Analysis of low velocity frontal impacts

Shrawan Kumar *, Yogesh Narayan, Tyler Amell

Ergonomics Research Laboratory, Department of Physical Therapy, University of Alberta, 3-75 Corbett Hall, Edmonton, Alta., Canada T6G 2G4

Received 29 November 2002; accepted 11 June 2003

Abstract

Objectives. The objectives of the study were to determine the phasic recruitment of cervical muscles with increasing magnitudes of low velocity frontal impacts, and to determine quantitative effects of awareness of impending impact in comparison to being impacted unawares.

Background. Biomechanics of low velocity frontal impact is poorly understood and requires more work.

Methods. Ten healthy young adults were subjected to frontal impacts causing accelerations of 5.3, 8.6, 11.0 and 14.0 m/s² (0.54, 0.88, 1.12 and 1.43 gs) while the subjects were unaware of the impending impact and after being told that they were going to be impacted. Electromyograph from sternocleidomastoids, splenius capitis and upper trapezius was recorded bilaterally. Triaxial accelerometers recorded the acceleration of the sled, torso and head of the subjects.

Results. The normalized electromyograph magnitude progressively rose with the level of acceleration whereas the time to onset generally decreased. At 14 m/s² sled acceleration the trapezius muscle generated 79% of their maximal voluntary contraction whereas the sternocleidomastoids generated 32% of their maximum voluntary contraction. The normalized peak electromyograph, the time to onset, the time of the peak electromyograph were significantly affected by the level of acceleration ($P < 0.01$), the expectation of impact ($P < 0.01$) and muscle group studied ($P < 0.01$). The subject gender did not have a significant effect. The kinematic variables and the electromyograph regressed significantly on acceleration ($P < 0.01$).

Conclusions. The muscle responses were greater with higher levels of acceleration, particularly the trapezius in frontal impacts. Since the muscular components play a significant and central role in head/neck complex motion abatement at higher levels of acceleration, it may be a primary site of injury at low velocity whiplash phenomenon.

Relevance

An understanding of the pattern of biomechanical loading may assist in a more specific treatment of the patient injured in a low velocity frontal impact.

© 2003 Elsevier Ltd. All rights reserved.

Keywords: Whiplash; Cervical muscles; Electromyography; Frontal impacts; Motor vehicle accidents

1. Introduction

Cervical injuries have become a major problem in our society. In Canada, United States, France, Italy, Japan, the Netherlands, the Republic of Ireland, Scandinavian countries and Switzerland there are huge numbers of whiplash patients reporting chronic pain (Ferrari, 1999). In Germany, Greece, Lithuania, New Zealand, and Singapore, however, even though whiplash patients report acute pain they seem to get better within weeks (Ferrari, 1999). Regardless of the cultural and social differences the cervical injuries are commonly reported in many parts of the world.

Approximately 65% of whiplash injuries are reported after low velocity impacts (Castro et al., 1997; Insurance Corporation of British, 1999). Although neck injury is mostly regarded as resulting from rear and collisions, almost one third of all neck injuries occur in frontal impacts (Kullgren et al., 2000). In a retrospective analysis of accidents in Sweden of 187 restrained front seat occupants in 143 frontal collisions it was reported that the crash pulse was an important variable influencing the risk of long-term disability of the neck (Kullgren et al., 2000). There are several studies, which have looked at the frontal collisions at high impact velocity (Magnusson et al., 1999;
McKinney, 1989; Svensson et al., 1993; Brorsson, 1989; Dischinger et al., 1993; Siegel et al., 1993).

