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Head and Neck Response of a Finite Element Anthropomorphic Test Device and Human Body Model During a Simulated Rotary-Wing Aircraft Impact

A finite element (FE) simulation environment has been developed to investigate aviator head and neck response during a simulated rotary-wing aircraft impact using both an FE anthropomorphic test device (ATD) and an FE human body model. The head and neck response of the ATD simulation was successfully validated against an experimental sled test. The majority of the head and neck transducer time histories received a CORrelation and Analysis (CORA) rating of 0.7 or higher, indicating good overall correlation. The human body model simulation produced a more biofidelic head and neck response than the ATD experimental test and simulation, including change in neck curvature. While only the upper and lower neck loading can be measured in the ATD, the shear force, axial force, and bending moment were reported for each level of the cervical spine in the human body model using a novel technique involving cross sections. This loading distribution provides further insight into the biomechanical response of the neck during a rotary-wing aircraft impact. [DOI: 10.1115/1.4028133]

Introduction

Since WWII, rotary-wing aircraft have become a staple in all major military conflicts including the Korean War, Vietnam War, Gulf War, and the War on Terror [1]. Aviator training is a very time-consuming and expensive process, typically requiring 1.5–2.5 yr of preparation and costing U.S. taxpayers as much as \$2 million per pilot [2]. Due to the large investment required to produce these highly skilled pilots, as well as the obvious concern for loss of life or injury, it is very important that these individuals be protected in the event of crash. Occupant response during rotary-wing aircraft crashes have been studied through full-scale, sled, and surrogate component-level experimental testing [3–11]. In many cases, the head and neck response were the focus of these studies.

Bass et al. conducted several series of ATD and cadaver tests to study and to evaluate the effects of head-supported mass (HSM) on neck injury risk during experimentally simulated rotary-wing aircraft impacts [3,4]. The results from a variety of head-neck component and whole-body experiments were used to both further elucidate the overall neck response and to develop a new injury criterion, the beam criterion (BC), for the lower neck [3,4]. For the component tests, injury severity was found to be less sensitive to the location of the HSM as compared to the initial neck angle and the peak sled acceleration. However, a linear relationship existed between the length of the moment arm, from the HSM to the lower neck, and both the peak force and moment. Lower neck anteroposterior (AP) shear force was not found to be a significant factor in injury predication, unlike axial force and flexion moment. In the whole-body cadaver tests, the cervical spine experienced sigmoid bending after the initial impact, due to gross

shearing motion in the lower neck, followed by a transition into C-shaped, simple bending.

Paskoff and Sieveka conducted an extensive series of rotary-wing aircraft sled tests to study the effects of ATD size, helmet mass properties, and impact severity on neck response [10,11]. Testing included a 50th percentile male, 95th percentile male, and 5th percentile female Hybrid III ATD with varying HSM configurations and three different severity sled pulses. The HSM did not have a statistically significant effect on the magnitude of the neck forces or moments for a given ATD size. However, both ATD size and pulse severity were found to have a statistically significant effect on most neck loading parameters.

The primary goal of this study was to evaluate the head and neck response of the Global Human Body Models Consortium (GHBMC) FE human body model during a simulated rotary-wing aircraft impact. The forces and moments at each cervical level of the neck were captured using a novel technique previously described by White et al. [12]. The seat geometry, belt configuration, and sled pulse were based on one of the experimental sled tests conducted by Paskoff and Sieveka [10,11]. The boundary conditions of this simulation were first validated through simulation of the experimental sled test with an FE ATD model. The head, neck, and chest response of the FE ATD model were quantitatively compared to the reported experimental results.

