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Effects of age and acute muscle fatigue on reactive postural control in healthy adults

Evan V. Papa [a,*], K. Bo Foreman [b], Leland E. Dibble [b]

[a] University of North Texas Health Science Center, 3500 Camp Bowie Blvd., Fort Worth, TX, 76107, USA
[b] University of Utah, 520 Wakara Way, Salt Lake City, UT, 84108, USA

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ABSTRACT

Background: Falls can cause moderate to severe injuries such as hip fractures and head trauma in older adults. While declines in muscle strength and sensory function contribute to increased falls in older adults, skeletal muscle fatigue is often overlooked as an additional contributor to fall risk. The purpose of this investigation was to examine the effects of acute lower extremity muscle fatigue and age on reactive postural control in healthy adults.
Methods: A sample of 16 individuals participated in this study (8 healthy older adults and 8 healthy young persons). Whole body kinematic and kinetic data were collected during anterior and posterior reproducible fall tests before (T0) and immediately after (T1) eccentric muscle fatiguing exercise, as well as after 15-min (T15) and 30-min (T30) of rest.
Findings: Lower extremity joint kinematics of the stepping limb during the support (landing) phase of the anterior fall were significantly altered by the presence of acute muscle fatigue. Step velocity was significantly decreased during the anterior falls. Statistically significant main effects of age were found for step length in both fall directions. Effect sizes for all outcomes were small. No statistically significant interaction effects were found.
Interpretation: Muscle fatigue has a measurable effect on lower extremity joint kinematics during simulated falls. These alterations appear to resolve within 15 min of recovery. The above deficits, coupled with a reduced step length, may help explain the increased fall risk in older adults.

1. Introduction

Falls are a major source of morbidity and disability in the aging population. Twenty to thirty percent of older adults who fall suffer moderate to severe injuries such as lacerations, hip fractures, and head traumas (Sterling et al., 2001). The economic burden of falls in older adults is very high and the direct medical cost of falls in 2012, adjusted for inflation, was $30 billion (CDC, 2012; Stevens et al., 2006). Despite the vast public health problem of falls in the older population, studies have generally neglected the impact that muscle fatigue may have on falls.

Acute muscle fatigue has been shown to alter postural control in healthy young and older individuals, which may increase the risk for falls (Bellew and Fenter, 2006; Nam et al., 2013; Paillard, 2012). More specifically, acute muscle fatigue is known to modify the peripheral proprioceptive system and the central processing of sensory inputs (Taylor et al., 2000) both of which are integral for reactive postural control.

Recovery from external perturbations requires reactive postural control, which is defined by modifying sensory and motor systems in response to changing tasks and externally induced postural demands (Shumway-Cook and Woollacott, 2012). The majority of falls in older adults occur in the context of tasks requiring reactive postural control (Niino et al., 2000). In addition, the chances of sustaining a fall are particularly high during slipping or tripping situations (Lord et al., 1993).

Several studies have experimentally examined the effect of acute muscle fatigue on reactive postural control in healthy older adults (Adlerton and Moritz, 2001; Davidson et al., 2009; Granacher et al., 2010; Mademli et al., 2008). Decrements such as increased center of mass (COM) displacements (Davidson et al., 2009), increased center of pressure (COP) displacements (Nam et al., 2013) and sway velocity (Parreira et al., 2013), increased antagonist muscle co-activation (Granacher et al., 2010), and decreased functional reflex activity (Granacher et al., 2010) have been demonstrated following acute muscle fatigue in older individuals in the context of externally induced perturbations. However, just one of these studies has examined the particular biomechanical aspects of postural control recovery at joint-specific anatomic locations (Mademli et al., 2008). This lack of biomechanical detail limits insight into the lower extremity segmental characteristics of postural instability. While Mademli et al. demonstrated alterations in the knee joint angle following muscle fatigue, they did not provide any indication of the duration of the fatigue effect. In general, investigations of muscle fatigue and postural control have neglected

* Corresponding author.
E-mail addresses: evan.papa@unthsc.edu (E.V. Papa), bo.foreman@hsc.utah.edu (K.B. Foreman), lee.dibble@hsc.utah.edu (L.E. Dibble).

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the recovery timeline of lower extremity proximal and distal joint kinematics.

The purpose of this investigation was to examine the effects of age and acute muscle fatigue on the biomechanic responses of reactive postural control. The secondary objective was to examine the timeline of postural control recovery following acute muscle fatigue. We hypothesized that regardless of age, acute muscle fatigue would result in declines in kinematic outcomes of reactive postural control but that the declines would be more substantial in older adults. We further hypothesized that these alterations would return to baseline within 20-min of post-exercise fatigue recovery (Yaggie and McGregor, 2002).

2. Methods

2.1. Study sample

Participants were recruited from the local community. Sample sizes were estimated using an effect size of 0.68 for peak center of pressure (COP) displacement following postural perturbations in previously published data (Davidson et al., 2009) and estimate tables (Portney and Watkins, 2009) based on an α level = 0.05 and 90% power. Inclusion criteria for the healthy older group required that individuals be older than 50 years of age (Toebes et al., 2012) and between 18 and 35 years of age for the healthy young group. Participants were excluded if they had participated in vigorous exercise 24 h prior to initiating the study. Subjects were also screened for musculoskeletal or neurologic impairments affecting postural stability via a health questionnaire and baseline functional mobility testing performed by a physical therapist trained in fall prevention. All participants were healthy, generally active individuals who were independent in activities of daily living. This study was reviewed by the University of Utah Institutional Review Board and participants provided informed consent prior to participation.

