mdiff = 14.6 cm, p

= 0.00). Similarly, angular displacements of the support limb were larger in HY than HO and PD persons for the hip (Mdiff = 71.4 deg, p = 0.00; Mdiff = 72.0 deg, p = 0.00), knee (Mdiff = 65.8 deg, p = 0.00; Mdiff = 76.7 deg, p = 0.00), and ankle (Mdiff = 25.4 deg, p = 0.00; Mdiff = 27.9 deg, p = 0.00). No between group differences were found for the PD and HO groups.

Statistically significant main effects of time were seen for hip angular displacement of the support limb (F = 3.124, p = 0.049, η_p^2 = .124) in the anterior LERT and M/L COP variability in the posterior LERT (F = 6.776, p = 0.000, η_p^2 = .235) (Table 10). Pairwise comparisons revealed that hip angular displacement was significantly increased in T1 compared to T0 (Mdiff = 12.1 deg, p = 0.024), and M/L COP variability was significantly increased in T30 compared to T0 (Mdiff = 4.1%, p = 0.002).

Discussion

The purpose of this study was to quantify the effects of acute muscle fatigue on anticipatory postural control in persons with PD and neurologically healthy young and older adults. The primary hypothesis was that persons with PD would perform worse on the experimental tasks and their performance would degrade more than the neurologically healthy groups. The results of both analytical methods demonstrated performance differences between the groups. The less conservative analysis suggested there were varied differential results of fatigue within groups, while the more conservative analysis indicated only an increase in lower-extremity joint angular displacements of the supporting limb due to immediate fatigue effects. This finding, however, was corroborated by the results of the less conservative analysis. Interestingly, there were few fatigue related alterations to anticipatory postural control in persons with PD, though significant postural control deficits were seen in the neurologically healthy cohorts.

Between-Group Effects (Less Conservative)

Synthesis of the between group comparisons for the posterior LERT appears to suggest that the PD participants as a whole were hypokinetic based on their reduced reach length (Figure 7). Such findings are not surprising given the consistent demonstration of this deficit in other movement tasks.[82–84] In addition, rather than consistently increasing variability and joint excursions, the persons with PD in this study appeared to adopt a "stiffening strategy" in their spatial and kinematic measures in both anterior and posterior reach tasks. Postural stiffening is common in persons with PD in response to external perturbations, and it has been linked to increased muscle co-activation[85, 86] as well as increased stiffness of intrinsic passive elastic muscle elements.[12] Studies have shown decreased displacement of ankle joints,[36] reduced initial COM velocity,[36] and increased surface shear forces[12] in the passive period (prior to EMG onset) in response to external and internally-driven perturbations. Such results are consistent with work by Martin et al. who demonstrated that persons with PD limit their COM excursions during gait initiation in order to compensate for deficiencies in movement.[87]

It has been previously demonstrated that the fear of falling may cause persons with PD to increase their degree of support limb postural stiffness relative to HO individuals during tasks requiring feedforward control of posture.[88] In addition, specific functional tasks requiring anticipatory postural control are all associated with fear of falling in persons with PD,[89] which may contribute to the decrease in support limb joint displacement relative to HO adults seen in this study.

Within-Group Effects (Less Conservative)

Parkinson's Disease

Contrary to our hypothesis, we found no significant changes induced by muscle fatigue in anticipatory postural control endpoints in persons with PD. One reason for this may be due to the inherent fear of falling in this cohort. Fear of falling has been shown to alter the magnitude of anticipatory postural adjustments in healthy adults[63] and is more evident in PD patients when compared with healthy individuals of similar age.[88] Additionally, specific mobility impairments requiring anticipatory postural control, including difficulty in rising from a chair, difficulty turning, and start hesitation, are all associated with fear of falling in persons with PD.[89] It is possible that the fear of falling may have resulted in hypokinetic movements, which diluted the muscle fatigue effects, leading to the lack of significant results in our cohort of PD subjects.

Neurologically Healthy Individuals

Acute muscle fatigue resulted in significant alterations in each category of biomechanical properties (temporal, spatial, kinematic) of anticipatory postural control measures in both healthy older and young individuals.

Spatial factors were the most common outcome measure altered by acute muscle fatigue in healthy adults. Of those, COP variability was the most consistently altered endpoint. Results of this study demonstrate that COP variability is increased following acute bouts of muscle fatigue in both A/P (Figure 6) and M/L (Figure 4) directions of anticipatory postural control tasks and in both healthy young and older adults.

If acute muscle fatigue caused a general decline in lower extremity coordination, one would expect to see alterations in both stance and reach limbs. In HO individuals in this study a 30% increase in M/L foot variability (prime mover) was accompanied by a 27% increase in M/L COP variability. Additionally, increased variability of the foot path of the reach limb following acute muscle fatigue was concomitant with alterations in the stability of the stance (postural) limb. Specifically, a 36% increase in knee angular displacement and a 21% increase in ankle angular displacement were noted in the support limb following muscle fatigue. The increased variability of the prime mover seen in HO individuals, in the setting of an enlarged M/L COP variability and increased joint angular displacements on the stance (postural) limb, indicate that muscle fatigue induces alterations in multiple components of the postural control system. Fatigue-induced errors in the stability of stance limb kinematics suggests that the central nervous system may not be able to produce functional adaptations to preserve postural stability in the presence of fatigue.[64, 65]

Another spatial factor altered by acute muscle fatigue in this study was the construct examining the difference between COP and COM. The COP-COM construct provides a unique ability to capture the interplay between momentum generation and dynamic stability arising from functional tasks requiring anticipatory postural control. The ability to move the COM by manipulating the COP is the catalyst for the early propulsive force of the APA toward the stepping limb. A large COP-COM difference will require a concomitantly sizeable moment arm for the ground reaction forces to act for momentum generation at the start of the task. However, the greater the difference, the greater the demands on the moment arm for the body-weight vector acting around centers of joint rotation.[90] In this state, a fine degree of anticipatory postural control is required to limit extraneous movement at joint centers to keep the COM within the functional base of support. The 160% smaller peak COP-COM difference in the postexercise state in this study suggests that older individuals may adopt a conservative strategy to limit the need for dynamic stability during fatigued conditions.

The effort to maintain a smaller COP-COM distance allows older subjects to reduce the mechanical and postural challenge of initiating the balance task. Han and Chou suggested that a reduced COP-COM difference would decrease the degree of muscular strength required for postural tasks because of the smaller moment arms created for the body weight vector acting around joint rotation centers in the support limb.[91] For older individuals who are known to have limitations in muscular strength,[92] particularly in fatigued states,[93] this strategy may be an important compensatory tactic for initiating and maintaining balance control during anticipatory functional tasks in fatigued conditions like rising from a chair, stepping over obstacles, and gait initiation.

The most common kinematic change seen in immediate response to the provocation of muscle fatigue was an increase in angular displacement of the support limb (Figure 5). Healthy older individuals presented with increased joint movement at the hip and knee during the postfatigue anterior LERT test and increases in knee and ankle displacements during the posterior LERT test. This increase in joint displacements suggests a decrement in the ability to properly support the body's mass during the fatigued condition. Increased joint angular displacement could be caused by a fatigue-induced impairment in proprioceptive acuity. Taimela et al. reported that a decrease in proprioceptive impulses after trunk extensor fatigue resulted in larger movements of the lumbar spine and, consequently, greater postural sway.[94] A decline in proprioception could decrease the sensory ability necessary to generate appropriate joint postural corrections[95] and, coupled with a decline in the ability to generate power due to muscle fatigue, could result in increased postural instability. Furthermore, this phenomenon could lead to an increased risk of falls in at-risk older adults when muscle fatigue is present, such as postexercise conditions or even at the end of the day.[40]

Recovery from Fatigue

The results of this less conservative analysis suggest that acute muscle fatigue alters anticipatory postural control beyond 30 minutes of rest. Regardless of age, task, or the presence of neurologic disease, each of the 12 postural control variables altered by acute muscle fatigue in this study failed to return to baseline.

These data contradict previously published reports examining the recovery of postural control after acute fatigue in neurologically healthy adults. Independent of age, studies examining general whole body fatigue have been shown to decrease postural control on average 14.6 minutes before returning to baseline.[96–100] The aggregate of studies examining localized muscle fatigue have shown that postural control returns to baseline across all age groups on average within 8.2 minutes.[31, 37, 49, 101, 102] This study, however, could not replicate these results. One reason for this prolonged fatigue effect may be due to the method of localized muscle fatigue employed.

Each of the aforementioned studies examining the effects of localized muscle fatigue on postural control utilized combinations of concentric and concentric/eccentric muscle contractions. The innovation in this design is that participants utilized a form of high force eccentric resistance exercise as a means of inducing skeletal muscle fatigue. Eccentric muscle contractions are capable of producing 2–3 times greater force than can be produced either isometrically or concentrically.[73, 74] Consequently, this intervention provided extremely high loads to the muscle in the shortest amount of time. Because eccentric exercise requires a much lower energetic cost and has reduced cardiovascular activation compared to traditional concentric resistance exercise,[75–78] it minimized the effects of cardiovascular causes to muscle fatigue while maximizing the effects of local muscular and neurologic contributions to fatigue. These heightened fatigue effects induced by eccentric muscle contractions may have been the cause of the prolonged recovery window for postural control measures to return to baseline.

Mixed-Design Effects (More Conservative)

Results of the group x time repeated measures ANOVAs provided a more conservative explanation for the effect of muscle fatigue on postural control measures during the anticipatory tasks. Just two measures were statistically significantly different in the main effect of time. Of those, just one demonstrated immediate pre-post fatigue effects, and it was in agreement with the results of the least conservative test. Regardless of group assignment, hip angular displacement of the support limb was significantly altered in the immediate post-fatigue exercise session, compared to baseline. As previously stated, the increases in joint angular displacement may be caused by a fatigue-induced impairment in proprioceptive acuity.[94]

Just one statistically significant result was found on the group and time interaction effect with the mixed design. The restriction of the lowering of the COM in the HY cohort while concomitantly increasing support limb lower-extremity joint angular displacements suggests that these individuals have the ability to alter their postural control strategy during a single limb stance task in the presence of fatigue. Meanwhile, the HO and PD cohorts increased the lowering of the COM in the immediate fatigued condition while also increasing joint angular displacements. This suggests a less adaptable approach to postural control tasks. Indeed, elderly individuals appear to be more restricted in modulating reflex responses during balance tasks,[103, 104] and persons with PD have demonstrated an inability to alter the H-reflex amplitude in response to self-initiated forward leaning.[105]

The lack of numerous statistically significant time and interaction effects in this method of analysis suggests that a number of type I errors were likely committed during the seminal analysis. In addition, this more conservative approach indicates that the study lacked adequate statistical power. The post hoc power analysis for the within-group (time) factor and the interaction effect in this more conservative analysis averaged 0.17 and 0.09 for the anterior and posterior LERT tasks, respectively.

From a research design standpoint, the lack of statistical power may have resulted from a small sample selection. Too few subjects were recruited because the sample sizes were estimated based on large effect sizes in healthy young and older persons.[45] A correction for this going forward would be to increase the sample population by making estimates based on the *lowest* effect size. In addition, the power in a statistical test is influenced by the variance within a data set. The goal for data collection of this anticipatory task was to examine the organic nature of the LERT, which included a de-constraining of the movements via fewer instructions on how to perform the task. However, this allowed for increased introduction of movement variability within persons and across groups. When the variability within individuals and groups is large, differences between groups become less obvious, and statistical power is attenuated.

In spite of this, multiple between-group differences were found. Moreover, the post hoc power for these differences averaged 0.98 for the anterior and posterior LERT tasks. While these differences might have been expected, it is insightful to note that HY persons moved faster and more robustly, demonstrating a more rapid reach velocity and a larger COP-COM difference than PD and HO individuals. In addition, the HY group moved with increased joint range of motion and center of pressure variability. The fact that these individuals can move in a swift and robust manner, while restricting the lowering of the COM in fatigued conditions, further supports their ability to dynamically alter their movement strategies in order to maintain an upright position. Meanwhile, HO individuals and persons with PD rely on a more rigid state for upright balance during tasks requiring anticipatory postural control.[84, 106] Persons with PD have demonstrated reduced magnitudes of movement and delayed timing of muscles during an anticipatory rise-to-toes task.[36] Meanwhile, HO individuals have been shown to rely predominately on proximal muscle strategies in comparison to HY individuals who employ a combination of postural strategies in the context of anticipatory postural adjustments.[107]

Conclusions

When interpreted conservatively, this study suggests that HY persons move dynamically, with more robust control of posture than HO individuals and persons with PD during anticipatory tasks. When introduced to bouts of acute muscle fatigue, the HY persons were able to alter their postural strategy compared to a less adaptable response by HO and PD persons to maintain an upright position. Regardless of group assignment, fatigue caused an increase in joint range of motion throughout the kinematic chain. However, few other statistically significant results were found on the effect of fatigue.

