

Neck Loads in Playground Activities in a Pediatric Population

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ABSTRACT

Neck injury assessment reference values (IARVs) and tolerance values for children have been specified using animal data compared to the loading of anthropomorphic test devices (ATDs). However, there is a paucity of data regarding the neck loads generated during non-injurious situations for children. Six males and six females aged 8-11 years old were equipped with a validated head sensor package and upper neck loads and moments were calculated from measured head kinematics while performing a series of playground-type activities. The maximum forces were 686 N in compression, 177 N in tension, and 471 N in shear, the maximum moments were 18.2 N-m in flexion, 6.0 N-m in extension, 6.4 N-m in lateral bending, and 12.1 N-m in axial twist. Female subjects exhibited similar loads and moments to their male counterparts, but larger N_{ij} values. The peak loads measured in this study are larger than or comparable to those undertaken with adult subjects participating in similar activities. These data provide details of the non-injurious upper neck loads that occur in pediatric subjects.

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INTRODUCTION

Neck injuries are an important issue in the pediatric population. While cervical spinal cord injuries are rare in the pediatric population, these injuries are associated with a high mortality rate [1, 2, 3, 4]. Spine trauma results in debilitating physical outcomes and substantial social and economic costs. Motor vehicle crashes are the leading cause of death and a common cause of cervical spine injuries in young children [2,5,6,7].

As demonstrated in numerous studies, children are not miniature adults. The anatomy and anthropometry of the young pediatric population are different than the adult. As an example, the mass and body proportions of a child differ from an adult, and continuously change throughout childhood [8-9]. Other differences between children and adults include increased cervical spine ligament laxity, unossified bone, decreased facet angle, size of the cervical spine, as well as differences in muscle strength, coordination, and behavior [10]. These factors affect the tolerance of the pediatric neck compared to the adult.

To investigate tolerance limits for the population, it is important to understand the mechanical loads during injurious and non-injurious situations. Investigations using animals and post mortem human subjects (PMHS) have been conducted to examine pediatric injury mechanics. Injurious neck loads have been examined using animal data, but few

studies have used PMHS to investigate pediatric injury biomechanics [12,13, 14,15, 16,17,18,19,20]. Because of the absence of data for pediatric cadaveric data, neck injury assessment reference values (IARVs) and tolerance values for children have been specified using animal data compared to the loading of anthropomorphic test devices (ATDs) [21,22,23,24]. However, there is a paucity of data regarding the neck loads generated during non-injurious situations for children.

In vivo testing of children during non-injurious everyday activities is feasible and avoids some of the limitations of other methods. ATD-based measurements are sensitive to biofidelity, instrumentation, and component design. Post-mortem tissue is difficult to obtain and muscle forces are difficult to incorporate. Previous studies have investigated the head accelerations experienced by pre-adults and adults in daily activities [25, 26, 27, 28, 29, 30, 31, 32, 33, 34, 35, 36, 37, 38], and a few also calculated neck loads. Adult volunteers have been used to characterize the non-injurious loads of the neck [39, 40, 41, 42]. Head trajectories and normal range of motion of the neck of child volunteers has been described in previous studies [43, 44, 45, 46]. Non-injurious head accelerations sustained by pre-adult participants in playground activities have also been quantified [47]. A recent paper has reported neck loads in children in frontal sled tests [48]; however, neck loads in everyday, non-injurious activities have not been characterized in children.

The purpose of this study is to measure non-injurious neck loads found in typical childhood activities for subjects 8 to 11 years of age. This study reports in-vivo neck kinetics from pediatric subjects during various activities using data recorded from a head sensor package.

METHODS

NECK LOADS OF CHILDREN

Head acceleration data was used from a previous study of everyday activities in child volunteers [47]. To summarize the methods, six male and six female pediatric (ages 8 to 11 years old) participants performed activities chosen for this study; none of the subjects knew any of the experimenters or had any interest in the outcome of our work. Upon arrival, each participant and their guardian listened to a standardized description of the activities and goal of the study, and provided their consent to participate and be videotaped. The participants were allowed to withdraw from the tests at any time and for any reason. An Institutional Review Board at Exponent approved the test procedures. The experimenters measured the participants' height, weight, head circumference, neck circumference, vertex-to-occiput height, and the height of the vertex to the T1 spinous process, and fitted the participant with the head sensor package.

