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Neck posture and muscle activity are different when upside down: A human volunteer study

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ABSTRACT

Rollover crashes are dynamic and complex events in which head impacts with the roof can cause catastrophic neck injuries. *Ex vivo* and computational models are valuable in understanding, and ultimately preventing, these injuries. Although neck posture and muscle activity influence the resulting injury, there is currently no *in vivo* data describing these parameters immediately prior to a head-first impact. The specific objectives of this study were to determine the *in vivo* neck vertebral alignment and muscle activation levels when upside down, a condition that occurs during a rollover. Eleven human subjects (6F, 5M) were tested while seated upright and inverted in a custom-built apparatus. Vertebral alignment was measured using fluoroscopy and muscle activity was recorded using surface and indwelling electrodes in eight superficial and deep neck muscles. *In vivo* vertebral alignment and muscle activation levels differed between the upright and inverted conditions. When inverted and relaxed, the neck was more lordotic, C1 was aligned posterior to C7, the Frankfort plane was extended, and the activity of six muscles increased compared to upright and relaxed. When inverted subjects were asked to look forward to eliminate head extension, flexor muscle activity increased, C7 was more flexed, and C1 was aligned anterior to C7 versus upright and relaxed. Combined with the large inter-subject variability observed, these findings indicate that cadaveric or computational models designed to study injuries and prevention devices while inverted need to consider a variety of postures and muscle conditions to be relevant to the *in vivo* situation.

1. Introduction

Rollovers are dynamic and complex events in which serious neck injuries, such as vertebral fractures and spinal cord injuries, can be caused by head contact with the vehicle interior (Bedewi et al., 2003; Hu et al., 2005). A common mechanism of neck injury in rollovers occurs when the vehicle roof contacts the ground and the upside down occupant continues head-first into the roof at their original velocity (Moffatt et al., 2003b; Raddin et al., 2009). Cadaver studies have shown that the axial force that develops in the neck from such inertial loading generates a spectrum of spine fractures and other injuries (Nightingale et al., 1997; Nusholtz et al., 1983; Saari et al., 2011; Yoganandan et al., 1986). Heterogeneous neck injury patterns also occur in real-world rollovers (Hu et al., 2005); yet, there is some disparity between the neck injuries observed in rollover accidents and cadaver experiments (Foster et al., 2012).

A number of studies have shown that the overall alignment of the head, neck, and torso have an effect on the neck injury mechanism observed (Alem et al., 1984; Hodgson and Thomas, 1980; Maiman et al., 2002; Nightingale et al., 1997; Nusholtz et al., 1981). Loss of the cervical lordosis (e.g. by flexing the spine forward) can increase susceptibility to compressive loads and thus burst fracture patterns of spinal column injury (Pintar et al., 1995a, 1989; Yoganandan et al., 1990), often with concomitant spinal cord injury (Chang et al., 1994; Pintar et al., 1995b). Although posture is known to influence the neck injury sustained in a head-first impact, cadaveric specimens are often tested upright or inverted

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using the neck's upright resting (lordotic) posture. In fact, the actual configuration of the cervical spine immediately before an inverted head-first impact remains unknown.

Ex vivo neck models of head-first impact lack neuromuscular control and the associated postural stability. Some cadaver tests use osteo-ligamentous spines, whereas others have left the muscle tissues intact to capture a passive muscle response (Alem et al., 1984; Maiman et al., 1983; Nusholtz et al., 1983). Overall muscle tone has been simulated with cables and springs, typically to restrain the head and maintain a particular cervical posture for axial testing (Ivancic, 2012; Saari et al., 2011; Yoganandan et al., 1990). Computational modeling has suggested that muscle activation influences neck injury risk in axial loading (Brolin et al., 2009) and may increase the neck fracture risk in rollovers (Hu et al., 2008). However, as with the cadaver experiments discussed above, muscle activation schemes are typically not based on human subject data. Indeed, the level of muscle contraction prior to an impact is unknown, and there is little understanding of how this muscle control alters the posture and stiffness of the cervical spine during a rollover (Lai et al., 2005).

The general assumption of existing head-first cervical spine impact models, likely due to the paucity of data, is that the upright and inverted necks have the same posture. However, if inversion influences neck posture, the initial conditions currently used in modeling head-first injury may not adequately replicate rollover conditions. Muscle activation levels while inverted may also be important inputs to these cadaveric and computational models. Therefore, the aim of this study was to measure and compare the neuromuscular activity and cervical spine posture in upside down configurations versus a neutral resting upright configuration. We hypothesized that, compared to the upright posture, inversion would change the head angle, neck curvature, and neck eccentricity (horizontal distance between C1 and C7), and increase neuromuscular activity; and further that these differences would depend on whether subjects were relaxed or maintaining a level gaze.

2. Methods

2.1. Subjects

Eleven asymptomatic subjects (6F, 5M, 33 ± 7 years, average ± standard deviation) participated in this study. Subjects were excluded if they had a history of neck injury or pain, or known muscle, nerve, or balance problems. The average group height was 172.9 ± 6.7 cm, their weight was 70.0 ± 15.7 kg, and their neck and head circumferences were 34.3 ± 4.2 cm and 56.4 ± 3.2 cm, respectively. One subject did not participate in the fluoroscopy portion of the experiment and another subject did not participate in the EMG portion. Subjects gave their written informed consent and the study was approved by the University of British Columbia's Clinical Research Ethics Board.

