ABSTRACT

Objective: To determine the electromyogram (EMG) response of the cervical muscles to a right lateral impact whiplash-type perturbation when the head is rotated.

Methods: Twenty healthy volunteers were subjected to right lateral impacts of 4.2, 8.1, 10.3, and 12.5 m/s² and were looking either left or right. Bilateral EMGs of the sternocleidomastoid, trapezius, and splenius capitis muscles were recorded. Triaxial accelerometers recorded the acceleration of the chair, torso at the shoulder level, and head of the participant.

Results: In a right lateral impact, muscle responses were of low magnitude with the head rotated to either the left or the right. At the highest acceleration of 12.5 m/s², all generated less than 39% of their maximal voluntary contraction EMG. The sternocleidomastoid muscle showed a greater EMG response than its counterpart and the muscles contralateral to the direction of impact had higher EMG responses. The time to onset of the EMG for the splenii capitis and trapezii generally decreased with increasing levels of acceleration. As anticipated, an increase in applied acceleration resulted in an increase in accompanying head accelerations (P < .05), and when the head acceleration increased, so too did the force equivalent exertions by the various muscles.

Conclusions: Overall, a right lateral impact with head rotation to either right or left appears to reduce the activity and thus the risk of muscle injury, perhaps because of “bracing” by muscles actively producing rotation or because of greater spinal stability from other structures when the head is in the rotated position. (J Manipulative Physiol Ther 2005;28:393-401)

Key Indexing Terms: Neck Muscles; Electromyography; Whiplash Injuries; Motor Vehicle Collision; Lateral Impact

Although it has been traditionally reported that rear impacts account for most cases of whiplash injury,¹ a large epidemiologic study has suggested that rear, lateral, and frontal collisions account for whiplash injury in roughly equal proportions.² One aspect of the relevance of understanding the cervical response to impacts is for the development of protective automobile features. There are no data currently on how side impact airbags, for example, specifically affect the cervical muscle response, although injury may occur. There are also no data on the effect that current seat belt restraints may have on altering the magnitude of the cervical muscles or which muscles are most activated in response to lateral impacts experienced by a seat-belted occupant.

As a first step in addressing these issues, it seems helpful to begin by understanding the cervical response to less complex impact conditions and then gradually introducing additional variables. Although there have been many impact studies,³ there have been little data on individual cervical muscle responses to neck perturbation. There are few studies that specifically reveal the electromyogram (EMG) response of individual muscles (eg, splenii capitis, sternocleidomastoids, and trapezii).⁶⁻¹⁰ Given ethical concerns with subjecting volunteers to injurious neck perturbations, most volunteer experiments have been necessarily limited to low-velocity or very low-velocity collisions, and most of these have been rear impacts.⁴⁻¹⁰

We have shown, however, that valuable information on both the electromyographic response of cervical muscles and the kinematic head response can be obtained by combining surface electromyography with regression techniques modeled on very low-velocity collisions.⁸⁻¹¹ We have been able to show, for example, that when a lateral impact occurs with
the head in neutral posture, the greatest muscle response arises from the contralateral splenius capitis muscle,\textsuperscript{10} with this muscle generating an EMG near its maximal voluntary contraction EMG. This compares to rear impacts\textsuperscript{8} where the greatest muscle response arises from the sternocleidomastoids, or frontal impacts,\textsuperscript{9} where the trapezius are most active in response to the perturbation. Thus, the direction of impact is highly relevant in determining which muscle group shows the greatest EMG response. We have also shown that whether or not one is expecting an impact affects the cervical muscle response (awareness reduces the head acceleration and muscle EMG activity).\textsuperscript{8-10}

Given these data, we are now able to add the variable of head rotation at the time of impact. We have begun a series of experiments to test this, and in this article we report on right lateral impacts with the head in rotation. One would expect that the muscles contralateral to the direction of impact will be most active in response, but it is not clear whether the head rotation will modify this response and to what degree.

\section*{Methods}

The methods for this study of right lateral impacts are similar to that used for our previous impact studies.\textsuperscript{8-11}

\subsection*{Subjects}

Twenty healthy 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 University Research Ethics Board. Subjects were asked to report any symptoms at the time of impacts or in 6 months’ follow-up.

