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Electromyographic and Kinematic Exploration of Whiplash-Type Neck Perturbations in Left Lateral Collisions

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Study Design. Seven healthy volunteers were subjected to left lateral impacts of 4.8, 7.9, 11, and 13.7 m/s^2 acceleration at two levels of expectation: expected and unexpected.

Objectives. The purpose of this study was to determine the response of the cervical muscles to increasing low-velocity left lateral impacts, and to compare the quantitative effects of expected and unexpected impact.

Summary of Background Data. The literature contains little information on the etiology of whiplash injuries. Animal and cadaver studies have yielded some insight into the phenomenon. However, *in vivo* studies of the cervical muscular response and head-neck kinematics to lateral impacts are rare.

Methods. Bilateral electromyograms of the sternocleidomastoids, trapezii, and splenii capitis were recorded bilaterally. Triaxial accelerometers recorded the acceleration of the chair, torso at the shoulder level, and head of the participant.

Results. At an acceleration of 13.7 m/s^2, the sternocleidomastoids and trapezii generated approximately ≤50% of their maximal voluntary contraction electromyogram in both the expected and unexpected impact conditions. Study participants exhibited lower levels of their maximal voluntary contraction electromyogram when the impact was expected. The splenius capitis behaved similar to these muscles in the expected condition, but when the impact was unexpected, the splenius capitis muscle contralateral to the impact (*i.e.*, right splenius in a left lateral impact) generated 94% of its maximal voluntary contraction electromyogram. Electromyographic variables were significantly affected by the levels of acceleration and expectation ($P < 0.001$). The onset time and peak electromyogram time for the sternocleidomastoid, splenii capitis, and trapezii on the side contralateral to the side of impact progressively decreased with increasing levels of acceleration. The onset time and peak electromyogram time for the sternocleidomastoid, splenii, and trapezii on the same side of impact progressively increased with increasing levels of acceleration. The kinetic variables and the electromyographic variables regressed significantly on the acceleration ($P < 0.01$). In response to left lateral impacts, muscle responses were greater with higher levels of acceleration, greater with unexpected impact conditions, and greatest for the splenius capitis muscle contralateral to the side of impact.

Conclusions. Because the muscular component of the head-neck complex plays a central role in the abatement of higher acceleration levels, it may be a primary site of injury in the whiplash phenomenon in lateral collisions. Expecting or being aware of imminent impact may play a role in reducing muscle responses in low-velocity impacts. [Key words: cervical muscles, electromyography, acceleration, motor vehicle collisions, lateral impacts, whiplash injury] **Spine 2004;29:650–659**

The Quebec Task Force on Whiplash-Associated Disorders impressed the need for more research studies on the whiplash syndrome, both on aspects of the acute injury and the development of chronic pain.[1] A major problem in conducting research in the area of whiplash, however, is that for "soft tissue" injuries (classified as Grade 1 or 2 whiplash-associated disorders [WAD] by the Quebec Task Force grading system) there is a lack of objective phenomena to measure. Grades 1 and 2 WAD are defined entirely according to subjective symptoms and physical examination findings that depend on subjective input such as tenderness and a limitation of range of movement of the spine according to, in part, pain perception. Castro *et al* have shown, moreover, that the acute whiplash syndrome, as measured by the presence of symptoms, can be reproduced in the presence of a nocebo collision (*i.e.*, fake collisions).[2] Thus, subjective symptoms and signs are less reliable as research measures.

The increasing use of surface electromyography (EMG), however, is providing promising research approaches concerning WAD. Surface EMG has been used, for example, to demonstrate the changes in sternocleidomastoid muscle (SCM) activity as a means to objectively visualize the neutral zone of spine movement in WAD grades 1 to 3.[3] The aim of such research is, in part, to develop methods to follow a given patient (who previously lacked the capacity to show the complete pattern of muscle activity, including motion into the elastic zone) and determine if there is an objective improvement, even if the patient yet has pain. This may be a useful method to reassure patients who are engaging in exercise therapy

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(which is often painful initially) that the muscle activity is improving.

Surface EMG studies in patients with whiplash have also helped in the investigation of some long-held (but potentially erroneous) beliefs. Nederhand *et al*, for example, have used EMG studies to show different muscular responses in the upper trapezius (TRP) muscles between a group of patients with WAD and a matched set of healthy control subjects.[4] In particular, the WAD group reacted with an increase in muscle activity in response to the performance of a dynamic physical exercise, indicating a decrease in the ability to relax the cervical muscles after being subject to a physical load. In a second study, this muscle "hyperreactivity" also appeared to be present in a group of patients with nonspecific neck pain without a traumatic onset.[5] Nederhand *et al* concluded that this muscle activation pattern is not related to the specific mechanism associated with a trauma.[5] This same group more recently analyzed the muscle activation patterns of the upper TRP muscles in a cohort of 92 subjects with acute whiplash patients in order to prospectively evaluate differences in muscular activation patterns between subjects who have recovered and those subjects who have not recovered following an acute injury and developed chronic neck pain.[6] Surface EMG measurements revealed no elevated muscle reactivity, either in the acute stage or during the follow-up period. They found an inverse association between the level of neck pain disability and EMG level. In subjects with future disability, the acute stage was characterized by a reorganization of the muscular activation of neck and shoulder muscles, possibly aimed at minimizing the use of painful muscles. This change of motor control is in accordance with both the (neurophysiological) pain adaptation model and (cognitive behavioral) fear avoidance model and contrary to the long-held belief that pain and increased muscle activity reinforce one another, resulting in a "vicious circle" of pain—muscle spasm—pain.