2.2. Conditions

Subjects were seated in a bucket seat (36 Series—Intermediate 20° Layback, Kirkey Racing Fabrication Inc., St. Andrew's West, ON), secured with a 75-mm wide 5-point harness (RCI Racer's Choice Inc., Tyler, TX) and held statically in one upright and two inverted postures (Fig. 1). Subjects were instructed to adopt three postures: (1) upright and relaxed (U-R), (2) inverted and relaxed (I-R), and (3) inverted and looking forward (I-F). The inversion trials were not randomized (I-R first, then I-F) and subjects were given about 30 s to experience the sensation of inversion before the experiment started.

2.3. Cervical spine posture

A fluoroscopic C-arm (OEC 9400, GE) was used to capture sagittal plane images of the cervical vertebra at 30 Hz. Each subject was exposed to a total of 12 s of fluoroscopy and on average a total effective dose of 0.0067 mSv for this experiment (a typical transatlantic flight is 0.03–0.045 mSv (De Beer, 2002)). Fluoroscopic images were corrected for distortion (Brainerd et al., 2010) and vertebral motions were tracked in the images using an automatic tracking algorithm based on Bifulco et al. (2009). Briefly, the image gradient was calculated and normalized cross-correlation was used to match a template of each vertebra to the fluoroscopic image of interest. Anatomical landmarks were identified for each vertebral body (Fig. 2) in the template images and the tracking algorithm was used to find the angle and displacements of the vertebral bodies and landmarks. Accuracy studies using cadaveric cervical vertebrae (unpublished observations) yielded maximum root mean squared (RMS) errors of 0.6° and 0.2 mm (Bifulco et al. (2009) reported values of 0.2° and 0.3 mm in the lumbar spine). Several images were tracked to ensure the subject remained static and one image near the end of each trial (at the mid-point of the EMG window—see below) was used for the statistical comparisons.

Cervical spine posture was quantified with three measures: the eccentricity (X_{ECC}), the curvature index (CI), and the C7 angle (Θ_{C7}). The eccentricity was the horizontal distance between the mid-superior point of C1 and the mid-inferior point of C7 vertebral bodies (positive = C1 anterior to C7) (Fig. 3). The curvature index was defined as the percentage difference between the arc and chord lengths of the spine (Fig. 3) (Klinich et al., 2004), where the arc length was the sum of the straight line segments passing through the superior and inferior mid-points of adjacent vertebral bodies from C2 to C7 and the chord length was distance from the top of the dens to the inferior mid-point of the C7 vertebral body. A higher curvature index indicates more curvature. The angle of the C7 vertebral body was with respect to the true horizontal and was an indication of the overall neck orientation (Fig. 3, positive = flexion).

2.4. Head orientation

Head orientation was measured using a bead array (six-2 mm beads) that was fastened to the head but projected into the field of view of the fluoroscope (Fig. 1). The Frankfort plane angle (Θ_F, unpublished RMS error = 0.74°) was defined as a line from the mid-point of the right and left tragus and the mid-point of the inferior margins of the right and left orbits and was measured relative to the true horizontal (Fig. 3) (positive = flexion). Prior to data collection, the beads and Frankfort plane landmarks were digitized using a FaroArm (B08-02, Lake Mary, FL) and the initial angle offset between the beads and the Frankfort plane was established. Angular motion of the Frankfort plane was thus equivalent to subsequent angular motions of the beads.

2.5. Electromyography

EMG activity was measured using both surface and indwelling electrodes. Surface electrodes (Ambu Blue Sensor, Ambu A/S, Ballerup, Denmark) were placed on the skin superficial to the left sternohyoid (STH) muscle (Siegmund et al., 2007). Indwelling fine-wire electrodes were inserted into the left sternocleidomastoid (SCM), trapezius (Trap), levator scapulae (LS), splenius capitis (SPL), semispinalis capitis (SsCap), semispinalis cervicis (SsCerv), and multifidus (MultC4) muscles. The indwelling electrodes consisted of pairs of PFA-coated Stainless Steel, 0.0055 in. diameter wire (A-M Systems, Inc., Sequim, WA) with 1 mm exposed wire and 2–3 mm spacing between the exposed ends of the two wires. The wires were inserted under ultrasound guidance into the center of each muscle belly at the C4/5 level (Blouin et al., 2007). Since the cross-section of the trapezius muscle at C4/C5 was typically only a few millimeters thick, these wires were inserted near the C5/C6 level. The EMG signals were amplified, band-pass filtered (wire: 50–1000 Hz; surface: 30–1000 Hz) and sampled at 2 kHz. The RMS of the muscle activity was calculated for a 500 ms window during the trial. A trigger pulse (emitted by the image acquisition card) enabled the fluoroscopic image of interest to be synchronized with the middle of this 500 ms window. Each muscle's activation was normalized to the maximum RMS activity recorded for that muscle in a maximum voluntary contraction (MVC).

2.6. Maximum voluntary contractions

Seated subjects were secured to a rigid backboard while wearing a snug skateboard helmet. The helmet was attached to a 6-axis load cell (45E15A-U760, JR3, Inc., Woodland, CA, nominal horizontal accuracy: ± 2.5 N) above the subject's head (Fig. 1). Isometric MVCs were performed with verbal encouragements (Gandevia, 2001) and real-time visual feedback of force/moment magnitude and direction. With a neutral head posture, MVCs were executed in seven directions: flexion, extension, left lateral bending, two 45° oblique combinations (flexion/left lateral bending, extension/left lateral bending), and right and left axial rotations. Each contraction was 3 s long and repeated twice. The EMG signals were amplified and filtered as described above and both the load cell and EMG were sampled at 2 kHz. A 500 ms window centered on the maximum force was used to calculate the RMS for each muscle's EMG in all seven directions (Fig. 4). The maximum RMS value for each muscle, regardless of direction, was used for normalization.