

# **Cervical vertebral kinematics and neck muscle responses during an inverted free fall simulating a vehicle rollover: pilot data from an *in vivo* human subject experiment**

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## **ABSTRACT**

*Vehicle rollovers account for 3% of motor vehicle crashes yet cause one-third of all crash-related fatalities. Despite advanced cervical spine injury models, a discrepancy exists between clinically reported injuries and cadaver test pathologies. One possible explanation for this discrepancy is that the intervertebral posture and simulated muscle tone used in cadaver models (and some computer models) typically mimic an upright and relaxed condition that may not exist during a rollover. The aim of this study was to characterize vertebral alignment and neck muscle responses in the cervical spine by studying a human subject in a simulated impending head-first impact, in an upside-down configuration. A custom inversion device was built to expose human subjects to a 312 ms inverted free-fall. An onboard fluoroscopic c-arm captured cervical vertebral motion while indwelling electromyography (EMG) captured the response of 8 superficial and deep neck muscles. The subject showed consistent muscular responses in 4 repetitions of the free-fall exposure. Moreover, the muscle response pattern was different from the scheme used in existing cervical spine injury models, and observed in previous quasi-static tests conducted in our lab. The general trends in muscle-induced changes to vertebral alignment were consistent with our previous work. C3-C6 translated anteriorly and inferiorly in response to the inverted free-fall stimulus, and the head moved into flexion. These observations suggest that, at the time of impact, the *in vivo* state of the neck may differ considerably from its initial alignment prior to the forewarned impact. The *in vivo* data set acquired from this experiment of vertebral and muscular responses could be used to improve and validate current injury models and advance injury prevention strategies in rollover crashes.*

## **INTRODUCTION**

Although rollovers account for only 3% of motor vehicle crashes (“Safercar.gov,” n.d.), they are responsible for 33% of all motor vehicle fatalities (Conroy, 2006) and 40% of serious cervical spine injuries (Yoganandan, 1989). Vehicle rollovers are a chaotic and complex type of motor vehicle collisions in which the exact injury mechanism is poorly understood. Neck injury can occur when the vehicle roof hits the ground and the inverted occupant strikes the interior of the vehicle with their head (Moffatt, 2003; Raddin, 2009). It’s been suggested that the cervical

spine is subsequently loaded by the incoming momentum of the torso (Bahling, 1995). Currently there is a disparity between lab-induced injuries and those reported from rollovers clinically. In cadaver studies, almost half of lab-induced injuries occur as vertebral body fractures while, in the rollover cases, unilateral or bilateral facet dislocations were common (39.1%) as were facet joint fractures (34.7%) but vertebral body fractures occurred to less than 15% of the case occupants (Foster, 2012). The inability to replicate real-world rollover injuries may be due to assumptions used to study cervical spine injury. In cadaveric tests, musculature is removed to aid visualization of the cervical spine during an impact test, although muscle tone may be simulated using systems of cables and pulleys. It's been suggested that the absence of neck musculature is likely responsible for the disparity between lab-induced and real-world injuries (Foster, 2012). Computational modelling has suggested that active neck musculature may shift the mode of injury (Nightingale, 2016a), and more than double the risk of neck fracture (Hu, 2008). The validity of these findings depends on the simulation of realistic muscle activation schemes. Current activation schemes aren't based on, or validated to, measured *in vivo* responses. Existing models assume 100% maximum voluntary contraction (MVC) (Hu, 2008) or a scheme optimized for upright, neutral posture (Chancey, 2003; Nightingale, 2016b), neither of which represent a vehicle rollover. The accurate simulation of muscle forces and spine postures from an impending head-first impact, in future cadaver tests, may drive the injuries from cadaveric tests towards concordance with clinically observed injuries. Previous work in our lab has shown that muscle activity and cervical spine posture are altered due to inversion alone (Newell, 2013) and differ yet again when voluntarily bracing after being instructed to brace for an impact under quasi-static conditions (Newell, 2014). However, the cervical spine posture and muscle activity of an occupant immediately before a head-first impact is unknown. Therefore, the objective of the current study was to capture the *in vivo* dynamic cervical spine re-alignment and muscle activity of a human subject in response to a simulated impending head-first impact.

## MATERIALS AND METHODS

### Subject

To date, one asymptomatic, 32-year-old, male human subject participated in this study. Additional subjects will be tested soon. The male subject met the following exclusion criteria: no prior whiplash injury, history of neck or back pain, known disease affecting muscles or nerves, history of balance problems, known heart condition, skin disease, history of cancer, or involvement in a study involving radiation in the last year. In the case of a female subject, pregnancy would be an additional exclusion criterion. The subject's height was 1.78 m, his weight was 80.9 kg, and his head and neck circumference were 0.585 m and 0.385 m, respectively. This study was approved by the University of British Columbia's Clinical Research Ethics Board, and the subject gave his written informed consent. The subject gave additional permissions to present photographs and video recordings of his participation in the experiment.

## Custom-built inversion device

A custom device was designed and built to expose human subjects to an inverted free-fall drop that simulates a short phase of a rollover crash. Subjects are first seated in an upright posture and then inverted and elevated to a fixed drop height by a feedback controlled linear motor. The linear motor is programmed to expose subjects to a controlled 312 ms inverted free-fall drop, before decelerating to rest (peak deceleration of 1.34g). An onboard fluoroscopic c-arm (OEC 9400, GE) retrofitted with a high-speed camera (Phantom V12.1M, Vision Research Inc., Wayne, NJ), was used to capture sagittal plane images of the cervical vertebra. A shelf mounted to the seat back accommodated pre-amplifiers used to collect Electromyography (EMG) data.

