Looking Away From Whiplash: Effect of Head Rotation in Rear Impacts

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Study Design. Twenty healthy volunteers in a laboratory setting were subjected to rear-end impacts 4.4, 7.9, 10.9, and 13.1 m/s² acceleration, with head rotation to the right and left.

Objective. The purpose of this study was to determine the response of the cervical muscles to increasing low-velocity rear impacts when the head is rotated at the time of impact.

Summary of Background Data. A previous study of rear impacts with head in neutral posture suggests that the burden of impact is borne primarily by the sternocleidomastoid muscles bilaterally. To improve automobile designs to prevent whiplash injury, we need to understand the response of the cervical muscles to whiplash-type perturbations in less-than-ideal conditions, such as when the head is rotated to the right and left at the time of rear-end impact.

Methods. Triaxial accelerometers recorded the acceleration of the sled, torso at the shoulder level, and head of the participant, while bilateral electromyograms of the sternocleidomastoids, trapezii, and splenii capitis were also recorded.

Results. For participants having a rear-end impact, whether having the head rotated to the left or right at the time of impact, the muscle responses increased with increasing levels of acceleration (P < 0.01). The time to onset and time to peak electromyogram for all muscles progressively decreased with increasing levels of acceleration (P < 0.001). Which muscle responded most to a whiplash-type neck perturbation was determined by the direction of head rotation. With the head rotated to the left, the right sternocleidomastoid generated 88% of the maximal voluntary contraction electromyogram (at least triple the response of other muscles). In comparison, the left sternocleidomastoid, both trapezi, and the splenii capitis generated on average only 10% to 30% of the maximal voluntary contraction electromyogram with head rotated to the left. On the other hand, with the head rotated to the right, the left sternocleidomastoid generated 94% of the maximal voluntary contraction electromyogram (again, at least triple the response of other muscles).

Conclusions. If the head is rotated out of neutral posture at the time of rear impact, the injury risk tends to be greater for the sternocleidomastoid muscle contralateral to the side of rotation. Measures to prevent whiplash injury may have to account for the asymmetric response because many victims of whiplash are expected to be looking to the left or right at the time of collision.

Key words: cervical muscles, electromyography, acceleration, motor vehicle collisions, rear impacts, whiplash injury. Spine 2005;30:760–768

In 1969, Ruedmann1 reported on the introduction of a new device in automobile design—the head restraint. This product was introduced to prevent whiplash injury in rear-end collisions. Whiplash injury remains an important health problem with significant economic and health burden.2 Researchers have long endeavored to understand the mechanism of whiplash injury. Since the early collision experiments with volunteers in the 1950s by Severy et al,3 there has been considerable work reported on the subject response to whiplash-type perturbations.4 This body of research has been primarily motivated by a desire to prevent or lessen the extent of whiplash injury, and is, for example, the basis for the introduction in 1969 of the head restraint.

Based on additional research with volunteers,5–22 it is suggested that for rear-end impacts, we can also understand the injury mechanism to be a whip’s lash, in the very sense that Crowe23 first postulated when he coined the term “whiplash” in 1928. As a result of a rear-end impact, the first movement of the body following collision is a forward movement of the torso, which places the lower cervical segments in extension and the upper cervical segments in flexion due to the inertia. As viewed from the lateral aspect, this produces an elongated S-shaped curve to the cervical spine rather than simply the lordotic C-curve that one would expect with neck extension alone.22,24 The entire cervical spine does not go into extension. It then begins to return to the neutral position, after which there may be flexion of the upper cervical spine before it again returns to neutral. During the initial translation of the head backwards to form this S-shaped curve, the muscles of the neck are being stretched, and a reflex contraction occurs during this stretch, again beginning at around 180 ms after impact. This is an eccentric contraction (i.e., contraction occurring during stretching) and is a possible injury mechanism. Alternatively, one may argue that the S-shaped curve itself, or the speed of these movements, produce an injury to ligaments and joints, or even that both may occur. Based on these data,
more innovative head restraints have been developed and are being evaluated.\textsuperscript{25} Seat design may also be relevant.\textsuperscript{26}

However, it is not known how well the aforementioned research applies to the injury mechanism when the head is rotated at the time of impact. The aforementioned studies considered the problem of whiplash injury with head in the neutral posture. The reality is that vehicle occupants are often not positioned in the neutral posture at the time of impact. Understanding the injury mechanism when the head is rotated may, thus, be a necessary step in the eventual development of the most effective prevention strategies for whiplash injury.