In a kinematic, electromyographic (EMG) and radiologic study of low velocity rear-end impacts ranging from 10 to 15 km/h, the motion of the head was reported to start 90 ms after the onset acceleration of the occupant compartment and the seat (Castro et al., 1997). The posterior cervical muscles became active 60 ms after the beginning of the acceleration of the vehicle (Castro et al., 1997). In another electromyographic study of rear-end impact at a single acceleration magnitude of 0.5 g EMG was recorded from superficial and deep cervical muscles using surface and inserted electrodes (Kumar et al., 2002a). Surface electrodes were used to record EMG from sternocleidomastoids and trapezius and wire electrodes for semispinalis capitis, splenius capitis and levator scapulae. The authors reported no significant differences between the two types of electrodes. They reported as short reaction times for EMG as 13.2 ms from head acceleration and 65.6 ms from sled acceleration. Kumar et al. (2001a) reported in their study that muscles played an important role in control of head neck motion. Kumar et al. (2002b) reported the EMG of superficial cervical muscles in isometric exertion from a neutral upright posture in eight different directions around the head. The authors demonstrated different response and magnitude profile in exertions in different directions. In a series of papers very different median frequencies, frequency spread and spectral parameters of sternocleidomastoids, splenius capitis and upper trapezius in exertions in sagittal, coronal and oblique planes have been demonstrated (Kumar et al., 2001b, in press(a,b)). Finally Kumar et al. (2002a) studied the same muscles in rear end low velocity impacts and demonstrated significantly higher proportion of activity of sternocleidomastoids in comparison to other muscles. Based on the observations made they suggested the muscles to be probable site of injury in such low velocity impacts.

Therefore, a study was designed where subjects were delivered frontal low velocity impacts in the range of 5.3 m/s² (0.54 g) and 13.7 m/s² (speed change due to impacts 0.96–3.60 km/h) in a random order in expected and unexpected conditions to determine head displacement, velocity and acceleration. An additional objective of the study was to discern the pattern and magnitude of superficial cervical muscle EMG response to the posture perturbation qualitatively and quantitatively through calibrated EMG.

2. Methods

2.1. Sample

Ten normal healthy subjects with no history of whiplash injury and no cervical spine pain in the past 12 months volunteered for the study. The mean age, height and weight of the sample was 24.8 (SD 2.1) years, 172.25 (SD 6.1) cm and 69.5 (SD 9.4) kg.

2.2. Tasks

Seated and stabilized subjects exerted their maximum effort in attempted flexion, extension and lateral flexion to the left and to the right as described by Kumar et al. (2001a). Subsequently, the sled with stabilized subjects was delivered accelerations of 5.3 (0.54), 8.6 (0.88), 11.0 (1.12) and 14.0 (1.43) m/s² (g) (speed change 0.96–3.60) in a random order by the pneumatic piston. The accelerations were delivered under either when volunteers were expecting (expected group) or were not expecting the impact (unexpected group).

2.3. Experimental set-up

The strength-measuring device has been described elsewhere (Kumar et al., 2001a). The EMG system consisted of surface electrodes, electrode cables, preamplifiers and amplifiers. Bipolar electrodes with an inter-electrode distance of 1 cm were used (Model MDI X10 NMRC, Boston, MA, USA). The electrodes were placed bilaterally on the most prominent aspect of the sternal head of the sternocleidomastoids (SCM) and the superior trapezii (TRP) at the C4 level. The electrodes were carefully applied to the identified areas after suitable preparation of the skin. A ground electrode was placed above the right acromion. The low noise and low non-linearity preamplifiers had a common mode rejection ratio of 130 dB and a wide bandwidth. These amplifiers fed to a low power, high accuracy instrumental amplifiers designed for signal conditioning and amplification. The amplifier had AC coupled inputs with single pole RC filter with a low cut off frequency of 8 Hz.

The acceleration device consisted of an acceleration platform and a sled (Fig. 1). The full details of the device are given in Kumar et al. (2000). This assembly allowed a maximum linear speed of up to 36 km/h. At one end of
the platform a pneumatic cylinder with a piston stroke length of 30 cm was connected to an air supply and was mounted rigidly to the acceleration platform. The device was calibrated for the delivery of known forces causing acceleration of 5.3 (0.54), 8.6 (0.88), 11.0 (1.43) and 14.0 m/s² (g). The opposite end of the platform was equipped with a high-density rubber stopper in the sled’s path to prevent it from sliding off the platform.

The sled consisted of a molded plastic seat with a backrest and four legs mounted to a rectangular sliding board coupled with the tracks for friction-reduced travel upon impact. The sled was equipped with a footrest. The seat was fitted with a four-point seat restraint system. The volunteers faced the impacting pneumatic cylinder for all experimental trials.

