Comparison of Head and Neck Kinematics and Electromyography Response for Low-Speed Frontal Impacts with Pediatric and Young Adult Volunteers

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ABSTRACT

No data exist on the electromyography (EMG) responses in children exposed to dynamic impacts and the resulting head and neck kinematics. The objective of the current study was to measure the EMG responses of the neck, torso and lower extremity in children and young adults during a low speed frontal impact using surface electrodes bilaterally and compare these muscle responses to key kinematic events. Low speed frontal impact tests were performed on seated and restrained pediatric (n=11, ages 8-14 years) and adult (n=9, ages 18-30 years) male subjects. Subjects experienced a maximum acceleration pulse of 4.9 g in 55.7 msec. The timing and magnitude of the EMG responses were compared between the pediatric and young adult groups. Since no bilateral differences were observed in muscle response, the left and right muscle responses were combined for the comparison with kinematics. The interquartile range (IQR) of maximum forward head excursion occurred at 183.8-224.6 msec. The IQR of onset times for the Cervical Paraspinous, Sternocleidomastoid and Upper Trapezius were 34.00-52.67 msec, 50.83-82.67 msec, and 48.17-98.50 msec, respectively. Also of note, the IQR for time of peak muscle activity for the Cervical Paraspinous, Sternocleidomastoid, and Upper Trapezius were 84.00-154.0 msec, 121.5-190.0 msec, and 116.2-169.3 msec. It can be seen that in the low-speed frontal impact tests, all the neck muscles under evaluation were activated (onset) and had reached their peak muscle response prior to the time of maximum forward head excursion. These preliminary results indicate that muscle response may influence head motion in the low-speed environment. Further analysis will examine the relationship between onset times, peak response times and maximum head rotation as well as between magnitude of muscle response and maximum head excursion.

INTRODUCTION

According to the US Centers for Disease Control, the leading cause of death for individuals 5-34 years is motor vehicle collisions (MVCs) [CDC, 2009]. Computational modeling provides a tool for the safety industry to examine occupant motion during MVCs and determine injury causation in an effort to design safety countermeasures. Accurate determination of injury causation may require an understanding of the role of muscle activation in occupant kinematics during MVCs.

Electromyography (EMG) provides a means to measure and evaluate muscle response during a dynamic event. Surface EMG (sEMG) electrodes are placed on the skin over the muscle of interest using a gel-based adhesive to reduce skin resistance and thereby facilitate conduction. These electrodes are able to detect, amplify and record variations in skin voltage due to underlying muscle contractions. Such data have never been collected for children in an automotive environment.

Currently, improvements in vehicle safety are achieved through analyses of post-mortem human subjects and anthropomorphic test device (crash test dummy or ATD) kinematics, which do not account for the effect of active musculature on occupant kinematics. Several studies have evaluated muscle activity of adult volunteers in response to dynamic events. Low-speed frontal impact tests on adult volunteers conducted by Ejima et al. [2007] showed a considerable influence of muscle response on the occupant kinematics. Occupants that were tensed prior to the impact demonstrated more controlled motion than when relaxed pre-impact. Choi et al. [2005] quantified the muscle tensing activity of adult occupants during pre-impact bracing and found good correlation with simulated muscle tensing behavior. These results were further used to validate computational human models with simulated muscle activity [Ejima et al., 2009; Choi et al., 2005]. Additionally, Kumar et al [2006] examined the effect of seat belt use on cervical muscle activity in response to whiplash-type events for multiple directions and varying magnitudes of acceleration, and found that with increasing acceleration, the time to onset of EMG decreased, indicating an effect of acceleration on muscle activation. These results were also supported by their study of low-velocity frontal impacts [Kumar et al, 2003]. In a study to evaluate the cervical muscle activity of volunteers in low-speed rear impacts, Magnusson et al [1999] concluded that due to the onset times of the cervical muscles, the muscles could influence injury patterns. They also determined that the location of the muscle relative to the spinal axis influences the muscle's reaction time. Bose et al. [2008] used musculoskeletal modeling to study the role of bracing on kinematics of restrained occupants. Their study demonstrated the utility of measuring muscle activity during dynamic events to incorporate muscle response in models optimizing adaptive restraint systems. They found that the injury outcomes of the occupant, as well as their interaction with the restraint system, were sensitive to pre-impact bracing by the occupant. Studies on occupant awareness have shown that unaware subjects have increased excursion than those with some level of awareness of the imminent perturbation [Siegmund et al, 2003; Kumar et al, 2000; Kumar et al, 2002].

All of the aforementioned studies were conducted on adult subjects; similar data do not exist in the pediatric population. There is also a dearth of information on the effect of age on the temporal nature of muscle activation during a dynamic event. Therefore, the objective of the current study was to measure the EMG responses of the neck, torso and lower extremity in children during a low speed frontal impact and compare those to similar data from young adults. We hypothesize that under low-speed frontal loading conditions, timing and magnitude of neck muscle activity can influence the occupant's head and neck kinematics.

METHODS

This study protocol was reviewed and approved by the Institutional Review Boards at The Children's Hospital of Philadelphia, Philadelphia, PA and Rowan University, Glassboro, NJ. Informed consent / assent was obtained from all the participants of this study.

Human Volunteer Instrumentation, Testing, and Data Processing

A comprehensive description of the testing method can be found in Arbogast et al [2009]. Briefly, low-speed frontal sled tests were conducted using 20 male human volunteers (11 pediatric subjects: 8-14 years old, 9 young-adult subjects: 18-30 years old). All subjects were between 5^{th} and 95^{th} percentile for their age-appropriate height, weight and body mass index [Centers for Disease Control, 2000; NHANES, 1994]. Subjects with existing neurologic, orthopedic, genetic, or neuromuscular conditions, any previous injury or abnormal pathology related to the head, neck or spine were excluded from the study.