\subsection*{Tasks}

Active surface electrodes with 10 times on-site amplification were placed on the belly of the sternocleidomastoids, upper trapezius at the C4 level, and splenius capitis in the triangle between sternocleidomastoids and trapezius bilaterally. 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 calibrating sled acceleration, the cervical strength of the volunteers was measured to develop a force-EMG calibration factor. The seated and stabilized subjects exerted their maximum isometric effort in attempted flexion, extension, and lateral flexion to the left and then right for force-EMG calibration, as described by previous studies that have obtained this normative data of maximal voluntary muscle contraction.\textsuperscript{12,13}

The seated and stabilized subjects were then exposed to right lateral sled accelerations of 4.2, 8.1, 10.3, and 12.5 m/s\textsuperscript{2} in a random order by a pneumatic piston. Subjects were exposed to right lateral impacts with their head rotated 45° to their left and then right at each acceleration generated in a random order by a pneumatic piston. The study presented here is one section of a larger study where multiple directions were investigated by random assignment of volunteers to different conditions to minimize risk of injury, if any, due to multiple exposures. To keep other variables constant, all participants in this study were aware of an impending impact to the extent that we did not block any visual or auditory cues.

\subsection*{Experimental Setup}

The acceleration device consisted of an acceleration platform and a sled, designed to reproduce the methodology of previous lateral impact studies.\textsuperscript{14} The acceleration platform had parallel tracks, 2 × 200 cm long, mounted lengthwise 60 cm apart. These tracks permitted smooth gliding of the sled on the rails, with a low coefficient of friction (0.03). At one end of the platform, a pneumatic cylinder with a piston stroke length of 30 cm was connected to an air supply and mounted rigidly on the acceleration platform. The device was calibrated for the delivery of known forces causing acceleration of 4.2, 8.1, 10.3, and 12.5 m/s\textsuperscript{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.

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 with a lap seat belt only so they could then be positioned out of neutral posture. Subjects were then

\begin{table}
\centering
\caption{Mean normalized peak EMG of cervical muscles in response to right lateral impacts}
\begin{tabular}{|llllllll|}
\hline
\textbf{Acceleration} & \textbf{Sternocleidomastoid} & \textbf{Splenius capitis} & \textbf{Trapezius} \\
\textbf{(m/s\textsuperscript{2})} & \textbf{(% MVC)} & \textbf{(% MVC)} & \textbf{(% MVC)} \\
\hline
\textbf{Head rotated to the right} & & & & & & & \\
4.2 & 19 (14) & 3 (2) & 9 (5) & 7 (3) & 12 (7) & 20 (14) \\
8.1 & 27 (17) & 3 (2) & 14 (9) & 9 (5) & 13 (7) & 28 (15) \\
10.3 & 30 (20) & 5 (5) & 17 (13) & 9 (7) & 16 (9) & 34 (19) \\
12.5 & 31 (24) & 5 (3) & 20 (17) & 14 (10) & 21 (11) & 39 (27) \\
\hline
\textbf{Head rotated to the left} & & & & & & & \\
4.2 & 3 (2) & 19 (11) & 9 (8) & 4 (2) & 19 (12) & 9 (6) \\
8.1 & 6 (6) & 26 (20) & 17 (12) & 6 (4) & 18 (14) & 11 (7) \\
10.3 & 9 (6) & 29 (17) & 18 (13) & 7 (5) & 24 (12) & 11 (6) \\
12.5 & 11 (7) & 31 (14) & 25 (13) & 8 (6) & 26 (17) & 12 (7) \\
\hline
\end{tabular}
\end{table}

Values in parentheses represent 1 SD. MVC, Maximal voluntary contraction.
outfitted with triaxial accelerometers (model no. CXL04M3, Crossbow technology, Inc, San Jose, Calif) 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.

