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ABSTRACT

Several studies have sought to investigate the biomechanics associated with “whiplash syndrome” by evaluating head kinematics in simulated low-speed rear-end collisions. However, the present study is the first to comprehensively measure head accelerations in six degrees of freedom for the purpose of estimating upper neck loads. In the first phase of the study, nine volunteers were instrumented with a sensor package to measure three-dimensional linear accelerations and angular velocities of the head during rear-end impacts while riding an amusement park bumper car. In the second phase, thirty volunteers were instrumented with the same sensors during selected vigorous activities, including hopping and skipping rope. The linear and rotational head accelerations as well as the calculated upper neck forces and moments for the two groups are presented and compared.

INTRODUCTION

Numerous investigations of volunteer exposures to low-speed rear-end collisions have been conducted to study human response to low-energy impacts [1-17]. These studies have made use of volunteers in order to evaluate occupant kinematics and injury potential. While the vast majority of volunteers had no complaints after participating in these studies, some volunteers complained of, at most, transient soreness or headache that disappeared within a few days. The main result common to all of these experimental tests was that the loads acting on the cervical spine during low-speed rear-end impacts were substantially lower than the thresholds for damage to the bones, ligaments, and discs of the spine, as reported in the biomechanical literature [18-23].

Nonetheless, cervical spine loads generated during low-speed rear-end collisions have been associated with a condition commonly referred to as “whiplash” or “whiplash syndrome.” The National Highway Traffic Safety Administration (NHTSA) has identified “whiplash syndrome” as the predominant injury sustained by occupants in rear-impact motor vehicle collisions. Although much effort has been focused on mitigating whiplash (for example, the advancements in head restraints have been shown to mitigate a large number of injuries), an estimated 272,000 persons involved in rear-impact collisions report sustaining whiplash annually [24]. A lack of objective symptoms makes whiplash syndrome difficult to diagnose, and the underlying injury mechanisms are still unclear. Consequently, there have been many attempts to define the tolerance or lower limits of whiplash injury by quantifying non-injurious neck ranges of motion, head accelerations, and neck loads and moments in volunteers subjected to simulated low-speed rear-end collisions [1-3, 5-9, 12-15].

Traditionally, low-speed rear-end collisions have been simulated by performing vehicle-to-vehicle collisions in a controlled environment or by employing sled test devices. More recently, several researchers have used amusement park “bumper cars” as analogues for low-speed rear-end collisions [4, 10], reporting that although the collisions tended to be more elastic than in automobiles, the occupant kinematics were biomechanically comparable. Despite their different approaches, these studies have typically used high-speed video to measure the range of cervical motion during rear-end impacts as well as to calculate head accelerations. However, this technique is limited because it cannot account for any out-of-plane motions and indirect calculation of the head accelerations is prone to accumulation errors. Likewise, arrays of linear accelerometers have been used by others to determine total acceleration (i.e., the linear and angular acceleration) of the head [7]. These systems typically require bulky apparatus to provide rigid connections between the sensor groups or very complex optimization routines [25]. To date, the only direct measures of total head accelerations have employed bite-plates which require the participant to actively clench their jaw during an entire trial for accurate readings. The measurement systems used previously lacked the ability to comfortably measure the total head acceleration in both collision environments and during longer activities.

Other researchers have related low-speed rear-end collision severity and occupant kinematics with calculated neck forces and moments. These results have been presented in the context of, or compared to existing tolerance thresholds for damage to the bones, ligaments, and discs of the spine; which are relevant to at best an upper bound for whiplash related injuries. However, the study of Allen, *et al.* (1994) sought to compare the head accelerations associated with a number of daily activities with the phenomenon of “whiplash” [26]. Although Allen, *et al*. did not collect any volunteer data from low-speed rear-end collisions by which to make this comparison, nor did they attempt to calculate neck loads and moments, their study established an initial reference point for comparing daily activities to head accelerations in low-speed rear-end collisions. Therefore, in the present study, it was proposed to expand on this concept by comparing volunteer head accelerations as well as neck loading data associated with both low-speed rear-end collisions and common vigorous activities. With these results, it is possible to establish a reference set of data by which to compare the loads and accelerations experienced in low-speed rear-end collisions to activities that are considered non-injurious.

METHODS

The study was conducted into two phases. In the first phase, nine volunteers were instrumented with a sensor package to measure three-dimensional linear accelerations and angular velocities of the head during rear-end impacts while riding an amusement park bumper car. In the second phase, thirty volunteers were instrumented with the same sensors while performing vigorous activities.

Both phases of the study used a head sensor package that was designed and constructed with micro-electromechanical (MEMs) gyroscopic rate sensors and linear accelerometers that were mounted to an adjustable lightweight headband. The sensor group (Blackdust Design, Los Angeles, California) had a range of ±10 Gs (resolution: 0.0003 Gs) along each axis and the gyroscopic rate sensors were capable of measuring ±5.24 rad/s (resolution of 0.00016 rad/s) about each axis. Each channel was low-pass filtered using a 100 Hz anti-aliasing filter then digitally sampled at 500 Hz. For testing, the headband was positioned such that the sensors were at the top of the head and its axes were coincident with the mid-sagittal plane and the coronal plane through both external auditory meati. The complete apparatus weighed 85 g (3 oz.).

To record the data, the sensors were connected to a small, lightweight (200 g) data logger located in a waist pack worn by the test subject. A flexible cable ran from the sensor package to the data logger along the subject’s back and was strain relieved with athletic tape at the back of the neck. Each individual verified that the placement and attachment of the cable did not impede normal head motion relative to the torso.

The data from the sensors were filtered digitally using a 50 Hz low-pass, zero phase shift filter. For all trials in both parts of the study, a power spectrum density (PSD) analysis was performed to determine if the sampling frequency was sufficient; all significant linear accelerations and angular velocities were captured for all trials, as the PSD magnitude decreased consistently by more than 30 dB beyond 15 Hz.

