

Implementation of Child Biomechanical Neck Behaviour into the Hybrid III Crash Test Dummy

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ABSTRACT

This research focuses on comparing the biomechanical response of the head and neck of the Hybrid III 3-year-old anthropometric test device finite element model and pediatric cadaver data, under flexion-extension bending and axial tensile loading conditions. Previous experimental research characterized the quasi-static biomechanical response of the pediatric cervical spine under flexion-extension bending and tolerance in tensile distraction loading conditions. Significant differences in rotational and linear stiffness were found between the Hybrid III model and the pediatric cadaver data. In this research the biomechanical child cadaver neck response has been implemented into the 3-year-old Hybrid III child dummy FE model. An explicit finite element code (LS-DYNA) and the modified Hybrid III model were used to numerically simulate the previous cadaver tests and validate the altered Hybrid III neck. Subsequent simulation of the child cadaver sled tests were completed with the child biomechanical test results implemented into the Hybrid III. The altered Hybrid III dummy illustrated a significant increase in flexion and distraction of the neck during the simulated sled test and a greater degree of chest deflection. The HIC evaluated using a 36 ms window for the altered Hybrid III dummy was observed to be 268 while the unaltered child dummy observed a HIC value of 194 evaluated using the same 36 ms window. The upper neck resultant forces computed from the simulation from the altered Hybrid III were observed to be 2.3 kN which, based upon the pediatric biomechanical response, would indicate tissue failure in this vicinity of the cervical spine.

INTRODUCTION

The National Highway Traffic Safety Administration (NHTSA) has found that motor vehicle accidents are the leading cause of death for persons aged 3 through 34

[1]. More specifically, children in the 3 to 4 year age group account for a greater number of passenger fatalities (45.5 percent) than any other age group [2]. Clinical observations documented from the Crash Injury Research Engineering Network (CIREN) calculated that motor vehicle accidents (MVC's) account for 50 percent of all pediatric trauma and 30-40 percent of pediatric cervical spine injuries [3, 4, 5, & 6]. The most common body part injured for children involved in motor vehicle frontal impact crashes is the head and the neck, followed by injuries of the torso and extremities [7]. For children under 11 years of age, motor vehicle accidents account for 38 percent of all cervical spine injuries [8]. Approximately 50 percent of injuries to the pediatric cervical spine sustained in MVC's are fatal [5]. In addition, injuries to the head and neck account for a greater number and severity of abbreviated injury scale (AIS) 2+ injuries [9]. When ranked by specific age, motor vehicle accidents are the leading cause of death for children aged 3 years old [1].

Anatomical, developmental and physiological differences between children and adults place children at an increased risk of injury as compared to adults in MVC's [10, 11, 12, & 13]. The ligaments of the pediatric cervical spine are underdeveloped lax and do not effectively contribute to the structural integrity of the cervical column [14]. The cervical vertebrae are not yet completely ossified making them more prone to separation. The facets are predominately horizontal which provide little restriction to dislocation and the posterior-lateral contours of the vertebral bodies are not developed and therefore can not restrict flexion-rotation forces as experienced in frontal MVC's [15, 16, 12, & 13]. In addition, the disproportionate size of the child's head relative to the rest of the body lends to cervical spinal injury. The ratio of head to body mass is 1:3 for newborns and 1:15 for adults. Therefore, the mass of a child's head is proportionately greater than that of an

adult thereby increasing the relative inertia in high acceleration environments such as in MVC's [17]. The relatively large head and shorter neck of children work in concert placing the fulcrum of the cervical spine within the Occ-C2 complex increasing the risk for Occ-C2 dislocation injuries [14]. As a result of these biomechanical factors, 60-80 percent of all pediatric vertebral injuries are in the cervical region. These factors in combination with the under developed musculoskeletal pediatric cervical spine put children at an increased risk for cervical spine injuries in MVC's.

