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Effect of neck flexor muscle activation on impact velocity of the head during backward falls in young adults

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ABSTRACT

Falls are a common cause of traumatic brain injuries (TBI) across the lifespan. A proposed but untested hypothesis is that neck muscle activation influences impact severity and risk for TBI during a fall. We conducted backward falling experiments to test whether activation of the neck flexor muscles facilitates the avoidance of head impact, and reduces impact velocity if the head contacts the ground. Young adults (n = 8) fell from standing onto a 30 cm thick gymnastics mat while wearing a helmet. Participants were instructed to fall backward and (a) prevent their head from impacting the mat ("no head impact" trials); (b) allow their head to impact the mat, but with minimal impact severity ("soft impact" trials); and (c) allow their head to impact the mat, while inhibiting efforts to reduce impact severity ("hard impact" trials).

Trial type associated with peak magnitude of electromyographic activity of the sternocleidomastoid (SCM) muscles ($p < 0.017$), and with the vertical and horizontal velocity of the head at impact ($p < 0.001$). Peak SCM activations, expressed as percent maximal voluntary isometric contraction (%MVIC), averaged 75.3, 67.5, and 44.5%MVIC in "no head impact", "soft impact", and "hard impact" trials, respectively. When compared to "soft impact" trials, vertical impact velocities in "hard impact" trials averaged 87% greater (3.23 versus 1.73 m/s) and horizontal velocities averaged 83% greater (2.74 versus 1.50 m/s). For every 10% increase in SCM %MVIC, vertical impact velocity decreased 0.24 m/s and horizontal velocity decreased 0.22 m/s.

We conclude that SCM activation contributes to the prevention and modulation of head impact severity during backward falls.

1. Introduction

Falls are the leading cause of deaths and hospitalizations due to traumatic brain injury (TBI). In the United States in 2013, falls accounted for more than twice as many TBI-related deaths and nearly four times as many TBI-related hospitalizations as the second greatest cause, motor vehicle accidents (Taylor et al., 2017). Furthermore, the age-adjusted rate of TBI in older adults has doubled over the past decade (Bouras et al., 2007; Harvey and Close, 2012; Saari et al., 2007).

Risk for TBI during a fall depends on whether impact occurs to the head, and how severely the head strikes the environment. Fortunately, in healthy people, falls from standing height rarely result in head impact (Feldman and Robinovitch, 2007; Hsiao and Robinovitch, 1998). The low chance for head impact is due presumably to neuromuscular protective responses that keep the head off the ground. In forward falls, we tend to rely on upper limb fall arrest to prevent head impact (Dietz et al., 1981; Hsiao and Robinovitch, 1998). In backward and sideways falls, we tend to supplement upper limb fall arrest by activating the core and neck muscle to keep the head off the ground (Feldman and Robinovitch, 2007; Robinovitch et al., 2004).

When compared to young adults, older adults have an increased risk for head impact during falls. Previous studies reported that head impact occurred in 37% of falls in long-term care homes, based on analysis on video footage of 227 falls experienced by older adults of mean age 78 (Robinovitch et al., 2013; Yang et al., 2017). Over 31% of cases were due to backward falls. Others have reported, based on review of emergency department records, that backward falls were 4-fold more likely than forward falls to cause moderate/severe traumatic brain injury (Hwang et al., 2015). Improved understanding is required of the biomechanical demands of protective responses for avoiding head impact in falls, and the effects of age-related changes in muscle strength, joint range of motion, reaction time, and muscle activation strategy.

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Several authors have proposed that risk for TBI in sports is increased by weak neck muscle strength (Benson et al., 2013; Mihalik et al., 2011), via the effect of the cervical muscles in altering head linear and rotational kinematics, and resultant brain loading during impact events (i.e., falls or player-on-player contacts). This is an intriguing possibility, since neck strength can be enhanced through resistance training (Mansell et al., 2005). However, in their 2013 review, Benson et al. highlighted that no clinical studies have examined the association between neck strength and TBI, and that no biomechanical studies had measured muscle activations during head impact to support a causal effect (Benson et al., 2013).

For falls, the challenge to date has been devising experimental approaches with living human participants that provide safe but accurate simulations of falls that result in head impact. In laboratory-based falling studies with living humans, participants either avoided impacting the head when falling on a mat (Feldman and Robinovitch, 2007; Hsiao and Robinovitch, 1998), or were prevented from doing so with a fall restraint harness (Ding and Yang, 2016; Liu et al., 2017; Liu and Lockhart, 2014; Lockhart et al., 2003; Rosenblatt and Grabiner, 2012; Troy and Grabiner, 2007). Researchers have studied the biomechanics of head impact in simulated falls with crash-test dummies (Caccese et al., 2014; Hajiaghamemar et al., 2015; O'Riordain et al., 2003) or cadavers (Hodgson and Thomas, 1971). These studies have provided essential measures of the peak accelerations of the head during impact with a hard surface (via accelerometers secured to the skull) – data that cannot be acquired ethically from falls with living humans. However, crash-test dummies and cadavers cannot be used to examine the complex effects of muscle activation on head impact severity during falls. A study reported head impact velocities based on digitization of video footage of real-life falls in older adults (Choi et al., 2015), but did not measure neck muscle activity.