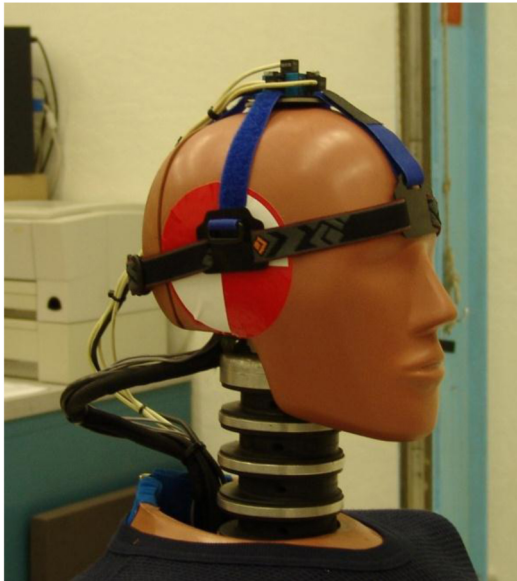


Figure 1. Head sensor package equipped on an ATD

Channel orientations were in accordance with the SAE J211 standardized ATD head coordinate system: x-axis positive forward, y-axis positive right, z-axis positive down [49]. 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 sensor array weighed 49 g (1.7 oz). The complete apparatus, including

head sensor, cabling, and waist-mounted data recorder, weighed approximately 800 g (28 oz).

The data acquisition system was capable of sampling at frequencies up to 100 kHz. Data were collected for 60 seconds for each activity. A flexible cable ran from the sensor package to the data logger along the subject's back. Each participant verified that the placement and attachment of the cable did not impede normal head motion relative to the torso.

Each participant voluntarily performed eight activities:

1. jumping as high as possible
2. running and jumping
3. skipping rope
4. swinging on a swing
5. spinning on a tire swing
6. riding a zip line
7. sliding down a slide
8. climbing on monkey bars or rings

Details of these activities can be found in Bussone et al. [47]. Jumping as high as possible involved three attempts, however the subjects chose to jump vertically as high as they were able. Running and jumping involved three attempts, however the subjects chose to jump as far as possible after a running start. Skipping rope involved the subjects attempting to jump for approximately 25 successful jumps, with as many consecutive successful jumps as possible. These jumping activities (vertical jump, run and jump, and skipping rope) were conducted with the subject on an asphalt sidewalk. Swinging on a swing involved the subjects receiving a push from a parent or guardian, and swinging at whatever height they were comfortable for a period of approximately one minute. Spinning on a tire swing involved the subjects receiving a twisting start from a parent or guardian, and spinning at their own discretion for approximately one minute, or until they wished to stop. Riding a zip line involved running and jumping onto a free-hanging handle, and riding a trolley device. The zip line was ridden down the length of the line, and then back. The subjects rode down an undulating slide for the slide task. Climbing on monkey bars or rings involved the subjects jumping or being lifted onto the first bar or rings, and attempting to traverse a gap as far as they were able, and then dropping down to the ground. Photographs and measurements of the equipment are provided in the [Appendix](#) as [Figure A1](#).

The subjects were asked to perform the tasks as they would normally, and each task was recorded using a real-time digital video camera recording at 29.97 frames per second. Subjects were cautioned not to jump from the swings during their dismount to prevent snagging of the cables or waist pack. The subjects' parents or guardians were asked to give the subjects an initial push on the swings and an initial spin

on the tire swing, because most participants could not touch their feet to the ground while seated on the swings. Before each activity, the subjects were asked to perform a set of head motions (rotating the head in the sagittal, axial, and coronal planes) that was used to synchronize the data with the video and to verify the relative alignment of the sensors. When jumping rope and during the two swinging activities, the participants were prompted to stop before the data collection ended. All other activities were completed by all the subjects in less than 60 seconds.