## Conditions

The subject was seated in a bucket seat (36 series – Intermediate 20-degree Layback, Kirkey Racing Fabrication INC., St. Andrew's West, ON) secured with a 75-mm wide 5-point harness (RCI Racers Choice Inc., Tyler, TX) with his arms strapped to his thighs and feet restrained using snowboard bindings. The subject adopted two static postures and one dynamic posture: upright with a neutral posture and relaxed muscle activity (U-R), inverted while maintaining a forward gaze (I-F), and inverted while subjected to an inverted free fall drop (D). The upright-relaxed condition is akin to initial conditions currently used in cadaveric and computational models while the inverted-forward condition represents inverted occupants maintaining their gaze on the road. The drop condition is intended to represent inverted occupants with pre-impact awareness. Prior to the onset of the free-fall drop, the subject was instructed to adopt a forward gaze. One trial was performed for each static condition, whereas four trials were performed for the drop condition with approximately 30 minutes rest between trials. Neck muscle activity and fluoroscopy were recorded throughout all trials.



Figure 1: Four video frames from the inverted drop condition. The first three frames show the subject being raised to the fixed drop height. The last frame shows the subject during the free-fall drop condition.

## **Electromyography (EMG)**

Muscle activity was measured using unilateral indwelling electrodes for 8 neck muscles: sternocleidomastoid (SCM), trapezius (Trap), levator scapulae (LS), splenius capitis (SPL), semispinalis capitis (SsCap), semispinalis cervicis (SsCerv), and multifidus (MultC4) (Blouin, 2007) on the left side only. Ultrasound guidance was used to insert the indwelling electrodes into the muscle bellies at the C5-C6 level (Trap) and C4-C5 level (remaining muscles). EMG signals were amplified at 100x gain and sampled at a frequency of 4000Hz. Two hardware filters band-passed frequencies between 50-2000Hz (NL844 & NL144, Digitimer Ltd., Hertfordshire, UK), after which a digital 4<sup>th</sup>-order high-pass Butterworth filter with a cutoff frequency of 50 Hz was applied. The onset of muscle activity was defined as the sample where the average value of rectified EMG in a 25 ms window around the sample was two standard deviations above the average resting EMG before the stimulus in each trial (Hodges, 1996). To gauge muscle activity relative to maximal muscle activity, the root mean square (RMS) of each signal was calculated for a moving 25 ms window centered around each point for every trial. For the static conditions, windows were chosen in which fluctuations in muscle contractions and neck posture were minimal. For the drop condition, windows were chosen to capture the maximal RMS activity of a muscle during the free-fall drop (0 to 312 ms). The RMS average using a 100 ms time window was also calculated for the 100 ms prior to the onset of free-fall in each trial to indicate pre-drop quasi-static muscle activity. Each muscle's RMS activity was normalized to the maximum RMS activity recorded for that muscle in a maximum voluntary contraction (MVC).

### *Maximum Voluntary Contractions (MVCs)*

The seated subject was secured with a chest strap to a rigid backboard while wearing a snug skateboard helmet that was attached to a 6-axis load cell (45E15A-UU760, JR3, Woodland, CA) above the subject's head (Newell et al., 2013). Two repetitions of 10 isometric contractions were performed with verbal encouragement (Gandevia, 2001) and real-time visual feedback representing force/moment magnitude and direction. The MVCs were performed for 3-5 seconds in a randomized order in 10 directions: flexion, extension, left lateral bending, right lateral bending, four 45° oblique combinations (flexion/left lateral bending, flexion/right lateral bending, extension/left lateral bending, extension/right lateral bending), and left and right axial rotations. The EMG signals were amplified, filtered, and collected as described above. A moving 25 ms window was used to calculate RMS activity at each point. The maximum RMS value for each muscle, regardless of direction, was used for normalization.

## **Cervical spine posture**

To determine cervical spine posture in the sagittal plane, the fluoroscopic c-arm recorded images of the cervical vertebra at 200Hz. Fluoroscopic images were corrected for spherical distortion using XMA Lab (Brainerd, 2010). Inertial loads due to the dynamic environment resulted in mechanical flexing of the c-arm, and consequent misalignment of the x-ray source and image intensifier. To correct for image distortion due to misalignment, a 10x10 array of

1mm steel beads spaced uniformly apart was mounted in the field of view. Images were imported into photo rectification software (PC-Rect, MEA Forensic, Richmond, BC) and rectified separately using a square formed by four of the steel beads. Photoshop (Photoshop CS4, Adobe Systems, San Jose, CA) was used to manually outline each vertebral body along high-contrast boundaries. To be able to make comparisons between images, two individual images were superimposed such that the beads from the bead array coincided, providing a common reference. Superimposed images were imported into video tracking software (TEMA Automotive, Image Systems AB, Linköping, Sweden) where vertebral corners, midpoints, and reference points were manually identified. A sample of the image analysis process is shown below in Figure 2. Four image comparisons were performed: upright-relaxed (U-R) vs. inverted-forward (I-F), and initial vs. final frames for three of the four drop trials. Initial frames were the last frame prior to free-fall onset, and final frames were the last frame prior to the onset of deceleration.