**Data Acquisition**

The data acquisition system consisted of an analog-to-digital board with a 100-kHz sampling capacity. The

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*Fig 1. Means of peak electromyographic activity (μV) for conditions of head rotated right and left, and 4 levels of applied acceleration. lscm, Left sternocleidomastoid; rscm, right sternocleidomastoid; lspl, left splenius capitis; rslpl, right splenius capitis; ltrp, left trapezius; rtrp, right trapezius.*
9 acceleration channels, 6 EMG channels, and the force channel were sampled at 1 kHz in real time. The sampled signals were stored on a computer with a large hard disc for storage and processing. The EMG and acceleration data were collected during the experimental trials. The peak and average EMG and acceleration values obtained from these sets of data were subjected to quantitative and statistical analysis. The time to onset was determined.

Fig 2. Normalized average and peak EMG (percentage of isometric maximal voluntary contraction), force equivalent of EMG (N), for right and left head rotation at 4 levels of applied acceleration.
when the EMG pertubation reached 2% of the peak EMG value. This method was chosen to avoid any false positives due to tonic EMG. In addition, this method was in agreement with projection of the line of slope on the baseline. Electromyogram amplitudes were normalized against the subjects’ maximal voluntary contraction EMG recorded before accelerative impacts. 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.

### Test Protocol

After the volunteers were seated and the accelerometers were affixed to the axes, the 3 accelerometers were aligned with the path of the chair. The pneumatic cylinder was aligned such that the piston head of the cylinder and the baseboard of the right side of the sled were in contact. The pneumatic piston delivered the appropriate acceleration to the sled. The data collection was initiated, and after 1 second the pneumatic piston was fired to accelerate the sled. The data analysis was initiated, and after 1 second the pneumatic piston was fired to accelerate the sled.

### Data Analysis

In the analysis, the sample of volunteers was collapsed across sexes because preliminary analysis showed no statistically significant differences in the peak EMG amplitudes between the men and women. 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

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

<table>
<thead>
<tr>
<th>Acceleration (m/s²)</th>
<th>Muscle</th>
<th>Sled</th>
<th>Shoulder</th>
<th>Head</th>
<th>Left</th>
<th>Right</th>
<th>Left</th>
<th>Right</th>
<th>Left</th>
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<td><strong>Head Right</strong></td>
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</tr>
<tr>
<td>4.2</td>
<td>53 (27)</td>
<td>67 (27)</td>
<td>81 (43)</td>
<td>169 (112)</td>
<td>281 (203)</td>
<td>128 (38)</td>
<td>197 (51)</td>
<td>144 (43)</td>
<td>139 (57)</td>
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<tr>
<td>8.1</td>
<td>34 (14)</td>
<td>48 (13)</td>
<td>73 (23)</td>
<td>127 (70)</td>
<td>224 (142)</td>
<td>107 (33)</td>
<td>150 (76)</td>
<td>108 (32)</td>
<td>134 (46)</td>
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<td>10.3</td>
<td>27 (11)</td>
<td>43 (14)</td>
<td>59 (29)</td>
<td>120 (48)</td>
<td>221 (147)</td>
<td>106 (23)</td>
<td>132 (60)</td>
<td>94 (25)</td>
<td>122 (43)</td>
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<td>12.5</td>
<td>27 (9)</td>
<td>40 (16)</td>
<td>53 (25)</td>
<td>114 (51)</td>
<td>219 (179)</td>
<td>95 (34)</td>
<td>127 (56)</td>
<td>82 (34)</td>
<td>112 (48)</td>
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<td><strong>Head Left</strong></td>
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<tr>
<td>4.2</td>
<td>49 (23)</td>
<td>63 (28)</td>
<td>93 (39)</td>
<td>122 (49)</td>
<td>167 (112)</td>
<td>108 (63)</td>
<td>264 (130)</td>
<td>140 (68)</td>
<td>228 (102)</td>
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<tr>
<td>8.1</td>
<td>34 (16)</td>
<td>46 (24)</td>
<td>78 (25)</td>
<td>113 (24)</td>
<td>151 (112)</td>
<td>107 (44)</td>
<td>231 (131)</td>
<td>137 (62)</td>
<td>221 (91)</td>
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<tr>
<td>10.3</td>
<td>30 (13)</td>
<td>40 (22)</td>
<td>66 (25)</td>
<td>112 (38)</td>
<td>130 (88)</td>
<td>104 (45)</td>
<td>223 (202)</td>
<td>126 (41)</td>
<td>159 (104)</td>
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<tr>
<td>12.5</td>
<td>26 (12)</td>
<td>37 (12)</td>
<td>59 (25)</td>
<td>110 (28)</td>
<td>125 (79)</td>
<td>99 (46)</td>
<td>204 (215)</td>
<td>125 (63)</td>
<td>151 (94)</td>
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</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 time to onset of EMG activity. Values in parentheses represent 1 SD.