PHASE I: BUMPER CAR COLLISIONS

Five female and four male participants volunteered for the bumper car portion of the study, which simulated a rear-end automobile collision using standard amusement park bumper cars. The participants ranged in age from 24 to 31 years old (mean 27 years), with heights measuring between 1.57 and 1.92 m (mean 1.73 m) and masses between 58 and 76 kg (mean 65 kg). Participants were informed as to the purpose of the study and the procedures involved, answered questions regarding their anthropometry and medical history, and provided their voluntary consent to participate in the study and be videotaped. Potential subjects were excluded if they had an increased risk of heart failure or joint problems. Also, the participants were allowed to withdraw from the tests at any time and for any reason. A Human Subject Research Committee at Exponent approved the test procedures. After the experimenters measured the participants’ height, weight, head circumference, neck circumference, vertex to occiput height, and vertex to T1 height, the participants were fitted with the head sensor package.

In addition to the head sensor package, sagittal plane vehicle and occupant kinematics were captured with a high-speed digital camera (RedLake Motion Xtra HG100K, Redlake, San Diego, California). The high-speed video was recorded at 250 frames per second at a resolution of 1504 x 1128 pixels; this corresponded to a spatial resolution of approximately 1 x 1 mm. Markers were placed on the side of the headband and at the shoulder, elbow, and dorsal aspect of the hand of the participants. Targets were also placed on the side of the vehicle, one at the front and one near the rear of the vehicle.

Bullet vehicle impact velocity and target vehicle acceleration were measured for each trial. The average bullet vehicle speed just prior to impact was measured with a floor-mounted speed trap constructed from contact switches. The target vehicle acceleration was measured with a triaxial accelerometer (Endevco 2262A-100) mounted on the floor of the target bumper car along the vehicle centerline; the range of the accelerometers was ±100 Gs (resolution: 0.0488 Gs) along each axis. Each channel was low-pass filtered using a 2 000 Hz anti-aliasing filter then digitally sampled at 10 000 Hz. A contact switch located at the rear of the target vehicle triggered both the data acquisition and a flash bulb, in order to synchronize the data acquisition with the high-speed video.

The target vehicle was positioned at the end of the bumper car rink approximately 2 feet away from the flat vertical edge bordering the rink. In order to reduce anticipation of the impact, an experimenter distracted the participant by discussing current events. Additionally, the bullet vehicle driver randomly selected the amount of time elapsed between trial initiation and impact by driving around the rink for varied lengths of time prior to impact. The same experimenter drove the bullet vehicle for all trials. The impact caused the target vehicle to move forward and contact the edge of the rink, which stopped the forward motion of the target vehicle. Each participant took part in three trials.

PHASE 2: VIGOROUS, NON-INJURIOUS ACTIVITIES

Fifteen male and fifteen female participants voluntarily performed the activities chosen for the second part of this study; none of the subjects took part in the first part of the study. The participants exhibited a wide range of age (from 18 to 44 years old), height (1.56 to 1.89 m, mean 1.72 m), and mass (51 to 106 kg, mean 73 kg). Upon arrival, each participant listened to a standardized description of the activities and goal of the study, provided his or her consent to participate and be videotaped, and provided relevant medical history for screening purposes. The participants were allowed to withdraw from the tests at any time and for any reason. As with the first part of the study, potential subjects were excluded if they had an increased risk of heart failure or joint problems. A Human Subject Research Committee at Exponent approved the test procedures. The experimenters measured the participants' height, weight, head circumference, neck circumference, vertex to occiput height, and vertex to T1 height and fitted the participant with the head sensor package.

The thirty participants voluntarily performed the activities chosen for the study: running with an abrupt stop, falling back into a soft chair, skipping rope, and hopping with two feet. For each participant, the tasks were assigned randomly in order to eliminate the possible influence of task order on measurements, i.e. to eliminate the possibility that fatigue from one task would influence head motions during the next task. The subjects were asked to perform the tasks in a quick and precise manner, and each task was recorded using a digital video camera. Before each activity, subjects were asked to keep their head stationary for 5 seconds and then perform a set of head motions used to synchronize the data with the video. These motions were also used to verify the relative alignment of the sensors.

DATA PROCESSING

All data processing was completed using Matlab (The MathWorks, Inc., Natick, Massachusetts). For each trial, the angular velocity data about each axis were differentiated to calculate the angular acceleration of the head. The analysis incorporated the fact that the output of the linear accelerometers was the vector sum of the sensor acceleration and the negative of the acceleration due to gravity. The linear acceleration at the sensor and angular velocity data were used to determine the linear accelerations at the approximate center-of-mass (COM) of the head using equation (1). The distance from the sensor to the COM of the head was estimated using the anthropometric measurements and regression formulae given in Zatsiorsky, 1985 [27].

$$\mathbf{a}_X = \mathbf{a}_{Sensor} + \boldsymbol{\alpha} \cdot \mathbf{r}_{X-Sensor} + \boldsymbol{\omega} \times \left(\boldsymbol{\omega} \times \mathbf{r}_{X-Sensor}\right) \quad (1)$$

where X is either the head COM or the AOJ, $\mathbf{a}_X$ is the linear acceleration at point X, $\mathbf{a}_{Sensor}$ is the linear acceleration measured by the sensor, α is the angular acceleration calculated from the sensor, $\mathbf{r}_{X\text{-}Sensor}$ is the vector from point X to the sensor center, and ω is the angular velocity of the head measured by the sensor. The head loading applied to the upper neck was calculated using Euler's equations; it was assumed that the head was a rigid body, that all forces and torques were applied between the head and the neck at the occipital condyles, and that the distance from the head COM to the AOJ was fixed (at 1.9 cm posterior and 6.2 cm inferior to the COM) [19, 28].

After all of the linear acceleration components were calculated at each location of interest, the resultant linear acceleration and angular rate vectors were determined. For each participant in each trial, the maximum linear acceleration, angular velocity, and angular acceleration at the head COM, and estimated upper neck loads were determined. The maxima were determined along each axis (or about each axis, in the case of angular data) and for the resultant vectors. Data were presented in terms of mean and standard deviation except where indicated.