The cautious interpretation of this analysis would suggest that fatigue has a constrained effect on tasks requiring anticipatory postural control. Future investigations should employ increased sample sizes, greater standardization of the anticipatory task, and greater control over the level of induced fatigue.

A more liberal interpretation of the data in this study suggests that acute muscle fatigue has deleterious effects on the feedforward control of posture in healthy young and older adults, with the primary impairments being seen in spatial and kinematic measures. Recovery of postural control during anticipatory tasks may extend beyond 30 minutes. These results have implications for neuromuscular rehabilitation involving balance and muscular fatigue components. Accordingly, clinical balance tests utilizing anticipatory postural control (e.g., the functional reach test) should be performed both before and after physical effort for HO adults at risk for falls. More research is needed in PD cohorts with improved construct validity of muscle fatigue and larger sample sizes.

Table 5. Characteristics of Study Participants

Characteristic	PD (n = 9)	HO (n = 8)	HY (n = 9)
Age (years)	69.5 ± 10.0 (62.4–76.6)[a, b]	59.7 ± 3.4 (54.6–62.4)[c]	26.0 ± 3.1 (23.6–28.4)
Body height (m)	1.7 ± 0.1 (1.6–1.8)	1.8 ± 0.1 (1.7–1.9)	1.7 ± 0.1 (1.6–1.8)
Body weight (kg)	75.2 ± 14.7 (64.7–85.7)	95.1 ± 20.7 (77.8–112.4)	73.4 ± 19.4 (58.6–88.3)
Body mass index (kg/m^2)	25.6 ± 5.4 (21.8–29.5)	28.3 ± 6.2 (23.1–33.5)	24.7 ± 4.8 (21.0–28.4)
UPDRS(motor)	25.6 ± 5.4 (21.7–29.5)		
Hoehn & Yahr	2.3 ± 0.8 (1.6–2.9)		

Values are mean ± SD (95% Confidence Intervals)
a-significant difference between PD and HO groups ($p < 0.05$)
b-significant difference between PD and HY groups ($p < 0.001$)
c-significant difference between HO and HY groups ($p < 0.001$)

Table 6. Total Work and Total Exercise Time for Each Participant, Categorized by Group

HY			**HO**			**PD**		
Subject	Total Work	Total Time	Subject	Total Work	Total Time	Subject	Total Work	Total Time
HY1	16,333	4:56	HO1	254,935	60:00	PD1	107,737	32:26
HY2	113,643	30:31	HO2	72,762	27:18	PD2	6,482	5:21
HY3	65,446	18:15	HO3	9,619	15:04	PD3	1,789	3:56
HY4	84,322	33:22	HO4	13,862	4:36	PD4	2,743	15:10
HY5	108,389	35:10	HO5	50,578	16:29	PD5	15,640	4:34
HY6	50,868	25:53	HO6	13,751	7:59	PD6	7,096	6:15
HY7	199,958	41:20	HO7	39,931	11:07	PD7	1,326	3:48
HY8	52,673	36:13	HO8	54,547	14:35	PD8	21,962	16:40
HY9	290,276	60:00				PD9	71,323	19:01
	109,100	31:44		63,748	19:33		26,233	11:54

PD: Parkinson's disease group; HO: Healthy older group; HY: Healthy young group

Table 7. Means ± Standard Deviations and Effect Sizes for Outcome Measures of the Anterior Lower-Extremity Reach Test (LERT), Organized by Biomechanical Category (Less Conservative Analysis)

Dependent measure	**HY**			**HO**			**PD**		
	PRE	**POST**	**ES**	**PRE**	**POST**	**ES**	**PRE**	**POST**	**ES**
SPATIAL									
Reach length (normalized) (m)	.376 ± .05	.369 ± .08	.25	.369 ± .09	.373 ± .08	.08	.337 ± .13	.355 ± .15	.49
M/L COP variability (%)	19.9 ± 8.1	19.2 ± 5.5	.15	22.0 ± 9.8	27.7 ± 12.4*	1.12	18.8 ± 6.7	20.9 ± 10.6	.38
A/P COP variability (%)	12.3 ± 4.7	14.1 ± 5.4*	1.21	8.4 ± 3.1	8.3 ± 1.3	.03	7.3 ± 1.9	6.4 ± 2.3	.48
M/L Foot variability (%)	16.4 ± 9.1	19.0 ± 10.1	.56	13.6 ± 5.9	18.5 ± 7.7	.70	12.9 ± 5.2	14.9 ± 5.8	.37
Peak APA (cm)	2.01 ± 3.0	2.08 ± 4.1	.25	2.33 ± 5.1	2.28 ± 1.5	.17	2.34 ± 3.0	2.42 ± 3.0	.36
COP/COM difference (step) (m)	-.034 ± .02	-.034 ± .03	.11	-.004 ± .04	-.006 ± .05	.21	-.006 ± .03	-.002 ± .04	.18
COP/COM difference (stance) (m)	.014 ± .02	.013 ± .02	.09	.005 ± .02	-.004 ± .02*	1.37	.003 ± .01	.007 ± .02	.47

Table 7. Continued

TEMPORAL									
Reach velocity (normalized) (m/s)	.160 ± .05	.189 ± .09	.53	.252 ± .09	.258 ± .12	.01	.144 ± .06	.211 ± .10	.70
Time_COP displacement_stance (s)	2.65 ± 1.1	1.90 ± .83*	.91	2.18 ± .53	2.26 ± .54	.14	3.35 ± .97	2.97± 1.08	.52
KINEMATIC									
Hip angular displacement (support) (deg)	86.5 ± 32.1	99.5 ± 27.5	.68	49.0 ± 14.0	65.2 ± 20.4*	.85	54.9 ± 19.4	62.0± 28.6	.39
Knee angular displacement (support) (deg)	145.1±42.0	149.8±37.8	.19	50.3 ± 12.7	67.6 ± 18.1*	.92	45.4 ± 12.5	51.6± 20.1	.28
Ankle angular displacement (support) (deg)	68.8 ± 15.3	67.3 ± 15.2	.13	29.2 ± 7.7	34.4 ± 8.9	.51	28.2 ± 6.1	28.9 ± 8.3	.08

PRE pre fatigue, *POST* post fatigue, *ES* effect size (Cohen's d), *M/L* medio-lateral, *COP* center of pressure, *Peak APA* peak anticipatory postural adjustment or COP shift toward stepping limb at the start of the task, *COP/COM difference* delta between COP at its peak and center of mass at the concomitant timepoint for movements occurring toward stepping and stance limbs, *Time_COP displacement_stance* time to achieve peak COP shift toward the stance limb, *Angular displacement (support)* joint angular displacement of support limb.

* Significant main effect of fatigue ($p < 0.05$)

Table 8. Means ± Standard Deviations and Effect Sizes for Outcome Measures of the Posterior Lower-Extremity Reach Test (LERT), Organized by Biomechanical Category (Less Conservative Analysis)

Dependent measure	Young			Older			PD		
	PRE	**POST**	**ES**	**PRE**	**POST**	**ES**	**PRE**	**POST**	ES
SPATIAL									
Reach length (normalized) (m)	.538 ± .10	.526 ± .09	.35	.393 ± .10	.426 ± .10	.82	.331 ± .16	.306 ± .17	.50
M/L COP variability (%)	19.1 ± 7.9	19.6 ± 4.5	.08	22.6 ± 10.2	28.7 ± 14.2*	1.09	15.7 ± 5.9	18.6 ± 7.1	.76
A/P COP variability (%)	10.0 ± 4.0	10.1 ± 4.1	.27	8.2 ± 2.7	7.7 ± 1.8	.15	6.9 ± 2.5	6.9 ± 4.1	.02
M/L Foot variability (%)	27.1 ± 11.0	25.0 ± 12.3	.19	14.5 ± 11.0	19.0 ± 11.9*	.96	20.9 ± 12.2	24.1 ± 14.3	.38
Peak APA (cm)	2.00 ± 4.0	2.10 ± 4.0	.35	2.42 ± 5.1	2.33 ± 4.0	.29	2.42 ± 3.0	2.29 ± 2.5	.67
COP/COM difference (step) (m)	-.040 ± .03	-.030 ± .03*	.71	.001 ± .04	-.001 ± .05	.71	-.009 ± .03	-.003 ± .03	.49
COP/COM difference (stance) (m)	.020 ± .02	.020 ± .02	.00	.000 ± .02	-.002 ± .02	.00	.008 ± .02	.007 ± .02	.14

Table 8. Continued

TEMPORAL									
Reach velocity (normalized) (m/s)	.238 ± .08	.236 ± .66	.09	.294 ± .15	.250 ± .08	.48	.207 ± .20	.213 ± .23	.52
Time_COP displacement_stance (s)	2.06 ± .66	2.21 ± .88	.12	2.53 ± 1.2	2.06 ± .51	.41	3.40 ± 1.6	3.26 ± 1.6	.19
KINEMATIC									
Hip angular displacement (support) (deg)	154.6 ± 26.1	147.3 ± 28.7	.33	79.0 ± 23.6	83.7 ± 18.8	.30	83.3 ± 40.3	75.5 ± 44.7	.41
Knee angular displacement (support) (deg)	124.1 ± 30.0	122 ± 22.3	.10	48.8 ± 11.3	66.7 ± 17.8*	1.08	49.2 ± 18.4	46.4 ± 14.8	.18
Ankle angular displacement (support) (deg)	58.6 ± 14.2	56.4 ± 11.8	.26	28.7 ± 7.2	34.6 ± 7.3*	.87	31.2 ± 10.9	30.1 ± 9.2	.12

PRE pre fatigue, *POST* post fatigue, *ES* effect size (Cohen's d), *M/L* medio-lateral, *COP* center of pressure, *Peak APA* peak anticipatory postural adjustment or COP shift toward stepping limb at the start of the task, *COP/COM displacement* delta between COP at its peak and center of mass at the concomitant timepoint for movements occurring toward stepping and stance limbs, *Time_COP displacement_stance* time to achieve peak COP shift toward the stance limb, *Angular displacement (support)* joint angular displacement of support limb.
* Significant main effect of fatigue ($p < 0.05$)

Table 9. Summary of Timeline for Recovery of All Anticipatory Postural Control Measures Altered by Acute Muscle Fatigue (Less Conservative Analysis)

VARIABLE	GROUP	FATIGUE				P ANOVA
		Pre (T0)	Post1 (T1)	Post2 (T15)	Post3 (T30)	
Anterior LERT						
M/L COPV (%)	HO	22.0 ± 9.7	*27.6 ± 12.4	27.1 ± 12.5	26.8 ± 12.4	0.01
COP-COM DIFF_STANCE (m)	HO	.005 ± .01	*-.003 ± .02	*-.002 ± .02	.001 ± .01	0.03
HIP_ANG_DISP (deg)	HO	49.9 ± 14.0	*65.2 ± 20.3	60.0 ± 23.1	64.0 ± 32.0	0.11
KNEE_ANG_DISP (deg)	HO	50.3 ± 12.7	*67.6 ± 18.1	54.4 ± 10.2	53.2 ± 7.3	0.01
A/P COPV (%)	HY	12.3 ± 4.7	*14.1 ± 5.4	13.1 ± 6.1	13.7 ± 5.4	0.26
TIME_COP_DISP_STANCE (s)	HY	2.65 ± 1.1	*1.89 ± 0.83	2.11 ± 0.67	2.34 ± 0.78	0.11

Table 9. Continued

Posterior LERT						
M/L COPV (%)	HO	22.5 ± 10.1	*28.6 ± 14.2	27.2 ± 12.2	*28.4 ± 11.7	0.01
M/L FOOT VARIABILITY (%)	HO	14.5 ± 11.0	*18.9 ± 11.8	18.7 ± 10.9	18.3 ± 14.5	0.33
KNEE_ANG_DISP (deg)	HO	48.8 ± 11.3	*66.7 ± 17.8	*56.7 ± 10.1	54.9 ± 10.6	0.01
ANKLE_ANG_DISP (deg)	HO	28.7 ± 7.2	*34.6 ± 7.3	32.3 ± 4.6	31.7 ± 8.9	0.20
PK_COP_DISP_STANCE (m)	HY	0.323 ± 0.08	*0.339 ± 0.08	0.332 ± 0.08	0.328 ± .08	0.03
COP-COM_DIFF_STEP (m)	HY	-0.038 ± 0.03	*-0.02 ± 0.03	-0.034 ± 0.03	-0.037 ± .03	0.05

Values are means ± SD. $p < 0.05$ indicates statistically significant main effect of time. * indicates statistically significant pairwise comparison from baseline. LERT Lower extremity reach test, HO Healthy older group, HY Healthy young group, M/L COPV Medial-lateral center of pressure variability, COP-COM_DIFF_STANCE Center of pressure-center of mass difference toward the step limb, HIP_ANG_DISP Hip angular displacement of support limb KNEE_ANG_DISP Knee angular displacement of support limb, A/P COPV Antero-posterior center of pressure variability, TIME_COP_DISP_STANCE time to peak center of pressure displacement toward stance limb, M/L FOOT VARIABILITY, Variability of reach limb foot, ANKLE_ANG_DISP Ankle angular displacement of support limb, PK_COP_DISP_STANCE Peak center of pressure displacement toward stance limb, COP/COM_DIFF_STEP Center of pressure-center of mass difference toward the reach limb

