



Kinematic and electromyographic response to whiplash-type impacts. Effects of head rotation and trunk flexion: Summary of research

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Abstract

Despite the fact that whiplash patients often report they had their head rotated or were in a twisted posture at the time of impact, the effect of these postures on the cervical muscle response to impact remains uninvestigated in impact studies. Prior impact studies have positioned the volunteers in the recommended driving position, for example, with head and trunk in a neutral posture. Using an approach of sled impacts with volunteers in very-low velocity impacts to describe the head kinematics and cervical muscle electromyography in response has provided a wealth of data. From this approach, the effect of varying impact direction and level of impact awareness can be discerned without subjecting the volunteers to injury. In part 1 of this review, a further series of results of impacts from eight directions is presented, revealing that the cervical electromyography response to whiplash-type impacts varies according to the presence and direction of head rotation. In part 2, additional data is summarised concerning whiplash-type impacts from 8 directions in the presence of trunk flexion. Contrary to a popular notion, head rotation or trunk flexion at the time of impact are factors that probably reduce injury risk. This data adds to attempts to approach an understanding of the human response to more complex scenarios of low-velocity road collisions.

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1. Introduction

Previously in this journal, Kumar et al. (2004a) described the history of volunteer studies of whiplash loading, particularly focusing on low-velocity impacts, which account for a substantial number of whiplash injury claims (Castro et al., 1997; Fast et al., 2002). The pathology of the acute whiplash injury remains unknown, even though the mechanism of acceleration and deceleration of the head is commonly used in the definition of

“whiplash injury” (Spitzer et al., 1995). Since the acute whiplash pathology is not readily detectable in most cases, engineering studies of low velocity impacts often rely on outcome measures such as symptoms as a surrogate of injury. This may be unreliable because symptoms may be non-specific and in fact can arise in placebo collisions (Castro et al., 2001). It still seems likely that muscle responses to acceleration of the head are reasonable candidates for injury in road collisions (Kumar et al., 2004a). While cadaveric and animal studies can define “damage” to tissue, muscle responses cannot be tested in cadaveric studies and thus require volunteers for study (MacNab, 1964; Gosch et al., 1972; Liu et al., 1973; Hu et al., 1977; Domer et al., 1979;

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Gennarelli et al., 1982; Liu et al., 1984; Dunsker, 1985; Kallieris and Schmidt, 1990; Panjabi, 1998; Panjabi et al., 2004).

Given ethical concerns with subjecting volunteers to injurious neck perturbations, however, surface electromyography (EMG) has been combined with regression techniques modelled on very-low velocity collisions. Using this approach, the results of the impact studies and the regression models are in good agreement with the available data concerning biomechanical responses documented in previous, small studies of higher velocity collisions (Ewing, 1975; Szabo and Welcher, 1992; Szabo et al., 1994; Szabo and Welcher, 1996; West et al., 1993; McConnell et al., 1993, 1995; Scott et al., 1993; Siegmund and Williamson, 1993; Siegmund et al., 1994; Siegmund et al., 2003; Matsushita et al., 1994; Rosenbluth and Hicks, 1994; Nielsen et al., 1997; Brault et al., 1998; Howard et al., 1998; Meyer et al., 1998; Kaneoka et al., 1999; Magnusson et al., 1999).

Thus far, it has been possible to map out the EMG responses of the splenius capitis, sternocleidomastoid, and trapezius muscles to eight directions of impact: frontal, rear, right and left lateral, right and left posterolateral, and right and left anterolateral (Kumar et al., 2002a, 2003, 2004b,c,d,e,f,g,h). It has been shown that the trapezii mainly bear the burden of the neck perturbation in frontal impacts (Kumar et al., 2003), the sternocleidomastoids bear most of the burden in rear impacts (Kumar et al., 2002a), and lateral impacts tend to more equally distribute the burden of impact against these latter muscles (Kumar et al., 2004c,d). More specifically in lateral impacts, the muscles contralateral to the direction of impact bear the impact burden more than their counterparts. The intermediate directions produce cervical muscles responses that are somewhat predictable from the findings of the cardinal directions of impact (Kumar et al., 2004e,f,g,h).

Yet, these studies were all conducted with the volunteers seated with their head and trunk in a neutral position, such as the recommended driving position. Anecdotally, whiplash patients report the belief that because they had their head turned or that they were leaning over to one side at the time of impact is the reason they have unilateral pain. This same expression of relevance of head and/or trunk position is commonly made in medicolegal reports as a reason why a low-velocity collision may have caused injury despite the seemingly minor nature of the collision: the effect of head or trunk position amplifies, it is presumed, the injury mechanism. Biomechanical studies of automobiles, and impact studies with dummies have indeed suggested that, in theory, being in a trunk-flexed posture at the time of rear impact may increase the risk and severity of an acute injury (Hu et al., 1977; Foret-Bruno et al., 1990; Warner et al., 1991). This is partly based on the finding that when the dummy's torso is in full contact with the seat, seat

deformation may reduce neck injury risk by virtue of less torso-head lag, and also that when there is a lack of full contact, the torso may "ramp up" the seat back and more neck extension may occur. Despite these considerations, there have to date been no volunteer impact studies of the cervical response to impacts with the head rotation or trunk flexion at the time of impact.

This article deals with the volunteer impact research illustrating the effect of head rotation to 45° on the cervical response to whiplash-type impacts, as well as the effect of trunk flexion to 45° right and left under the same impact conditions. Some of this work remains in progress and some has been published (Kumar et al., 2005a,b,c,d,e,f). These data may help to clarify some of the concepts concerning the effect of occupant position on the response to neck perturbation.

2. Methods

As previously described (Kumar et al., 2004a), the ethical problem of injury to volunteers has been circumvented by using other specialized techniques to understand the response of individual neck muscles to being loaded with the forces of sudden accelerations. These techniques include electromyography, accelerometry, special sleds, tracks, impacting mechanisms, incremental impacts in the safe limits and predictive techniques to discern responses at higher impact velocities. An acceleration device is used as in Fig. 1, consisting of an acceleration platform and a sled. The acceleration platform has parallel tracks that permit smooth gliding of the sled on the rails, with a low coefficient of friction of 0.03. This assembly allows a maximum linear speed of up to 22 mph (36 km/h) without significant speed loss. At one end of the platform, a pneumatic cylinder is connected to an air supply and mounted rigidly on the acceleration platform. The device is calibrated for the delivery of known forces causing specific sled accelerations, from 4 to 14 m/s². The opposite end of the platform is equipped with a high-density rubber stopper in the sled's path to absorb the impact and prevent it from sliding off the platform (see Fig. S1 online).

The sled consists 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 is further equipped with a footrest and four buckled straps to stabilise the lower extremities, while the seat is fitted with a five-point seat restraint system. Before sled acceleration, the cervical strength of the volunteers is measured to provide some "yardstick" by which to gauge the magnitude of the EMG response to impact (Kumar et al., 2001). That is, it is expected that the more a muscle's EMG response exceeds the physiological maximal voluntary contraction level, the more likely it is to be injured. If the EMG response to

