Effect of Trunk Flexion on the Occupant Neck Response to Anterolateral Whiplash Impacts

ABSTRACT


Objective: The purpose of this study was to determine the response of the cervical muscles to increasing low-velocity anterolateral impacts with the volunteer’s trunk flexed to the right and left.

Methods: A total of 20 healthy volunteers were subjected to left anterolateral impacts of 4.0, 7.6, 10.7, and 13.4 m/sec² and, sequentially, with trunk flexed either left or right. Bilateral electromyograms (EMGs) of the sternocleidomastoids, trapezius, and splenii capitis were recorded.

Design: At an acceleration of 13.4 m/sec², with the trunk flexed left, the left trapezius generated 48% of its maximal voluntary contraction EMG, whereas the right trapezius (contralateral to the left anterolateral impact) generated 38% of this variable. All other muscle generated ≤ 23% of their maximal voluntary contraction EMG, a significant difference from the trapezius (P = 0.005). Similarly, with the trunk flexed to the right under these same conditions, the left trapezius generated 26% and the right trapezius 35% of their maximal voluntary contraction EMG. Again, all other muscles generated significantly less EMG activity, ≤ 22% (P = 0.009). Overall, the EMG responses were of low magnitude compared with known data with the trunk in neutral posture in this direction of impact.

Conclusions: When the subject sits with trunk flexed out of neutral posture at the time of an anterolateral impact, the cervical muscle response is reduced compared with anterolateral impacts with the trunk in neutral posture.

Key Words: Neck Muscles, Electromyography, Accidents, Traffic, Whiplash Injuries
Previously in this journal, Fast et al. conducted a cadaveric study and reported on lumbar spinal strains associated with whiplash injury. As they note, many whiplash claims arise out of low-velocity collisions, but there is a paucity of biomechanical data concerning the biomechanics of spinal injury in low-velocity collisions. Physiatrists and other medical specialists are frequently involved in providing opinions dealing with a number of controversial topics concerning whiplash injury, a problem that costs billions of dollars each year in the United States alone and is an internationally recognized economic and healthcare burden. One question encountered by physiatrists asked to give medicolegal opinions is, “Did the accident cause the whiplash syndrome?” or “Are the patient’s symptoms consistent with the collision injury mechanisms?” As difficult as these questions may seem, one source of information for the clinician may come from engineering studies. A problem that has long been encountered, however, is that the pathology of the acute whiplash injury remains unknown, although the mechanism of acceleration and deceleration of the head is commonly used in the definition of the “whiplash injury mechanism.” Because the acute whiplash injury is not readily demonstrated in most cases in patients or volunteers in studies, engineering studies of low-velocity impacts often rely on outcome measures such as symptoms as a surrogate of injury, although this may be unreliable because symptoms may be nonspecific and, in fact, can arise in placebo collisions. Still, from available data of volunteers subjected to impacts, it seems likely that muscle responses to acceleration of the head are reasonable candidates for injury mechanisms. Although cadaveric and animal studies can define damage to tissue, muscle responses cannot be tested in cadaveric studies and thus require volunteers for study.

Given ethical concerns with subjecting volunteers to injurious neck perturbations, however, to conduct investigations elucidating the kinematics and electromyographic response in volunteers, we have used surface electromyography (EMG) combined with regression techniques modeled 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 response that has been gathered in previous, small studies of high-velocity collisions.

Although this solves the ethical concerns, additional problems in deriving conclusions from volunteer impact studies arise from the fact that most research with volunteers concerns rear or frontal impacts, with few considering other directions of impact. Moreover, even rear and frontal impacts may be offset by some angle so that they are, for example, anterolateral or posterolateral impacts. Furthermore, patients often report that they were turned or twisted to one side at the time of impact, yet impact studies are conducted with the head and trunk in the neutral posture (i.e., the recommended driving posture).

Therefore, considering these more complicated impact scenarios is increasingly relevant in understanding what happens to people in road collisions. Previous volunteer studies of the cervical muscle response and head–neck kinematics to frontal impacts have indicated, for example, that the exact direction of frontal impact is relevant to the measured responses. In straight-on frontal impacts, the trapezius muscles show the greatest muscle response to head acceleration, bearing a greater proportion of the force of the neck perturbation than other neck muscles. It has also been shown, however, that anterolateral impacts (offset by 45 degrees) result in not only increased EMG generation in both trapezi, but the splenius capitis contralateral to the direction of impact also bears part of the force of the neck perturbation.