Finally, surface EMG is providing valuable data in so-called whiplash collision experiments.[7–13] In the setting of controlled collision events, and in combination with other objective measurements, surface EMG studies have helped to model the mechanism of acute muscle injury in low-velocity collisions. Ideally, one would devise experiments in which volunteers are subjected to collisions of progressively higher velocities where the injury threshold is reached and surpassed, while EMG data are collected. There have been limited experiments with volunteers where this has been done, but usually the volunteers have been members of the research team or military volunteers.[14] With other volunteer groups, the collision velocities have been necessarily kept at ≤8 km/h for rear-end collisions.[15]

Given the ethical considerations and the difficulties in subjecting volunteers to injury, another approach to the problem that we have been investigating has included the use of regression techniques modeled on incremental very low-velocity range impacts. As it can be shown that the regression models are in good agreement with the available data that have been gathered in previous, small studies of higher-velocity collisions, the use of linear regression extrapolation methods may have a role, providing more understanding of what happens to the neck muscles in various collision types and yet avoiding subjecting volunteers to injury itself.

We have previously reported on the results of this approach with data for rear-end collision experiments with volunteers,[12] and in this study we report the data for left lateral impacts.

Methods

The methods for this study of left lateral impacts are the same as that used previously for rear-end impact study.[12] The details have thus been described elsewhere and will be given in brief here.

Sample. Seven healthy normal subjects with no history of whiplash injury and no cervical spine pain during the preceding 12 months volunteered for the study. The study was approved by the university health research ethics board.

Tasks. As described elsewhere,[12] seated and stabilized subjects were exposed to left lateral sled accelerations of 4.9, 8.8, 10.8, and 13.7 m/s^2 in a random order by the pneumatic piston. The accelerations were delivered under two conditions: volunteers were either expecting (expected group) or not expecting (unexpected group) the impact. As the results for left lateral impact are likely to be of more consequence to the driver of vehicles, it was studied first.

Experimental Setup. The acceleration device consisted of an acceleration platform and a sled. The full details of the device are given by Kumar *et al*.[12] An abbreviated description follows. The acceleration platform had parallel tracks 2 × 200-cm long, mounted lengthwise 60 cm apart. These tracks permitted smooth gliding of the sled on the rails, with a low coefficient of friction (0.03). This assembly allowed a maximum linear speed up to 36 km/h. At one end of the platform, a pneumatic cylinder with a piston stroke length of 30 cm was connected to an air supply and mounted rigidly on the acceleration platform. The device was calibrated for the delivery of known forces causing acceleration of 4.9, 8.8, 10.8, and 13.7 m/s^2. The opposite end of the platform was equipped with a high-density rubber stopper in the sled's path to prevent it from sliding off the platform.

The sled consisted of a molded plastic seat with a backrest and four legs mounted to a rectangular sliding board coupled with the tracks for friction-reduced travel on impact. The sled was equipped with a footrest and four buckled straps to stabilize the lower extremities. The seat was fitted with a four-point seat restraint system. The volunteers faced 90° from the direction of travel for all experimental trials, and the trials were conducted so that the impacts could be experienced from the left for each subject. Three high-performance triaxial accelerometers with a full-scale nonlinearity of 0.2% were used in the study. They had a dynamic range of ± 5 g, a sensitivity of 500 mV/g, and a resolution of 5 mg within bandwidth DC 100 Hz.

Table 1. Mean Normalized Peak Electromyogram of Cervical Muscles in Response to Simulated Left Lateral Impacts Impacts

Impact Acceleration (m/s^2)	Sternocleidomastoid (% MVC)		Splenius Capitis (% MVC)		Trapezius (% MVC)	
	Left	Right	Left	Right	Left	Right
Unexpected						
4.9	4 (2)	6 (4)	15 (6)	26 (22)	21 (9)	28 (14)
8.6	19 (26)	25 (22)	21 (10)	52 (38)	30 (15)	45 (29)
10.9	19 (17)	37 (36)	28 (13)	69 (52)	41 (26)	52 (39)
13.7	41 (51)	58 (48)	36 (28)	94 (29)	54 (26)	59 (30)
Expected						
4.9	9 (2)	8 (2)	12 (7)	15 (12)	20 (6)	16 (4)
8.6	8 (3)	13 (9)	13 (7)	15 (12)	21 (4)	18 (3)
10.9	9 (3)	23 (24)	14 (5)	20 (9)	23 (5)	20 (5)
13.7	9 (3)	47 (61)	15 (8)	29 (10)	27 (11)	26 (11)

Values in parentheses represent 1 standard deviation.
MVC = maximal voluntary contraction.

Data Acquisition. The data acquisition system consisted of an analog-to-digital board with a 100-kHz sampling capacity. Each of the nine acceleration channels and six EMG channels as well as the force channel were sampled at 1 kHz in real time. The sampled signals were stored on a computer with a large hard disc for storage and processing. The EMG and acceleration data were collected during the experimental trials. The peak and average EMG and acceleration values obtained from these sets of data were subjected to quantitative and statistical analysis.

Test Protocol. After the experiment was discussed and informed consent obtained, the age, weight, and height of each volunteer were recorded. The volunteers then were seated on the chair and stabilized in neutral spinal posture. Two triaxial accelerometers were fixed to the volunteer: the one immediately inferior to the seventh cervical vertebra at the level of the shoulder and the other immediately superior to the glabella region of the frontal bone of the skull. The accelerometers were affixed to the volunteers with strong self-adhesive tapes. The axes of the three accelerometers were aligned with the path of the chair. The pneumatic cylinder was aligned such that the piston head of the cylinder and the baseboard of the rear of the sled were in contact. The pneumatic piston delivered the appropriate acceleration to the sled. The subjects in the "expected" group were informed about the forthcoming impact magnitude in qualitative terms: very slow, slow, medium, and fast. The subjects in the "unexpected" group were blindfolded and provided a portable stereo with engaging music playing loud enough to block any auditory cues. The data collection was initiated, and after 1 second the pneumatic piston was fired to accelerate the sled.

