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Cervical muscle response to whiplash-type right anterolateral impacts

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Abstract Frontal impacts are a common cause of whiplash injury. Yet, volunteer studies of the cervical muscular response and head–neck kinematics to frontal impacts are uncommon, and specifically, the effect of an offset (anterolateral) frontal impact on the resultant muscle responses is unknown. The purpose of this study was to determine the response of the cervical muscles to increasing low-velocity frontal impacts offset by 45° to the right, and to compare the quantitative effects of expected and unexpected impact. Ten healthy volunteers were subjected to frontal impacts, offset by 45° to the subject's right, of 5.1-, 8.7-, 12-, and 13.7-m/s² peak acceleration at two levels of expectation: expected and unexpected. Bilateral electromyograms of the sternocleidomastoids, trapezii, and splenii capitis were recorded. Triaxial accelerometers recorded the acceleration of the chair, torso at the shoulder level, and head of the participant. At a peak acceleration of 13.7 m/s², with an unexpected impact, the contralateral trapezius (i.e., left trapezius in a right anterolateral impact) generated 83% of its maximal voluntary contraction electromyogram, whereas all other muscles generated 50% or less of this variable. Although it generated less EMG, the splenius capitis muscle also tended to show an asymmetric EMG response, with the left (contralateral) splenius capitis generating a higher percentage (46%) of its max-

imal voluntary contraction electromyogram than the ipsilateral (right) splenius capitis. In comparison, the sternocleidomastoid muscles behaved symmetrically and generated 25% or less of this variable under all impact conditions. Similarly, the times to onset and times to peak electromyogram for the contralateral (left) splenius capitis and (left) trapezius progressively decreased with increasing levels of acceleration ($p<0.01$). Subjects exhibited lower levels of their maximal voluntary contraction electromyogram when the impact was expected ($p<0.01$). The kinetic variables and the electromyographic variables regressed significantly on the acceleration ($p<0.01$). In response to right anterolateral impacts, muscle responses were greater with higher levels of acceleration, and more specifically, when a frontal impact is offset to the subject's right, it results in not only increased EMG generation in the contralateral trapezius, but the splenius capitis contralateral to the direction of impact also bears part of the force of the neck perturbation. Expecting or being aware of imminent impact plays a role in reducing muscle responses in low-velocity anterolateral impacts.

Keywords Cervical muscles · Electromyography · Motor vehicle collisions · Right anterolateral impacts · Whiplash injury

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Introduction

Although it has been traditionally reported that rear impacts account for most cases of whiplash injury [15], large epidemiological study has suggested that rear, lateral, and frontal collisions account for whiplash injury in approximately equal proportions [2]. The 1995 Quebec Task Force on Whiplash-Associated Disorders [15] indicated that an important research objective is understanding the mechanism of whiplash, and direction of impact may be an important variable. There have been a number of studies conducted with cadavers, animals, and human volunteers to attempt to elucidate the injury mechanisms, mainly for rear impacts, but these have suffered from a number of methodological issues, most notably the difficulties extending results to living humans or that the volunteer population has been nonrepresentative [4]. Given ethical concerns with subjecting volunteers to injurious neck perturbations, most volunteer experiments have, for example, been conducted with military personnel and members of the research team. A few other experiments have been done with other volunteer groups but were necessarily limited to low- or very low-velocity collisions [1, 3, 4, 7, 9, 10, 14].

In order to overcome this ethical dilemma and still conduct investigations to elucidate the kinematics and electromyographic response to neck perturbation in human volunteers, we have shown that surface electromyography (EMG) can be combined with the use of regression techniques modelled on very low-velocity collisions [5, 7, 9]. Using this approach, we have conducted rear and frontal-impact studies and the regression models are in good agreement with the available data that has been gathered in previous, small studies of higher-velocity collisions.

Furthermore, we have been able to show that direction of impact is highly relevant to the muscle group which shows the greatest EMG response. In straight-on rear impacts, for example, the greatest muscle response arises from the sternocleidomastoids rather than the trapezii or splenius capitis muscles [5, 7]. On the other hand, in straight-on frontal impacts, the trapezius muscles show the greatest muscle response and are most likely to be the first injured in frontal collisions [9].