The subjects were contacted three weeks after testing. Responding subjects were asked whether they had soreness or pain in their head, neck, or body immediately after, within one day, and within three weeks after the tests.

DATA PROCESSING

During testing, each channel was low-pass filtered using a 400 Hz low-pass anti-aliasing filter, then digitally sampled at 1000 Hz. All data processing was completed using Matlab (The MathWorks, Inc., Natick, Massachusetts).

Appropriate frequency cutoffs for each channel of data were analyzed on a per task and per subject level by an automated routine in accordance with Winter's residual analysis method, as described in Bussone, et al (2010b) [50]. At rest, all linear accelerometers measured zero output except the z-axis linear accelerometer, which measured +1 g. The linear acceleration and angular velocity and acceleration data were used to determine the linear accelerations at the approximate center-of-mass (COM) of the head using the three-axis vector [equation \(1\)](#).

$$a_{COM} = a_{Sensor} + \alpha \times r_{COM-Sensor} + \omega \times (\omega \times r_{COM-Sensor}) \quad (1)$$

The distance from the sensor to the COM of the head was estimated using the anthropometric measurements and regression formulae given in Zatsiorski and Selujanov [51], where COM is the head COM, a_{COM} is the linear acceleration at the COM, a_{Sensor} is the linear acceleration measured by the sensor, α is the angular acceleration calculated from the sensor, $r_{COM-Sensor}$ is the vector from the COM to the sensor center, and ω is the angular velocity of the head measured by the sensor. Each variable indicates a three-axis vector quantity.

NECK LOAD CALCULATION

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 condyle, and that gravity was accounted for within the measured linear accelerations ([equations 2 and 3](#)). The distance from the head COM to the atlanto-occipital junction (AOJ) was calculated by normalizing the measured vertex-to-occiput and head circumference lengths to the Hybrid-III 50th male ATD length. Anterior-posterior head COM-to-occiput distance was

calculated using normalized circumference, and inferior-superior head COM-to-occiput distance was calculated using normalized vertex-to-occiput lengths. Head masses and moments of inertia were also scaled from Hybrid-III 50th male ATD values, according to [equation 4](#) and [equation 5](#), using inertial values and reference geometry and scaling methods from the literature [52-53]. Head circumference was used as the x- and y-axis characteristic length, and vertex-to-occiput length was used as the z-axis characteristic length. The measured and calculated values for the Hybrid-III 6-year-old and Hybrid-III 10-year-old are consistent with these methods.

$$F_{AOJ} = m_{head} * a_{COM} \quad (2)$$

$$M_{AOJ} = I_{head} * \alpha_{head} + F_{AOJ} \times d_{COM-AOJ} \quad (3)$$

$$\lambda_m = \lambda_l^3 = \lambda_c \times \lambda_c \times \lambda_z \quad (4)$$

$$\lambda_l = \lambda_l^5 = \lambda_m \times \lambda_l^2 \quad (5)$$

The 3-year-old was used as the IARV reference point, because the N_{ij} risk curves were originally scaled from this age to other age groups. N_{ij} values were calculated for each activity on a per-subject basis, using subject-specific thresholds normalized based on Mertz et al. [24], with the 3-year-old in-position thresholds as the reference values. In-position values were chosen because all activities were performed with the subjects aware and braced. Scaling factors for failure stress for each subject was linearly interpolated as a function of subject age, using data in Mertz et al., and the characteristic length scale was as a ratio of subject neck circumference to the listed 3-year-old neck circumference.

STATISTICS

The significance of differences in anthropometry and biomechanical responses was determined using the standard *t*-test ($\alpha=0.05$, $p \leq 0.05$ for significance), with assumption of equal variance subject to *F*-test ($\alpha=0.05$) to verify or refute homoscedasticity. One-sided tail distribution was assumed with a negative *F*-test, and two-sided distribution was assumed with a positive *F*-test. Tests were performed using the standard analysis package in Microsoft Excel.

RESULTS

NECK LOADS OF CHILDREN

The participants exhibited a range of ages (from 8 to 11 years old, mean 10 years old), heights (1.27 to 1.52 m, mean 1.41 m), and masses (25 to 43 kg, mean 32 kg). Subject

anthropometry and calculated anthropometric variables are reported in the [appendix](#)- (Tables A1-A2).

