INJURY BIOMECHANICS RESEARCH
Proceedings of the Thirty-Sixth International Workshop

Electromyography Responses of Pediatric and Adult Volunteers in Low Speed Frontal Impacts

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ABSTRACT

Several studies have examined the electromyography (EMG) responses of the neck muscles during frontal, rear and side impacts in the adult population (Kumar et al, 2002; Kumar et al., 2004; Kumar et al., 2006; Magnusson et al., 1999, Siegmund et al., 2003). No data exists on the EMG responses in children exposed to dynamic impacts. The objective of the current study was to measure the EMG responses of the neck, torso and lower extremity muscles in children and adults during a low speed frontal impact. Low speed volunteer testing of pediatric (n=4, ages 8-12 years) and adult (n=3, ages 18-24 years) male subjects were performed using a pneumatically actuated – hydraulically controlled sled. Safe limits were established from measurement of bumper car accelerations at an amusement park ride (4.9 g, 55.7 msec rise time, 110 msec duration), a sub-injurious activity to the adult and child amusement park population. We recreated the bumper car environment in the laboratory, by developing a low-speed hydro-pneumatic sled. As an added measure of safety, our average maximum cart acceleration was 3.7 g for children and 3.5 g for adults, thus producing occupant loads that are approximately 25% less than the bumper car amusement park ride. Surface EMG electrodes were placed bilaterally on the neck (sternocleidomastoid, cervical paraspinous, trapezius), lower torso (erector spinae) and lower extremity (rectus femoris) muscles. Maximum voluntary isometric contraction (MVIC) measurements were made while the subjects exerted their maximum isometric effort in attempted neck flexion, neck extension, torso extension and leg extension. Mean MVIC for each muscle was computed by averaging with a 25 ms moving window over the middle 6 sec duration of the entire isometric contraction trial. The EMG responses were normalized against the subjects' mean MVIC values. The mean MVIC, and the timing and magnitude of the normalized EMG responses during frontal impacts, were compared between the pediatric and adult groups. These data could be used to model active musculature in computational models used in impact biomechanics studies.

INTRODUCTION

Traumatic brain and skull injuries are the most common serious injuries sustained by children in motor vehicle crashes regardless of age group, crash direction, or restraint type (Arbogast et al., 2005; Arbogast et al., 2002; Durbin et al., 2003; Howard et al., 2003; Orzechowski et al., 2003; Arbogast et al., 2004). Head injuries are responsible for one-third of all pediatric injury deaths (Adekoya et al., 2002; Thompson and Irby, 2003) and are particularly relevant clinically as the developing brain is difficult to evaluate and treat. The neck muscles play an important role in the postural control of the head and neck motion during a dynamic crash event (Kumar et al., 2001). Several studies have examined the electromyography (EMG) responses of the neck muscles during frontal, rear and side impacts in the adult population.

In response to rapid acceleration, a centrally generated response is responsible for the postural control of the head, neck and trunk (Forssberg and Hirschfeld, 1994; Magnusson et al., 1999; Vibert et al., 2001). For forward trunk acceleration, it has been shown that sensory inputs (visual, vertibular and proprioceptive) that detect body movement elicit cervicocollic (CCR) and vestibulocollic (VCR) reflexes to interact with the central nervous system to modulate the neck muscle response (Siegmund et al. 2002; Blouin et al., 2002; Forssberg and Hirschfeld, 1994; Vibert et al., 2001). Although, cadaver models have been used to study head and neck responses to rapid trunk acceleration (Yoganandan et al., 1999), they cannot demonstrate appropriate neuromuscular control strategies required for head and neck stabilization. It has been demonstrated that the neck muscle responses and magnitude varies during isometric activity in eight different directions around the head (Kumar et al., 2002). Also, the neck muscle response slightly varies with torso posture during low speed frontal impacts. However, it has been shown that a forward-flexed trunk posture does not increase the likelihood of cervical spine muscle injury when compared to a neutral trunk posture (Kumar et al., 2006).

Several studies on low speed rear-impacts on human volunteers have shown the influence of muscle contractions on the head and neck kinematics. It has been hypothesized that the muscle reflex would stiffen the head-neck complex and thereby reduce the risk of injury by decreasing excursion (Brault et al., 2000; Kaneoka et al., 1999; Kumar et al., 2002; Magnusson et al., 1999). This hypothesis was largely based on relative kinematics of the head and thorax in relation to timing of the EMG activity of the neck muscles (Stemper et al., 2005). Studies have reported higher EMG onset timing and latency for the neck muscles during voluntary experiments (Mazzini and Schieppati, 1992; Siegmund et al., 2001) as compared to whiplash-like perturbations (Ono et al., 1997; Magnusson et al., 1999; Brault et al., 2000). Attenuation of the EMG activity, also known as habituation have also been observed with repeated exposure in postural (Nashner, 1976; Keshner et al., 1987; Hansen et al., 1988; Woollacott et al., 1988; Allum et al., 1992; Bisdorff et al., 1994; Timmann and Horak, 1997), startle (Landis and Hunt, 1939; Davis, 1984; Brown et al., 1991) and whiplash-like experiments (Kumar et al., 2000, 2002; Siegmund et al., 2003;) in adult human volunteers. No data exists on the EMG responses in children exposed to dynamic impacts.

The objective of this research was to develop a methodology to measure the EMG responses of the neck, torso and lower extremity muscles in children and adults exposed to a dynamic sub-injurious frontal crash pulse. This paper describes the methodological development of the test protocol and provides exemplary data from both the child and adult test subjects.

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.

Test device

A pneumatically actuated – hydraulically controlled 'low-speed acceleration seating buck' (LASB) shown in Figure 1a, was designed to subject restrained human volunteers to a sub-injurious, low-speed frontal crash pulse.