Table 10. Means ± Standard Deviations for Anticipatory Postural Control Measures Altered Across All Timepoints by Acute Muscle Fatigue (More Conservative Analysis)

VARIABLE	GROUP	FATIGUE				P ANOVA		
		Pre (T0)	Post1 (T1)	Post2 (T15)	Post3 (T30)	*Group*	*Time*	*Interaction*
Anterior LERT								
HIP_ANG_DISP (deg)	PD	54.9 ± 19.4	62.0 ± 28.6	61.2 ± 37.2	55.7 ± 24.6	**0.009**[a, c]	**0.049**[d]	0.880
	HO	49.9 ± 14.0	65.2 ± 20.3	60.0 ± 23.1	64.0 ± 32.0			
	HY	86.4 ± 32.1	99.5 ± 27.4	98.2 ± 33.5	98.9 ± 35.4			
COM_MIN (cm)	PD	0.45 ± 0.34	0.65 ± 0.59	◆0.42 ± 0.48	0.51 ± 0.49	**0.000**[a, c]	0.187	**0.042**
	HO	0.33 ± 0.42	0.43 ± 0.54	◆0.33 ± 0.44	0.28 ± 0.31			
	HY	11.5 ± 7.5	9.98 ± 7.1	10.7 ± 8.2	*11.7 ± 9.3			
Posterior LERT								
M/L COPV (%)	PD	15.6 ± 5.8	18.6 ± 7.1	19.1 ± 9.5	21.4 ± 9.8	0.143	**0.000**[f]	0.100
	HO	22.5 ± 10.1	28.6 ± 14.2	27.2 ± 12.2	28.4 ± 11.7			
	HY	19.1 ± 8.3	19.3 ± 4.7	18.3 ± 6.5	19.7 ± 6.8			

bold- indicates statistically significant effect, a) Tukey HSD post hoc difference between PD and HY groups, b) Tukey HSD post hoc difference between PD and HO groups c) Tukey HSD post hoc difference between HO and HY groups, d) Bonferroni pairwise comparisons difference between T0 and T1, e) Bonferroni pairwise comparisons difference between T0 and T15, f) Bonferroni pairwise comparisons difference between T0 and 30, g) Bonferroni pairwise comparisons difference between T1 and T15, ◆ indicates return to baseline, HIP_ANG_DISP angular displacement of support limb hip joint, COM_MIN lowest point of COM descent in superior-inferior direction, M/L COPV medio-lateral center of pressure variability

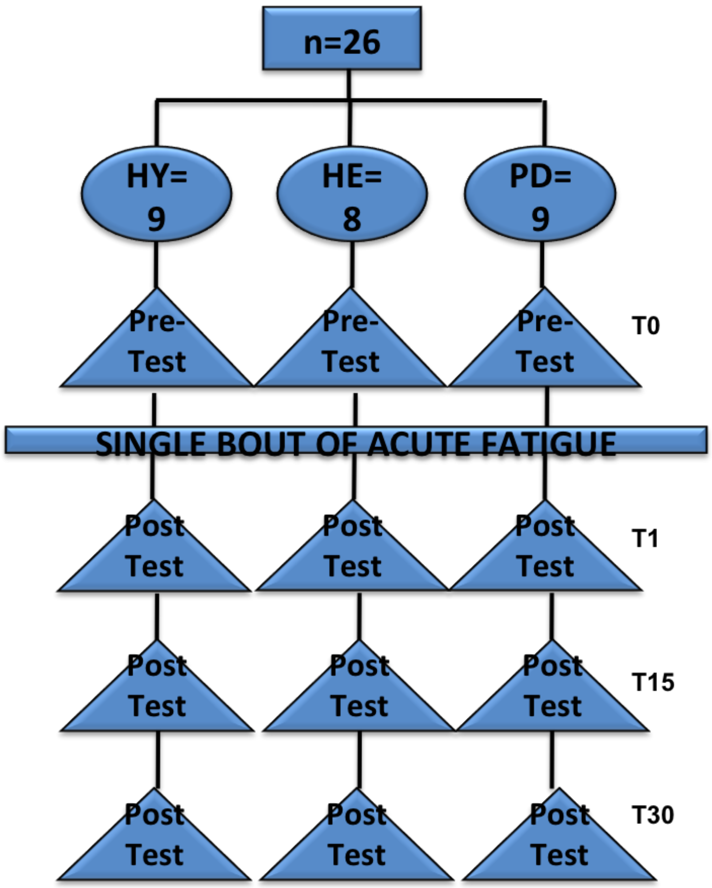


Figure 3. Flow Diagram of Study Procedures

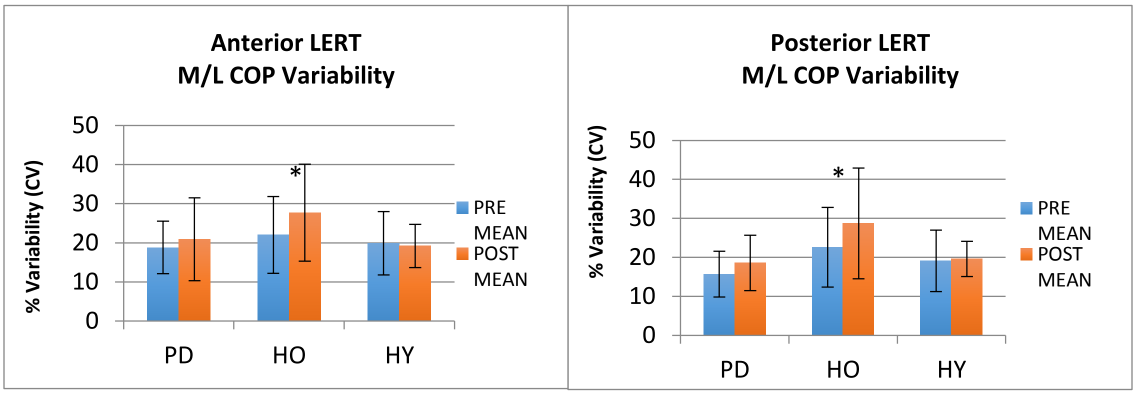


Figure 4. Center of Pressure (COP) Variability in Medio-Lateral (M/L) Directions for Lower-Extremity Reach Tasks (LERT) Performed in Anterior and Posterior Directions (Less Conservative Analysis)

PD: Parkinson's disease group, HO: Healthy older group; HY: Healthy young group
* Significant difference ($p < 0.05$)

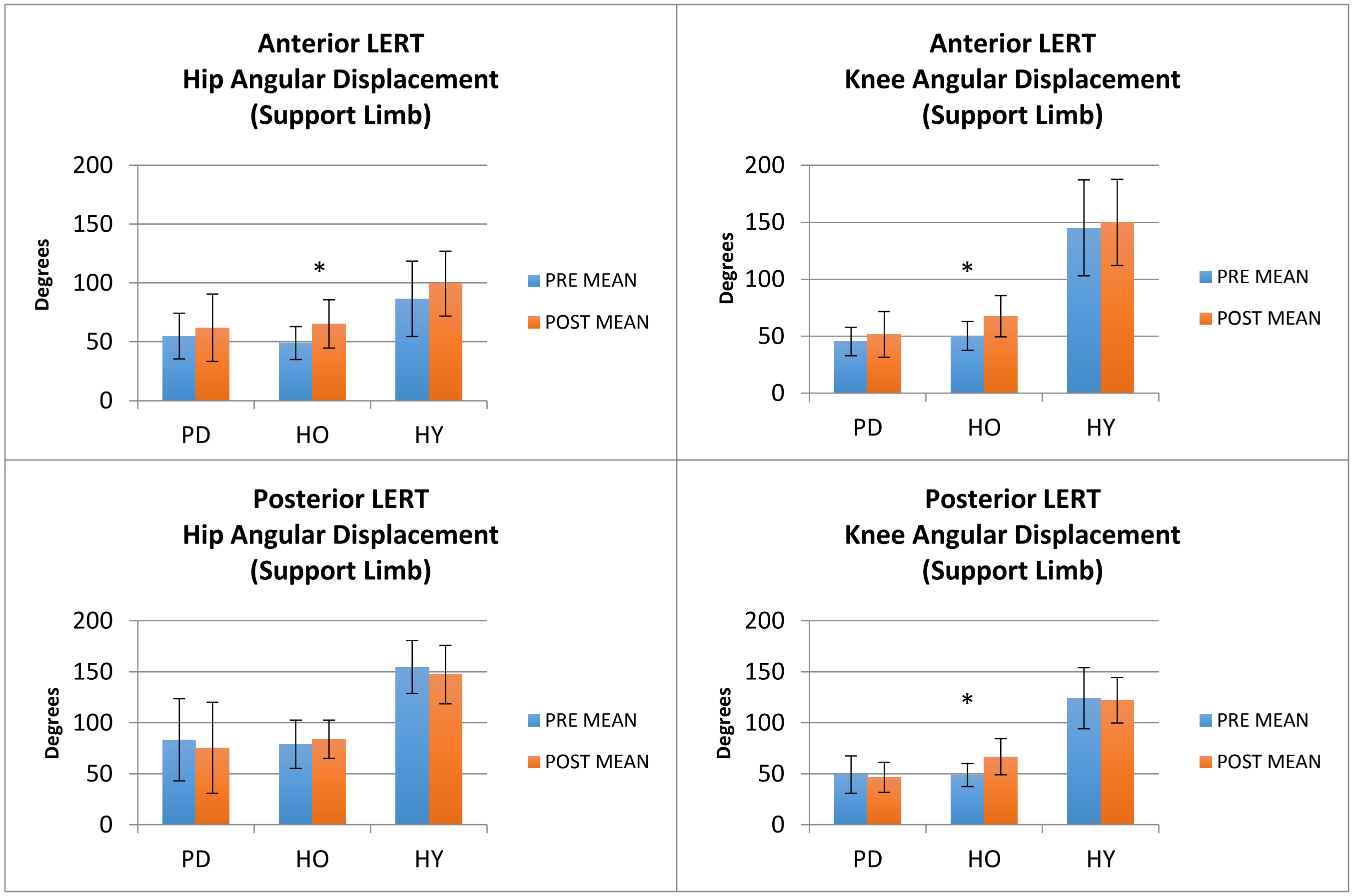


Figure 5. Angular Displacements of the Hip and Knee of the Support Limb During Anterior and Posterior Lower-Extremity Reach Tasks (LERT) (Less Conservative Analysis)

PD: Parkinson's disease group; HO: Healthy older group; HY: Healthy young group
* Significant difference ($p < 0.05$)

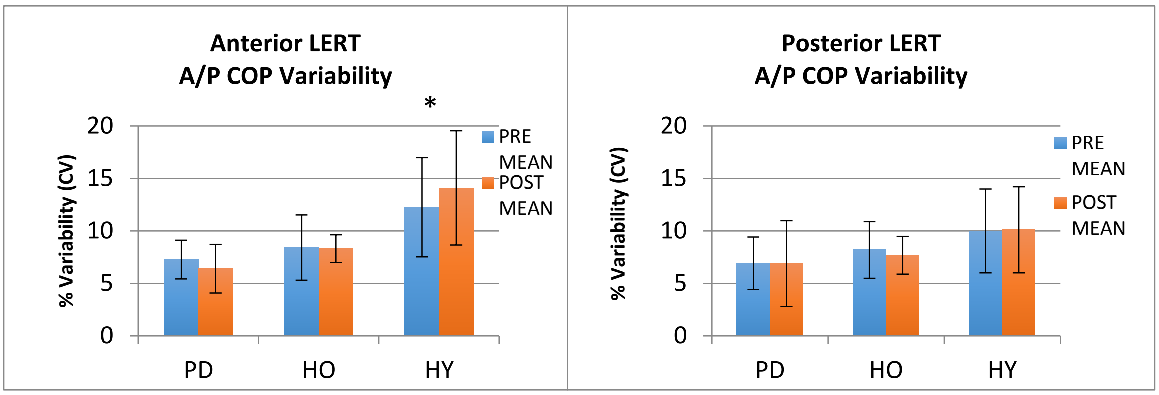


Figure 6. Center of Pressure (COP) Variability in Antero-Posterior (A/P) Directions for Lower-Extremity Reach Tasks (LERT) Performed In Anterior and Posterior Directions (Less Conservative Analysis)

PD: Parkinson's disease group; HO: Healthy older group; HY: Healthy young group
* Significant difference ($p < 0.05$)

