



Effects of submaximal fatiguing contractions on the components of dynamic stability control after forward falls

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ABSTRACT

The present study aimed to investigate the effect of lower extremity muscle fatigue on the dynamic stability control of physically active adults during forward falls. Thirteen participants (body mass: 70.2 kg, height: 175 cm) were instructed to regain balance with a single step after a sudden induced fall from a forward-leaning position before and after the fatigue protocol. The ground reaction forces were collected using four force plates at a sampling rate of 1080 Hz. Kinematic data were recorded with 12 vicon cameras operating at 120 Hz. Neither the reaction time nor the duration until touchdown showed any differences ($p > 0.05$). The ability of the subjects to prevent falling did not change after the fatigue protocol. In the fatigued condition, the participants demonstrated an increase in knee flexion during the main stance phase and an increased time to decelerate the horizontal CM motion (both $p < 0.05$). Significant ($p < 0.05$) decreases were seen post-fatigue in average horizontal and vertical force and maximum knee and ankle joint moments. The fatigue related decrease in muscle strength did not affect the margin of stability, the boundary of the base of support or the position of the extrapolated centre of mass during the forward induced falls, indicating an appropriate adjustment of the motor commands to compensate the deficit in muscle strength.

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1. Introduction

Falls are an inherent part of life for most everyone. A number of factors may contribute to falls. Many studies have examined static and dynamic balance situations in an attempt to learn more about how the human system controls stability and prevents falls.

Although aging is a common focus of balance literature, a loss of muscular strength can also be induced through means other than aging, such as fatigue (Yeung et al., 1999; Moritani et al., 1990). Several studies reported that loss of muscle strength may alter the capacity of the human system to generate rapid force for balance corrections after sudden perturbations (Granacher et al., 2008; Pijnappels et al., 2005; Simoneau and Corbeil, 2005). Karamanidis and Arampatzis (2007) reported that muscle–tendon capacities of the lower limbs contribute about 33% to balance recovery after forward induced falls. However, postural corrections after a sudden perturbation involve sensorimotor adaptational responses which include mechanisms responsible for maintaining the dynamic stability and thus muscle weakness may be partly compensated for by proper planning and execution of the used locomotion strategy. By using fatigued younger subjects to test

the effects of muscular strength on forward falls we can control for the other changes that occur with aging. Three mechanisms have been presented (Hof, 2007) by which stability may be maintained; by (1) increasing the base of support with relation to the extrapolated centre of mass, (2) counter rotating segments around the CM, and (3) applying an external force other than the ground reaction force (e.g. grasping a handrail or secure object).

Examining the human's capabilities to regain balance after sudden release from a forward inclined body position is a common way to assess dynamic stability (Hsiao and Robinovitch, 1999; Theilen et al., 1997; Wojcik et al., 2001). The margin of stability during locomotion can be quantified by the position of the extrapolated centre of mass (CM) in relation to the base of support (Hof et al., 2005). When the position of the extrapolated CM exceeds the anterior boundary of the base of support, the result is a loss of stability (Hof et al., 2005). Although studies regarding either fatigue or balance are numerous, to date only one study we found has examined the effects of fatigue on dynamic stability control (Mademli et al., 2008). This study found that fatigue had no effect on the components of dynamic stability. However, the fatigue protocol for this study used the knee extension exercise and therefore only fatigued the knee extensor muscles. Preventing a forward fall requires a motion from the front leg that is similar to a lunge and requires not only the knee extensor muscles, but also hip and ankle muscles. The knee extension is a single joint movement requiring

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force production from one muscle group and has little relevance to normal human movement. Additionally, it has been reported that to regain balance during forward falls, extensor muscles from the major lower extremity joints (hip, knee and ankle) are all important (Pijnappels et al., 2005; Madigan and Lloyd, 2005).

Therefore the purpose of the present study is to experimentally investigate the effect of fatigue of the lower extremity, induced by submaximal fatiguing contractions, on the dynamic stability control of young adults during forward falls. We hypothesize that dynamic stability will be compromised when the lower extremity is fatigued. This could be seen in a decrease in the ability to regain balance after an induced forward fall.

2. Methods

Thirteen participants (7 men and 6 women), age 27.7 ± 4.5 years, body mass: 70.2 ± 13.1 kg, body height: 175 ± 11.1 cm, participated in this study. Before the initiation of this experiment, the procedures were explained to the subjects and informed consent protocol was followed in accordance with the rules of the University Human Subjects Board.