The LASB is primarily comprised of three sub-assemblies, namely frame, actuator and seating buck. The frame for the LASB was constructed of extruded aluminum tubing (MiniTec Framing Systems LLC, Victor, NY). The structural framework included a platform (for the actuator assembly) which was rigidly connected to two 18 feet long parallel support rails with equally spaced cross members for rigidity. A steel bar between the two support rails served to slow the sled to a stop following the primary acceleration pulse. The actuator assembly was comprised of a pneumatic actuator (Mc Master-Carr, Robbinsville, NJ) (diameter

– 4 inches, stroke length – 20 inches, operating pressure – 200 psi) connected to an opposing dual hydraulic piston-cylinder (Model TZ22, Vickers Cylinders, Eaton Corporation, Cleveland, OH) arrangement using a rigid frame. A 2-way high dynamics proportional throttle cartridge valve (Model LIQZO-LE, Atos, Italy) was used in the custom-designed hydraulic circuit to control the displacement profile of the pneumatic actuator. When the pneumatic actuator was fired, it delivered the impact force to the seating buck.

The seating buck assembly (Figure 1b) framework was also constructed using extruded aluminum tubing (MiniTec Framing Systems LLC, Victor, NY). It was comprised of a moving platform mounted on the two support rails by means of six low friction linear bearings. A custom-built impact fixture was mounted on the platform to transfer the force from the pneumatic actuator to the moving platform. A rigid low-back padded seat, an adjustable height shoulder belt anchor post (similar to a B-pillar in an automobile), lap belt anchors and an adjustable footrest were mounted on the platform. The low-back seat allowed for the motion analysis markers along the spine to be visible to the cameras. A standard automotive three-point belt system was attached to the lap belt and shoulder belt anchor points. An onboard pneumatic braking system was provided to interact with the braking rail to decelerate the moving platform. In order to limit the excursion of the subject during rebound associated with braking, a nylon strap was attached to two vertical bars behind the seat (at the location of T4).

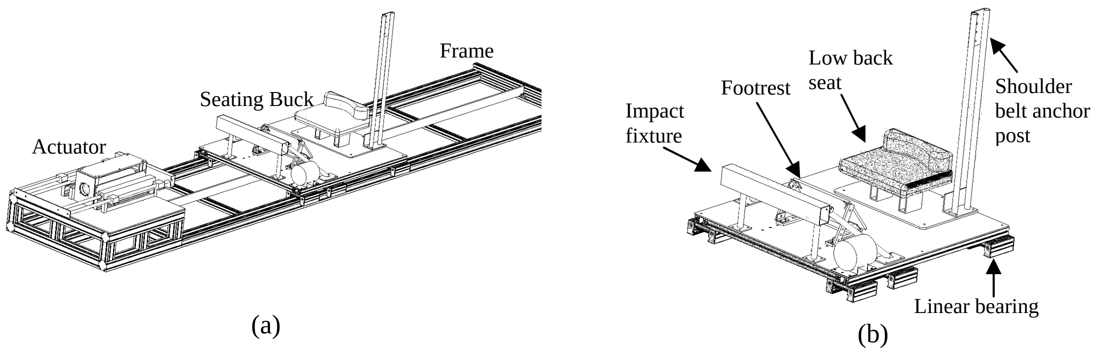


Figure 1: (a) Schematic of the low-speed acceleration seating buck (LASB) (b) Schematic of the seating buck assembly.

Safe volunteer crash pulse

An amusement park bumper car ride was studied to provide a benchmark of a crash-like situation commonly and safely used by children for recreation and enjoyment. Safe limits on the volunteer crash pulse were defined from measuring a bumper car-to-wall impact in an amusement park (Six Flags Great Adventure, Jackson, NJ). An accelerometer was secured to the rigid cross-member of the steering assembly of a bumper car. The car was used in its typical usage patterns, impacting the wall of the arena, another vehicle head to head, and another vehicle in a T-type configuration. This process was repeated with two different bumper car vehicles. The maximum pulse obtained was 4.9 *g* in 55.7 msec (Figure 2). This was defined as the envelope of safety for the human volunteers. The hydraulically controlled – pneumatic powered actuator system was designed to deliver an acceleration pulse with a maximum acceleration of less than 4.5 *g* with a rise time of 50-70 msec – within the defined safety envelope. However, the subjects received a slightly lower pulse (shown in the results section). Several safety checks also ensured that the LASB delivered the appropriate pulse.

Human Subjects

Inclusion criteria. Specific inclusion criteria were male subjects aged between 6 and 40 years whose height, weight and BMI were within 5th and 95th percentile for the subject's age (based upon CDC growth charts for children (CDC Growth Charts, 2000) and CDC NHANES data for subjects 18+ years (NHANES data, 1994)). Subjects with existing neurologic, orthopedic, genetic, or neuromuscular conditions, any previous injury or abnormal pathology relating to the head, neck or spine were excluded from the study. Subjects were recruited from flyers placed in the community and throughout CHOP and Rowan sites. Prior to the testing dates, telephone interviews were conducted with the adult subjects and parent / guardian of child subjects to confirm eligibility.

For the analyses presented herein, a total of 7 male subjects – four subjects in 8-12 years and 3 subjects in 18-24 years age groups – were tested. Upon arrival at the test site, the study was explained in detail to all subjects including a demonstration of how the LASB functions by firing the sled without an occupant. The adult subjects were given a self-consent letter and the parent / guardian of the child subjects were given a parental consent letter with a child subject assent. After the subjects had been consented, height and weight were measured to verify that their height, weight and body mass index (BMI) were consistent with the inclusion criteria. The subjects were asked to remove their shirt(s) and anthropometric measurements of the head, neck, torso and extremities were recorded.

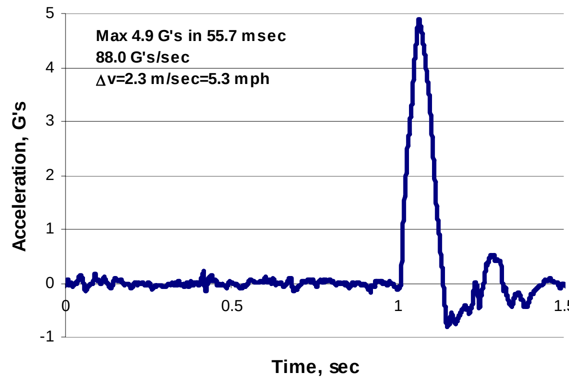


Figure 2: Bumper car to wall acceleration pulse.