RESULTS

The sensors did not interfere with the participants' head range of motion or their ability to complete the tasks in either phase of the study. All thirty-nine participants were able to complete the tasks and no participant complained of headaches, neck pain, or back pain during or immediately after the series of tasks.

All acceleration data are presented as head accelerations at the head COM, with a basis that is corotational with the head. In anatomical terms, the x direction is perpendicular to the coronal plane, the y direction is perpendicular to the sagittal plane, and the z direction is perpendicular to the transverse plane. The following abbreviations are used: AxLin, AyLin, and

AzLin, for linear accelerations in the anterior-posterior, medio-lateral, and inferior-superior anatomic directions, where anterior, right, and inferior are positive, respectively; AxRot, AyRot, and AzRot for rotational accelerations about the anterior-posterior axis, medio-lateral axis, and inferior-superior axis, respectively; Vx, Vy, Vz for positive velocities in the anterior, right, and inferior anatomic directions, respectively. The right-hand rule applies for determination of the direction of positive rotation. ArLin and ArRot designate the resultant vectors of the linear and rotational accelerations, respectively. As previously described, the linear and rotational accelerations are used to estimate the forces and moments of the head acting on the neck.

For the upper neck loading results, all components are reported for the force and moment applied on the neck by the head. Fx designates the shear force acting on the neck at the AOJ along the anterior-posterior axis, where anterior shear is positive and posterior shear is negative; Fz designates the superior-inferior axial force acting on the neck, where tension is positive and compression is negative; and, Mocy designates the moment component about the medio-lateral axis acting on the upper neck by the head at the occipital condyles, where flexion is positive and extension is negative.

PHASE I: BUMPER CAR COLLISIONS

During the rear-end collisions, the change in velocity of the target vehicle ranged from 3.9 to 5.9 kph and the peak vehicle resultant acceleration ranged from 1.2 to 6.9 G. The peak resultant linear head accelerations of the nine volunteers ranged from 2.8 to 6.5 G (mean 4.0 G), with the largest component always in the positive x-direction. Peak resultant angular accelerations ranged from 107 to 308 rad/s^2. In general, the lateral linear accelerations of the head were much lower than the other linear components, usually less than 1 G. However, no clear correlation between peak vehicle acceleration and peak linear head accelerations was observed even for a single volunteer at different vehicle accelerations. The average peak neck extension moment was 8.1 Nm, the average peak anterior shear was 139 N, and the average axial compression was 92.7 N.

As previously described, the rear impact of the target bumper car by the bullet bumper car was followed a few seconds later by a frontal impact of the target bumper car with the edge of the rink. However, for almost all parameters in all collisions, the rear impact associated volunteer data peaks were much greater than the subsequent frontal impacts. The vehicle accelerations and velocities measured during an example bumper car impact as well as the associated neck loads and head accelerations are shown as a function of time for an example impact in the Appendix (Figure A1 and Figure A2).

PHASE II: VIGOROUS, NON-INJURIOUS ACTIVITIES

The peak resultant linear acceleration during the vigorous activities ranged from 2.8 to 9.9 G and resultant angular accelerations ranged from 58 to 427 rad/s^2. The resultant linear and angular accelerations during the vigorous activities were much larger than those found in the simulated rear-end collisions, often exceeding 4 G and 100 rad/s^2. Three males generated resultant linear accelerations greater than 7 G and angular accelerations greater than 300 rad/s^2, but this phenomenon only occurred during the hopping task. Example traces of the neck loads and head accelerations as a function of time for each of the vigorous activities are presented in the Appendix (Figure A3).

COMPARISON OF BUMPER CAR REAR-END IMPACTS AND VIGOROUS ACTIVITIES

The parameter values obtained during vigorous activities were compared to the corresponding values obtained during bumper car testing to determine if statistically significant differences existed. If a statistically significant difference from the bumper car data was obtained in a two-tailed t-test ($p<0.05$) it was indicated on the following figures using the "*" symbol.

The average peak neck flexion and extension moments about the occipital condyles for the four vigorous activities are shown in relation to the average maximum neck moments obtained during the bumper car impacts (Figure 1). The largest average peak flexion moment was obtained during hopping. This was significantly different from the flexion moment measured during bumper car testing. The bumper car average peak flexion moment was greater than that of skipping rope and falling into a chair, but less than running with an abrupt stop. Although the average peak extension moment recorded during the bumper car testing was of the same order of magnitude, it was significantly greater than that of the four vigorous activities.

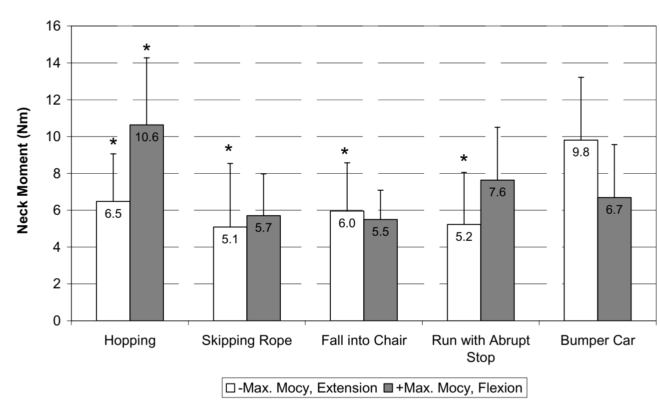

Figure 1. Average peak neck flexion and peak extension moment about the occipital condyles for all five activities.

The average peak axial compression and axial tension at the occipital condyles are shown in Figure 2. The bumper car average peak axial compression force was smaller than that of the vigorous activities, underscoring the fact that the bumper car impact is horizontally directed, with little vertical component. This difference was statistically significant. Tensile axial forces across the exercises were similar, with the exception of falling into a chair and running with an abrupt stop. There was a significant statistical difference found for average peak tension between the bumper cars and a fall into a chair and running with an abrupt stop. The former may be explained by the lack of rebound in this exercise compared to the other exercises, in part due to the energy absorption of the chair. The average peak axial compression was the highest for the hopping activity.