A pneumatically actuated, hydraulically controlled low-speed acceleration volunteer sled (Figure 1) consisting of a moving platform with a low back padded seat, lap-shoulder belt and an adjustable foot rest was used to subject restrained human volunteers to a sub-injurious, low-speed frontal crash pulse. The maximum linear acceleration was 25% below the maximum acceleration measured during an amusement park bumper car ride (4.3 g in 61.9 msec) [Arbogast et al, 2009].

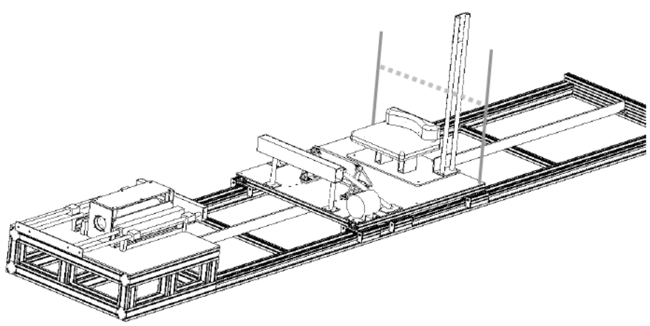

Figure 1: Schematic of low-speed acceleration sled

Several anthropometric measurements were obtained from the subjects prior to testing. Photo-reflective targets were placed on anatomical landmarks including the head, spine, shoulders, sternum, and legs and were tracked using a 3D motion analysis system at 100 Hz (Model Eagle 4, Motion Analysis Corporation, Santa Rosa, CA).

EMG measurements were obtained for all trials. Prior to sEMG electrode placement, the subject's skin was cleaned by applying Skin Prepping Gel (NUPREP, Weaver and Co., Aurora, CO). Disposable, self-adhesive dual surface EMG electrodes (20 mm inter-electrode distance) (Noraxon, Inc., Scottsdale, AZ) were placed bilaterally (left-L, right-R) on key muscle groups of the neck (Sternocleidomastoid, Cervical Paraspinous and Upper Trapezius), lower torso (Erector Spinae), and lower extremities (Rectus Femoris) to measure the muscle response of the subjects (Figure 2). A grounding electrode was centered over the right mastoidale. Signals from the muscle leads were passed to two battery-operated eight-channel FM transmitters (TeleMyo 2400T V2, Noraxon, Scottsdale, AZ) and recorded throughout each trial at 1,500 Hz per channel. Within the EMG acquisition system, the signals were amplified (gain 1000) with a single-ended amplifier (impedance >10 Mohm) and filtered with a fourth-order Butterworth filter (10–500 Hz) and common mode rejection ratio of 130 dB at direct current (minimum 85 dB across entire frequency of 10–500 Hz). Prior to sled testing, for each subject, the maximum voluntary isometric contraction (MVIC) for these muscles was measured during 10 seconds of attempted neck flexion, neck extension, torso extension and leg extension. Mean MVIC for each of the bilateral muscles was calculated by averaging with a 25 ms window over the middle six seconds of the entire duration of the isometric contraction trial.

Each subject performed a total of six dynamic trials on the low speed sled with an interval of 10 minutes between trials. Subjects were encouraged to fully relax prior to each trial and were provided with a countdown leading up to trigger press.

Signals from the accelerometers, angular rate sensor (ARS) and load cells were sampled at 10,000 Hz using a T-DAS data acquisition system (Diversified Technical Systems Inc., Seal Beach, CA) with a built-in anti-aliasing filter (4,300 Hz). The time series motion analysis and T-DAS data were imported into MATLAB (Mathworks, Inc., Natick, MA) for kinematic calculations using a custom written program.

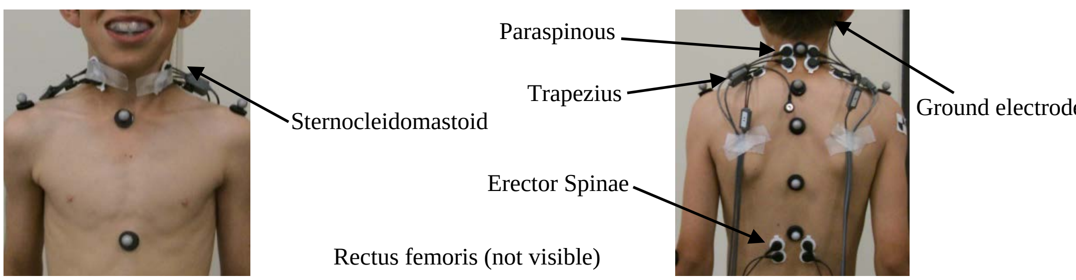

Figure 2: Surface EMG electrode locations on a child subject.

Using a customized MATLAB algorithm, the raw EMG signals were filtered using a Band-pass (20–510 Hz) Finite Impulse Response (using a Kaiser Window method) filter [Winter et al, 1980; Merletti et al, 1999; DeLuca et al, 2010]. A root-mean-squared (RMS) method with a 25 ms moving average smoothing window was applied, as it had the least effect on the EMG onset times (Figure 3). This process also served to rectify the signal. The EMG signals were analyzed with respect to several key event time points such as event onset (time zero, beginning of acceleration pulse) and time of maximum head excursion. The mean of the prestimulus noise (i.e. EMG signal from event onset to one second prior) for each muscle was subtracted from its respective signal. For each subject, the processed EMG signals were normalized by their respective MVIC.