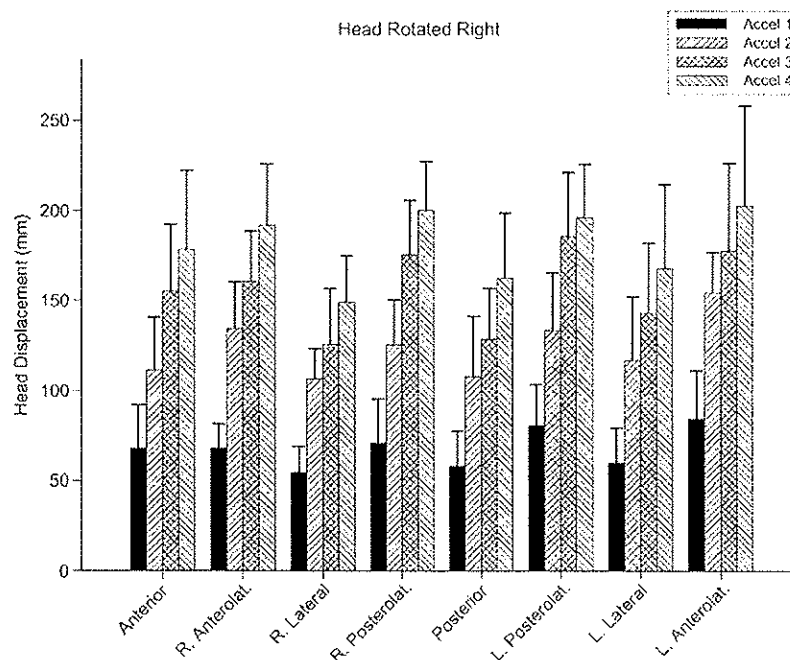


Fig. 1. Peak head displacement in response to accelerative impacts for 8 directions of impact with head rotated right. Applied acceleration is approximately 13 m/s^2 . Directions: 1 = rear, 2 = right posterolateral, 3 = right lateral, 4 = right anterolateral, 5 = frontal, 6 = left anterolateral, 7 = left lateral, 8 = left posterolateral.

an impact is well within the envelope of maximal voluntary contraction, then it is assumed that injury is less likely to occur to that muscle.

The seated and stabilized subjects exert their maximum effort in attempted flexion, extension, and lateral flexion to the left and the right. This data is then used to calculate a maximal voluntary contraction EMG for each of the sternocleidomastoid, splenius capitis, and trapezius muscles, by which the peak EMG response to impact for each muscle will be normalised.

Volunteers are subjected to sudden accelerations in a manner which mimics vehicle collisions in terms of the suddenness and duration of movement. The sled device is such that volunteers can face the direction of travel, or be turned 180° around to travel backward, and further can be seated at angles varying from 45° to 135° from the direction of travel.

Two triaxial accelerometers are fixed to the volunteer: 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 data collection is initiated, and after 1 s the pneumatic piston is fired to accelerate the sled. At the same time, the neck muscles are studied by using standardized active surface electrodes attached to the subject's skin, over specific neck muscle groups, including bilateral sternocleidomastoid (SCM), splenius capitis (SPL), and trapezius muscles (TRP). The seated and stabilized subjects are then exposed to sled accelerations ranging from 4 to 14 m/s^2 in a random order by a pneu-

matic piston. To examine the effect of head rotation, subjects are exposed in different experiment series to eight directions of impact with their head rotated 45° to their left and then right at each acceleration generated in a random order.

The methodology is the same for studies of impacts with trunk flexion, with the one variation being the subject positioning at the time of impact. There is no attempt to have the subjects completely relaxed with the neck fully flexed (i.e., slumped posture), as it is not expected this would be typical of road collisions. Volunteers are positioned in 45° forward flexion and 45° lateral flexion either to the left or to the right. The subjects were asked to look down at their right or left foot. (The positioning of the subject is illustrated in Fig. S2 online.) Though a component of head rotation cannot be excluded in this approach, the goal is to produce a posture that was clearly different from the neutral head and trunk posture, and yet within the range of what might occur in a vehicle occupant who, for example, was reaching for an object at the time of impact. In the series of experiments presented here, all participants in this study were aware of an impending impact. This is known to reduce the overall EMG response compared to impact studies where we have blocked the auditory and visual cues for the subject (Kumar et al., 2000, 2002a,b, 2003, 2004b,c,d,e,f,g,h). Yet, there is complete data for this type of scenario and while the magnitude of EMG response may vary with awareness of impact, the pattern of the muscle response (i.e., which muscle

has the greatest response to impact) does not change with awareness of impact.

We used a lap belt in this study in order to standardise the subject positioning without interference from the shoulder strap. When occupants lean over from their driving position, for example, or try to pick up something from the vehicle floor, they are sliding out from behind the restraint of the shoulder strap, and effectively are restrained only by the lap portion. Still, studies are underway to compare the effect of different restraint systems on cervical muscles response as this may remain relevant.

The data on the peak and average accelerations in all three axes of the sled, C7, and head and four levels of accelerative impacts are measured. The EMG signals from a total of 6 muscles (3 pairs) are analysed. 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 peak of the EMG in relation to the solenoid firing are measured and analyzed.

The subjects are exposed to incremental accelerations from different directions in a random order. The accelerations used are high enough to generate muscle responses from impact, but not high enough to cause injury. From the data collected, statistical regression is undertaken to extrapolate what types of neck muscle forces, head accelerations, etc. will occur in the low velocity range.

3. Results part 1. Impacts with head rotation

3.1. Kinematics in 8 directions of impact and two directions of head rotation

As stated above, besides subjecting volunteers to 8 directions of impact, it is also possible, for any given direction of impact, to study the variable of head rotation. Fig. 1 (and see Fig. S3 online) presents the head displacement for 8 directions of impact and for right and left head rotation at one level of acceleration. Comparing these responses to previous figures with the head and trunk in neutral position (Kumar et al., 2004a), it is apparent that head displacement is less when the head is rotated, a factor which may affect injury risk. Similarly Figs. 2 and S4 online present the head acceleration under these conditions. Again, both head velocity and acceleration are decreased when the impact is experienced with the head rotated at the time of impact as compared to when the head is in the neutral position.

3.2. EMG findings of head rotation at time of impact

Tables 1 and 2 and Tables S1 and S2 online show the time to onset of EMG and time to peak EMG for 6 muscles in straight-on rear and frontal impacts with the head rotated right or left. The onset of EMG is the time at which the EMG activity is 2% above baseline. Table 1 reveals, for example, the times to EMG onset in a rear

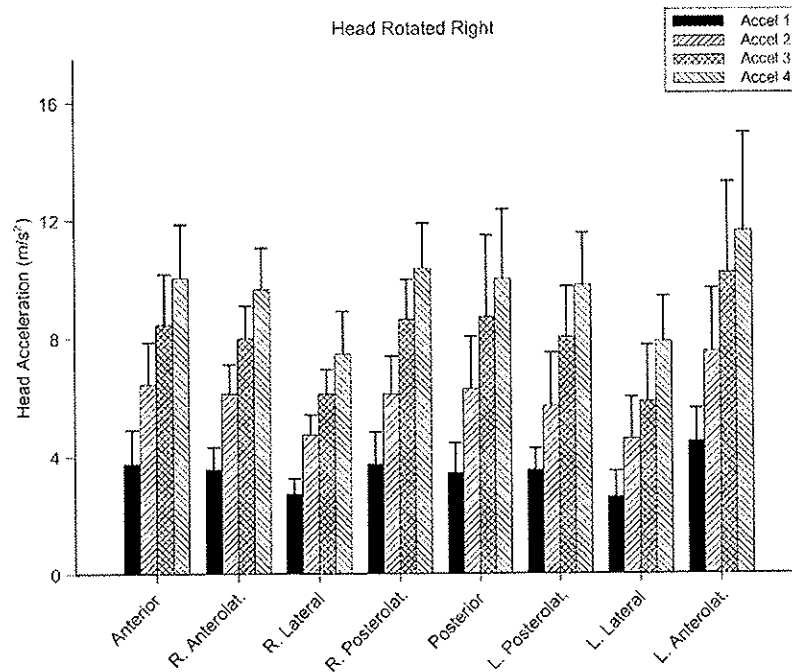


Fig. 2. Peak head acceleration in response to accelerative impacts for 8 directions of impact with head rotated right. Applied acceleration is approximately 13 m/s^2 . Directions: 1 = rear, 2 = right posterolateral, 3 = right lateral, 4 = right anterolateral, 5 = frontal, 6 = left anterolateral, 7 = left lateral, 8 = left posterolateral.