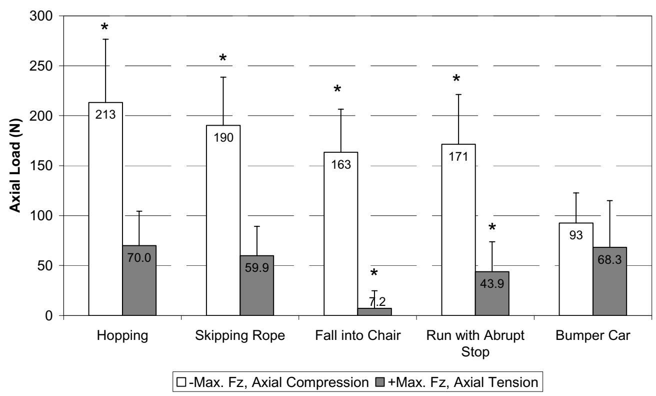

Figure 2. Average peak axial tension and peak axial compression for all five activities.

The average maximum posterior and anterior neck shear at the occipital condyles for the four vigorous activities and bumper car impacts are shown in Figure 3. The maximum anterior shear occurred during the bumper car impacts. The difference between the maximum anterior shear obtained during the bumper car rear-end impacts and the maximum anterior shear obtained during each of the activities was statistically significant. The maximum posterior shear occurred during hopping.

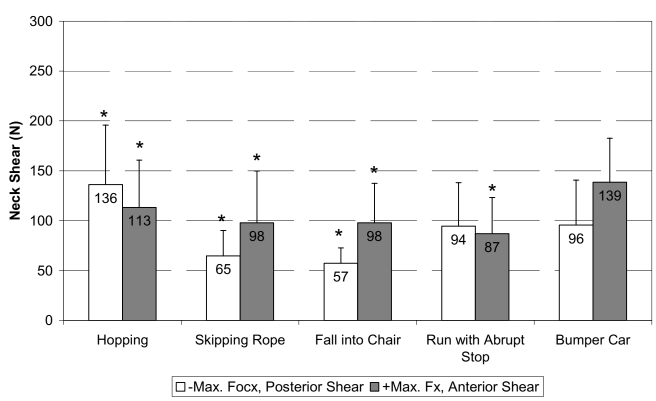

Figure 3. Average peak anterior and peak posterior shear for all five activities.

The average peak rotational accelerations in each axis of the head of the volunteers for the vigorous activities and bumper car impacts are shown in Figures 4 through 6. The peak rotational accelerations were greatest about the y-axis, followed by the x-axis and the z-axis for all exercises with the exception of running with an abrupt stop. Running with an abrupt stop resulted in a slightly greater average peak resultant rotational acceleration than that of hopping. The average peak resultant rotational acceleration in the bumper car tests exceeded those of skipping rope and falling into a chair. There were statistically significant differences between the resultant rotational accelerations obtained during the bumper car impacts and the resultant rotational accelerations for hopping, falling into a chair, and running with an abrupt stop.

The peak rotational acceleration about the x-axis was similar for each of the exercises, with the bumper car having the lowest overall magnitude (Figure 4). Running with an abrupt stop recorded the greatest rotational accelerations about this axis, nearly three times greater than that of the bumper cars tests. It is notable that hopping caused rotational accelerations about the x-axis approximately twice that of the bumper car tests. The differences between rotational accelerations about the x-axis for both hopping and running with an abrupt stop and the rotational accelerations for the bumper car impacts were statistically significant for both the positive and negative directions. There was no significant difference found between the rotational accelerations about the x-axis in the positive direction and those in the negative direction for each of the vigorous activities. There was a statistically significant difference between those in the positive direction and those in the negative direction during the bumper car impacts.

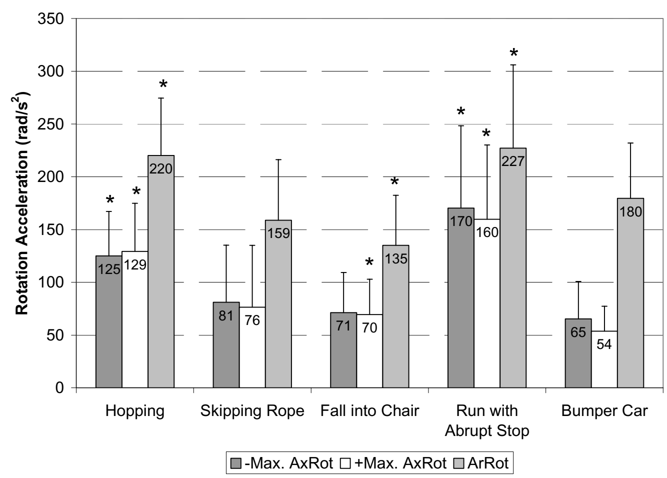

Figure 4. Average peak rotational acceleration about the x-axis and average peak resultant rotational acceleration.

The peak rotational acceleration about the y-axis was greatest in running with an abrupt stop, followed closely by the hopping task (Figure 5). The greatest negative rotational acceleration occurred during running with an abrupt stop, followed closely by the hopping task. Despite the rotation about the y-axis caused by the rear-end impact, the angular accelerations during the bumper car collisions were not the greatest in either the negative or positive direction of the y-axis.

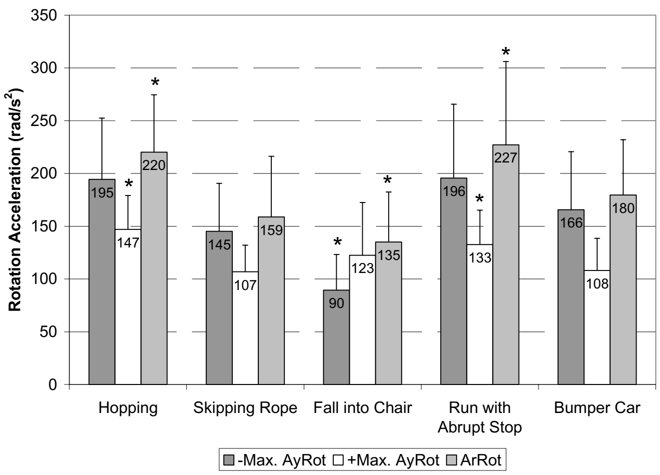

Figure 5. Average peak rotational acceleration about the y-axis and average peak resultant rotational acceleration.