The advancement of our understanding of what happens to individuals in motor vehicle collisions, however, has been partly hampered by 2 main problems: (1) experiments with animals, dummies, and cadavers cannot provide the same value of data from volunteer experiments, while (2) there are ethical concerns with subjecting volunteers to injurious neck perturbations. To overcome this ethical dilemma and yet conduct investigations to elucidate the kinematics and electromyographic response to neck perturbation in volunteers, we have shown that surface electromyography (EMG) can be combined with the use of regression techniques modeled on very-low velocity collisions.\textsuperscript{27–29} Using this approach, we have reported on rear impact studies with the head in the neutral position, and the regression models are in agreement with the available data that have been gathered in previous, small studies of higher velocity collisions.\textsuperscript{3} We have also shown that if the subject is expecting an impact, this mitigates the risk of injury.

To address a void in current knowledge, we undertook a study to assess the cervical muscle response for rear-end impacts in which the head is rotated to the right and left. The study presented here is one section of a larger study in which multiple directions were investigated by the random assignment of volunteers to different conditions to minimize risk of injury, if any, due to multiple exposures.

\section*{Methods}

\textbf{Sample}. Twenty healthy, normal subjects (10 males and 10 females) with no history of whiplash injury and no cervical spine pain during the preceding 12 months volunteered for the study. The 20 subjects had a mean age of 23.6 ± 3.0 years, a mean height of 172 ± 7.7 cm, and a mean weight of 69 ± 13.9 kg. The subjects were all right-hand dominant. The study was approved by the Health Research Ethics Board.

\textbf{Tasks}. Small active surface electrodes with 10 times on-site amplification were placed on the belly of the sternocleidomastoïds, upper trapezius at C4 level, and splenius capitis in the triangle between sternocleidomastoids and trapezi bilateral. The fully isolated amplifier had additional gain settings up to 10,000 times, with frequency response DC- 5 kHz and common mode rejection ratio of 92 dB. Before calibrated sled acceleration, the cervical strength of the volunteers was measured to develop force-EMG calibration factor.\textsuperscript{30,31} The seated and stabilized subjects exerted their maximum isometric effort in attempted flexion, extension, and lateral flexion to the left and the right for force-EMG calibration, as described by Kumar \textit{et al.}\textsuperscript{30} The acceleration device consisted of an acceleration platform and a sled (Figure 1). The full details of the device and the EMG data collection are given by Kumar \textit{et al.}\textsuperscript{27}

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 the neutral spinal posture. Subjects were then outfitted with triaxial accelerometers (Model \# CXL04M3, Crossbow Technology, Inc., San Jose, CA) on their glabella and the first thoracic spinous process. Another triaxial accelerometer was mounted on the sled. The accelerometers had a full scale nonlinearity of 0.2\%, dynamic range of ± 5 g, with a sensitivity of 500 mV/g, resolution of 5 mg within a bandwidth of DC-100 Hz, and a silicon micromachined capacitive beam that was quite rugged and extremely small in die area. Subjects were then exposed to rear-impacts with their head rotated 45° to their left and right at accelerations of 4.4, 7.9, 10.9, and 13.1 m/s\textsuperscript{2} generated in a random order by a pneumatic piston. The accelerations involved in this experiment are low enough that injury is not expected, but the acceleration impulse is delivered in a way that mimics the time course seen in motor vehicle collisions and occurs fast enough to produce eccentric muscle contractions.