Three high performance triaxial accelerometers with a full-scale nonlinearity of 0.2% were used in the study. Their dynamic range was ±5 g with a sensitivity of 500 mV/g, a resolution of 5 mg within bandwidth DC-100 Hz.

2.4. Data acquisition

The data acquisition system consisted of an analogue to digital board with a 100 kHz sampling capacity. Each of the nine acceleration channels, six EMG channels and the force channel were sampled at 1 kHz in real time. The sampled signals were stored on a personal computer. The sampling period was 5 s. Strength data was converted to units of force (N). The corresponding peak and average magnitude EMG was measured and used in the normalization of the EMG. Force and EMG data were collected during the strength tests while EMG and acceleration data were collected during the experimental trials. The peak and average strength, EMG and acceleration values obtained from these sets of data were subjected to quantitative and statistical analysis.

2.5. Test protocol

After obtaining informed consent, the age, body weight and height of each volunteer was recorded. The volunteers were then seated on the chair and stabilized in neutral spinal posture for the cervical muscle strength measurement. Simultaneous strength and EMG was made for normalization of EMG in rest of the experiment. The details of the technique are described elsewhere (Kumar et al., 2001a). After the strength testing two triaxial accelerometers were fixed to the volunteer; one immediately inferior to the 7th cervical vertebra at the level of the shoulder and other immediately superior to the glabella region of the frontal bone of the skull. The accelerometers were affixed to the volunteers using strong self-adhesive tapes. The axes of the three accelerometers were referenced with the path of the chair. The pneumatic cylinder was aligned such that the piston head of the cylinder and the baseboard of the front of the sled were in contact. The pneumatic piston delivered the appropriate acceleration to the sled. The subjects in the “expected” group were informed about the forthcoming impact and the magnitude in qualitative terms (very slow, slow, medium and fast). The subjects in the “unexpected” group were blindfolded, and provided with a portable stereo with engaging music playing loud enough to block any auditory cues.

2.6. Data analysis

Since there was no statistically significant differences in the EMG variables between males and females the data from two genders were pooled. Data analysis was performed in three stages. In the first stage, the peak EMG amplitude of the SCM, TRP and SPL (right and left) were measured in response to maximal isometric flexion, extension and lateral flexion (right and left) from the neutral posture. The EMG amplitude corresponding to the peak force in each direction (flexion: SCM; extension: TRP; and right and left lateral flexion: right and left SPL) was given a value of 100%. The EMG amplitudes recorded during the acceleration trials were normalized against these maximal values.

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

In the third stage, a statistical analysis was carried out using the SPSS statistical package to calculate descriptive statistics, correlation analysis between EMG and head acceleration, ANOVA of the EMG slope, time to peak EMG, EMG onset time, peak EMG, average EMG and the force equivalents generated in these muscles.

3. Results

3.1. Accelerations

The accelerative response of the sled, volunteer’s torso and head were recorded in sagittal, coronal and
vertical planes. However, only the peak responses in fore/aft plane are presented here in Table 1 for the four levels of acceleration under unexpected and expected conditions. With an increase in the magnitude of the impact there was a progressive rise in the acceleration of both the sled and the head (Fig. 2A). A backward acceleration of the sled caused a forward acceleration of the head. At backward acceleration of 14.0 m/s² of the sled the torso experienced a forward acceleration of 4.99 and 4.23 m/s² in the unexpected and expected conditions respectively. The corresponding head accelerations for the unexpected and expected conditions were 8.42 and 7.79 m/s². The head acceleration was consistently higher in the unexpected conditions with respect to the expected conditions. The time to onset of the head acceleration was slightly longer in the expected condition in comparison to the unexpected condition. The time of the peak acceleration of the head was significantly higher than that of the sled ($P < 0.01$) and it progressively decreased with increasing acceleration (Fig. 2B).