About the z-axis, there was no significant difference between rotational accelerations in the negative and positive directions in any of the activities or in the bumper car impacts (Figure 6). Running with an abrupt stop resulted in the greatest magnitude in rotational accelerations about this axis.

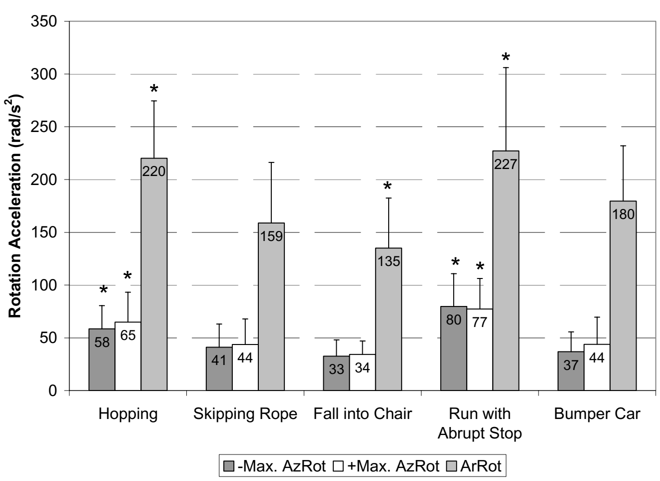

Figure 6. Average peak rotational acceleration about the z-axis and average peak resultant rotational acceleration.

The average peak linear accelerations at the head COM of the volunteers for the vigorous activities and bumper car impacts are shown in Figure 7 through Figure 9. The largest linear accelerations in the positive x- and positive z-directions were obtained during the bumper car impacts. The largest linear accelerations in the negative x- and positive z-directions were obtained during hopping. The largest linear accelerations in the positive and negative y-directions were obtained during running with an abrupt stop. The maximum resultant linear acceleration was obtained during hopping.

The largest average maximum linear acceleration in the positive x-axis occurred during the bumper car impact (Figure 7). The differences between the positive linear accelerations obtained during the bumper car impacts and those obtained during each of the activities were statistically significant. The largest negative linear accelerations along the x-axis occurred during the hopping task. The next largest negative linear accelerations occurred during the bumper car impacts.

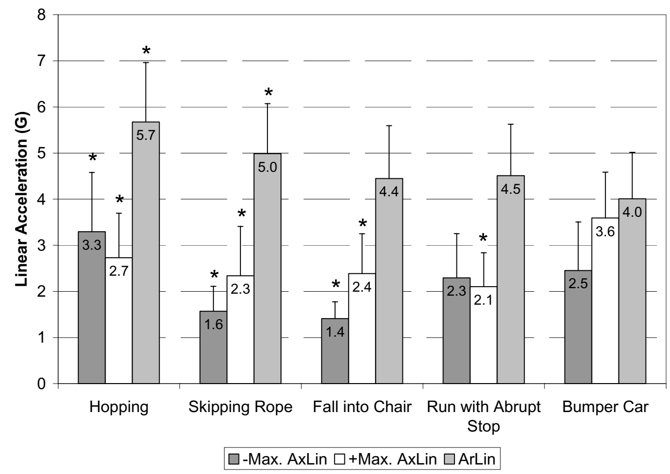

Figure 7. Average peak positive and peak negative linear acceleration in the x-axis and average peak resultant linear acceleration.

The positive and negative linear accelerations along the y-axis are shown in Figure 8. For each activity, these values are nearly equivalent, indicating that these activities had little lateral component, similar to what was found in the rotational accelerations shown previously. There was no significant difference found between positive and negative linear accelerations in the y-axis for each of the vigorous activities. There was a statistically significant difference between the positive and negative linear accelerations obtained during the bumper car impacts.

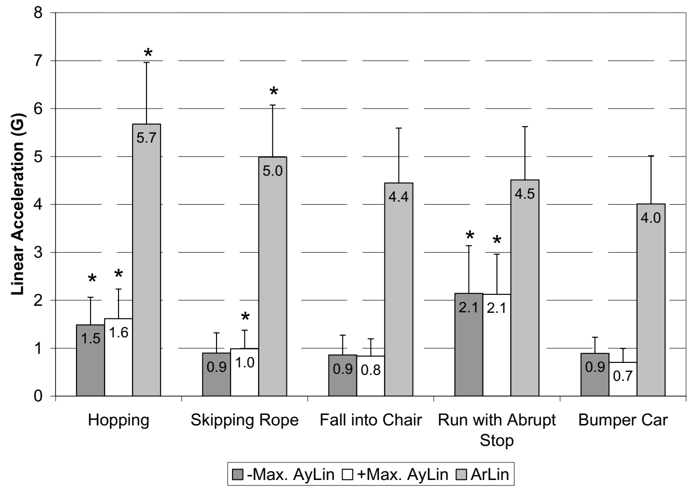

Figure 8. Average peak positive and peak negative linear acceleration in the y-axis and average peak resultant linear acceleration.

The average peak linear accelerations in the z-axis are shown in Figure 9. The average peak linear accelerations in the negative z-direction measured during the bumper car impacts were significantly lower than those measured during each of the activities. This follows from the fact that all the vigorous activities included some component of vertical displacement, while the bumper car task was primarily an anterior/posterior activity. The magnitudes of the negative z-axis linear acceleration were significantly lower than the magnitudes of the positive z-axis linear acceleration for the vigorous activities.