Instrumentation

Subject. Prior to EMG electrode placement, each subject's skin was cleaned by applying NUPREP Skin Prepping Gel (Weaver and Co., Aurora, CO). Disposable, self-adhesive dual surface EMG electrodes (Noraxon, Inc., Scottsdale, AZ) were placed bilaterally on key muscle groups of the neck (Sternocleidomastoid – SCM, Paraspinous – PS and Trapezius – TZ), lower torso (Erector Spinae – ES), and lower extremities (Rectus femoris – RF) to measure the muscle response of the subjects to the loading environment (Figure 3). 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 1,000 Hz per channel. 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). For each subject, the maximum voluntary isometric contraction (MVIC) for these muscles was measured during attempted neck flexion, neck extension, torso extension and leg extension prior to sled testing.

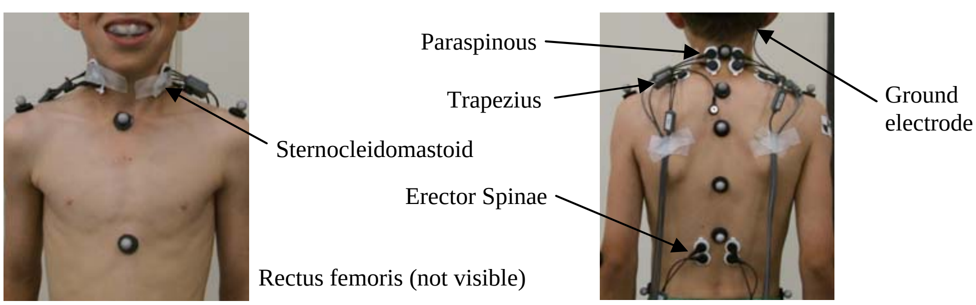


Figure 3: Surface EMG electrode locations on a child subject.

Spherical reflective markers (10 mm diameter) were placed on various anatomical landmarks on the head, neck, torso, upper and lower extremities and various locations on the seating buck and tracked using a 3D motion analysis system (Model Eagle 4, Motion Analysis Corporation, Santa Rosa, CA). An angular rate sensor – ARS (Model ARS-300, DTS Inc, Seal Beach, CA) was mounted via a custom fixture to a subject-specific athletic mouth guard to measure the head rotational velocity.

LASB. Spherical reflective markers were also placed on various locations on the seating buck and tracked using a 3D motion analysis system (Model Eagle 4, Motion Analysis Corporation, Santa Rosa, CA). A piezoresistive accelerometer (Model 7264-200, Endevco, San Juan, CA) was mounted to the moving platform frame to record the acceleration of the LASB. Lightweight belt webbing load cells (Model 6200FL-41-30, Denton ATD Inc, Rochester Hills, MI) were attached five inches from the D-ring location on the shoulder belt and on the inboard and outboard locations on the lap belt, respectively. Six-axis load cells was placed under the seat pan (Model IF-217, FTSS, Plymouth, MI) and footrest (Model IF-234, FTSS, Plymouth, MI), respectively to measure the reaction forces exerted by the subjects.

A high-speed video camera (MotionXtra HGTH, Redlake, San Diego, CA) was placed sagittally to record the event at a rate of 1,000 frames per second (fps). In addition, two standard video camcorders (Model DC20, Canon Inc., Japan) were used to capture the frontal and sagittal views at 30 fps. The hydraulic controller, Motion Analysis, T-DAS, EMG and high speed camera systems were triggered synchronously using a custom made circuit.

Testing

After the instrumentation setup was completed, the subjects were seated in the LASB as shown in Figure 4. The torso and knee angles were maintained at 110 degrees by adjusting the position of the footrest and nylon strap to mimic the posture of a seated occupant in an automobile (Reed et al. 2005). The shoulder belt angle at the D-Ring (defined as the angle the shoulder belt makes with the horizontal) and lap belt buckle angle (defined as the angle the lap belt buckle makes with the horizontal) were set at 70 degrees at initial position for all the subjects. In order to minimize the effect of initial head position, the subjects were asked to focus at a point placed directly in front of them at the level of their nasion. The lap and shoulder belts were then adjusted and secured to fit optimally for the subject's size.

The experimental procedure with the LASB is a series of six tests, with each successive test designed to encourage the occupant to relax their muscles and allow the restraints to support their weight during the acceleration event, thus simulating the condition of an unbraced occupant in a frontal vehicle crash whose inertial forces are supported by the restraint system in the vehicle. Subjects were aware of the impending impact and received a countdown in each test prior to firing of the actuator. Each subject was given the option to either continue or withdraw from further testing at the completion of each test run. All the tests were conducted identically with a rest period of approximately 10 minutes between subsequent tests.

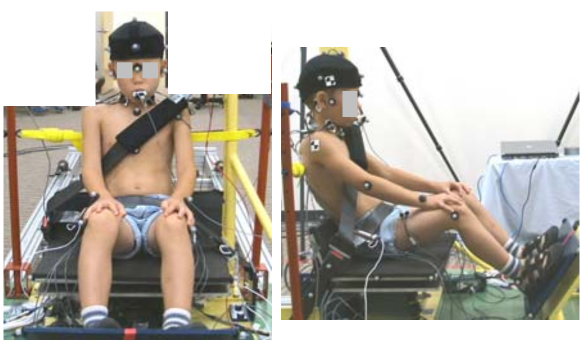


Figure 4: Child subject seated in LASB.

Data acquisition and analyses

Signals from the ARS, accelerometer 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 Motion Analysis data were acquired at 100 Hz and analyzed using EVaRT5 software (Motion Analysis Corporation, Santa Rosa, CA). MyoResearch XP software was used to export the EMG data into ASCII format. The T-DAS, Motion Analysis and EMG data processing were automated using MATLAB 8.0 (The Mathworks Inc, Natick, MA). The high-speed video data were analyzed with Falcon software (Falkner Consulting for Measuring Technology GmbH, Gräfelfing-Lochham, Germany).

