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The effect of asymmetrical body orientation during simulated forward falls on the distal upper extremity impact response of healthy people

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ABSTRACT

The occurrence of distal upper extremity injuries resulting from forward falls (approximately 165,000 per year) has remained relatively constant for over 20 years. Previous work has provided valuable insight into fall arrest strategies, but only symmetric falls in body postures that do not represent actual fall scenarios closely have been evaluated. This study quantified the effect of asymmetric loading and body postures on distal upper extremity response to simulated forward falls. Twenty participants were suspended from the Propelled Upper Limb fall ARest Impact System (PULARIS) in different torso and leg postures relative to the ground and to the sagittal plane (0°, 30° and 45°). When released from PULARIS (hands 10 cm above surface, velocity 1 m/s), participants landed on two force platforms, one for each hand. Right forearm impact response was measured with distal (radial styloid) and proximal (olecranon) tri-axial accelerometers and bipolar EMG from seven muscles. Overall, the relative height of the torso and legs had little effect on the forces, or forearm response variables. Muscle activation patterns consistently increased from the start to the peak activation levels after impact for all muscles, followed by a rapid decline after peak. The impact forces and accelerations suggest that the distal upper extremity is loaded more medial-laterally during asymmetric falls than symmetric falls. Altering the direction of the impact force in this way (volar-dorsal to medial-lateral) may help reduce distal extremity injuries caused when landing occurs symmetrically in the sagittal plane as it has been shown that volar-dorsal forces increase the risk of injury.

1. Introduction

Fractures to the distal upper extremity are now among the most common traumatic injuries experienced by the adult population (van Staa et al., 2001; Shauver et al., 2011), significantly contributing to the more than $12.5 Billion associated with accidental/orthopaedic injuries (Canadian Orthopaedic Foundation, 2015). Distal upper extremity injuries frequently occur due to impacts with the ground following a fall onto the outstretched arm in an attempt to arrest the momentum of the falling body. It has been estimated that approximately 95% of fractures of the proximal humerus, elbow, and wrist are a result of a fall, with wrist fractures alone costing more than an estimated US $500 Million annually (Burge et al., 2005).

A variety of impact methods have been evaluated in the literature which allow researchers to study the response of the upper extremities of living people, without increasing the risk of injury to the participants appreciably. However, many of the previously reported methods which have been used to study forward fall-induced impacts, released participants from a state of zero horizontal velocity (Chiu and Robinovitch, 1998; Chou et al., 2001; DeGoede and Ashton-Miller, 2002) an initial condition that misrepresents the pre-fall kinematics commonly seen during a fall *in vivo*. To overcome this limitation, Burkhart et al. (2012) developed a Propelled Upper Limb Fall ARrest Impact System (PULARIS) that provided a pre-fall horizontal velocity to participants before being released and impacting floor-mounted force platforms. Using PULARIS, Burkhart and Andrews (2013) found that young healthy individuals were capable of selecting a fall arrest strategy that minimized the impact forces applied to the distal upper extremity. However, this investigation was limited in that it only analysed falls that occurred in the sagittal plane. While falls most frequently occur in the forward direction (Nevitt and Cummings, 1994), they are likely to occur in an asymmetric manner (Palvanen et al., 2000). Troy and Grabiner (2007a, 2007b) were able to show slight differences in loading of the distal upper extremity when the hands impacted the ground in an asymmetric manner; this study, like

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those described above, was limited by its state of initial zero horizontal velocity.

Burkhart and Andrews (2013) also found that muscle activation levels from six upper extremity muscles (biceps brachii, brachioradialis, triceps brachii, anconeus, flexor carpi radialis, and extensor carpi radialis) demonstrated a preparatory muscle activation response prior to impact, showing that peak muscle activation levels occurred prior to the peak impact loads. It was theorized (Burkhart and Andrews, 2013) that the ability to contract the muscles prior to impact may allow the faller to stabilize the joints in an attempt to arrest the fall in a relatively controlled manner (*i.e.*, energy absorption through changing joint angles); however, they did not investigate asymmetric falls in multiple planes of motion.

Therefore, the purpose of the current study was to investigate the effects of sagittal and horizontal plane asymmetric forward falls on the impact kinetics, and muscle activation patterns of the upper extremity using a fall simulation method that enables pre-impact kinematics that more closely reflect those experienced during a forward fall.

2. Methods

2.1. Participants

The Propelled Upper Limb fall ARrest Impact System (PULARIS) previously described by Burkhart et al. (2012) was used to simulate the impact phase of symmetric and asymmetric forward falls of 9 male (mean (SD) height: 1.74 (0.11) m; body mass: 76.5 (14.2) kg) and 11 female (mean (SD) height: 1.65 (0.06) m; body mass: 64.6 (9.7) kg) university-aged participants. Participants had no history of upper extremity injuries, which was verbally confirmed by the participants. Participants provided written informed consent prior to testing and all procedures were approved by the University of Windsor's Research Ethics Board.