Data Analysis. In the analysis, the sample of volunteers was collapsed across gender because preliminary analysis showed no statistically significant differences in the peak EMG amplitudes between the men and women. Data analysis was performed in two stages.

In the first stage, the velocity and acceleration of the sled subsequent to the pneumatic piston impact and the rubber stopper impact were calculated. The time of the peak acceleration from the firing of the piston was measured. The data on the peak and average accelerations in all three axes of the sled, shoulder, and head for all four levels of accelerative impacts and for both levels of expectation (expected and unexpected) were measured. For the amplitude analysis, the magnitudes of the full-wave rectified, averaged, and linear envelope-detected EMG signals were subjected to 7-point segment polynomial smoothing repeated once. From such traces, peak EMG, average EMG, and the slope depicting the rise of the EMG traces were obtained. Also, the time relations of the onset and peaking of the EMG in relation to the piston firing were measured and analyzed.

EMG amplitudes were normalized against the subjects' maximal voluntary contraction electromyogram, these voluntary contraction electromyograms having been determined in the same experiment with these subjects before accelerative impacts, providing strength measurement results in Newtons for each muscle.[16,17] The ratio (percentage) of the EMG amplitude *versus* the maximal contraction normalized EMG activity for that subject allowed us to determine the force equivalent generated due to the impact for each muscle. In the second stage, a statistical analysis was performed using the SPSS statistical package (SPSS Inc., Chicago, IL) to calculate descriptive statistics, correlation analysis between EMG and head acceleration, analysis of variance of the EMG slope, time of peak EMG, EMG onset time, peak EMG, average EMG, and the force equivalents.

Results

The seven subjects had a mean age of 25.7 ± 2.7 years, a mean height of 173 ± 5.3 cm, and a mean weight of 71 ± 14.9 kg.

Electromyogram Amplitude in Left Lateral Impacts

The mean peak (normalized) EMG amplitude of the cervical muscles tested in this experiment for the expected and unexpected impacts at each applied acceleration level is presented in Table 1. Figure 1 illustrates the EMG recorded under these conditions. As the level of applied acceleration in a left lateral impact increased, the magnitude of the EMG recorded from the right splenius capitis increased progressively and disproportionately. The EMG was greater during unexpected impacts.

The normalized EMG showed that the percentage of SCM, right splenius capitis, and TRP magnitude in-

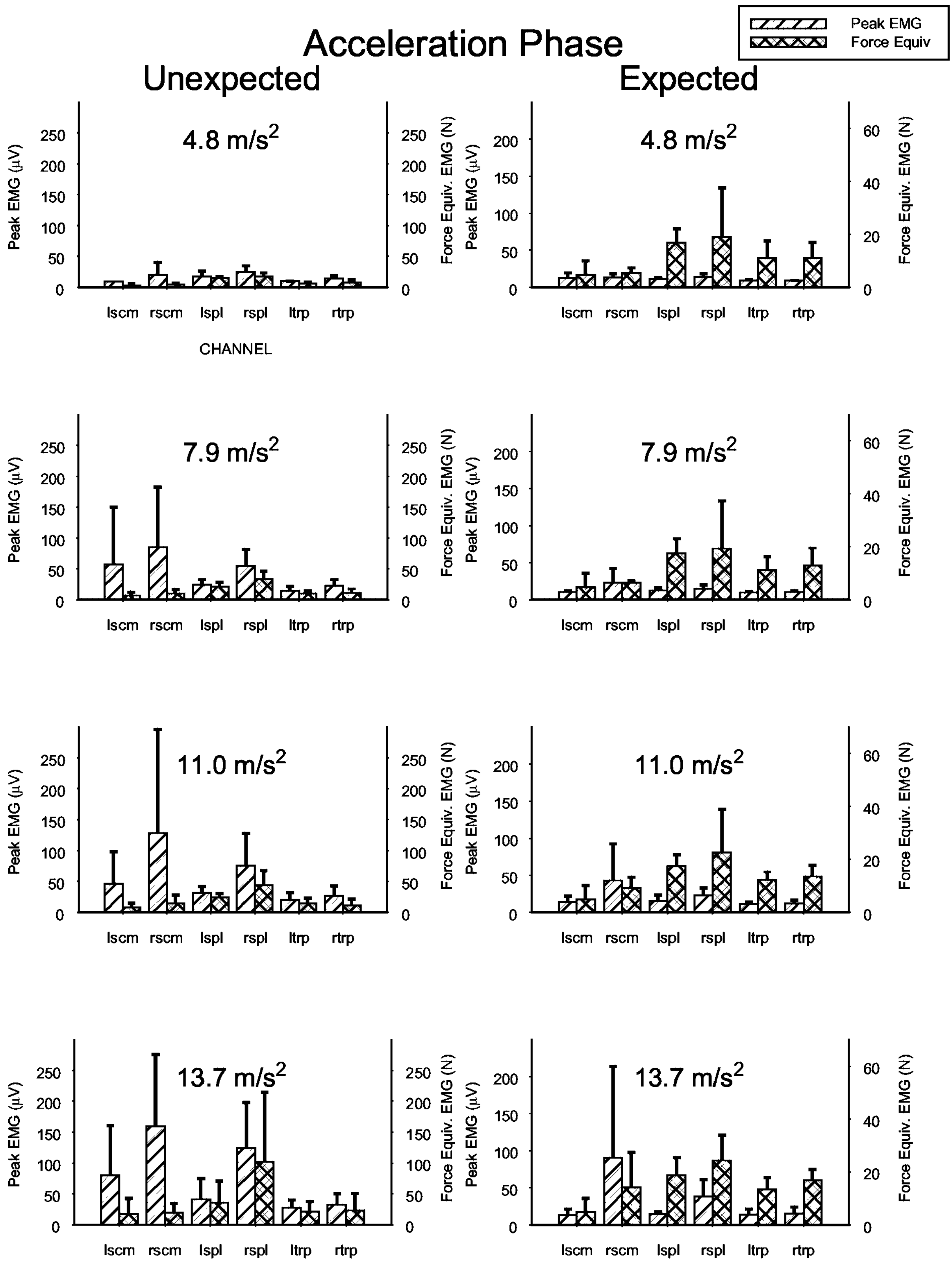

Figure 1. Means of peak electromyographic activity (μV) at two levels of expectation and four levels of applied acceleration. LSCM = left sternocleidomastoid; RSCM = right sternocleidomastoid; LSPL = left splenius capitis; RSPL = right splenius capitis; LTRP = left trapezius; RTRP = right trapezius.