### Table 3. Mean time (milliseconds) at which peak EMG occurred after the firing of the solenoid of the pneumatic piston

<table>
<thead>
<tr>
<th>Acceleration (m/s²)</th>
<th>Muscle EMG</th>
<th>Sled</th>
<th>Shoulder</th>
<th>Head</th>
<th>Left</th>
<th>Right</th>
<th>Left</th>
<th>Right</th>
<th>Left</th>
<th>Right</th>
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<tbody>
<tr>
<td><strong>Head Right</strong></td>
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<tr>
<td>4.2</td>
<td>306 (95)</td>
<td>742 (516)</td>
<td>257 (61)</td>
<td>535 (436)</td>
<td>282 (110)</td>
<td>278 (40)</td>
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<tr>
<td>8.1</td>
<td>265 (84)</td>
<td>698 (508)</td>
<td>247 (55)</td>
<td>442 (364)</td>
<td>262 (73)</td>
<td>250 (55)</td>
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<tr>
<td>10.3</td>
<td>263 (72)</td>
<td>617 (438)</td>
<td>243 (37)</td>
<td>424 (233)</td>
<td>257 (47)</td>
<td>248 (44)</td>
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<tr>
<td>12.5</td>
<td>227 (46)</td>
<td>564 (355)</td>
<td>231 (50)</td>
<td>397 (224)</td>
<td>246 (42)</td>
<td>240 (44)</td>
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<tr>
<td><strong>Head Left</strong></td>
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<tr>
<td>4.2</td>
<td>246 (55)</td>
<td>564 (322)</td>
<td>346 (219)</td>
<td>678 (416)</td>
<td>632 (517)</td>
<td>833 (336)</td>
<td></td>
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</tr>
<tr>
<td>8.1</td>
<td>243 (68)</td>
<td>455 (253)</td>
<td>310 (137)</td>
<td>620 (317)</td>
<td>468 (315)</td>
<td>537 (315)</td>
<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>10.3</td>
<td>236 (31)</td>
<td>409 (249)</td>
<td>302 (167)</td>
<td>593 (275)</td>
<td>400 (233)</td>
<td>512 (293)</td>
<td></td>
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<td></td>
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<tr>
<td>12.5</td>
<td>234 (32)</td>
<td>314 (179)</td>
<td>279 (109)</td>
<td>577 (266)</td>
<td>376 (205)</td>
<td>496 (269)</td>
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</tbody>
</table>

Values in parentheses represent 1 SD.
solenoid was also measured, as were 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 and for both directions of head rotation (right and left). 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. In addition, the time relations of the onset and peak of the EMG in relation to the solenoid firing were measured and analyzed.

Electromyogram amplitudes were normalized against the subjects’ maximal voluntary contraction EMG (voluntary strength data reported in Refs 12,13). 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. Finally, a statistical analysis was performed using the SPSS statistical package (SPSS Inc, Chicago, Ill) to calculate descriptive statistics, t test, \( \chi^2 \) with Yates correction, correlation analysis between EMG and applied acceleration, analysis of variance of the EMG slope, time to EMG onset, time to peak EMG, time to average EMG, and the force equivalents. Correction was made for multiple comparisons in determining significance.