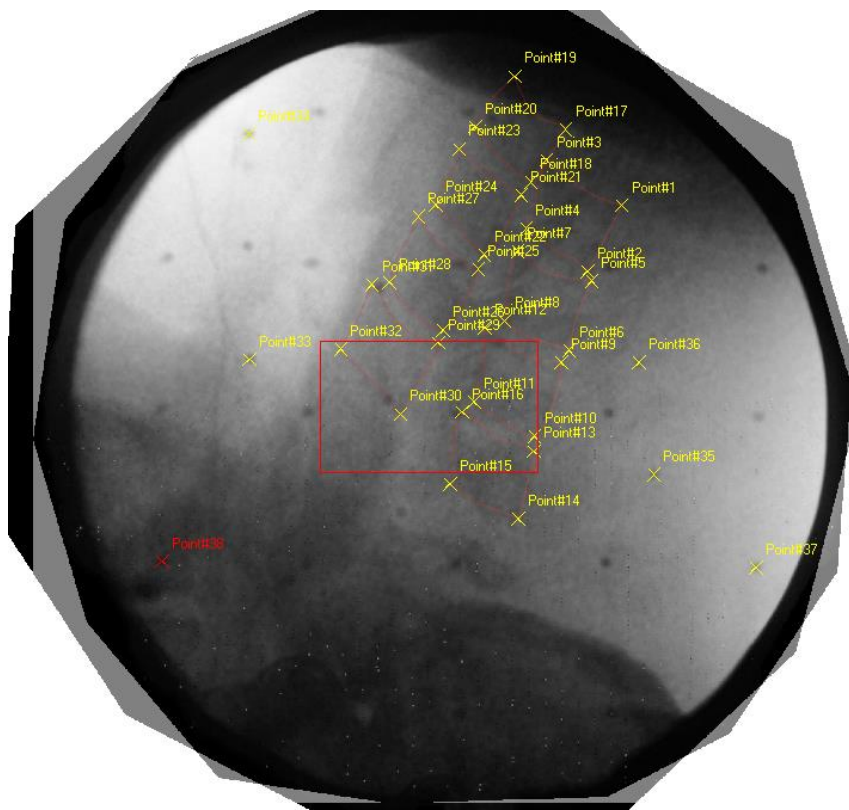


Figure 2: Sample of the image analysis process after the images have been corrected for spherical distortion, vertebral bodies outlined (faint red), images rectified, and superimposed to allow comparison between the identified vertebral corners. The beads from the bead array of the initial image coincide with the beads from the bead array in the final image. The 2mm steel bead array related to the Frankfort plane angle is visible for both frames. The common reference line connects points 37 and 38.

Changes in cervical spine posture were quantified with two measures: the displacement of vertebral bodies and change in vertebral angle ( $\theta_v$ ) between images. The displacement was measured between the inferior midpoints of each vertebra. Distances were normalized to a reference line common to all frames and reported as a fraction thereof. Vertebral angles were

determined by a line drawn through the two inferior corners of each vertebra and measured relative to the common reference line, as shown in Figure 3.

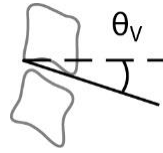


Figure 3: Vertebral angles were determined by drawing a line through the two inferior corners of the vertebra and measured relative to the common reference line. The dotted horizontal line is parallel to the common reference line described in Figure 2.

### Head orientation

To determine head orientation, a bead array (six-2mm beads) was fastened to the head and projected into the field of view of the fluoroscope. The Frankfort plane angle ( $\theta_F$ ) was defined as a line from the mid-point of the left tragus and inferior midpoint of the left orbit and was measured relative to the common reference line (positive=extension) (Figure 4). Prior to the experiment, the beads and Frankfort plane landmarks were digitized using a 3D motion capture system (Optotrak Certus, Northern Digital Inc., Ontario, Canada) to establish the initial offset. Thus, angular motion of the beads could be used to determine angular motion of the Frankfort plane.

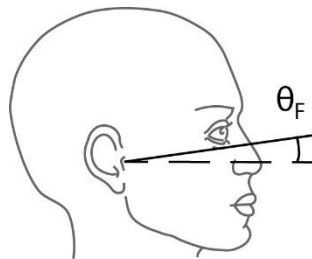


Figure 4: Frankfort plane angle was measured relative to the common horizontal (the dotted line is parallel to the common horizontal reference line), with extension in the positive direction.

### Data processing notes

The upright-relaxed, inverted-forward, and drop conditions were performed before the MVCs. During the MVC portion of the experiment, the indwelling electrode wires for STH and SsCap may have been damaged as the subject transferred from the inversion device to the MVC set-up. Cervical vertebra C1-C6 were visible in the 23cm field of view for all trials, except C6 in the upright-relaxed condition. Prior to the experiment, the mirror which optically couples the camera and image intensifier had unknowingly come loose. As a result, images were dark and out-of-focus. Due to image quality, the boundaries of C1-C2 and C7 were not discernable and could not be analyzed. Additionally, fluoroscopic images were not recorded for the free-fall duration of Trial 2 due to human error.

All head/neck posture variables and EMG signals were analyzed using Matlab (R2017a, MathWorks Inc., Natick, MA).

## RESULTS

### Muscle activity

In each trial, unprocessed EMG signals showed a large initial burst of activity followed by several moderately sized bursts. Muscle activity approached minimal levels prior to the onset of deceleration (

Figure 5). The onset of muscle activity occurred within the first 100 ms of each trial. There appeared to be a consistent onset of muscle activity after the free-fall stimulus over the four trials (Figure 6).