creased steadily with the increasing magnitude of the impact and generally for both the expected and unexpected conditions (Table 1; Figure 2). In a left lateral impact, with unexpected condition, at an acceleration of 13.7 m/s^2, the left SCM exerted 41% and the right SCM 58% of the mean normalized maximal voluntary con-

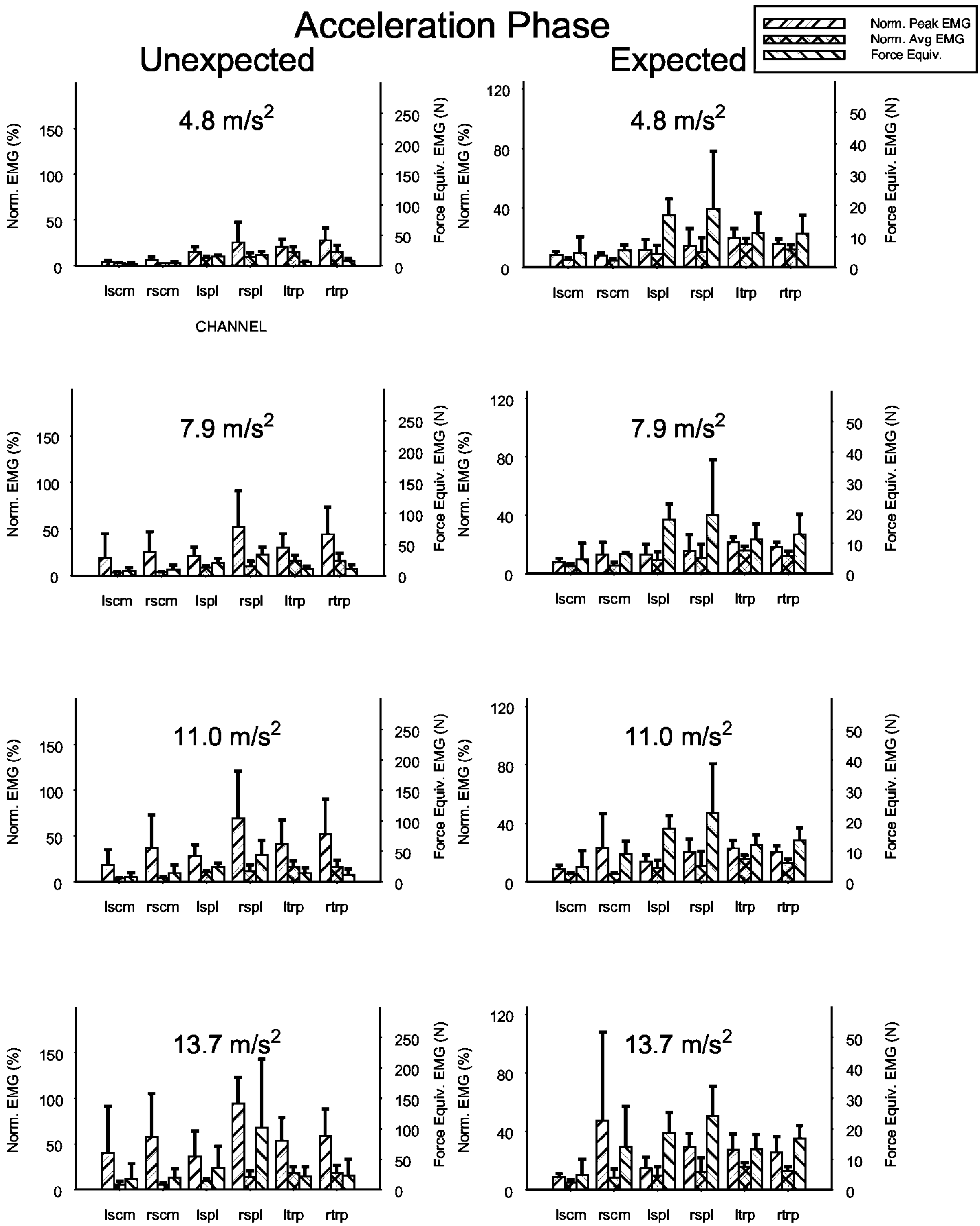

Figure 2. Normalized average and peak electromyogram (EMG) (percentage of isometric maximal voluntary contraction), force equivalent of EMG (N), and levels of expectation and applied acceleration. LSCM = left sternocleidomastoid; RSCM = right sternocleidomastoid; LSPL = left splenius capitis; RSPL = right splenius capitis; LTRP = left trapezius; RTRP = right trapezius.

traction, whereas the right splenius exerted 94% and the left splenius 36% of the mean normalized maximal voluntary contraction and the TRP never exceeded 60% of this variable. The percentage of exertion was higher in the SPL than in the TRP or SCM for both expectation conditions at the highest level of applied acceleration. In terms of force equivalents, the SPL contralateral to the side of impact were required to resist the lateral impact at a level near their maximal voluntary contraction capability.