Materials and Methods

Overview. In the current study, two low-severity rotary-wing aircraft impact simulations were conducted using LS-DYNA R4.2.1 (LSTC, Livermore, CA) to examine the head and neck response of an ATD and human body FE model. The sled pulse, sled configuration, seat geometry, and belt preload were based on the low-severity, 7.62 m/s delta-V, sled test (#530) conducted by Paskoff and Sieveka [10,11]. This particular experimental test simulated a 30 deg pitch-down helicopter ground impact using a

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Fig. 1 Positioning at the time of sled pulse initiation ($t = 70$ ms) for the (a) ATD and (b) GHBM simulations

50th percentile Hybrid III ATD with 1.8 kg of HSM proximate to the head center of gravity (CG). A generic five-point restraint system was then modeled according to established military aircraft personnel restraint design principles [13]. In order to simulate the experimental preloading of the restraint system, modeled pretensioners applied approximately 181 N of belt loading prior to any sled acceleration. To allow this initial belt tension to be achieved, as well as model settling due to gravity, both simulations were run for 70 ms prior to the initiation of the sled pulse (Fig. 1).

ATD FE Model. The Humanetics (Plymouth, MI) 50th percentile Hybrid III ATD FE model (v8.0) was used to validate the boundary conditions of the simulation by comparing the model and experimental responses. This FE model has been extensively validated on the component, subassembly, and whole ATD levels [14]. The original seated, driving posture of the ATD model was modified, increasing the neck extension angle by 15 deg and the thorax extension angle by 3 deg. The upper arms were rotated until they were in line with the torso and the lower legs were flexed until the feet were parallel with the sled floor. These modifications aligned the ATD model more closely to the experimental test initial position. Lumped mass elements were equally distributed around the head accelerometer mounting block, representing the HSM in the experimental test. Neck forces and moments were extracted from the upper and lower load cell transducers and transformed to the occipital condyle and base of the neck, respectively.

Prior to data analysis, the experimental data was zeroed and time shifted 70 ms, matching the start time of the simulation sled pulse. The time intervals for analysis were defined from 70 to 270 ms for acceleration and force data and 70–290 ms for moment data. These time intervals were chosen since they focus on the most relevant parts of the respective time histories. A quantitative comparison analysis was then conducted to compare the experimental results with the ATD simulation results using CORA, v3.6.1 (Partnership for Dummy Technology and Biomechanics, Germany).

The CORA analysis uses two different methods to quantitatively compare signals [15–17]. The corridor method rates, from 0 to 1, how well the subject curve fits within a narrow inner corridor and a wide outer corridor, both defined about the benchmark curve. For purposes of the current study, the subject curve is the simulation time history and the benchmark curve is the experimental test time history. A rating of 1 is assigned if the subject curve falls completely within the narrow corridor and a 0 if it falls completely outside of the wide corridor. The cross-correlation method analytically

analyzes the characteristics of the signal including phase shift, size, and shape using individual weighing factors. For each of these three submethods, a rating between 0 and 1 is also calculated. An overall, global rating is then calculated as the weighted sum of the corridor and correlations ratings, ranging from 0 for a poor match to 1 for a perfect match. While there is no definitive rating indicative of a “good” match, correlations with a rating of 0.7 or greater have been designated as a good match in past studies [16]. It is important to note that the CORA rating is dependent on the control variables within the program. Default global settings were used for both the corridor and cross-correlation methods.

Human Body FE Model. The GHBM seated, 50th percentile male FE model (v3.5) was used to simulate the experimental ATD sled test [18–20]. This state-of-the-art human body model was developed by an international team of research institutions to investigate injury to the human body during dynamic impact events. It has been validated on both segmental and full cervical spine levels [21–24]. Additionally, whole-body validation has been conducted in a number of different impact scenarios [25–28]. The model, as a whole, is composed of 1.3×10^6 nodes, 1.95×10^6 elements, and 847 parts. The neck model includes the seven cervical vertebrae, intervertebral discs, ligaments, and muscles. The cancellous bone is modeled with hexahedral solid elements and the cortical bone with shell elements; both with plastic-kinematic material behavior. The intervertebral discs include the annulus fibrosis and nucleus pulposus. The fiber layers of the annulus fibrosis are modeled with a series of shell layers and the ground substance between fiber layers with an Ogden-rubber solid component. The nucleus pulposus is modeled as a solid, visco-elastic material. Multiple 1D nonlinear rate dependent tension-only beam elements represent the ligaments. The facet joint articular cartilage is modeled as a superior and inferior layer of visco-elastic solid elements. The joint synovial fluid is simulated using a squeeze-film model. Twenty-six neck muscles are modeled with solid elements using an Ogden-rubber material, representing the passive response. Active response is simulated with corresponding embedded Hill-type 1D elements.