Across all subjects and all activities, a maximum axial force of 686 N, a maximum flexion-extension torque of 18.2 N-m, and a maximum N_{ij} value of 0.37 were observed (Figures 2-3). The maximum resultant force was 726 N and the maximum resultant moment was 18.8 N-m (Figures 4-5). The largest compression forces, flexion moments, and N_{ij} occurred during landing activities, such as running and jumping, jumping rope, and landing from the monkey bars. Means and maxima across all activities are shown in Figures 4, 5, 6. Data in the swing task for subject 3 are not plotted, as such 3 jumped off the swing, contrary to instructions. Data for this subject were not the maxima for forces, moments, or N_{ij} for the swing task. Cumulative N_{ij} values are shown in Figure 7.

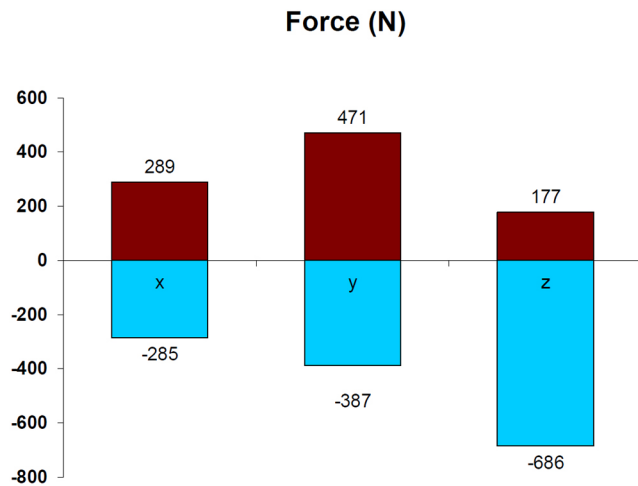


Figure 2. Maximum forces by direction. Forces are positive in forward shear (x), left lateral shear (y), and tension (z).

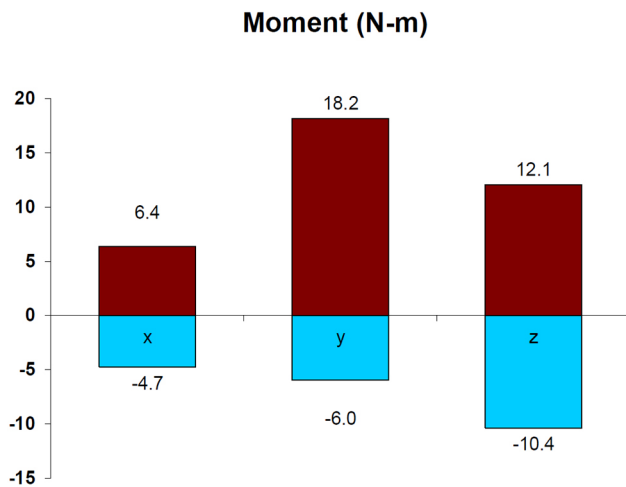


Figure 3. Maximum moments by direction. Moments are positive in left roll (x), flexion (y), and left yaw (z).

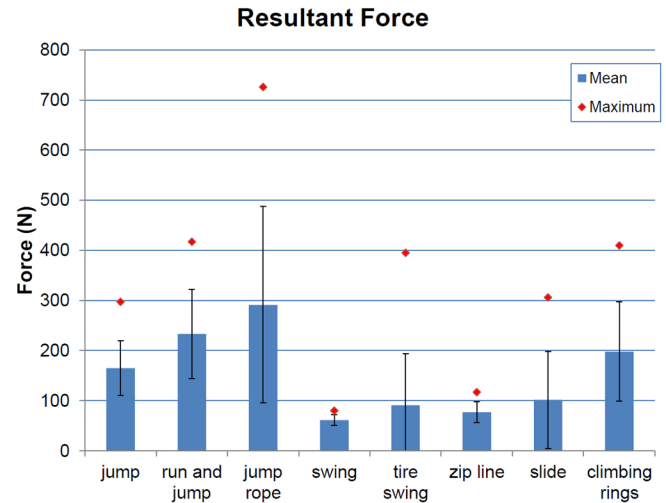


Figure 4. Mean and maximum resultant force. Results shown are mean $\pm$ standard deviation. Maxima are shown as red diamonds.