Table 2. Mean Time to Onset (msec) of Acceleration and of Muscle EMG From the Firing of the Solenoid of the Pneumatic Piston

Impact Acceleration (m/s^2)	Acceleration			Sternocleidomastoid		Splenius Capitis		Trapezius	
	Sled	Shoulder	Head	Left	Right	Left	Right	Left	Right
Unexpected									
4.9	128 (35.4)	452 (168.5)	450 (99.8)	171 (115)	303 (249)	326 (166)	288 (216)	260 (312)	716 (948)
8.6	106 (14.7)	401 (105.5)	419 (27.6)	214 (142)	323 (359)	201 (87)	156 (29)	336 (373)	180 (61)
10.9	84.7 (11.7)	278 (4.9)	407 (41.2)	154 (39)	150 (34)	342 (268)	130 (16)	832 (1375)	152 (20)
13.7	42.5 (8.4)	283 (24.1)	415 (44.8)	635 (1001)	140 (35)	643 (936)	140 (48)	769 (892)	146 (51)
Expected									
4.9	113 (13.8)	305 (143.7)	453 (79.9)	302 (186)	212 (173)	871 (935)	203 (146)	1364 (1082)	1318 (978)
8.6	85.4 (6.5)	303 (115.9)	419 (27.6)	123 (153)	209 (98)	979 (956)	104 (39)	1007 (836)	712 (1164)
10.9	78.2 (9.6)	318 (135.1)	407 (41.2)	166 (99)	520 (707)	1124 (1386)	102 (30)	928 (1453)	652 (1170)
13.7	95.2 (34.3)	352 (75.4)	415 (44.8)	1172 (1097)	165 (59)	1688 (963)	153 (10)	677 (1080)	664 (1002)

Times for the sled, shoulder, and head represent the time at which acceleration in z-axis (direction of travel) began. Times for the cervical muscles represent the onset time for EMG activity. Values in parentheses represent 1 standard deviation.

Electromyogram Slope

During the lateral impacts with increasing magnitude of applied acceleration, the slope increase in the EMG response for all muscles was significantly less in the expected condition. In comparison, the incline of the right SPL (that contralateral to the left lateral impact) was higher than the slopes for the SCM and TRP slopes for all levels of applied acceleration and expectation.

Timing

The time of the sled, shoulder, and head acceleration onset in the z-axis (axis along lateral impact direction) and the EMG signals of the six muscles examined are presented in Table 2. The time of onset was measured from the firing of the pneumatic piston. The time of the sled, torso, and head acceleration onset decreased with increased applied acceleration. Similarly, the time of the EMG onset decreased with increased applied acceleration. The mean times at which peak EMG occurred for all the experimental conditions are presented in Table 3.

Head Acceleration

The kinematic response of the head to the four levels of applied acceleration in the expected and unexpected conditions is shown in Figure 3. As anticipated, an increase in applied acceleration resulted in an increase in excursion of the head and accompanying accelerations. The head acceleration response was greater in the unexpected than in the expected condition. The association between the force equivalent EMG response of each muscle and the head acceleration is shown in Table 4.

Statistical Analyses

The applied acceleration, the muscles examined, and the volunteer's expectation had significant main effects on the peak EMG activity ($P < 0.001$). To justify the combination of the male and female EMG responses to applied acceleration, gender was entered into the analysis, and the results were nonsignificant, indicating that gender did not confound the results. Clearly, with greater magnitude of impact, there was a greater teleologic response and each muscle responded differently. Awareness of the impending impact significantly reduced the muscle activity ($P = 0.014$).

The levels of acceleration, the muscles examined, and the level of expectation had significant main effects on both the slope of EMG and onset time ($P < 0.02$). The onset time of the right splenius muscle in a left lateral impact was the only one affected by acceleration. Ini-

Table 3. Mean Time (msec) at Which Peak Electromyogram Occurred After the Firing of the Solenoid of the Pneumatic Piston

Acceleration (m/s^2)	Sternocleidomastoid		Splenius Capitis		Trapezius	
	Left	Right	Left	Right	Left	Right
Unexpected						
4.9	1233 (528)	491 (307)	758 (398)	499 (404)	739 (555)	456 (263)
8.6	409 (347)	931 (1158)	690 (938)	242 (32)	331 (139)	323 (57)
10.9	180 (89)	246 (98)	704 (360)	238 (36)	576 (464)	286 (69)
13.7	279 (164)	213 (46)	581 (535)	205 (39)	744 (451)	264 (71)
Expected						
4.9	1876 (151)	368 (295)	860 (709)	202 (78)	728 (101)	845 (938)
8.6	450 (453)	786 (1018)	1122 (622)	209 (48)	1465 (670)	800 (627)
10.9	1175 (867)	615 (784)	1112 (751)	227 (25)	684 (131)	1224 (1135)
13.7	611 (381)	424 (388)	1172 (498)	229 (32)	431 (354)	484 (323)

Values in parentheses represent 1 standard deviation.