Table 1
Mean time to onset (ms) of muscle EMG from the time of motion onset in a rear impact

| Acceleration (m/s ²) | Muscle | | | | | |
|----------------------------------|---------------------|----------|------------------|-----------|-----------|-----------|
| | Sternocleidomastoid | | Splenius capitis | | Trapezius | |
| | Left | Right | Left | Right | Left | Right |
| <i>Head rotated right</i> | | | | | | |
| 4.2 | 128 (38) | 157 (29) | 128 (44) | 234 (118) | 254 (117) | 263 (90) |
| 8.1 | 101 (30) | 124 (22) | 96 (26) | 215 (100) | 235 (133) | 254 (105) |
| 10.4 | 100 (29) | 120 (19) | 92 (30) | 191 (127) | 196 (117) | 186 (103) |
| 13.0 | 97 (27) | 118 (20) | 78 (41) | 154 (79) | 152 (96) | 164 (101) |
| <i>Head rotated left</i> | | | | | | |
| 4.2 | 155 (25) | 100 (54) | 225 (55) | 118 (45) | 213 (108) | 248 (112) |
| 8.1 | 128 (24) | 97 (30) | 218 (70) | 89 (33) | 206 (76) | 228 (131) |
| 10.4 | 124 (23) | 95 (33) | 206 (72) | 84 (36) | 193 (90) | 160 (88) |
| 13.0 | 123 (18) | 92 (31) | 203 (95) | 82 (34) | 163 (84) | 116 (71) |

Times for the cervical muscles represent the time to onset for EMG activity. Values in parentheses represent one standard deviation. (See Kumar et al., 2005b for further details.)

Table 2
Mean time (ms) to peak muscle EMG from the from the time of motion onset in a rear impact

| Acceleration (m/s ²) | Muscle | | | | | |
|----------------------------------|---------------------|----------|------------------|-----------|-----------|-----------|
| | Sternocleidomastoid | | Splenius capitis | | Trapezius | |
| | Left | Right | Left | Right | Left | Right |
| <i>Head rotated right</i> | | | | | | |
| 4.2 | 226 (17) | 240 (13) | 322 (231) | 567 (339) | 715 (366) | 496 (306) |
| 8.1 | 213 (33) | 213 (18) | 238 (89) | 515 (340) | 700 (346) | 487 (318) |
| 10.4 | 195 (15) | 207 (17) | 223 (82) | 434 (321) | 605 (613) | 480 (210) |
| 13.0 | 192 (17) | 192 (15) | 223 (61) | 425 (263) | 561 (349) | 454 (231) |
| <i>Head rotated left</i> | | | | | | |
| 4.2 | 230 (24) | 229 (30) | 451 (304) | 247 (63) | 566 (362) | 650 (304) |
| 8.1 | 206 (20) | 199 (23) | 438 (199) | 215 (61) | 510 (296) | 494 (253) |
| 10.4 | 205 (20) | 198 (15) | 405 (215) | 206 (55) | 507 (287) | 489 (236) |
| 13.0 | 201 (16) | 195 (18) | 390 (263) | 202 (18) | 496 (167) | 428 (249) |
| 13.0 | 226 (17) | 240 (13) | 322 (231) | 567 (339) | 715 (366) | 496 (306) |

Times for the cervical muscles represent the time to onset for EMG activity. Values in parentheses represent one standard deviation. (See Kumar et al., 2005b for further details.)

impact. The SCMs are known to generate the greatest EMG response in a rear impact (see below and Kumar et al., 2002a) and these muscles show an overall decrease in time to onset of EMG as the acceleration levels increase. At an acceleration of 4.2 m/s², for example, with the head rotated to the right, the mean time to onset of EMG for the left sternocleidomastoid muscle (SCM) is 128 ms (SD 38) and for the right SCM it is 157 ms (SD 29). At the higher acceleration of 13.0 m/s², the mean time to onset of EMG for the left SCM is 97 ms (SD 27) and for the right SCM it is 118 ms (SD 20). The TRP and SPL muscles are not as active or relevant in a rear impact and they have onset times that bear less correlation to the applied acceleration, but even here they show a trend to decreased times to onset with increasing applied acceleration. When we compare these times to EMG onset with head rotation to those previously shown in this journal (Kumar et al., 2004a) for impacts with the head in neutral posture, there are no

significant differences overall. This suggests that head rotation does not delay the rate of initial muscle response to the neck perturbation.

For a frontal impact, TRP show the greatest EMG response (see below and Kumar et al., 2003). With the head rotated to the right, at an applied acceleration of 4.2 m/s², in a frontal impact, the left TRP has an EMG onset time of 122 ms (SD 42) and the right TRP 115 ms (SD 40). When the applied acceleration is 13.0 m/s², the left TRP has an EMG onset time of 90 ms (SD 36) and the right TRP 92 ms (SD 30). For the SCMs and SPLs, which are not as relevant in a frontal impact, there appears to be less correlation between applied acceleration and time to onset of EMG (see Table S1 online).

Similar patterns are observed for time to peak EMG in rear impacts as shown in Table 2, with the muscle groups showing a trend towards decreased times to peak EMG with increasing applied acceleration. The timing

of the muscle response does not appear to be affected by head rotation at the time of impact. Also, as shown in Table S2 (online), the onset of peak EMG activity occurs sooner with increasing applied acceleration in a frontal impact. Again, comparing to EMG peak times with head rotation to those previously shown in this journal (Kumar et al., 2004a) for impacts with the head in neutral posture, there are no significant differences overall. This is further evidence that head rotation does not delay the timing of the muscle response to the neck perturbation.

As revealed in studies of other impact directions (Kumar et al., 2005a,b,c,d,e), the more active muscles in response to a given impact tend to show similar results to the above in terms of correlations of timing to applied acceleration.

With each direction of impact, the muscle that exhibits the greatest response via eccentric contraction varies. Fig. 3 provides polar plots of the EMG responses of the pairs of SCM, SPL and TRP at a specified acceleration with impacts in 8 different directions and with head rotated right or left. The plots emphasize the directional nature of the cervical responses. When this polar plot is compared to the plot of impacts with the head in neutral position under the same impact circumstances, we see that the directional nature of the response persists, but the overall EMG magnitude is lessened for each muscle.

This directional response and the incremental response to increasing acceleration is further shown in Figs. 4–6 (also Figs. S5–S7 online). Here we plot the EMG response of the cervical muscles for 8 different directions, under conditions of head rotated right or left. Again, the direction of impact significantly alters the pattern of cervical response. In a rear impact (see in particular Figs. 4 and S5 online), the SCMs generate the

highest EMG responses, but not in a symmetrical fashion. At an acceleration of 13.0 m/s^2 , with the head rotated to the left, the right SCM generated 88% of its maximal voluntary contraction (MVC) electromyogram (at least triple the response of other muscles). In comparison, the left SCM, both TRPs, and the SPLs generated on average only 10–30% of their MVC electromyogram with head rotated to the left. On the other hand, with the head rotated to the right, the left SCM generated 94% of its MVC electromyogram (again, at least triple the response of other muscles). Here then is a significant difference from rear impacts with the head in neutral position: when the head is rotated an asymmetric muscle response occurs among the pair of SCM muscles.

Similarly, in a frontal impact (see in particular Figs. 5 and S6 online), with the head rotated to the left, the left TRP generated 77% of its MVC EMG (more than double the response of other muscles). In comparison, the right TRP generated only 33% of its MVC. The right SCM (the one responsible for head rotation to the left) was more active than its counterpart. On the other hand, with the head rotated to the right, the right TRP generated 71% of its MVC EMG, while the left TRP generated only 30% of this value. Again, the left SCM (being responsible for head rotation to the right) was more active than its counterpart. This is different to frontal impacts with the head in neutral position, where pairs of muscles behaved symmetrically (Kumar et al., 2003).

