HABITUATION DURING REPEATED EXPOSURE TO BALANCE RECOVERY FROM A FORWARD LOSS OF BALANCE IN YOUNGER ADULTS

by

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ABSTRACT

The purpose of this study was to begin to develop within session volume parameters for perturbation-based balance training by determining the minimum number of exposures needed for participants to habituate to balance recovery from a substantial perturbation. Two young adult participants were exposed to 15 substantial perturbations induced via release from a static forward lean. All participants were instructed to attempt to recover balance by taking a single rapid step. A scalable anatomical model consisting of 36 degrees-of-freedom and 94 muscle actuators was used to compute kinematics and joint moments from motion capture and force plate data. Margin of stability was calculated at heel strike and maximum knee joint flexion to quantify balance recovery performance. Balance recovery trials were divided in to three blocks (early, middle, and late) with 5 trials in each block and static optimization was used to compute estimates of the mean and peak force generated by key muscle groups during recovery for each block. Participant margin of stability declined rapidly during early balance recovery trials and plateaued between trials 5-8. Hip abductor forces remained relatively consistent across trials while the force produced by the Vastus Femoris group decreased during the mid and late trial blocks. Mean force produced by the Soleus during balance recovery decreased across trial blocks. In contrast, the peak force and impulse generated by the Soleus increased across trial blocks. We interpret these data to reflect an adaptation in balance recovery coordination strategy, which appears to occur sometime between the early and middle trial blocks and stems from young participants desire to minimize the effort associated with successful recovery of balance. These results further suggest participants became habituated to balance recovery after exposure to 5-8 substantial perturbations.

CHAPTER ONE

INTRODUCTION

Falls and Older Adults

It is estimated that 29 million older adults, those age 65 and older, experience a fall each year (Bergen et al., 2016), and once an older adult has experienced a fall their risk of suffering another fall is doubled (O'Loughlin et al., 1993). Three million older adults are treated by emergency departments each year for fall related injuries (Centers for Disease Control and Prevention [CDC], n.d.) with falls being the number one cause of traumatic brain injuries (Jager et al., 2000) and more than 95% of hip fractures are caused by falls (Hayes et al., 1993). In the US the death rate from falls increased by 30% from 2007 to 2016. If this trend were to continue, the estimated fall death rate would reach 7 deaths per hour by 2030 (Bergen et al., 2016). Falls also incur a large financial burden on public health care systems. The medical costs associated with falls are estimated to be more than $50 billion each year and are projected to increase as the proportion of older adults in the population continues to rise (Florence et al., 2018). Not only is the growing fall rate associated with increased medical spending, but also increased resource utilization. Resources such as hospital beds, nursing home spaces and, importantly, the time of health care professionals who are caring for them, all of which must be allocated away from other areas to care for those suffering from fall-related injuries.

Fall Causes and Risk Factors

Falls occur as a result of a loss of balance from which an individual cannot recover. Both younger and older adults experience loss of balance. However, older adults experience falls more often than younger adults as a result of their diminished balance recovery capacity (Arampatzis et al., 2008; Carty et al., 2011; Madigan & Lloyd, 2005; Thelen et al., 2000; Thelen, 2003). The reason for older adults diminished balance recovery is primarily due to a progressive decline in sensory-motor function, which stems from factors such as age-related declines in vestibular function, proprioception, vision, muscle function and reaction time (Christofi et al., 2001; Grabiner et al., 2008; Seidler & Stelmach, 1995). This age-related decline in balance and balance recovery leads to elevated fall risk. While the causes of falls are multifactorial, risk factors are generally divided into external and internal risk factors.

External risk factors are those which are related to the environment such as poor lighting, weather, slippery floors, or poor choice in footwear. Internal factors are those which relate directly to an individual such as poor vision, Parkinson's disease, dementia or polypharmacy (Enderlin et al., 2015). While it is not feasible to reduce most external risk factors or certain internal risk factors such as polypharmacy and Parkinson's disease in a meaningful way, there are some internal risk factors that are relatively modifiable. Balance disorders and poor balance have been reported as one of the leading causes of falls in older adults (Rubenstein et al., 1994). However, balance and specifically poor balance is a risk factor which can be modified. As such, increasing balance performance has been the focus of many recent fall prevention programs such as those outlined in the American College of Sports Medicine (ACSM) recommendations on physical activity for older adults which include many balance related exercises (Chodzko-Zajko et al., 2009).

Fall Prevention Paradigms

Exercise-Based Balance Training

There is strong evidence to suggest physical activity improves general health and wellness and decreases all-cause mortality of older adults (Chodzko-Zajko et al., 2009; Nocon et al., 2008). Exercise-based balance training such as resistance exercise, yoga, tai-chi or static balance exercises including tandem standing, one-legged standing and heel or toe stands are commonly recommended by entities such as the ACSM and HHS to reduce fall risk for older adults (Chodzko-Zajko et al., 2009; US Department of Health and Human Services [HHS], n.d.). These types of exercise-based fall preventions programs have been moderately successful. Lower extremity strengthening has been linked with decreased fall risk and increased balance recovery ability (Karamanidis et al., 2008; Pijnappels et al., 2008) and exercise-based balance training programs generally result in a 17-24% decrease in fall rate (Sherrington et al., 2008).

Perturbation-Based Balance Training

Perturbation-based balance training (PBT) is an emerging fall prevention strategy which exposes participants to gait and postural perturbations in order to cause loss of balance and trigger a compensatory stepping response (Mansfield et al., 2015). Whereas conventional balance training (CBT) is comprised primarily of volitional quasi-static exercises such as tandem stance or heel to toe walking and occurs within the participants region of stability (i.e., they do not actually lose balance), PBT is dynamic and occurs outside the participants region of stability.

Because PBT simulates a real-world fall in a safe and controlled environment, it allows participants to experience a real loss of balance and practice the mechanisms required for dynamic balance recovery in a specific manner. Thus, PBT provides participants with a strong learning stimulus for balance recovery at a notably smaller dose than CBT. The recommended dose for CBT is two hours a week for 6 months or approximately 50 hours total (Tiedemann et al., 2010) in contrast adaptations from PBT interventions have been observed after a single training session (Barrett et al., 2012a; Grabiner et al., 2008; Maki et al., 2008; Yungher et al., 2012). Current literature has demonstrated PBT is an efficacious fall prevention strategy (Gerards et al., 2017a) and participants have retained substantial balance improvements for up to 6 months after a single PBT session (Bhatt et al., 2012).

Development of a Problem

The efficacy of PBT interventions to date (Mansfield et al., 2015; McCrum et al., 2017) has even lead some authors to suggest that PBT has the potential to inoculate older adults from falls (Pai et al., 2010). However the parameters of PBT interventions have been highly variable thus far, in terms of perturbation type, magnitude, volume, and frequency. Due to the high variability, parameters such as perturbation magnitude, volume and training frequency has yet to be established (Gerards et al., 2017b).

Further, the current understanding with regard to the mechanisms of improvement as well as the safety of long-term participation in PBT are also unknown (Gerards et al., 2017b). Although there have been no adverse events reported as a result of PBT thus far, Graham et al. (2016) reported hip joint contact loads during recovery from large perturbations, which were above the range of mechanical failure for cadaveric femurs from older adults as reported by

Schileo et al. (2014). This is suggestive of an inherent risk associated with participation in PBT, especially during repeated exposure to large perturbations, that has yet to be addressed. The development of PBT parameters such as minimum effective dose will inform future research, optimize PBT interventions and mitigate potential risks associated with participation by minimizing participant exposure to perturbations.

Purpose

The purpose of the present study is to begin to develop within session volume parameters for PBT by determining the minimum number of exposures needed for participants to become habituated to balance recovery. This purpose will be achieved by evaluating Margin of Stability during repeated exposure to recovery from substantial tether release induced perturbations.

CHAPTER TWO

REVIEW OF RELATED LITERATURE

The Causes of Falls

The are many reasons older adults experience falls. Risk factors for falls are generally divided into intrinsic or extrinsic risk factors. Extrinsic risk factors are environmental factors such as slippery surfaces, weather conditions, poor lighting, clutter on the floor or poor footwear all of which may cause loss of balance and result in a fall. Intrinsic risk factors are those which are tied directly to the individual and include age-related physiological changes and health problems that affect the systems involved balance performance (Enderlin et al., 2015). Intrinsic risk factors associated with falls in older adults include chronic conditions such as heart disease, diabetes, stroke, arthritis, Parkinson’s disease and dementia (Lord et al., 2004). Intrinsic risk factors can also include pharmacological factors such as the use of antihypertensive medications or polypharmacy (Hill & Wee, 2012) as well as vision-related factors including binocular vision disorders such as strabismus, amblyopia, diplopia and nystagmus (Pineles et al., 2015).

Although some extrinsic factors are modifiable (i.e. choice of footwear), it is not feasible to mitigate many extrinsic factors, such as icy or poorly lit sidewalks, in a meaningful way. While intrinsic factors such as Parkinson’s disease, stroke, dementia, and polypharmacy are not feasibly modifiable, factors such as balance and more specifically the ability to recover balance appears to be a highly modifiable fall risk factor for older adults. Poor balance and dizziness have been reported as the second and third leading causes of falls in community dwelling older adults, respectively (Rubenstein et al., 1994). Further, a 2010 meta-analysis regarding risk factors for falls in community dwelling older adults (74 prospective studies) found that problematic gait and the use of walking aids, both of which are suggestive of poor balance during gait, were predictive of falls and the leading cause of falls in community dwelling older adults (Deandrea et al., 2010).