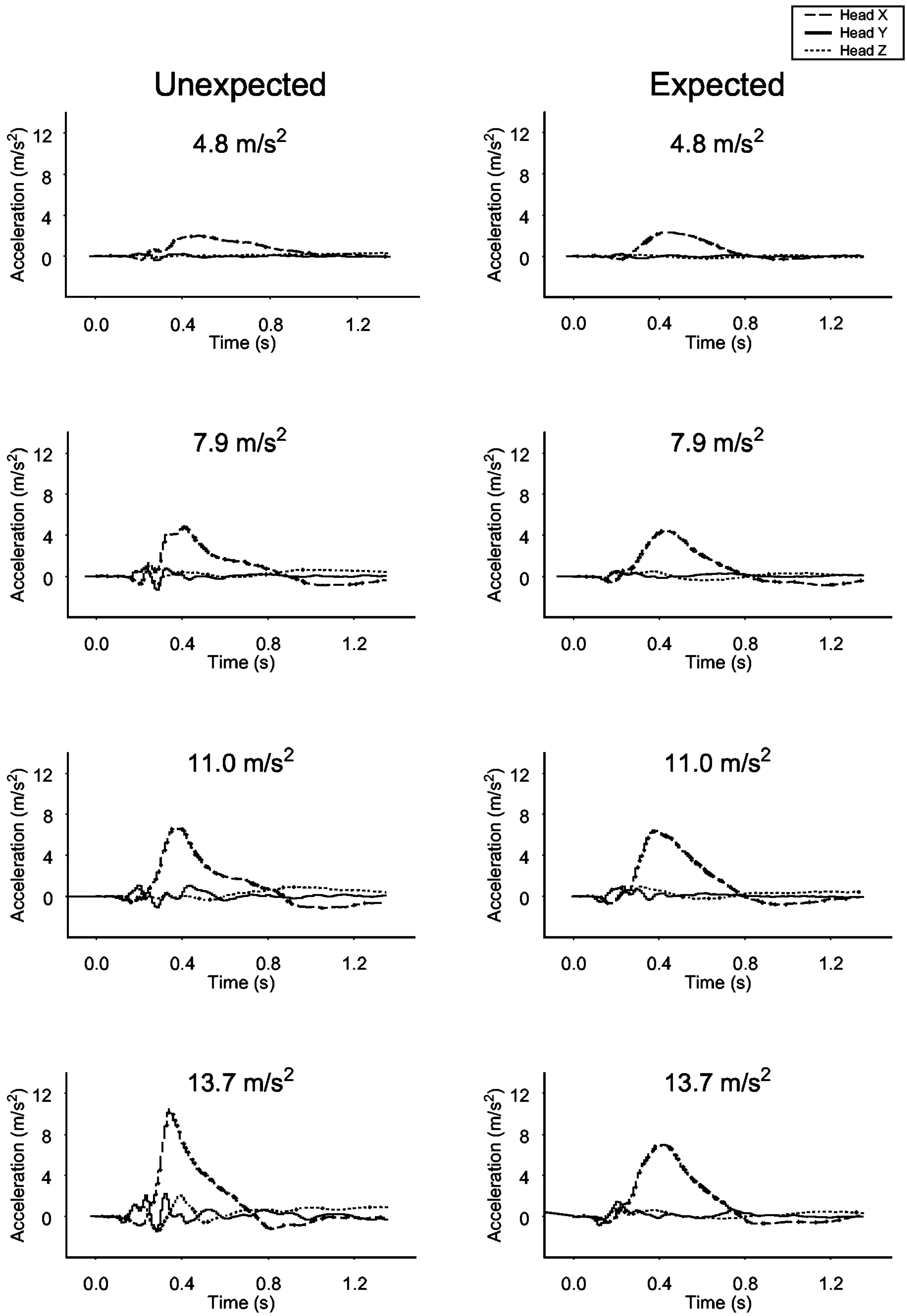

Figure 3. Head acceleration in the x-, y-, and z-axes of one study participant in response to the level of expectation and applied acceleration. The z-axis is parallel, the x-axis orthogonal, and the y-axis vertical to the direction of travel. Head X = head acceleration in the x-axis; Head Y = head acceleration in the y-axis; Head Z = head acceleration in the z-axis.

Table 4. Mean Force Equivalents and Mean Head Accelerations at Time of Maximal EMG in Direction of Travel for Left Lateral Impact

		Force Equivalents for Muscle (N)					
		Sternocleidomastoid		Splenius Capitis		Trapezius	
Chair Acceleration (m/s^2)	Head Acceleration (m/s^2)	Left	Right	Left	Right	Left	Right
Unexpected							
4.9	3.0 (0.55)	3 (3)	4 (2)	16 (2)	18 (6)	7 (2)	8 (5)
8.6	6.8 (1.8)	7 (6)	10 (7)	21 (8)	34 (12)	10 (5)	11 (7)
10.9	8.1 (0.91)	8 (7)	15 (14)	25 (6)	44 (24)	14 (9)	11 (10)
13.7	13.9 (6.44)	18 (25)	20 (15)	36 (35)	101 (114)	21 (17)	23 (28)
Expected							
4.9	2.6 (0.38)	5 (5)	6 (2)	17 (5)	19 (19)	11 (6)	11 (6)
8.6	4.7 (0.62)	5 (5)	7 (1)	18 (5)	19 (18)	11 (5)	13 (7)
10.9	7.2 (1.8)	5 (5)	9 (4)	17 (4)	23 (16)	12 (3)	14 (4)
13.7	6.4 (1.2)	5 (5)	14 (13)	19 (7)	24 (10)	13 (5)	17 (4)

Values in parentheses represent 1 standard deviation.

tially, regression analyses were performed only up to 13.7 m/s^2 using linear, quadratic, cubic, power, and exponential functions. The kinematic variables of head displacement, velocity, and acceleration in response to applied acceleration were calculated (Figures 4, 5). Using the obtained regression equations, the responses of the left and right muscle groups were extrapolated to more than twice the applied acceleration value.

■ Discussion

Electromyographic investigation provides useful information for the investigation of whiplash mechanisms and patients. In the current study, EMG information has been used to examine the muscle response to very low-velocity impact and to extrapolate that response to higher-velocity impacts. EMG studies also allow one to examine muscle group responses and patterns, rather than simply describe head or other body region accelerations. The experimental design we have used to study neck perturbations to very low-velocity change is not intended to mimic vehicle occupant position but rather to allow for the initial exploration of the role of EMG in assessing neck perturbations. Because it is not yet possible to objectively identify the acute whiplash injury thought to underlie Grades 1 and 2 WAD, current injury models are based on evaluation of volunteers in collisions. With so many parameters available for modulation in attempting to approximate road collisions, the task of developing a model for the acute whiplash injury is daunting. One starting place, however, is the use of objective measurements such as EMG in a laboratory setting where other confounding variables have been accounted for or eliminated. In time, more variables can be introduced and studied with this approach.

