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Does chronic ankle instability influence lower extremity muscle activation of females during landing?

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ABSTRACT

Much remains unclear about how chronic ankle instability (CAI) could affect knee muscle activations and interact with knee biomechanics. Therefore, the purpose of this study was to assess the influence of CAI on the lower extremity muscle activation at the ankle and knee joints during landings on a tilted surface. A surface electromyography system and two force plates were used to collect lower extremity muscle activation of 21 young female individuals with CAI and 21 pair-matched controls during a double-leg landing with test limb landing on the tilted surface. In the pre-landing phase, compared to controls, CAI participants displayed a reduced ankle evertor activation that could place CAI at a high risk of giving way or sprain injury. In the landing phase, an increased tibialis anterior activation of CAI led to increased co-contraction of ankle muscles in the sagittal and frontal plane. A greater ankle muscle co-contraction could increase the ankle stability during landings but may adversely influence the knee muscle activations (e.g., a greater co-contraction ratio of quadriceps to hamstrings). Relevant training programs (e.g., increasing pre-landing peroneal activation, and optimizing activation ratio of quadriceps to hamstrings) may help individuals with CAI improving ankle stability and reduce atypical knee loading during landings.

1. Introduction

Chronic ankle instability (CAI) usually develops after an initial acute ankle sprain [Hertel, 2002; van Rijn et al., 2008]. The common symptoms of CAI include pain, feeling of instability, episodes of giving way and ankle weakness [Hubbard et al., 2007; Mitchell et al., 2008]. These symptoms could be related to changes in tissues (e.g., elongation of the anterior talo-fibular ligament and damage to the cartilage [Hintermann et al., 2002]), deficits in proprioception [Lee and Lin, 2008; Witchalls et al., 2012], and/or reduced ankle muscle strength [Willems et al., 2002]. Consequently, CAI may provoke a more serious condition than initially thought, because it could be related to functional impairment [Simon et al., 2012] and decreased physical activity levels [Hubbard-Turner and Turner, 2015].

Moreover, CAI may be related to knee injuries (e.g., anterior cruciate ligament (ACL) injury: Kramer et al., 2007; Söderman et al., 2001) due to the alterations of lower extremity biomechanics during high-impact movements [Gribble and Robinson, 2010; Gribble and Robinson, 2009; Terada et al., 2014]. In a previous study that focused on knee biomechanics of CAI [Li et al., 2017], the researchers demonstrated that in comparison with healthy controls, reduced ankle energy dissipation of CAI individuals resulted in a greater eccentric knee extensor moment and work and increased internal rotation moments. These altered knee kinetics could be related to the mechanisms of ACL strain [DeMorat et al., 2004; Fleming et al., 2001].

One possible reason for altered biomechanics is differences in neuromuscular control. Neuromuscular differences of ankle muscles for CAI group compared to controls have been observed, though conflicting findings exist. First, CAI group has demonstrated atypical muscle activation magnitudes during various weight-bearing movements [Delahunt et al., 2007; Lin et al., 2011; Hopkins et al., 2012]. Increased tibialis anterior activation was found in CAI during the stance phase of walking [Louwerens et al., 1995; Hopkins et al., 2012]. Among CAI studies utilizing landing activities, researchers observed lower pre-landing peroneal activations [Caulfield et al., 2004; Delahunt et al., 2006; Suda et al., 2009] in CAI groups compared with controls.

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However, for the landing phase, conflicting results among studies were reported for soleus [Brown et al., 2004; Delahunt et al., 2007], peroneal [Lin et al., 2011; Suda et al., 2009] and tibialis anterior activation [Delahunt et al., 2007; Suda et al., 2009]. Second, differences in neuromuscular reaction time between CAI participants and controls have been investigated. A trapdoor device was commonly used to measure the reaction time of the peroneal activation when the ankle was suddenly inverted. A greater latency of peroneal onset was found for CAI [Karlsson and Andreasson, 1992; Konradsen and Ravn, 1990]; however, others did not observe timing differences [Vaes et al., 2002]. Third, reduced eversion muscle strength measured by an isokinetic dynamometer has been reported in CAI participants [Rottigni and Hopper, 1991; Willems et al., 2002].

In this context, the ankle neuromuscular alterations of CAI could influence the ankle biomechanics and further interact with the knee biomechanics and muscle activations, because the function and dysfunction of one joint can affect the function of the adjacent proximal joint of the kinetic chain [Kaminski and Hartsell, 2002; Terada et al., 2014, 2013]. Given that previous studies have found altered neuromuscular control at the ankle joint, it could be that those with CAI may also be demonstrating decreased energy dissipation at the ankle [Li et al., 2017]. Consequently, greater knee extensor activation could be needed to achieve an increase in energy dissipation at the knee during landing.

However, much remains unclear about how CAI could affect knee muscle activations and further influence knee biomechanics and ACL loading. To our knowledge, few studies investigated knee muscle activities of CAI individuals [Delahunt et al., 2007, 2006]. One study reported greater rectus femoris activity in the pre-landing and landing phases of lateral hopping, but did not provide any detailed explanation for this observation.