3.2. EMG amplitude in the frontal impacts

The normalized mean peak EMG amplitudes of the cervical muscles tested in this experiment for the unexpected and expected conditions are presented in Fig. 3. The normalized peak EMG for sternocleidomastoids in both the unexpected and expected conditions was low level, under 30%. However, those for the unexpected conditions reached a magnitude double of the expected conditions. The splenius capitis demonstrated a magnitude and pattern similar to those of the sternocleidomastoids. The maximum activity was observed for the trapezius muscles, which ranged between 38% and 79% increasing progressively with the increasing acceleration for the unexpected condition. For the expected condition the pattern remained the same but the magnitude ranged from 32% at the slowest acceleration and 53% with the fastest acceleration. In terms of force equivalents the sternocleidomastoids represented between 4 and 11 N in the unexpected and 2–6 N in the expected conditions. The splenii ranged between 17 and 30 N in the unexpected and 9–16 N for the expected condition. The trapezii scored 13–31 N in the unexpected and 14–24 N for the expected condition. These force outputs represented 2.8–8.6% of MVC and 5.7–15.7% of MVC for sternocleidomastoids in expected and unexpected impacts. For the trapezii on the other hand the force represented 13.5–23.3% in expected and 12.6–30% in unexpected conditions. The force output for the splenius capitis in expected conditions ranged 9–21.3% and in unexpected conditions 22–40% of the maximum voluntary contraction.

3.3. The EMG slope

In the frontal impacts generally with increasing magnitudes of acceleration there was an increasing
incline in the slope of the cervical muscles. The EMG activity was taken to have commence when it rose by 1% of maximum voluntary construction (MVC) over the stable base line. The EMG slopes for the expected conditions were lower than those of the unexpected conditions. The differences between the EMG slopes of the expected and unexpected conditions were more pronounced for the sternocleidomastoids where they rose from 76–85 to 491–553 lV/s in the unexpected trials and ranged 9–165 lV/s in the expected trials. In the expected conditions the slopes of EMG rise in the sternocleidomastoids did not follow the pattern of progression of the magnitude of the accelerative impacts.

3.4. The timing

The time to onset of the sled, torso and head accelerations in the fore-aft plane and the EMG signals for the six cervical muscles studied are presented in Table 2. The time to onset is measured in real time from the moment of the firing of the piston. The time to onset of the sled, torso and head progressively declined with the increasing acceleration. The time to onset of the trapezius muscles in the unexpected condition also progressively declined with the increasing acceleration. In the expected condition however, it was relatively stable through the range of the acceleration values studied.
The time to onset of the sternocleidomastoids demonstrated a trend of increase with increasing levels of acceleration, in both the unexpected and expected conditions. The splenius capitis muscle did not demonstrate a consistent pattern but the time to onset values generally remained in the same range. The mean times to peak EMG for the cervical muscles are presented in Table 3. The peak EMG activities of the trapezii muscles significantly affected the time to peak sternocleidomastoids were significantly different from those of the splenii and trapezii (P < 0.01). However, the splenii and trapezii were not significantly different from each other.

The levels of acceleration, the muscles examined, and the level of expectations had significant main effects on the slope of the EMG activity (P < 0.001) as well as on the time to onset (P < 0.001). However, gender did not affect either of these variables. For both these variables (slope of the EMG, time to onset) the sternocleidomastoids were significantly different from those of the splenii and trapezii (P < 0.001), but the splenii and trapezii were not significantly different from each other. Finally, the time at which peak EMG occurred was significantly affected by the gender (P < 0.02), the acceleration (P < 0.01), the muscle examined (P < 0.01), and the expectation (P < 0.01) as shown by the multivariate analysis of variance.

The regression analyses were initially carried out up to the acceleration value of 14 m/s² using linear,
quadratic, cubic, exponential and power functions. The best relationships were obtained by power functions, though all others were significant as well. The kinematic variables of displacement of the head, head velocity and head acceleration upon applying acceleration were calculated. The regression equations explained over 98% of variability in all three variables. Subsequently, using these equations, extrapolation up to just above twice the value of the highest applied acceleration was plotted. The percent variability accounted for the raw peak EMG, normalized peak EMG and the force equivalents ranged between 70% and 100% except for the right sternocleidomastoids, which was between 50% and 59%. The slope of the EMG rise was accounted for between 74.8% and 99.9%. The time to onset and the time at which peak occurred were generally predicted well, explaining large amount of variability except the right sternocleidomastoid and left splenius for time to onset and the left trapezius for the time of the peak EMG.

