Head/Neck Kinematics and Muscle Responses for Relaxed Drivers and Passengers in Frontal and Rear Impacts.

Jason B. Fice, Daniel W. H. Mang, Jóna M. Ólafsdóttir, Karin Brolin, Peter A. Cripton, Jean-Sébastien Blouin, Gunter P. Siegmund.

I. INTRODUCTION

The risk of whiplash-associated disorders (WADs) for drivers is 1.4 to 1.9 times higher than the risk for passengers during a rear-end impact [1-2]. No similar difference has been observed for frontal impacts [3]. While these differences in the relative risk of WAD in drivers and passengers may be due to a range of factors, e.g., postures, awareness, bracing, etc., the goal of this study was to quantify the isolated influence of holding onto a steering wheel on the head/neck kinematics and neck muscle responses of relaxed volunteers exposed to low-speed frontal and rear-end impacts.

II. METHODS

Eleven healthy subjects (3F, 8M) volunteered for the study, which was approved by the University of British Columbia Clinical Research Ethics Board. Subjects were seated in a 2005 Volvo S40 driver's seat that was mounted on a feedback-controlled sled ([4] Fig. 1). The acceleration pulse generated a speed change of 0.77±0.03m/s and a peak acceleration of 2.01±0.09g over a duration of 65.5±1.2ms (Fig. 2A). All subjects experienced frontal and rear impacts in a driving posture (relaxed arms, hands on the steering wheel, Fig. 1) and a passenger posture (arms and hands resting on their lap). The sound of an actual vehicle-to-barrier crash (109dB) accompanied the perturbations. Five minutes was left between consecutive trials to minimise habituation. This study was part of a larger experiment examining risk factors for WADs [4].

Neck muscle electromyographic (EMG) activity was recorded with indwelling electrodes inserted unilaterally under ultrasound guidance (Sonosite, Bothell, USA) into the sternohyoid (STH), sternocleido-mastoid (SCM), splenius capitis (SPL), semispinalis capitis (SSCAP), and C4 multifidus (MU) muscles. EMG signals were amplified (×100), notch filtered (60Hz), bandpass filtered (50-2000Hz) and recorded at 4000Hz. Before the perturbations, subjects performed neck maximum voluntary isometric contractions (MVICs) in 10 directions for EMG normalisation. Head accelerations were measured with a nine-accelerometer array (Kistler, Amherst, USA) in a 3-2-2-2 configuration [5]. Torso kinematics were measured with triaxial angular rate sensors (ATA, Albuquerque, USA) and triaxial accelerometers (Summit, Akron, USA). An Optotrak system (NDI, Waterloo, Canada) tracked infrared markers fixed to the head sensors, torso sensors and sled. Sensor data were acquired at 4000Hz and Optotrak data were acquired at 200Hz.

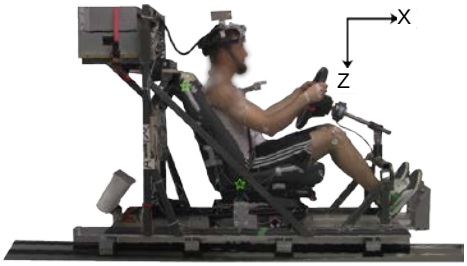

Fig. 1. Experimental setup with the lab coordinate system shown.

The root mean square (RMS, 20ms moving window) of the EMG data were normalised by the maximum RMS EMG measured during the MVICs. EMG onsets were determined with a log-likelihood ratio approach [6] and peak EMG was the maximum RMS value after impact. Zero-phase low-pass filters were applied to the linear accelerometer signals (100Hz) and angular rate signals (30Hz). Gravity was removed from the accelerometer signals using Optotrak-derived orientations, and the pre-impact bias was then removed from all kinematic signals. Head angular accelerations were calculated from the linear accelerometers [5]. Head and torso kinematics were transformed to the head CoG and C7-T1 respectively in the lab reference frame (Fig. 1). All data are reported as median (interquartile ranges) and differences between driver and passenger trials were tested using Wilcoxon signed rank tests with separate analyses for each variable and impact direction. The test statistic, W, is the sum of signed ranks.

J.B. Fice completed his PhD and D.W.H. Mang is a PhD student in Kinesiology at the University of British Columbia in Vancouver, British Columbia, Canada. J.M. Ólafsdóttir completed her PhD and K. Brolin is a Professor in Mechanical Engineering at Chalmers University of Technology in Gothenburg, Sweden. P.A. Cripton is a Professor in Mechanical Engineering and J.-S. Blouin is a Professor in Kinesiology at the University of British Columbia in Vancouver, British Columbia, Canada. G.P. Siegmund is the Director of Research at MEA Forensic Engineers & Scientists in Richmond, British Columbia, Canada (gunter.siegmund@meaforensic.com, +1 604 277 3659).

III. INITIAL FINDINGS

Driver head positions were 6.5mm (4.4-13.5mm, W_{11}=62, p=0.003) forward of the passenger head positions before rear impacts only. Fore-aft head position and angles were otherwise similar (p>0.206). Pre-impact neck muscle activity was lower in drivers than passengers in SPL (0.81 vs 0.82%MVIC, W_{11}=50, p=0.024) before rear impacts, and in STH (1.21 vs 1.44%MVIC, W_{11}=50, p=0.024) and SCM (1.36 vs 1.37%MVIC, W_{11}=48, p=0.032) before frontal impacts.