For the analyses presented herein, only the sled acceleration data and EMG data will be discussed. The sled acceleration data were filtered at SAE channel frequency class (CFC) 60, as recommended by the SAE J211 standards. The raw EMG data were rectified and high-pass filtered (60 Hz) to remove motion artifacts before calculating the root mean squared (RMS) values. The RMS of each muscle's EMG signal was calculated using a 25, 50, 100, 150 and 200 ms sliding windows in order to determine the optimal smoothing method. For each muscle, the RMS amplitude of the EMG signal was calculated for the interval between EMG onset and 600 ms following EMG onset. The mean values of the middle 6 s of rectified EMG data from the pre-test MVIC measurements were used to normalize the test EMG data. The EMG onset was defined as the time at which the RMS amplitude reached 5% of its maximum value. This onset time was also confirmed visually. EMG latency was the time delay between event onset and EMG onset. Pre-stimulus activity was quantified using the RMS amplitude of the EMG signal over the entire duration preceding the EMG onset and subtracted from the pre-normalized smoothed EMG signal.

RESULTS

The age, height, weight, BMI and the percentiles for the subjects whose data are presented herein are listed in Table 1.

Table 1.
Height, Weight, BMI and percentile for subjects

Subject #	Age years	Height cms	Height Percentile	Weight kgs	Weight Percentile	BMI kg/m^2	BMI Percentile
17	8	140	94	34.4	90	17.6	79
18	10	144	68	33.1	47	16	32
16	12	165	92	50.3	74	18.5	54
19	12	155	68	40.3	41	16.8	29
21	22	172	38	64.8	14	21.7	31
23	22	176	51	86.6	65	28	66
22	24	169	22	73.4	37	25.8	47

For pediatric subjects, ages 6-18 years: Height, Weight and BMI percentiles were calculated using the CDC growth charts – Published May 30, 2000

For Adult subjects, ages >20 years: Height, Weight and BMI percentiles were calculated using the NHANES data 1988 – 1994

The individual and averaged sled acceleration pulse for a set of six trials on a single subject is shown on Figure 5a. The activation of the synchronous trigger (henceforth called 'time zero') was followed by a time delay before the movement of the sled (event). The time delay (approximately 100 msec) was attributed to the response lag associated with the LASB hydraulic system. Event onset (vertical line in Figure 5a) was defined as the time at which the sled acceleration reached 5% of its peak value.

The five phases of the event are outlined below:

1. Acceleration – This is the first phase of the event that immediately follows event onset and corresponds to the pre-programmed acceleration pulse of the sled.
2. Restraint loading – The subject loads the seatbelt restraints.
3. Rebound – After maximum excursion, the subject rebounds back and interacts with the nylon strap behind the seat.
4. Coasting – During this phase, the sled coasts on the rails.
5. Braking – The pneumatic brakes are applied during this phase causing the gradual deceleration of the sled.

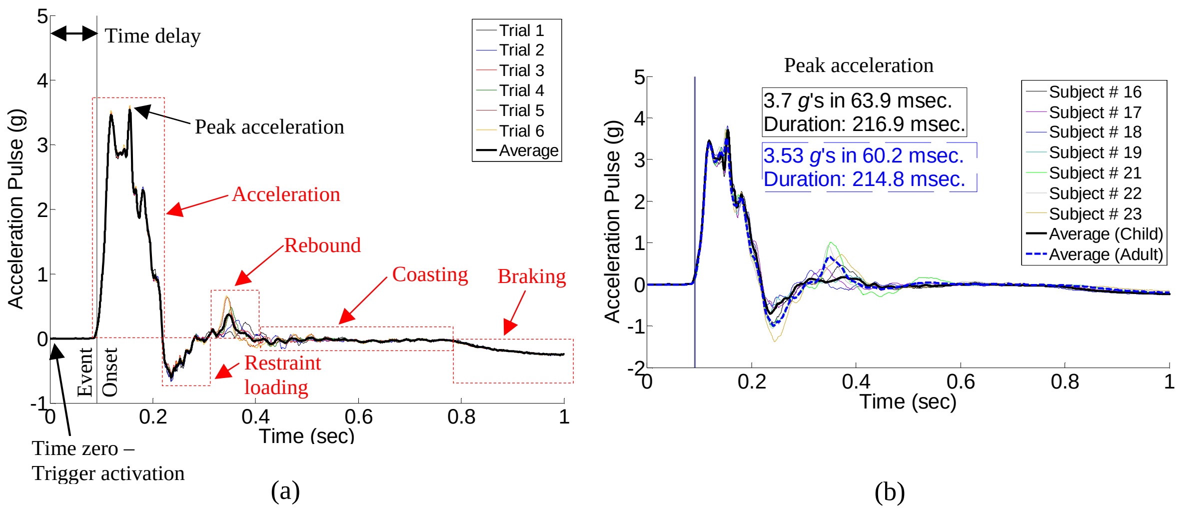


Figure 5: (a) Sled acceleration pulse of six individual trials and their average for a single human volunteer. Various phases of the event are shown in red – dashed boxes (b) Average sled acceleration pulse with peak values for child and adult subjects.

The sled acceleration pulse from the six trials for each subject were averaged and plotted for the two age groups – child and adult (Figure 5b). The children had slightly higher average peak acceleration (3.7 g) with slightly longer rise time (63.9 msec) compared to adults (3.5 g in 60.2 msec).

The rectified raw EMG data for the right SCM muscle along with the RMS signals using 25, 50, 100, 150 and 200 ms sliding windows for a single subject are shown on Figure 6. The 25 ms RMS sliding window was chosen as smoothing algorithm for further analysis as it had the least amount of time shift when compared to the raw data. Figure 7 shows the raw EMG data and the 25 ms smoothed signals superimposed on sled acceleration.

The average EMG onset, event onset and EMG latency for all muscle groups (averaged bilaterally) for the child and adult subjects across six trials are shown in Table 2. On an average, the SCMs were observed to have a shorter latency when compared to the other neck muscles (PS and TZ) in the child and adult subjects. The unnormalized EMG data for the right SCM muscle for a child and adult subject are shown on Figures 8a and 8b, respectively. In order to compare the EMG data between the two age groups, the data were normalized using the mean MVIC for each specific muscle. The normalized data for bilateral SCM are shown on Figures 9a and 9b, for the child and adult subjects. Although the unnormalized child data (for SCM) have a greater peak value when compared to adults, the normalized data show an opposite trend.

To study the effect of repeated exposure on EMG response, the normalized EMG responses for the right SCM were plotted for all six trials for a child and adult subject (Figures 10a and 10b). No significant attenuation of the EMG signals was observed with repeated exposure. The individual mean normalized EMG response and their average for the right SCM for the child and adult subjects across six trials is shown in Table 3.