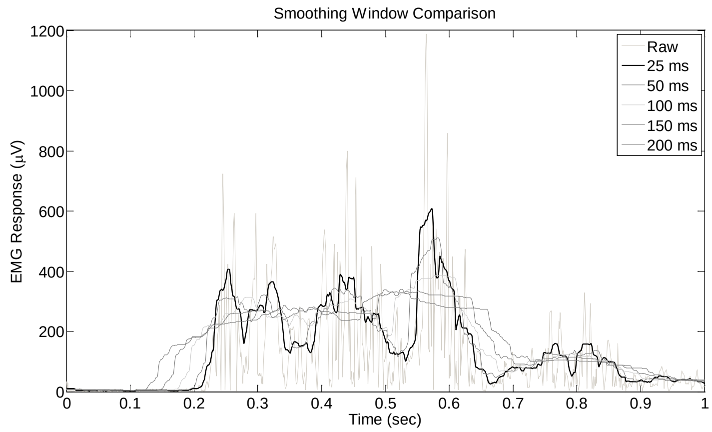

Figure 3: Raw EMG and Root mean squared (RMS) plots using 25, 50, 100, 150 and 200 ms smoothing windows for the right SCM muscle for a subject

Maximum forward head excursion (and the corresponding time) is defined as the maximum head top excursion in the x-direction. Maximum head rotation (and the corresponding time) is the maximum difference from the initial angle made by the nasion to external auditory meatus (EAM) mid-point relative to horizontal. Normalized peak EMG (and the corresponding time) was defined as the maximum magnitude of the normalized signal between event onset and time at maximum head excursion. EMG onset was defined as the time (between event onset and the time at peak EMG, or secondarily between trigger and event onset) at which the magnitude of the signal was 5% of the normalized peak value. These parameters were evaluated for all subjects, with age as a continuous variable. These parameters can be seen in Figure 4 below.

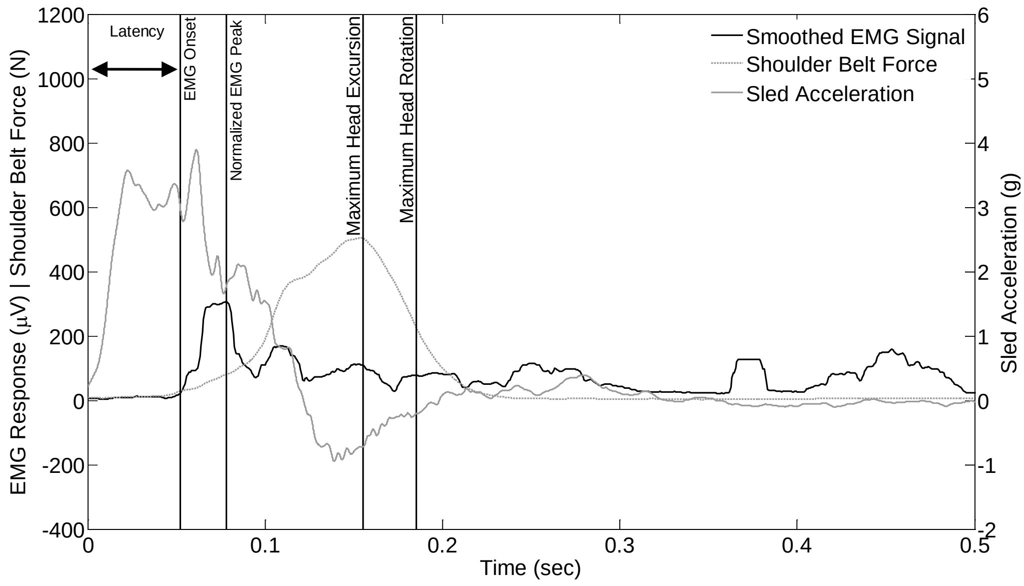

Figure 4: Sled acceleration (solid gray) superimposed over smoothed 25 ms RMS (solid black) data and CFC 60 filtered shoulder belt force (dashed gray) for an exemplar human volunteer's Cervical Paraspinous muscle, where event onset is time zero.

RESULTS

Eleven children and nine adult subjects were tested. The age and anthropometric variables for each subject are summarized in Table 1. Mean and standard deviation values for each EMG parameter are found in Appendix A, Tables 2-8.

No age-based differences were seen in any of the EMG parameters therefore the remaining analyses combined data from all subjects, all trials. Figures 5-7 compare the timing of the EMG and kinematic events with respect to each of the neck muscles. Figures 8-10 depict the relationship between the degree of maximum head rotation and each of the neck muscle's normalized peak EMG magnitude. All of the below figures are inclusive of all subjects, all trials, all measurements from the left and right electrodes.

Table 1: Key Anthropometric Subject Parameters

Subject #	Age	Height	Height %ile	Weight	Weight %ile	BMI	BMI %ile
	years	cm		kg			
1	8	140	94	34.5	90	17.6	79
2	10	144	68	33.1	47	16	32
3	10	138	27	40.4	80	21.5	92
4	11	151	83	35.8	41	15.5	14
5	11	134	8	31.3	18	17.3	50
6	12	165	92	50.3	74	18.5	54
7	12	155	68	40.4	41	16.8	29
8	12	143	12	40.4	43	20	76
9	13	164	85	65.3	94	24	93
10	13	150	20	46.7	54	20.8	78
11	14	173	67	60.8	68	20.4	59
8-14 yrs Average:		**150.6**	**56.7**	**43.5**	**59.1**	**18.9**	**59.6**
12	18	185	90	73.5	37	21.4	11
13	19	179	69	83.5	58	25.7	46
14	22	172	37.8	64.9	14.53	21.7	31.4
15	22	176	50.9	86.6	64.6	28	66.3
16	22	180	69.3	106.6	93.8	32.8	93.3
17	22	166	11	64.9	15	23.4	27
18	24	169	22	73.5	36.6	25.8	47.3
19	24	165	6	68.0	23	25	40
20	30	180	69	80.7	53	24.8	39
Adult Average:		**174.7**	**47.2**	**78.0**	**43.9**	**25.4**	**44.6**