RESULTS
Electromyogram Amplitude

The mean peak (normalized) EMG amplitude of the cervical muscles for the impacts with head rotated right and left at each applied acceleration level are presented in Table 1. In a right lateral impact, with the head rotated 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 maximal voluntary contraction EMG. For the impacts where the head was rotated to the right, the left sternocleidomastoid muscle (the one primarily responsible for head rotation to the right) had a much higher activity than its counterpart (\( P < .05 \)). With the head rotated to the left (looking in the impact direction), again all muscles responded to the impact with relatively low EMG activity, 26% or less of the subject’s maximal voluntary contraction. The sternocleidomastoid muscles reversed their pattern from above, the right sternocleidomastoid (being primarily responsible for head rotation to the left) was the more active when the head was rotated to the left (\( P < .05 \)). The left splenius capitis (the one contralateral to the direction of impact) had the higher EMG response as compared with the right splenius capitis when the head was rotated to the left, but not when the head was rotated to the right.

Fig 1 illustrates the EMG recorded under these conditions. As the level of applied acceleration in a right lateral impact increased, for both directions of head rotation, the magnitude of the EMG recorded increased progressively and significantly for most muscles (\( P < .05 \)).

The normalized EMG results reveal that muscle activity never exceeded 39% of the maximal voluntary contraction (Table 1 and Fig 2). In terms of force equivalents, the burden of impact was widely distributed over the measured muscles, except for the sternocleidomastoid not responsible for head rotation, which had the lowest force equivalents.

Electromyogram Timing

The time to onset of the sled, shoulder, and head acceleration onset in the z-axis (axis along lateral impact direction) and the EMG signals of the 6 muscles examined...
are presented in Table 2. The time to the sled, torso, and head acceleration onset decreased with increased applied acceleration. Similarly, the time to EMG onset and peak EMG showed a trend toward decreasing with increased applied acceleration, not reaching statistical significance (ie, $P > .05$). The mean times at which peak EMG occurred for all the experimental conditions are presented in Table 3.

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

<table>
<thead>
<tr>
<th>Chair acceleration (m/s²)</th>
<th>Head acceleration (m/s²)</th>
<th>Force equivalents for muscle (N)</th>
<th>Sternocleidomastoid</th>
<th>Splenius capitis</th>
<th>Trapezius</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Left (N)</td>
<td>Right (N)</td>
<td>Left (N)</td>
<td>Right (N)</td>
</tr>
<tr>
<td>Head right</td>
<td></td>
<td>10 (9)</td>
<td>6 (2)</td>
<td>20 (7)</td>
<td>19 (8)</td>
</tr>
<tr>
<td>4.2</td>
<td>2.7 (0.5)</td>
<td>8.1</td>
<td>4.7 (0.7)</td>
<td>13 (9)</td>
<td>6 (4)</td>
</tr>
<tr>
<td>10.3</td>
<td>6.1 (0.8)</td>
<td>10.5</td>
<td>7.5 (1.4)</td>
<td>14 (9)</td>
<td>7 (5)</td>
</tr>
<tr>
<td>12.5</td>
<td>7.5 (1.4)</td>
<td>12.5</td>
<td>7.5 (1.4)</td>
<td>16 (8)</td>
<td>7 (7)</td>
</tr>
<tr>
<td>Head left</td>
<td></td>
<td>4.2</td>
<td>2.6 (0.6)</td>
<td>6 (4)</td>
<td>11 (6)</td>
</tr>
<tr>
<td>8.1</td>
<td>5.2 (1.7)</td>
<td>8.1</td>
<td>5.2 (1.7)</td>
<td>7 (5)</td>
<td>12 (7)</td>
</tr>
<tr>
<td>10.3</td>
<td>6.9 (1.3)</td>
<td>10.3</td>
<td>6.9 (1.3)</td>
<td>8 (6)</td>
<td>15 (7)</td>
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<tr>
<td>12.5</td>
<td>7.8 (2.2)</td>
<td>12.5</td>
<td>7.8 (2.2)</td>
<td>9 (5)</td>
<td>15 (10)</td>
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</tbody>
</table>

Values in parentheses represent 1 SD.