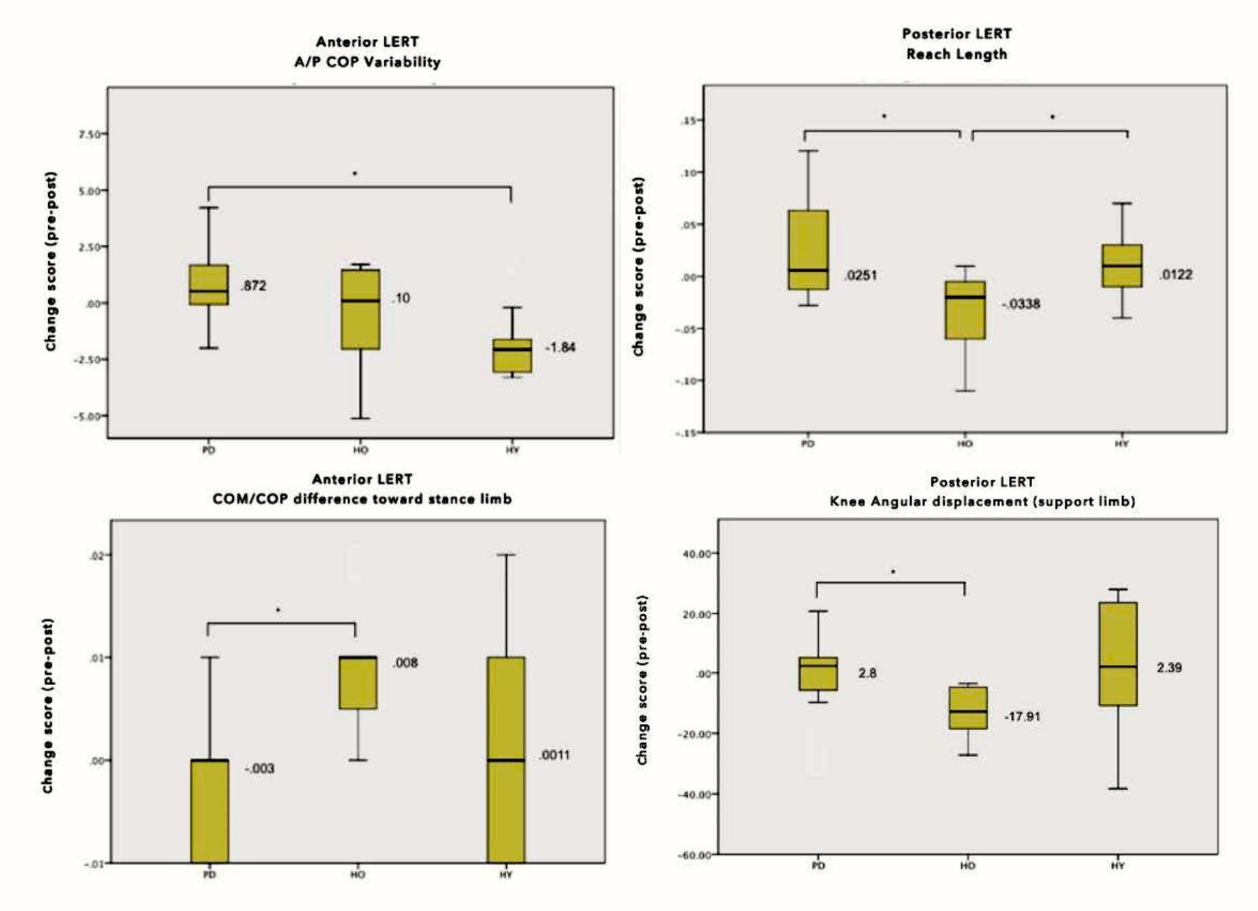


Figure 7. Change Scores (Pre-Post Fatigue) for Outcome Measures of the Lower-Extremity Reach Test (LERT) (Less Conservative Analysis)

PD: Parkinson's disease group; HO: Healthy older group; HY: Healthy young group
Whiskers represent 95% confidence intervals * - significant main effect of fatigue

CHAPTER 4

EFFECTS OF ACUTE MUSCLE FATIGUE ON REACTIVE POSTURAL CONTROL IN PERSONS WITH PARKINSON'S DISEASE AND HEALTHY ADULTS

Introduction

Falls are not a new problem in Parkinson's disease (PD). Indeed, James Parkinson made several references to falls in his seminal description of PD or "paralysis agitans."[108] Risk factors for falls in persons with PD include disease duration, previous falls, dementia, and loss of arm swing.[109] Preventing falls has become one of the most important unmet needs in PD, and it has been suggested that potential strategies to prevent falls should focus on intrinsic (patient-related) factors.[18] Despite the need for investigations into the intrinsic causes of falls in this population, studies have generally neglected the impact that personal physiologic conditions, such as muscle fatigue, have made on postural control.

While no studies have examined the acute effects of fatigue on postural control in persons with PD, postural control is known to diminish following intense bouts of resistance exercise in healthy individuals.[110–113] Muscle fatigue has specifically been shown to modify both the peripheral proprioceptive system and the central processing of sensory inputs,[27] both of which are integral for the response-oriented control of posture. Additionally, acute muscle fatigue has been shown to alter postural control in healthy adults in response to slip-induced falls.[39] Recovery from these external perturbations requires reactive postural control, which is employed on a daily basis in response to changing environmental demands. Indeed, it has been reported that the majority of falls in older adults occur in the context of tasks requiring reactive postural control appropriations.[16] Furthermore, the chances of sustaining a fall are particularly high during slipping or tripping situations in fatigued conditions[38] such as may be present at the end of the day.[40]

Despite substantial evidence in healthy populations, investigators have yet to examine fatigue's influence on reactive postural control for persons with PD. Persons with PD exhibit smaller than normal postural stability margins,[114] and their fall risk propensity is around twice that of individuals in the general population.[18] Meanwhile, persons diagnosed with Parkinson's disease are becoming increasingly advised by medical providers to seek strength and mobility training interventions clinically.[66, 67] Intuitively, this would suggest that in the immediate postexercise period muscle fatigue may push individuals with PD closer to their already lowered falls threshold. The purpose of this investigation was to characterize the previously unexplored effects of acute muscle fatigue on reactive postural control in persons with Parkinson's disease and to compare those effects to neurologically healthy older and young adults. Additionally, this investigation sought to examine the chronology of recovery from acute muscle fatigue relative to reactive postural control settings. Such a design was intended to shed light on the effect that acute muscle fatigue may have on persons with known postural instability and provide insight into these fatigue effects across the lifespan. It was hypothesized that muscle fatigue would cause declines in reactive postural control outcomes for all groups, with the greatest decrements being seen in individuals with Parkinson's disease.

Methods

Study Sample

Twenty-six participants were recruited from the local community including nine individuals with Parkinson's disease, eight healthy older adults (HO), and nine healthy young adults (HY). Sample sizes were estimated using an effect size of 0.81 (large effect) for time-to-peak center of pressure (COP) displacement following postural perturbations in previously published data[45] and estimate tables[70] based on an α level = 0.05 and 90% power. Participants were screened for self-reported musculoskeletal disorders or neurologic impairments beyond PD that could affect postural stability. Inclusion criteria for both PD and HO groups required that individuals be older than 50 years of age and between 18–35 years of age for the HY group. In addition, all persons were required to have the ability to flex their knees greater than 90 degrees and stand on one leg for greater than 5 seconds without assistance. Hoehn and Yahr stages 1.5–3.0 were also used as inclusion criteria for persons with PD. Exclusion criteria for persons with PD included any previous surgical management of PD (pallidotomy, DBS) or motor fluctuations and/or dyskinesias uncontrolled by medications. All persons were exempt from the investigation if they participated in vigorous exercise 24 hours prior to initiating the study. This study was reviewed by the University of Utah Institutional Review Board, and participants provided informed consent prior to participation.

Instrumentation

Whole body kinetic and kinematic data were collected during all postural control assessments using a Vicon 10-camera motion analysis system (Vicon Motion Systems; Oxford, UK). Kinematic data were collected at a 250 Hz frame rate by tracking movement of subjects instrumented with reflective markers based on a standardized gait analysis marker set (Plug-In-Gait marker set; Vicon Motion Systems; Oxford, UK). Kinetic data were measured using two AMTI OR6 series force platform systems (AMTI; Watertown, MA) at a rate of 1000 Hz. Kinematic and kinetic data were filtered using a fourth order, low-pass, zero-phaseshift Butterworth filter at 6 and 20 Hz for trajectory and analog data, respectively.

Postural Control Assessments

The reactive postural control task utilized a tether-release model, which forced the subject to incorporate a protective step to regain stability (Appendix C). The tether-release protocol has been used previously to investigate balance recovery from a forward perturbation[115–117] and has been utilized and validated for posterior balance recovery from a posterior perturbation.[14] The protocol for this study consisted of securing one end of a tether to a trunk harness at the level of the xiphisternal joint. The other end of the tether was connected to a force sensor and electromagnet that was fixed to the wall. Participants were asked to lean against the tether, the length of which was adjusted to provide an initial lean between 9–12% of their body mass. This value has been shown to exceed sway-based recovery abilities.[14] Once the subject was in position for the trial, they were given the following instruction: “When the tether is released try to recover your balance with a single step.” Release of the tether was randomized between 1–20 seconds from the time they were in position to limit anticipation of the release time. Five consecutive trials were performed in each of the backward and forward falling directions, with data analysis being performed on three successful trials for each direction. Successful trials were those where the individual was able to recover from the lean-induced fall independently or without assistance from the overhead harness and the joint markers were visible throughout the trial. Reactive postural control outcome measures included: step length normalized by height, step length velocity, peak COP displacement toward stance limb, center of pressure-center of mass (COP-COM) difference, reaction time, and joint angular displacements. The *peak COP displacement* represents the force necessary for transferring the load from the stepping limb to the supporting limb during the fall for the maintenance of postural stability. The *COP-COM difference* examined the difference between the COP position at its peak displacement in the mediolateral direction and the COM at the concomitant timepoint. The COP is considered to be a significant controller of body kinematics.[71] The COM is derived from the mass of all body segments. When considered together, the COP-COM construct provides a unique ability to capture the interplay between postural dyscontrol and dynamic stability. A large COP-COM difference is indicative of robust postural control. With changes in body position during the fall task, the distance between COP and COM increases. Because the individuals in this study were required to stay upright for successful completion of the task, effective postural control was required to limit excessive straying of the COM outside of the functional base. Meanwhile, a small COP-COM difference represents a conservative approach to postural tasks, in that the performer does not feel stable enough to allow separation of the COP and COM. More detailed operational definitions for the outcome measures in this study can be seen in Appendix D, along with explanations of how each of the variables was calculated.

Reactive postural control outcome measures were captured continuously throughout the tether-release task, but due to the bipedal nature of the task the following nomenclature was developed to articulate a more clear distinction between anatomical limbs and task phases. The *swing phase* of the postural task refers to the time between when the heel of the stepping foot leaves the force platform to the point at which the same foot strikes the second force platform upon landing. The *support phase* represents the point from when the stepping foot strikes the second force platform upon landing until the individual's center of mass stops moving in the direction of the fall. Diagrams for these phases can be seen in Appendix D.

Fatigue Protocol

The principal muscle groups fatigued in this study were those required for the control of center of mass stability in posturally dynamic positions, namely the quadriceps and hip extensors. Lower extremity resistance exercise was performed on a motorized, isokinetic ergometer (EccentronTM, BTE Technologies, Inc., Hanover, MD) that appears like a normal seated ergometer (Appendix B). Participants resisted a motorized foot pedal that moved toward them at a self-selected pace between 20–40 rpm and experienced eccentric muscle contractions about the knee and hip extensors. Eccentric muscle contractions are capable of producing 2–3 times greater force than can be produced either isometrically or concentrically.[73, 74] In addition, because eccentric exercise requires a much lower energetic cost and has reduced cardiovascular activation compared to traditional concentric resistance exercise,[75–78] it minimizes the effects of cardiovascular contributions to fatigue while maximizing the effects of local muscular fatigue. Fatigue was determined by real-time biofeedback of a 30% drop in individual participants' maximal voluntary contraction (MVC), which has been shown to induce a deterioration in postural control following localized muscle fatigue.[79] This was accomplished in the present study by asking participants to "resist the pedals as hard as you can for 10 seconds" prior to beginning the fatiguing bout of exercise. Biofeedback was provided on a computer monitor with the average of four maximal effort pedal strokes being represented by a horizontal line. An additional line indicated a 30% decline in their baseline peak torque. Subjects were required to perform the exercise until they dropped below the 30% level for four consecutive pedal strokes. An image of the computer screen participants viewed during exercise can be seen in Appendix B.

Procedure

Postural control assessments were performed before (T0) and immediately after (T1) muscle fatiguing exercise, as well as after 15-minutes (T15) and 30-minutes (T30) after exercise (Figure 3). Reactive postural control was assessed using the tether-release model, as described previously. Prior to participants entering the Motion Capture Laboratory, the motion capture cameras and force plates were calibrated. After calibration, each participant was asked to wear form-fitting clothing and instrumented with light reflective markers over bony prominences using a modified plug in gait marker set (Plug-In-Gait marker set; Vicon Motion Systems; Oxford, UK). In addition, for all tether-release trials, a fall restraint tether was attached from a trunk harness to a ceiling support to prevent any unsuccessful recoveries from the postural control tests. Prior to baseline testing participants were exposed to 1–3 trial sessions of the tether-release test in order to become familiar with the testing procedure and to overcome the fear of falling. After performing the baseline (T0) assessments, markers on the posterior aspect of the trunk and pelvis were removed, and masking tape was applied to the outer garment to allow for the seated fatiguing exercise to be performed without the threat of losing markers and to ensure accurate re-application of the markers after exercise. Immediately (<2 mins) after exercise the markers were re-applied and the immediate post (T1) postural control assessments were performed in a randomized order.