Balance and Balance Recovery

Balance declines with advancing age due a progressive degradation of sensory-motor function (Christofi et al., 2001; Seidler & Stelmach, 1995). The factors which amount to this decline in sensory-motor function are age-related declines in proprioception, vision, vestibular function, muscle function and reaction time (Grabiner et al., 2008). While both younger and older adults can experience loss of balance, older adults are more likely to experience a fall as a result of loss of balance due to their markedly reduced capacity for balance recovery. The diminished ability of older adults to recover from perturbations of the same magnitude as younger adults has been repeatedly demonstrated in the literature by studies which have investigated compensatory stepping responses to perturbations induced via waist pulls (Pijnappels et al., 2005; Troy et al., 2009), trips during walking (Troy et al., 2009), slips during walking (Yang et al., 2012), and release from static forward lean (Carty et al., 2011)

Despite the diminished balance recovery capacity of older adults, the ability to adapt and improve reactive stepping response does not appear to diminish with age. A meta-analysis conducted by Bohm et al. (2015), which consisted of 18 studies and over 1000 participants, found no significant difference in general locomotor adaptability between younger and older adults. Additionally, investigators reported no significant difference in the predictive or adaptive ability of older adults compared to younger adults. The similarity in adaptive ability of older and younger adults suggests that despite age-related deficits in balance recovery performance, older adults are capable of increasing balance recovery performance via adaptive balance training. For example, Dijkstra et al. (2015) found that older adults were able to improve their compensatory stepping response at similar or in some instances greater rates than younger adults. Similarly, Mansfield et al. (2010) demonstrated improvements in compensatory stepping response after a loss of balance in older adults after a 6-week perturbation-based balance training program. Further, an investigation by McCrum et al. (2016) concluded the ability of older adults to adapt to perturbations via reactive gait adjustments did not appear to be limited to a specific type of motion (i.e., gait, sit-to-stand or stance). Taken in summation, these finding suggest that interventions involving adaptation and learning paradigms which also utilize the mechanisms required of reactive dynamic balance recovery may increase recovery performance during loss of balance and therefore reduce risk of fall.

Current Balance Training Guidelines for Older Adults

Because balance is a modifiable fall risk factor, it is not surprising balance improvement has been the emphasis of many fall prevention programs for older adults. Exercises which challenge balance are generally accepted as central components in exercise-based balance training (BT) programs for older adults (Chodzko-Zajko et al., 2009; HHS, n.d.). The current ACSM recommendations on balance training for older adults (Chodzko-Zajko et al., 2009) include: 1) progressively difficult postures that gradually reduce the base of support (e.g., two-legged stand, semi-tandem stand, tandem stand, one-legged stand); 2) dynamic movements that perturb the center of gravity (e.g., tandem walk, circle turns); 3) stressing postural muscle groups (e.g., heel stands, toe stands); and 4) reducing sensory input (e.g., standing with eyes closed). Examples of recommended BT activities from the HHS include walking backwards or sideways, yoga, tai chi and aqua aerobics classes (HHS, n.d.).

Exercise-based balance training has been shown to be an effective fall prevention strategy for older adults. A systematic review and meta-analysis conducted by Sherrington et al. (2008) included 44 trials with 9603 total participants and estimated a 17% decrease in fall rates after participation in exercise-based balance training. Further, data from 25 studies in the same systematic review which included only conventional balance training with no resistance or aerobic exercise, resulted in a 24% decrease in fall rates (Sherrington et al., 2008).

While conventional and exercise-based balance training have been moderately effective for reducing falls in otherwise healthy older adults, there is conflicting evidence as to its effectiveness for older adults suffering from frailty and limited evidence for older adults with Parkinson's disease or those who are recovering from a stroke. A randomized controlled trial conducted by Faber et al. (2006) investigated the effects of exercise-based interventions on falls in pre-frail and frail older adults. Two hundred and seventy-eight older adults were assigned to a control group or one of two exercise-based interventions, which were based on either functional walking and mobility or on the principles Tai Chi. The interventions lasted for 20 weeks and falls were recorded during a 52 week follow-up period after the intervention. The functional walking group reported a post intervention fall incidence of 3.3 falls/year, which was higher than the Tai Chi group (2.4 falls/year).

However, the control group reported a post intervention fall incidence of 2.5 falls/year, which was lower than the functional walking group and very similar to the Tai Chi based group. None of the results regarding differences in falls by intervention were statistically significant. However, the risk of becoming a faller increased significantly (hazard ratio [HR]=2.95; 95% confidence interval [CI], 1.64−5.32) for participants classified as frail in both exercise-based groups compared to control. A 2015 systematic review of randomized controlled trials regarding exercise-based interventions in frail older adults with 837 total participants (De Labra et al., 2015), reported conflicting evidence for the efficacy of such interventions on fall prevention for frail older adults. As detailed above Faber et al. (2006) found a significant increase in fall risk for frail participants throughout the intervention. Additionally, Fairhall et al. (2014) reported the intervention group had a higher post intervention fall rate (1.54 falls/person) compared to control (1.5 falls/person), although this difference was not statistically significant. Only one study found a decreased fall rate in frail older adults, Cadore et al. (2014), who reported a statistically significant increase in balance performance and decrease in fall incidence in after a 12-week multi-component exercise program.

In addition to the conflicting evidence for exercise-based balance training in frail older adults, evidence for similar interventions in adults with Parkinson disease is also mixed. In a randomized controlled trial on exercise for fall prevention in older adults, Canning et al. (2015) stratified participants based on fall history in the preceding 12 months and randomly assigned them to an exercise-based intervention group or control group. The 6-month intervention consisted of three 40-60-minute sessions per week. Sessions were primarily composed of progressive balance training, lower limb strengthening exercises and gait training. The control group received normal care from their usual medical practitioner. No statistically significant difference was found in falls during the intervention between the exercise (467 falls; 4.1 falls/person) and control groups (810 falls; 7 falls/person) although the 27% lower fall rate reported for participants in the exercise group was likely clinically significant. Interestingly, a subgroup analysis revealed a statistically significant interaction for levels of disease severity on fall rate and proportion of fallers. In participants with lower disease severity a 69% reduction in falls was observed in the exercise group compared to control. However, in participants with higher disease severity in the exercise group a trend towards more falls was observed as the intervention progressed and a significantly higher proportion of fallers (RR = 1.28, 95% CI: 1.01–1.62, $p = 0.04$) was seen.

Similarly conflicting evidence has been found on the effects of exercise-based fall prevention programs in older adults who are recovering from stroke (Verheyden et al., 2013). The limited and inconsistent evidence as to the efficacy of exercise-based balance programs for older adults with frailty and other pre-existing conditions has led some to suggest that exercise-based balance programs may lack specificity necessary to increase their balance recovery performance (Grabiner et al., 2014).

APAs and the Specificity of Balance Recovery

Due to the multisegmented nature of the body, voluntary movement alone will impose some level of perturbation to posture. Anticipatory postural adjustments (APAs) are integrated into voluntary movements to compensate for internal postural perturbations and ensure accurate movement (Gahéry & Massion, 1981). Belen’kiĭ et al. (1967) first reported the presence of APAs by demonstrating that muscles responsible for producing APAs were activated before those responsible for creating the voluntary movement. APAs are triggered before voluntary movement and are both flexible and task specific (Ingimar & Oddsson, n.d.; Layne & Abraham, 1991). Meaning, different voluntary movements may be associated with different postural adjustments depending on the context in which they are performed. For instance, in a voluntary stepping motion an APA which shifts the trunk over the stance limb precedes the motion in order to maintain balance (Maki & McIlroy, 1997). Similarly to internal perturbations, external perturbations to posture such as a slip or trip trigger APAs which are specific to the magnitude, direction and type of perturbation (Diener et al., 1983; Keshner et al., 1988; McIlroy & Maki, 1993). For example, in response to smaller postural perturbations APAs known as "ankle strategies," adjustments made by the musculature surrounding the ankle, are used to regain balance. Or, for recovery from larger postural perturbations a "hip strategy," which consists of APAs involving the musculature surrounding the pelvis, may be used to regain balance. If balance is not recovered by these strategies, a stepping response is initiated to recover from loss of balance and when required a stepping response can be initiated immediately (McIlroy & Maki, 1993). The ability to perform a rapid compensatory stepping response is a critical skill for recovery from a loss of balance, which appears to decline with age (Maki & McIlroy, 2006). This type of recovery strategy is rapid, reactive and involuntary (Mcllroy & Maki, 1995), meaning it is not under volitional control and therefore cannot be refined through voluntary movements. Yet, most exercise-based fall prevention programs are comprised of exercises which are both static or quasi-static and volitional in nature. When considered summatively this suggests that exercise-based balance training may lack a high degree of specificity to the mechanisms used during recovery from a loss of balance.

Perturbation-Based Balance Training

PBT is an emerging fall prevention strategy which exposes participants to gait or postural perturbations in order to cause loss of balance and trigger a reactive stepping response (Mansfield et al., 2015). Whereas exercise-based balance training is comprised predominantly of static or quasi-static exercises which are volitional in nature and therefore allows participants to remain within their limits of stability (i.e., they do not typically experience actual loss of balance). In contrast, PBT is both dynamic and reactive in nature and provides participants with a much stronger challenge to balance. By allowing participants to experience loss of balance in a controlled environment, PBT simulates a real-world fall which allows participants to practice and improve the mechanisms required for dynamic balance recovery in a highly specific manner, in terms of speed, range of motion, stability and type of movement.

Pai & Bhatt (2007) suggested that improvements in stability resulting from PBT may be due to a newly acquired predictive form of adaptive control which emerges following repeated exposure to perturbations. This theory was supported by Scheidt et al. (2001) who demonstrated the emergence of feed-forward behaviors in response to repeated exposure to external perturbations. Subjects exhibited feed-forward behavior by responding to external perturbations in a predefined manner which improved performance by adapting current motor commands and relying on stored information from previous experiences. It has been postulated that the newly formed level of adaptive control improves balance recovery performance in two ways. Firstly, that adaptations from PBT create APAs which alter limb and trunk position prior to exposure to hazardous situations which improve stability prior to the onset of a perturbation and reduce the need for a compensatory stepping response following a perturbation (Pai et al., 2003). Secondly, that PBT can lead to increased post-perturbation stability during a compensatory stepping response and therefore increase balance recovery performance, which in conjunction with pre-perturbation APAs has been highly correlated with a reduction in loss of balance (Bhatt et al., 2006). These findings suggest the high degree of specificity to real world falls provided by PBT generates a strong stimulus for the development of the sensorimotor and sensory information processing capabilities necessary for participants to adapt and improve their proactive and reactive control of balance.