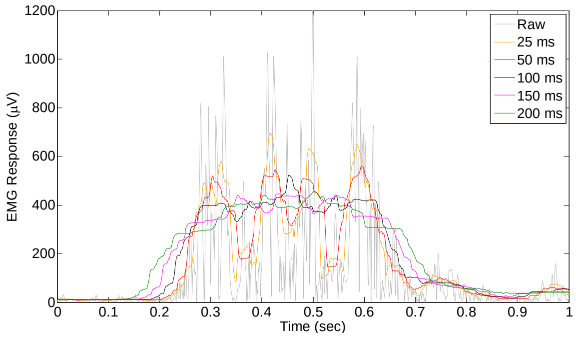


Figure 6: Raw EMG and Root mean squared (RMS) plots using 25, 50, 100, 150 and 200 ms smoothing windows for the right SCM muscle for a subject.

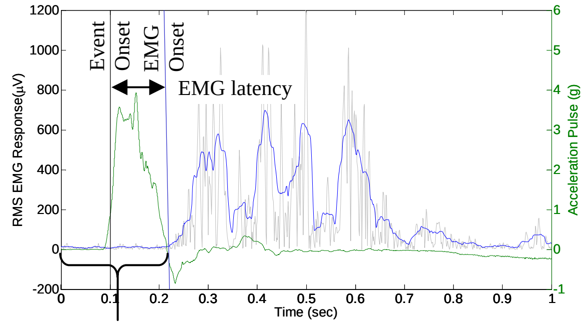


Figure 7: Sled acceleration superimposed over raw EMG data and smoothed 25 ms RMS data for a single human volunteer.

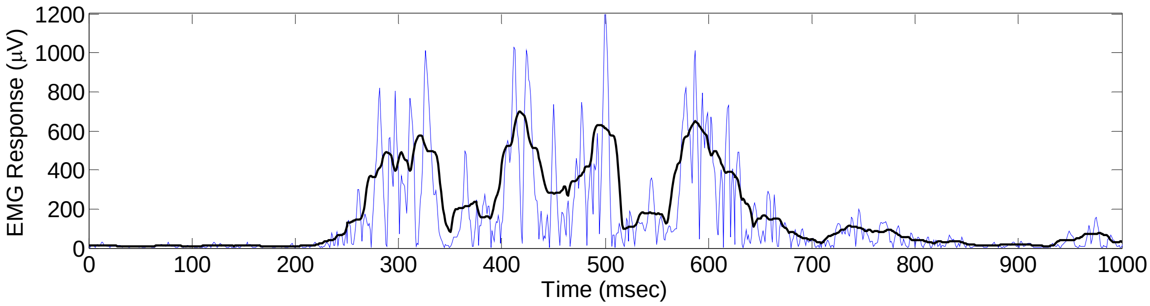
Table 2. (a) Average EMG and event onset and (b) average EMG latency for the child and adult subjects across six trials.

(a)

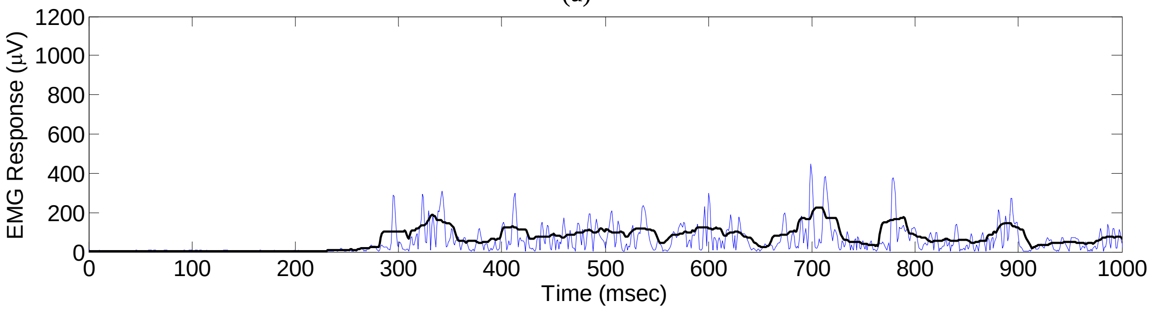
Subject #	Age (years)	Average EMG Onset Across Six Trials (sec)					Average Event Onset (sec)
		SCM	PS	TZ	ES	RF	
17	8	0.0940	0.2026	0.1599	0.1248	0.1906	0.0917
18	10	0.1900	0.2390	0.1742	0.2047	0.1024	0.0899
16	12	0.2180	0.1926	0.1888	0.2059	0.2143	0.0928
19	12	0.1476	0.2286	0.1030	0.2170	0.1677	0.0907
21	22	0.1675	0.2442	0.2451	0.1166	0.1762	0.0932
23	22	0.1681	0.1723	0.1908	0.1768	0.1901	0.0923
22	24	0.2040	0.2755	0.2461	0.2395	0.1974	0.0916

(b)

Subject #	Age (years)	Average EMG Latency (sec)				
		SCM	PS	TZ	ES	RF
17	8	0.0023	0.1109	0.0683	0.0331	0.0989
18	10	0.1002	0.1491	0.0843	0.1148	0.0125
16	12	0.1253	0.0998	0.0960	0.1131	0.1215
19	12	0.0569	0.1379	0.0123	0.1263	0.0770
21	22	0.0743	0.1510	0.1518	0.0234	0.0830
23	22	0.0758	0.0800	0.0985	0.0846	0.0978
22	24	0.1123	0.1839	0.1545	0.1479	0.1058