Again, these studies considered the occupant only in a neutral head and trunk posture. To address a void in current knowledge, we undertook an EMG study to repeat a previous study with anterolateral (45-degree offset) impacts but examined the cervical muscle response when the trunk is flexed forward and to the subject’s right and left to mimic circumstances of “out-of-position” vehicle occupants. To place the magnitude of the muscle response into perspective, with the assumption that a greater muscle contraction in response to impact indicates a greater risk of muscle injury, we compare the EMG responses with the subject’s maximal voluntary contraction EMG. It is assumed that EMG responses lower than the maximal voluntary muscle contractions are less likely to be injurious compared with impacts that cause EMG responses that exceed these physiologic maximum contractions.

SUBJECTS AND METHODS

The methods for this study of left anterolateral impacts (frontal impact offset by 45 degrees) with trunk flexed to the left or right are the same as that used for previous anterolateral impact studies.

Sample

A total of 20 healthy, normal subjects with no history of whiplash injury and no cervical spine pain during the preceding 12 mos volunteered for the study. The 20 subjects (ten women, ten men) had a mean age of 23.6 ± 3.0 yrs, a mean height of 172 ± 7.7 cm, and a mean weight of 69 ± 13.9 kg.
All were 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 at the 6-mo follow-up.

Methods

Active surface electrodes with ten-times on-site amplification were placed on the belly of the sternocleidomastoids, upper trapezius at 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 of 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.19,20 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 et al.19,20

The acceleration device consisted of an acceleration platform and a sled (Fig. 1). The full details of the device and the EMG data collection are given elsewhere.15–18 After the experiment was discussed and informed consent obtained, the age, weight, and height of each volunteer was recorded. The volunteers then were seated on the chair with a lap seatbelt only so that they could then be positioned out of neutral posture. Subjects were then outfitted with triaxial accelerometers (CXLO4M3, Crossbow Technology, 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 left anterolateral impacts (45-degree offset frontal impact) with their trunk flexed forward and to either their left or right at accelerations of 4.0, 7.6, 10.7, and 13.4 m/sec² generated in a random order by a pneumatic piston (Fig. 2). We did not attempt to have the subjects completely relaxed with the neck fully flexed (i.e., slumped posture), as we expected this would not be typical of road collisions. We positioned each of the volunteers in 45 degrees of flexion and 45 degrees of rotation either to the left or to the right. The subjects were asked to assume a position of trunk flexion (forward and lateral) and to look down at their right or left foot. As shown in previous studies, there is a greater muscle response to impact if all visual and auditory cues are blocked before impact.15–18 This is referred to as an unexpected impact status, although even though the volunteers know they will experience an impact at some time. The expected impact status of the previous anterolateral impact study to which we are comparing means the subjects did not have any blocking of auditory and visual cues.17,18 The impact severity and posture positions were randomly varied between the four levels of acceleration, as was done previously.17,18 The accelerations involved in this experiment are again 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.

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 were measured. In the analysis, the sample of volunteers was collapsed across sexes 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 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. This method was chosen to avoid any false positives due to tonic EMG. 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 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.

Statistical analysis was performed using the SPSS statistical package (SPSS, Chicago, IL) to calculate descriptive statistics, 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. In addition, a linear regression analysis was performed for the kinematic variables of head displacement, head velocity, head acceleration, and EMG variables. Initially, all regressions were carried out to the level of exposure, and subsequently, they were extrapolated to twice the applied acceleration. Significance is defined at a P value of <0.01; this level was chosen to correct for the fact that we conducted multiple comparisons. A trend is considered when the P value lies between 0.01 and 0.05.

RESULTS

The kinematic response of the head to the four levels of applied acceleration are shown in Figure 3. As anticipated, an increase in applied acceleration resulted in an increase in excursion of the head and accompanying accelerations. The accelerations in these impacts were not associated with any reported symptoms in the volunteers after the experiment and up to 6 mos later.

In a left anterolateral impact, with the trunk flexed 45 degrees to the right or left, the trapezii were the most active of all muscles (P = 0.004). That is, in left trunk flexion at the time of impact, the left trapezius generated 48% of its maximal voluntary contraction EMG, whereas the right trapezius (contralateral to the left anterolateral impact) generated 38% of this variable. All other muscles generated ≤23% of their maximal voluntary contraction EMG, a significant difference from the trapezius (P = 0.005). Similarly, with the trunk flexed to the right under these same conditions, the left trapezius generated 26% and the right trapezius 35% of their maximal voluntary contraction EMG. The remaining muscles generated ≤22% of their maximal voluntary contraction EMG, a significant difference from the trapezius (P = 0.009).

Although the magnitude of the right splenius capitis was numerically higher, than the left splenius capitis, this was not statistically significant. This trend suggests that the splenius capitis contralateral to the direction of impact has a greater EMG response.