The LS-PrePost (v4.0, LSTC, Livermore, CA) mass trimming function was used to equally distribute the additional head mass about the skull while maintaining the original head CG. Unlike in ATDs, there are no load cells to directly measure neck forces and moments in the GHBM model since it is based off of the human anatomy. To capture this neck loading, cross sections were defined at each cervical level of the neck using a methodology

Fig. 2 Midsagittal cut of the GHBM head and neck with transverse cross sections indicated by the solid lines. The LCSYS for C1 and C7 is visualized, along with the cross section at the level of C5.

Table 1 Quantitative comparisons between the benchmark experimental data and the ATD simulation data

Transducer		CORA		
		Corridor	Correlation	Overall
Acceleration (magnitude)	Head	0.761	0.803	0.782
	Chest	0.826	0.849	0.838
Upper neck loading	F_x	0.736	0.903	0.819
	F_z	0.721	0.751	0.736
	M_y	0.589	0.818	0.704
Lower neck loading	F_x	0.578	0.577	0.578
	F_z	0.698	0.701	0.700
	M_y	0.803	0.914	0.858

previously developed by White et al. [12]. Effectively, each of these seven cross sections represents a virtual load cell in the GHBM model neck. For each cervical level, a local coordinate system (LCSYS) was defined according to SAEJ211 sign convention [29] with the origin located at the CG of the corresponding vertebra (Fig. 2). The positive x -direction was directed along the midsagittal plane from the origin to the midpoint of the anterior vertebral body. The positive z -direction was directed inferiorly from the origin, orthogonal to the local x -axis. A cross section was then defined in the local xy -plane for each level of the cervical spine using the appropriate LS-DYNA keywords. The axial and shear forces reported for each of these planes is the summation of the internal nodal forces from the respective cross-sectional interface, reported in the respective LCSYS. The moments reported for each of these planes is calculated as the cross product of the internal nodal force and the vector from the centroid of the given cross section to the location of this force. The moment about the respective local origin was then calculated

using the concept of equivalent force systems. These forces and moments include the cross-sectional loading contributions from the bones, ligaments, active muscles, passive muscles, and surrounding flesh for a given cervical level.

The upper neck loading was reported at the C1 level with bending moment reported about the CG of the C1 vertebra. Unlike the upper neck load LCSYS in the ATD, which rotated with the head, the C1 loads were reported in a LCSYS that rotated with the vertebra. This method of reporting upper neck loading provided a more realistic load response in the human body model since the majority of the neck loading was transmitted through the bony skeleton, either directly or indirectly through muscle and ligament attachments. The lower neck loading was reported at the C7 level with bending moment reported about the CG of the C7 vertebra.

Data Analysis. The results of each simulation were postprocessed using Oasys T/HIS (v11, Arup, London, UK). For consistency with the reported experimental results, all simulation data was filtered using a 100 Hz, eight-pole Butterworth filter. The filtered cross-sectional force and moment results from the GHBM model were then processed using in-house code written in MATLAB (v7.12.0, R2011a, MathWorks, Natick, MA).

Results

Overview. The head and neck response is reported for the experimental and simulated rotary-wing aircraft impacts, with the upper and lower neck load polarities reported in accordance with SAE J211 sign convention [29]. Data is only reported prior to the time of head to head restraint contact. Results of the quantitative comparison analyses between the experimental test and corresponding ATD simulation are reported in Table 1. Global extrema for upper and lower neck loading, as well as N_{ij} (neck injury criteria) and BC values are reported in Table 2 for the experimental test and both simulations.