Figure 5. Mean and maximum resultant moment. Results shown are mean $\pm$ standard deviation. Maxima are shown as red diamonds.

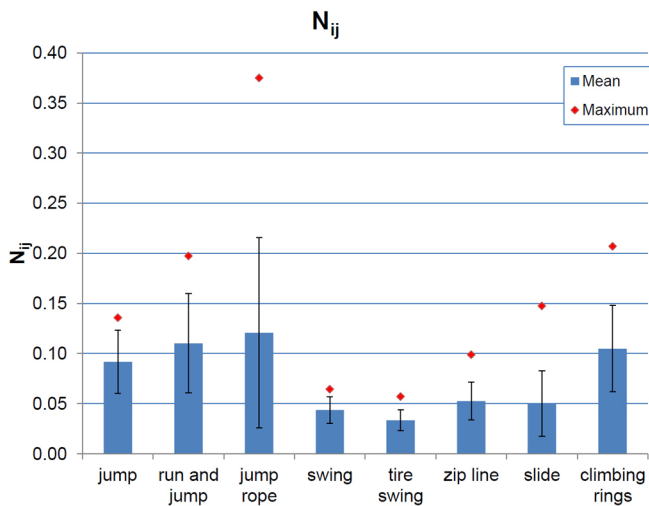


Figure 6. Mean and maximum N_{ij} . Results shown are mean $\pm$ standard deviation. Maxima are shown as red diamonds.

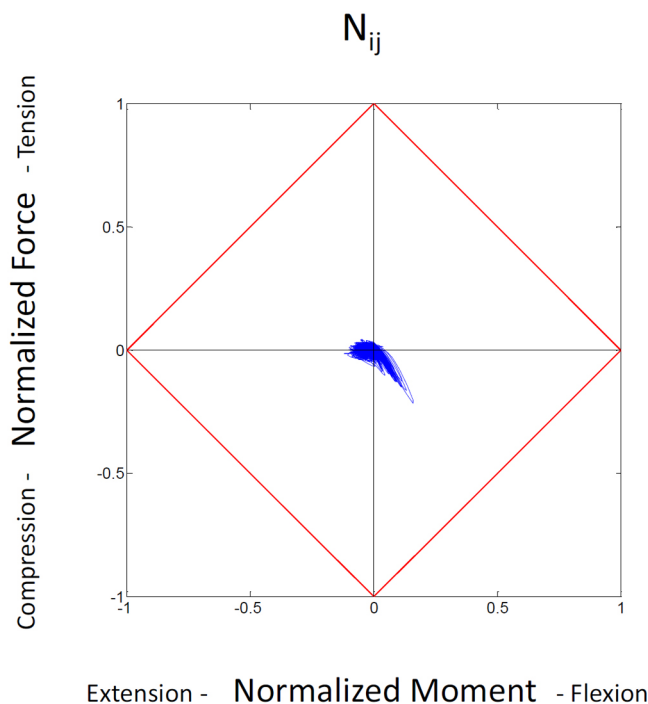


Figure 7. Cumulative plot of N_{ij} across all subjects and all activities.

The largest axial forces and flexion-extension moments were observed in the jump rope activity. The largest neck loads occurred during jumping activities, such as running and jumping and jumping rope (Figures 4-5). Data for each activity data are shown in Tables A3, A4, A5.

Certain data records were influenced by brief contacts between the head sensor array and objects in the test environment, such as the jump rope or the chains of the tire swing. These records were detected by iteratively identifying

outliers in the data (two standard deviations from mean), and reviewing the data and video recordings for the causes, such as head contacts. Data associated with head contacts were not considered in this analysis. If the effect on the data from a removal of a portion of the signal generated new outliers, those new outliers were also reviewed.