In lateral impacts (see in particular Figs. 6 and S7 online), the burden of impact is more widely distributed with the contralateral SPL (contralateral to the direction of impact) being more active than its counterpart. As may be predicted from a biomechanical perspective, the intermediary directions of right and left posterolat-

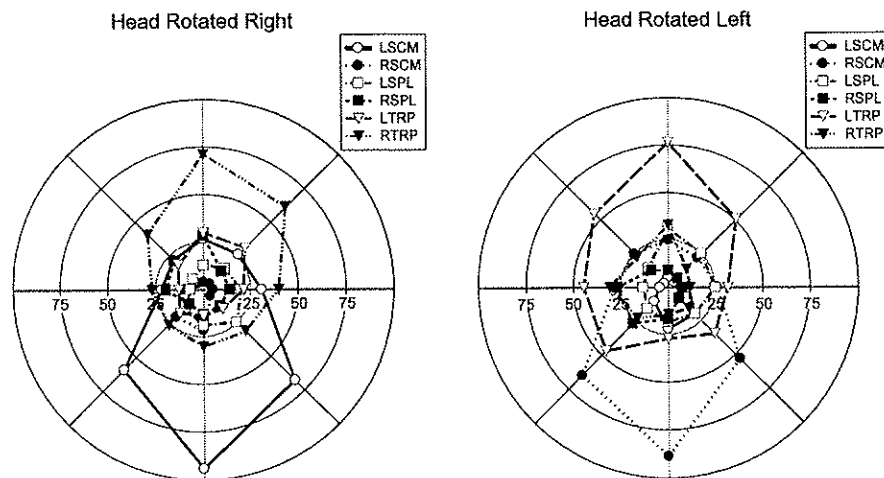


Fig. 3. Polar plot of EMG response to impacts with head rotation (right and left) under accelerations of 13.7 m/s^2 . LSCM, left sternocleidomastoid; RSCM, right sternocleidomastoid; LSPL, left splenius capitis; RSPL, right splenius capitis; LTRP, left trapezius; RTRP, right trapezius.

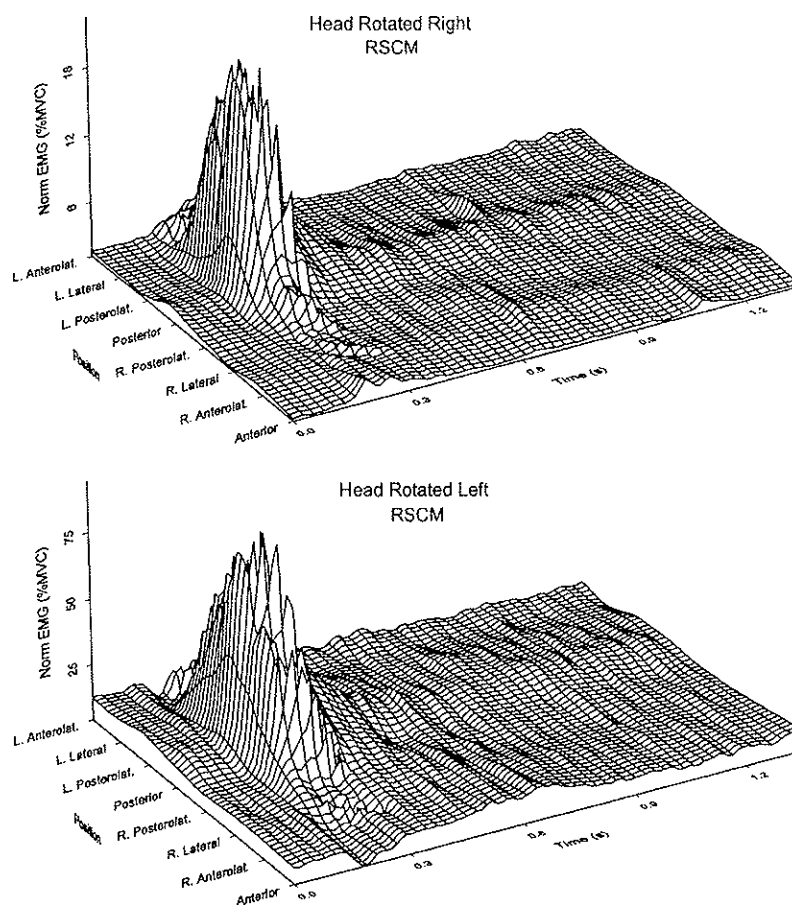


Fig. 4. Muscle activity of the right sternocleidomastoid muscle (RSCM) for 8 directions of impact with head rotated either right or left.

eral and anterolateral produce cervical responses which blend the findings of rear/lateral and frontal/lateral impacts, respectively.

Figs. 4–6, show moreover, that for any given direction of impact, the cervical response to whiplash loading may be mitigated by head rotation at the time of impact in the volunteer. Under the conditions of head rotation, the head experiences less acceleration and peak EMGs are lower than when the head is in neutral posture at the time of impact, all else being equal. The fact that there is asymmetry in the paired muscles responses, particularly noticeable with head rotation in rear and frontal impacts, would lend credence to the notion that asymmetric injury may occur. Yet, the head rotation itself, despite not affecting the timing of muscle responses, reduces the magnitude of the EMG response, and thus may mitigate injury risk.

This effect cannot be ascribed to an adaptation by the volunteers to repeated impacts. These impacts are carried out in random orders of severity. If subjects were adapting to repeated impacts with increasingly shorter EMG onset times, for example, the relationship between time to EMG onset and level of acceleration would be

lost. Clearly it is evident that a relationship between level of acceleration and timing of EMG onset or peak EMG remains.

4. Results part 2. Impacts with trunk flexion

4.1. Kinematics in 8 directions of impact and two directions of trunk flexion

As stated above, besides subjecting volunteers to 8 directions of impact, it is also possible, for any given direction of impact, to study the variable of trunk flexion. Figs. 7 and S8 online present the head displacement for 8 directions of impact and for right and left trunk flexion at one level of acceleration. If we compare these responses to data with the head and trunk in neutral position (Kumar et al., 2004a), it is apparent that head displacement is less when the trunk is flexed, suggesting this position may be protective, since it has long been assumed that head displacement relative to the trunk is a quintessential element of the whiplash mechanism. Similarly, Figs. 8 and S9 online present the head

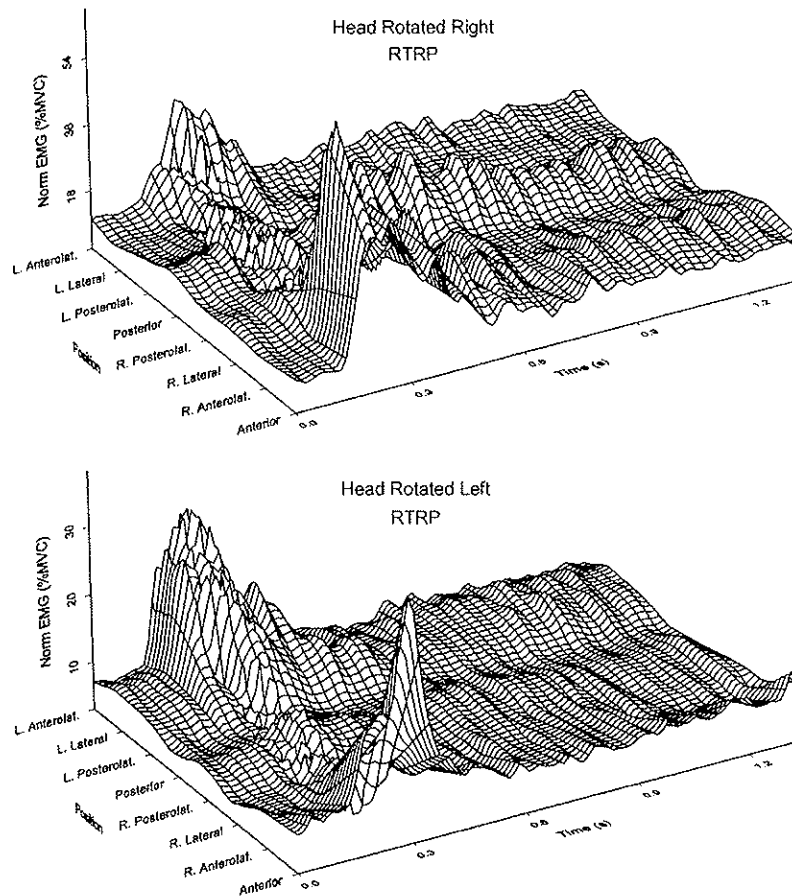


Fig. 5. Muscle activity of the right trapezius muscle (RTRP) for 8 directions of impact with head rotated either right or left.