Therefore, the purpose of the present study was to assess the influence of CAI on lower extremity muscle activation at the ankle and knee joints during landings onto a tilted surface. We utilized a tilted surface (25°) because landing on an inverted surface has been suggested as a more demanding situation for individuals with CAI [Chen et al., 2012] and could result in altered CAI muscle activation patterns. We hypothesized that individuals with CAI would exhibit some different ankle and knee muscle activation patterns compared to healthy controls.

2. Methods

2.1. Participants

Twenty-one females with CAI participated in the study. The inclusionary and exclusionary criteria established by the International Ankle Consortium were utilized for identifying those with CAI (Gribble et al., 2013). Twenty-one healthy control participants that were pair-matched with the CAI participants, based on their gender, height (± 2.5 cm), body mass (± 5 kg), and physical activity levels (± 2 h/week of similar intensity of physical activity) were recruited. All participants were healthy, without having had a serious lower extremity injury or dysfunction, and had experience in landing-related sports (e.g., basketball, volleyball, soccer, etc.). The participants' characteristics are presented in Table 1.

2.2. Instrumentation

A wireless surface electromyography (EMG) system (sampling rate = 2040 Hz, CMRR > 80 dB; Delsys Trigno™ System, Natick, MA, USA) was used to measure the EMG of lower extremity muscles. Six surface EMG electrodes (37 mm by 27 mm) were attached to the muscle belly of tibialis anterior, gastrocnemius lateralis, peroneus longus, rectus femoris, vastus lateralis and biceps femoris of the test limb (see details in Section 2.3). We used gastrocnemius lateralis rather than medialis to calculate the ankle muscle co-contraction index (described in Section 2.4) suggested by previous studies [Lin et al., 2011; Suda et al., 2009].

Table 1
Demographical Data, Ankle Instability Scores and MVIC Moments (mean ± SD) of the Participants.

Variable	Control	CAI	*p*-value
Sample size	21	21	NA
Body mass (kg)	64.4 ± 11.9	64.4 ± 12.4	0.809
Height (cm)	165 ± 6	164 ± 6	0.774
Age (yr)	21 ± 2	21 ± 2	0.895
CAIT score	29.5 ± 0.9	19.3 ± 6.0	0.000*
IdFAI score	1.3 ± 2.1	22.2 ± 9.2	0.000*
Ankle dorsiflexor (Nm)	18.0 ± 5.3	14.1 ± 3.5	0.003*
Ankle plantarflexor (Nm)	22.2 ± 10.6	17.1 ± 4.7	0.054
Ankle evertor (Nm)	6.3 ± 2.8	4.7 ± 1.7	0.007*
Knee extensor (Nm)	95.7 ± 39.7	76.5 ± 33.4	0.156
Knee flexor (Nm)	33.0 ± 8.2	31.7 ± 8.3	0.590

Note: MVIC = maximum voluntary isometric contraction; CAI = chronic ankle instability participants; CAIT = Cumberland Ankle Instability Tool; IdFAI = Identification of Functional Ankle Instability.
* Indicates statistical significant ($p < .05$).

To define the instant of initial contact and determine landing trial eligibility, vertical ground reaction forces (GRF) of each foot were collected using one force plate per foot (Bertec Corp., Columbus, OH, USA) at 2040 Hz. As shown in Fig. 1, one side of the force plate was tilted downwards at 25° in the lateral direction and the other force plate flat such that the centers were at the same height. EMG and ground reaction force data were captured and synchronized using the Vicon Nexus™ 2.2 software (Vicon Motion Systems Ltd., Oxford, UK).

2.3. Procedures

The study was approved by the Institutional Review Board (IRB approval number: STUDY00002113), and all participants provided informed consent before data collection. Participants completed the Identification of Functional Ankle Instability (IdFAI) questionnaire and the Cumberland Ankle Instability Tool (CAIT) to determine the eligibility of the CAI participants and the level of chronic ankle instability of each limb. The limb with less ankle stability (higher IdFAI and lower CAIT scores) was chosen as the test limb for CAI participants. For a pair-matched control, the test limb was the limb that had the same limb dominance as the corresponding CAI test limb. The dominant limb was classified as the limb that the participants verbally chose as their 'kicking' leg [Yeow et al., 2010].

A proper skin preparation was carried out prior to electrode placement, which was done by shaving any hair on the skin, then the skin was wiped with isopropyl alcohol to remove oils and surface residuals. The electrodes were placed on the participants at the locations suggested in Cram et al. (1998) guidelines.

To normalize and compare the EMG data across participants, maximum voluntary isometric contraction (MVIC) tests were conducted. For all MVIC tests, the participants were placed in the test position and a hand-held dynamometer (FCE Series Medical Dynamometer, AMETEK, Inc., Berwyn, PA, USA) was used to create isometric resistance and obtain the MVIC moments. The same investigator performed all MVIC testing by holding the dynamometer against the participants' testing limb as resistance was provided. A five-second EMG signal (one trial) was captured for each muscle MVIC test [Dai et al., 2012] and a 30-s break was taken between the tests. A detailed MVIC test procedure and body position were described in Table 2.

After five minutes of treadmill jogging warm-up, the participants practiced performing the drop landing task, then performed ten acceptable drop landing trials. For a given landing trial, the participant stood on a box 30 cm above the foot landing targets, then stepped