Mechanisms of neck injury include extreme hyperflexion which leads to rupture of the tectorial membrane and separation of vertebrae which leads to atlantooccipital dislocation (AOD) [18]. Separation occurs in the immature pediatric cervical spine due to the column not yet having formed a contiguous interlocking structure through the normal processes of growth and development such as ossification and geometric changes. This structural inadequacy in high axial tension-forward flexion environments as experienced in MVC's damages the spinal cord by placing it in a state of tension or complete transection [9]. The combination of tension and forward flexion is an injury mechanism for Occ-C2 dislocation injuries. As a result, Occ-C2 dislocations occur more frequently in children due to their relatively small occipital condyles, ligamentous laxity and flat articulation between the occiput and the atlas.

Research and development in the field of child occupant crash protection relies heavily on the biofidelity of anthropomorphic test devices (ATD's) and the ability of researchers to relate measured parameters on the ATD to injury [19]. In previous studies comparing the head-neck responses of the Hybrid III adult-male ATD and adult human cadaver, the neck of the Hybrid III adult-male ATD was found to be between 2 to 4 times stiffer and transmitted approximately 48.5 percent more force to the lower neck under axial compression as compared to the adult human cadaver [20 & 21]. In addition it has been shown that the thoracic spine of the Hybrid III six-year-old ATD is overly stiff and results in high neck forces and moments that are not representative of the true injury potential [22].

Previous researchers have found the response of the Hybrid III dummy to overstate the stiffness of the human cervical spine under compression loading conditions. No research has made comparisons to stiffness of the Hybrid III dummy neck under axial tensile or flexion loading conditions. There have been no documented studies comparing the stiffness of the Hybrid III 3-year-old dummy finite element model neck to human pediatric cervical spine cadaver data. In addition, precise failure tolerances for the neck of the Hybrid III dummy and model are lacking and attempts to incorporate the material properties of the human cervical spine obtained from recent pediatric cadaver research into the Hybrid III three-year old finite element model have not yet been made.

Previous research suggests that there is a strong need for improving the biofidelity of the neck of the Hybrid III 3-year-old child ATD model. Therefore, the specific purpose of this research is to compare the kinematic neck response of the Hybrid III three-year-old child finite element model with known real-life pediatric cadaver data. The biomechanical response of the human cervical spine under flexion and tensile loading conditions has recently been provided through human pediatric cadaver research [23]. It follows that this new invaluable information on the biomechanical response and tolerance of the pediatric cervical spine be compared to that of the Hybrid III 3-year-old child finite element model and be incorporated in the child models design to improve its biofidelity and usefulness as a research tool. With more life-like child surrogates, researchers will be better able to evaluate the efficacy of child restraint systems and other interventions to mitigate the risk of injury and death.

EXPERIMENTAL METHODS

HYBRID III THREE-YEAR-OLD CHILD DUMMY

Experimental sled testing of the Hybrid III 3 year old child dummy was completed at Graco Corporation's sled testing facilities in accordance with Federal Motor Vehicle Safety Standards (FMVSS) 213. The Hybrid III 3-year-old dummy was procured from Denton ATD, Inc. and was equipped with three uniaxial accelerometers in the head, chest and pelvis. The accelerometers were rated for a capacity of 2000 g's. The sampling rate for each channel was 10 kHz [24].

The child dummy was restrained using a five point restraint system in a convertible forward facing Child Restraining System (CRS). The CRS was secured to the sled using the lower anchorage and tether for children (LATCH) system. In addition a mandatory top tether was also used that was fastened to the sled bench. The child was reclined at an angle of approximately 20 degrees to the vertical due to the decline of the CRS backing. Sled impact testing was completed at a speed of 11.6 m/s [25].