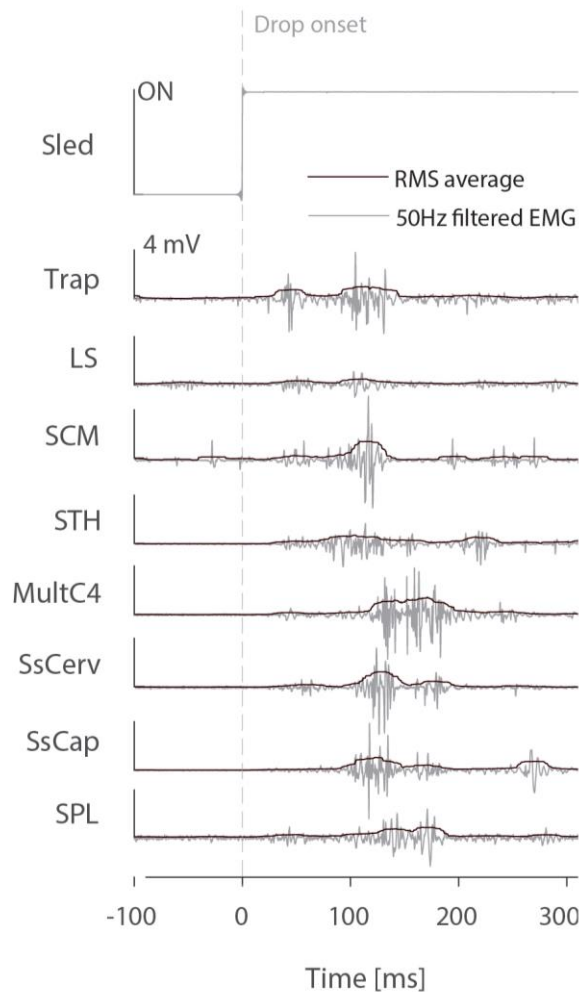


Figure 5: Unprocessed EMG signal from Trial 1 after being high-pass filtered at 50Hz. Rms EMG is included and has been overlaid. Vertical scale bars represent 4mV. The onset of deceleration is 312ms.

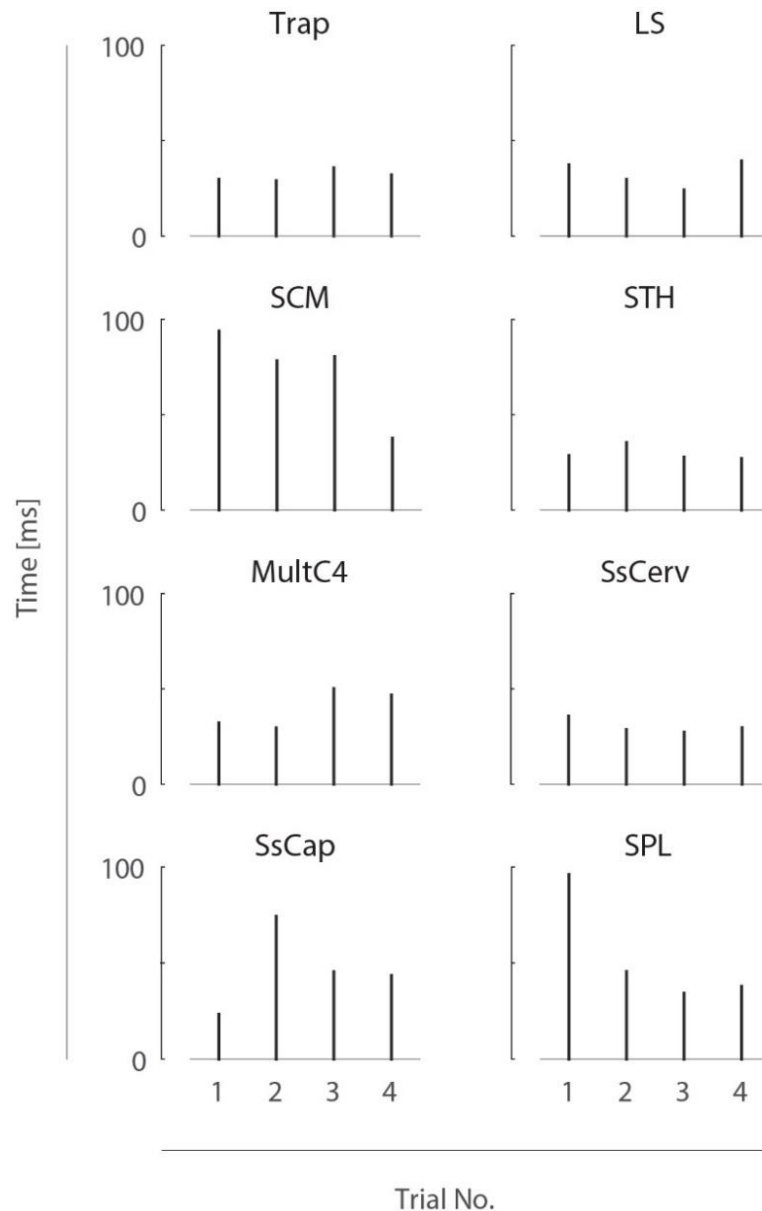


Figure 6: Onset of Normalized rms EMG for each trial. The onset of muscle activity was consistent across the four trials.