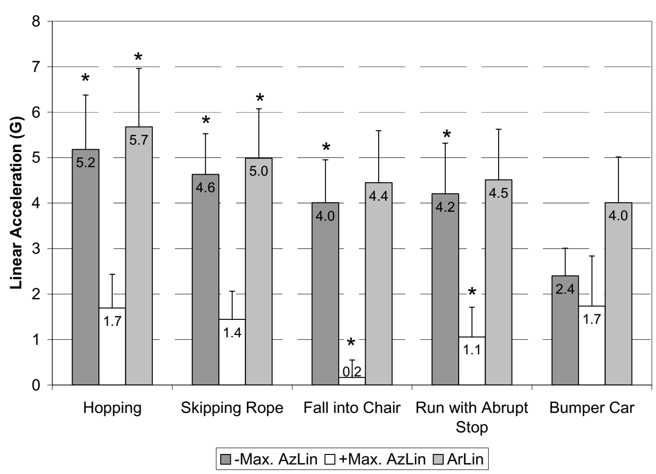

Figure 9. Average peak positive and peak negative linear acceleration in the z-axis and average peak resultant linear acceleration.

DISCUSSION

For both phases of the present study, the levels of upper neck forces and moments were significantly less than reported neck injury tolerances. In particular, the results of this study indicate that the upper neck moments experienced by the volunteers exposed to the bumper car collisions of the present study were similar to or lower than those recorded in other rear-end collision studies of similar magnitude (Figure 10). As in these other studies, none of the participants of the current study presented with any symptoms related to their exposure to the rear-end collisions. Although these other studies were comparable in collision magnitude, the authors are not aware of any other study to measure head accelerations directly and to use those data to calculate the estimated upper neck loads and moments. Additionally, this is the first known study to directly compare volunteer data from both low-speed rear-end collisions and vigorous activities.

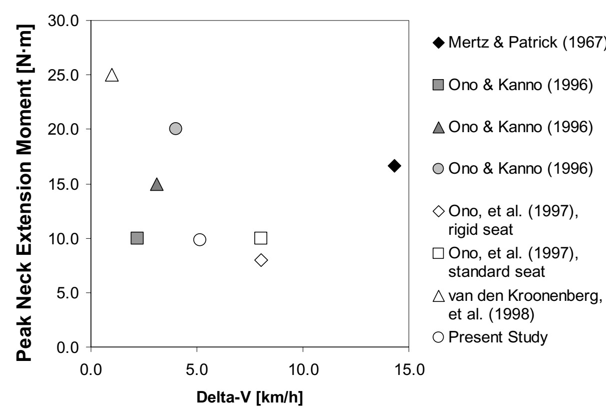

Figure 10: Peak extension moments observed during low-speed rear-end volunteer exposures.

With the exception of anterior shear loads and extension moments, the volunteers exposed to the bumper-car collisions experienced comparable or lower upper neck loads than the volunteers who participated in the vigorous activities outlined above. However, the shear loads were not considered significantly different between the two groups. Additionally, comparison of supra-tolerance cadaveric testing to human voluntary limit data revealed that neither shear forces nor axial forces are significant factors in neck injuries associated with rear-end collision [2]. This may be due to the neck stiffness compared to the dynamic load: the neck is stiff enough to cause the head acceleration to produce a significant moment before significant axial or shear forces are produced (all relative to voluntary maxima).

The linear and angular head accelerations experienced in both the simulated rear-end collisions and the selected vigorous activities of the present study were again well below published injury tolerance limits [29]. Additionally, the accelerations recorded in the present study were orders of magnitude lower than the accelerations that have been measured in a variety of contact sports [25, 30, 31]. Therefore, considering both the forces and moments acting on the upper neck during low-speed rear-end collisions of similar severity to the bumper car collisions of the present study, the risk of sustaining a head or neck injury is minimal in such an event and is likely no greater than the risk associated with participating in a typically non-injurious, vigorous activity.

Although these results indicate that it is extremely unlikely that an individual would sustain a head or neck injury in a low-speed rear-end collision, it is possible that these results may only be relevant to the population selected for this study. However, no significant age-related differences were observed in the linear and angular head accelerations recorded in the vigorous activities selected for this study. Additionally, it is noteworthy that the levels of head accelerations that the volunteers experienced were similar to those recorded by Allen *et al*. in more benign daily activities, such as sneezing, coughing, and being jostled in a crowd [26].

It is notable that in the placebo study conducted by Castro *et al*. (2001), in which participants were led to believe they had been in a moderate-speed rear-end collision when in fact they had only been exposed to an aural and visual simulation of a rear-end collision, nearly 20% of their test subjects reported symptoms, including neck pain [32]. Their findings suggest that there exists a psychological component to the origin of some “whiplash” symptoms. The results of the present study lend biomechanical support to these findings.

The present system of measurement provides significant advantages over those of previous studies, including the minimal weight and interference imposed by the acceleration measurement apparatus. However, in the present study, resultant neck loads were calculated from head motion as it was not possible to take direct measurements. The effects of muscle contractions, co-contractions of synergistic muscle groups, the effect of volunteer anticipation and expectation, and the effect of repeat exposures were not considered. As such, the reported upper neck loads may be regarded as a lower limit of possible loading on a cross-section of the neck.

CONCLUSION

In summary, the head accelerations and calculated upper neck loads and moments recorded during the low-speed rear-end bumper car collisions were found to be comparable to or lower than those experienced during vigorous activities as may be expected of healthy adults. Therefore, it is unlikely that low-speed rear-end collisions of severity comparable to the present studies are mechanically related to neck injuries in a comparable population.