Against this background, our goals in the current study were: (1) to measure the velocity of the head at impact during backward falls experienced by young adults; and (2) to test the hypothesis that magnitude of activation of the neck flexor muscles associates with the head impact velocity during falls.

2. Methods

2.1. Participants

Healthy young men (n = 3) and women (n = 5) aged between 19 and 35 years participated. Participants' age, body weight and height averaged 26.5 years (SD 5.3), 63.9 kg (SD 8.1), and 170.5 cm (SD 7.2), respectively. Individuals were excluded if they had recent injuries to the head (i.e., concussion) or musculoskeletal conditions causing joint or soft tissue pain. Participants were recruited through flyers posted at Chapman University Irvine campus. The experimental protocol was approved by the Institutional Review Board at Chapman University, and all participants provided written informed consent.

2.2. Protocol

The experimental protocol was designed based on previously-reported laboratory-based falling experiments, where participants observed and mimicked the falling behavior of older adults in real-life falls captured on video (Aziz et al., 2014; Choi et al., 2015; Robinovitch et al., 2013). During the trials, participants fell backward from standing onto a 30 cm thick gymnastics mat (Fig. 1b). Each participant viewed videos of backward falls involving head impact experienced by older adults in long-term care (Fig. 1a) and practiced until they feel comfortable in mimicking the falls on the mat (no data was collected during these practice trials). The stiffness of the mat was similar to that used during athletic gymnastics. Furthermore, participants wore helmets and wrist guards for additional protection against potential injuries. All falls were self-initiated and not produced by a biomechanical perturbation to balance. There is a considerable range in the types of perturbations to balance (both internal and external in nature) that results in real-life falls, and incorrect weight shifting (which a self-induced fall mimics) is the most common type of imbalance event leading to falls among older adults in long-term care (Robinovitch et al., 2013).

The protocol included falling trials that were designed to result in head impact. In particular, each participant experienced three series of trials, each of which involved unique instructions. In "no head impact" trials, participants were instructed to "fall backward and prevent your head from impacting the mat". In "soft impact" trials, participants were instructed to "fall backward and allow your head to impact the mat, but with minimal impact severity". In "hard impact" trials, the instruction was to "fall backward and allow your head to impact the mat, while inhibiting your efforts to reduce head impact severity". The order of presentation of the trials was randomized, and three repeated trials were acquired in each condition.

Our intent was to study falls where participants did not utilize their upper limbs to arrest the fall (and thereby isolate the protective effect of neck muscle activation), but to otherwise measure the natural falling behavior of our participants. Accordingly, additional instructions to participants for all trials were to try not to use their arms and hands to break the fall, and to "fall backward in a manner similar to what you observed in the video of older adult's fall".

In each trial, a ten-camera motion analysis system (Raptor system, Motion Analysis Corp, Santa Rosa, CA) was used to measure (at 250 Hz) the time-varying three-dimensional positions of reflective markers placed on the top of the helmet, and on the greater trochanters. As described below (in Data analysis) these markers were used to analyze head impact kinematics and to define fall initiation, respectively. Additional markers not used in the current analysis were placed at the spinous process of the 7th cervical vertebra and bilaterally on the feet, ankles, knees, wrists, elbows and shoulders. Muscle activations of the left and right sternocleidomastoid (SCM) were measured through surface electromyography (EMG) at 2000 Hz (DTS system, Noraxon, Scottsdale, AZ). We used a self-adhesive, figure 8-shaped, Noraxon dual Ag/AgCl snap electrode (dimension of adhesive 4 × 2.2 cm, inter-electrode distance 1.75 cm). The SCM muscles were first identified when participants flexed and rotated their head against resistance from the examiner, and the electrode was placed at the belly of the SCM (half way down from the mastoid process of temporal bone) with the long axis of the electrode approximately parallel to the SCM muscle fiber direction.

Just prior to the falling trials, we conducted manual muscle testing to measure EMG activations during maximum voluntary isometric contraction (MVIC) of the right and left side SCM muscles. Only one trial was conducted due to fatigue and safety considerations. During these measures, participants lay supine with their arms at their sides and head supported on a table. Participants were then instructed to flex and rotate their neck 40 degrees, respectively, and to then maximally resist the pressure applied by the investigator's hand to the temple of the head, and hold for 2 s. The specific instruction was "do not let me push your head down".

2.3. Data analysis

Outcome variables from each falling trial included the vertical and horizontal velocities of the head at the instant of head impact (our measures of head impact severity), and the EMG onset time, and the timing and magnitude of peak EMG activity in the SCM muscles, expressed as %MVIC. We used the velocity of the head at the initial instant of ground contact as our measure of impact severity (as opposed to subsequent peak accelerations of the head during impact), since impact velocity is not influenced by the cushioning of the head provided by the gymnasium mattress. In trials that resulted in head impact, EMG analysis was conducted over the time period between fall initiation and head impact. For trials that did not result in head impact,