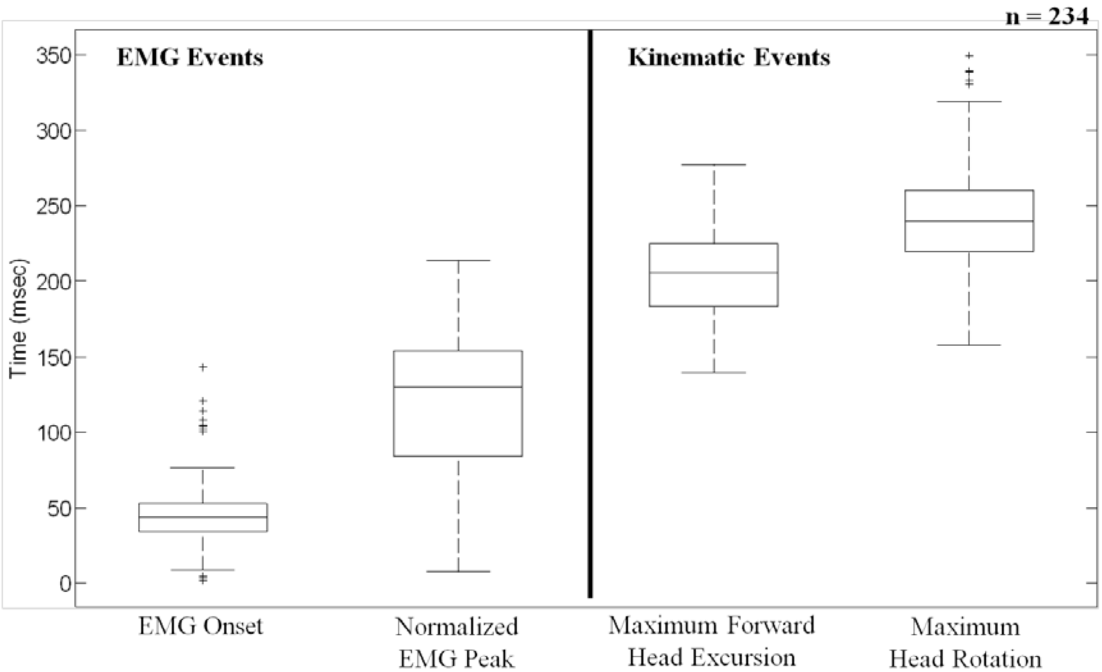

Figure 5: Comparison of EMG and Kinematic event timings for the Cervical Paraspinous

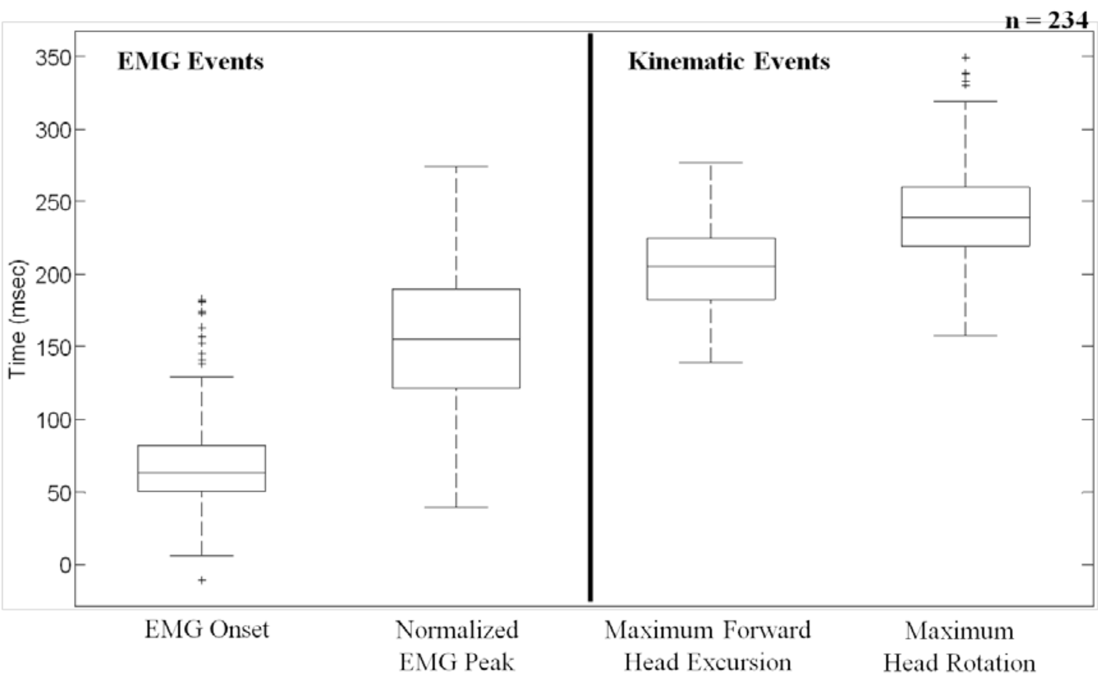

Figure 6: Comparison of EMG and Kinematic event timings for the Sternocleidomastoid