Data were compared by gender using t -tests ($\alpha=0.05$, $p \leq 0.05$ for significance), for each activity. Positive and negative neck moment maxima were not significantly different between male and female subjects. T-statistics between males and females for all activities are shown in Table A6. Females demonstrated significantly larger neck compression values in the jumping, running and jumping, and climbing rings tasks. Maximum N_{ij} was significantly higher in female subjects in the jumping, swinging, and zip line tasks.

Two female subjects climbed on monkey bars as opposed to climbing rings. The climbing apparatus was changed starting with subject 3 when it was found that the bars became too hot to hold under direct sunlight. The results from monkey bars and climbing rings were combined and reported under the climbing ring activity. The maximum neck loads during the climbing rings and zip line activities often occurred during the dismount phase. The maximum neck loads during the swing activity most often occurred at the peak amplitude of the swing cycle. However, one subject ignored the experimenters' instructions and jumped from the swings. That subject generated the largest resultant linear and angular accelerations for the swing activity (120 N, 4.8 N-m), which coincided with the landing from the jump.

One subject reported transient dizziness immediately after the tire swing activity, which resolved prior to performing the next task. None of the subjects reported any head or neck pain during the testing. Of the nine subjects who could be contacted three weeks after the testing, none reported any head or neck pain within that three week time period.

DISCUSSION

The present understanding of neck injury thresholds for children is limited by a lack of data on tolerable levels of loading. The purpose of this study was to better understand the neck accelerations experienced by children during non-injurious activities. To this end, upper neck kinetics of children were measured during vigorous activities. Discussion of head kinematics for these subjects can be found in Bussone, et al. [47].

Ideally, the anthropometric and inertial properties of the subjects' heads would be measured directly, or via finite element analysis of medical imaging data with assumptions made for material masses and densities, as inverse dynamics processes using scaled inputs from standard anthropometric datasets can exhibit large error rates [52-53]. However, as direct measurements require post-mortem decapitation, such data for volunteer studies are unavailable. As the present study used healthy volunteers in an otherwise normal setting,

ethical obligations prevented the use of radiological scanning techniques to create individual finite element arrays. However, scaling from ATD values instead of directly measuring or digital recreating individual inertial data was not debilitating, as the purpose of the present study was to compare to similar data generated using ATDs. Also, as noted in Loyd et al., the Hybrid-III 6-year-old and 10-year-old exhibited head contours similar to CT-scanned cadavers of equivalent ages, and the volumetric scaling techniques used to scale between dummies, and to generate inertial data in the current study, also correlated well with the cadavers. The masses, I_{yy} moment of inertia, and cg-occiput offsets of the current subjects were comparable to similarly-aged subjects in Loyd et al and Seacrist et al [48,55].

The results of the present study demonstrate that children can generate non-injurious neck loads more than one-third as large as IARV values (maximum axial force of 686 N, a maximum flexion-extension torque of 18.2 N-m, and a maximum N_{ij} value of 0.37). Averaged across activities and participants, the forces and moments are also similar to or larger than values reported in previous studies involving adults [32,37,39, 40, 41, 42].

The largest neck loads were observed in the tasks involving jumping and landing (run and jump, jump rope, climbing on rings). In some instances, participants included jumping activities in non-jumping tasks. For example, the largest values during a swinging task occurred when a subject jumped off the moving swing. Our finding that the largest neck loads occur during "jumping" tasks corresponds well to previous adult data; Funk, et al., and Vijayakumar et al. [37,41] found that falling and jumping activities generated the largest non-voluntary values. Our data support the conclusion that single-axis neck forces greater than 400 N and moments greater than 10 N-m, and N_{ij} values greater than 0.1, are not unexpected during vigorous activities that include a jumping, landing, or falling component.