acceleration under these conditions. Again, both head velocity and acceleration are decreased when the impact is experienced with the trunk flexed at the time of impact as compared to the head and trunk being in the neutral position.

4.2. EMG findings of trunk flexion at time of impact

Tables 3 and 4 to Tables S3 and S4 (online) show the time to onset of EMG and time to peak EMG for 6 muscles in straight-on rear and frontal impacts with the trunk flexed right or left. The onset of EMG is the time at which the EMG activity is 2% above baseline. Table 3 reveals, for example, the times to EMG onset in a rear impact. At an acceleration of 4.2 m/s^2 , with the trunk flexed to the right, the mean time to onset of EMG for the SCM is 146 ms (SD 26) and for the right SCM it is 307 ms (SD 155). At the higher acceleration of 13.4 m/s^2 , the mean time to onset of EMG for the left SCM is 132 ms (SD 20) and for the right SCM it is 178 ms (SD 100). The remaining muscles tend to either show a trend towards or have decreased times to EMG onset with increasing applied acceleration. The same is true with the trunk flexed to the left. For a fron-

tal impact, the TRP show the greatest EMG response (see below and Kumar et al., 2003). With the trunk flexed to the right, at an applied acceleration of 4.4 m/s^2 , in a frontal impact, the left TRP has an EMG onset time of 105 ms (SD 69) and the right TRP 114 ms (SD 62). When the applied acceleration is 13.3 m/s^2 , the left TRP has an EMG onset time of 102 ms (SD 24) and the right TRP 80 ms (SD 34). For all muscles there is a trend to decreasing time to EMG onset with increasing applied acceleration (see Table S3 online). The same is true for the trunk flexed to the left. For either rear or frontal impacts, when we compare these times to EMG onset with trunk flexion to those previously shown in this journal (Kumar et al., 2004a,b,c,d,e,f,g,h) for impacts with the head and trunk in neutral posture, there are no significant differences overall in these trends except there is a trend for the SCMs and TRPs to have an earlier onset of EMG activity with the trunk in flexion. This suggests that trunk flexion does not significantly delay the rate of initial muscle response to the neck perturbation.

Similar patterns are observed for time to peak EMG in rear impacts as shown in Table 4, with the muscle groups showing a trend towards decreased times to peak

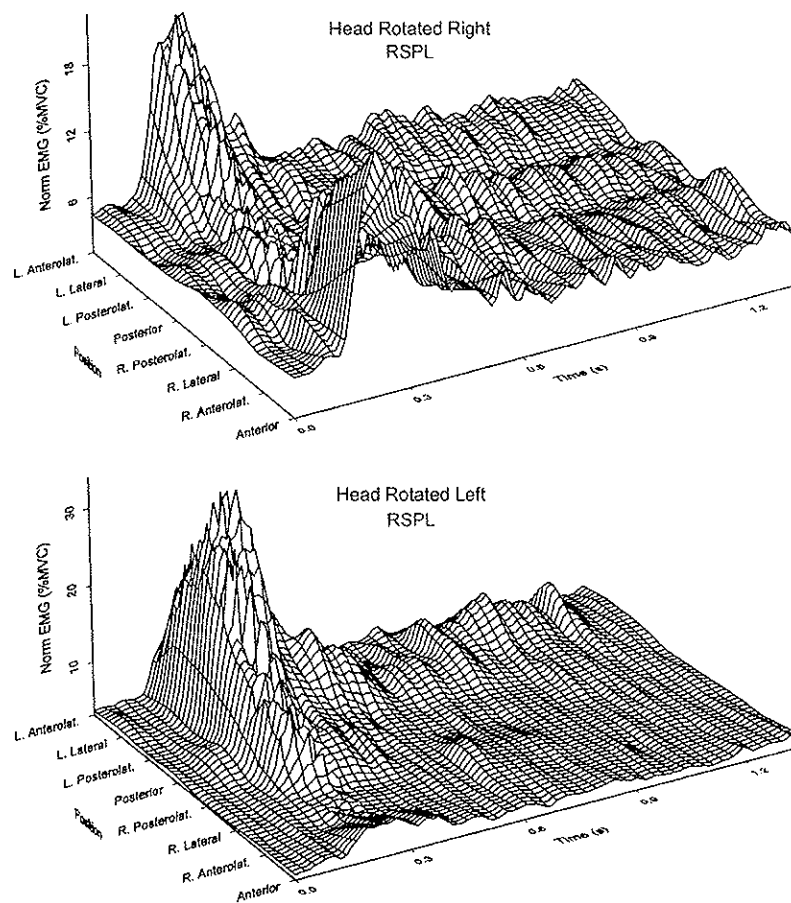


Fig. 6. Muscle activity of the right splenius capitis muscle (RSPL) for 8 directions of impact with head rotated either right or left.

EMG with increasing applied acceleration. Also, as shown in Table 8, the onset of peak EMG activity occurs sooner with increasing applied acceleration in a frontal impact. When we compare these times to EMG onset or peak EMG with trunk flexion to those previously shown in this journal (Kumar et al., 2004a) for impacts with the head and trunk in neutral posture, again there are no significant differences overall. This is further evidence that trunk flexion does not delay the timing of the muscle response to the neck perturbation.

In as yet unpublished data, we have found that for other impact directions (lateral, anterolateral, and posterolateral), the more active muscles in response to a given impact tend to show similar results to the above in terms of correlations of timing to applied acceleration.

With each direction of impact, the muscle that exhibits the greatest response via eccentric contraction varies. Fig. 9 provides polar plots of the EMG responses of the pairs of SCM, SPL and TRP at a specified acceleration with impacts in 8 different directions and with head rotated right or left. The plots emphasize the directional nature of the cervical responses. When this polar plot is compared to the plot of impacts with the head and

trunk in neutral position under the same impact circumstances, we see that the directional nature of the response persists, but the overall EMG magnitude is markedly lessened for each muscle.