Due to the strength of the stimulus provided by PBT adaptations in balance recovery and long term retention of adaptations from PBT interventions have been observed after a single training session (Barrett et al., 2012a; Grabiner et al., 2008; Maki et al., 2008; Yungher et al., 2012). In contrast, the minimum recommended dose for exercise-based balance training is about 2 hours a week for 6 months, or approximately 50 hours (Tiedemann et al., 2010).

Though results have been demonstrated from single sessions of PBT, there are several studies involving larger doses of PBT which also bear mention. In a study conducted by Bhatt et al. (2012), 48 community dwelling older adults participated in a PBT intervention. 100% of participants experienced backwards loss of balance from the first perturbation and 44% experienced a fall. After the initial perturbation, 23 additional perturbations were induced within a 90-minute training session. After the single training session, the incidence of falls from backwards loss of balance was reduced to 0%. In a similar study, Grabiner et al. (2012) randomly assigned 52 community dwelling elderly women to either a PBT intervention or a control group. During post-intervention assessment of balance recovery, 27% of women in the control group experienced a fall compared to 4.5% of women in the PBT group. Further, Bhatt et al. (2012) demonstrated substantial retention of balance improvements 6 months after a single PBT session, and that the addition of a single "booster" perturbation at 3 months post intervention led to better balance at 6 months compared to a single session conventional balance training group. These results led authors to speculate that PBT could be a way to "inoculate" older adults of falls (Pai et al., 2010) with a considerably lower dose than required for conventional balance training. Further, a prospective study with over 200 community-dwelling older adults (Carty et al., 2015), demonstrated that balance recovery strategy after a forward loss of balance strongly predicted falls in the following year. The same study also reported the use of a single step as opposed to multi-step, balance recovery strategy from equivalent forward lean angles, as well as the maximal lean angle from which balance could be recovered with a single step, were strong predictors of falls while traditional measures of balance were not. A 2013 investigation with over 450 community dwelling older adults reported performance in a choice reaction time stepping task (stepping onto one of four panels which were illuminated in random order) was a strong and independent predictor of future falls (Sturnieks et al., 2013). Considered collectively, these findings provide evidence that by allowing participants to practice the compensatory stepping responses used during balance recovery in a controlled environment, PBT can improve older adults balance recovery capacity and is an effective strategy for reducing real-world falls.

The tether release method is a common approach for inducing loss of balance in PBT studies (Hsiao-Wecksler, 2008). During a tether release induced perturbation, the participant is tilted forward into a static forward lean by a horizontal tether. After a random time interval has elapsed the tether is released and the participant takes one or more rapid steps to recover their balance (Do et al., 1982). A commonly used assessment of maximal balance recovery is Maximum Recoverable Lean Angle (MRLA) which is the maximum lean angle from which a subject can successfully recover balance in a single step from a tether release induced perturbation (Thelen et al., 1997). Recently Carty et al., (2015) demonstrated that balance recovery performance from anterior perturbations via the tether release method at MRLA was significantly and independently predictive of future falls. Additionally, Margin of Stability (MoS) during recovery at MRLA is often used as a primary outcome measure in balance recovery studies (Barrett et al., 2012b).

As described above there is considerable evidence to suggest that PBT is an efficacious fall prevention strategy. Thus far there have been no reported adverse events as a result of PBT in healthy, high fall risk older adults in the literature. However, there is some evidence to suggest an inherent risk associated with PBT. Bergmann et al. (1993), reported hip joint contact loads as high as 8.7 bodyweights during balance recovery from an unexpected trip experienced during walking in patients with an instrumented hip replacement. Additionally a study by Graham et al. (2016), computed hip joint contact loads in older adults during compensatory step recoveries from a forward loss of balance, reported peak hip joint contact loads of between 4.3 to 12.7 bodyweights. The hip joint contact loads reported by Bergmann et al., and Graham et al., are within the mechanical failure range of the femur of 5.5-14.5 bodyweights reported by Schileo et al. (2014), who tested the maximal load bearing capacity of femurs from older adults in positions similar to the stance phase of gait. Despite the lack of adverse outcomes from PBT thus far, these findings indicate there may be some level of inherent injury risk for older adults participating in PBT programs. Determining training parameters such as minimum effective dose will help mitigate the potential risks associated with PBT outlined above by minimizing participants exposure to large perturbations.

Measuring Stability and Balance Recovery Performance

Margin of Stability (MoS) is frequently used to quantify stability and performance during balance recovery tasks (Gerards et al., 2017b). MoS is quantified as the distance between the extrapolated center of mass (XCoM) and the anterior boarder of the Base of Support (BoS) and defined by the equation:

$$MoS = |\mu_{max} - XcoM|$$

Where BoS is the area beneath the foot contacting the ground. XCoM is the velocity adjusted position of the CoM and defined by the equation:

$$XCoM = x + \frac{v}{\omega_o}$$

where x is the vertical projection of the CoM, v is the velocity of the CoM and ω_o is equal to acceleration of gravity divided by the height of the whole-body CoM (Hof et al., 2005a). Prior to the advent of MoS, stability was typically assessed by whether CoM was within the BoS at any given instant. However, Pai & Patton (1997) demonstrated that during dynamic tasks the velocity of the CoM must be taken into account to more accurately assess stability. This is because during dynamic tasks there are instances in which the CoM is outside the BoS, but if the velocity of the

CoM is directed inward towards the BoS balance can still be achieved. Conversely, there are situations where the CoM may be inside the BoS but if its velocity is directed outwards balance may be unachievable (Hof et al., 2005a). By adjusting for the velocity of the CoM, MoS more accurately quantifies stability during dynamic tasks such as balance recovery.

Musculoskeletal Modeling

Musculoskeletal modeling is the process of representing a recorded movement of interest using a virtual musculoskeletal model. Musculoskeletal models are constructed to represent the dynamics of a physical system and are generally comprised of rigid body segments, joints and the forces which act upon those bodies and joints to re-produce the motion of interest. The applications of musculoskeletal modeling are vast. In terms of clinical applications musculoskeletal models have been used to better understand normal (Anderson & Pandy, 2003) and pathological movement patterns (Piazza & Delp, 1996), predict outcomes of surgeries such as joint replacements (Delp et al., 1994, 1996; Piazza & Delp, 2001) and gain insight into the factors contributing to risk for injuries such as ACL tears (Ali et al., 2014; Bulat et al., 2019; McLean et al., 2004). Musculoskeletal modeling has also been used to provide insight in fields such as ergonomics and workplace biomechanics (Chaffin, 1987; Chang & Wang, 2007; Schult et al., 2013), sport performance (Herzog, 2009; Preatoni et al., 2012; Schwameder, 2008) and even in paleontology (Allen et al., 2013; Gignac & Erickson, 2017).

Musculoskeletal modeling typically involves placing reflective markers on the subject's bony landmarks, the trajectory of which are then recorded during the movement of interest. The recorded trajectories are used to create the segments and joints of the model and recreate the motion of interest and calculate kinematic or kinetic (when combined with measured ground reaction forces) variables of interest. Several software programs are available to perform musculoskeletal modeling, one of which is OpenSim (Delp et al., 2007).

An OpenSim musculoskeletal model consist of components which correspond to parts of that physical system including bodies, joints, forces, constraints, and controllers. Rigid body segments are the primary structural components of the model and each model has a set of rigid body segments and segment is defined by a set of virtual markers. A joint set is also needed to define all joints in the model. Joints specify the relationship between each of the rigid bodies. Constraints can be placed on joints to prevent any non-physiologic movement of the model during simulation. Forces, which are defined in the force set section of the model, specify the forces that will be applied to the model during simulation. In general there are two types of forces: passive forces such as springs and dampeners and active forces such as springs and idealized linear or torque actuators. Active forces require a set of controls, called actuators, and must be defined by the user (Hicks, 2012).

Muscle actuators are used to represent muscles in the musculoskeletal system. Because the force-producing properties of muscle are complex and non-linear, lumped-parameter, dimensionless muscle models which are capable of representing a large range of muscle architectures are often used during musculoskeletal modeling (Zajac, 1989). Muscle models include a set of muscle points which define where the muscle attaches to bone (represented by the bodies) and are used to calculate the lengths and velocities of the muscle actuator. Parameters for muscle activation and contraction dynamics as well as muscle states are also commonly included in muscle models and their control values are typically bounded excitations that range from 0 to 1 and lead to changes in activation and therefore force. One such muscle model is the

Thelen 2003 Muscle Model (Thelen, 2003). The Thelen model is based on the Hill-Type Musculotendon Model (Hill, 1938) and consists of three components: a contractile component, parallel component, and a series component. In the Thelen muscle model, muscle force is generated as a function of the activation value, the normalized length of the muscle unit and the normalized velocity of the muscle unit. Musculotendinous parameters such as maximum isometric force, optimal muscle fiber length, tendon slack length, maximum contraction velocity, and pennation angle of the muscle model can be specified by the user based on subject specific measures or normative values.

Once a generic model has been constructed it can be scaled to the anthropometrics of the subject using relative differences in the three-dimensional position of reflective markers placed on subject's bony landmarks prior to motion capture. Segments of the model are scaled so the distances between the virtual markers which define each segment match the distances of the corresponding experimental markers. During scaling a cost function is used to minimize the total error between virtual markers and corresponding experimental markers. To account for error in marker placement from factors such as skin movement artifact, a weighting system can be applied to the markers during scaling. The weighting system penalizes errors on virtual markers with a higher weighting factor in the cost function. As a result a marker with a greater weighting factor will have lower residual error and be placed closer to its experimental marker during scaling, whereas a marker with a lower weighting factor may have a larger residual and be placed relatively farther from its corresponding experimental marker (Delp et al., 2007; Lu & O'Connor, 1999).