While upright and relaxed, the subject had minimal muscle activity with SsCap most active at 12% MVC (Table 1). When inverted and maintaining a forward gaze, muscle activity increased in five of the eight muscles. Activity in the extensor muscles Multc4, SsCap, and SsCerv decreased compared to the upright and relaxed condition. In the drop trials, all muscles were minimally active in the 100 ms prior to the onset of free-fall. During free-fall, maximum activity levels of all muscles were significantly higher than compared to either static posture (Table 1). Most muscles increased activity by an average of more than 40%, excluding LS and

SPL, which increased by less than 20%. The muscle activation pattern was generally consistent across all four trials, apart from SsCap and SPL in Trial 2 (Figure 7).

Table 1: Normalized EMG rms average based on a 100ms window, applied to the 100ms prior to the drop onset. Maximum %MVC values recorded for each muscle across all 4 trials and both static postures (U-R = upright-relaxed, I-F = inverted-forward gaze)

Muscle	%MVC 100ms prior to drop onset				Maximum %MVC				%MVC	
	Trial 1	Trial 2	Trial 3	Trial 4	Trial 1	Trial 2	Trial 3	Trial 4	U-R	I-F
	[%]	[%]	[%]	[%]	[%]	[%]	[%]	[%]	[%]	[%]
Trap	2.72	3.96	4.50	4.57	31.67	57.83	66.02	42.06	3.15	9.29
LS	3.51	2.59	2.18	5.44	13.69	24.47	19.03	22.87	6.96	13.20
SCM	5.35	6.22	5.05	7.12	50.36	65.83	53.79	43.41	1.32	12.66
STH	3.45	5.62	4.83	3.68	98.18	127.96	95.16	72.24	4.52	12.52
MultC4	1.09	1.12	1.19	1.09	42.49	68.10	50.90	39.20	5.11	1.29
SsCerv	1.87	1.26	1.32	1.22	48.42	52.40	56.53	66.23	1.88	1.41
SsCap	2.09	2.07	2.16	2.13	102.17	304.23	87.99	110.92	11.59	2.81
SPL	2.00	1.79	1.92	1.79	17.52	52.37	12.80	15.12	2.64	3.33

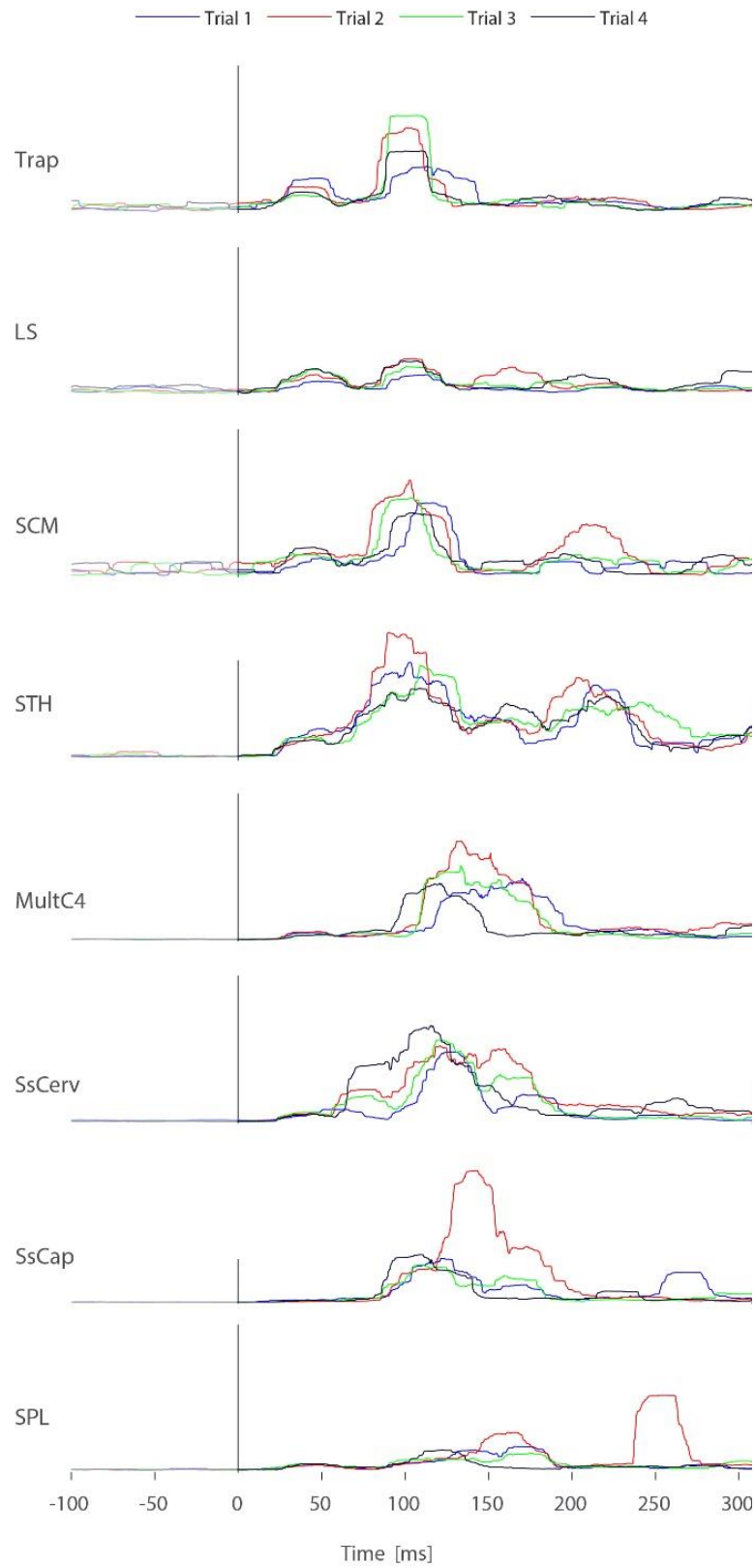


Figure 7: Normalized rms EMG for each muscle across all four trials. Muscle scale bars indicate 100% of maximal voluntary contractions (MVCs).