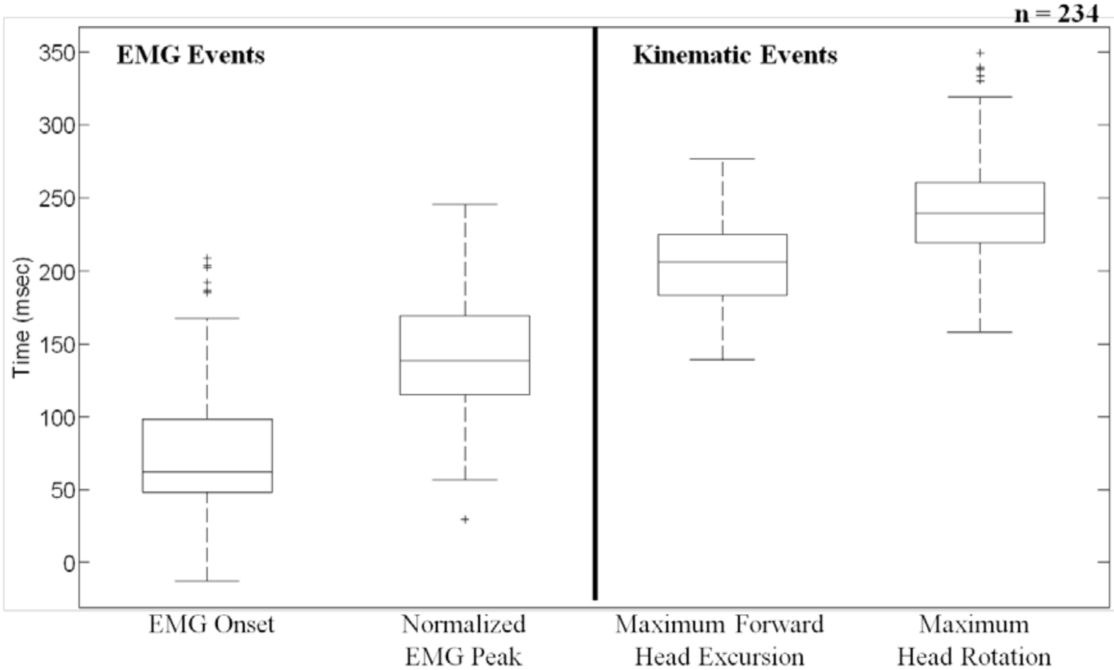

Figure 7: Comparison of EMG and Kinematic event timings for the Upper Trapezius

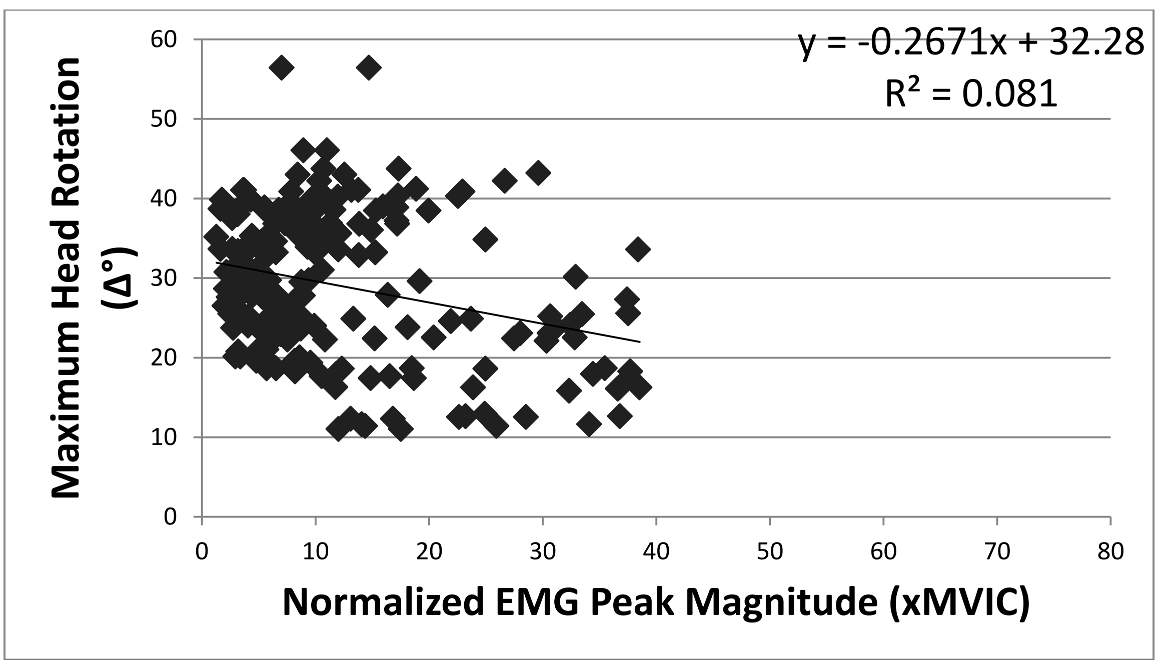

Figure 8: Maximum Head Rotation versus Normalized EMG Peak Magnitude for the Cervical Paraspinous