Once the model has been scaled, the inverse kinematics (IK) tool is used to generate generalized coordinates (joint angles and translations) which best recreate the raw marker trajectories from recorded during motion caption. Generalized marker coordinates are calculated by solving an IK problem. The IK problem solves the equation:

$$Squared\ Error = \sum_{i=1}^{markers} w_i\left(x_i^{\to subject} - x_i^{\to model}\right)^2 + \sum_{i=1}^{joint\ angles} w_j\left(\theta_j^{\to subject} - \theta_j^{\to model}\right)^2$$

which is a least-squares cost function that minimizes the weighted squared error between the virtual marker locations and the experimental marker locations for each frame of the experimental kinematics. Where $x_i^{\to subject}$ and $x_i^{\to model}$ are the three-dimensional positions of the *i*th marker or joint center for the subject and model, $\theta_j^{\to subject}$ and $\theta_j^{\to model}$ are the values of the *j*th joint angle for the subject and model, and w_i and w_j are factors that allow markers and joint angles to be weighted differently. Similarly to scaling, a weighting system is also to be applied to the cost function during IK to account for marker error and tune joint coordinates within the joint constraints (Lu & O'Connor, 1999).

Due to experimental error and assumptions made during modeling there are often dynamical inconsistencies between measured ground reaction forces and joint moments and simulated model kinematics. The Reduced Residual Algorithm (RRA) is used calculate generalized joint coordinates which are more dynamically consistent with measured kinematics. During RRA, residuals are calculated and averaged over the duration of the motion. Calculated averages are then used by the algorithm to recommend changes in model mass parameters, such as the position of trunk CoM which in turn lower the average residual values for the simulation. After model mass parameters have been adjusted, a control problem is solved where all joints in the model are actuated by idealized joint moments. Additionally three residual forces and three residual moments are applied to a user specified segment to control the six degrees of freedom between the model and the ground. RRA can often track experimental kinematics with little or no error when no constraints are placed on residual values. However, users often place limits on the magnitude of the residuals which can result in a new dynamically consistent set of altered kinematics with reduced residual values. The altered kinematics computed by RRA are optimized by the cost function:

$$Squared\ Error = \sum_{j=1}^{joints} \Omega_j \cdot (\ddot{q}_j^{desired} - \ddot{q}_j^{model})^2$$

which minimizes the squared error between model accelerations and desired accelerations by distributing errors across the joint angles. Where Ω_j is a factor weighting the relative importance of the *j*th joint, and $\ddot{q}_j^{desired}$ is the desired acceleration of the *j*th degree of freedom (Delp et al., 2007; Thelen et al., 2003).

The altered kinematics computed by the RRA tool can be used to solve for individual muscle forces at each instant in time using the Static Optimization Tool. The Static Optimization Tool is an extension of inverse dynamics (ID). ID uses the known values for the models generalized positions, velocities, and accelerations to solve an adapted version of the classical equations of motion for the unknown generalized forces. The generalized forces computed during ID are net forces and moments acting at each joint in the model. The Static Optimization Tool distributes the net forces and moments acting at a joint across contributing muscles using the equations of motion constrained to muscle activation to force conditions of from either ideal force generators:

$$(\sum_{m=1}^{n}(a_m F_m^0) r_{m,j} = \tau_j)$$

or force-length-velocity properties:

$$(\sum_{m=1}^{n}[a_m f(F_{m,}^0\ l_{m,}\ v_{m,})] r_{m,j} = \tau_j)$$

while minimizing a cost function which is typically the sum of the activations squared:

$$(J = \sum_{m=1}^{n}\ (a_m)^P)$$

Where n is the number of muscles in the model; a_m is the activation level of muscle m at a discrete time step; F_m^0 is its maximum isometric force; l_m is its length; v_m is its shortening velocity; $(F_{m,}^0\ l_{m,}\ v_{m,})$ is its force-length-velocity surface; $r_{m,j}$ is its moment arm about the jth joint axis; τj is the generalized force acting about the jth joint axis; and p is a user defined constant. When computing individual muscle force the Static optimization Tool assumes that the tendon is inextensible and does not include contribution from the parallel elastic element of the muscle (Hicks, 2012). It should be noted that there are many sets of muscle forces which can recreate the joint forces and that moments from ID and the muscle forces computed via static optimization are simply one set of possible forces which minimized the cost function the most out of all possible sets of forces.

Summary

Falls can substantially diminish quality of life for older adults and incur a large burden to public health care systems both in terms of financial costs and resource utilization. The cost of falls is projected to grow as the proportion of older adults in the population rises. The reasons older adults experience falls are multifactorial, many of which cannot be feasibly mitigated. However, balance and specifically the ability to recover from a loss of balance can is a risk factor which can be improved via training. Exercise-based balance training has been moderately effective at reducing falls and fall risk in older adults but may lack specificity to the neuromotor mechanisms utilized for balance recovery during a real-world fall. Perturbation-based balance training is an alternative paradigm for improving balance and the ability to recover from a loss of balance. Due to its dynamic non-volitional nature, PBT provides a high level of specificity and therefore a more powerful stimulus for improvement of balance recovery performance. The efficacy of PBT in terms of fall reduction has been demonstrated repeatedly in the literature. Due to the varied nature of PBT programs thus far parameters regarding perturbation intensity, volume, frequency, and minimum effective dose have yet to be established and the risks associated with long term participation are not yet known. Further, there is evidence to suggest that the forces experienced during high intensity PBT are potentially injurious. Establishing volume parameters such as minimum effective dose will help mitigate potential risks associated with PBT by minimizing the number of perturbations experienced and improve the efficacy of PBT by informing future balance recovery research and PBT programs.

CHAPTER THREE

CHANGES IN MARGIN OF STABILITY DURING REPEATED EXPOSURE TO FORWARD LOSS OF BALANCE IN YOUNG ADULTS

Contribution of Authors and Co-Authors

Manuscript in Chapter 3

Author: Justin M. M. Whitten

Contributions: Primary investigator and developer of study methodology, primary processor and analyzer of data, and author of manuscript.

Co-Authors: Dawn S. Tarabochia

Contributions: Dr. Tarabochia reviewed the study design, results, analysis, and conclusions and recommended changes; and reviewed the study proposal and final manuscript and recommended changes.

Co-Authors: John G. Siefert

Contributions: Dr. Siefert reviewed the study design, results, analysis, and conclusions; and the study proposal and final manuscript.

Co-Authors: David F. Graham

Contributions: Dr. Graham assisted with development of the study methodology and recommended changes, assisted with data analysis and development of conclusions; and reviewed the study proposal and final manuscript.

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Abstract

The purpose of this study was to begin to develop within session volume parameters for perturbation-based balance training by determining the minimum number of exposures needed for participants to habituate to balance recovery from a substantial perturbation. Two young adult participants were exposed to 15 substantial perturbations induced via release from a static forward lean. All participants were instructed to attempt to recover balance by taking a single rapid step. A scalable anatomical model consisting of 36 degrees-of-freedom and 94 muscle actuators was used to compute kinematics and joint moments from motion capture and force plate data. Margin of stability was calculated at heel strike and maximum knee joint flexion to quantify balance recovery performance. Participant margin of stability declined rapidly during early balance recovery trials and plateaued between trials 5-8. Step length and step execution time demonstrated by participants during balance recovery as well as the forward lean angle from which they were able to recover were exceptional relative to previously reported values for young adults which is suggestive of high levels of inherent stability. We suggest the observed decline in margin of stability was a result of young adults documented desire to minimize the effort associated with a successfully recovery of balance. We further suggest that these results are indicative of participants habituation to balance recovery after exposure to 5-8 substantial perturbations, although further work is needed to verify these results.

Introduction

Falls represent a major public health crisis. An estimated 2.9 million older adults experience falls each year; they are the leading cause of injury and death in older adults in the United States and incur a cost of more than $50 billion on the US health care system each year (Bergen et al., 2016; Florence et al., 2018). Falls occur as a result of a loss of balance. While younger and older adults alike experience loss of balance, older adults experience falls more frequently due to their diminished capacity to recover from a loss of balance (Arampatzis et al., 2008; Carty et al., 2011; Madigan & Lloyd, 2005; Thelen et al., 2000; Thelen, 2003). Older adults reduced balance recovery capacity, in part, stems from their inability to produce an adequate recovery step response. In contrast to younger adults, recovery steps generated by older adults are typically shorter in length and slower in time (Carty et al., 2011; Karamanidis et al., 2008) which in conjunction with their limited trunk control (Bieryla et al., 2007; Crenshaw et al., 2012) results in reduced stability during recovery and subsequently an elevated fall risk.

The benefits of physical activity in terms of health and wellness for older adults has long been understood (Chodzko-Zajko et al., 2009; Nocon et al., 2008). Further, lower extremity muscle and tendon strengthening has been associated with decreased fall risk and increased balance recovery ability (Karamanidis et al., 2008; Pijnappels et al., 2008) and exercise-based balance training programs have resulted in a 17-24% reduction in fall rate (Sherrington et al., 2008). Recently a new paradigm for improving older adults balance recovery deficits has emerged. This paradigm is Perturbation-based balance training (PBT). PBT is an emerging fall prevention strategy which exposes participants to gait or postural perturbations to cause loss of balance and trigger a reactive stepping response (Mansfield et al., 2015). PBT has been very efficacious in fall prevention and fall risk reduction and has resulted in improved balance recovery capacity after a single session (Barrett et al., 2012a; Grabiner et al., 2008; Maki et al., 2008; Yungher et al., 2012). These results are not surprising considering the importance of compensatory step response balance recovery has been supported by several prospective studies which demonstrated the capacity to recover balance in a single step compared to multiple steps strongly predicted future fall incidence for older adults (Carty et al., 2015; Hilliard et al., 2008; Mille et al., 2013).

To date PBT interventions have used many methods of inducing perturbations including: release from a static forward lean (Carty et al., 2011; Karamanidis & Arampatzis, 2007; Thelen et al., 1997), slips induced with movable floor platforms translations (Bhatt et al., 2012; Pai et al., 2010; Parijat & Lockhart, 2012) and rotations (Gage et al., 2008; Keshner et al., 1988), ground surface compliance changes (Bierbaum et al., 2010, 2011), treadmill accelerations and decelerations (Grabiner et al., 2012; Lee et al., 2018; Lurie et al., 2013; Rieger et al., 2020; Sakai et al., 2008), waist pulls (Fujimoto et al., 2015; Martelli et al., 2016; Pai et al., 1998) and weight drop cable pulls (Mansfield & Maki, 2009).