The normalized EMG for the sternocleidomastoid, splenius capitis, and trapezius muscles are shown in Figure 4. As the level of applied acceleration in the impact increased, the magnitude of the EMG recorded from the trapezius increased progressively and disproportionately compared with other muscles (P = 0.009). With trunk flexed to the right or left, the sternocleidomastoid muscles were the least active in response to the EMG (P = 0.008).

To place the magnitude of the EMG responses in perspective, in Figure 5 we show the normalized EMG percentages of the trapezius for two conditions (all data from impacts in which the impact was expected): head and trunk in neutral posture and trunk flexed forward and laterally right or left. The comparison data are from a previous left anterolateral impact study. With the head and trunk in neutral posture in an anterolateral impact, the right trapezius EMG contraction was 74% of the maximal voluntary contraction and the left trapezius contraction was 55% of the maximal voluntary contraction. In the current study, the corresponding values are 38% and 48%, respectively. Thus, trunk flexion significantly reduces the trapezius EMG response (P = 0.009).

The time to onset of the sled, torso, and head acceleration decreased with increased applied acceleration (P = 0.05), though this trend was not statistically significant at the P = 0.01 level. Similarly, the time to onset of the EMG decreased with increased applied acceleration. The mean times at which peak EMG occurred show a trend to earlier times of peak activity with increasing acceleration, but this again did not reach statistical significance.

To obtain the force equivalency of a muscle response due to impact, we first performed a linear regression analysis on the graded EMG data obtained in the maximal voluntary contraction trials. This resulted in an equation for force/EMG ratio. EMG values from each muscle as measured in this impact study were then entered into the equation, giving us a force equivalent value for each muscle. This helps to gauge the magnitude of the response compared with physiologic force exertions. 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. The head acceler-
tions were correspondingly lower than the sled accelerations in this experiment. For very-low-velocity impacts, this is to be expected, as it is usually only when the sled acceleration exceeds $5 \times g$ (5 times acceleration due to gravity) that head acceleration begins to exceed sled acceleration. This experiment involved $<2 \times g$ accelerations.

We used a linear regression model to plot the available data and extrapolate from the experimental accelerations to accelerations on the order of 30 m/sec$^2$. Initially, regression analyses were performed only up to 13.4 m/sec$^2$ using a linear function. The kinematic variables of head displacement, velocity, and acceleration in response to applied acceleration were calculated (Fig. 1). In addition, we also regressed the EMG magnitudes on acceleration. The responses of the left and right trapezius muscles were extrapolated to more than twice the

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**FIGURE 3** Trunk flexed to left and right. 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.
applied acceleration value (Table 1). It is of note that the EMG magnitudes remain low over this range compared with previous studies with the head and trunk in neutral posture. In particular, the sternocleidomastoid muscles show little increase in EMG activity despite increasing acceleration.

**DISCUSSION**

In this study, in which volunteers underwent impacts from the frontal direction offset by 45 degrees to the left (i.e., left anterolateral impacts), we found that if the subjects are flexed to the left or right at the time of impact, then the muscle EMG response is highest from the trapezius, with the contralateral splenius capitis being the next most active muscle. From a biomechanical perspective, we suspect that with the trunk flexed forward and laterally, the out-of-position condition activates the trapezius more because of their role in maintaining this position. In this experiment, we kept the

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FIGURE 4  Trunk flexed to left and right. Normalized peak and average electromyogram (EMG, percentage of isometric maximal voluntary contraction), force equivalent of EMG, and applied acceleration. lscm, left sternocleidomastoid; lspl, left splenius capitis; ltrp, left trapezius; rscm, right sternocleidomastoid; rspl, right splenius capitis; rtrp, right trapezius.
direction of impact constant (left anterolateral) but had the subjects out of position, with the trunk flexed to the right or left about 45 degrees. This was done to determine if the occupant position itself amplified or mitigated the magnitude of cervical muscle response already shown for frontal and anterolateral impacts.\textsuperscript{15–18} In a straight-on frontal impact, for example, the burden of the impact is primarily responded to by the trapezii, and the sternocleidomastoids are not significantly active.\textsuperscript{16} When the frontal impact is offset 45 degrees to the subject’s left or right, however, part of the impact burden is experienced more by the contralateral splenius capitis.\textsuperscript{17,18} Thus, direction of impact determines which muscles respond and the proportionality of the response among the different muscle groups. In this experiment, trunk flexion alters the muscle response significantly.