EMG onset occurred earliest in STH (58ms, 55-67ms) and latest in MU (102ms, 79-120ms) for the passengers during rear impacts (Fig. 2A & B). EMG onset did not vary between drivers and passengers for either frontal or rear impacts ($W_{8\text{-}11}$=1-30, p=0.193-1.000). Peak EMG ranged from 12 (8-41) %MVIC in SSCAP for passengers during rear impacts to 113 (42-157) %MVIC in SCM for passengers during frontal impacts. Peak EMG did not vary between drivers and passengers in either impact direction ($W_{9\text{-}11}$=2-42, p=0.067-0.966).

In frontal impacts, drivers had greater head accelerations (1.50 vs. 0.94g, W_{10}=55, p=0.002) and torso accelerations (2.82 vs. 1.31g, W_{10}=55, p=0.002), lower peak torso angles (5.8 vs. 9.6°, W_{10}=47; p=0.014) and earlier peak kinematics (11.4-85.2ms, W_{10}=43-55, p=0.002-0.027) except for torso acceleration (Fig. 2A and C). In rear impacts, drivers had greater torso accelerations (2.31 vs. 2.22g, W_{10}=41, p=0.037), head angles (20.1 vs. 18.8°, W_{10}=43, p=0.027), and torso angles (15.4 vs. 12.3°, W_{10}=45, p=0.020). The timing of the peak kinematics was not significantly different between drivers and passengers in rear impacts (W_{10} = 1-37, p=0.065-1.000).

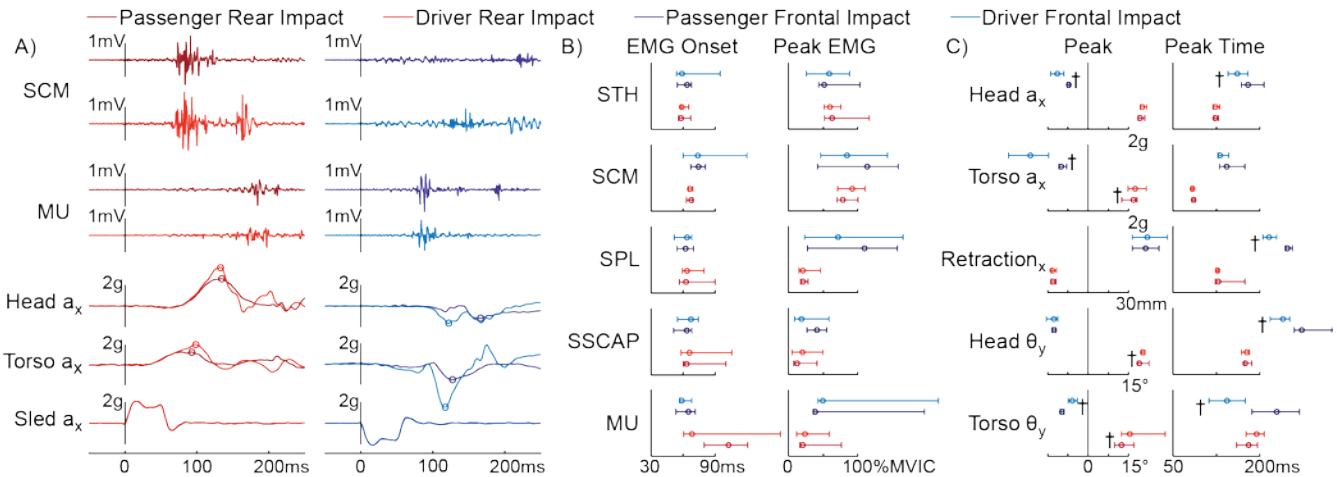

Fig. 2. Exemplar data traces from one subject (A). The median and interquartile range of the volunteers' EMG (B) and kinematic (C) responses (passenger and driver postures). † Denotes a significant difference (p<0.05).

IV. DISCUSSION

In this study, we found that neck muscle activity (both onset and peak) was not different for the seated postures of drivers and passengers in either impact direction. These results suggest that neck muscle activity does not contribute to the previously reported differences in WAD injury risk between drivers and passengers in rear-end impacts. Nevertheless, neck muscle activity could still contribute to these injury risk differences when other factors, e.g., non-neutral postures, awareness, and bracing, are considered.

In frontal impacts, drivers exhibited higher head and torso accelerations and similar head flexion angles despite lower torso flexion angles. This pattern suggests higher neck forces and increased neck tissue strains even though there is no evidence of increased injury risk in field data [3]. Smaller increases in torso acceleration, head angle, and torso angle were seen in drivers during rear impacts, where field data do show a higher risk of injury in drivers than passengers [2]. These discrepancies between our findings and the field data suggest that isolated differences in arm posture do not explain the higher WAD risk of drivers in the field studies. Drivers are more likely to report being in non-neutral head postures at the time of impact [2] and they can brace against the steering, both of which are factors that are known to increase injury risk [2-3].

V. REFERENCES

[1] Jonsson B et al, IATSS, 2013.
[2] Jakobsson L et al, TIP, 2008.
[3] Jakobsson L et al, TIP, 2004.
[4] Fice J, PhD Thesis, 2019.
[5] Padgaokar A et al, J. Appl. Mech. 1975.
[6] Staude G, IEEE Trans Biomed, 2001.