Table 2 Maximum upper and lower neck loads for the experiment and simulation along with established IARVs. Time is reported in milliseconds.

Loading mode		IARV	Parameter	Experiment	ATD	GHBMC
Upper neck loading	AP shear F_x (N)	3100	t max	132	124	136
			F_x max	496.6	371.6	790.3
		−3100	t min	204	204	176
			F_x min	−538.6	−690.2	−678.8
	Tension/compression F_z (N)	4170	t max	178	174	182
			F_z max	665.5	809.4	735.5
		−4000	t min	134	121	134
			F_z min	−1604.5	−1560.1	−898.2
	Flexion/extension moment M_y (Nm)	190	t max	192	211	189
			M_y max	53.7	63.8	77.2
		−96	t min	142	146	149
			M_y min	−34.1	−20.9	−1.6
N_{ij}	1	t max	136	122	188	
		N_{ij} max	0.36	0.32	0.35	
Lower neck loading	AP shear F_x (N)	3100	t max	126	90	180
			F_x max	182.9	46.1	198.8
		−3100	t min	202	204	136
			F_x min	−731.7	−594.3	−491.9
	Tension/compression F_z (N)	4170	t max	178	174	179
			F_z max	752.3	908.3	1416.7
		−4000	t min	133	121	134
			F_z min	−1945.4	−2261.5	−1312.7
	Flexion/extension moment M_y (Nm)	380	t max	213	206	196
			M_y max	128.8	155.8	110.5
		−192	t min	127	127	146
			M_y min	−33.2	−25.6	−25.5
	BC	1	t max	208	205	193
			BC max	0.93	1.19	0.98

Fig. 3 Midsagittal head rotation for the two FE simulations. Head rotation from the experimental test was not available for comparison.

Fig. 5 Resultant chest acceleration for the experimental test and simulations

Fig. 4 Resultant head CG acceleration for the experimental test and simulations

Fig. 6 Upper neck AP shear force for the experimental test and simulations

Head and Chest Response. While not reported in the experimental test, midsagittal head rotations for the two simulations are presented in Fig. 3. Head rotation remained minimal prior to 70 ms, during the settling phase of the simulations. While the GHBM head underwent approximately 11 deg of extension during the beginning of the sled acceleration, head flexion was the primary form of rotation for both simulations. Maximum head flexion occurred 16 ms later for the GHBM simulation than in the ATD simulation. Chin-to-chest contact did not occur in either of the simulations.

The resultant head CG accelerations are presented in Fig. 4. The global peak accelerations occurred during the first half of the time history plot for both the experimental test and ATD simulation and were almost identical in magnitude. Unlike the ATD simulation, the global peak in the experimental test exhibited bimodal-behavior. The overall CORA rating between the head accelerations in the experimental test and ATD simulation indicates good correlation between the two time histories. While the resultant head acceleration in the GHBM simulation followed a similar trend to both ATD time histories, the global peak acceleration occurred during the second half of the simulation.

The peak chest resultant linear accelerations were very similar in magnitude between the experimental test and ATD simulation (Fig. 5). The overall CORA rating indicated good correlation between the two acceleration time histories. A bimodal-peak occurred for both the experimental test and GHBM simulation, around the time of peak chest acceleration.

Upper and Lower Neck Response. Upper neck loading is reported in Figs. 6–8. The CORA analysis of the upper neck

loading showed good agreement between the experimental test and ATD simulation time histories for all loading modes. A period of positive AP shear and compression followed by a period of negative AP shear and tension occurred in the experimental test and both simulations. The change in load direction appeared to correlate with head rotation for the simulations. During compression, the experimental test exhibited bimodal-behavior, similar to the head acceleration time history. The experimental test and ATD simulations experienced a short period of extension followed by a more pronounced flexion moment, while the GHBM simulation experienced primarily a flexion moment. The N_{ij} for the upper neck was calculated using critical intercepts defined in FMVSS 208 [30]. Neither the peak loading values nor the N_{ij} values approached established injury assessment reference values (IARVs).