This directional response and the incremental response to increasing acceleration is further shown in Figs. 10–12 and S10–S12 online. Here we plot the EMG response of the cervical muscles for 8 different directions, under conditions of trunk flexed right or left. Again, the direction of impact significantly alters the pattern of cervical response. In a rear impact (see in particular Figs. 10 and S10 online), the results are surprising. With the trunk flexed to the right or left, the TRP muscles showed the greatest EMG response compared to the remaining muscles, suggesting that the forward-flexed posture primes these neck extensors for action more so than other muscles. Recall from previous data (Kumar et al., 2004a), that the SCMs muscles are the most active and the TRP are the least active in response to a rear impact with the head and trunk in neutral position. As the level of applied acceleration in the rear impact increased, the magnitude of the EMG recorded from most muscles increased, but the normalised (to

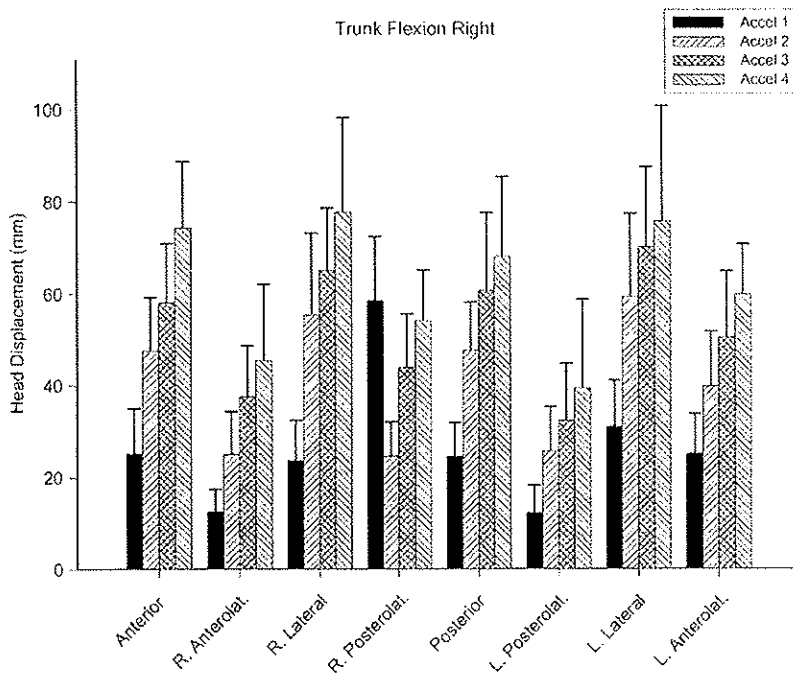


Fig. 7. Peak head displacement in response to accelerative impacts for 8 directions of impact with trunk flexed right. Applied acceleration is approximately 13 m/s^2 . Directions: 1 = rear, 2 = right posterolateral, 3 = right lateral, 4 = right anterolateral, 5 = frontal, 6 = left anterolateral, 7 = left lateral, 8 = left posterolateral.

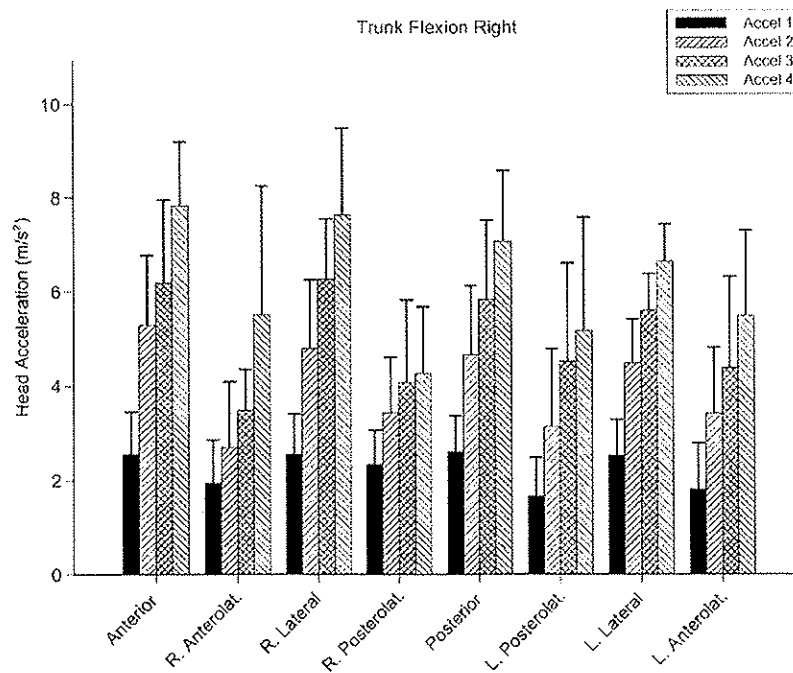


Fig. 8. Peak head acceleration in response to accelerative impacts for 8 directions of impact with trunk flexed right. Applied acceleration is approximately 13 m/s^2 . Directions: 1 = rear, 2 = right posterolateral, 3 = right lateral, 4 = right anterolateral, 5 = frontal, 6 = left anterolateral, 7 = left lateral, 8 = left posterolateral.

the muscle’s maximum voluntary contraction) EMG showed that the EMG was very low for either trunk

flexed to the left or right, the TRP generating 28% or less of the maximal voluntary EMG.

Table 3
Mean time to onset (ms) of muscle EMG from the time of motion onset in a rear impact

| Acceleration (m/s ²) | Muscle | | | | | |
|----------------------------------|---------------------|-----------|------------------|-----------|-----------|----------|
| | Sternocleidomastoid | | Splenius capitis | | Trapezius | |
| | Left | Right | Left | Right | Left | Right |
| <i>Trunk flexed right</i> | | | | | | |
| 4.4 | 146 (26) | 307 (155) | 172 (81) | 257 (49) | 180 (86) | 218 (33) |
| 7.9 | 146 (19) | 281 (73) | 136 (77) | 253 (51) | 178 (102) | 215 (83) |
| 10.7 | 139 (14) | 198 (109) | 122 (68) | 247 (89) | 173 (75) | 208 (75) |
| 13.4 | 132 (20) | 178 (100) | 121 (50) | 238 (111) | 158 (94) | 194 (72) |
| <i>Trunk flexed left</i> | | | | | | |
| 4.4 | 156 (73) | 162 (82) | 255 (47) | 142 (108) | 221 (52) | 185 (80) |
| 7.7 | 147 (51) | 136 (43) | 246 (61) | 133 (58) | 218 (81) | 184 (70) |
| 10.9 | 134 (44) | 123 (38) | 227 (82) | 131 (54) | 211 (96) | 169 (81) |
| 13.4 | 119 (68) | 112 (38) | 218 (102) | 120 (96) | 207 (93) | 163 (75) |

Times for the cervical muscles represent the time to onset for EMG activity. Values in parentheses represent one standard deviation. (See Kumar et al., 2005f for further details.)

Table 4
Mean time (ms) to peak muscle EMG from the from the time of motion onset in a rear impact

| Acceleration (m/s ²) | Muscle | | | | | |
|----------------------------------|---------------------|-----------|------------------|-----------|-----------|-----------|
| | Sternocleidomastoid | | Splenius capitis | | Trapezius | |
| | Left | Right | Left | Right | Left | Right |
| <i>Trunk flexed right</i> | | | | | | |
| 4.4 | 225 (42) | 709 (425) | 490 (387) | 608 (430) | 595 (414) | 410 (83) |
| 7.9 | 220 (21) | 709 (366) | 454 (279) | 532 (188) | 448 (221) | 397 (82) |
| 10.7 | 213 (20) | 427 (272) | 447 (229) | 504 (181) | 439 (210) | 382 (136) |
| 13.4 | 211 (21) | 320 (135) | 331 (126) | 476 (159) | 476 (159) | 372 (203) |
| <i>Trunk flexed left</i> | | | | | | |
| 4.4 | 720 (522) | 366 (163) | 577 (123) | 334 (88) | 460 (133) | 508 (247) |
| 7.7 | 596 (277) | 226 (37) | 565 (184) | 325 (117) | 424 (167) | 404 (152) |
| 10.9 | 528 (331) | 221 (35) | 562 (246) | 317 (129) | 393 (140) | 387 (129) |
| 13.4 | 470 (409) | 217 (32) | 505 (223) | 301 (137) | 378 (212) | 381 (138) |

Times for the cervical muscles represent the time to onset for EMG activity. Values in parentheses represent one standard deviation.