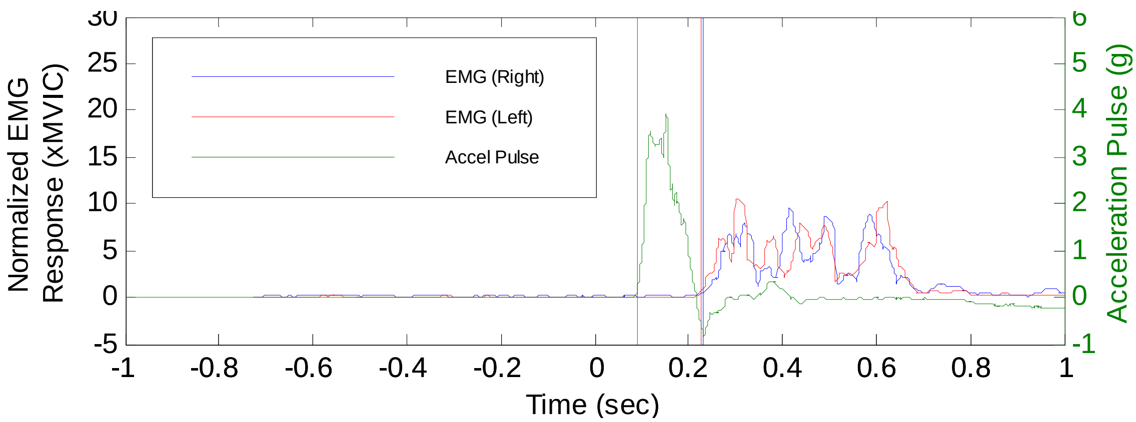


(a)

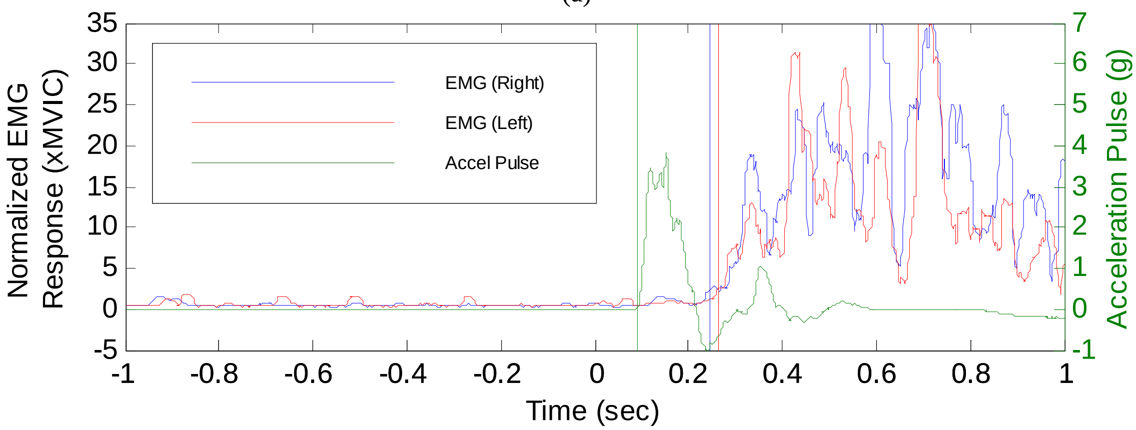


(b)

Figure 8: Unnormalized raw and smoothed EMG data for right-SCM muscle for a (a) child and (b) adult subject.

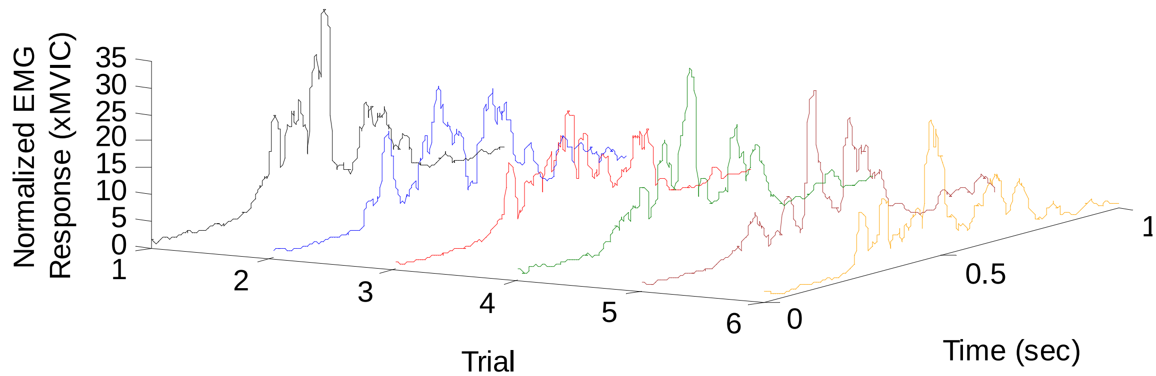


(a)

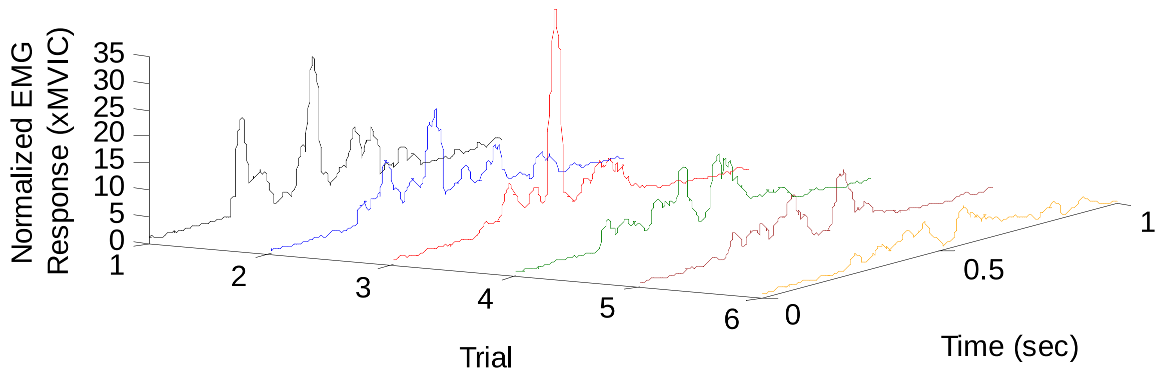


(b)

Figure 9: Sled acceleration along with Normalized EMG data for left and right SCM muscles for a (a) child and (b) adult subject.



(a)



(b)

Figure 10: Individual traces of normalized EMG response for the right SCM muscle for a (a) child and (b) adult subject.

Table 3. Individual and average normalized mean EMG data for the right SCM muscle the child and adult subjects across six trials.