By subjecting volunteers to very low-velocity impacts, one can then subject the resulting data to mathematical techniques such as regression. This allows one to use the curve that describes the data and extend this curve out to acceleration values that were not directly tested. In doing so, one is of course assuming that the mathematical result is indeed a good predictor of what happens further along the curve. It may or may not be, and is best viewed as an approximation. Without subjecting volunteers to high accelerations, this is the only currently available, ethical approach to volunteer studies. In the current study, we extended the use of regression extrapolation methods to examine the effect of offset frontal impacts on the cervical muscle response. One of the clinical authors (R.F.) has encountered reports from whiplash patients that their frontal

collision took place with the offending driving striking not in a straight-on approach, but rather at an angle, striking the right front corner of the vehicle bumper usually when the collision occurred in an intersection. Patients have reported anecdotally their belief that this increased their injury risk and that it caused a tendency for mainly unilateral neck pain.

We thus undertook a study to assess the cervical muscle response for frontal impacts where the direction of impact is offset 45° to the subject's right in order to determine if this scenario alters the muscle response relative to straight-on frontal impacts.

Materials and methods

The methods for this study of offset frontal impacts are the same as that used previously for our previously straight-on impact studies [5, 7, 9]. The details have thus been described elsewhere and are given briefly here.

Sample

Ten healthy normal subjects (3 males, 7 females, all right-hand dominant) 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 [5, 7, 9], seated and stabilized subjects were exposed to frontal impacts with their seating offset 45° from the direction of peak sled accelerations of 5.1, 8.7, 12, and 13.7 m/s^2 in a random order by the pneumatic piston. No one has documented at what range of angles motor vehicle frontal collisions occur. We chose an angle of 45° because we had previously conducted frontal collisions (with results known), rear collisions (with results known), and lateral collision experiments pending. We elected to choose a point mid-way between the lateral and frontal direction as the most logical point to test for an interaction between the two vectors of impact direction. By studying cardinal points of impact in a 360° circle, one can extrapolate to other angles in between. The accelerations were delivered under two conditions: volunteers were either expecting (expected group) or not expecting (unexpected group) the impact.

Experimental set-up

The acceleration device consisted of an acceleration platform and a sled. The full details of the device are given by Kumar et al. [5, 7, 9]. An abbreviated description follows. The acceleration platform had parallel tracks $2 \times 200\text{-cm}$ 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 5.1, 8.7, 12, 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 and the volunteers faced 45° from the direction of travel for all experimental trials. Three high-performance triaxial accelerometers (two on subjects and one on the sled with aligned axis as described in "Test protocol") 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.

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 for 5 s. An RMS filter with 25 ms time constant was applied to raw EMG data. The RMS sampled signals were subjected to a 9-point linear moving average filter. 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 was 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 third accelerometer was mounted rigidly on the sled. The axes of the three accelerometers were aligned with the path of the chair. Fixed inter-electrode distance (10 mm) active knife-edge electrodes were placed on the belly of the sternocleidomastoids, upper trapezius at C4 level, and splenius capitis in the triangle between sternocleidomastoids and trapezii bilaterally. The subjects were grounded at acromioclavicular joint bony prominence. All electrodes were applied after suitable skin preparation. Subsequently resisted primary activity of each muscle was recorded to detect any cross talk. Once clean signals were identified the experiment proceeded.

The pneumatic cylinder was aligned such that the piston head of the cylinder and the baseboard of the front of the sled were in contact at an angle of 45°. 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 s 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 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 measured. The time of the peak acceleration from the firing of the piston was also 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 time to onset and peaking of the EMG in relation to the piston firing were measured and analyzed.

The 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 prior to accelerative impacts, providing strength measurement results in Newtons for each muscle [6, 8]. The ratio (percentage) of the EMG amplitude vs 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, Chicago, Ill.) to calculate descriptive statistics (with correction for multiple comparisons), correlation analysis between EMG and head acceleration, analysis of variance (ANOVA) of the EMG slope, time of peak EMG, time to onset of EMG, time to peak EMG, average EMG, and the force equivalents.

Table 1 Mean normalized peak electromyogram of cervical muscles in response to simulated right anterolateral impacts. *MVC* maximal voluntary contraction

Acceleration (m/s ²)	Normalized muscle EMG					
	Sternocleidomastoid (%MVC)		Splenius capitis (%MVC)		Trapezius (%MVC)	
	Left	Right	Left	Right	Left	Right
Unexpected						
5.1	8 (6)	4 (1)	26 (17)	17 (7)	38 (13)	32 (20)
8.7	12 (9)	5 (2)	29 (6)	19 (8)	63 (26)	40 (25)
12.0	19 (18)	7 (3)	44 (25)	20 (11)	79 (36)	54 (45)
13.7	23 (18)	13 (6)	46 (31)	30 (23)	83 (32)	58 (35)
Expected						
5.1	9 (4)	7 (2)	12 (6)	21 (14)	32 (8)	30 (21)
8.7	7 (3)	8 (2)	16 (7)	24 (17)	44 (14)	38 (22)
12.0	13 (10)	11 (4)	18 (5)	22 (13)	48 (14)	40 (26)
13.7	12 (6)	11 (2)	23 (7)	23 (13)	53 (16)	44 (19)