The largest compressive neck forces (686 N), flexion moments (18.2 N-m), and N_{ij} values (0.375) were generated by subject 4 during the jump rope activity. Traces of the loads and forces for this subject are included in the [appendix](#) as [Figures A2-A3](#). Subject 4 was 8 years of age and weighed 25 kg. Though she was the youngest and lightest subject, her neck loads, independent of head or neck size, were some of the largest calculated. Unlike the other subjects, the maximum head accelerations and neck forces for subject 4 coincided with substantial head rotation during the landing phase of the jump. If this subject were excluded, the second largest compressive neck forces (429 N) and N_{ij} values (0.223), and third largest flexion moment (11.8 N-m) were produced by subject 6, who was the oldest female subject, and was, amongst the female subjects, above-average in height and weight. Subject 6 was notable for jumping higher than the other subjects when jumping rope. Peak loads during the jump rope activity are likely dependent upon the

proficiency and experience of the subject in performing the activity.

Normalization of the subject data can be accomplished using scaling methods [22]. Assuming constant density and geometric similarity, a scaling factor λ_c can be based upon neck circumference and a scaling factor λ_σ can be based upon age, used as a proxy for failure stress. In this manner, normalized force is equal to measured force multiplied by normalized failure stress and the square of normalized neck circumference, and normalized moments are equal to measured moments multiplied by normalized failure stress and the cube of normalized neck circumference. N_{ij} values are explicitly scaled during their calculation, and are already normalized. While the scaled multipliers varied considerably - 0.94 to 1.32 for forces and 0.86 to 1.52, with 1.00 corresponding to the Hybrid-III 10-year-old - normalized forces were not correlated to subject size. The subjects that generated the highest peak neck loads (subject 4 and 6) were slightly smaller than the mean and had scaling factors between 0.91 and 1.00. The subjects with the largest scaling multipliers (subjects 5 and 8) did not demonstrate raw or normalized neck loads that were substantially different from those of the other subjects. The data do not suggest a correlation between peak acceleration and subject size.

Neck load tolerance can also be compared between the adult and pediatric population on the basis of age (as a proxy for failure stress) and neck circumference [24]. According to the scaling methods published in the literature, equivalent neck forces for the adult are 137-201% of the 8-11 year old, and equivalent neck moments are 159-282% of the 8-11 year old, based on the larger and slightly stronger neck of adults. In other words, the adult neck is one-and-one-third to twice as strong as the 8-11 year old neck in resisting forces, and one-and-one-half to three times as strong in resisting moments or torques. The calculated values from children in compression and flexion, when scaled to adult loads, are larger than the recorded adult tolerable values [32,37,39, 40, 41, 42].

With the exception of compressive forces during jumping activities, load data for male subjects did not differ significantly from data for female subjects. This is likely because at the subjects' ages, 8 to 11 years old, there is little difference in stature, weight, and other anthropomorphic measures between males and females. In general, gender was not found to be a factor in determining load maxima in this study.

The converse was true for N_{ij} . Because loads were similar across gender, but because the male subjects tended to have larger neck circumferences, the female subjects had larger N_{ij} values. All activities with the exceptions of running and jumping and sliding down a slide exhibited differences between males and females at least approaching significance ($p \leq 0.1$). We observed some other differences between males and females. For example, male subjects produced larger x-axis accelerations, but lower z-axis accelerations

during the run and jump task, which reflected a greater tendency to look down when landing [47]. However, while females did exhibit larger compression loads, peak shear loads did not differ between genders during landing. Another area where gender may have played a role was in the jump rope task. The largest female x-axis linear accelerations and y-axis angular accelerations were larger than the largest male accelerations. Review of the video footage showed that the female subjects had greater head motion during the landing phase [47]. This is reflected in the larger shear forces, axial forces, and moments in the female subjects. The male subjects had significantly larger neck circumferences, and lower head to neck circumference ratios. This is a similar finding to that of Stemper et al [56], and may explain the extra motion and increased loads in female subjects during the more energetic tasks such as jump rope and running and jumping. That paper noted that female subjects, when matched for seated height or head circumference with male subjects, had smaller cervical spine cross-sectional areas than the males and thus potentially lower spinal stiffness independent of other differences, such as muscle mass and ligamentous material properties. However, it was also noted that female subjects had shorter necks and smaller lever arms, which would tend to increase spinal stiffness, and potentially confounded by male populations that varied in an internally inconsistent manner from the female subjects. Regardless, it is unclear what effect potential differences in adult spinal geometry have on an 8-11 year old population that is substantially pre-pubescent. It is noteworthy that task familiarity may have also played a role. Generally, the female subjects appeared more adept at jumping rope than the male subjects. Male accelerations were somewhat lower on the climbing rings task as well. This is likely because more male subjects than female subjects successfully climbed across all the rings, and therefore did not drop to the ground during this activity and did not generate the typically higher accelerations that occur during a landing.