Statistical Analyses

Separate paired *t*-tests were used to determine within group differences in pre-post fatigue outcome measures. In order to examine between group differences, change scores were calculated on pre-post differences for all outcome measures. Due to small sample sizes and unequally sized groups, a nonparametric Kruskal–Wallis ANOVA on the change scores was performed. Significant main effects were further examined using Mann–Whitney U post hoc pair-wise comparisons. Effects were considered statistically significant when $p \leq 0.05$. The classification of effect sizes for between-group comparisons on change scores were calculated using partial eta-squared. The above analyses were performed using Statistical Package for Social Sciences (SPSS) version 21.0. Effect sizes for within group differences were calculated in Microsoft Excel (version 14.3.7) using Cohen's d. Statistical power for within group differences was estimated using G*Power[81] (version 3.1.3) and effect sizes that were calculated from paired *t*-test means and standard deviations, $\alpha = 0.05$ and n = sample sizes.

A subsequent analysis was performed in order to assess the potential for the commitment of type I errors in the preceding statistical methodology. Separate 3x4 (group x time) ANOVAs with repeated measures on the time factor were used to determine between-group, within-group, and interaction effects for all postural control variables of the tether-release tests. If the assumption of sphericity was failed in specific variables, a Greenhouse–Geiser correction was used. Post hoc analyses were performed using the Bonferroni correction to identify differences in the time factor (T0, T1, T15, T30). Between-group differences were analyzed with Tukey's HSD post hoc comparisons or the Games–Howell assessment when equal variances were not assumed. Effect sizes were calculated using partial eta-squared. The above statistical analyses were performed using SPSS (version 21.0). Post hoc statistical power was estimated using G*Power[81] (version 3.1.3).

Results

Table 5 describes the anthropometric characteristics of each group. Statistical between-group differences were found for age in PD versus HY ($p = 0.02$), PD versus HO ($p < 0.001$), and HO versus HY groups ($p < 0.001$). The groups did not differ on height, weight, or BMI factors.

Descriptive characteristics of total work performed and total exercise time for each group can be seen in Table 6. These data demonstrate that HY and HO cohorts generally produced more force than the participants in the PD group.

Less Conservative Analyses

Anterior Tether-Release

Within-Group Analysis

The anterior tether-release testing demonstrated significant alterations in neurologically healthy individuals, including five kinematic outcome measures and one spatial end-point (Table 11). Stepping limb angular displacements of the knee and ankle during the support phase were increased 37% ($p = 0.02$) and 23% ($p = 0.008$) after fatigue in HO individuals. Likewise, fatigue resulted in a number of changes in HY individuals during the support phase of the fall, including a 19% increase in stepping hip angular displacement ($p = 0.048$), a 20% increase in stepping knee angular displacement ($p = 0.03$), and a 26% increase in stepping ankle angular displacement ($p = 0.016$). In the singular spatial outcome measure to reach statistical significance, a 19% increase in COP-COM difference during the swing phase of the tether-release was noted in the HO group following muscle fatigue ($p = 0.014$).

No statistically significant changes were seen in the PD group.

Between-Group Analysis

No between group differences were noted for any of the change scores. Effect sizes for all dependent variables were small (≤ 0.12).

Posterior Tether-Release

Within-Group Analysis

Three outcome measures were altered by acute muscle fatigue in the HO group (Table 12). COP-COM difference during the swing phase of the tether-release was increased 17% ($p = 0.036$) from pre-to-post fatigue. Likewise, a 16% increase ($p = 0.049$) was seen in knee angular displacement of the stepping limb during the support phase and an 11% increase ($p = 0.035$) was seen during the swing phase of the same limb after fatigue.

No statistically significant changes occurred in the HY or PD groups. Effect sizes for all dependent variables in the HY group ranged from 0.01–0.61 with an average effect size of 0.21. Accordingly, statistical power was low with an effective yield of 0.14 in this group.

Between-Group Analysis

No between group differences were noted for any of the change scores, and effect sizes for all dependent variables were small ($\leq$0.10).

Recovery from fatigue. Three out of eight reactive postural control measures (38%) that were altered by acute muscle fatigue returned to baseline within the 30-minute post-fatigue window. Each of the three outcome measures were endpoints of stepping limb kinematic control during the support phase of the anterior fall (tether-release) task (Table 13). Ankle angular displacement returned to baseline within the first 15-minute rest period (T15) in HO adults ($p = 0.01$). Knee ($p = 0.005$) and ankle ($p = 0.004$) angular displacements also returned to baseline within the first 15-minute rest period (T15) in HY adults (Figure 8).

More Conservative Analyses

No statistically significant interaction effects were found. Statistically significant main effects of group were found for several outcomes of the anterior tether release, including COP-COM displacement in the swing phase ($F = 4.95$, $p = 0.016$, $\eta_p^2 = .301$), normalized step length ($F = 6.53$, $p = 0.016$, $\eta_p^2 = .362$), step length velocity ($F = 8.98$, $p = 0.001$, $\eta_p^2 = .439$), and knee angular displacement of the stepping limb during the swing phase ($F = 18.95$, $p = 0.00$, $\eta_p^2 = .622$). Tukey's HSD post hoc comparisons revealed that

HY persons allowed a larger separation of the COP-COM during the swing phase than persons with PD (Mdiff = 5.6 cm, $p = 0.02$). Step length and step length velocity were larger in the HY group compared the PD cohort (Mdiff = 5.6 cm, $p = 0.007$, Mdiff = 0.34 m/s, $p = 0.009$). HO persons also used a faster recovery step than PD participants (Mdiff = 0.29 m/s, $p = 0.018$). Finally, knee angular displacement of the stepping limb in the swing phase was larger for the HY group than the PD cohort (Mdiff = 28.9 deg, $p = 0.000$) and the HO group (Mdiff = 16.3 deg, $p = 0.002$). HO persons also demonstrated larger knee angular displacement with the stepping limb in the swing phase than the individuals with PD (Mdiff = 12.5 deg, $p = 0.045$).

Significant main effects of group were also seen in the posterior tether-release task, including normalized step length ($F = 20.77$, $p = 0.00$, $\eta_p^2 = .654$), step length velocity ($F = 24.62$, $p = 0.00$, $\eta_p^2 = .691$), COP-COM displacement during the swing ($F = 6.38$, $p = 0.006$, $\eta_p^2 = .367$), and support phases ($F = 10.13$, $p = 0.001$, $\eta_p^2 = .480$), and stepping limb angular displacement of the hip ($F = 23.38$, $p = 0.00$, $\eta_p^2 = .680$) and knee ($F = 14.66$, $p = 0.00$, $\eta_p^2 = .571$) during the swing phase of the fall. Post hoc comparisons revealed that PD persons used shorter and slower recovery steps than HY (Mdiff = 13.6 cm, $p = 0.00$; Mdiff = .73 m/s, $p = 0.00$) and HO individuals (Mdiff = 9.9 cm, $p = 0.005$; Mdiff = .62 m/s, $p = 0.001$). HO individuals also produced shorter steps relative to HY persons (Mdiff = 3.6 cm, $p = 0.041$). Hip angular displacement of the stepping limb during the swing phase was larger in HY than PD (Mdiff = 19.2 deg, $p = 0.00$) and HO groups (Mdiff = 10.4 deg, $p = 0.004$). The HO group also used a larger hip angular displacement with the stepping limb during the swing phase than the PD group (Mdiff = 8.7 deg, $p = 0.014$). Finally, COP-COM displacement during the swing and support phases of the posterior tether-release was smaller in PD versus HY persons (Mdiff = 6.0 cm, $p = 0.036$; Mdiff = 11.5 cm, $p = 0.003$) and HO individuals (Mdiff = 5.9 cm, $p = 0.040$; Mdiff = 11.7 cm, $p = 0.002$).

Several main effects of time were found for the anterior tether-release test, including COP-COM difference in the swing phase ($F = 10.50$, $p = 0.00$, $\eta_p^2 = .314$), knee angular displacement of the stepping limb during the support phase ($F = 7.38$, $p = 0.00$, $\eta_p^2 = .243$), and ankle angular displacement of the stepping limb during the support phase ($F = 5.99$, $p = 0.001$, $\eta_p^2 = .207$) (Table 14). Pairwise comparisons revealed that COP-COM difference continued to increase across time for all 3 groups (T0 vs. T1, $p = 0.046$; T0 vs. T15, $p = 0.013$; T0 vs. T30, $p = 0.000$). Knee angular displacement of the stepping limb during the support phase was increased in T1 compared to T0 ($p = 0.007$) but returned to baseline in T15 (T1 vs. T15, $p = 0.004$). Likewise, ankle angular displacement of the stepping limb during the support phase also returned to baseline by T15 (T1 vs. T15, $p = 0.005$).

A statistically significant main effect of time was seen for the posterior tether-release in COP-COM difference during the swing phase ($F = 7.245$, $p = 0.002$, $\eta_p^2 = .248$) (Table 14). In a similar manner as the anterior tether-release test, pairwise comparisons revealed that COP-COM difference continued to increase across time (T0 vs. T15, $p = 0.012$; T1 vs. T15, $p = 0.003$; T0 vs. T30, $p = 0.003$).

Discussion

The purpose of this study was to quantify the acute effects of muscle fatigue on reactive postural control in persons with Parkinson’s disease and neurologically healthy young and older adults. The central hypothesis of this investigation was that acute muscle fatigue would cause greater alterations to reactive postural control measures in persons with PD than in neurologically healthy populations because of the inherently smaller postural stability margins for persons with PD.[69] The results did not reveal any significant alterations to reactive postural control in persons with PD as a result of muscle fatigue. However, acute muscle fatigue did lead to significant postural control deficits in the neurologically healthy cohorts. In general, the results indicate that acute muscle fatigue has deleterious effects on reactive postural control in healthy young and older individuals and that some of these effects are alleviated with rest.

Between-Group Effects (Less Conservative)

The lack of statistically significant differences between groups in this study is likely due to an issue of power and sample size. For a sample size of nine persons in this study, statistical power ranged from 0.05–0.76, with a mean of 0.12 across all outcome measures of the anterior tether-release test. The outcome measures of the posterior tether-release in this population demonstrated statistical power ranging from 0.05–0.68, with a mean of 0.18. The ability to reject the null hypothesis would be stronger with a larger sample size and an effectively greater degree of statistical power. From a research design standpoint, too few subjects may have been recruited because the sample sizes were estimated based on work done in healthy young and older persons.[45] Given the prevalence of hypokinesia and bradykinesia characteristic of persons with PD, this study may have been underpowered in its ability to detect minute biomechanic changes in this population from the beginning.

Within-Group Effects (Less Conservative)

Parkinson's Disease

Contrary to our hypothesis, we found no significant changes induced by muscle fatigue in reactive postural control endpoints in persons with PD. Similar to the between-group effects, one reason for this may be due to a lack of statistical power, which is related to the small effect sizes in this analysis. Effect sizes for all dependent variables in the PD group were small, ranging from 0.02–0.62 and an average effect size of 0.28. Consequently, statistical power was equally low in the PD cohort (0.19).

Neurologically Healthy Individuals

The most common changes noted in neurologically healthy cohorts following acute muscle fatigue occurred at the kinematic level during the support (landing) phase of the lean-induced falls. During this phase, increases were seen in joint excursions throughout the lower extremity kinematic chain in the stepping leg in both HO and HY cohorts (Figure 9). These findings are in agreement with work by Mademli et al., who reported increased knee flexion angles in healthy young and older individuals during an anterior fall following acute muscle fatigue of the quadriceps femoris muscles.[43] Previous research has demonstrated that muscle fatigue induced by repetitive contractions causes a reduction in the force generating capacity of the muscle,[3] making it more difficult to maintain body weight through the support limb during a fall.