\textbf{Data Analysis}. The data on the peak and average accelerations in all 3 axes of the sled, shoulder, and head for all 4 levels of accelerative impacts were measured. In the analysis, the sample of volunteers was collapsed across gender because preliminary analysis showed no statistically significant differences in the EMG amplitudes between the men and women. The sled velocity and its acceleration subsequent to the pneumatic piston impact and the rubber stopper impact were measured. All timing data were referred to the solenoid firing. The time of the peak acceleration was measured. Also, the time relations of the onset and peak of the EMG were measured and analyzed. The time to onset was determined when the EMG perturbation reached 2% of the peak EMG value to avoid false-positives due to tonic activity. This method was in agreement with projection of the line of slope on the baseline. EMG amplitudes were normalized against the subjects’ maximal voluntary contraction EMG. The ratio percentage of the EMG amplitude \textit{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.

Statistical analysis was performed using the SPSS statistical package (SPSS Inc., Chicago, IL) to calculate descriptive statis-
tics, correlation analysis between EMG and head acceleration, analysis of variance of the time to EMG onset, time to peak EMG, average EMG, and the force equivalents. Additionally, a linear regression analysis was performed for the kinematic variables of head displacement, head velocity, and head acceleration and EMG variables. Initially, all regressions were performed to the level of exposure, and, subsequently, they were extrapolated to twice the level of acceleration used in the study.

■ Results

Head Acceleration

The kinematic response of the head to the 4 levels of applied acceleration are shown in Figure 2. As anticipated, an increase in applied acceleration resulted in an increase in excursion of the head and accompanying accelerations.

EMG Amplitude

In a rear impact with the head rotated 45° to the right or left, the contralateral sternocleidomastoid muscle shows the greatest EMG response compared to the ipsilateral sternocleidomastoid, the splenii capitis, and the trapezius bilaterally ($P < 0.01$), suggesting that the contralateral sternocleidomastoid carries the higher burden of the effects of the head and neck perturbation. The mean peak (normalized) EMG amplitude of the cervical muscles tested in this experiment at each applied acceleration

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Figure 2. Head rotated to the right and left. Head acceleration in the x, y, and z axes of one subject in response to the level of 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.
level are presented in Figure 3. As the level of applied acceleration in the rear impact increased, the magnitude of the EMG recorded from the sternocleidomastoid contralateral to the head rotation increased progressively and disproportionately compared to other muscles ($P < 0.01$).

The normalized EMG showed that the percentage of sternocleidomastoid, splenius capitis, and trapezius magnitude increased steadily with the increasing magnitude of the impact (Figure 3). In a rear impact, with head rotated to the right, for example, the left sternocleidomastoid exerted 94% and the right sternocleidomastoid 21% of mean normalized maximal voluntary contraction, while the splenius capitis and trapezius bilaterally generated 30% or less of mean normalized maximal voluntary contraction. The reverse occurred when the head was rotated to the left, in which the right sternocleidomastoid instead generated 88% of the maximal voluntary contraction, and the remaining muscles generated 30% or less of mean normalized maximal voluntary contraction. In terms of force equivalents, the sternocleidomastoid contralateral to the side of impact was required to resist the impact at a level near the maximal voluntary contraction capability.

**Timing**

The time to onset of the sled, shoulder, and head acceleration onset in the z-axis (axis along impact direction) and the EMG signals of the 6 muscles examined for head rotated to the left or right are presented in Table 1. The
time to onset was measured from the firing of the solenoid of the pneumatic piston. The time to onset of the head acceleration decreased with increased applied acceleration. Similarly, the time to onset of the EMG decreased with increased applied acceleration. Mean times at which peak EMG occurred for all the experimental conditions are presented in Table 2. The times at which peak EMG occurred showed a trend to being reduced with increasing acceleration, but this did not reach statistical significance.

The relationship between the force equivalent EMG response of each muscle and the head acceleration are shown in Table 3. The kinematic responses show that very-low velocity impacts produce less force equivalent than the maximal voluntary contraction for the same subject, and, thus, this experimental approach allows us to gather valuable data without exposing subjects to any foreseeable injury.