**FIGURE 5** Trunk flexed to left and right. Normalized peak electromyogram (EMG, percentage of isometric maximal voluntary contraction) at the highest level of acceleration for two conditions: head and trunk in neutral posture and trunk flexed forward and laterally left or right. The example shown here is for the left and right sternocleidomastoid (lscm and rscm) and trapezius muscles (ltrp and rtrp).
The biomechanical rationale for reduced EMG response, implying lesser torque on the cervical spine, is strong. The postural configuration, in which the trunk is flexed, brings it closer to the direction of the force vector of the impact. Although this may increase the axial compression on spinal units, for which they are well adapted, the magnitude of this compression remains minor compared with their tolerance. This posture then reduces the magnitude of the moment arm significantly, which acts as multiplier for the force to develop torque, causing a forced postural perturbation. With a considerably smaller moment arm, the torque is reduced to a large extent. Because the torque tending to cause postural perturbation of the cervical spine is small, the EMG response of the cervical muscles is also small, mitigating risk of injury. The observation in the current study of significantly lower cervical muscle EMG supports this reasoning.

There are limitations to this study. First, we are using results from a previous impact study\(^{18}\) with the trunk in neutral posture (historical controls) in comparing the effect of trunk flexion on the cervical response. We believe this is a reasonable approach given that we have used the exact methodology and equipment in both studies and that there is no difference in the mean ages, sex distribution, or mean weight of volunteers between the studies. Furthermore, the data were normalized and are thus comparable across the subjects. A second limitation may be that the overall EMG activity is lower partly because the impacts were expected (no blocking of visual or auditory cues), and this is known to reduce EMG responses.\(^{15–18}\) Here again, however, when we compare muscle responses between this study and that of a previous study with the trunk in neutral posture, in both cases we are comparing volunteers in the expecting state. Trunk flexion reduces the EMG response when compared with the neutral posture due to a decrease in the moment arm of the impacting force. Third, we only used a lap belt in this study, and the results may have been different if we used a lap and shoulder belt. We argue, however, that 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, we are now conducting additional studies comparing the effect of different restraint systems on cervical muscles response because this may remain relevant. Finally, we are assuming, in extrapolating our results into the low-velocity impact range that the muscle behavior remains linear. Nonlinear regressions were developed as well, but they did not significantly better the linear ones. All extrapolations are approximations of the truth, and the only way to confirm them is to make measurements at the extrapolated ranges. We do know, however, that our extrapolations closely match those from small volunteer studies in which higher velocities were used with symptoms produced.\(^{8,11–14}\) This suggests that regression techniques may allow for extrapolations into low-velocity ranges and may obviate the need for exceeding ethical concerns with experimental designs that could cause volunteer injury. It is recognized that extrapolation may lead one into an erroneous realm if they are not confirmed with available data and exceed the physiologic range. In this instance, neither is the case.

Although we cannot measure injury in this experiment, we have no indication of any significant muscle response, even in the extrapolations that would exceed the normal physiologic limits of strength of these muscles. There are no studies in

<table>
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<th>Force Equivalent EMG, N</th>
<th>Normalized Peak EMG, %</th>
<th>Peak EMG, μV</th>
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LTRP, left trapezius; A, applied acceleration; RTRP, right trapezius; LSCM, left sternocleidomastoid; RSCM, right sternocleidomastoid.
animals in which EMG responses to impact are compared with maximal voluntary EMG. This is not surprising because maximal effort cannot be produced in animals. As we have no other means by which to measure injury potential, we are proceeding on the assumption that the more a muscle’s EMG response exceeds the physiologic maximal voluntary contraction level, the more likely it is to be injured. If the EMG response to 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.

This is the first study to document the cervical muscle response to impact when the occupant is out of the neutral posture, and we are unable to compare with any other approaches to this problem. There has been a review of data from dummy experiments suggesting that being out of position at the time of impact may increase the risk of injury in a frontal impact, not only from impact with the vehicle interior, but through effects of increased cervical extension when the occupant is seated with most of the torso away from the seat and rebounds into the seat after the impact.21 Obviously, it is difficult to extrapolate these results to humans, and our results would not support an increased risk of injury. It may be that at higher velocities, the biomechanics are quite different, and our results likely hold for only very–low-velocity impacts or low-velocity impacts, as confirmed with published data.8,11–14

CONCLUSIONS

Unless the trunk-flexed posture increases the risk of interior vehicle impact, it is likely protective in low-velocity impacts. Compared with occupants sitting with head and neck in the neutral posture, when they are out of position at the time of impact, EMG responses to impact are lower. Thus, the current study finds no evidence to indicate that being out of position at the time of a low-velocity frontal impact amplifies injury risk. If anything, this positioning probably lowers injury risk. It remains unclear, therefore, why it is assumed, as it often seems to be in medicolegal reports, that being out of the neutral posture is more likely to cause injury. There are, to date, no data to support this for low-velocity impacts, and more studies are warranted to determine how occupant position affects whiplash injury risk.

REFERENCES

2. Spitzer WO, Skovron ML, Salmi LR, et al: Scientific mono-

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