Due to the varied nature of perturbation-based balance training paradigms thus far current understanding with regard to the mechanisms of balance recovery improvement as well as parameters such as perturbation volume, type, magnitude and the long term consequences of participation are relatively unknown (McCrum et al., 2017). Therefore the purpose of the present study is to improve the understanding around volume parameters for PBT. We will accomplish this by evaluating margin of stability during repeated exposure to balance recovery from large perturbations induced via tether release from a static forward lean.

Methods

Participants

Participants were two college-aged adults (age: 22.5 ± 1.5 years, height: 1.71 ± 0.03m, mass: 70.5 ± 1.5kg) recruited from the area. All participants had normal vision. Individuals that reported neurological, cognitive, metabolic, or musculoskeletal impairment were excluded. Ethics approval was obtained from the Institutional Review Board at Montana State University, Bozeman.

Experimental procedures

Perturbations were induced using the tether release method, which is a commonly used method for induing perturbations (Carty et al., 2011; Karamanidis & Arampatzis, 2007; Thelen et al., 1997). During a tether-release induced perturbation, participants are leaned forward to a static angle by a tether attached to a harness at waist level, the tether is released after an unknown time interval, the participant then attempts to recover their balance in a single rapid step. The furthest angle at which a participant can lean forward and successfully recover their balance in a single step is known as their maximum recoverable lean angle and is often used as a criterion to evaluate balance recovery capacity.

Participants wore a harness with a safety rope attached to the ceiling to prevent them from falling in the event of a failed balance recovery. Participants stood barefoot with their feet shoulder width apart and were then leaned forward by an inextensible cable attached to safety harness at waist level on one end and a safety support frame, which was bolted to the lab floor, located directly behind the participant on the other. Once the desired lean angle was achieved participants were instructed to maintain a rigid posture with their heels fully contacting the floor, ankles, knees, hips and trunk in alignment and shoulder blades pulled back. Once the prescribed posture was achieved participants remained static and the cable was released after a random interval of between 2-10s. Center of pressure locations was displayed in real time and visually inspected prior to release to ensure anticipatory actions such as antero-posterior or medio-lateral weight shifts were not present prior to cable release.

Participants performed 2 familiarization trials of balance recovery from a minimal forward lean angle. Following familiarization participants underwent 15 balance recovery trials from a substantial lean angle. The substantial lean angle was achieved by progressively leaning participants forward until they could no longer maintain heel contact with the ground. Trajectories of 51 reflective markers attached to participants bony landmarks and grounds reaction forces under the foot were recorded simultaneously using a 10-camera motion capture system (Motion Analysis Corp.) recording at 220 Hz and three force plates (Advanced Mechanical Technologies, Inc.) recording 100 Hz. The first plate was located beneath the participants right foot in the initial forward lean position and the second was located beneath the left foot. The third plate was anterior to the first plate in order to record the ground reaction forces associated with the stepping foot.

Specific events during balance recovery were defined as follows. Cable Release (CR) was identified as an increase in force on the two plates beneath the participant's feet after a static forward lean had been achieved. Heel Strike (HS) was identified from a force recorded in excess of 20 N recorded on the anterior force plate. Knee Joint Maximum (KJ_{max}) was identified as the point, after heel strike, at which the greatest knee flexion angle was calculated by inverse kinematics for the stepping leg.

<u>Data analysis</u>

Data analyses were performed using OpenSim (version 4.0) (Delp et al., 2007) in conjunction with custom MATLAB scripts (Version 2019a, The Math Works, USA). A scalable anatomical model with 17 bodies, 17 joints, 94 muscle actuators and 36 degrees of freedom (pelvis: 6, neck: 3, Lumbar joints: 3, hip: 3, shoulder joints: 3, wrist: 2, elbow: 1, radioulnar: 1, knee: 1, and ankle: 1), as described by Hamner et al., (2010), was used as the generic model for analysis. A wrap object was embedded in the generic model to prevent non-physiologic moment arms for the erector spinae muscle during trunk flexion (Daggfeldt & Thorstensson, 2003; Graham et al., 2014).

Model Scaling and Inverse Kinematics (Lu & O'Connor, 1999) were performed by fitting the anatomical model to measured 3D marker positions with a high weighting on virtual markers which defined the joint center of the hip, knee and ankle. Joint centers were estimated from experimental marker trajectories: the regression equations of (Harrington et al., 2007) were used for the hip joint, while the knee and ankle joint centers were identified as the midpoints of the femoral condyles and the medial and lateral malleoli respectively.

To quantify performance during balance recovery, margin of stability was calculated at Heel Strike and Knee Joint Maximum for each trial (Arampatzis et al., 2008; Carty et al., 2011), as described by Hof et al., (2005). The anterior border of the base of support was determined by the position of the first metatarsal marker of the stepping foot, in the antero-posterior direction. Center of mass orientation, position and velocity were recorded using a Body Kinematics Analysis in OpenSim (Delp et al., 2007). For analysis tether release trials were grouped into early (trials 1-5), middle (trials 6-10) and late trial periods (trials 11-15).

Results

Participants used their right limb for the first step for all trials. Mean step time was 0.18s and was consistent across all trial periods. Mean step length was 0.92 m across all trials and longest during early trial periods (0.96 m). Forward lean angle across all trials was 36˚ on average. A full summary of temporal-spatial variables can be seen in Table 1.

Table 1. Margin of Stability, values reported as mean ± standard deviation

Margin of Stability (cm)	Early (1-5)	Middle (6-10)	Late (11-15)	All Trials
Heel Strike	17.4 ± 7	16.5 ± 2.3	13.2 ± 4.6	15.7 ± 5.4
Knee Joint Maximum	33.63 ± 12.1	32.6 ± 2.7	26.6 ± 8.2	30.9 ± 9.1

Margin of Stability was greater at Knee Joint Maximum than Heel Strike. At Heel Strike, Margin of Stability was highest during the first trial and declined in trials 2 and 3 and then remained between 10 and 20cm for the rest of the trials (figure 1). Similarly to at Heel Strike, Margin of Stability at Knee Joint Maximum decreased during earlier trials although the decline appeared to be more gradual.

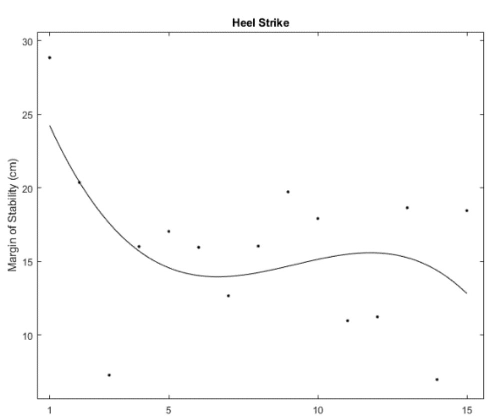

Figure 1. Margin of Stability at Heel Strike

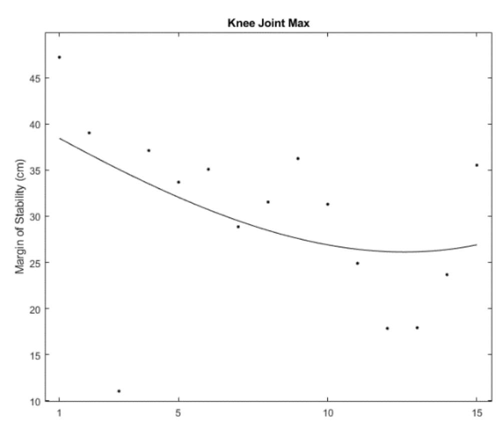

Figure 2. Margin of Stability at Knee Joint Max

Margin of Stability at Heel Strike and Knee Joint Maximum was largest during early trial periods and declined during the mid and late trial periods. Table 2 provides mean and standard deviations for margin of stability during all trial periods.

Table 2. Temporal-Spatial variables, values reported as mean ± standard deviation

Trial period	Early (1-5)	Middle (6-10)	Late (11-15)	All Trials
Step length (m)	0.96 ± 0.9	0.88 ± 0.7	0.92 ± 0.05	0.92 ± 0.08
Step time (s)	0.18 ± 0.02	0.18 ± 0.02	0.18 ± 0.01	0.18 ± 0.02
Lean angle (deg.)	35.8 ± 2	36 ± 2	36.1 ± 2	36 ± 2

Discussion

Results from this study indicate that MoS would increase during early trials and plateau during mid and late trials. Interestingly, average MoS was largest during the early trial period and declined during the mid and late trial periods at both Heel Strike and Knee Joint Maximum. It was particularly notable that MoS was highest during the first trial for Heel Strike and Knee Joint Maximum and subsequently declined sharply during the second and third trials. At Heel Strike, this decline appears to be more rapid than at Knee Joint Maximum. After its rapid decline during the early trial period, MoS at Heel Strike appears to level off and remain within 10 – 20 cm for the mid and late trials with the exception of trial 14. In contrast, MoS at Knee Joint Maximum appears to decline more gradually during the early trial period and continues to decline during the middle trial periods and only begins to plateau during the late trial periods. Initially these results seem to suggest that participants' balance recovery performance decreased as balance trials progressed. However, an alternative interpretation which will be explored in this discussion is that participants were inherently very stable as indicated by a large MoS in trials 1 and 2 and uncommon temporal-spatial movement characteristics, which are described in below.

Further, the decline in margin of stability across trials may reflect the younger participants desire to minimize the effort associated with balance recovery while still achieving a successful recovery each trial.

Interestingly, the lean angle participants were able to recover from (Carty et al., 2011; Graham et al., 2014; Hsiao-Wecksler & Robinovitch, 2007), as well as their step length (Carty et al., 2011; Graham et al., 2014; Rogers et al., 2001), were greater than previously reported values for younger adults during balance recovery. Additionally, step time was considerably shorter than previously reported times for younger adults (Carty et al., 2011; Graham et al., 2014; Hsiao-Wecksler & Robinovitch, 2007; Hsiao & Robinovitch, 1999; Rogers et al., 2001; Thelen et al., 2000). Participants were leaned forward as far as possible while maintaining heel contact with the floor, yet they were stable (MoS > 0cm) during balance recovery for all trials. When considered with the magnitude of the temporal-spatial variables outlined above relative to previous studies, this could suggest that participants were inherently very stable and further that their inherent stability exceeded the demands of the balance recovery task. Although step length and lean angle in this study were greater than commonly reported values for younger adults, our findings are consistent with previous literature in the sense that many studies have reported an increase step length as a result of increase perturbation magnitude (Do et al., 1982; Hsiao & Robinovitch, 1999; Maki et al., 1996; Thelen et al., 1997). Therefore it stands to reason that lean angles greater than those commonly reported would result in step lengths greater than those commonly reported.