Fig 4. Extrapolated regression plots of the effect that applied acceleration has on the head motion variables of displacement (A) (mm), velocity (B) (m/s), and acceleration (C) obtained (m/s²).
Head Acceleration

The kinematic response of the head to the 4 levels of applied acceleration with head rotated to the right and left is shown in Fig 3. As anticipated, an increase in applied acceleration resulted in an increase in accompanying head accelerations \((P < .05)\). The relationship between the force equivalent EMG response of each muscle and the head acceleration is shown in Table 4. As this table shows, when the head acceleration increased, so too did the force equivalent exertions by the various muscles \((P < .05)\).

Statistical Analyses

With greater magnitude of impact, there was a trend toward greater EMG response and each muscle responded differently. Initially, regression analyses were performed only up to 12.5 m/s² using linear, quadratic, cubic, power, and exponential functions. The power function, however, was chosen as it fit the data best. The other functions also produced 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. Fig 4 provides extrapolated regression plots of the effect that applied acceleration has on the head motion variables of displacement, velocity, and acceleration. These data further reveal that the head displacement, velocity, and acceleration were greatest in the direction of the impact.

Discussion

Motor vehicle collisions are complex. Besides direction of impact, whether or not one is aware if the impact coming, the restraint systems, body positioning, anthropometrics, and head position might affect the cervical muscle response. To begin examining more complex scenarios mimicking real collisions, we have analyzed a group of volunteers subjected to a right lateral impact but have in this study added the variable of head rotation to the left or right at the time of impact.

Whiplash patients frequently remark on their belief that there is a significance to their head position at the time of impact. Some believe that the fact that they “saw it coming” because they were looking to their right at the time of a right lateral impact, for example, protected them from more serious injury. Others believe that having their neck “twisted” at the time of impact is the reason they have unilateral neck pain. Both may be correct, although more experimental and epidemiologic research is needed to determine this.

In the current study, designed to determine what happens when the head is rotated to the right or left at the time of a right lateral impact, the muscles all generated on average 39% or less of their maximal voluntary contraction EMG, with the muscles contralateral to the direction of impact still generally showing greater EMG activity. Thus, whiplash patients may be correct in their assertion of unilaterality of neck pain as a consequence of impact direction and/or head position, but we also see from this study that the magnitude of muscle EMG activity is less with head rotation than in a previous study of right lateral impacts with the head in the neutral position.\(^{14}\) That is, we have previously shown that in a right lateral impact, at an acceleration of 13.7 m/s², and the head in the neutral position, the contralateral (right) splenius capitis generated up to 84% of its maximal voluntary contraction EMG, the highest activity observed for any of the muscles.\(^{14}\) This occurs, however, only if the impact is unexpected. When the impact is expected, all muscles generate a lesser EMG response, about half that in the unexpected impact condition. In the current study, the subjects were also expecting the impact, so this may explain why there was less muscle activity observed. The other possibility is that when the head is rotated there is less overall capacity for head displacement, so less stretch of muscles in response to impact, and less muscle activation overall. Other neck structures prevent the full range of motion when the head is rotated. These structures, like facet joints, can tolerate significantly higher forces than muscles can before injury occurs. In addition, head rotation requires muscle activity in cervical muscles providing additional stability. 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. More studies are needed to consider this.

We believe, nevertheless, that this is best achieved by using very low-velocity experimental design, and as none of our subjects reported symptoms of concern, it is hoped that we will be able to increase the acceleration magnitude in future studies. With this approach, we are able to extrapolate through linear regression to predict the head accelerations and forces likely to be experienced by neck perturbations at higher velocities. Our extrapolations seem to match those from small volunteer studies where higher velocities were used with symptoms produced.\(^3\) Regression techniques, although they provide approximations only, thus avoid the need for exceeding ethical concerns with experimental designs that could cause volunteer injury. This, combined with the fact that EMG studies also allow one to examine muscle group responses and patterns, rather than simply describe head or other body region accelerations, makes this approach useful in terms of the broad scope of data. With so many parameters available for modulation in attempting to approximate road collisions, the use of objective measurements such as EMG in a laboratory setting where other confounding variables have been accounted for or eliminated is viable. In time, more variables can be introduced and studied with this approach, and the current study is but one of a series of attempts to discern the effects of a number...
of impact variables. Future studies will examine the effects of different seats and restraint systems on the cervical muscle response.

REFERENCES