Subjects were instructed to regain balance with a single step after a sudden induced fall from a forward-leaning position both before and after a fatiguing protocol targeting the lower extremity. The experiment was performed at the maximal individual lean angle of each participant (maximal forward incline position was where the subjects were able to regain balance with a single step after the sudden release of a support cable which supported them from behind). The force on the cable at maximum lean position was between 25–40% bodyweight. The maximal individual lean angle was determined experimentally before the main experiment. After a short rest (~5 min) subjects were tested at their previously measured maximum lean angle. The experimental design of the balance recovery task has been previously described (Karamanidis and Arampatzis, 2007). Following the determination of the subject's maximum lean angle from which they can be released and still catch themselves, all subjects performed a maximum voluntary isometric squat. Subjects performed the isometric squat with a knee angle of approximately 110° . The knee angle was controlled using a goniometer. The isometric squats were performed on a force platform with an adjustable squat bar that was fixed at a height so that the subject's knee angle was approximately 110° . The maximum measured force as measured by the force platform was considered the isometric squat maximum.

For the fatigue protocol, subjects were instructed to perform dynamic squats from an extended knee angle of approximately 180° to a lower position of approximately 90° . The weight used during the fatiguing protocol was 30% of their maximum isometric squat. The dynamic squats were performed with free weights. Once they reached the extended leg position they immediately started moving down again to eliminate the chance of 'resting' during the fatigue protocol. Three spotters were present during the fatigue protocol to help the subject with the last repetitions and to help place the bar on the squat rack. The dynamic squat sets were completed until voluntary exhaustion. After 1 min of rest another set of squats was performed and this was repeated until each given subject performed four sets. Directly after the fatigue protocol, the maximum isometric squat of each subject was tested again to measure the extent of the fatigue. Then, immediately after the post maximum isometric squat (~30 s) the ability of the subject's to regain balance with a single step after a sudden induced fall was tested again at their maximum lean angle to see if fatigue affected their dynamic stability control. Following the post induced fall trial, the maximum isometric squat was again tested to document any recovery that may have occurred.

3. Equipment

Kinematic data were tracked using a Vicon Motion Capture System using 12 vicon cameras operating at 120 Hz. Thirty-eight reflective markers (diameter 14 mm) were fixed to landmarks of each subject to aid in motion capture. This included a headband that had four markers on it. The whole body model was used to calculate joint angles, the parameters of dynamic stability and the resultant joint moments through inverse dynamics. We used data provided by Zatsiorsky and Selujanov (1983) to calculate masses and moments of inertia of the body segments.

For the trials of each subject, the release, touchdown and minimum knee joint angle were identified. The time of release was defined as the moment the cable holding the subjects in a forward lean was released. Touchdown was defined as the moment the recovery leg contacted the ground after the step. This event was identified using a Kistler force plate (60×90 cm, Kistler, Winterthur, Switzerland, trigger was force exceeding 20 N). Minimum knee angle corresponded with the vertical velocity of the subject reaching a value close to 0 m/s. Using those three points, two recovery phases were identified: (1) phase until touchdown and (2) main stance phase. The reaction time was defined as the time from moment of release to the moment the midpoint of the foot exceeded an acceleration of 1.5 m/s/s.

The ground reaction force of both legs was collected using four Kistler force plates at a sampling rate of 1080 Hz. At the beginning of each trial subjects stood on two force plates and when the fall was induced they used a one step strategy and stepped forward onto another force plate. Ground reaction forces of both feet were inspected before each 'fall' to confirm that subjects were standing symmetrically with their weight evenly distributed between their right and left legs.

The margin of stability in the anteroposterior direction was calculated according to Hof et al. (2005) as follows:

$$b_x = U_{x\max} - X_{CM}, \text{ where } X_{CM} = P_{XCM} + \frac{V_{XCM}}{\sqrt{g/l}}$$

b_x is the margin of stability in the anteroposterior direction, $U_{x\max}$ is the anterior boundary of the base of support (i.e. horizontal component of the projection of the toe from the recovery limb to the ground; a zero value represents the position of the toe before release), and X_{CM} is the position of the extrapolated CM in the anteroposterior direction. P_{XCM} is the horizontal (anteroposterior) component of the projection of the CM to the ground, V_{XCM} is the horizontal (anteroposterior) CM velocity, g is the acceleration of gravity, and l is the distance between CM and centre of the ankle joint in the sagittal plane. The term $\sqrt{g/l}$ is known as the eigenfrequency. Hof et al. (2005) used the eigenfrequency for much smaller angles, but more recently it has also been shown valid for larger angles of the inverted pendulum (Arampatzis et al., 2008; Curtze et al., 2010; Hof, 2008; Hof et al., 2007). The eigenfrequency is used to calculate the extrapolated center of mass, which tells us where the base of support needs to be to attain stability in a moving (falling) subject. Postural stability is maintained in circumstances where the position of the extrapolated CM is within the base of support (positive values of margin of stability) while stability is lost in cases where the extrapolated CM surpasses the anterior boundary of the base of support.

4. Statistics

Paired *T*-tests were performed to determine pre-fatigue/post-fatigue differences in the components of dynamic stability at both touchdown and minimum knee angle. Additionally, paired *t*-tests were used to compare, the margin of stability, the maximum hip,

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