4. Discussion

Whiplash injuries are commonplace in our society with a significant economic burden, and yet they are enigmatic. A great deal of the puzzling nature of this affliction arises from a lack of knowledge of its causation and therefore lack of a precise treatment. However, some have argued that this lack of knowledge is inconsequential (Ferrari, 2001). In his editorial Ferrari (2001) argues that the most effective treatment regimens are nonspecific exercise regimens and general advice (McKinney, 1989; Bonk et al., 2000; Mealy et al., 1986; Vendrig et al., 2000). By showing a drastic difference between different jurisdictions with respect to the different treatments and chronicity, it has been argued that little treatment is needed by the victims of whiplash in Lithuania, Greece and Germany to recover (Ferrari, 2001). The argument states further that it is unlikely that a North American patient is so different to warrant intensive treatment (McKinney, 1989; Borchgrevink et al., 1998; Ferrari and Schrader, 2001; Keidel et al., 2001; Partheni et al., 2000; Martin et al., 2000). The discrepancy so clearly seen in different countries has been assigned to unnamed factors (Ferrari, 2001). Whereas the influence of factors other than physical, on chronic whiplash may be apparent, it can be argued that these mystical factors assume a larger role as the knowledge regarding the mechanism of causation of these injuries is lacking.

Four possible mechanisms of cervical whiplash have been proposed in the published literature. First, in a porcine model, it was demonstrated that after a sudden and violent rearward motion, a considerable, high intracranial pressure was created, which led to neuronal degeneration (Svensson et al., 1993). It has also been reported, in other studies, that during rear-end impacts the cervical vertebrae create a “S” shaped curve causing considerable ligamentous injury (Obelieniene et al., 1999; Ono and Kanno, 1996; Panjabi, 1998). Damage to the facet joints has been proposed as yet another mechanism of cervical whiplash (Panjabi et al., 1998; Yoganandan et al., 1999). In low velocity rear-end impacts, Kumar et al. (2002a) have suggested that the muscles may be the primary site of injury, as sternocleidomastoids exerted 179% of their isometric MVC. All of the foregoing theories have one thing in common that they all have investigated rear-end impacts.

### Table 3

<table>
<thead>
<tr>
<th>Levels of expectation</th>
<th>Levels of acceleration (m/s²)</th>
<th>Sternocleidomastoids</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>
</tr>
<tr>
<td>Unexpected</td>
<td>5.3</td>
<td>555 (386)</td>
<td>1669 (751)</td>
<td>737</td>
</tr>
<tr>
<td></td>
<td>8.6</td>
<td>706</td>
<td>818</td>
<td>796</td>
</tr>
<tr>
<td></td>
<td>11.0</td>
<td>355</td>
<td>377</td>
<td>748</td>
</tr>
<tr>
<td></td>
<td>14.0</td>
<td>299</td>
<td>42</td>
<td>334</td>
</tr>
<tr>
<td>Expected</td>
<td>5.3</td>
<td>1032</td>
<td>787</td>
<td>1027</td>
</tr>
<tr>
<td></td>
<td>8.6</td>
<td>717</td>
<td>352</td>
<td>498</td>
</tr>
<tr>
<td></td>
<td>11.0</td>
<td>953</td>
<td>1022</td>
<td>233</td>
</tr>
<tr>
<td></td>
<td>14.0</td>
<td>1079</td>
<td>743</td>
<td>283</td>
</tr>
</tbody>
</table>

### Table 4

<table>
<thead>
<tr>
<th>Source</th>
<th>Degrees of freedom</th>
<th>F-Value</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gender</td>
<td>1</td>
<td>1.21</td>
<td>NS</td>
</tr>
<tr>
<td>Acceleration</td>
<td>3</td>
<td>4.35</td>
<td>0.001</td>
</tr>
<tr>
<td>Muscle</td>
<td>5</td>
<td>13.04</td>
<td>0.000</td>
</tr>
<tr>
<td>Expectation</td>
<td>1</td>
<td>19.32</td>
<td>0.000</td>
</tr>
<tr>
<td>Muscle × Expt</td>
<td>5</td>
<td>2.44</td>
<td>0.035</td>
</tr>
<tr>
<td>Muscle × Accl</td>
<td>15</td>
<td>1.21</td>
<td>NS</td>
</tr>
<tr>
<td>Accl × Expt</td>
<td>3</td>
<td>3.74</td>
<td>0.012</td>
</tr>
</tbody>
</table>
Contrary to the popular belief, that the whiplash injuries occur mainly as a result of rear-end collisions, there are studies which indicate that significant number of whiplash injuries are caused by frontal collisions (Kullgren et al., 2000; Dischinger et al., 1993; Yoganandan et al., 2001; Buzeman et al., 1998; Cassidy et al., 2000; Hill et al., 1995; Kullgren et al., 1995). Despite such common association of the direction of impact with the whiplash sequale the mechanism of causation of whiplash injuries in frontal collisions remains poorly understood.