**Statistical Analyses**
The applied acceleration and the muscles examined had significant main effects on the peak EMG activity ($P < 0.01$). We used a linear regression model to plot the available data and extrapolate from the experimental accelerations to accelerations approximately 30 m/s$^2$. Initially, regression analyses were performed only up to 13 m/s$^2$ using a linear function. The kinematic variables of head displacement, velocity, and acceleration in response to applied acceleration were calculated (Figure 4). Additionally, we also regressed the EMG magnitudes on acceleration. The responses of the left and right muscle groups were extrapolated to more than twice the applied acceleration value (Figures 5 and 6). This result likely explains why the contralateral sternocleidomastoid muscle shows the highest response because it has a position and function that would be most likely to carry the greatest burden of this perturbation.

**Discussion**
To develop preventative measures for whiplash injury, like novel head restraints, the mechanism of injury needs to be understood. Volunteer collisions were partly the

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**Table 1. Rear Impact with Head Rotated to the Right and Left. Mean Time to Onset (ms) of Acceleration and of Muscle EMG From the Firing of the Solenoid of the Pneumatic Piston**

<table>
<thead>
<tr>
<th>Impact Acceleration (m/s$^2$)</th>
<th>Muscle</th>
<th>Sled</th>
<th>Shoulder</th>
<th>Head</th>
<th>Sternal</th>
<th>Splenius</th>
<th>Trapezius</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Left</td>
<td>Right</td>
<td>Left</td>
<td>Right</td>
<td>Left</td>
<td>Right</td>
<td>Left</td>
</tr>
<tr>
<td>Head rotated to right</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>4-2</td>
<td>54 (19)</td>
<td>80 (29)</td>
<td>102 (41)</td>
<td>128 (28)</td>
<td>157 (29)</td>
<td>128 (44)</td>
<td>234 (118)</td>
</tr>
<tr>
<td>8-1</td>
<td>32 (13)</td>
<td>50 (14)</td>
<td>82 (29)</td>
<td>101 (30)</td>
<td>124 (22)</td>
<td>96 (26)</td>
<td>215 (100)</td>
</tr>
<tr>
<td>10-4</td>
<td>29 (11)</td>
<td>47 (19)</td>
<td>77 (28)</td>
<td>100 (29)</td>
<td>120 (19)</td>
<td>92 (30)</td>
<td>191 (127)</td>
</tr>
<tr>
<td>13-0</td>
<td>28 (11)</td>
<td>45 (14)</td>
<td>68 (27)</td>
<td>97 (27)</td>
<td>118 (20)</td>
<td>78 (41)</td>
<td>154 (79)</td>
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<td></td>
<td></td>
</tr>
<tr>
<td>4-2</td>
<td>48 (25)</td>
<td>79 (32)</td>
<td>107 (32)</td>
<td>155 (25)</td>
<td>100 (54)</td>
<td>225 (55)</td>
<td>118 (45)</td>
</tr>
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<td>49 (23)</td>
<td>77 (32)</td>
<td>128 (24)</td>
<td>97 (30)</td>
<td>218 (70)</td>
<td>89 (33)</td>
</tr>
<tr>
<td>10-4</td>
<td>30 (10)</td>
<td>48 (13)</td>
<td>73 (22)</td>
<td>124 (23)</td>
<td>95 (33)</td>
<td>206 (72)</td>
<td>84 (36)</td>
</tr>
<tr>
<td>13-0</td>
<td>29 (9)</td>
<td>46 (8)</td>
<td>68 (17)</td>
<td>123 (18)</td>
<td>92 (31)</td>
<td>203 (95)</td>
<td>82 (34)</td>
</tr>
</tbody>
</table>

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 one standard deviation.