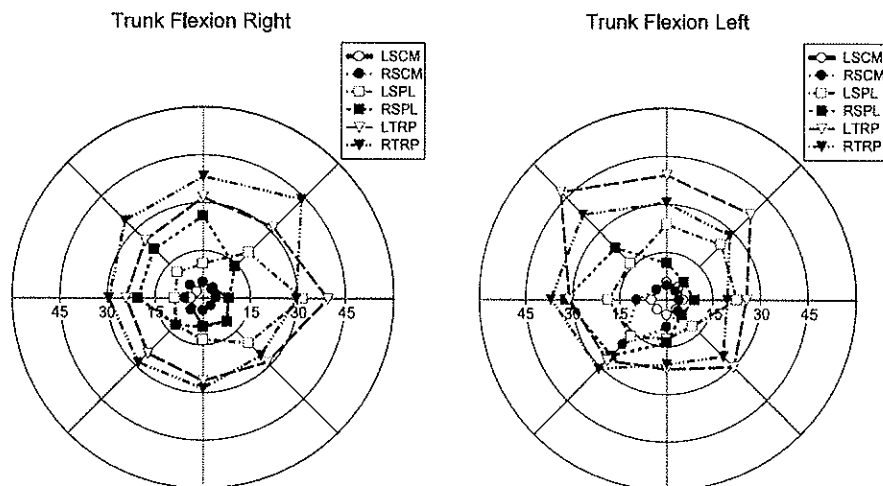


Fig. 9. Polar plot of EMG response to expected and unexpected accelerative impacts of 13.7 m/s². LSCM, left sternocleidomastoid; RSCM, right sternocleidomastoid; LSPL, left splenius capitis; RSPL, right splenius capitis; LTRP, left trapezius; RTRP, right trapezius.

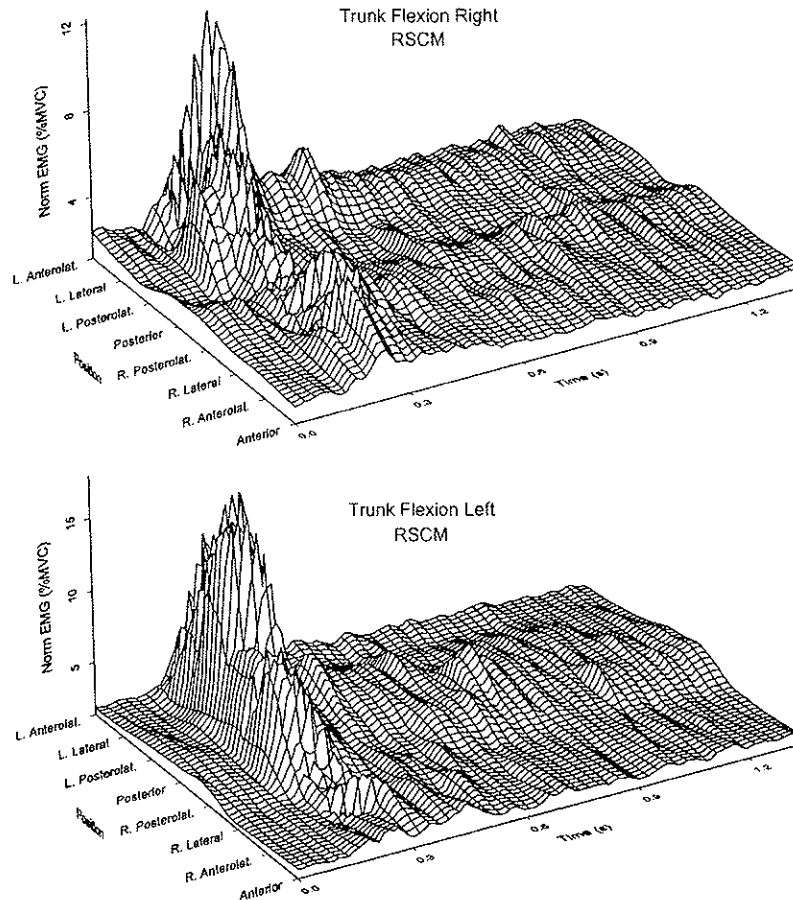


Fig. 10. Muscle activity of the right sternocleidomastoid muscle (RSCM) for 8 directions of impact with trunk flexed either right or left.

Similarly, in a frontal impact (see in particular Figs. 11 and S11 online), as the level of applied acceleration in the impact increased, the magnitude of the EMG recorded from the TRP increased progressively and disproportionately compared to other muscles. In a frontal impact, with trunk flexed to the right, the most significant finding is that the muscle responses were generally of low magnitude (38% or less for all muscles), the TRP muscles having the highest EMG activity. The ipsilateral SPL (the right one in right trunk flexion) had a greater EMG response than its counterpart. The same occurred when the trunk was flexed to the left, where the muscles had a 39% or less normalised peak EMG response, and the ipsilateral (to trunk flexion) SPL had a higher response than its counterpart. This is different from frontal impacts with the head and trunk in neutral position, where pairs of muscles behave symmetrically.

In lateral impacts (see in particular Figs. 12 and S12 online), the burden of impact is more widely distributed with the contralateral SPL (contralateral to the direction of impact) being more active than its counterpart. As an example, in a right lateral impact, with the trunk flexed to the right, there is an overall low magnitude of EMG

activity, with all of the muscles studied exhibiting peak EMGs at 39% or less of their MVC EMG. For the impacts where the trunk was flexed to the right, the left SPL muscle (the one contralateral to the impact direction) had a much higher activity than its counterpart, 31% versus 8%, respectively. With the trunk flexed to the left, again all muscles responded to the impact with relatively low EMG activity, 25% or less of the subject's MVC. The left SPL muscle was still more active than the right SPL, 22% versus 9%, respectively.

As may be predicted from a biomechanical perspective, the intermediary directions of right and left posterolateral and anterolateral produce cervical responses which blend the findings of rear/lateral and frontal/lateral impacts, respectively.

Figs. 10–12 and S10–S12 online, show, as alluded to above, that for any given direction of impact, the cervical response to whiplash loading is mitigated by trunk flexion at the time of impact in the volunteer. Under the conditions of trunk flexion, the head experiences less acceleration and the force equivalents for the neck muscles are lesser than when the head and trunk are in neutral posture at the time of impact, all else being equal.

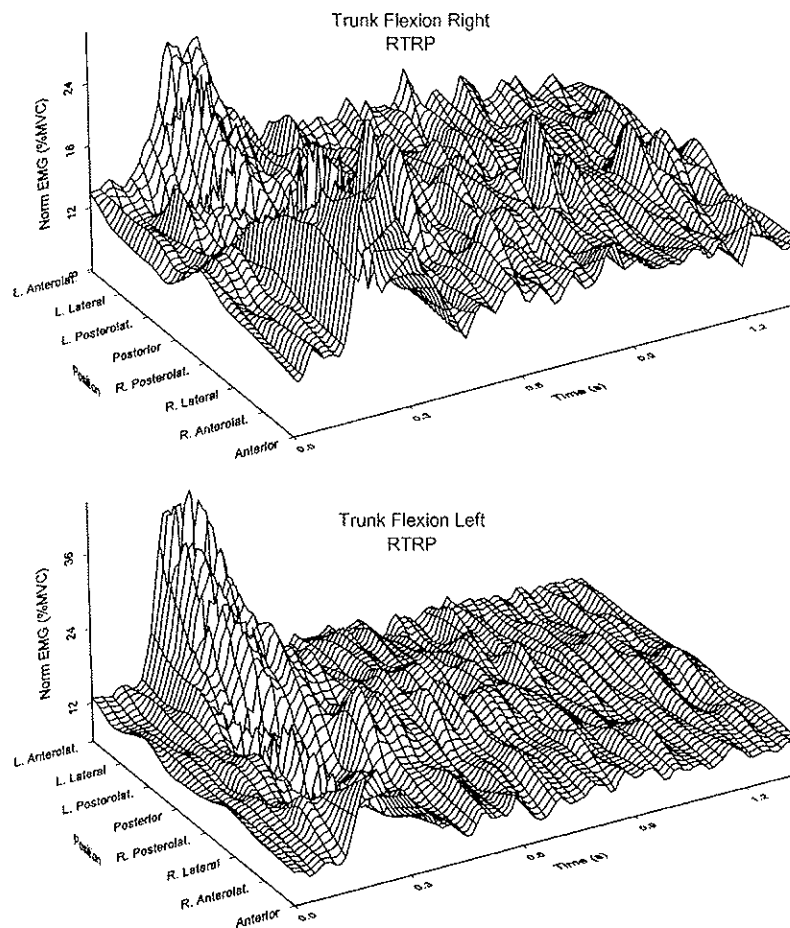


Fig. 11. Muscle activity of the right trapezius muscle (RTRP) for 8 directions of impact with trunk flexed either right or left.