In a study similar to the present investigation, Hsiao & Robinovitch (1999) modeled balance recovery by stepping to assess the relationship between effort, which they defined as leg contact force, and step length and step time. It was demonstrated that temporal-spatial characteristics of young participants during recovery matched those predicted by a model which minimized the leg contact force. Specifically, young participants utilize longer step lengths, which increase the stepping legs moment arm and therefore the restorative torque it is able to produce, and shorter step times which reduce the time the trunk is rotating forward and therefore the force that must be absorbed by the leg to recover balance. Authors concluded this represented young participants attempt to minimize the effort associated with balance recovery.

Hsiao & Robinovitch's conclusion comports with our interpretation of the results of this study. The observed decline in MoS as exposure to balance recovery increased may reflect young participants desire to minimize the effort associated with balance recovery. Therefore, we suggest a secondary analysis which would investigate participants effort across balance recovery trial periods to provide more clarity to our current results. While Hsiao & Robinovitch (1999) elegantly used peak leg contact force as a proxy for effort, we believe conducting muscle level analysis similar to that of Graham et al. (2014) who estimated the force generated by key muscle groups and their contribution to recovery, may offer more specific insights to our results.

Limitations

Limitations to our study include our sample of two participants which could have led to higher variability in our results. An additional limitation is the tether-release rig, which may not have been suitable for inherently stable younger adults to achieve maximum recoverable lean angle. As mentioned above, participants were leaned as far forward as possible while maintaining heel contact with the lab floor. Yet participants were stable and successful during all recovery trials, which indicates that they may have been too stable to be suitably challenged by the maximal perturbation our tether release rig was able to incur, meaning their maximum recoverable lean angle may be greater than the angle to which they were able to lean for this study.

Conclusion

In summation, MoS was largest during the early trial period and declined during the middle and late trial periods. Participants step length and lean angle were larger and step time was faster than typically reported in the literature for younger participants. These unusual temporal-spatial measures suggest participants had high levels of inherent stability and the decrease in MoS as trial periods progress may have been a result of young participants desire to minimize the effort associated with balance recovery as outlined by Hsiao & Robinovitch (1999). This would further suggest that although Margin of Stability declined, habituation occurred somewhere between trials 5-8 as indicated by the subsequent plateau in Margin of Stability. Further analysis should be conducted, however, to support these interpretations.

CHAPTER FOUR

CHANGES IN MUSCULAR EFFORT DURING REPEATED EXPOSURE TO FORWARD LOSS OF BALANCE IN YOUNG ADULTS

Contribution of Authors and Co-Authors

Manuscript in Chapter 4

Author: Justin M. M. Whitten

Contributions: Primary investigator and developer of study methodology, primary processor and analyzer of data, and author of manuscript.

Co-Authors: Dawn S. Tarabochia

Contributions: Dr. Tarabochia reviewed the study design, results, analysis, and conclusions and recommended changes; and reviewed the study proposal and final manuscript and recommended changes.

Co-Authors: John G. Siefert

Contributions: Dr. Siefert reviewed the study design, results, analysis, and conclusions; and the study proposal and final manuscript.

Co-Authors: David F. Graham

Contributions: Dr. Graham assisted with development of the study methodology and recommended changes, assisted with data analysis and development of conclusions; and reviewed the study proposal and final manuscript.

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Abstract

The purpose of this study was to begin to develop within session volume parameters for perturbation-based balance training by determining the minimum number of exposures needed for participants to habituate to balance recovery from a substantial perturbation. Two young adult participants were exposed to 15 substantial perturbations induced via release from a static forward lean. All participants were instructed to attempt to recover balance by taking a single rapid step. A scalable anatomical model consisting of 36 degrees-of-freedom and 94 muscle actuators was used to compute kinematics and joint moments from motion capture and force plate data. Margin of stability was calculated at heel strike and maximum knee joint flexion to quantify balance recovery performance. Balance recovery trials were divided in to three blocks (early, middle, and late) with 5 trials in each block and static optimization was used to compute estimates of the mean and peak force generated by key muscle groups during recovery for each block. Participant margin of stability declined rapidly during early balance recovery trials and plateaued between trials 5-8. Hip abductor forces remained relatively consistent across trials while the force produced by the Vastus Femoris group decreased during the mid and late trial blocks. Mean force produced by the Soleus during balance recovery decreased across trial blocks. In contrast, the peak force and impulse generated by the Soleus increased across trial blocks. We interpret these data to reflect an adaptation in balance recovery coordination strategy, which appears to occur sometime between the early and middle trial blocks and stems from young participants' desire to minimize the effort associated with successful recovery of balance. These results further suggest participants became habituated to balance recovery after exposure to 5-8 substantial perturbations.

Introduction

An estimated 3 million older adults fall each year. Falls are the leading cause of injury and death in older adults in the United States and cost the health care system over $50 billion each year (Bergen et al., 2016; Florence et al., 2018). Both younger and older adults are exposed to losses of balance during activities of daily living, while older adults are more likely to experience a fall because of their compromised ability to recover from a loss of balance (Arampatzis et al., 2008; Carty et al., 2011; Madigan & Lloyd, 2005; Thelen et al., 2000; Thelen, 2003). This is in part due to older adults reduced capacity to produce an adequate recovery step response in terms of both step length and time (Carty et al., 2011; Karamanidis et al., 2008) and poor trunk control (Bieryla et al., 2007; Crenshaw et al., 2012) which lead to a reduction in stability during balance recovery and subsequently an elevated fall risk.

While benefits of physical activity in regard to the health and wellness of older adults are well documented (Chodzko-Zajko et al., 2009; Nocon et al., 2008) and lower extremity strengthening has been associated with reduced fall risk (Karamanidis et al., 2008; Pijnappels et al., 2008) exercise-based balance training programs only resulted in a 17-24% reduction in fall rate (Sherrington et al., 2008). Perturbation-based balance training (PBT) is an emerging fall prevention paradigm which exposes participants to gait or postural perturbations to cause loss of balance and trigger a reactive stepping response (Mansfield et al., 2015). PBT has shown promising results in terms of fall prevention and fall risk reduction and has improved balance recovery capacity after a single session (Barrett et al., 2012a; Grabiner et al., 2008; Maki et al., 2008; Yungher et al., 2012).

Thus far, PBT interventions have been highly variable, perturbations have been induced via release from a static forward lean (Carty et al., 2011; Karamanidis & Arampatzis, 2007; Thelen et al., 1997), slips induced with movable floor platforms translations (Bhatt et al., 2012; Pai et al., 2010; Parijat & Lockhart, 2012) and rotations (Gage et al., 2008; Keshner et al., 1988), ground surface compliance changes (Bierbaum et al., 2010, 2011) , treadmill accelerations and decelerations (Grabiner et al., 2012; Lee et al., 2018; Lurie et al., 2013; Rieger et al., 2020; Sakai et al., 2008), waist pulls (Fujimoto et al., 2015; Martelli et al., 2016; Pai et al., 1998) and weight drop cable pulls (Mansfield & Maki, 2009). Due to the variability of PBT interventions, to date much is unknown regarding training parameters for PBT such as perturbation type, magnitude, volume, frequency and its mechanism of improvement (Gerards et al., 2017a). However, the importance of step response balance recovery capacity is generally known and supported by several prospective studies which have demonstrated that the capacity to recover balance in a single step compared to multiple steps strongly predicted future fall incidence for older adults (Carty et al., 2015; Hilliard et al., 2008; Mille et al., 2013).

In an attempt to improve the understanding around PBT parameters, chapter three investigated Margin of Stability (MoS) during repeated exposure to balance recovery from substantial perturbations in young adults. Unexpectedly, participants demonstrated a decline in MoS as exposure to balance recovery increased. Interestingly, the step length (Carty et al., 2011); Graham et al., 2014; Rogers et al., 2001), step time (Carty et al., 2011; Graham et al., 2014; Hsiao-Wecksler & Robinovitch, 2007; Hsiao & Robinovitch, 1999; Rogers et al., 2001; Thelen et al., 2000), and lean angle (Carty et al., 2011; Graham et al., 2014; Hsiao-Wecksler & Robinovitch, 2007) observed in chapter three were more extreme than commonly reported values for younger participants. Authors suggested subjects inherent stability may have exceeded the balance recovery demands of the perturbations and the observed decrease in MoS may have been representative of younger participants' desire to minimize the effort associated with balance recovery. Their interpretation of the results is supported by Hsiao & Robinovitch (1999), who predicted recovery step length and time for younger steppers using a model that minimized the muscular effort associated with each recovery.

To build on the results of chapter three and add clarity to the authors interpretation of said results, a secondary analysis was suggested to estimate muscular effort during repeated exposure to balance recovery. Whereas, Hsiao & Robinovitch (1999) used predicted peak contact force to estimate effort, authors suggested a muscle level analysis, similar to that conducted by Graham et al., (2014) who estimated the force generated by key muscle groups during balance recovery, would be more appropriate to estimate muscular effort.

Therefore the purpose of this study is to perform a secondary analysis which will build on the results reported in chapter three regarding habituation during repeated exposure to balance recovery. This purpose will be accomplished by estimating the force generated by key muscle groups and assessing whether muscular effort decreased during repeated exposure to balance recovery.

Methods

Participants

Participants consisted of two college-aged adults (age: 22.5 ± 1.5 years, height: 1.71 ± 0.03m, mass: 70.5 ± 1.5kg) from chapter three. All participants had normal vision. Individuals that reported neurological, cognitive, metabolic, or musculoskeletal impairment were excluded.

Ethics approval was obtained from the Institutional Review Board at Montana State University, Bozeman.