Subject #	Age (years)	Mean RMS EMG Response (x MVIC)						Average (x MVIC)
		Trial 1	Trial 2	Trial 3	Trial 4	Trial 5	Trial 6	
17	8	0.1497	1.0806	1.0071	0.8350	0.6671	1.3963	0.8560
18	10	3.1319	2.1055	2.2240	1.1754	1.5840	1.6759	1.9828
16	12	4.0024	3.1299	3.6036	4.1344	5.3070	7.2292	4.5677
19	12	2.7347	3.0487	2.4895	2.6943	3.6433	2.9247	2.9226
21	22	3.5479	2.7341	2.9917	3.3687	3.0104	3.5779	3.2051
23	22	0.8915	0.9826	0.8603	0.8484	0.2887	0.3449	0.7027
22	24	1.1958	4.1307	1.4776	1.6821	2.8670	2.5352	2.3147

DISCUSSION AND CONCLUSIONS

This paper describes the development of a method and device capable of providing a safe frontal pulse to restrained pediatric and adult human volunteers. During this dynamic sub-injurious frontal crash pulse, the EMG responses of the neck, torso and lower extremity muscles were measured. While adult volunteers have previously been used in impact biomechanics, this effort represents the first to use child subjects in this manner. The envelope for a safe volunteer crash pulse was derived using a novel approach – determining the "pulse" associated with a bumper car to wall impact in an amusement park setting. From this envelope, a custom designed sled was constructed that allowed for the safe conduct of low speed frontal sled tests for the volunteers. Across the six trials for a single subject, the acceleration pulse was very repeatable. Both adult and child volunteers experienced similar accelerations however the mass differences between the subject groups led to slightly greater restraint loading and rebound phases for the adults. From the preliminary analyses of the EMG data, age-based differences were observed in the unnormalized and normalized EMG data between the child and adult subjects.

Previous rear impact studies of adult human volunteers exposed to repeated acceleration of similar magnitude demonstrated a habituation response of the neck muscles, thereby leading to muscle relaxation with subsequent exposure (Blouin et al., 2003). In this study the pediatric and adult volunteers were subjected to a series of six frontal impacts of equal magnitude. No significant pattern of attenuation in neck muscle responses were observed in these tests. Also, the degree of muscle relaxation directly influences head excursion. Future work will correlate the magnitude of the dynamic EMG activity to the head and neck kinematics. The time sequence of the head trajectory data could be compared to the timing of the neck muscle activity.

The different muscle groups displayed varied EMG onset and latency characteristics. The response time of muscles depend on the type of muscle fibers, as well as the dynamics of the activity. Gender and age effects on the dynamic muscle response also needs further clarity. Normalization schemes using anthropometric measures will shed insight into whether this variability is truly age dependent or can be explained by differences in size. Future studies will correlate anthropometry measures and load to the dynamic muscle response.

Several limitations of this approach need to be discussed. First, the normalizations were based on maximum muscle response during an isometric activity. Since, the magnitude of muscle response is rate dependent, an isokinetic-based normalization scheme would be more appropriate for a dynamic event. However, the challenge would be to choose an activity whose rate is similar to the actual event. Second, measurement of muscle activity with surface EMG electrodes may introduce crosstalk from muscles other than the ones being studied. In the past, limited studies have used fine wire electrodes in adult human subjects. Although more accurate measurements could be gathered using this approach, it may be difficult to recruit pediatric volunteers to participate in studies involving such intrusive methods. Third, the volunteers were instructed to relax prior to the start of each test run. Pre-impact bracing would affect the initial level of EMG activity between each run and may also cause considerable variation between subjects. For this reason, the pre-stimulus noise was subtracted from the overall EMG response for all subjects.

In the absence of traditional efforts to define biomechanical response for children using pediatric PMHS, this approach represents a novel means by which to obtain important data that is needed for the design of biofidelic ATDs. By subjecting living child volunteers to sub-injurious dynamic loading, we gain a quantitative understanding of how real children move compared to adults. The human volunteer work described herein is part of a larger project in collaboration with University of Virginia and Takata Corporation in which adult PMHS will be subjected to crashes similar to those experienced by the volunteers and then those same PMHS will be loaded at crash relevant speeds. The synthesis of the volunteer data with the adult PMHS data using either traditional scaling methods and/or computational models will greatly increase our knowledge of the biomechanics of child occupants, leading to better tools for optimizing protection of these occupants in motor vehicle crashes.

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, Richard Kent, PhD (University of Virginia) for his assistance in design of the LASB, Ewout van der Laan, MSc (Technical University of Eindhoven, The Netherlands) for helping design the sled control system and Gunter P. Seigmund, PhD (MEA Forensics) for his guidance with EMG data collection and processing. 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.

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DISCUSSION

PAPER: **Electromyography Responses of Pediatric and Adult Volunteers in Low Speed Frontal Impacts**

PRESENTER: ***Sriran Balasubramanian, Center for Injury and Prevention, The Children's Hospital of Philadelphia***

QUESTION: *Andrew Kemper, Virginia Tech*
I have a question about your repeated loading. Was it the same condition for each of the six trials for the same person?

ANSWER: That is correct.

Q: Were they all conducted on the same day or did you--? Do you know if there would be a difference if you wait a few days and do it again? I could understand there would be a conditioned response, but I'm wondering—

A: In terms of the crash pulse or the response?

Q: Right. For the same subject, if you're trying to compare multiple--

A: We tested the same subject six times in a span of about one hour so there was a time duration of 10 minutes between each test.

Q: I see. I mean I could definitely see the advantage of looking at that, especially if you want to look at multiple conditions, if there's going to be a conditioned response. But, do you know if it would make a difference—if it would help to span that over a couple days or have them come in, like, once a week with the response? Would you start to see conditioning still or not? Do you know?

A: It would be nice if we could do a longitudinal study like that for one subject. But as part of this study, we are constrained by time and availability of the subject. This testing takes about two and a half hours per subject so we are constrained by that. And as part of this study, we are not doing that, but that would definitely be interesting to know though.

Q: Thank you. Very interesting.

Q: *Guy Nusholtz, Chrysler*
Why did you choose smoothing routine? First question is: What is the smoothing routine you're using? And then, why did you choose that over, say, a filter? And then third thing: Why did you choose to rectify it as opposed to, say, leaving it unrectified and looking at it in either the frequency or the wavelength domain?

