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Individuals with diminished hip abductor muscle strength exhibit altered ankle biomechanics and neuromuscular activation during unipedal balance tasks

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ABSTRACT

Coordinated control of the hip and ankle is important for maintaining postural stability. The purpose of the study was to compare postural stability between individuals with contrasting hip abductor strength during unipedal balance tasks and to determine whether diminished hip abductor strength results in greater utilization of the ankle strategy to maintain balance. Forty-five females (276 ± 35 years) participated in the study. Participants were ranked based on their isometric hip abductor muscle strength. The top 33% of the participants were categorized as the strong group ($n = 15$) and the lower 33% as the weak group ($n = 15$). Each subject performed a static and a dynamic unipedal balance task, during which mean COP displacement, peak ankle invertor and evertor moments, and neuromuscular activation of the lower leg muscles were assessed. Two-way mixed analyses of variance tests with task as a repeated factor were performed to detect the effects of task and group on the variables of interest. When averaged across tasks, mean medial-lateral COP displacement was significantly greater in the weak group (136 ± 117 vs. 98 ± 60 mm, $p = 0.05$). The weak group also exhibited greater peak ankle invertor and evertor moments (0.31 ± 0.10 vs. 0.25 ± 0.11 N m/kg, $p = 0.03$; 0.04 ± 0.06 vs. −0.02 ± 0.07 N m/kg, $p = 0.01$), and increased peroneus longus activation (46 ± 12 vs. 36 ± 15%, $p < 0.01$). Our results demonstrate that individuals with diminished hip abductor muscle strength demonstrated decreased medial-lateral postural stability, and exhibited a shift toward utilizing an ankle strategy to maintain balance during unipedal tasks.

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1. Introduction

Coordinated control of the hip and ankle is important for maintaining postural stability in response to multi-directional perturbations [1]. It has been suggested that the hip provides a greater degree of modulation of body center of mass (COM), while the ankle allows fine-tuning of body COM location and maintains effective foot contact to the supporting surface [2]. Riemann et al. has reported that hip control demand is accentuated under more challenging balance conditions (i.e. standing on unstable surfaces) [3].

The influence of hip muscle performance on postural stability has been described in previous studies. Gribble and Hertel reported that fatiguing the hip abductors and adductors resulted in a significant increase in center of pressure excursion velocity during a single-leg standing task [4]. Additionally, the authors reported that effects of hip muscle fatigue on postural stability was accentuated in persons with chronic ankle instability [5]. Similarly, Bisson et al. reported that fatigue of the hip flexors and extensors resulted in increased center of pressure excursion velocity in the frontal plane [6].

Although existing literature suggests that diminished hip muscle performance is associated with decreased postural stability, the associated compensatory adaptations in lower extremity control have not been elucidated. For example, it is plausible that persons with diminished hip muscle strength may exhibit increased reliance on the ankle to reposition the body COM. Such a strategy would place greater biomechanical and neuromuscular demands on the ankle and may contribute to increased risk of ankle injury. This premise is supported by a number of clinical studies reporting that hip muscle weakness is associated with ankle sprain and soft tissue injuries [7,8].

The purpose of the current study was to compare postural stability during static and dynamic unipedal balance tasks in females with contrasting levels of hip abductor muscle strength. In addition, we sought to determine the influence of hip abductor

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muscle strength on frontal plane kinetics and neuromuscular activation at the ankle. We hypothesized that individuals with diminished hip abductor strength would exhibit: (1) increased medial-lateral center of pressure (COP) displacement, (2) increased peak ankle invertor and evertor moments, and (3) increased neuromuscular activation of the ankle invertor and evertor muscles during static and dynamic unipedal balance tasks.

2. Methods

2.1. Subjects

Forty-five female participants between the ages of 23–34 years participated. Only females were recruited in order to control for the potential sex difference in hip joint structure and function [9]. All participants were recreationally active with no current musculoskeletal complaints.

Participants were excluded from participation if they presented: (1) history of lower extremity or lower back surgery, (2) current lower extremity pain during physical activity, (3) conditions that influence balance, and (4) any other conditions that impair participant's ability to safely perform the activities required. Prior to participation, the objectives, procedures, and risks of the study were explained to each participant. Informed consent as approved by the Institutional Review Board of the University of Southern California Health Science Campus was obtained.

2.2. Instrumentation

Lower extremity joint kinematics were collected using a 11-camera Qualisys® digital motion capturing system (Qualisys Track Manager ver. 25, Qualisys AB, Gothenburg, Sweden). Ground reaction force and center of pressure (COP) data were obtained using an AMTI® force platform (#OR6-6-1, Advanced Mechanical Technology, Inc., MA, USA). Kinematic data were sampled at 250 Hz. Ground reaction force and COP signals were sampled at 1500 Hz.