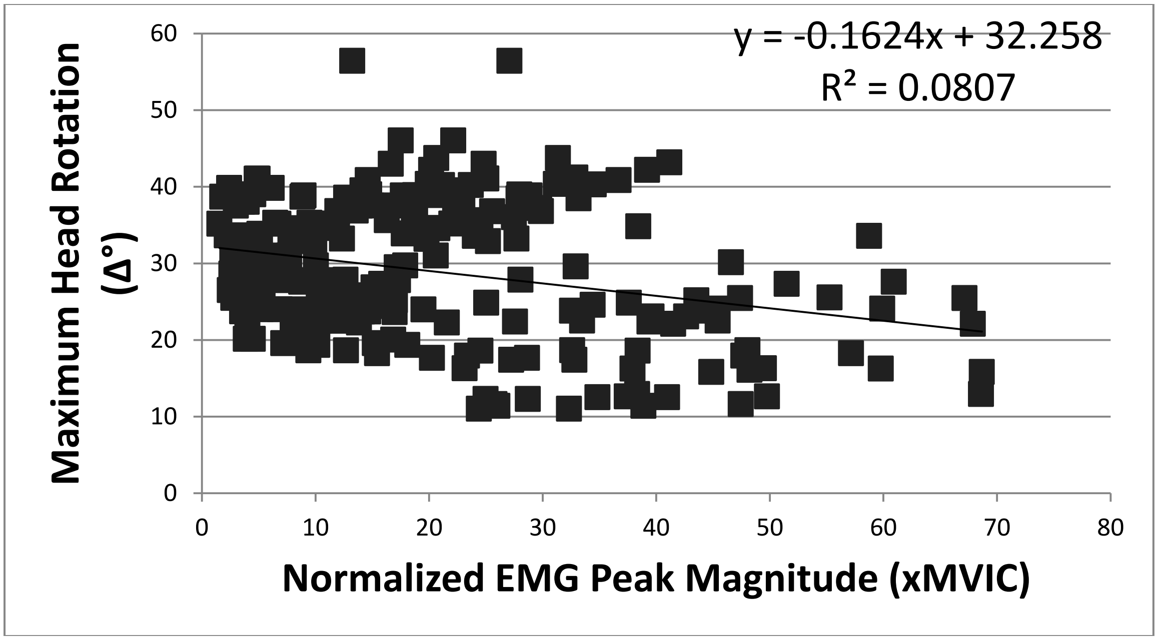

Figure 9: Maximum Head Rotation versus Normalized EMG Peak Magnitude for the Sternocleidomastoid

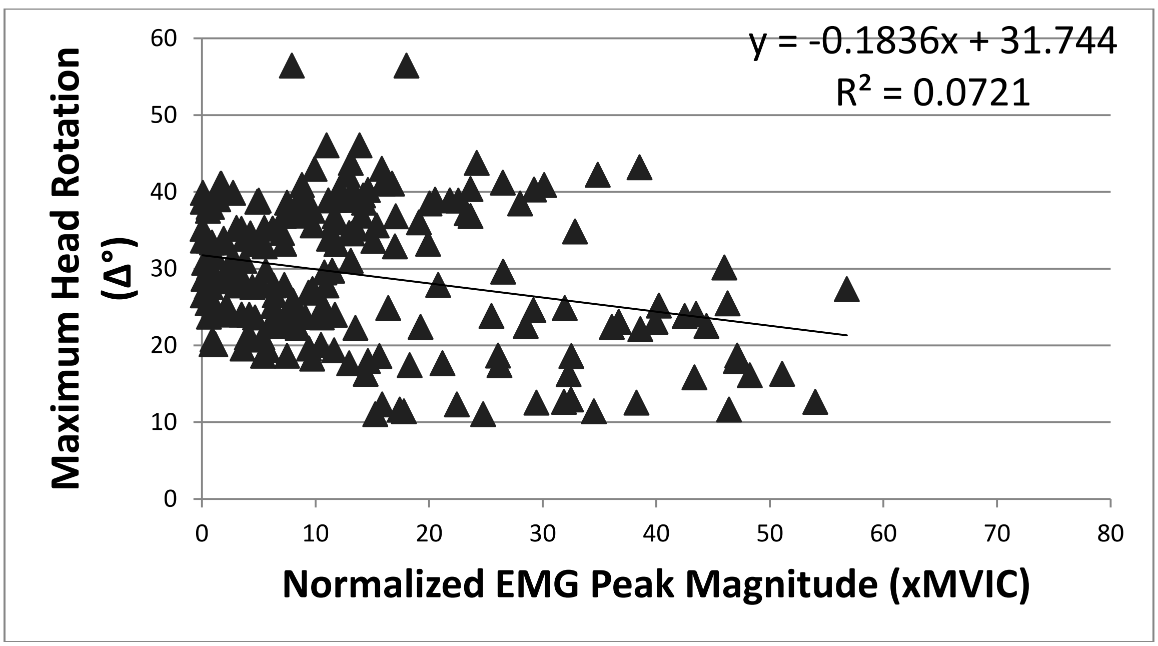

Figure 10: Maximum Head Rotation versus Normalized EMG Peak Magnitude for the Upper Trapezius

DISCUSSION

This study compared the muscle responses of the cervical spine with the head and neck excursions of children and adults in response to a low speed frontal impact.

From figures 5-7 it can be seen that all of the neck muscles achieve peak activation prior to time of maximum head rotation and maximum forward head excursion. Magnusson et al [1999] found that the Sternocleidomastoid reached its peak magnitude within the time of maximum head acceleration, when subjected to 0.5g. These findings indicate that in the low-speed frontal loading condition, muscle activation can influence occupant kinematics for the head and neck.

Additionally, a decreasing trend can be observed from figures 8-10, demonstrating that subjects with increased normalized EMG peak magnitude had a decreased maximum head rotation. Greater amplitudes of normalized EMG magnitude are indicative of greater resistive muscle force, or tenser muscles, which would thereby minimize skeletal movement. This could be extrapolated to other key body regions such as the torso and lower extremity. Further analysis should be performed to determine the effect of muscle activation in the torso and lower extremity on the respective region’s kinematics.