Kumar et al. (2002a) argued that since the osteoligamentous preparations of the head and the cervical spine has been shown to be able to support only 1/5th to 1/4th of the head weight (Ono and Kanno, 1996; Panjabi, 1998), the cervical muscles likely play a central role in injury causation. The hierarchical model suggested that the muscles may be the first casualty in a long chain of affected tissues which will get progressively involved with increasing velocity of impact (Kumar et al., 2002a). Likely stretch reflexes will be modulated by either muscle spindles or Golgi tendon organs or both. Thus an establishment of precise injury mechanism may help us in primary prevention as well as the remedial efforts. One of the most common methods of obtaining the cause and effect relationship is to take the tested structure of the volunteering subject to the injury level. However, such an approach is unethical. Hence the approach of testing the subject at several incremental impacts at sub-injury threshold to determine the behavior of the involved structures for the sake of extrapolating variable response up to a reasonable is a valid one.

The threshold injury causing velocity change in frontal impact has been stated to be much higher than that of the rear-end impacts. Therefore, the acceleration values used in the current study, though similar to Kumar et al. (2002a) used for rear-end impacts, are deemed to be proportionally much lower than the possible injury threshold levels in frontal collisions. In the context of this experiment, a higher head acceleration in the unexpected conditions was consistently recorded in comparison to the head acceleration achieved during expected impacts, as also reported elsewhere (Kumar et al., 2000). In the expected conditions the subjects, likely by anticipating the motion, volitionally accelerated with the sled for “catching the motion”. In an unexpected condition the head tended to continue to stay in its original position until the inertia of rest was overcome, at which time it suddenly underwent motion registering a significantly higher acceleration.

The normalized peak EMG activities recorded in this project revealed considerably higher values for the trapezius, which scored up to 53% of the MVC in the expected and 79% of the MVC in the unexpected conditions at such low levels of acceleration. These values did neither reach nor exceed the MVC as the sternocleidomastoid did in rear-end impacts at approximately the same velocity (Kumar et al., 2002a). It should be noted that the trapezius muscle is significantly larger than the sternocleidomastoid, and 13.7 m/s² is relatively closer to the threshold acceleration for a discomfort/injury precipitation in rear-end impact as compared to the 14 m/s² in a frontal collision. Fig. 4 demonstrates as to how the head motion and trapezius EMG are expected to vary with increasing level of acceleration beyond the level tested. The trapezius EMG score will rapidly exceed the MVC level and possibly setting it up for an injury. This suggestion is also supported by the observation of progressive decrement in time to onset of the trapezii with increasing level of acceleration (Table 2) and rising slope of the EMG in the unexpected conditions. In the expected conditions the trapezius show much smaller peak EMG scores (Fig. 3) with much smaller slope and less variable time to onset (Table 2). Since the expected conditions allow the subject to develop a strategy for facing the impact by pretensing the muscles and resisting the motion under relatively much lower velocity impacts for frontal

![Fig. 4. Extrapolated regression plots of the effect of applied acceleration on the left and right trapezius muscle for the variables of the peak EMG (µV), normalized EMG (% isometric MVC) and force equivalents (N).](image-url)
impacts, they generate a less variable response from the EMG.

The time to onset of EMG presented an interesting pattern. The shortest time to onset was obtained for the trapezius, which were generally higher in the unexpected conditions. It generally ranged between 39 and 57 ms in the unexpected conditions whereas in the expected conditions they ranged between 11 and 32 ms. Since the frontal impact tends to propel the subjects backwards their head moves forward. A forced forward motion of the head activates the trapezius first to control or resist the motion. Thus subsequent to the impact when the head has undergone some motion the trapezius come on, likely due to a reflex action. However, in the expected conditions as the subjects were aware of the impact they braced themselves for it quickly as such a faster response. In these impacts the sternocleidomastoids came on much later, after significant motion had occurred. The sternocleidomastoids may be involved in controlling or resisting the head motion backwards during the rebound phase.