2.2. Instrumentation

Six pairs of Kendall Ag/Ag-Cl rectangular (23 mm × 33 mm) surface electrodes (Tyco Healthcare Group LP, Mansfield, MA; ES40076-H59P) were placed over the muscle bellies of seven upper extremity muscles in the direction of their lines of action (2 cm inter-electrode distance). Muscle activation levels were collected from the Biceps Brachii (BB), Brachioradialis (BR), Triceps Brachii (lateral head) (TrLa), Anconeus (AN), Extensor Carpi Ulnaris (ECU), Flexor Carpi Ulnaris (FCU), and Flexor Digitorum Superficialis (FDS). Details regarding the collection of the maximal voluntary exertions (MVEs) can be found in Burkhart and Andrews (2013). The EMG signals were differentially amplified (±2.5 V; AMT-8 Bortec Calgary Canada; Bandwidth 10 –1000 Hz, CMRR = 115 dB at 60 Hz, input impedance = 10 GΩ), full wave rectified and filtered with a dual pass 2nd order Butterworth filter (cut-off frequency of 2.5 Hz (Burkhart and Andrews, 2013)) (Table 1). While EMG signals are commonly normalized to the participant's MVE in the literature, pilot testing revealed a propensity for the muscle activation during impact to be greater than 100% MVE. This is likely due to the very dynamic nature of the impacts studied compared to the static MVE protocol typically used (Burkhart and Andrews, 2013). To accommodate this, the EMG signals were also normalized to a resting muscle activation level (% EMG_{rest}) and a baseline level collected as the participants were propelled towards the impact surface (%EMG_{base}) (see description below).

The PULARIS was used to propel the participants towards three tri-axial (F_x: medio-lateral; F_y: anterior-posterior; F_z: inferior-superior) strain gauge force platforms (Advanced Mechanical Technology Inc. Watertown, MA, model # OR6-7; 2200 N capacity in the x and y axis and 4250 in the z direction; natural frequency of 1 kHz) which were rigidly mounted to the laboratory floor (Fig. 1). Two tri-axial accelerometers (MMA1213D and MMA3201D, Freescale Semiconductor, Inc, Ottawa, ON, Canada; range of ±50 G and ±40 G, respectively) were firmly secured onto the skin overlying the right radial styloid and the right olecranon process using double sided tape. The transducers were pressed snuggly to the underlying bone with a 45 N load applied with a Velcro™ strap (Burkhart and Andrews, 2010a, 2010b). Accelerations were measured in the axial (parallel with the long axis of the forearm), off-axis (perpendicular to the long axis of the forearm in the volar-dorsal directions) and the medial-lateral directions (Burkhart and Andrews, 2013). Force and acceleration data were filtered with a dual pass, 4th order Butterworth filter and the cut-off frequencies for the forces and accelerations were determined separately for each data channel by residual analysis (Burkhart et al., 2011) (Table 1).

2.3. Data collection protocol

Following instrumentation, the participants were instructed to lay prone on a torso harness which was positioned directly under and in-line with PULARIS. A strap was secured around the torso (mid-sternum), with another placed around the legs (just below the knees). These were subsequently attached to two separate solenoid-controlled quick releases. The quick releases themselves were connected to the lower tracking of the PULARIS system (Burkhart et al., 2012; Burkhart and Andrews, 2013) (Fig. 1) through the use of inverted steel c-channel tracking that was mounted perpendicular to the bottom of PULARIS.

Once the participants in the harness were attached to the quick releases, an automatic hoist was used to raise the PULARIS so that the hands of the participants were located approximately 0.10 m from the force platforms when the shoulders were in 45° of flexion. A section of foam (0.36 m × 0.36 m × 0.10 m) was located at the front edge of the force platforms (Fig. 1) and used as a guide to ensure participants kept their hands at a consistent height of 10 cm prior to release. The release location (*i.e.*, the horizontal location at which the participant would be dropped) was determined as the position where approximately half of the participants' hands were over the front edge of the force platform directly in front of them. From this location, participants were moved backwards 1.9 m to achieve a final horizontal velocity of 1.0 m/s (Burkhart and Andrews, 2013). This velocity was used as it represented the limits of the PULARIS but also ensured a safe fall for the participants. Furthermore, the ratio of horizontal:vertical hand and hip velocity agree well with standing height falls (Burkhart et al., 2012). Upon confirming the appropriate postures, the PULARIS was propelled forward and the participants were dropped (the torso quick release was set to drop 150 ms prior to the legs) as they passed the release location. Participants were instructed to adopt a straight arm elbow posture at impact; this position was considered a worse-case scenario fall (DeGoede and Ashton-Miller, 2002; Burkhart and Andrews, 2010a, 2013). Participants were instructed to maintain all postures throughout the fall and impact.

Each participant experienced three repetitions of three different horizontal plane torso angles (0° (symmetric); 30°; and 45°) (Fig. 2a) and two different leg to torso height ratios (2:1 and 1:1) (Fig. 2b), resulting in 18 impacts per participant. The asymmetric falls were achieved by altering the medio-lateral position of the leg quick releases within the perpendicular steel track that was mounted beneath the inferior steel track of PULARIS (Fig. 1). The leg to torso height ratios were achieved by making adjustments to the leg and torso straps that supported these segments from