It is important to put these findings in context of response in an actual motor vehicle crash. Lopez-Valdes et al [2010] reported pulse duration of 135 msec with peak acceleration occurring at 57.5 msec in high-speed (40 kph) frontal acceleration sled tests with adult post-mortem human subjects (PMHS). Maximum head excursion was observed between 125-130 msec. The current study reports neck muscle activation IQR as 42-79 msec. The IQR for normalized peak muscle activation for the Cervical Paraspinous, Sternocleidomastoid, and Upper Trapezius spanned from 111-170 msec, encompassing the range of maximum head excursion during a high-speed frontal impact (Figure 11). Therefore, it could be inferred that the neck muscles may activate and achieve peak activation prior to time of maximum head excursion in a high-speed collision. This should be further examined to determine if muscle activity can govern the occupant kinematics during a motor vehicle collision.

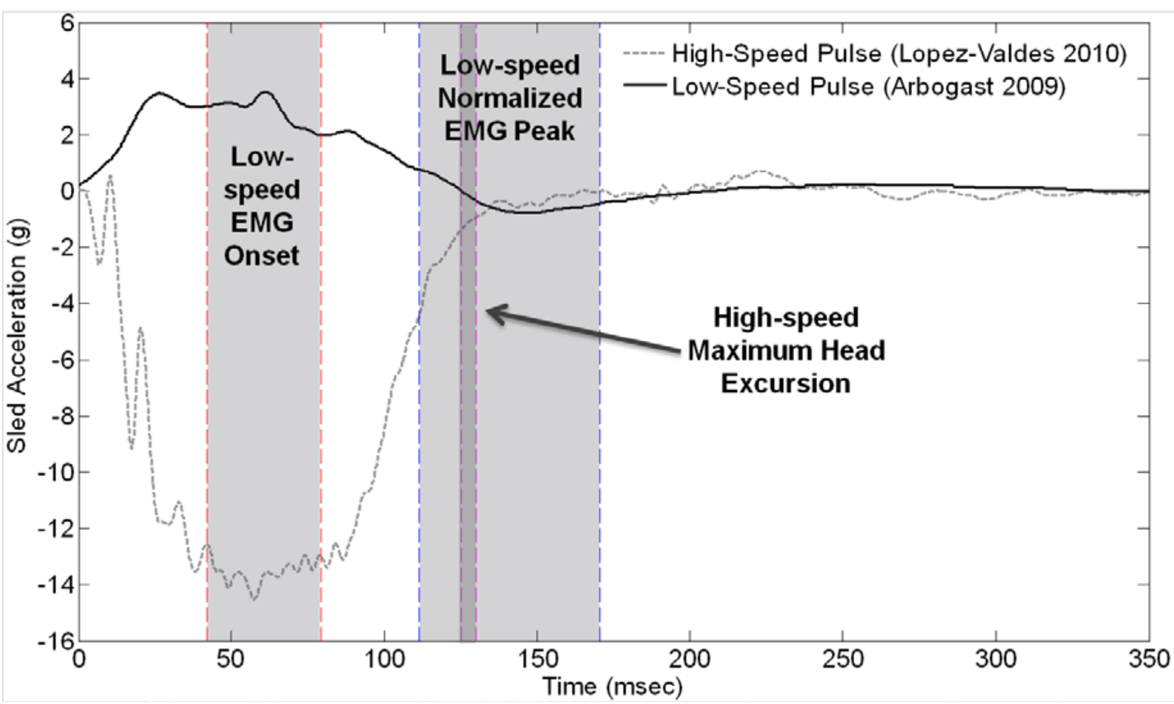

Figure 11: Comparison of low-speed and high-speed pulse, demonstrating EMG relevance in high-speed collision

There were several limitations regarding this study mainly associated with the data quality of sEMG signals. The use of sEMG, in contrast to fine wire EMG which is inserted directly into the muscle belly, allows for cross-talk between electrodes as well as the addition of noise from the overlaying skin and fat. Previous studies indicate that 20 mm inter-electrode spacing will create greater cross-talk amplitudes than 10 mm spacing [DeLuca et al, 2012]. To account for the cross-talk, we implemented the band-pass filter with a Kaiser window as described above, which is consistent with existing methods for processing signals from electrodes spaced 20 mm apart [DeLuca, 2010; Wittek et al, 2001]. The benefit of sEMG however is that it allows for signal acquisition across a greater surface area for a muscle of interest, allowing for a more comprehensive signal from the muscle as a whole rather than isolated data from a particular location within the muscle as seen with fine wire EMG. Signal processing helps to minimize the noise but may alter the relationship between key parameters and muscle group or age. Hence, processing algorithms were carefully chosen to ensure minimum alteration of the data signal. Additionally, normalization by mean MVIC may not be optimized due to the subject’s level of effort. Although subjects were encouraged to exert their maximum effort during the MVIC tests, there was no guarantee that the subjects follow these instructions. While the MVIC data may not be optimized, it was still subject specific and therefore indicative of that particular subject’s relative ability. By nature of the test, MVIC was measured as a static response but for the purposes of EMG data analysis, it was applied to a dynamic event.

Timing of muscle response in a crash event is crucial to understanding neuromuscular influence on the occupant kinematics. Activation and peak activation times reported in this study can be utilized in computational modeling of a motor vehicle occupant to activate key muscles in the model and introduce active human response to the simulation.

CONCLUSION

This data set provides a unique, exploratory glance at the relationship between neck muscle activity and head and neck kinematics for pediatric and young adult human volunteers subjected to low-speed frontal loading conditions. The timing of muscle activation onset and peak muscle activation is prior to maximum forward head excursion and maximum head rotation. It can also been seen that subjects with increased normalized peak EMG magnitude experienced decreased maximum head rotation. These data can be useful inputs into computational models of motor vehicle occupants to activate key muscles in the simulation.