Another explanation for the increased knee angular displacement could be due to the reduction of functional reflex activity (FRA) after acute muscle fatigue, which has been shown to contribute to the degree of joint laxity and joint stability.[118] Granacher et al. demonstrated that ankle fatigue diminished FRA in the tibialis anterior muscle of young and elderly men, resulting in reduced joint stability during reactive gait perturbations, as measured by increased maximal angular velocity of the ankle joint complex.[47] Interestingly, the researchers also reported increased co-activation of agonist and antagonist muscles surrounding the ankle joint complex following fatigue and proposed that this was an effort to enhance joint stiffness and stability to compensate for fatigue. These findings suggest that the stability of ankle, knee, and hip joints of the stepping limb during the support phase of a fall is altered by acute muscle fatigue and may be a significant contributor to the increased risk of falls in HO persons seen at the end of the day[41] or during slipping or tripping situations.[38]

These results also indicate that the COP-COM difference is increased during the swing phase in HO individuals during an externally induced fall after fatiguing exercise (Figure 10). The COP-COM construct in this particular task is best understood in the context of the two phases of the task. During the swing phase, the individual is leaning to a degree that would cause the COM to be positioned outside the functional base of support. During this phase a large COP-COM difference is expected. During the support (landing) phase, however, the individual is attempting to recover from the horizontal acceleration of COM without falling. The COP-COM difference offers the greatest utility during this phase because it provides a meaningful indication of postural control recovery. In this context, a large COP-COM difference during the support phase indicates a robust degree of postural control because the definition of a successful trial in this study required that the individual recover upright stance. Contrariwise, a small COP-COM difference during the support phase would indicate a conservative approach to controlling the COM within the base of support and may provide an indicator of falls in at-risk populations.

Chang and Krebs reported a peak COP-COM difference during gait initiation of 21 cm for HO adults and about 16 cm for older adults with disability.[119] These data are in line with what has been seen in this report, with the HO subjects averaging 19.2 cm displacements during the swing phase of the anterior tether-release test. Interestingly, during the support phase the HO subjects reduced the COP-COM difference to a 17.6 cm average displacement. Indeed, with increasing age and disability the spatial parameters of step initiation are thought to become smaller and more variable, leading to decreased separation between the COP and COM.[90] This conservative approach to COP-COM control may tacitly serve to keep the COM within the functional base of support during recovery from externally induced falls as well, but it indicates an overall lack of robust control of dynamic stability.

Recovery from Fatigue

The results of this discussion suggest that rest may not ameliorate alterations to reactive postural control induced by acute muscle fatigue within 30 minutes. Regardless of age, task, or the presence of neurologic disease, the majority (63%) of postural control variables altered by acute muscle fatigue in this study failed to return to baseline.

These data contradict previously published reports examining the recovery of postural control after acute fatigue in neurologically healthy adults. Independent of age, studies examining general whole body fatigue have been shown to decrease postural control on average 14.6 minutes before returning to baseline.[96–100] The aggregate of studies examining localized muscle fatigue have shown that postural control returns to baseline across all age groups on average within 8.2 minutes.[31, 37, 49, 101, 102] This study however, could not replicate completely these results. A small percentage (38%) of postural control outcome measures altered by fatigue returned to baseline using a 30-minute recovery window. One reason for this prolonged fatigue effect may be due to the method of localized muscle fatigue employed.

Each of the aforementioned studies examining the effects of localized muscle fatigue on postural control utilized combinations of concentric and concentric/eccentric muscle contractions. The innovation in this design is that participants utilized a form of high force eccentric resistance exercise as a means of inducing skeletal muscle fatigue. Eccentric muscle contractions are capable of producing 2–3 times greater force than can be produced either isometrically or concentrically.[73, 74] Consequently, this intervention provided extremely high loads to the muscle in the shortest amount of time. Because eccentric exercise requires a much lower energetic cost and has reduced cardiovascular activation compared to traditional concentric resistance exercise,[75–78] it minimized the effects of cardiovascular causes to muscle fatigue while maximizing the effects of local muscular and neurologic contributions to fatigue. These heightened fatigue effects induced by eccentric muscle contractions may have been the cause of the prolonged recovery window for postural control measures to return to baseline.

Despite the predominance of a lack of recovery of fatigue effects, there were 3 outcome measures that did return to baseline following rest, and deserved to be addressed here. It is noteworthy that each of the measures that were ameliorated by rest were outcomes of stepping limb kinematic control during the support (landing) phase. The enhanced recovery of fatigue in this area may be related to the use of muscle synergies used during reactive stepping responses.

Results of reactive postural control research suggest that the CNS combines independent, though related muscles into synergies (e.g., ankle and stepping strategies),[120, 121] which may partially circumvent the acutely fatigued quadriceps and hip extensor muscles used for control of posture. In this study, the return to baseline seen in the outcomes of support phase kinematics may have been ameliorated by the selection of muscle synergies, allowing for a more efficient recovery period and improved performance following acute bouts of fatiguing exercise.

Mixed-Design Effects (More Conservative)

The results of the more conservative statistical test corroborated the effect of fatigue on lower-extremity kinematics seen in the less conservative analyses. Specifically, increases in knee and ankle angular displacement were found in the immediate postfatigue test of the anterior fall. It appears that, regardless of group assignment, individuals in this study had a more difficult time maintaining their body weight through the stepping limb in the support phase of a fall following fatigue. This may be due to the reduction in force generating capacity of the muscle following fatigue induced repetitive contractions.[3] This effect is limited, however, because both knee and ankle angular displacements returned to baseline after 15 minutes of rest.

Several between-group differences were found, most commonly between HY persons and the older cohorts. Of interest, however, were several group differences between HO individuals and persons with PD. HO individuals stepped further and faster than persons with PD and used a kinematic strategy incorporating larger hip and knee angular displacements than those with neurologic impairment during the externally induced falls. These data are supported by previous reports of stiffening in persons with PD in response to externally-induced slipping/tripping scenarios.[122, 123] This stiffening response may be caused by changes in passive elastic properties[123] or increases in co-activation of antagonist muscles during postural responses.[12] The results of this between-group comparison should be interpreted cautiously, however, because the HO and PD cohorts in this study were not age-matched.

No statistically significant interaction effects of group and time were found in this analysis. The average statistical power for the anterior and posterior tether-release tests was 0.10 in the interaction analysis. This is slightly larger than the statistical power of the interaction effects in the anticipatory tests, which is likely due to decreased movement variability inherent in the repeatable tether-release task. However, the very low effect sizes in this study made it difficult to detect statistically significant changes. These effect sizes were a product of large pooled standard deviations, which could have been caused by differing levels of fatigue within and between groups. In addition, it is possible that the lack of control for medication timing in the PD group and the differing levels of fatigue across groups had an effect on the variability of individual performances.

Conclusions

The purpose of this investigation was to characterize the previously unexplored effects of acute muscle fatigue on reactive postural control in persons with Parkinson's disease and to compare those effects to neurologically healthy adults. The combined data suggest that acute muscle fatigue has a deleterious effect on lower-extremity joint kinematics of the stepping limb during the support (landing) phase of a fall, regardless of age or the presence of neurologic disease. However, the within-group effect of acute muscle fatigue on reactive postural control in persons with PD remains unclear. Further research is needed with larger sample sizes and greater controls for threats to internal validity (level of muscle fatigue, control for medication status, age matching). A conservative interpretation of these findings suggests that muscle fatigue may alter lower-extremity joint kinematics following a fall, though this effect is brief. A more liberal view suggests that acute muscle fatigue may increase the risk for externally induced falls in neurologically healthy individuals. These results should also serve to heighten the awareness of clinicians regarding the potential negative effects of muscle fatigue for at-risk populations during clinical exercise settings.

Table 11. Means ± Standard Deviations and Effect Sizes for Outcome Measures of the Anterior Tether-Release Test, Organized by Biomechanical Category (Less Conservative Analysis)

Anterior Tether-Release Dependent measure	Young			Older			PD		
	PRE	**POST**	**ES**	**PRE**	**POST**	**ES**	**PRE**	**POST**	**ES**
SPATIAL									
Step Length (normalized) (m)	.309 ± .03	.309 ± .03	.02	.278 ± .02	.286 ± .03	.40	.247 ± .05	.251 ± .05	.07
COP/COM difference (swing phase) (m)	-.193 ± .05	-.205 ± .04	.24	-.175 ± .03	-.209 ± .05*	1.14	-.138 ± .04	-.158 ± .04	.57
COP/COM difference (support phase) (m)	.176 ± .04	.166 ± .04	.44	.173 ± .05	.179 ± .02	.15	.130 ± .06	.137 ± .05	.12
TEMPORAL									
Step Length Velocity (normalized) (m/s)	1.20 ± .11	1.17 ± .14	.54	1.15 ± .10	1.14 ± .06	.33	.774 ± .31	.779 ± .30	.04
Reaction Time (s)	.179 ± .03	.192 ± .03	.49	.184 ± .02	.191 ± .03	.29	.193 ± .04	.194 ± .03	.08

Table 11. Continued

KINEMATIC									
Stepping limb, Hip angular displacement (support phase) (deg)	20.3 ± 4.2	24.3 ± 8.1*	.85	18.4 ± 5.4	23.8 ± 9.9	.49	19.4 ± 10.9	19.5 ± 9.5	.01
Stepping limb, Knee angular displacement (support phase) (deg)	31.4 ± 7.0	37.8 ± 8.1*	.96	26.4 ± 8.1	36.2 ± 9.1*	1.07	27.4 ± 9.6	30.8 ± 12.9	.33
Stepping limb, Ankle angular displacement (support phase) (deg)	18.8 ± 5.2	23.7 ± 5.1*	1.11	17.5 ± 5.7	21.5 ± 5.2*	1.30	18.9 ± 8.6	19.9 ± 9.4	.12

PRE prefatigue, *POST* postfatigue, *ES* effect size, COP/*COM difference* delta between center of pressure at its peak and center of mass at the concomitant timepoint, calculated during swing and support phases of the task, *Reaction time* time from tether-release to the point when the heel comes off the force plate, *Stepping limb, angular displacement (support phase)* joint angular displacement of the stepping limb during the support phase of the task.

*Significant main effect of fatigue ($p < 0.05$)

Table 12. Means ± Standard Deviations and Effect Sizes for Outcome Measures of the Posterior Tether-Release Test Organized by Biomechanical Category (Less Conservative Analysis)

Posterior Tether-Release	Young			Older			PD		
Dependent measure	PRE	POST	ES	PRE	POST	ES	PRE	POST	ES
SPATIAL									
Step Length (normalized) (m)	.323 ± .04	.324 ± .05	.02	.299 ± .02	.304 ± .02	.24	.181 ± .06	.194 ± .08	.48
COP/COM difference (swing phase) (m)	-.162 ± .04	-.172 ± .04	.19	-.155 ± .03	-.182 ± .03*	.92	-.116 ± .06	-.115 ± .07	.02
COP/COM difference (support phase) (m)	.207 ± .06	.197 ± .04	.25	.218 ± .04	.200 ± .05	.77	.094 ± .09	.082 ± .11	.18
TEMPORAL									
Step Length Velocity (normalized) (m/s)	1.44 ± .19	1.43 ± .16	.05	1.37 ± .14	1.37 ± .12	.01	.699 ± .42	.665 ± .37	.26
Reaction Time (s)	.213 ± .06	.199 ± .03	.25	.203 ± .04	.197 ± .02	.26	.222 ± .03	.200 ± .04	.62

Table 12. Continued

KINEMATIC									
Stepping limb, Hip angular displacement (support phase) (deg)	25.8 ± 6.8	25.7 ± 6.3	.01	24.4 ± 9.7	29.0 ± 13.2	.30	39.2 ± 22.0	36.1 ± 18.4	.30
Stepping limb, Knee angular displacement (support phase) (deg)	39.1 ± 7.9	44.9 ± 10.7	.61	38.8 ± 9.9	41.9 ± 9.4	.34	55.8 ± 38.6	54.5 ± 34.5	.14
Stepping limb, Ankle angular displacement (support phase) (deg)	29.1 ± 4.0	28.7 ± 3.0	.08	30.4 ± 7.4	31.0 ± 5.7	.17	35.7 ± 23.6	29.7 ± 13.7	.42

PRE prefatigue, *POST* postfatigue, *ES* effect size, *COP/COM difference* delta between center of pressure at its peak and center of mass at the concomitant timepoint, calculated during swing and support phases of the task, *Reaction time* time from tether-release to the point when the heel comes off the force plate, *Stepping limb, angular displacement (support phase)* joint angular displacement of the stepping limb during the support phase of the task.
* Significant main effect of fatigue ($p < 0.05$)

Table 13. Summary of Timeline for Recovery of all Reactive Postural Control Measures Altered by Acute Muscle Fatigue (Less Conservative Analysis)

VARIABLE	GROUP	FATIGUE				P ANOVA
		Pre (T0)	Post1 (T1)	Post2 (T15)	Post3 (T30)	
Anterior Tether-Release						
KNEE_ANG_DISP (deg)	HO	26.4 ± 8.1	*36.2 ± 9.1	28.8 ± 5.6	34.2 ± 14.3	0.07
ANKLE_ANG_DISP (deg)	HO	17.5 ± 5.7	*21.5 ± 5.2	◆*16.3 ± 5.6*	16.4 ± 8.5	0.01
COP-COM_DIFF_SW_PHS (m)	HO	-0.175 ± .03	*-0.208 ± .04	-0.204 ± .04	*-0.217 ± .05	0.03
HIP_ANG_DISP (deg)	HY	20.4 ± 4.2	*24.3 ± 8.1	21.1 ± 6.7	20.6 ± 6.2	0.15
KNEE_ANG_DISP (deg)	HY	31.4 ± 7.0	*37.8 ± 8.1	◆*29.0 ± 5.3*	31.5 ± 5.7	0.005
ANKLE_ANG_DISP (deg)	HY	18.8 ± 5.2	*23.7 ± 5.1	◆*17.6 ± 4.6*	17.5 ± 4.5	0.004
Posterior Tether-Release						
COP-COM_DIFF_SW_PHS (m)	HO	-.155 ± .03	*-.181 ± .03	*-.190 ± .03	*-.201 ± .03	0.001
KNEE_ANG_DISP_SW (deg)	HO	56.7 ± 10.5	*63.2 ± 12.7	58.6 ± 10.4	58.1 ± 11.7	0.02