## Neck and head posture

Neck vertebral postures were different in all three conditions. In the inverted-forward condition, vertebral bodies moved posteriorly and superiorly, when compared in the same orientation, to the upright-relaxed condition. Prior to the onset of free-fall, the initial inverted-forward vertebral postures were consistent with the static inverted-forward condition. After experiencing free-fall and prior to the onset of deceleration, vertebral bodies moved anteriorly and inferiorly – indicating neck flexion (Figure 8). Vertebral angles increased with inversion, and more so with exposure to free-fall. Frankfort plane moved in concert with the vertebrae, indicating head flexion with exposure to free-fall (Table 2).

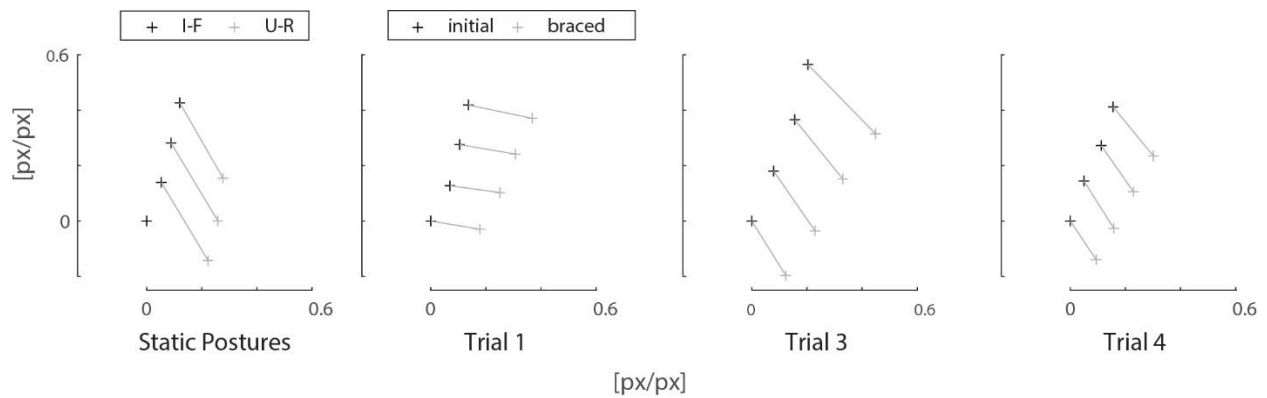


Figure 8: Vertebral translations between the initial posture prior to the onset of free-fall, and the final posture prior to the onset of deceleration. Distances are expressed as a fraction of the common horizontal reference line. The static postures are upright-relaxed (U-R) and inverted-forward (I-F). Vertebra C3-C6 were identified for all conditions except for the upright-relaxed condition, of which C6 was missing. The conditions have been plotted in the same orientation for comparison; C3 is plotted as the top-most vertebra with C4 immediately below, and so on.

Table 2: Vertebral and Frankfort plane angles were measured relative to the reference horizontal (bottom-most row of bead array plate) reference line. Decreasing Frankfort plane angles indicate flexion. The Frankfort plane assumes that the Image Intensifier and Optotrak's Y-Z plane are parallel.

	U-R	I-F	Trial 1		Trial 2		Trial 3		Trial 4	
			Initial	Final	Initial	Final	Initial	Final	Initial	Final
<b>Vertebral angles [deg]</b>										
<b>C1</b>	-	-	-	-	-	-	-	-	-	-
<b>C2</b>	-	-	-	-	-	-	-	-	-	-
<b>C3</b>	19.75	22.30	22.85	29.90	-	-	24.22	42.29	27.98	40.78
<b>C4</b>	22.58	30.00	17.27	29.14	-	-	24.24	40.38	24.07	40.75
<b>C5</b>	20.48	23.81	24.00	31.58	-	-	31.20	38.45	21.46	35.88
<b>C6</b>	-	22.71	26.28	34.93	-	-	24.74	37.14	30.42	45.70
<b>C7</b>	-	-	-	-	-	-	-	-	-	-
<b>Frankfort plane angles [deg]</b>										
	1.62	13.40	12.22	7.22	-	-	6.30	-4.54	8.80	1.62

## DISCUSSION

The objective of this study was to measure *in vivo* muscle activation patterns and the realignment of human cervical vertebrae in response to an inverted, free-fall, impending head-first impact. Overall, we observed sub-maximal increases in muscle activity followed by muscle-induced anterior motion of the cervical spine and combined flexion-retraction of the head. These observations support our previous findings that the *in vivo* state of the neck, at a time relevant to a head-first impact during a rollover crash, may differ considerably from its initial alignment prior to a forewarned impact (Newell, 2014).

In previous work done in our lab, inverted subjects were instructed to ‘brace for impact’ under quasi-static conditions (Newell, 2014). Their responses were performed without any actual threat and depended solely on their interpretation of the instructions. The current study applies a free-fall stimulus under the threat of impending head-first impact, generating a more realistic reflexive neck muscle response. These dynamic conditions are more likely to reflect the posture and muscle state immediately before an inverted head-first impact. Applied to vehicle rollovers, the end of free-fall approximates the instant of head contact with the vehicle roof, and thus the state of the spine at this instant is potentially relevant to catastrophic neck injuries in rollover crashes.