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APPENDIX

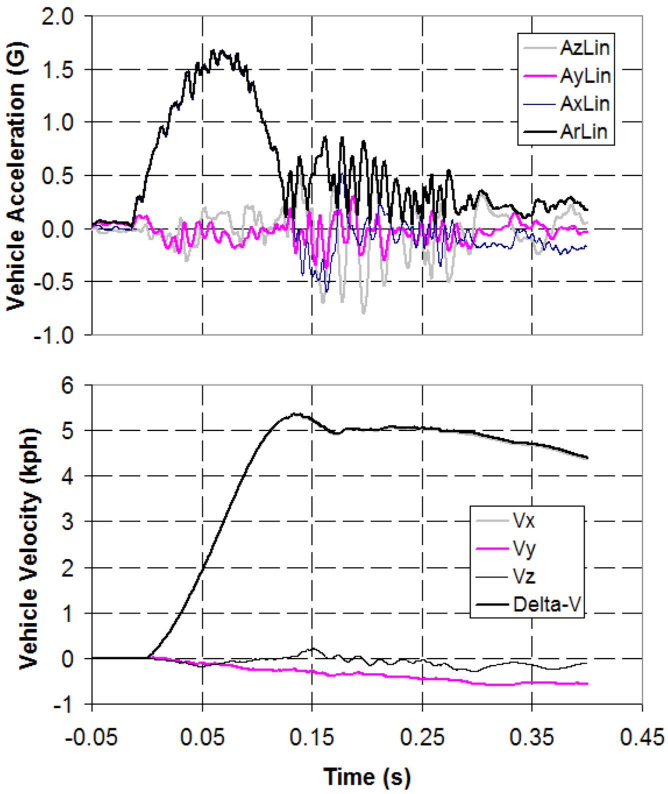


Figure A1. Representative plots of bumper car vehicle acceleration and velocity.

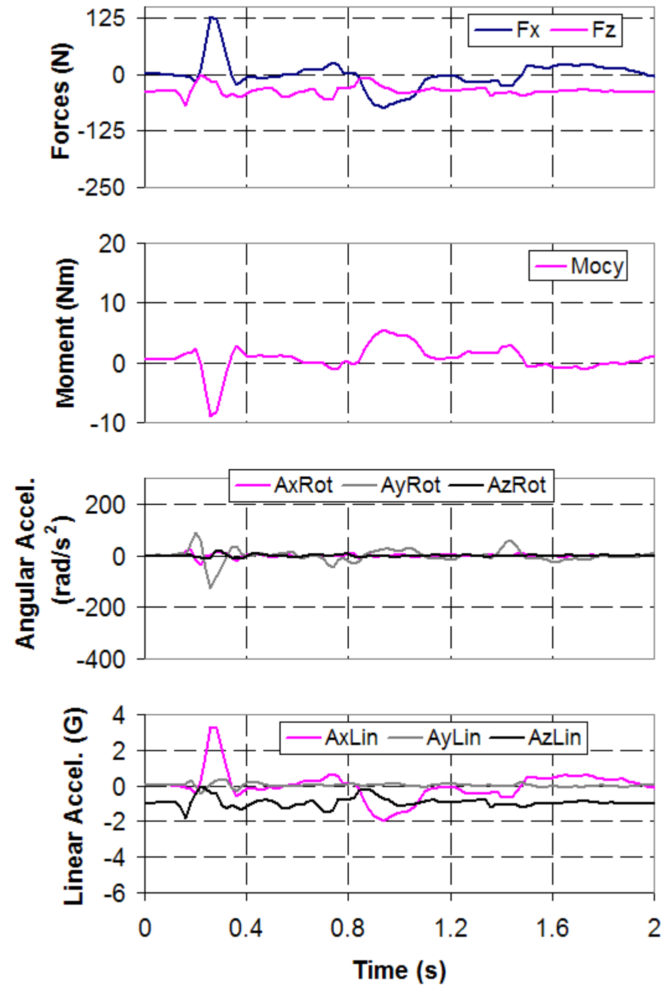
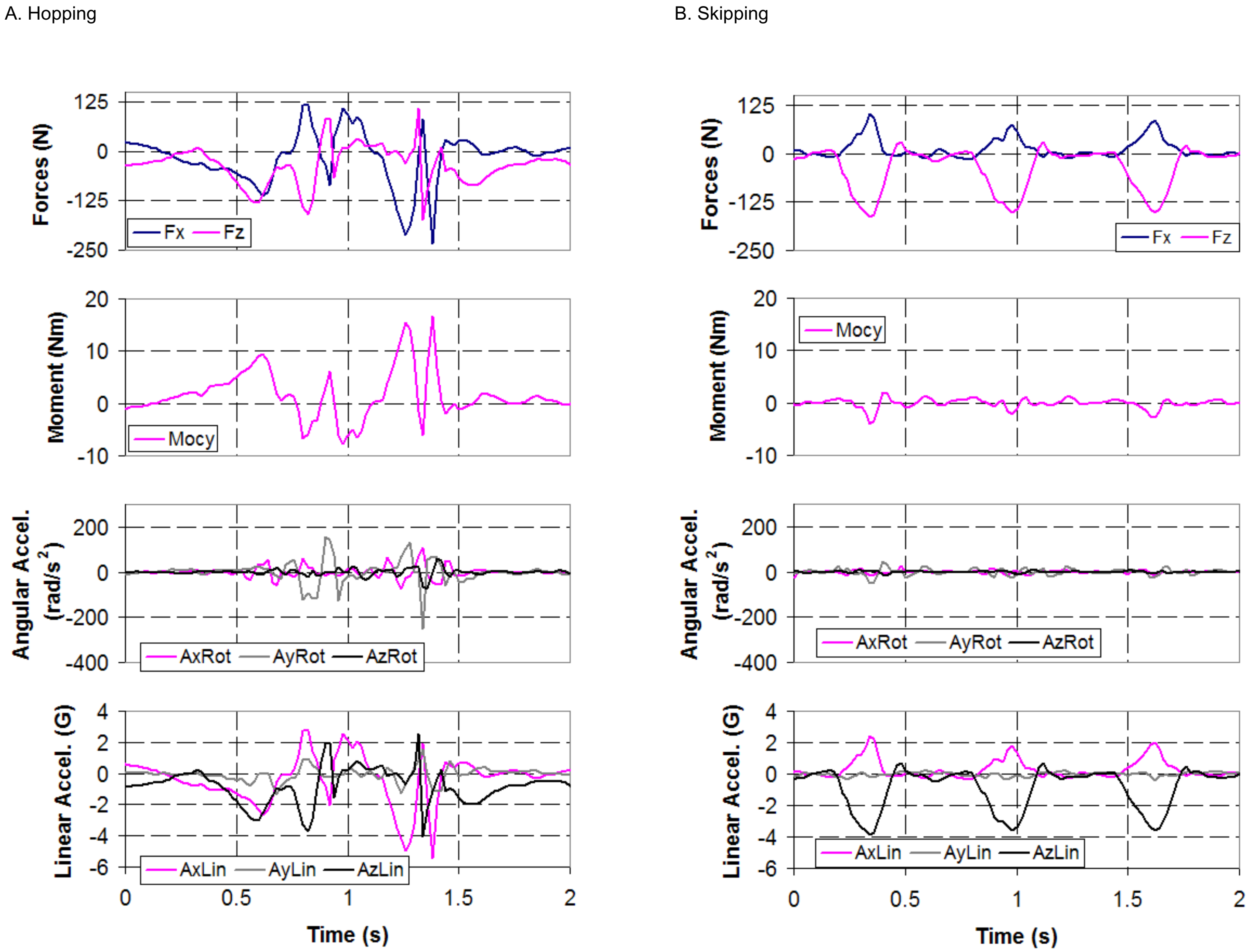
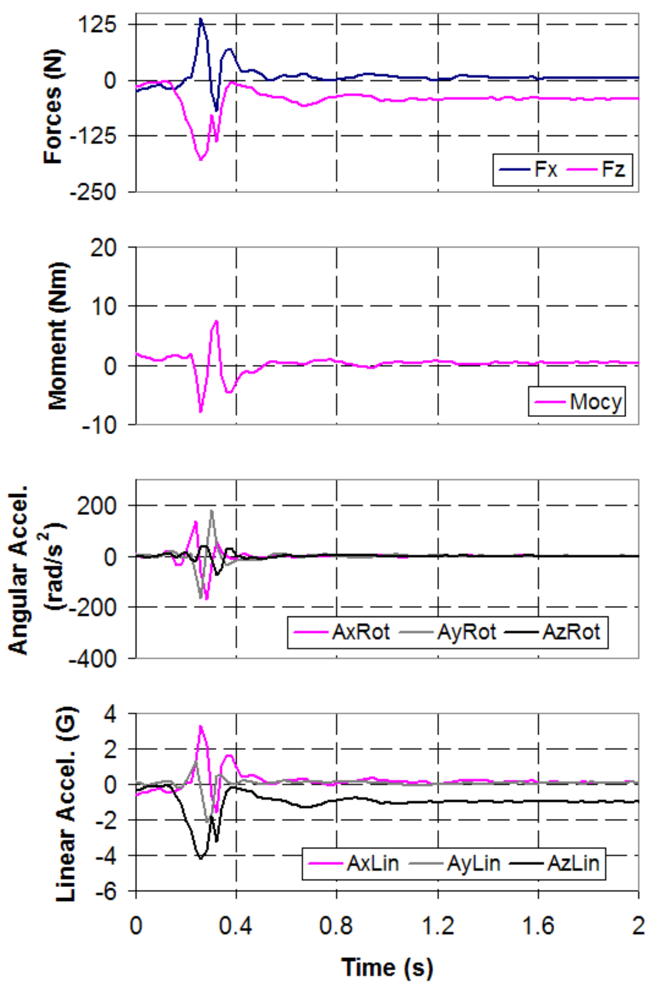