The magnitude of the peak loads, as in any analysis of self-generated kinetics, is dependent on the effort put forth by the subjects. Effort level is a known limitation during the running and jumping tasks [27,33,47], where inter-subject effort was notably variable and could to some extent be observed during the testing. Although effort was not explicitly quantified or directly measured, based upon qualitative observations during testing, there was far less effort variability when measuring jump rope accelerations, as that task had a minimum performance requirement necessary to clear the rope. It is important for future studies to consider attempting to regulate effort level via instruction, repetition, competition, or by choosing activities that require a certain level of effort to perform successfully.

Several other research groups have also measured neck loads during a variety of everyday non-injurious activities in adults. Vijayakumar et al. and Mkandawire, et al. [32,37] performed a similar study in adult subjects, comparing six-axis head and neck loads between daily living tasks and rear

and lateral impacts in bumper car riders. Vijayakumar reported bumper car impacts with average peak moments ranging as high as 9.8 N-m in extension and 6.7 N-m in flexion, and average peak forces as high as 139 N in shear and 93 N in compression. They also reported everyday activities with average peak moments ranging as high as 6.5 N-m in extension and 10.6 N-m in flexion, and average peak forces as high as 136 N in shear and 213 N in compression. Mkandawire reported for lateral bumper cars mean peaks of 13.3 N-m with a maximum of 29.5 N-m in extension, a mean of 5.8 N-m with a peak of 14.8 in flexion, a mean of 8.5 N-m with a peak of 36.2 N-m in lateral bending, maximum shear loads of approximately 480 N, and maximum compression and tension values of approximately 430 N. Funk et al. [41] reported compressive loads of up to approximately 400 N in compression, 15 N-m in flexion, and 15 N-m in extension for a variety of soccer ball heading, seat tip, and fall into chair activities.

Mertz and Moroney reported static tolerance maxima [39-40,42]. Mertz reported a static compressive and tensile tolerance of approximately 1100 N, a maximum flexion of approximately 32 N-m, and a maximum extension of approximately 15 N-m. Moroney reported a maximum flexion of approximately 10 N-m, and a maximum extension of approximately 25 N-m.

Seacrist et al. reported head angular accelerations, axial and anterior-posterior shear upper neck forces, and fore-aft bending moments for children and adults in approximately 2 m/s change in velocity (ΔV), 3.6 g sled decelerations [48]. The sled decelerations resembled a low-speed bumper car impact [45,48]. They found that angular accelerations decreased and My bending moments decreased with increasing age and decreasing head-to-neck ratio. Upper neck shear forces increased and upper neck axial forces decreased with age. The authors reported axial forces of 106 ± 20 N for 6-8 year olds and 72 ± 20 for 9-11 year olds, shear forces of -133 ± 13 N for 6-8 year olds and -128 ± 20 N for 9-11 year olds, and bending moments of 7.1 ± 1.0 N-m for 6-8 year olds and 8.2 ± 1.2 N-m for 9-11 year olds. This paper did not report maxima. The averaged values of the 6-11 year olds are similar to those in the jumping tasks (jump, run and jump, jump rope) during the present study. This is similar to the rear bumper car analysis of Vijayakumar for adults, which found that the neck forces and moments in bumper car impacts were similar to more vigorous activities of daily living [37]. It is not surprising that a similar observation holds true for children as well.

The maximum forces and moments measured during the present study were comparable to the reported values in previous everyday activity studies. It is difficult to compare head accelerations across studies because of differences in activities, instrumentation, filtering, and effort level. The present study differs from prior studies in that the playground-type activities involve much more running, jumping, and landing than the vehicular, office task-based, or treadmill-based activities typically performed by volunteers