Values in parentheses represent one standard deviation

Fig. 1 An timeline of the acceleration impulse (the example is an unexpected impact of 13.7 m/s^2) and the timing of various muscle responses. *LSCM* left sternocleidomastoid, *RSCM* right sternocleidomastoid, *LSPL* left splenius capitis, *RSPL* right splenius capitis, *LTRP* left trapezius, *RTRP* right trapezius

Results

The ten subjects had a mean age of 24.5 ± 2.0 years, a mean height of 172 ± 5.2 cm, and a mean weight of 70 ± 12.6 kg.

Electromyogram amplitude in right anterolateral 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 are presented in Table 1.

Figure 1 illustrates the an approximated timeline of the acceleration impulse (one example of 13.7 m/s^2) and the timing of various muscle responses, here given as an example of the unexpected impact, showing in effect how the volunteer's muscle were loaded. As the level of applied acceleration in a front impact increased, the magnitude of the EMG recorded from the left trapezius and left splenius capitis increased progressively and disproportionately ($p < 0.05$). The EMG was greater during unexpected impacts ($p < 0.01$).

The normalized EMG showed that the percentage of sternocleidomastoids (SCMs), splenii capitis (SPLs), and trapezii (TRPs) magnitude increased steadily with the increasing magnitude of the impact and generally for both the expected and unexpected conditions (Fig. 2; Table 1). In a right anterolateral impact, with unexpected condition, at an acceleration of 13.7 m/s^2 , the left TRP exerted 83% and the left SPL 46% of the mean normalized maximal voluntary contraction, whereas the right TRP exerted 58% and the right SPL 30% of the mean normalized maximal voluntary contraction. Meanwhile, the SCMs never exceeded 23% of this variable. In terms of force equivalents, the TRP contralateral to the side of impact were required to resist the impact at a level near its maximal voluntary contraction capability.

Electromyogram timing

The time to onset of the sled, shoulder, and head acceleration in the z-axis (axis along impact direction) and the EMG signals of the six muscles examined are presented in Table 2. The time of the sled, torso, and head acceleration onset decreased with increased applied acceleration. Similarly, the time to onset of the EMG decreased with increasing applied acceleration, and the time to onset showed a trend in being shorter with unexpected impacts, but this did not reach statistical significance. The mean times at which peak EMG occurred for all the experimental conditions are presented in Table 3, and these were shorter with unexpected impacts ($p < 0.05$).

Sled and head acceleration

The sled acceleration impulse duration was inversely related to the magnitude of the pneumatic cylinder impact due to the piston excursion. It was 1.0, 0.6, 0.5, and 0.4 s for the acceleration levels of 1–4; however, the times to peak acceleration at these four levels of impact were 118, 87, 82, and 80 ms, respectively.

The kinematic response of the head to the four levels of applied acceleration in the expected and unexpected conditions is shown in Fig. 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 ($p < 0.01$). The relationship between the force equivalent EMG response of each muscle and the head acceleration is shown in Table 4. All force equivalents were well within the force generated by these muscles (determined in previous studies with these when same subjects being asked to perform maximal voluntary contractions) [6, 8].

Fig. 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

Statistical analyses

The applied acceleration, the muscles examined, and the volunteer's expectation had significant main effects on the peak EMG activity ($p < 0.05$). To justify the combination of the male and female EMG responses to applied acceleration, gender was entered into the analysis, and the re-

sults were nonsignificant, indicating that gender did not confound the results. Awareness of the impending impact significantly reduced the muscle activity ($p < 0.01$).