Values are means ± SD. $p < 0.05$ indicates statistically significant main effect of time. *indicates statistically significant pairwise comparison from baseline. ◆ indicates return to baseline measure, HO Healthy older group, HY Healthy young group, KNEE_ANG_DISP Knee angular displacement of stepping limb during support phase, ANKLE_ANG_DISP Ankle angular displacement of stepping limb during support phase, COP-COM_DIFF_SW_PHS Center of pressure-center of mass difference of the stepping limb during swing phase, HIP_ANG_DISP Hip angular displacement of stepping limb during support phase, KNEE_ANG_DISP_SW Knee angular displacement of stepping limb during swing phase

Table 14. Means ± Standard Deviations for Reactive Postural Control Measures Altered Across All Timepoints by Acute Muscle Fatigue (More Conservative Analysis)

VARIABLE			FATIGUE				P ANOVA		
		GROUP	Pre (T0)	Post1 (T1)	Post2 (T15)	Post3 (T30)	*Group*	*Time*	*Interaction*
Anterior Tether-Release	COP-COM_DIFF_SW_PHS	PD	-0.137 ± .04	-0.157 ± .04	-0.162 ± .04	-0.163 ± .04	**0.016**[a]	**0.000**[d, e, f]	0.423
		HO	-0.175 ± .03	-0.208 ± .04	-0.204 ± .04	-0.217 ± .05			
		HY	-0.184 ± .04	-0.199 ± .04	-0.222 ± .06	-0.242 ± .05			

Table 14. Continued

VARIABLE									
Anterior Tether-Release	KNEE_ANG_DISP (deg)	PD	27.4 ± 9.5	30.7 ± 12.9	23.9 ± 6.9	28.8 ± 9.1	0.346	**0.000**[d, g]	0.587
		HO	26.4 ± 8.1	36.2 ± 9.1	28.8 ± 5.6	34.2 ± 14.3			
		HY	31.4 ± 7.0	37.8 ± 8.1	◆*29.0 ± 5.3*	31.4 ± 5.7			
	ANKLE_ANG_DISP (deg)	PD	18.9 ± 8.5	19.9 ± 9.3	15.2 ± 4.1	16.7 ± 4.2	0.743	**0.001**[g]	0.533
		HO	17.5 ± 5.7	21.5 ± 5.2	◆*16.3 ± 5.6*	16.4 ± 8.5			
		HY	18.8 ± 5.2	23.7 ± 5.1	◆*17.6 ± 4.6*	17.5 ± 4.5			

Table 14. Continued

VARIABLE									
Posterior Tether-Release	COP-COM_DIFF_SW_PHS (m)	PD	-0.115 ± .05	-0.115 ± .06	-0.143 ± .06	-0.117 ± .05	**0.006**[a, b]	**0.002**[e, f, g]	0.190
		HO	-0.155 ± .03	-0.181 ± .03	-0.190 ± .03	-0.201 ± .03			
		HY	-0.164 ± .04	-0.181 ± .03	-0.191 ± .02	-0.196 ± .03			

bold- indicates statistically significant effect, a) Tukey HSD post hoc difference between PD and HY groups, b) Tukey HSD post hoc difference between PD and HO groups c) Tukey HSD post hoc difference between HO and HY groups, d) Bonferroni pairwise comparisons difference between T0 and T1, e) Bonferroni pairwise comparisons difference between T0 and T15, f) Bonferroni pairwise comparisons difference between T0 and 30, g) Bonferroni pairwise comparisons difference between T1 and T15, ◆ indicates return to baseline, COP-COM_DIFF_SW_PHS center of pressure-center of mass difference during the swing phase, KNEE_ANG_DISP angular displacement of the stepping limb knee joint during the support (landing) phase, ANKLE_ANG_DISP angular displacement of the stepping limb ankle joint during the support (landing) phase

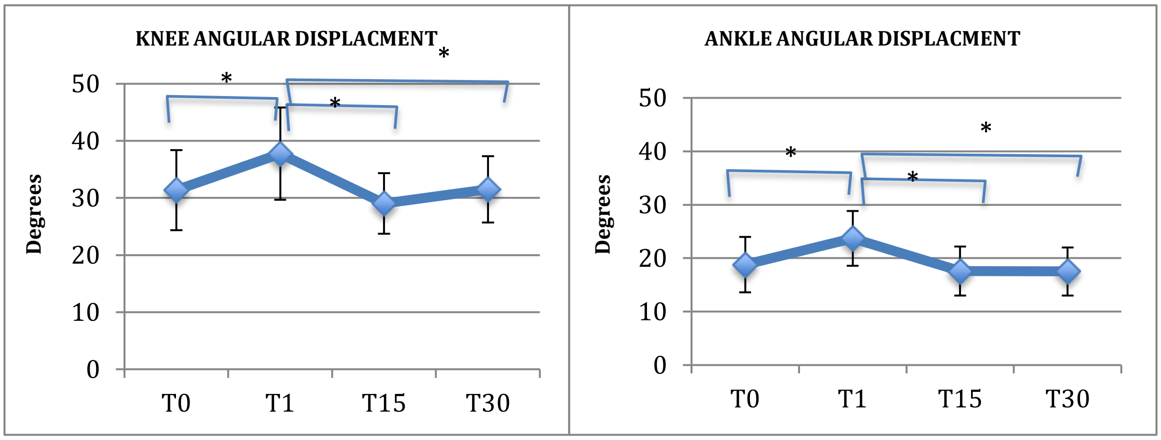


Figure 8. Angular Displacements of the Stepping Limb Knee and Ankle Joints During the Support Phase of the Anterior Tether-Release Test in Healthy Young Individuals (Less Conservative Analysis)

* - statistically significant pairwise comparison ($p < 0.05$)

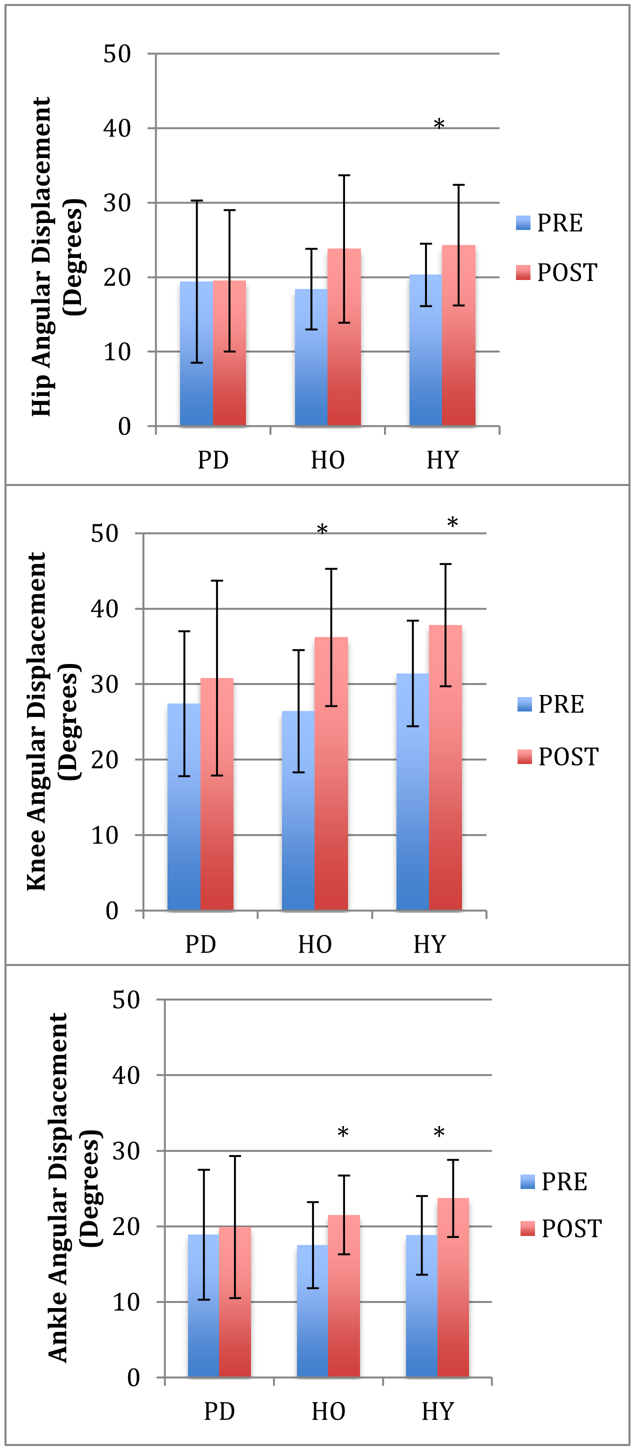


Figure 9. Means and Standard Deviations of Hip, Knee, and Ankle Angular Displacements in the Stepping Limb of the Support (Landing) Phase During Simulated Falls in the Anterior Direction Before and After Muscle Fatigue (Less Conservative Analysis)

PD: Parkinson's disease group; HO: Healthy older group; HY: Healthy young group

* Significant difference ($p < 0.05$)

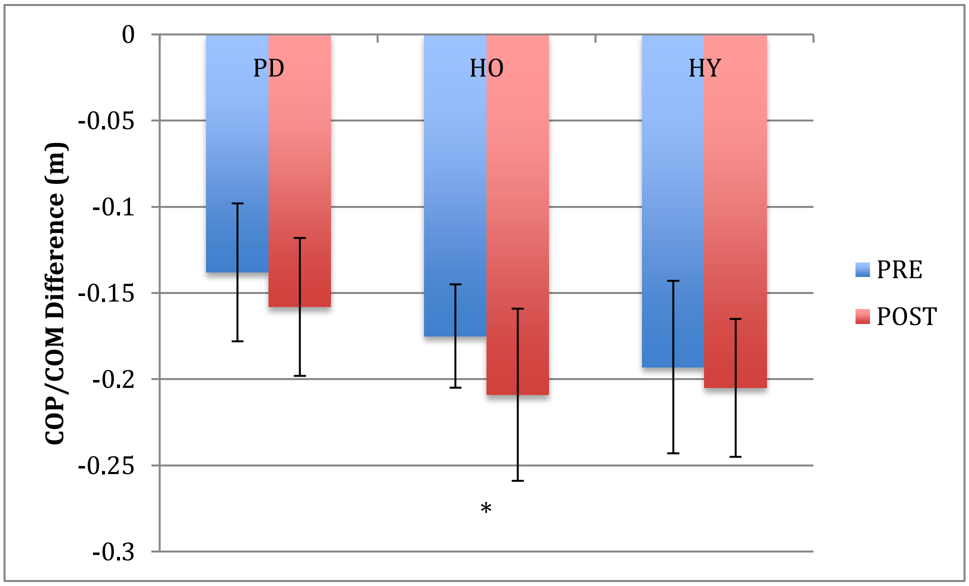


Figure 10. Means and Standard Deviations of Center of Pressure-Center of Mass (COP-COM) Difference in the Swing Phase of the Anterior Tether-Release Task (Less Conservative Analysis)

PD: Parkinson's disease group; *HO:* Healthy older group; *HY:* Healthy young group
COP-*COM difference* is calculated as 'COP-COM' and takes the COM position when COP is at its peak in the mediolateral direction.
* Significant difference ($p < 0.05$)

CHAPTER 5

SUMMARY AND PERSPECTIVES

Clinical Impact

Accidental or environment-related falls are the most frequently cited cause of falling in older individuals, accounting for 30–50% of cases. The second most common cause is postural instability and/or gait problems.[17] When muscle fatigue is added to these inherent fall risks, older individuals become increasingly susceptible to falls.[38, 39, 41] The composite results of this study indicate that fatigue induces postural control deficits in older individuals and potentially in persons with Parkinson's disease, during anticipatory and reactive postural control tasks. These results are important to clinical fall risk examinations, postexercise precautions, and to identify potential targets for therapeutic intervention.

The results of this investigation and previous examinations of postural control in older individuals in fatigued states, coupled with studies reporting the alteration of the effectiveness of sensory inputs and motor output of postural control,[27] strongly suggest that fatigue has a measurable clinical effect on stability and potentially on fall risk. Given that the aging population is becoming increasingly advised to seek strength and mobility training interventions clinically,[124, 125] it becomes critical to have guidance on the time course of postural control recovery for those who seek clinical interventions for strengthening. Composite results of previous studies suggest that recovery of postural control in healthy populations occurs on average within 8.2 minutes following localized muscle fatigue.[31, 37, 49, 101, 102] The results from this study, however, should serve to caution clinicians that recovery in at-risk older adults could last beyond 30 minutes.