Trunk flexion itself, despite not affecting the timing of muscle responses, reduces the magnitude of the EMG response.

5. Conclusion

Despite the anecdotal reports of importance of head and trunk position at the time of impact, both EMG and kinematic measures do not support the notion of greater muscle response or head perturbation in response to the impact. Using the techniques described above, research suggests that, if anything, having the head rotated or trunk flexed at the time of impact reduces the apparent result of the perturbation, if this result is measured by degree of EMG response and head displacement, etc.

Head rotation is not a passive event, but an active process, and require muscle contraction not only to achieve, but to maintain. The reason that head rotation may thus reduce the cervical response magnitudes is that it may be equivalent to being more “prepared” for the

impact. Previously, we reviewed how important simply having auditory and visual cues are to reducing the EMG response to impact (Kumar et al., 2004a). Holding the head in a fixed position may prepare the individual in a similarly protective way. It may also be that when the head is rotated there is less overall capacity for head displacement due to, for example, articular restriction, so less stretch of muscles in response to impact, and less muscle activation overall. Yet, the tolerances of capsular ligaments and joints is likely higher than that of muscles, so this reduced capacity for head displacement should not increase injury risk when the forces involved are well within the physiological envelope of muscle activity.

At the time of an impact, whiplash victims may also be leaning forward or leaning over as a result of watching for traffic or speaking with other occupants, etc. With trunk flexion, the magnitudes of the EMG response are lower than the responses previously shown for when the head and trunk are in neutral posture. For example, when the head and trunk are in neutral posture in a frontal impact, both TRPs generate EMG

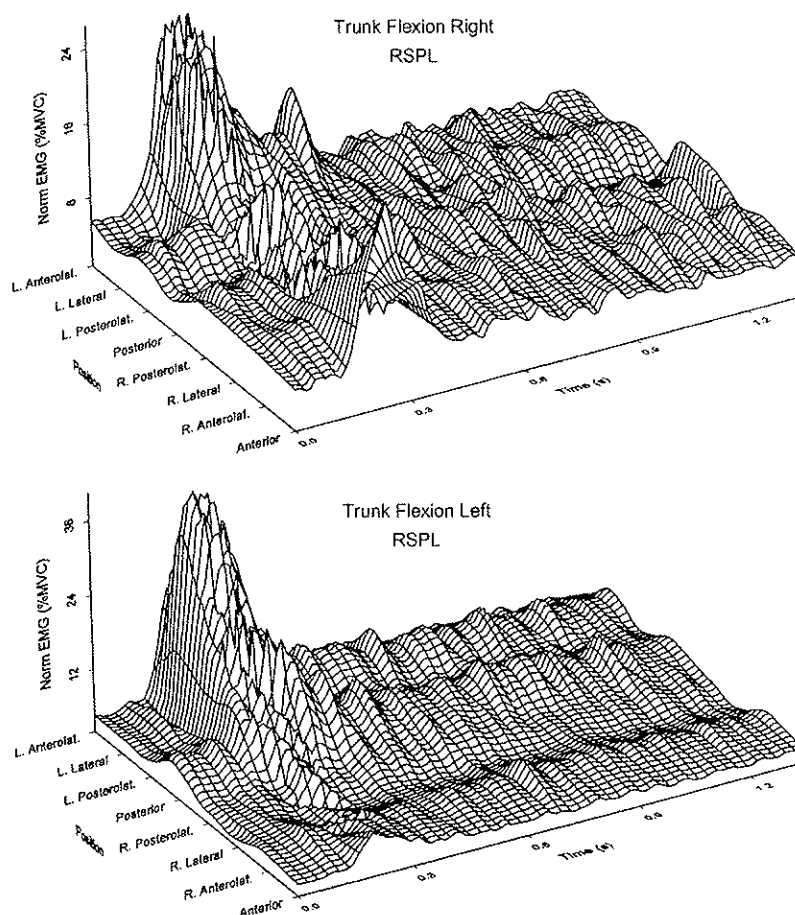


Fig. 12. Muscle activity of the right splenius capitis muscle (RSPL) for 8 directions of impact with trunk flexed either right or left.

activity on the order of 50% or more of their MVC EMG values (Kumar et al., 2003). When the trunk is in neutral posture and the head is rotated in a frontal impact, the TRP ipsilateral to the direction of head rotation generates a response on the order of 75% of its MVC EMG (Kumar et al., 2004a). As seen in this series of experiments with trunk flexion, even the most active muscles do not exceed 39% of their MVC EMG. The SCMs, by their attachment and action, are least likely to undergo eccentric contraction in the presence of what we expect is much less head-torso lag in the trunk-flexed posture. In contrast, the attachment and action of the trapezii, cervical extension being one action, are likely in a “pre-stretched” position in the trunk flexed posture with the subject looking downward. This posture is thus expected to generate more response and a higher likelihood of eccentric contraction in the TRPs than the SCMs.

It is suggested that the forward flexed trunk posture, with the subject looking down, does not increase the likelihood of cervical muscle injury as compared to impacts with the trunk in neutral position, at least not for low-velocity impacts, and assuming that EMG activ-

ity is a measure of that injury risk. Certainly, with no direct measure of injury, this surrogate measure may not be entirely valid, but nothing about the EMG and kinematics in response to impacts while the occupant is “out-of-position” points to a greater injury risk. Even when we extrapolate to twice the impact velocity and thus into the low-velocity range, the EMG responses are well within physiological tolerances of maximal voluntary contraction. This is contrary to previous research findings (Foret-Bruno et al., 1990). Previous research, however, focused on dummy responses, which may explain the difference in our findings, and also some of the dummy experiments were much higher velocity impacts. With low-velocity impacts, one does not expect any significant rebounding of the subject back into the seat, and from our extrapolations, a trunk-flexed posture, assuming no bodily impact otherwise, does not otherwise appear to increase the risk of cervical muscle injury compared to occupant positioning in the neutral posture.

Nevertheless, symptoms are reported even after low-velocity impacts, and these lead to as many as 60% of injury claims (Castro et al., 1997). Considerably more

research will be needed to determine how, low-velocity impacts become associated with symptoms. It remains to be seen how this data will ultimately be used in the prevention and treatment of whiplash injury, but the approach provides numerous avenues of research now, because we can introduce many variables systematically into the whiplash experiment. It also offers clinicians, particularly those involved in medicolegal assessments, an evidence base from which to draw some more meaningful conclusions. Certainly, road collisions are more complex than the impacts described here, but additional parameters can be readily modulated with this approach, and will continue to answer questions on the nature of the human response to low-velocity impacts, clarifying both facts and myths.

Conflict of interest statement

The authors declare that there are no financial and personal relationships with other people or organisations that could inappropriately influence (bias) their work. There was no funding source for the writing of this article.

Supplementary data

Supplementary data associated with this article can be found, in the online version, at doi:10.1016/j.clinbiomech.2005.02.007.

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