Initially, regression analyses were performed only up to 13.7 m/s² using linear, quadratic, cubic, power, and exponential functions; however, the power function was chosen as it fit the data best. The other functions also pro-

Table 2 Mean time to onset (in milliseconds) of acceleration and of muscle EMG from the firing of the solenoid of the pneumatic piston in a right anterolateral impact

Acceleration (m/s ²)	Acceleration Onset Time			Muscle Activity Onset Time					
	Sled	Shoulder	Head	Sternocleidomastoid		Splenius capitis		Trapezius	
				Left	Right	Left	Right	Left	Right
Unexpected									
5.1	78 (15)	119 (22)	190 (69)	342 (417)	1552 (1127)	244 (121)	301 (258)	197 (39)	155 (47)
8.7	41 (11)	93 (24)	153 (31)	334 (392)	493 (534)	243 (183)	332 (265)	116 (18)	162 (93)
12.0	37 (7)	77 (13)	153 (56)	335 (530)	1140 (1558)	141 (15)	219 (175)	130 (11)	140 (19)
13.7	35 (15)	83 (24)	141 (59)	440 (815)	229 (126)	92 (50)	190 (163)	126 (25)	129 (29)
Expected									
5.1	65 (9)	112 (12)	150 (36)	1162 (1342)	225 (318)	195 (142)	127 (147)	123 (51)	180 (134)
8.7	42 (7)	74 (15)	125 (24)	667 (1155)	238 (199)	130 (31)	161 (323)	133 (18)	110 (15)
12.0	41 (19)	84 (13)	158 (74)	756 (1098)	835 (986)	114 (23)	786 (1055)	120 (7)	122 (13)
13.7	30 (11)	83 (25)	120 (16)	651 (942)	1208 (1181)	113 (5)	1146 (1172)	122 (4)	109 (11)

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 time to onset of EMG activity. Values in parentheses represent one standard deviation

Table 3 Mean time (in milliseconds) at which peak electromyogram occurred after the firing of the solenoid of the pneumatic piston

Acceleration (m/s ²)	Muscle EMG					
	Sternocleidomastoid		Splenius capitis		Trapezius	
	Left	Right	Left	Right	Left	Right
Unexpected						
5.1	937 (708)	873 (1098)	227 (67)	731 (368)	284 (64)	446 (293)
8.7	206 (87)	991 (860)	216 (9)	981 (770)	253 (67)	423 (316)
12.0	621 (756)	302 (76)	243 (51)	444 (512)	266 (23)	255 (44)
13.7	190 (51)	290 (110)	231 (77)	365 (295)	246 (37)	352 (283)
Expected						
5.1	967 (561)	756 (398)	762 (956)	1051 (767)	229 (28)	862 (923)
8.7	974 (729)	654 (448)	601 (621)	623 (372)	238 (116)	290 (102)
12.0	622 (628)	765 (378)	219 (42)	797 (781)	227 (51)	233 (54)
13.7	499 (205)	784 (268)	217 (42)	567 (347)	228 (108)	212 (75)

duced good fit, but extrapolation using those functions did not always seem to follow the intuitive mechanical behavior. The kinematic variables of head displacement, velocity, and acceleration in response to applied acceleration were calculated (see Figs. 4, 5 with the TRP muscles as an example). 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

This study did not take into consideration the many other parameters which occur in vehicle collisions, including seat design and restraint systems. These factors may obviously affect muscle responses as well. We have, however, begun the process of a larger series of investigations by showing what effect increasing acceleration, impact direc-

tion, and expectation has on muscle responses when other factors are held constant (i.e., seat and restraint type). Future studies can build on this and determine how different seat design or other factors that exist in vehicles affect muscle responses when other factors (such as acceleration, expectation, and direction) are held constant. The 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

Fig. 3 Head acceleration in the x, y, and z axes of one subject 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

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.

The chief purpose of the study was to see what effect an offset impact had on muscle responses, knowing as we

did that in straight-on rear impacts the greatest muscle response arises from the sternocleidomastoids rather than the trapezii or splenius capitis muscles [5, 7], whereas in straight-on frontal impacts, the trapezius muscles show the greatest muscle response, in a symmetrical fashion [9]. Although one might predict the response from the trapezii in a right anterolateral impact, the human response

Table 4 Mean force equivalents and mean head accelerations at time of maximal EMG in direction of travel for right anterolateral impact

Chair acceleration (m/s ²)	Head acceleration (m/s ²)	Force equivalents for muscle (<i>n</i>)					
		Sternocleidomastoid		Splenius capitis		Trapezius	
		Left	Right	Left	Right	Left	Right
Unexpected							
5.1	2.4 (0.5)	3 (3)	4 (3)	21 (8)	16 (5)	13 (4)	9 (3)
8.7	4.2 (0.4)	5 (3)	4 (3)	23 (5)	17 (6)	21 (7)	11 (4)
12.0	6.6 (2.3)	6 (4)	5 (4)	33 (11)	19 (8)	26 (9)	13 (11)
13.7	7.3 (1.5)	8 (5)	7 (6)	34 (12)	20 (5)	31 (15)	17 (10)
Expected							
5.1	2.1 (0.4)	5 (5)	6 (1)	11 (11)	19 (18)	17 (9)	16 (8)
8.7	3.7 (0.7)	6 (5)	5 (3)	13 (15)	21 (18)	24 (11)	19 (6)
12.0	5.2 (0.6)	5 (5)	6 (4)	15 (12)	21 (19)	24 (10)	21 (9)
13.7	5.5 (0.9)	6 (6)	6 (3)	19 (15)	20 (17)	29 (21)	25 (9)