BIOMECHANICAL PEDIATRIC CADAVER NECK DATA

The experimental procedure to determine the biomechanical response of the pediatric cervical spine under bending and tensile loading was conducted by Ouyang et al., (2005) [23]. Briefly, cervical head and neck complexes were obtained from pediatric donors aged 2 through 12 years of age. Neck musculature was removed and the cervical spines were subjected to quasi-static tensile tests. A pure tensile load was applied by the test apparatus that allowed for full anterior-posterior translation and flexion-extension rotational degrees for freedom. The head-neck complex was subjected to a destructive test in which the displacement was increased at a rate of 5 mm/s until a 10 percent reduction in axial tensile load was observed. Axial tensile force data was collected via a load cell at the

base of T2. The average linear stiffness for tensile loading was calculated to be 34.7 N/mm with no statistically significant difference found for pediatric age.

PEDIATRIC CADAVER SLED TEST

Frontal impact sled tests utilizing pediatric cadaver specimens were conducted by Kallieris et al, (1976) [26]. The child cadaver was seated in a forward facing CRS restrained with a lap belt. In addition, a molded polystyrene block was positioned over the child's abdomen to protect the abdomen from lap belt injuries. The crash pulse was approximately trapezoidal in shape. From 0 to 20 ms the pulse ramped up to an approximate value of 17 g's. The acceleration remained relatively constant from 20 ms to 55 ms. At 55 ms, the pulse began a downward ramp to 0 g's at 80 ms. The acceleration remained at a value of 0 g's for the duration of the test. The trajectory of the head was measured from the centre of gravity of the head.

NUMERICAL METHODS

A brief overview of the numerical modeling technique is presented here. Further information can be found in reference [27]. The Hybrid III finite element model has been previously validated and excellent numerical predication capabilities were observed [24, 25 & 27]. The child seat was modeled using Computer Aided Design (CAD) surfaces provided by the Century/Graco Corporation. The child safety seat, as well as all other components of the numerical model including the seat buck, the CRS webbing and the CRS foam pad was meshed using Finite Element Model Builder (FEMB). The mesh of the child seat comprised of 12,728 nodes and 13,379 shell elements. The seatbelt was modeled to pass through a series of openings and channels of the CRS and to fit around the child dummy which had the same configuration as outlined in the FMVSS 213 standards [4]. The lower ends of the three portions of the seatbelt that passed through the child dummy's groin and wrapped around the child dummy's legs were connected directly to the child safety seat as in the actual CRS. The buckle in front of the dummy's chest was modeled to permit sliding motion of the webbing. The buckle at the waist of the dummy was modeled on the assumption that there was no relative movement between the waist buckle and the seatbelt. A foam pad inserted between the polypropylene shell and the seat fabric of the CRS was also incorporated into the FE model of the CRS. The material model was obtained from Turchi et al. (2004) [24]. Two parts, namely a rigid seat buck frame and a deformable seat buck pad, were created in to model the seat buck (vehicle seat bench).

UNALTERED HYBRID III THREE-YEAR-OLD MODEL

The Hybrid III 3-year-old finite element child dummy model used in this study was developed by First Technology Safety Systems (FTSS). The mass of the Hybrid III 3-year-old child dummy is 15.4 kg and the stature height is 94.5 cm. The FE model is completely

deformable containing 12,172 elements and 11,698 nodes. The FE mesh was unaltered employing the default values for the material, joints and element formulations. Detailed information of the Hybrid III child dummy can be found in reference [28]. Positioning of the model in the CRS was completed using FEMB.

ALTERED HYBRID III THREE-YEAR-OLD MODEL

Pediatric cadaver material properties were implemented into the neck region of the altered Hybrid III 3-year old child finite element model. The material model of the neck cable was changed from material type 1 (*MAT_ELASTIC) to material type 67 (*MAT_NONLINEAR_ELASTIC_DISCRETE_BEAM) of LS-Dyna. This material model is defined for simulating the effects of non linear viscous beams and permits the use of a load curve to define the translational forces and rotational moments for relative displacements. The load curve was obtained by averaging a sub set of five pediatric cadaver donors ages ranging from 2 to 4 years of age. Load deflection curves for axial tensile distraction loading were experimentally obtained by Ouyang et al., (2005) [23]. A cubic spline interpolation was performed using Mathcad for each subject to produce a polynomial function with continuous values for derivatives of displacement. The 5 curves were then fitted and averaged to produce an average load-deflection curve for the 5 subjects of interest which was used as the load curve for the material model in the altered model. In addition, the elastic bulk modulus and short term and long term shear modulus were changed from there original values of 0.030 Pa, 0.012 Pa and 0.0014 Pa to 0.425E-04 Pa, 0.239E-04 Pa and 0.237E-04 Pa respectfully.