There is no direct way to measure forces exerted by muscles due to neck perturbation and subsequent muscle activity, but examining EMG activity generated allows one to compare this to EMG activity in voluntary contractions. This in turn allows one to relate the muscle responses to normal muscle forces in various physiologic ranges of activity. It is not surprising that subjects in our experiment experienced no adverse symptoms in relation to the experimental design. The velocities used were by

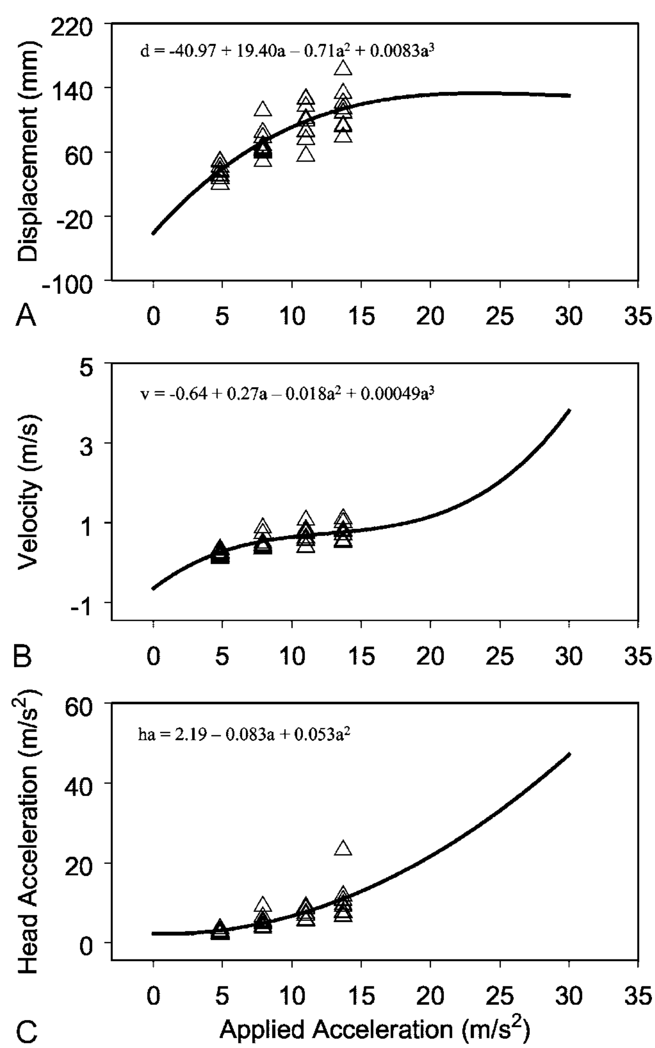

Figure 4. Extrapolated regression plots of the effect that applied acceleration has on the head motion variables of displacement (mm) (**A**), velocity (m/s) (**B**), and acceleration (m/s^2) (**C**) obtained.