Lower neck loading is reported in Figs. 9–11. The CORA analysis of the lower neck loading showed good agreement between the experimental test and ATD simulation for both the axial force and bending moment time histories. The degree of agreement between AP shear time histories was less clear. The experimental test and simulations experienced a period of compression followed by tension, similar to the upper neck axial loading. Again, the change in load direction appeared to correlate with head rotation. A small extension moment occurred in the experimental test and simulations, followed by a more pronounced flexion moment. While the peak loading values did not approach established IARVs, the BC value approached or exceeded 1, indicating approximately a 50% risk of a moderate injury to the lower cervical spine [3,4].

Fig. 7 Upper neck axial force for the experimental test and simulations

Fig. 10 Lower neck axial force for the experimental test and simulations

Fig. 8 Upper neck bending moment for the experimental test and simulations

Fig. 11 Lower neck bending moment for the experimental test and simulations

Fig. 9 Lower neck AP shear force for the experimental test and simulations

Fig. 12 Cross-sectional AP shear force for each cervical level of the GHBMC simulation. The forces are reported in their respective LCSYS.

GHBMC Cervical Column Response. While only the upper and lower neck loading can be reported from the ATD, loading throughout the cervical column can be captured in the GHBMC model. To further investigate the loading distribution throughout the entire neck, cross-sectional AP shear forces (Fig. 12), axial forces (Fig. 13), and bending moments (Fig. 14) are reported for each cervical level in the GHBMC simulation with peak axial and moment values reported in Table 3. The solid vertical line ($t = 167$ ms) indicates the initiation of primary head rotation and the dashed vertical line ($t = 231$ ms) indicates maximum head rotation. The lower cervical levels tended to experience greater axial forces and bending moments than the upper levels. During

the first 150 ms a marked difference is noted between both the polarity and peak values of the AP shear force between the upper and lower cervical levels. Incremental, midsagittal neck kinematics with respect to the C7 local origin are presented in Fig. 15, illustrating the change in cervical spine curvature during the different phases of loading.

Discussion

Simulation Configuration. Each simulation was allowed to settle under the effects of gravity and belt pretensioning for 70 ms,

Fig. 13 Cross-sectional axial force for each cervical level of the GHBMC simulation. The forces are reported in their respective LCSYS.

Fig. 14 Cross-sectional bending moment for each cervical level of the GHBMC simulation. Each moment is reported about the CG of its LCSYS.

prior to the onset of the sled acceleration pulse. While some head and chest acceleration and neck loading occurred during the settling phase of the simulation, almost all of the time histories returned to zero at the onset of the sled acceleration pulse ($t = 70$ ms). Combining both the settling and sled acceleration into a single simulation allowed for the initial stresses, particularly the belt pretensioning, to be maintained. This process avoided some of the complications associated with applying initial stress and nodal positioning obtained from separate settling runs.

Only one HSM-configuration was evaluated in the current study with the location of the CG remaining relatively close to the original head CG. Since added head mass and CG location had no statistically significant effect of the magnitude of the measured forces or moments for the experiments as reported by Paskoff and

Fig. 15 Incremental, midsagittal neck kinematics of the GHBMC simulation. The position of the local origin for each vertebra is plotted with respect to the C7 local origin at different points in time.

Sieveka [10,11], any minor differences in CG location for the simulations were considered negligible.