Our results align with previous work done in our lab that the *in vivo* muscle activation levels of inverted subjects differ considerably from those of upright subjects. All muscle activation levels for the static conditions are within two standard deviations of previous studies (Newell, 2013) except for STH, SsCap, and SPL in the upright-relaxed condition. Activity levels associated with STH and SsCap are expected to be artificially high as the indwelling wires may have been damaged prior to the MVC procedure.

Under dynamic conditions, we observed a different muscle activation scheme than the one seen in inverted subjects who voluntarily adopted a quasi-static bracing posture (Newell, 2014). In the quasi-static bracing task, Trap and SPL increased the most (mean increase of 36.2% and 22.7%, respectively), whereas we observed six muscles with mean increases greater than 40%, excluding SPL. Another of these muscles was MultC4, which had previously increased the least (mean increase of 4.3%) in the quasi-static task. The quasi-static bracing task reported high levels of between-subjects variability, with confidence intervals as high as 68%, making it difficult to attribute the difference in muscle response to quasi-static vs. reflexive contractions.

The observed muscle activation pattern was consistent across all four free-fall trials, except for SPL and SsCap in Trial 2. After consulting the raw data, the peaks in these two muscles did not resemble EMG activity and should not be interpreted as such. The consistency of the reflexive muscle response across all four trials suggests that between-subjects variability may be reduced in a reflexive response.

It should be noted that the EMG onset of Trap, LS, SCM, STH is not a true onset, as the muscles were already active while the subject tried to maintain an inverted-forward gaze (Figure 7). Cervical vertebral translations followed the same trend as the quasi-static bracing task (Newell, 2014). Unfortunately, a comparison of distances is precluded by the missing vertebra (C1-C2, C7) and the yet undetermined scaling factor between the image intensifier and the

subject plane. The decrease in Frankfort plane angle was not observed in previous work, suggesting a reflexive response generates a different head posture than a quasi-static tensing.

Most existing cervical spine injury models assume an upright posture with muscle activity simulated at 100% MVC (Hu, 2008) or to represent an upright-relaxed posture (Chancey, 2003; Nightingale, 2016b). These assumptions give little consideration to the effect of pre-impact awareness. Our results show the *in vivo* muscle activity (Figure 6) and cervical spine posture (Figure 8) during free-fall differ from the upright-relaxed condition. Aside from artificially high activations in SPL and SsCap, none of our muscles exceeded 70% MVC, yet all exceeded upright-relaxed activation levels. This difference between inverted occupants maintaining a forward gaze and those preparing for impact (Figure 7) illustrates the effect of pre-impact awareness. While different from one another, both conditions showed increased muscle activity and a shift in cervical spine posture when compared with the upright-relaxed condition. Thus, the initial conditions used in current cervical spine injury models may not be representative of a head-first impact in a rollover.

Cadaver studies have shown that injuries to the spinal column are sensitive to the overall spinal eccentricity. Average eccentricities of -5mm, 1mm, 23mm, and 53mm are reported to result in compression-extension, vertical compression, compression-flexion, and hyperflexion injuries, respectively (Maiman, 2002). Thus, the cervical spine posture represented in the initial conditions should be carefully considered as they are likely to change the injury outcome. Further work is needed to evaluate how much the eccentricity of subjects exposed to our free-fall stimulus changes and how relevant these changes will be to the risk of different neck injuries.

For this study, a single human subject was exposed to an inverted free-fall intended to simulate a short phase of a vehicle rollover. These results should be interpreted carefully as more subjects are needed before reaching definitive conclusions. Additionally, rollovers are dynamic and complex events, and pre-rollover dynamics are not captured in this study. Centripetal accelerations in a rollover environment could influence both muscular and postural responses and future experiments are necessary to understand these effects. The aim of this study was to capture muscle activity and posture in a condition which may exist immediately before an inverted head-first impact. Our findings are relevant to other circumstances of inverted, impending head-first impact as it provides evidence that a reflexive response to a free-fall stimulus can generate significant change in muscle activity and cervical spine posture.

## CONCLUSION

A custom inversion device was built to simulate impending head-first impacts and to capture cervical spine posture and muscle activity. An *in vivo* data set of vertebral and muscular responses, in the context of pre-impact in a rollover environment, was collected. Sub-maximal increases in muscle activity were observed, followed by muscle-induced anterior motion of the cervical spine and flexion of the head. These results indicate the initial conditions used in current cervical spine injury models may not reflect those present during an inverted headfirst impact.

An *in vivo* data set of vertebral and muscular responses could be used to improve and validate current injury models and advance injury prevention strategies.

## ACKNOWLEDGMENTS

Thank-you to Jeff Nickel and Mircea Oala-Florescu of MEA Forensic Engineers & Scientists for their help and support in the development of the inversion device.