It is interesting to note that the gender of the subject did not influence any of the biomechanical parameters significantly. However, as shown previously the expectation significantly affected the head acceleration ($P < 0.01$) (Kumar et al., 2000, 2002a) and the peak EMG, time to onset, and time of the peak EMG (Kumar et al., 2002a).

The data presented in this paper reinforces the findings of Kumar et al. (2002a) in terms of predictability of pattern of the head acceleration and the progressive increase in the electromyographic activity of the trapezius. The normalized peak EMG of the trapezius muscle will reach and possibly exceed its maximal capacity at about twice the level of acceleration used in this experiment and may become the likely site of injury. In this experiment, as elsewhere (Kumar et al., 2002a), a molded plastic chair was used which had no upholstery and its backrest was upright. The selection of chair was motivated by the intent to take a systematic approach of studying the problem with a readily available standard chair, which will have no effect on modifying the human response. Subsequently actual car seats of various makes will be deployed (currently under progress) to discern and quantify the differences. However, it is also suggested that the nature of chair is unlikely to modify the response of the subjects significantly in a frontal collision, as they, under this condition, will tend to be separated from the chair. Therefore, the data presented in this experiment may not have to be adjusted significantly. The variability in the time to onset and also between right and left muscle pairs for the trapezius muscle obtained with different acceleration levels in this experiment supports the argument presented before that the onset of the cervical muscle activities was not a central response (Kumar et al., 2002a). This assessment derives further support from the observation that the expected conditions did not reduce the time to onset of the EMG. A large difference between the onset times of the left and right sides also indicates to the fact that these responses are more likely to be modulated by the stretch reflex upon receiving the mechanical stimulus rather than being controlled centrally. It appears that the somatosensory, visual, and vestibular cues do not have a primary role in this mechanism, though they may modify response to some undetermined extent.

The regression analyses revealed a significant power function relationship between the head motion variables and the acceleration, both applied and projected. It may be pointed out that the projected values reported may get modified by other tissue resistance and geometry in this segment. However, if these factors were not to play a modified role by the virtue of their geometry or modulus of elasticity, it may be possible to estimate the threshold range of acceleration where the injuries are likely to occur in frontal collisions. A similar regression analysis of the EMG variables may help us discern the level of acceleration at which some injury may be likely to precipitate. This will be even more likely when one were to consider the ultimate tensile stress of the trapezius (16 g/mm$^2$). A consideration of the ultimate tensile stress of other structures such as ligaments, fascia and tendons lend support to this argument. Yamada (1973) reported the ultimate tensile stress of ligamentum nuchae (160–320 g/mm$^2$) depending on location of the sample aponeuroses (1.11 kg/mm$^2$), fascia (5.3 kg/mm$^2$) and tendon (5.4 kg/mm$^2$). Compared these foregoing values, the ultimate tensile stress of trapezius muscle is significantly smaller. When all structures are responsible for safety of the neck in the same motion, it is not inconceivable that the failure of tissues follow a pattern of their mechanical properties.

The limitations of the current study lie in use of a plastic chair rather than a real car seat and the mode of delivery of the impact where the sled was accelerated by propulsion of the pneumatic cylinder rather than a real impact. However, it has been pointed out that the molded plastic chair is likely to be less significant in this case as the impact led to a separation of the subjects’ body from the chair. With respect to the impact, the pulse of the acceleration achieved in the experimental conditions were compared with those of car-to-car collisions and were found to have quite comparable characteristics. A further limitation of the dataset presented here is absence of steering wheel in the set-up where drivers would place their hands. This will tend to magnify the responses recorded in the current study which otherwise be mitigated in real life. Finally, it is cautioned the response of trapezius muscles beyond the loading studied is extrapolated. The actual responses may be different. However, if the initial loading conditions were to prevail the extrapolated values are deemed as reasonable estimates.
References