Numerical Validation of the ATD Simulation. From a visual standpoint, the salient features of the experimental time histories appeared to be adequately simulated by the ATD model. This observation is supported by the quantitative analysis where an overall CORA rating of 0.7 or greater was achieved for all acceleration, force, and moment time histories, except for lower neck shear loading. The chest acceleration, upper neck shear loading, and lower neck bending moment received an overall rating of greater than 0.8. This relatively high degree of correlation validates the boundary conditions of this simulated rotary-wing aircraft impact.

Between 100 and 150 ms, distinct bimodal-peaks occurred in the experimental head and chest accelerations, as well as the upper and lower neck axial loading. The magnitude of the second peak was always larger than the first peak and the local minimum between these peaks occurred at approximately 130 ms. Even though this bimodal-peak behavior was not reproduced by the ATD model, the simulation peaks always aligned in time with the first, smaller peaks of the experimental bimodal-response. While the experimental belt loading was not reported in the literature, it was examined in the ATD simulation. It appeared that the ATD began to pitch forward and load the shoulder harness at around 130 ms, the same time the local minimum occurs in the experimental bimodal-response. Due to the high degree of correlation between the experimental test and simulation in terms of head and neck response, it is reasonable to deduce that the simulated belt loading approximated the experimental loading. Therefore, the bimodal-response in the experiment data may be the result of the initial shoulder harness interaction with the ATD, creating a sharp, momentary deceleration of the upper torso. Minor differences in

Table 3 Maximum cross-sectional axial force and bending moment for each cervical level in the GHBMC simulation

Level	Compression		Tension		Extension		Flexion	
	Time (ms)	F_z (N)	Time (ms)	F_z (N)	Time (ms)	M_y (Nm)	Time (ms)	M_y (Nm)
C1	134	-898.2	182	735.5	149	-1.6	189	77.2
C2	134	-1110.9	182	900.3	149	-3.6	189	78.1
C3	135	-1383.9	181	1040.4	148	-11.8	190	84.5
C4	135	-1352.0	178	1271.6	150	-34.8	193	96.7
C5	134	-1319.9	177	1344.3	150	-39.9	195	106.5
C6	134	-1334.0	178	1399.6	149	-33.0	195	106.4
C7	134	-1312.7	179	1416.7	146	-25.5	196	110.5

belt placement, pretensioning, and friction properties may explain why the bimodal-response was not produced in the simulation.

The largest discrepancy between the experimental test and simulation response occurred for the lower neck AP shear, as was evident by the low overall CORA rating. A clear, visual difference between the experimental test and simulation responses occurred between 100 and 150 ms. In the experimental test, a short duration positive AP shear peak occurred at 126 ms, just prior to a large, almost instantaneous reverse in polarity at approximately 130 ms. The simulation produced neither this distinct positive AP shear peak nor such a sharp polarity change. The disparity in simulation and experiment responses was most likely due to a difference in initial shoulder harness interaction with the ATD.

GHBMC Biomechanical Response. The GHBMC model produced a more biofidelic neck response during the simulated impact. Similar to the upper and lower neck loads reported at the C1 and C7 levels, the AP shear, axial, and bending moment loads were also reported for the C2–C6 levels; effectively capturing the response of the entire GHBMC neck. Prior to onset of head flexion rotation, all levels underwent axial compression with peak levels occurring at approximately the same time, between 134 and 135 ms. Maximum compression was noticeably larger for C3–C7 levels than for the C1 and C2 levels. At approximately 136 ms, the C1 and C2 levels experienced maximum positive AP shear as the C3–C7 levels experienced a local peak negative AP shear. From the incremental kinematics (Fig. 15), it appeared that the cervical spine takes on an S-shaped curvature during this period of time, with the transition in curvature occurring between the C3 and C4 levels. This transition was even more pronounced at the time of maximum extension moment. Similar observations in spinal curvature were reported in the cadaver component and sled tests by Bass et al. [4]. This S-shaped curvature can be explained using basic beam deflection theory. From the simulation animation, it was observed that the global head CG translation from the onset of the sled pulse to the onset of head flexion occurred almost exclusively in the horizontal direction. This creates a paradigm where the neck can be represented as a beam fixed at the C7 level and free, but guided, at the C1 level. From the corresponding beam deflection equation, it can be shown that the beam, in this case the neck, will bend into a sigmoidal shape. Transition of the curvature occurs at the midpoint of the beam, which would correspond most closely to the C4 level.