Electromyographic (EMG) signals of the Peroneus Longus (PL) and Tibialis Anterior (TA) were recorded using a surface EMG system (MA-300, Motion Lab Systems, LA, USA). EMG data were collected at 1500 Hz using pre-amplified, bipolar electrodes consisting of two 9 mm silver/silver chloride discs with a 20 mm inter-electrode spacing (Norotrode 20, Myotronics Inc., WA, USA). The preamplifiers had a double-differential input design, CMRR >100 dB at 65 Hz, gain at 1 kHz $\times$ 20 $\pm$ 1%, input impedance >100,000,000 Ω, and a signal bandwidth from 20 Hz to 3000 Hz (MA-420, Motion Lab System).

EMG signals were transmitted from the first stage pre-amplifier to a second stage receiver unit attached to the back of the subject. From the receiver unit, the signal was hardwired to a 16-bit analog to digital converter. Ankle kinematics, ground reaction force, COP and EMG data were collected simultaneously using a motion analysis program (Qualisys Track Manager ver. 25, Qualisys Motion Capture Systems, Gothenburg, Sweden).

2.3. Procedures

Data were collected at the Jacquelin Perry Musculoskeletal Biomechanics Research Laboratory at the University of Southern California. To control for the influence of footwear, participants were fitted with standardized athletic shoes (New Balance Inc., MA, USA).

Each participant participated in two testing sessions. Hip abductor strength was assessed during the first visit. To minimize the potential effect of muscle fatigue induced from the muscle strength assessment, the biomechanical and balance testing took place on a separate day (average interval = 57 $\pm$ 44 days).

2.3.1. Hip abductor muscle strength testing

Hip abductor strength was evaluated using a previously described weight bearing method. Detailed description as well as the reliability data of this method have been published in a previous study [10]. Briefly, the hip muscle strength testing was performed utilizing a force transducer connected to a non-stretchable fabric belt wrapping around the distal ends of both femurs. Hip abductor muscle strength was assessed with subjects in a weight bearing squat position (50° knee flexion and 30° hip flexion). Participants were instructed to maintain their natural lordotic curvature, and place their feet parallel to each other, shoulder-width apart. Participants were then instructed to push outward against the resistance belt "as hard as possible" for 5 s. A 1 min rest period was given between trials. A total of 3 trials were collected.

2.3.2. Balance testing

The skin surface above TA and PL was shaved and clean with isopropyl alcohol. Electrodes for the PL were placed on the muscle belly inferior from the fibular head at a distance approximately 1/3 of the total muscle length. Electrodes were oriented so they were parallel to the PL muscle fibers. Electrodes for the TA were placed on the muscle belly approximately 3 cm inferior and 2 cm lateral to the tibial tuberosity, and were parallel to the TA muscle fibers. The electrodes and pre-amplifiers were secured to the skin with pre-wrap to minimize movement artifacts during testing. To standardize the level of EMG activity, subjects performed two 5-second maximal voluntary isometric contractions (MVIC) for each muscle. Resisted contraction of the TA (ankle dorsiflexion) was performed against an immobile object in a standing position with the tested ankle in neutral plantarflexion/dorsiflexion, and the hip and knee in 0° of flexion. Resisted contraction of the PL (ankle eversion) was performed with subjects in a seated position with the hip and knee flexed to 90° and the ankle in neutral plantarflexion/dorsiflexion). Resistance to ankle eversion was provided by a non-stretchable strap placed around the forefoot.

Following EMG preparation, opto-reflective markers (spheres, 14 mm diameter) were placed on the following anatomical landmarks: end of second toes, first and fifth metatarsal heads, medial and lateral malleoli, medial and lateral epicondyles of the femurs, greater trochanters, anterior superior iliac spines, iliac crests, and the L5-S1 junction. Tracking cluster markers were attached to neoprene bands secured around the thigh and shank. Additional tracking cluster markers were secured to the heel counter of the shoe. Once all markers were attached, a standing calibration trial was captured. Following collection of the calibration trial, all markers were removed except for the tracking clusters and the markers on the iliac crest and L5-S1 junction.

For the static balance task, participants were instructed to stand on one leg with the hip and knee of the stance leg extended, and the arms folded across the front of the chest (Fig. 1). Only the dominant leg (defined as participant's preferred leg for jumping and landing) was tested. Care was taken to ensure that the stance foot was oriented in line with the fore-aft direction of the lab coordinate system. During this task, participants were instructed to prevent the legs from bracing against each other and to look ahead at an object on the wall about five meters away. Subjects were informed that the goal of the task was to stand as steady as possible for 20 s. A total of 3 static balance trials were collected.

Dynamic postural stability was assessed during a unipedal step-down task. Participants were instructed to lower themselves from an elevated force platform, touch their heel on the lower step, then return to the starting position over a 2-second period (Fig. 2). The