## REFERENCES

- BAHLING, G. S. ., BUNDORF, R. T. ., MOFFATT, E. A. ., & ORLOWSKI, K. F. (1995). The influence of increased roof strength on belted and unbelted dummies in rollover and drop tests. *Journal of Trauma*, 38(4), 557–563.
- BLOUIN, J.-S., SIEGMUND, G. P., CARPENTER, M. G., & INGLIS, J. T. (2007). Neural Control of Superficial and Deep Neck Muscles in Humans. *Journal of Neurophysiology*, 98(2), 920–928. <https://doi.org/10.1152/jn.00183.2007>
- BRAINERD, E. L., BAIER, D. B., GATESY, S. M., HEDRICK, T. L., METZGER, K. A., GILBERT, S. L., & CRISCO, J. J. (2010). X-ray reconstruction of moving morphology (XROMM): precision, accuracy and applications in comparative biomechanics research. *Journal of Experimental Zoology Part A: Ecological Genetics and Physiology*, 313A(5), 262–279. <https://doi.org/10.1002/jez.589>
- CHANCEY, V. C. ., NIGHTINGALE, R. W. ., VAN EE, C. A. ., KNAUB, K. E. ., & MYERS, B. S. (2003). Improved estimation of human neck tensile tolerance: Reducing the range of reported tolerance using anthropometrically correct muscles and optimized physiologic initial conditions. *Stapp Car Crash Journal*, 47, 135–153.
- CONROY, C., HOYT, D. B., EASTMAN, A. B., ERWIN, S., PACYNA, S., HOLBROOK, T. L., ... VELKY, T. (2006). Rollover crashes: Predicting serious injury based on occupant, vehicle, and crash characteristics. *Accident Analysis & Prevention*, 38(5), 835–842. <https://doi.org/10.1016/j.aap.2006.02.002>
- FOSTER, J. B., KERRIGAN, J. R., NIGHTINGALE, R. W., FUNK, J. R., CORMIER, J. M., BOSE, D., ... CRANDALL, J. R. (2012). Analysis of cervical spine injuries and mechanisms for CIREN rollover crashes. In *Proceedings of the International Research Council on the Biomechanics of Injury conference* (Vol. 40, pp. 61–75). Retrieved from [http://www.ircobi.org/downloads/irc12/pdf\\_files/15.pdf](http://www.ircobi.org/downloads/irc12/pdf_files/15.pdf)

- GANDEVIA, S. C. (2001). Spinal and supraspinal factors in human muscle fatigue. *Physiological Reviews*, 81(4), 1725–1789.
- HODGES, P. W., & BUI, B. H. (1996). A comparison of computer-based methods for the determination of onset of muscle contraction using electromyography. *Electroencephalography and Clinical Neurophysiology/Electromyography and Motor Control*, 101(6), 511–519. [https://doi.org/https://doi.org/10.1016/S0921-884X\(96\)95190-5](https://doi.org/https://doi.org/10.1016/S0921-884X(96)95190-5)
- HU, J., YANG, K. H., CHOU, C. C., & KING, A. I. (2008). A Numerical Investigation of Factors Affecting Cervical Spine Injuries During Rollover Crashes: *Spine*, 33(23), 2529–2535. <https://doi.org/10.1097/BRS.0b013e318184aca0>
- MAIMAN, D. J., YOGANANDAN, N., & PINTAR, F. A. (2002). Preinjury cervical alignment affecting spinal trauma. *J Neurosurg Spine*, 97(1), 57–62.
- MOFFATT, E. A., COOPER, E. R., CROTEAU, J. J., ORLOWSKI, K. F., MARTH, D. R., & CARTER, J. W. (2003). Matched-Pair Rollover Impacts of Rollcaged and Production Roof Cars Using the Controlled Rollover Impact System (CRIS). *Society of Automotive Engineers*, 2003-1–172.
- NEWELL, R. S., BLOUIN, J.-S., STREET, J., CRIPTON, P. A., & SIEGMUND, G. P. (2013). Neck posture and muscle activity are different when upside down: A human volunteer study. *Journal of Biomechanics*, 46(16), 2837–2843. <https://doi.org/10.1016/j.jbiomech.2013.08.013>
- NEWELL, R. S., SIEGMUND, G. P., BLOUIN, J.-S., STREET, J., & CRIPTON, P. A. (2014). Cervical Vertebral Realignment When Voluntarily Adopting a Protective Neck Posture. [Miscellaneous Article]. *Spine*, 39(15). <https://doi.org/10.1097/BRS.0000000000000384>
- NIGHTINGALE, R. W., SGANGA, J., CUTCLIFFE, H., & BASS, C. R. “DALE.” (2016a). Impact responses of the cervical spine: A computational study of the effects of muscle activity, torso constraint, and pre-flexion. *Journal of Biomechanics*, 49(4), 558–564. <https://doi.org/10.1016/j.jbiomech.2016.01.006>
- NIGHTINGALE, R. W., SGANGA, J., CUTCLIFFE, H., & BASS, C. R. “DALE.” (2016b). Impact responses of the cervical spine: A computational study of the effects of muscle activity, torso constraint, and pre-flexion. *Journal of Biomechanics*, 49(4), 558–564. <https://doi.org/10.1016/j.jbiomech.2016.01.006>

RADDIN, J., CORMIER, J., SMYTH, B., CROTEAU, J., & COOPER, E. (2009). Compressive neck injury and its relationship to head contact and torso motion during vehicle rollovers. *SAE Technical Paper No. 2009-01-0829*. <https://doi.org/10.4271/2009-01-0829>

SAFERCAR.GOV. (n.d.). Retrieved December 1, 2016, from <http://www.safercar.gov/Vehicle-Shoppers/Rollover/Fatalities>

YOGANANDAN, N., HAFFNER, M., MAIMAN, D. J., NICHOLS, H., PINTAR, F. A., JENTZEN, J., ... SANCES, A. (1989). Epidemiology and Injury Biomechanics of Motor Vehicle Related Trauma to the Human Spine. <https://doi.org/10.4271/892438>