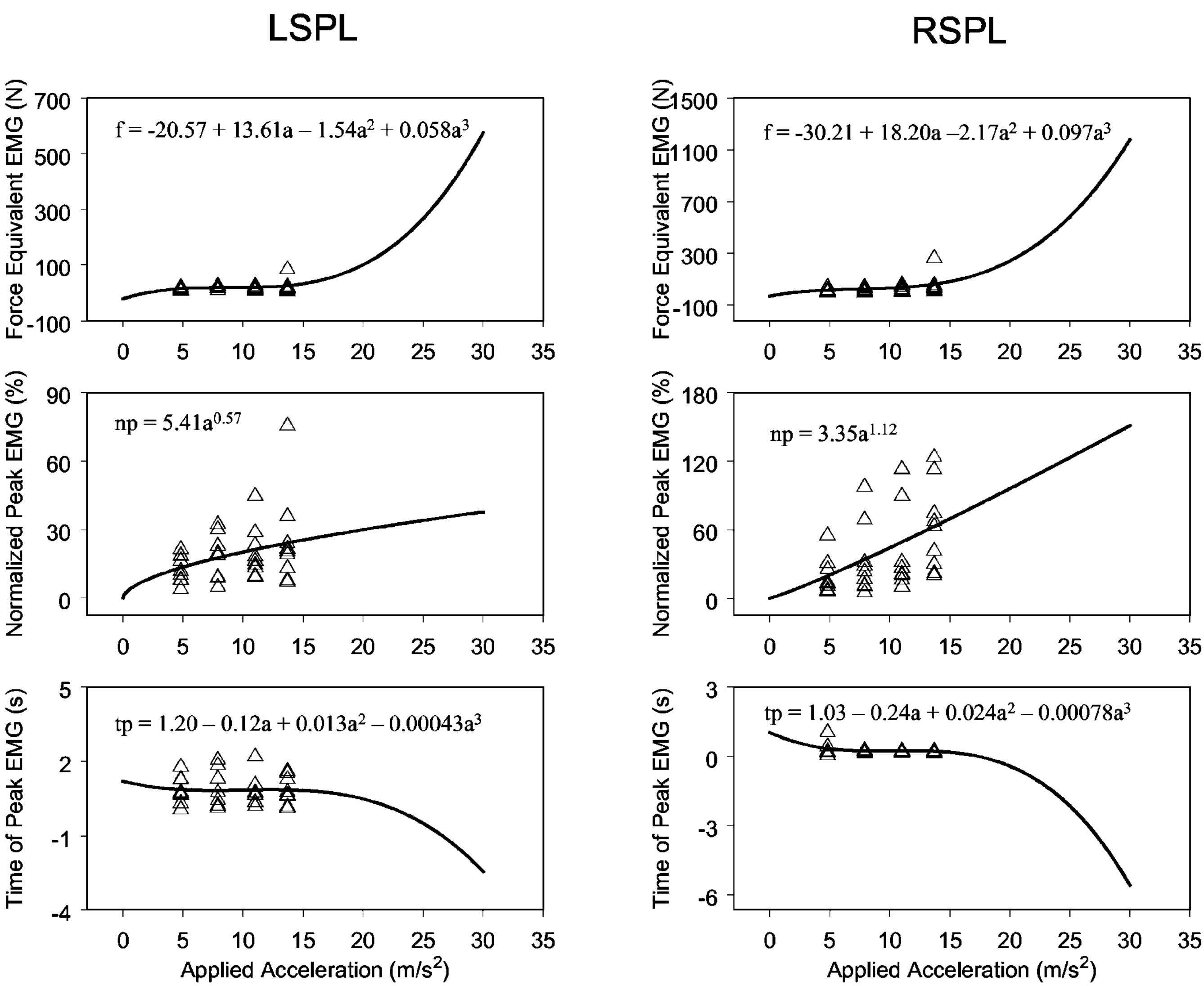

Figure 5. Extrapolated regression plots of the effect that applied acceleration has on the left and right splenius capitis muscles for the variables of time of peak electromyogram (EMG)(s), normalized EMG (percentage of isometric maximal voluntary contraction), and force equivalent of EMG (N).

design meant to avoid potential injury. Moreover, we have previously measured the force exertions of neck muscles in healthy volunteers, and we see that voluntary cervical force exertions exceed the force equivalent exposures we observed in these low-velocity experiments.

Nevertheless, using incremental very low-velocity experimental design, we are able to extrapolate through linear regression to predict the head accelerations and forces likely to be experienced by neck perturbations at higher velocities within the low-velocity range. Our extrapolations closely match those from small volunteer studies where higher velocities were used with symptoms produced.[14] Siegmund *et al* have also found very low-velocity collision setups to be useful in studying specific variables of collisions.[7] They used an experimental design, for example, in which head acceleration was expected to be very low, on the order of 1.5 g. Studying 66 seated subjects undergoing a single forward horizontal perturbation and using surface electromyography to measure the SCM and cervical paraspinal muscle activity, they found that while the kinematic (timing) responses of aware (alerted and unalerted) subjects were not significantly different, cervical paraspinal amplitudes were 260% larger and angular head accelerations in flexion were 180% larger in surprised subjects than in alerted subjects. While more studies and extrapolation are needed to assess this phenomenon in the higher-velocity and accelerations seen in road collisions, these data and our own suggest that aware occupants may have a different whiplash injury potential than unprepared occupants in real collisions.

Studies suggest a central role may be played by the cervical muscles in injury causation during low-velocity collisions.[8,9] Because the muscles are the first in the line of defense for the cervical region, they are likely to be the first in casualty as well. In our study, the data reveal that the splenius capitis muscle contralateral to the side of impact is at greatest risk for injury in low-velocity collisions. In rear-end impacts, we have shown that the SCMs are at greatest risk for injury.[13] Thus, direction of impact or head position relative to direction of impact (*i.e.*, a driver looking to the left during a rear-end impact) may determine which muscles are injured. The current authors propose that whiplash injuries are complex and progressive. Muscles, ligaments, facet joints, and the brain may be injured in sequence with increasing magnitude of impact. In the series of lateral impacts described in this report, although the EMG magnitudes for low acceleration levels were small, they rose rapidly as a power function in a nonlinear fashion. At lower acceleration levels, the unexpected conditions engendered a higher level of EMG activity than the expected conditions. This changed even further at the highest acceleration tested.

As we have discussed in detail elsewhere with a review of the relevant EMG data,[12,13] studies in neck perturbations suggest that the cervical muscle response is triggered by peripheral input of muscle stretch.

Because of their origin and insertion, the contralateral muscles undergo lengthening in a lateral impact. Stretching is likely to be a very effective mechanism for triggering stretch receptors. This does not rule out any central input. Driving a motor vehicle is a learned behavior that involves significant training and conditioning of the somatosensory mechanisms in particular. Therefore, it is likely to have a role in strengthening or modifying the peripheral response. This is clearly shown by the data reported in the current experiment. The EMG onsets in unexpected conditions were later than those in the expected conditions and magnitudes were higher ($P < 0.002$).

The regression analysis showed a function relation between the motion variables of the head and the applied and projected acceleration. The projected values are hypothetical and likely to be affected by the ligaments and joint geometry in a manner different from that recorded in the experiment. Nonetheless, the experiment provided a sense of the head's behavior. With additional experiments uncovering more information regarding the modulus of elasticity of various ligaments and capsules, it may be possible to estimate the threshold level or range of acceleration at which injuries are likely to occur. A similar regression analysis of EMG amplitude (raw, normalized, and force equivalents) assists in discerning the level of acceleration most likely to precipitate some injury. It must be emphasized that no experimental results of the current study help in determining this. However, further research may enable the achievement of this goal. The applied acceleration was a sudden push laterally rather than a lateral impact to the volunteer's vehicle and seat similar to what would be expected from a lateral collision. Such an impact could be attained through modification of the experimental design, by separating the piston from the sled, but when we have done this (data not shown), we found that the ensuing acceleration curves were not significantly different from the regimen used in this experiment. Otherwise, future experiments will examine the effect of other directions of impact (*e.g.*, right lateral impact).

■ Key Points

- With lateral impact, the splenius capitis muscle on the side contralateral to the impact is activated more than the trapezii or sternocleidomastoid muscles.
- The splenius capitis contralateral to the impact reaches 94% of its maximal voluntary contraction with an acceleration of 13.7 m/s^2.
- The trapezii, sternocleidomastoids, and ipsilateral splenius capitis did not generally rise above 50% of their maximal voluntary contraction.
- It is surmised that during left lateral impact such a mechanism is likely to injure the contralateral splenius capitis before other injuries occur.
- Awareness of impending impact significantly reduced the muscle response, and lack of awareness had the opposite effect.

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