Values in parentheses represent one standard deviation

Fig. 4 Extrapolated regression plots of the effect that applied acceleration has on the head motion variables of displacement (in millimeters), velocity (in millimeters per second), and acceleration (in millimeters per second squared) obtained

to impacts and the neck structure is seemingly complex enough that it cannot always be assumed to be as predicted. Our study methodology allowed for direct testing

of the response rather than assumptions. In a straight-on front impact, the burden of the impact is again primarily responded to by the trapezii and the sternocleidomastoids and splenius muscles are not significantly active [9]. With a right anterolateral impact (the frontal impact is offset 45° to the subject's right), however, part of the impact burden is experienced more by the left splenius capitis, the contralateral muscle. Under these conditions, the left trapezius still generated the greatest EMG, but only up to 83% of the maximal voluntary contraction, and the left splenius capitis instead became more active and reached a level of 46% of this variable. This is greater than the response of the splenius capitis in straight-on impacts; thus, direction of impact determines which muscles respond and the proportionality of the response among the different muscle groups.

Until recent years, electromyographic studies were seldom employed in studying the controversial and important problem of whiplash injury. Recent studies have shown promise in applying electromyographic approaches to the clinical and biomechanical issues that arise in whiplash patients. Surface EMG has been used, for example, to demonstrate the changes in sternocleidomastoid muscle (SCM) activity as a means to objectively visualise the neutral zone of spine movement in whiplash-associated disorders I–III [11]. 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 to determine if there is an objective improvement, even if the patient still has pain.

Surface EMG studies in whiplash patients 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 muscles between a group of WAD patients and a matched set of healthy control subjects [12]. 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

Fig. 5 Extrapolated regression plots of the effect that applied acceleration has on the left and right trapezius muscles for the variables of peak EMG (V), normalized EMG (percentage of isometric maximal voluntary contraction), and force equivalent of EMG (N)

the ability to relax the cervical muscles after being subject to a physical load. In a second study this muscle “hyper-reactivity” also appeared to be present in a group of non-specific neck pain patients without a traumatic onset, leading the authors to conclude that this muscle activation pattern is not related to the specific mechanism associated with a trauma [13]. This same group more recently analyzed the muscle activation patterns of the upper trapezius 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. Surface EMG measurements revealed no elevated muscle reactivity either in the acute stage, or during the follow-up period. They found an inverse relationship between the level of neck pain disability and EMG level and that 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 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. 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. Studies suggest a central role may be played by the cervical muscles in injury causation during low-velocity collisions [16, 17].

As we have discussed in detail elsewhere with a review of the relevant EMG data [5, 7], studies in neck perturbations suggest that the cervical muscle response is triggered by peripheral input of muscle stretch. 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. As in straight-on frontal impacts, we also found that being unaware of the impact (unexpected condition) significantly increased aspects of the muscle response (peak EMG and normalized EMG); thus, expectation may also play a role in probability of injury in low-velocity impacts.

Although there is no direct way to measure forces exerted by muscles due to neck perturbation and subsequent muscle activity, examining the 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 physiological ranges of activity. We are providing here extrapolations only. All extrapolations are approximations of the truth and are always subject to a degree of error for various reasons. Because one cannot test the higher accelerations for ethical reasons, the best one can do currently is to compare to the small volunteer studies that were done previously. Further studies with larger samples and perhaps somewhat higher accelerations (within ethical limits) will allow to determine further how reasonable these extrapolations are. This experiment is a first approximation. 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. We may be able to discern 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; otherwise, future experiments will examine the effect other directions of impact and nonneutral head positions at time of impact.

Conclusion

Overall, the increasing use of surface EMG is providing promising research approaches and data in so-called whiplash collision experiments. 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 is collected. Because of the ethical concerns in doing so, our approach as demonstrated in this current study and previous studies is more useful. Our extrapolations closely match those from small volunteer studies where higher velocities were used with symptoms produced. This suggests that regression techniques may allow for extrapolations into low-velocity ranges, and may obviate the need for exceeding ethical concerns with experimental designs that could cause volunteer injury.

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