Altered model head and neck component test analysis

Initial head and neck component numerical testing with the unaltered Hybrid III 3-year-old FE model revealed significant differences in stiffness of the neck. When subjected axial distraction loading, the unaltered Hybrid III 3-year old model was 435 times stiffer than the pediatric cadaver data yielding a maximum load of 185 kN at 13 mm displacement as compared to 425 N for pediatric cadaver data. The stiffness of the unaltered model for axial distractile tensile loading was calculated to be 14,413 N/mm as compared to 34.7 N/mm for the pediatric cadaver data. For flexion and extension loading, the unaltered model exhibited 168 percent and 128 percent difference greater moments for flexion and extension respectfully as compared to the pediatric data. Based on these findings, implementation of the pediatric cadaver data the Hybrid III 3-year-old FE model was warranted.

To validate the altered numerical model to the pediatric cadaver data, the experimental testing conducted by Ouyang et al., (2005) [23] was numerically simulated. Preparation of the FE model head and neck complex was achieved by importing the altered Hybrid III 3-year old child FE model into FEMB and disarticulating the

child model at the adapter plate. Head and neck component testing was performed for the Hybrid III 3-year-old altered and unaltered FE models under conditions similar to the axial distraction experimental procedure outlined by Ouyang et al. (2005) [23] to validate the kinematic response of the altered model. Results of axial distraction loading are illustrated in Figures 1a and 1b. A load versus deflection curve yields a maximum load at 13 mm displacement of 450 N for the altered model, 425 N for the average pediatric cadaver data and 185 kN for the unaltered model.

The altered model and pediatric data exhibited similar linear load-deflection profiles and the maximum values were within 6 percent difference of one another. Least Squares Linear Regression Analysis was utilized to compare the slopes (stiffness) of the altered and pediatric load-deflection curves. The resulting slopes (stiffness) were 34.5 N/mm for the altered model and 32.5 N/mm for the pediatric data. The numerically calculated stiffness for the altered model and the sub set of pediatric data used in this research are both in good agreement with the average stiffness calculated by Ouyang et al. (2005) [23] of 34.7 N/mm.