Experimental procedures

The balance recovery protocol conducted was the same as previously described in chapter three. Participants were fitted to a safety harness and the tilted into a static forward lean by a horizontal cable attached to a safety support rig located behind the participant on one end and the participant's harness at waist level on the other. Participants were instructed to maintain a rigid posture, with their head, hips, knees, and ankles in alignment prior to release and instructed to recover their balance as best they could with a single step following release. Once the desired posture was achieved, participants were released after a random time period (2-10s). Center of pressure location was displayed in real time and visually inspected prior to cable release to ensure anticipatory actions such as medio-lateral or antero-posterior weight shifting was not evident prior to cable release. The participants harness was attached to the ceiling by a safety rope to prevent the participant from contacting the ground in the event of a failed recovery.

Participants were given two familiarization trials from a minimal forward lean. Following the familiarization trials each participant completed 15 releases from a substantial lean angle. Lean angle was determined as the furthest angle by which participants could lean while still maintaining heel contact with the floor. Trajectories of 51 reflective markers attached to participants bony landmarks and grounds reaction forces under the foot were recorded simultaneously using a 10-camera motion capture system (Motion Analysis Corp.) recording at 220 Hz and three force plates (Advanced Mechanical Technologies, Inc.) recording at 1000 Hz. The first plate was located beneath the participants right foot in the initial forward lean position and the second was located beneath the left foot. The third plate was anterior to the first plate in order to record the ground reaction forces associated with the stepping foot.

Specific events during balance recovery were defined as follows. Cable Release (CR) was identified as an increase in force on the two plates beneath the participant's feet after a static forward lean had been achieved. Heel Strike (HS) was identified from a force recorded in excess of 20 N recorded on the anterior force plate. Knee Joint Maximum (KJ_{max}) was identified as the point, after heel strike, at which the greatest knee flexion angle was calculated by inverse kinematics for the stepping leg.

Data analysis

Data analyses were performed using OpenSim (version 4.0) (Delp et al., 2007) in conjunction with custom MATLAB scripts (Version 2019a, The Math Works, USA). A scalable anatomical model with 17 bodies, 17 joints, 94 muscle actuators and 36 degrees of freedom, as described by Hamner et al., (2010), was used as the generic model for analysis. A wrap object was embedded in the generic model to prevent non-physiologic moment arms for the erector spinae muscle during trunk flexion (Daggfeldt & Thorstensson, 2003; Graham et al., 2014).

Model Scaling and Inverse Kinematics (Lu & O'Connor, 1999) were performed by fitting the anatomical model to measured 3D marker positions with a high weighting on virtual markers which defined the joint center of the hip, knee and ankle. Joint centers were estimated from experimental marker trajectories: the regression equations of (Harrington et al., 2007) were used for the hip joint, while the knee and ankle joint centers were identified as the midpoints of the femoral condyles and the medial and lateral malleoli, respectively.

Residual Reduction Analysis (RRA) was performed to improve the dynamic consistency of the mass and accelerations of the model and the measured ground reaction forces (Delp et al., 2007). Mass adjustments were made to each of the models segments according to the recommendations produced by RRA. The final average residuals following RRA were: Fx = -3.2 N, Fy = 110.6 N, Fz = -55.7 and Mx = -72.1, My = -68.4, Mz = 6.7. Muscle force estimates were computed by minimizing the sum of squared activations within the force-length-velocity constraint for each muscle, using the RRA mass adjusted model. Static optimization is commonly used to estimate muscle during movements of interest including balance recovery (Dorn et al., 2012; Graham et al., 2014, 2016). All muscle actuator force estimates are reported relative to body weight. Muscle force analysis was performed for the time from cable release to knee joint maximum. Key muscle groups were identified based on their contribution to balance recovery as reported by Graham et al., (2014) and were grouped as follows: Hip Abductors of the stance leg (HAB – gluteus minimus and medius), Soleus of the stepping leg (SOL) and Vastus Femoris of the stepping leg (VAST – vastus lateralis, vastus intermedius and vastus medius). For analysis tether release trials were grouped into early (trials 1-5), middle (trials 6-10) and late (trials 11-15).

Results

Step length was longest during trials 1-5 and shortest during trials 6-10, mean step length was 0.92 m, mean step time was 0.185 s and mean lean angle was 36˚. A summary of all spatial-temporal variables is displayed in Table 1.

Of the key muscle groups, largest mean, and peak normalized muscle forces from cable release to knee joint max were generated by the Hip Adductors of the stance leg. Mean HAB force was highest during the middle trials and slightly lower during the early and later trials while peak HAB force remained relatively consistent across all trials (figure. 1). Peak and mean normalized forces for the Vastus Femoris group were highest during the early trials and declined into the middle and late trials (figure 1). Soleus peak force increased from early to middle trials and mid to late trials. Whereas, mean Soleus force decreased from early to middle trials and remained relatively constant between middle and late trials (figure 1).

Table 1. Margin of Stability, values reported as mean ± standard deviation

Margin of Stability (cm)	Early (1-5)	Middle (6-10)	Late (11-15)	All Trials
Heel Strike	17.4 ± 7	16.5 ± 2.3	13.2 ± 4.6	15.7 ± 5.4
Knee Joint Maximum	33.63 ± 12.1	32.6 ± 2.7	26.6 ± 8.2	30.9 ± 9.1

Similarly to their peak and mean force the Vastus Femoris group produced the largest mean total impulse during trials 1-5 and the smallest impulse during trials 10-15. Impulse created by the Soleus was also greatest during trials 10-15 and smallest during trials 1-5. Hip Abductor impulse was greatest during trial 6-10 and smallest during trials 10-15. Ensemble averages of the normalized force produced by the Hip Abductors, Vastus Femoris and Soleus from cable release to knee joint max as well as the average impulse are displayed in figure 2.

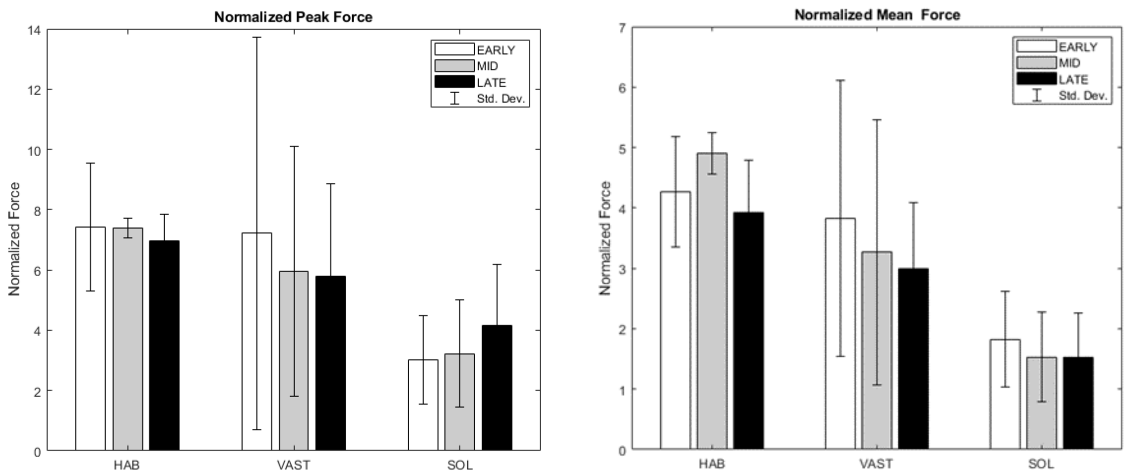

Figure 1. Normalized mean and peak force by muscle group. All forces are expressed as proportion of bodyweight. Muscles presented are stance side Hip Abductors (HAB), stepping side Vastus Femoris (VAST) and stance side Soleus (SOL)

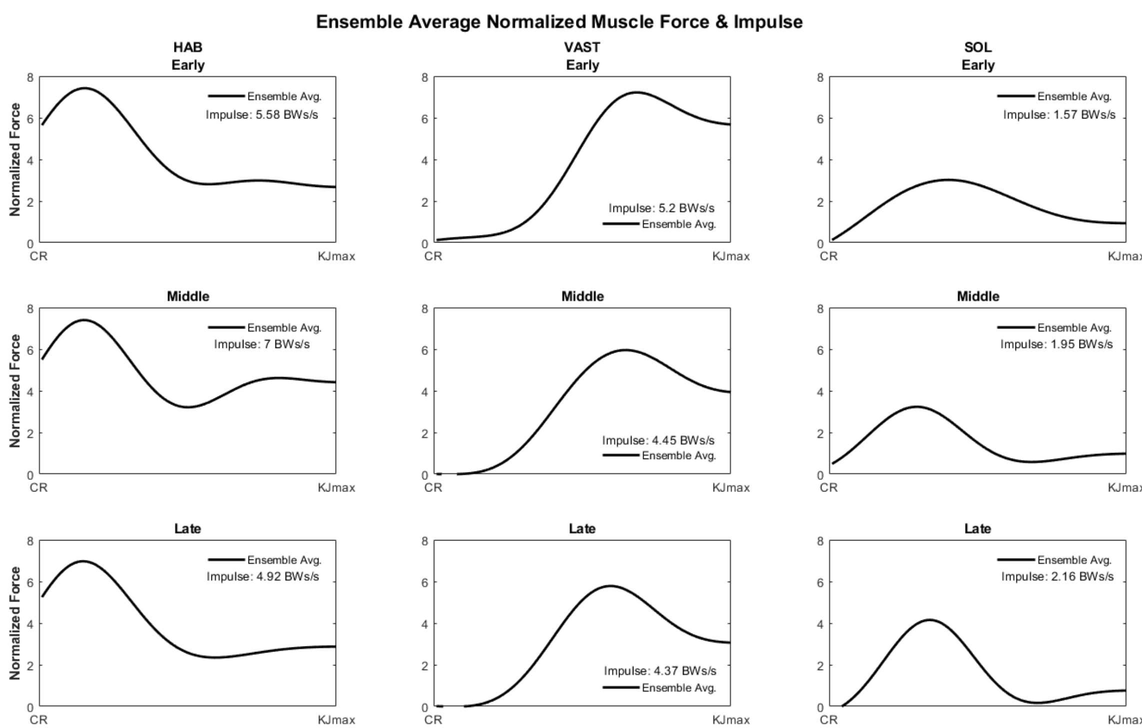

Figure 2. Normalized muscle force ensemble average and mean total impulse for early, middle, and late trials. All forces and impulse are expressed as proportion of bodyweight. Muscles presented are stance side Hip Abductors (HAB), stepping side Vastus Femoris (VAST) and stance side soleus (SOL)

Discussion

We have described the changes in muscular forces produced during compensatory stepping response after repeated exposure to forward loss of balance in a group of younger adults. The largest normalized force was generated by the Hip Abductors of the stance leg, which is consistent with the findings of Graham et al., (2014). Of the muscle groups analyzed, the Hip Abductors were the only group that did not decrease in mean force in later trials, they produced the largest mean forces during the middle trials and similar mean forces during the early and late trials. However, we demonstrated that mean force and mean total impulse of the Vastus Femoris group as well as Soleus mean force were highest during the early trials and decreased during the mid and late trial periods.