2.2. Postural control assessments

The reactive postural control task utilized a tether-release model, which forced the subjects to incorporate a protective step to regain stability. The tether-release protocol has been used previously to investigate balance recovery from anterior (Carty et al., 2011; Madigan, 2006; Madigan and Lloyd, 2005) and posterior falls (Hsiao and Robinovitch, 2001). The protocol for this study consisted of securing one end of a tether to a trunk harness at the level of the xiphisternal joint. The other end of the tether was connected to an inline force sensor and electromagnet that was fixed to the wall. Participants were asked to lean against the tether until 9–12% of their body mass registered through the inline force sensor. This value has been shown to exceed sway-based recovery abilities (Hsiao-Wecksler, 2008). Once the subjects were in position for the trial, they were given the following instruction: "When the tether is released try to recover your balance with a single step." Release of the tether was randomized between 1 and 20 s from the time they were in position to limit anticipation of the release time. Five consecutive trials were performed in each of the anterior and posterior falling directions, with data analysis being performed on three successful trials for each direction. A trial was considered successful if the individual was able to recover from the tether-release independently in one step, without assistance from the overhead harness, and the joint markers were visible throughout the trial.

For analysis, the reactive postural control task was divided into the following phases: 1) The *swing phase* represented the time between when the stepping foot left the force platform to the point at which the same foot struck the second force platform upon landing (Fig. 1). 2) The *support phase* represented the point from when the stepping foot struck the second force platform upon landing until the individual's COM stopped moving in the direction of the fall.

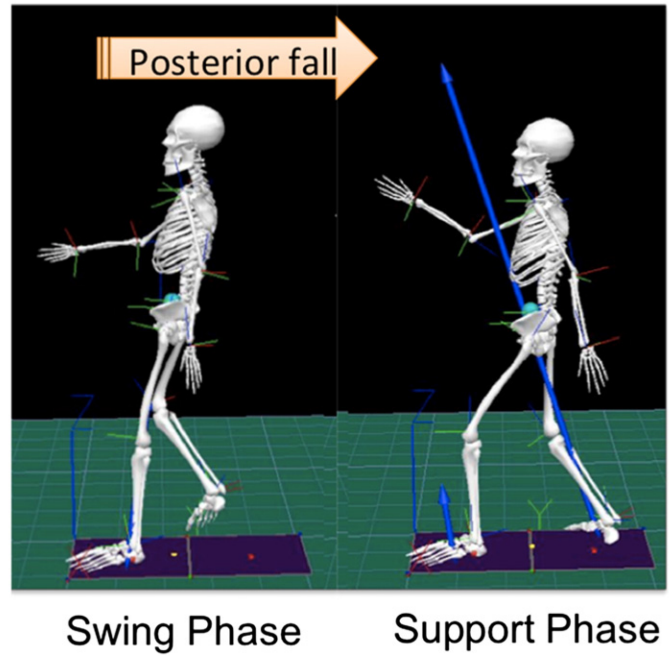

Fig. 1. Diagram of a posterior tether-release task, demonstrating both swing and support phases.

2.3. Data collection and procedures

Following consent, subjects were instrumented with reflective markers based on a standardized gait analysis marker set defining 15 body segments (Plug-In-Gait marker set; Vicon Motion Systems; Oxford, UK). Data were acquired using a Vicon 10-camera motion analysis system (Vicon Motion Systems; Oxford, UK) and 2 AMTI OR6 series force platform systems (AMTI; Watertown, MA) at a rate of 250Hz. Data were recorded and synchronized using Vicon Nexus (Vicon Motion Systems; Oxford, UK) and post-processing occurred using Visual3D (C-Motion, Germantown, MD). Marker and force platform data were filtered using a fourth-order, low pass, zero-phaseshift Butterworth filter at 6 Hz and 20 Hz, respectively (Winter, 2009).

Reactive postural control assessments were performed before (T0) and immediately (<2min) after muscle fatiguing exercise (T1), as well as after 15-min (T15) and 30-min (T30) of rest.

Prior to baseline testing participants were exposed to 1–3 trial sessions of the tether-release test in order to become familiar with the testing procedure, and to overcome the fear of falling. After performing the baseline (T0) assessments, markers on the posterior aspect of the trunk and pelvis were removed for seated fatiguing exercises (described below). To assure accurate re-application of the markers, the base of each marker was traced prior to removal. Immediately after exercise the markers were re-applied and the immediate post (T1) postural control assessments were performed. Following 15 min of rest the T15 assessments were performed. After a second 15-min rest, the T30 postural control assessments were completed.

2.4. Postural control outcomes

Outcomes collected during the swing phase of the tether-release task included: step length normalized by height and step velocity. Outcome measures collected during the support phase included: hip, knee, and ankle joint angular displacements, peak COP displacement, and center of pressure-center of mass (COP-COM) difference. The COP-COM difference examined the difference between the COP position at its peak displacement in the medio-lateral direction and the COM at