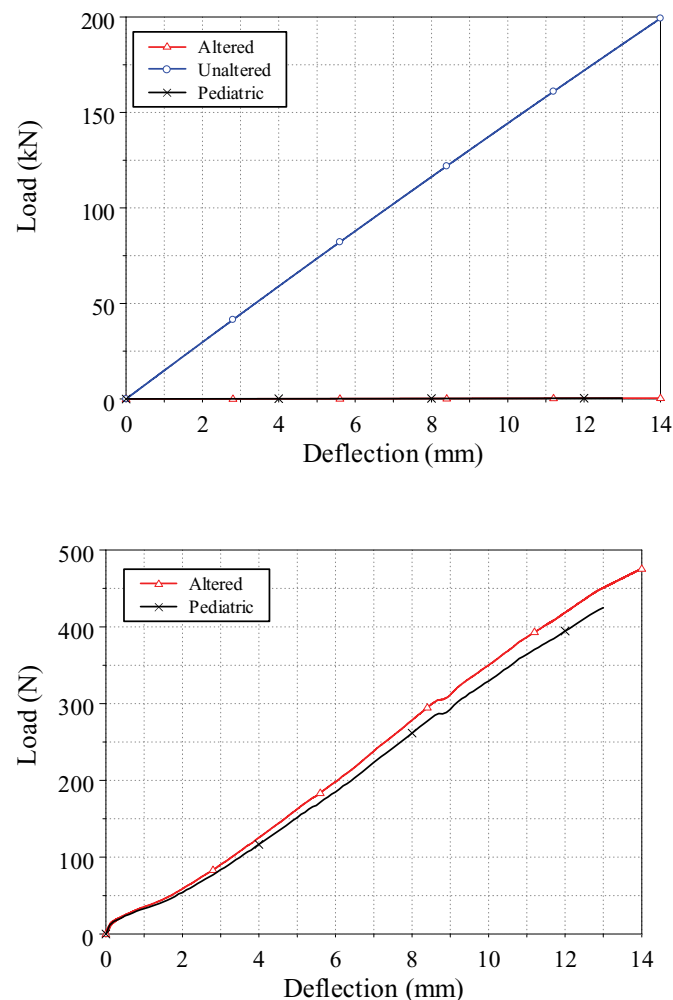


Figure 1a and 1b. Axial neck force load-deflection curve for altered unaltered and pediatric data.

Further head and neck component testing validation was performed for the Hybrid III 3-year-old altered FE model under conditions similar to the flexion-extension bending experimental procedure outlined by Ouyang et al. (2005) [23]. Analyses of the kinematic response under flexion-extension bending are illustrated in Figure 2. The unaltered model exhibited flexion-extension moments of 29.7 N·m for flexion and 11.4 N·m for extension. The altered model exhibited flexion-extension bending moments of 2.6 N·m for flexion and 2.4 N·m for extension. The experimentally obtained flexion extension bending moments for the pediatric data was 2.4 N·m for flexion and extension. The percentage differences for the altered model were 8 percent and 4 percent greater for flexion and extension respectively. In addition, the average rotational stiffness was calculated using Least Squares Regression Analysis for the altered model and was 0.04 N·m/deg which was identical to the calculated value of 0.04 N·m/deg for the experimental pediatric cadaver data as reported by Ouyang et al. (2005) [23].

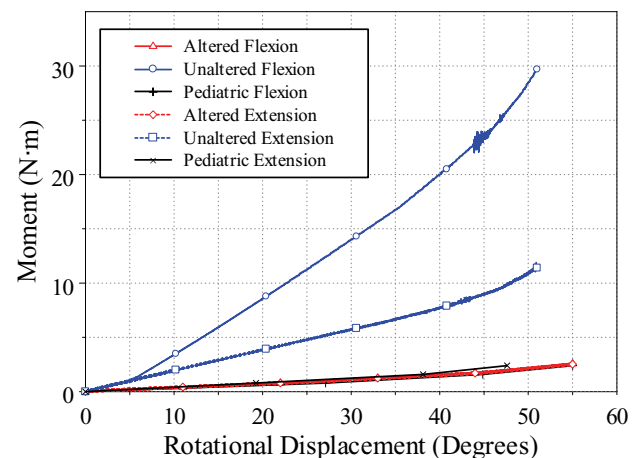


Figure 2. Angular neck moment load vs. rotational displacement curve for altered unaltered and pediatric data under flexion and extension bending.

Simulation procedure

A brief overview of the simulation procedure is presented here. For further details in regards to the numerical modeling of the CRS, tethers and the seat buck, the reader should consult reference [27]. Tightening of the front-adjusting harness strap was simulated quasi-statically to position the child dummy into the CRS. This process was completed prior to application of the acceleration pulse to the seat buck through dynamic relaxation. The number of iterations between the convergence checks and the convergence tolerance for dynamic relaxation were specified as 1000 and 0.0001 respectively. In this point in the simulation, the front-adjusting harness strap as well as the upper and lower tethers were preloaded. Loads assigned to the LATCH and the top tether in the dynamic relaxation phase were of a magnitude of 200 N and 50 N [27]. Federal Motor Vehicle Safety Documents respectively in the direction of the tethers.