**Table 2. Rear Impact with Head Rotated to the Right. Mean Time (ms) at Which Peak EMG Occurred After the Firing of the Solenoid of the Pneumatic Piston**

<table>
<thead>
<tr>
<th>Muscle EMG</th>
<th>Acceleration (m/s$^2$)</th>
<th>Sternal</th>
<th>Splenius</th>
<th>Trapezius</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head rotated to right</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>4-2</td>
<td>226 (17)</td>
<td>240 (13)</td>
<td>322 (231)</td>
<td>567 (339)</td>
</tr>
<tr>
<td>8-1</td>
<td>213 (33)</td>
<td>213 (18)</td>
<td>238 (89)</td>
<td>515 (340)</td>
</tr>
<tr>
<td>10-4</td>
<td>195 (15)</td>
<td>207 (17)</td>
<td>223 (82)</td>
<td>434 (321)</td>
</tr>
<tr>
<td>13-0</td>
<td>192 (17)</td>
<td>192 (15)</td>
<td>223 (61)</td>
<td>425 (263)</td>
</tr>
<tr>
<td>Head rotated to left</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>4-2</td>
<td>230 (24)</td>
<td>229 (30)</td>
<td>451 (304)</td>
<td>247 (63)</td>
</tr>
<tr>
<td>8-1</td>
<td>206 (20)</td>
<td>199 (23)</td>
<td>439 (199)</td>
<td>215 (61)</td>
</tr>
<tr>
<td>10-4</td>
<td>205 (20)</td>
<td>198 (15)</td>
<td>405 (215)</td>
<td>206 (55)</td>
</tr>
<tr>
<td>13-0</td>
<td>201 (16)</td>
<td>195 (18)</td>
<td>390 (263)</td>
<td>202 (18)</td>
</tr>
</tbody>
</table>

Values in parentheses represent one standard deviation.
basis for the introduction of the head restraint in 1969, and further experiments have shown the need for improvements in the head restraint design. In rear impacts, victims of whiplash may be looking to the left or right, either as a result of watching for traffic or speaking with other occupants. This movement affects the neck muscle response to accelerations. In the current study, in which we have used EMG measurements to study the cervical muscle when the head is rotated to the right or left at the time of a rear impact, we find that the sternocleidomastoid contralateral to the direction of head rotation generates the highest EMG and force equivalents, and other muscles on the side contralateral to the direction of head rotation also become more active. Thus, head rotation creates a potential for a contralateral muscle injury. We have previously shown that in a rear impact, with the head in neutral posture, both sternocleidomastoids generate activity higher than the maximal voluntary contraction EMG and much more so than the splenius capit is or trapezius muscles.

The question is whether or not having the head rotated at the time of a rear impact increases the risk of whiplash injury. From the experimental design conditions here, one can state that if an injury is to occur, it is most likely to occur first to the sternocleidomastoid muscle contralateral to the direction of head rotation. Otherwise, the magnitude of sternocleidomastoid muscle EMG activity in the volunteers of this study is less than in a previous study of rear impacts with the head in the neutral position. That is, in a previous study, we found that both sternocleidomastoids respond to a rear impact, with EMG activity that is up to 179% as great as normal maximal static muscle contraction strength when the impact is unexpected and the subject is looking forward.

Table 3. Rear Impact with Head Rotated to the Right and Left. Mean Force Equivalents and Mean Head Accelerations at the Time of Maximal EMG in Direction of Travel

<table>
<thead>
<tr>
<th>Chair Acceleration (m/s²)</th>
<th>Head Acceleration (m/s²)</th>
<th>Sternocleidomastoid</th>
<th>Splenius Capitis</th>
<th>Trapezius</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Left</td>
<td>Right</td>
<td>Left</td>
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<tr>
<td>Head rotated to right</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>4-2</td>
<td>3-4 (0.6)</td>
<td>17 (8)</td>
<td>5 (4)</td>
<td>23 (17)</td>
</tr>
<tr>
<td>8-1</td>
<td>6-3 (1.8)</td>
<td>24 (11)</td>
<td>6 (4)</td>
<td>27 (20)</td>
</tr>
<tr>
<td>10-4</td>
<td>8-7 (2.9)</td>
<td>30 (10)</td>
<td>10 (5)</td>
<td>30 (18)</td>
</tr>
<tr>
<td>13-0</td>
<td>10-0 (2.4)</td>
<td>34 (15)</td>
<td>11 (5)</td>
<td>33 (22)</td>
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<tr>
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</tr>
<tr>
<td>4-2</td>
<td>4-4 (0.6)</td>
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<td>10-4</td>
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<td>11 (7)</td>
<td>25 (12)</td>
<td>24 (10)</td>
</tr>
<tr>
<td>13-0</td>
<td>13-1 (1.5)</td>
<td>13 (9)</td>
<td>37 (20)</td>
<td>27 (14)</td>
</tr>
</tbody>
</table>