A: Unrectified or rectified, your right ottomas will give you the same thing because it's acquired. We chose to rectify so that we can fit it on a graph that's more visible. So why did we chose the ottomas or smoothing over other filtering? We looked into the literature data and we wanted to follow the basis of what is being used in all these dynamic activities and how they analyze the EMGs. One of the referenceable methods that they use is the ottomas-type filter, and 25 millisecond is also one of the referenceable and workable EMG analysis methods. We chose to use that so that we can reference it.

Q: It's historical. If you use, say, a non-causal filter, you won't run into the problem of extenuating backwards in time and so you won't have that type of distortion. So you might get a little bit more power or capability of looking into your signal if you use some of the filters. The other thing is: Why didn't you look at it unrectified and say you did the frequency of the wavelength domain? You're looking at it in the time domain. You could do a 48 transform and look at the unrectified signal in the frequency domain and see if there's a frequency shift between, say, adults and children. Or since it's a finite time, you might even consider a wavelet decomposition and then look at the change in the wavelet power for the purposes of comparison.

A: Sure. This is a preliminary data analysis that we've done. This is, in no way, finalized by any means. There are many other techniques, like you said, that involve more complicated mathematical analysis so we can definitely look into how all this plays in our future analysis.

Q: Okay. Thank you.

Q: *Joel Stitzel, Wake Forest University Center for Injury Biomechanics*
This may be a bad question because I'm not an EMG expert. Is the EMG measurement sensitive at all to how far the transducer is placed from the muscle? Would there be influence of skin thickness or subcutaneous fat? I just noticed that the older folks had lower signals and I was wondering if that could be a product of the skin thickness or subcutaneous fat or other products of just being an adult and having an adult body type.

A: Very valuable question. We chose the bi-polar electrodes. There is also a reason to believe that if you increase the distance between the electrodes. Ours was a 1 cm distance between the bi-polar electrodes. So that is kind of standard to reduce any sort of noise pick-up from other cross-talk from other muscles. This was placed in the belly of the muscle, as recommended by one of the other obligations from NASA. So we placed it in the belly of the muscle so we don't pick up cross-talk or heartbeat. So we don't want EKG in that as well. The second part was: Is there a difference in skin thickness? What one study has shown is that they put fine-wire electrodes and they put surface electrodes, and they compared the two. They found that the surface electrodes produced equally reliable measures as the fine-wire electrodes, which are [put] directly into the muscle for the neck muscles. You don't have a way to quantify subcutaneous muscle fat so here is impedance because of the fat in the voltage that's measured. We don't have to wait to measure it, but it seems that it's reliable from the literature. Surface electrodes are as reliable as fine-wire for the neck muscles.

Q: Great. Thanks.

Q: *Gunter Siegmund, MEA Forensic Engineers & Scientists*
First of all, a comment in answer to that question about the muscle response attenuating: We've looked at muscle responses a week apart, 20 to 30 seconds apart, which you saw earlier today, 15 minutes apart, and an hour apart. At a week, there's no habituation. Clearly at an hour, there's none. At 15 minutes, it's sort of subject-dependent; and at 20 to 30 seconds, it's very much present. So, just to answer that question.

A: Okay. Thank you.

Q: And a separate question: During your MVCs, did you measure the force of the moment that the subjects were generating. And, how do you coax a child to generate an MVC? It's very difficult with an adult.

A: I think it's one of your papers or one your group's that does say that verbal coaching helps. I'm not sure if it's your paper or what you told us. So we do verbal coach the adults and child subjects and we do repeat measurements also. So we take two MVC measurements and take the average of the two, and that is the one that we use to normalize. They do attempt at neck flexion and extension on a load cell and we do verbally encourage them to keep pushing harder and harder. So we collect it for 10 seconds and we take the middle six seconds to control for any artifacts, for initialization, and at the end. So we take the middle six seconds of the MVC collection for two trials and average that, and that's the data that we use.

Q: Did you compare the forces or moments they generate with what's published in the literature?

A: Yes. We do have some preliminary data. I can share some slides with you later, but that's not being presented here.

Q: Thank you.

Q: *John Melvin, Tandelta*
I'm just curious. How do you get informed consent from a minor?

A: Initially when we started this project, we approached the Institutional Review Board of the hospital. They suggested that the IRB protocol should be okay because this is an activity that kids and adults participate in on a voluntary basis and it's very safe. So we have a written Informed Consent form that shows that the participant and their parents have to be present at all times and can read. And, we show the sled moving without anybody on it first. And if they are willing to participate, they can ride the sled always belted once first; and then, we ask them if they would like to do it for six more trials. So it is a sequential process where we run it with nobody, run it once with them in it; and if they choose to go on, we continue to instrument them and they ride it six more times. We've had a different problem. We've had kids that don't want to get off the sled! [laughter]

Q: I understand that, but kids can be convinced to do some pretty dumb things as well.

Q: *Kristy Arbogast, Children's Hospital of Philadelphia*
John, just to answer that. We do consent the parent and the parent is present during the event. And any child above seven, we ask for their ascent. So as Sri has described, they see what's going to happen, their parent consents, and then they must give their ascent to participate.

Q: *John Melvin, Tandelta*
I understand, but I'm still sure it's a rather controversial situation. There was a situation in Detroit recently where parental consent was given at a gun show. The five year-old kid had been taught how to shoot a rifle. They gave him an Uzi and he shot himself in the head because he held the trigger down and it just went around. So, parents don't necessarily have good sense either.

Q: *Erik Takhounts, NHTSA*
I have a quick question for you. You compare the muscle latency, compare sled acceleration with muscle response basically. I was wondering if you are going to compare, actually, acceleration of the head. That's one. That will probably reduce the latency. And #2, Why do you compare to acceleration and why not displacement? What is that inside of the head or the brain that actually make it responds to acceleration? Velocity? Displacement?

A: We are. You're absolutely right. That is from the kinematic data that we have not analyzed for this study so I didn't have—So we just defined EMG latency for now as the time difference between the EMG onset and the event onset, but there is something happening. I mean when the sled starts to move and the head starts to move, so we will compare it with the head displacement.

Q: Okay.

A: You're right. That will reduce that.

Q: Okay.