ACKNOWLEDGEMENTS

The authors would like to thank all the human volunteers who participated in this study for their patience and willingness to take part in this research, Gunter P. Siegmund, PhD (MEA Forensics) for his guidance with EMG data collection and processing, and Department of Health and Exercise Science, Rowan University for use of its facilities. The authors would like to acknowledge Takata Corporation, Japan for their collaboration and financial support for this study. The results presented in this report are the interpretation solely of the author(s) and are not necessarily the views of Takata Corporation.

REFERENCES

ARBOGAST, K.B.; BALASUBRAMANIAN, S.; SEACRIST, T.; MALTESE, M.R.; GARCIA-ESPANA, F.; HOPELY, TERRANCE; CONSTANS, ERIC; LOPEZ-VALDES, F.J.; KENT, R.W.; TANJI, H.; HIGUCHI, K. Comparison of Kinematic Responses of the Head and Spine for Children and Adults in Low-Speed Frontal Sled Tests. Stapp Car Crash Journal 2009; Vol. 53.

BOSE, D; CRANDALL, J.R. Influence of Active Muscle Contribution on the Injury Response of Restrained Car Occupants. Proceedings of 52nd Annual Conference of the Association for the Advancement of Automotive Medicine Conference 2008; 61-72.

Centers for Disease Control and Prevention. 2010. "Motor Vehicle Safety." http://www.cdc.gov/Motorvehiclesafety/index.html

Centers for Disease Control and Prevention. 2000. "Growth Charts – 2000 CDC Growth Charts." http://www.cdc.gov/GrowthCharts/

CHOI, H.Y.; SAH, S.J.; LEE, B.; CHO, H.S.; KANG, S.J.; MUN, M.S.; LEE, I.; LEE, J. Experimental and Numerical Studies of Muscular Activations of Bracing Occupant. Proceedings of Enhanced Safety of Vehicles Conference. 2005;

DELUCA, C.J.; GILMORE, L.D.; KUZNETSOV, M.; ROY, S.H. Filtering the Surface EMG Signal: Movement Artifact and Baseline Noise Contamination. Journal of Biomechanics 2010; 43(8): 1573-1579.

DELUCA, C.J.; KUZNETSOV, M.; GILMORE, L.D.; ROY, S.H. Inter-electrode Spacing of Surface EMG Sensors: Reduction of Crosstalk Contamination During Voluntary Contractions. Journal of Biomechanics 2012; 45: 555-561.

EJIMA, S.; ONO, K.; HOLCOMBE, S.; KANEOKA, K.; FUKUSHIMA, M. A Study on Occupant Kinematics Behavior and Muscle Activities during Pre-Impact Braking Based on Volunteer Tests. Proceedings of International Research Council on the Biomechanics of Injury 2007; 31-45.

EJIMA, S; ZAMA, Y; ONO, K; KANEOKA, K; SHIINA, I; ASADA, H; Prediction of Pre-Impact Occupant Kinematic Behavior Based on the Muscle Activity during Frontal Collision. Proceedings of Enhanced Safety of Vehicles Conference. 2009;

KUMAR, S; NARAYAN, Y; AMELL, T. Role of Awareness in Head-Neck Acceleration in Low Velocity Rear-End Impacts. Accident Analysis and Prevention 2000; 32: 233-241.

KUMAR, S; NARAYAN, Y; AMELL, T. An Electromyographic Study of Low-Velocity Rear-End Impacts. Spine 2002; 27(10): 1044-1055.

KUMAR, S.; NARAYAN, Y.; AMELL, T. Analysis of Low Velocity Frontal Impacts. Clinical Biomechanics 2003; 18: 694-703.

KUMAR, S.; FERRARI, R.; NARAYAN, Y.; JONES, T. The Effect of 3-Point Harness Restraint and Car Seat in Whiplash-Type Lateral Impacts. Spine 2006; 31(1): E11-E18.

LOPEZ-VALDES, F.J.; LAU, A; LAMP, J; RILEY, P; LESSLEY, D.J.; DAMON, A; KINDIG, M; KENT, R; BALASUBRAMANIAN, S; SEACRIST, T; MALTESE, M.R.; ARBOGAST, K.B; HIGUCHI, K; TANJI, H; Analysis of Spinal Motion and Loads during Frontal Impacts. Comparison between PMHS and ATD. Proceedings of the 54th Annual Conference of the Association for the Advancement of Automotive Medicine Conference 2010.

MAGNUSSON, M.L.; POPE, M.H.; HASSELQUIST, L.; BOLTE, K.M.; ROSS, M.; GOEL, V.K.; LEE, J.S.; SPRATT, K.; CLARK, C.R.; WILDER, D.G. Cervical Electromyographic Activity During Low-Speed Rear Impact. European Spine Journal 1999; 8: 118-125.

MERLETTI, R. Standards for Reporting EMG Data. Journal of Electromyography and Kinesiology 1999; 9(1).

NHANES, 1994. "Center for Disease Control and Prevention Anthropometric reference data (1988-1994)." http://www.cdc.gov/nchs/about/major/nhanes/anthropometric_measures.htm

SIEGMUND, G; SANDERSON, D; MYERS, B; INGLIS, J; Awareness Affects the Response of Human Subjects Exposed to a Single Whiplash-Like Perturbation. Spine 2003; 28(7): 671-679.

WITTEK, A.; ONO, K.; KAJZER, J.; ORTENGREN, R.; INAMI, S. Analysis and Comparison of Reflex Times and Electromyograms of Cervical Muscles Under Impact Loading Using Surface and Fine-Wire Electrodes. IEEE Transactions on Biomedical Engineering 2001; 48(2): 143-153.

WINTER, D.A.; RAU, G.; KADEFORS, R.; BROMAN, H.; DELUCA, C.J. Units, Terms, and Standards in the Reporting of EMG Research: A Report by the AdHoc Committee of the International Society of Electrophysiology and Kinesiology 1980;