Limitations

The lack of a predefined minimal workload for inclusion into the study may have influenced the results of this investigation. Attempts were made to follow the guidelines proposed by Paillard et al[79] of a 30% decline in maximal voluntary contraction (MVC) during exercise to induce the fatigue-altering changes proposed in postural control. However, certain subjects in the PD group were able to minimally exert themselves during their baseline measurement. Perhaps out of PD-related apathy[126, 127] or fear of postexercise discomfort, this strategy enabled them to more quickly obtain the 30% decline in their MVC during the fatiguing session, without providing a clear indication to the researchers that they were acutely fatigued. The use of a predefined minimal level of energy expenditure (calories) or workload equivalence (joules), which all participants would have been required to attain, would have enabled us to be more confident in the achievement of acute muscle fatigue across all participants.

In this investigation, attempts were made to standardize the use of dopamine replacement medication in the inclusion criteria, but regulations regarding the timing of the ingestion of antiparkinsonian treatment were overlooked. Patients with Parkinson's disease, in both early and late stages of the disease, are prone to experiencing fluctuations in their response to levodopa known as "wearing off" or "end of dose" deterioration.[128] This wearing off is dependent on the timing of dopamine replacement ingestion and the pharmacokinetics of the particular drug. The lack of control on the time of testing relative to the time of levodopa medication ingestion could have contributed to the inconsistency and lack of statistical significance seen in the full cohort of PD persons.

Another limitation that affected this study was the differences on age between the PD and HO groups. The fact that the PD group was statistically significantly older than the HO group could diminish any between-group differences on outcomes that were reported in this study.

Suggestions for Future Research

In the future, studies examining the effect of acute muscle fatigue on postural control should establish a minimal threshold of energy expenditure or workload equivalence that all participants must acquire for inclusion into the study, thereby improving the construct validity of muscle fatigue for studies of postural control.

The recovery of postural control could be mediated, in part by the method of fatigue induced exercise. This study employed a novel form of eccentric muscle contractions to alter postural control, and it appears that this form of exercise induced a prolonged recovery window. Future studies in localized muscle fatigue should compare the effects of concentric and eccentric exercise protocols on postural control recovery.

The degradation of postural control by acute muscle fatigue would appear to reveal a potential target for intervention. If exercise programs were explicitly designed to make lower extremity muscles more fatigue resistant, the participant might derive postural control benefits. To date, several chronic muscle endurance-training studies have been employed using a combination of postural control outcomes.[51–55] However, these studies have employed clinical balance correlates like static stance posture, gait speed, the Berg balance test, the Dynamic Gait Index, and others, which fail to incorporate measures of reactive postural control. Although multidimensional fall risk assessment and exercise interventions have shown promise in reducing falls,[56] these interventions are generally composites of neuromuscular reeducation and lower extremity muscle strength and endurance activities. Because of this, the differential benefits of muscle endurance training versus coordination training are unclear. Controlled trials are needed to examine the efficacy of training regimens on muscle fatigue induced instability.

Conclusions

In summary, this investigation has examined the effect of localized muscle fatigue on components of anticipatory and reactive postural control in persons with Parkinson's disease and neurologically healthy adults. The results indicate that there are clear deteriorations in both anticipatory and reactive postural control in healthy young and older populations and potentially in persons with Parkinson's disease, following acute fatiguing exercise of the lower extremities. The results of this study challenge the composite results of previous investigations suggesting that postural control returns to baseline within 8.2 minutes of acute fatiguing exercise. In addition, these results should caution clinicians and leaders of community based exercise settings to be aware of the potential negative effects of acute muscle fatigue in older adults at-risk for falls.

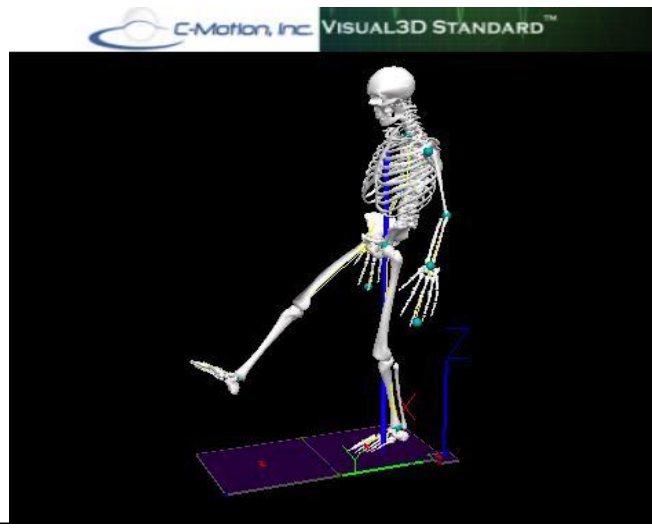
APPENDIX A

OUTCOME MEASURES AND OPERATIONAL DEFINITIONS DEVELOPED FOR THE LOWER-EXTREMITY REACH TEST (LERT)

Table 15 Outcome Measures and Operational Definitions Developed for the Lower-Extremity Reach Test (LERT)

LERT Outcome Measures	
Variable	*Definition*
SPATIAL	
-Reach length (m)	-defined as the maximum distance in the antero-posterior (A/P) direction traveled by the toe marker of the stepping leg. Normalizing reach length was done by dividing the subject's step length by their height.
-Peak anticipatory postural adjustment (APA) (m) -Peak COP displacement (stance) (m)	-defined as the peak displacement of the center of pressure (COP) in the mediolateral (M/L) direction toward the *stepping* limb, starting with the initiation of movement away from zero (bipedal stance). -defined as peak displacement of the center of pressure (COP) in the mediolateral (M/L) direction toward the *stance* limb.
-COP-COM difference (step) (m) -COP-COM difference (stance) (m)	-defined by taking the COM position when COP is at its peak. COM and COP are used in M/L direction. One construct examines the delta toward the *stepping* limb, and the other construct observes the delta toward the *stance* limb, providing an indication of effectiveness of momentum generation (step) and overall dynamic stability (stance) during a postural control task.
-A/P COP variability (%) -A/P foot variability (%) -M/L COP variability (%) -M/L foot variability (%)	-defined using the coefficient of variation* of the COP and the great toe marker of the reaching limb in antero-posterior (A/P) and mediolateral (M/L) directions from zero (bipedal stance) to end of trial.
TEMPORAL	
-Reach velocity (m/s)	-reach length of stepping limb / reach length time (defined from heel off to max reach length)
-Time_COP displacement (stance) (s)	-time it takes to achieve peak COP displacement toward the stance limb.

Table 15 continued

KINEMATIC	
-Joint angular displacements (swing) (deg) -Joint angular displacements (support) (deg)	-joint angular displacements of the hip, knee and ankle for both stepping limb and support limb. Displacements include flexion and extension.
* The CV is the ratio of standard deviation to the mean (SD / mean) x 100. It is a measure of relative variability, expressed as a percentage.	
	

APPENDIX B

METHOD OF INDUCING ACUTE MUSCLE FATIGUE VIA ECCENTRIC ERGOMETRY

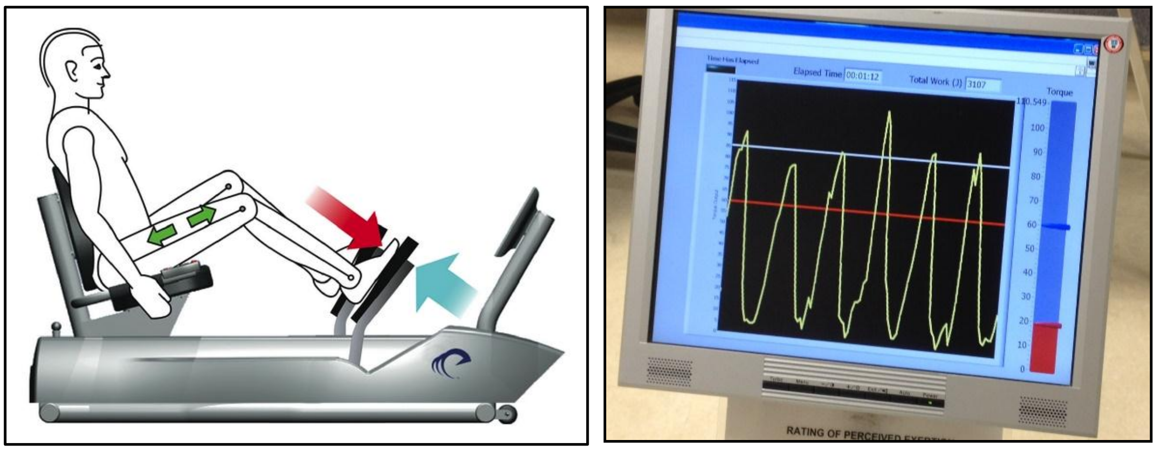


Figure 11 Method of Inducing Acute Muscle Fatigue via Eccentric Ergometry

Image on right depicts the screen that participants viewed while resisting pedals, providing real-time biofeedback of each pedal stroke and an indicator for the investigator of when muscle fatigue was accomplished.

APPENDIX C

TETHER-RELEASE METHOD FOR REACTIVE POSTURAL CONTROL ASSESSMENT

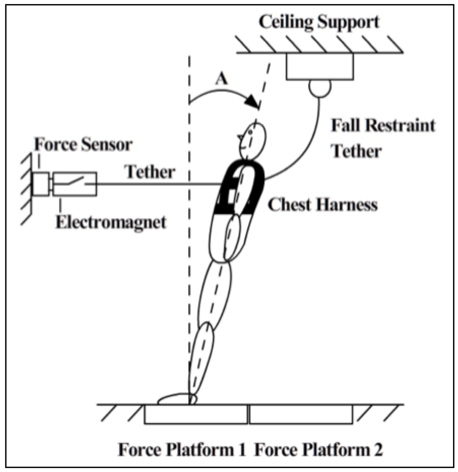


Figure 12 Tether-Release Method for Reactive Postural Control Assessment

"A" represents the lean associated with 12% body mass registered at the force sensor.

APPENDIX D

OUTCOME MEASURES AND OPERATIONAL DEFINITIONS

DEVELOPED FOR THE TETHER-RELEASE TEST

Table 16 Outcome Measures and Operational Definitions Developed for the Tether-Release Test

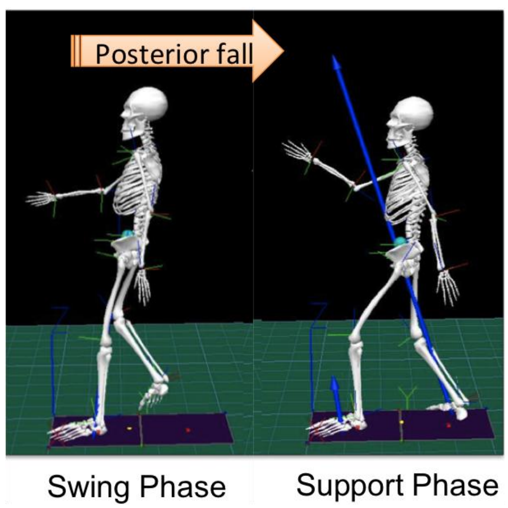
Tether-Release Outcome Measures	
Variable	***Definition***
SPATIAL	
-Step length (normalized) (m)	-distance in the antero-posterior (A/P) direction traveled by the toe marker of the stepping leg. Normalizing step length was done by dividing the subject's step length by their height.
-Peak COP displacement (m)	-peak displacement of the center of pressure (COP) in the mediolateral (M/L) direction. Calculated during swing and support phases of the task.
-COP-COM difference (swing phase) (m) -COP-COM difference (support phase) (m)	-defined by taking the COM position when COP is at its peak. COM and COP are used in M/L direction. One construct examines the delta toward the *stepping* limb, and the other construct observes the delta toward the *stance* limb, providing an indication of effectiveness of momentum generation (swing phase) and overall dynamic stability (support phase) during a postural control task.
TEMPORAL	
-Step length velocity (m/s)	-step length of reaching limb / step length time (defined from heel off to max step length)
-Reaction Time (s)	-time from tether release to when the heel comes off the force plate, as defined as the point when the lateral ankle marker of the stepping leg exceeds .4 m/s in the A/P direction.

Table 16 continued

KINEMATIC	
-Joint angular displacements (deg)	-joint angular displacements of the hip, knee, and ankle of the stepping limb. Displacements include flexion and extension. Calculated for both swing and support phases of the task.

Reactive postural control outcome measures were captured continuously throughout the tether-release test but due to the bipedal nature of the task, we developed the following nomenclature to articulate a more clear distinction between anatomical limbs and task phases.

- The ***swing phase*** refers to the time between when the heel of the stepping foot leaves the force platform to the point at which that same foot strikes the second force platform upon landing.
- The ***support phase*** was defined as the point from when the stepping foot strikes the second force platform upon landing until the individual's center of mass stops moving in the direction of the fall.



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