Figure A2. Representative plots of calculated forces and moments at AOJ, and head rotational and linear accelerations at head COM in bumper car studies.


A. Hopping
B. Skipping
Forces (N)
Moment (Nm)
Angular Accel. (rad/s²)
Linear Accel. (G)
Time (s)
Fx
Fz
Mocy
AxRot
AyRot
AzRot
AxLin
AyLin
AzLin
125
0
-125
-250
20
10
0
-10
200
0
-200
-400
4
2
0
-2
-4
-6
0
0.5
1
1.5
2

C. Falling Into a Chair



D. Running with an Abrupt Stop

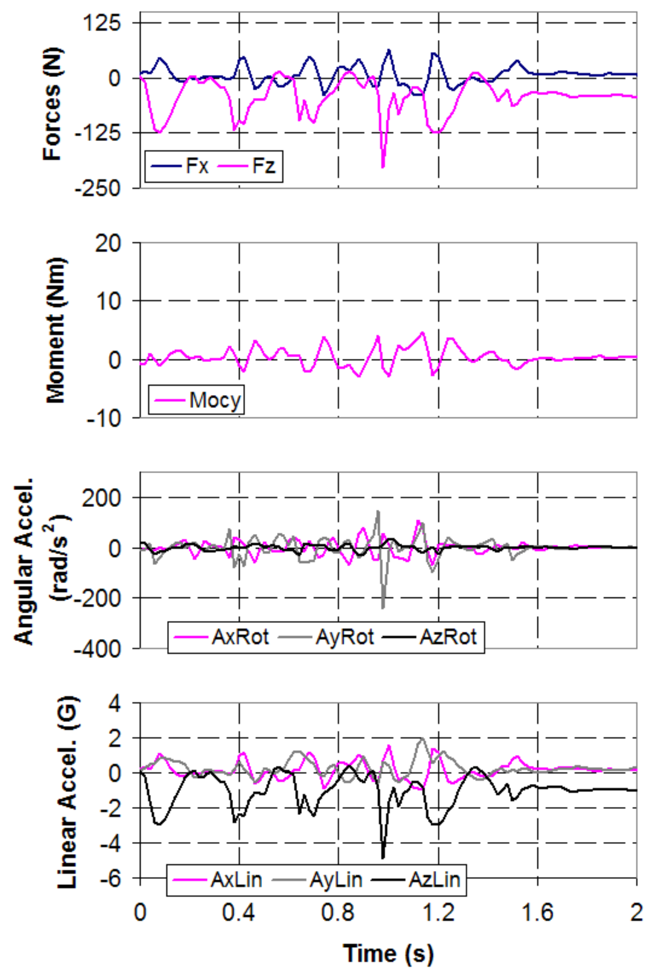


Figure A3. Representative plots of calculated forces and moments at AOJ, and head rotational and linear accelerations at head COM for the four vigorous activities: A, hopping; B, skipping rope, C, falling into a chair, and D, running with an abrupt stop.