The onset of axial tension began at approximately 160 ms for the C4–C7 levels and 167 ms for C1–C3 levels. This delay between the lower and upper levels appeared in the incremental kinematics illustration, where the lower vertebrae began to straighten out prior to the onset of head flexion. Straightening in the upper levels occurred by the time of maximum tension, creating a C-shaped curve in the cervical spine, indicative of simple beam bending. Maximum tension was reached at approximately the same time for all levels, with the C4–C7 levels reaching greater peak values than the C1–C3 levels. Shortly after maximum tension occurred, maximum flexion moment was achieved for all levels. The largest peak moments were achieved at the C5–C7 levels, due to the longer moment arm from these levels to the head. Peak flexion moments were smallest for the C1–C3 levels, due to the shorter moment arm.

To assess injury risk, the N_{ij} was calculated for the upper neck and the BC for the lower neck. The N_{ij} was originally developed to assess the risk of upper neck injury during a frontal automotive crash through a linear combination of normalized axial force and sagittal bending moment. Federal regulations set a threshold of 1.0 for the N_{ij} in the upper neck, corresponding to a 22% risk of a serious injury [30]. Using the axial loading and bending moment from the C1 cross section, the calculated N_{ij} value was well below this injury threshold. The BC was developed to assess the risk of lower neck injury during combined vertical-frontal crashes with through a linear combination of normalized axial force and

sagittal flexion moment [3]. Using the axial loading and flexion moment from the C7 cross section, the calculated BC value approached a value of 1, indicating a 50% risk of moderate injury to the lower cervical spine.

Limitations. The current work is not without limitations. One is that there is no direct method to validate the cross-sectional forces and moments calculated at the different levels of the GHBMC neck. Even if experimental cadaver tests were conducted using the same setup as the ATD, these values could not be measured in the actual cadavers. While it is possible to use inverse kinematic methods in to estimate the forces and moments at the occipital condyle [31] and T1 [32] in cadavers, there is no direct measurement method. Traditionally, inverse kinematic approaches have been acceptable. Cross sections are a predictive tool, as is the model itself, which estimate the loading response of the cervical spine during a simulated dynamic impact. A second limitation is that there is no intrinsic method to position the joints of the GHBMC model so that the initial position perfectly matches that of the ATD in the experimental sled test. However, given the overall head, neck, chest response of the GHBMC model followed that of the ATD, any difference in model response due to initial positioning is considered minimal.

Conclusions

The head and neck response of an FE ATD during a simulated rotary-wing aircraft impact was successfully validated against experimental test data. A quantitative analysis was conducted to compare the responses of the simulation and experimental test. All but one time history received a CORA rating of 0.7 or higher, indicating good overall correlation. The neck response of a human body FE model was also investigated during a simulated rotary-wing aircraft impact. Detailed neck loading for each cervical level was reported. While the maximum axial and AP shear forces and bending moments did not exceed existing IARVs, the BC value calculated from the C7 axial loading and flexion moment indicated a 50% risk of moderate injury to the lower cervical spine. The loading distribution throughout the cervical spine may help to explain the kinematic response, including change in spinal curvature. Sigmoidal curvature of the cervical spine was most prominent during the early part of the simulation when compression and extension moment were greatest. At this time, the AP shear force was positive in the upper cervical spine and negative in the lower cervical spine. Relating load distribution in the cervical spine to kinematic response has potential to explain the mechanism of injury during dynamic impact. These observations highlight the importance of human body modeling in understanding the underlying biomechanics of potential neck injury, a much harder task to complete when using an ATD.

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