Values in parentheses represent one standard deviation.

Figure 4. The kinematic variable of head displacement, velocity, and acceleration in response to applied acceleration.
When the impact is expected, this EMG activity decreases to about 80% of maximal muscle contraction strength. In the current study, the subjects were also expecting the impact, so this may explain why there was less muscle activity observed. Yet, there was less muscle activity for all muscles when the head is rotated than when compared to a rear-impact with the head in the neutral position. This result suggests that it is not simply expecting the impact that reduces the muscle response, but the fact 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. The reader can appreciate that with the head rotated to the right or the left, one is not able to flex or extend the neck as much as when looking straightforward.

Furthermore, head rotation is a volitional activity achieved by exerting the muscles and tensing them, thereby putting them in a state of readiness to resist any
perturbation. Other neck structures prohibit the range of motion when the head is rotated. These structures, like facet joints, can tolerate higher forces than muscles can before injury occurs. Thus, having the head rotated at the time of impact may be protective, just as expecting an impact may be protective against injury. Obviously, with higher velocity impacts, these protective factors may be overcome, resulting in more serious injury. Therefore, more studies are needed to ascertain what additional strategies need to be used for improving head restraint design in the prevention of whiplash injury, considering that the muscle responses vary greatly under different impact conditions.

The experimental design we have used to study neck perturbations to very low-velocity change is, nevertheless, useful for further studies into these matters. There is a wealth of information obtained through this approach without potentiating any volunteer injury. The study is not intended to mimic injury causing impacts but rather to allow for the initial exploration of the role of EMG for assessing neck perturbations. Because it is not yet possible to identify objectively the acute whiplash injury thought to underlie Grades 1 and 2 whiplash associated disorders, current injury models are based on evaluation of volunteers in collisions. For ethical reasons, we are not able to subject deliberately volunteers to higher accelerations that may cause injury. Thus, we rely on extrapolation techniques from noninjurious, very low-velocity impacts. All extrapolations are approximations of the truth and, therefore, represent a weakness of this approach.

Using a very low-velocity experimental design, however, we are able to extrapolate through nonlinear regression to predict the head accelerations and forces likely to be experienced by neck perturbations at higher velocities, and our extrapolations closely match those from small volunteer studies in which higher velocities were used with symptoms produced. This result 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.

With so many parameters available for modulation attempting to approximate road collisions, the task of developing a model for the acute whiplash injury is daunting. However, one starting place is the use of objective measurements, such as EMG, in a laboratory setting where other confounding variables have been accounted for or eliminated. With time, more variables can be introduced and studied with this approach. It is also known that not expecting an impact accentuates the head acceleration and muscle activity more so than when the impact is expected.

In this experiment, the subjects were aware of an impending impact, so future studies will also evaluate the effect of head rotation in the unexpected impact as well.

Key Points

- With head rotation at the time of rear impact, the sternocleidomastoid muscle opposite the direction of head rotation is activated more than the trapezius, splenii capitis, or ipsilateral sternocleidomastoid muscles.
- The sternocleidomastoid contralateral to the direction of head rotation reaches near the maximal voluntary contraction, with an acceleration of 13.1 m/s².
- If the head is rotated out of the neutral posture at the time of rear impact, the injury risk tends to be higher for the sternocleidomastoid muscle contralateral to the side of rotation.

References