Similarly to its mean force, the peak force generated by Vastus Femoris was largest during early trials and decreased during later trials. The mean total impulse generated by Vastus Femoris also decreased from early to late trial periods. In contrast Vastus Femoris, and unlike its mean force, the peak force generated by the Soleus was lowest during early trials and highest in later trials. However, the mean total impulse generate by the Soleus increased from early to late trial periods.

Ensemble averages of each trial period indicate that Vastus Femoris begins to produce force earlier during recovery and drops from its peak more rapidly in middle and later trials than early trials. Similarly to the Vastus Femoris, ensemble averages indicate that force generated by the Soleus in later trials declined more rapidly in later trials than earlier trials. Further, it appears that participants reached peak force more rapidly after cable release during later trials despite the higher peak forces generated by the Soleus.

The relatively stable forces generated by the Hip Abductors in conjunction with the decrease in impulse generated by the Vastus Femoris and increase in impulse generated Soleus from the early trial period to the mid and late trial periods suggest that participants may have adapted their recovery stepping strategy sometime after or during the early trials to produce less force but still recover balance. Specifically, it appears that by generating larger peak forces more rapidly with the soleus and subsequently lowering force production demands on the Vasti later in recovery, participants were able to reduce the muscular effort needed to achieve successful recovery in later trial periods.

Limitations

Average residual forces and moments following RRA were kept as low as possible but exceeded the standard tolerable threshold outlined in the OpenSim documentation (Hicks, 2012). This was in part due to technical issues in the lab that resulted in an offset between the measured ground reaction forces and kinematic data. The offset was addressed to the best of our abilities during data processing, but the smallest of offsets could create dynamic inconsistencies and result in elevated residuals such as ours. A further limitation to this study is the use of static optimization to compute muscle force estimates. Static optimization assumes that muscles were recruited in an order that minimizes muscle activation, which may not be the way muscles are recruited during balance recovery from large perturbations. An additional limitation to this study is the model used for analysis. The generic model was originally developed for gait analysis and was not strong enough to accurately simulate the rapid movements and extreme ranges of motion demonstrated by our participants during balance recovery. Subsequently, the model had to be globally strengthened prior to static optimization. Due to the model strengthening the absolute values for our muscle force estimate are inflated and non-physiologic, however, the relative differences in estimates are still valid. Finally, the small sample size of this study (n=2) was likely not representative of the larger population of young steppers. However this limitation needed to be balanced with the time required to perform the large number of simulations involved in our analysis.

Conclusion

Considered with the results of the previous analysis from chapter three our results suggest that participants adopted a new strategy or updated their existing recovery step response strategy to require less muscular effort from key muscle groups during mid and late trials compared to early trials, while still achieving a successful single step recovery result in each trial. This supports our interpretation in chapter three that habituation occurred somewhere between trials 5-8, although further work is needed to confirm these results.

CHAPTER FIVE

GENERAL DISCUSSION

The purpose of this study was to begin to develop within session volume parameters for PBT by determining the minimum number of exposures needed for participants to become habituated to balance recovery. We have demonstrated that younger adults experienced a decrease in Margin of Stability during repeated exposure to balance recovery from substantial perturbation. Where average MOS, for both Heel Strike and Knee Joint Max, was largest during early trials and declined in each of the subsequent trial periods and appeared to plateau after 5-8 trials. We further demonstrated that the muscular effort associated with successful balance recovery declined as participants exposure to balance recovery increased, where the force production patterns of the stance side Soleus and stepping side Vasti changed after the early trial period.

Temporal-Spatial Factors

As highlighted in chapters three and four, the temporal spatial characteristics demonstrated by participants during recovery were generally more extreme than those commonly reported in previous studies on balance recovery in younger adults (Carty et al., 2011; Graham et al., 2014; Hsiao-Wecksler, 2008; Rogers et al., 2001; Thelen et al., 2000). Since the magnitude of the perturbations used in this study were also larger than that of those previously reported, these characteristics are consistent with the previous studies which demonstrated an increases in step length and decrease in step time during recovery from perturbations of increasing magnitude (Do et al., 1982; Hsiao & Robinovitch, 1999; Maki et al., 1996; Thelen et al., 1997).

Notably, Hsiao & Robinovitch (1999) demonstrated that modulation of step length and step time as perturbation magnitude increased was associated with a desire to reduce the effort of balance recovery in young subjects. Participants took shorter and quicker steps as perturbation magnitude increased and changes in step length and time were associated with a decrease in predicted leg contact force. Specifically, these changes were most consistent with a model which predicted step length and time by minimizing predicted leg contact force. Authors interpreted their findings as younger subjects' desire to minimize the effort associated with successful balance recovery (Hsiao & Robinovitch, 1999).

Muscle Force Estimates

In chapter four, we demonstrated a change in the estimated average impulse and force generated by key muscle groups during balance recovery, including the stance side Soleus and the stepping side Vastus Femoris group. Specifically, it was demonstrated that the average and peak force as well as average impulse generated by the Vastus Femoris group was greatest during early trials and decreased in subsequent trial periods. The average force produced by the Soleus was also greatest during early trials. However, the peak force and average impulse generated by the Soleus were lowest during early trial periods and increased during the mid and late trial periods.

Ensemble averages of each trial period, reported in chapter four, indicate that Vastus Femoris began to produce force earlier during recovery in the mid and later trials than early trials. Additionally, the Soleus appeared to reach peak force production earlier during recovery and decline more rapidly from its peak in later trials despite producing larger peak forces. Further, since all recoveries were successful, the reduction in average force generated by the

Soleus coupled with the increase in its peak force and average impulse production, indicate it may have been used more efficiently during recovery in mid and later trials.

Synthesis and Interpretation

We believe it is reasonable to suggest the decrease in MoS reported in chapter three may be interpreted as an adaptation of balance recovery coordination, stemming from the desire of younger subjects to minimize the effort associated with a successful recovery. This is supported by Hsiao & Robinovitch (1999), as well as the changes in the force and impulse generated by key muscle groups as reported in chapter four.

If we are to accept this interpretation of the results, it would suggest that despite the decrease in MoS, participants may still have become habituated to balance recovery in the form of learning to reduce the effort needed for a successful recovery of balance. Specifically, it appears that habituation may have occurred between trial 5-8, where the difference in impulse force was greatest in the soleus and vastus femoris group in chapter four and the plateau in MoS was observed in chapter three.

Limitations

There are several limitations which must be considered when interpreting the findings of this study. The first and most apparent limitation to this study is the sample size (n=2). There were only two participants in this study and the results are not generalizable to young adults in general as they may display different balance recovery characteristics or trends in balance recovery performance. This is limitation had to be weighed against the computational cost of running the large number of simulations required for data analysis, as well as the risk of bringing additional subjects into the Laboratory during the height of the COVID-19 pandemic.

An additional limitation to this study is the final average residuals from RRA, which exceeded the tolerance outlined in the OpenSim documentation (Hicks, 2012). We suspect this was primarily due to an offset between the recorded ground reaction forces and marker trajectories, which was discovered after data collection. The offset was addressed to the best of our abilities during data processing with custom MATLAB scripts which re-synchronized force and marker data to account for the offset and RRA was conducted several times to minimize the final average residuals as much as possible within the time constraints of the study. However, even minor remaining differences in force and marker data that were undetected after data processing could result in dynamic inconsistency between the recorded forces and marker trajectories and produce residual errors like those reported in chapter 4.

A final limitation which must be considered is the model used for this study. The generic model used for analysis was originally based off of the gait2392 OpenSim model (Delp et al., 2007) which was has been used primarily for the simulation and analysis of gait (Hamner et al., 2010). Derivations of this model have been used extensively to simulate tether-release balance recovery (Barrett et al., 2012a; Cronin et al., 2013; Graham et al., 2014, 2015, 2016). However as described in previous sections the forward lean angle and step length experienced by participants in this study exceeded those commonly reported. Subsequently, during initial attempts to solve static optimization several actuators encountered constraint violations such as muscle length-tension and optimal fiber length violations during simulation. In order to successfully simulate the relatively extreme temporal-spatial balance recovery characteristics demonstrated by participants in this study, the model was globally strengthened. Consequentially, the magnitude of the muscle force estimates reported in this were elevated, however the relative difference in force generated by muscle groups between trial periods remains valid. Future analysis should utilize a model which is better equipped to handle the ranges of motion experienced by younger participants during balance recovery from a substantial forward lean.

Conclusion

This study demonstrated that younger adults experience a decline in Margin of Stability during repeated exposure to substantial perturbations. It was further demonstrated that this decline in stability was accompanied by an adaptation in balance recovery coordination strategy. We interpret this decline in stability as a consequence of the adaptations in balance recovery coordination strategy which reflects an attempt to reduce the effort associated with successful recovery of balance. Further it appears these adaptations occur after exposure to 5-8 perturbations; future work is needed to verify this interpretation. Future efforts to better understand training parameters for PBT should aim to replicate this protocol with a larger sample size to verify our results. Additionally, future research should be extended to include older populations in order to better understand habituation to balance recovery for older adults.

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