HUMAN HEAD-NECK INJURY ASSESSMENT USING A MULTIBODY MODELING AND SIMULATION

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Abstract

The increasing number of vehicle accidents and injuries caused to the occupants has lead the researchers to perform research work using different methods of studies in this field. The major source of permanent disabilities or even death sometimes are injuries to head, brain, and neck of the vehicle occupants. Different software's and methods are being implemented to minimize these risks of injuries. The use of mathematical models has been proving beneficial in these studies. Models resemble the motions of the body of humans, and prove to be advantageous being cheaper, quicker, and more detailed in results, and are also capable of predicting many different crash situations. In this study, a multi-body model of the cervical spine is being developed which is a discrete parameter mathematical model of human cervical spine. The elements of the head-neck system are considered as a linkage of rigid bodies. The multi-body model corresponds to a 50th% adult male. The model is designed in Pro/E and then imported to Visual Nastran 4D. The constraints are given between the multi-bodies. The input acceleration given to the torso system follows a curved path and the outputs are noted at the head, as a measure of injury severity. The outputs are compared with the NAMRL experimental results the similarities between the model response and the experimental performance were found. The analysis of the neck with the negative accelerations using -6G sled test is performed.

Key words: Head-Neck, Multibody, Modeling, Head Injury Criteria (HIC), 6G Sled test.

I. INTRODUCTION

Head Injury Criteria (HIC)

The response to a study by Versace on comparison of the WSTC and the SI, a new injury criterion for the head was defined by the U.S government, the Head Injury Criterion (HIC). HIC assists in developing improved design standards to reduce injuries in vehicles crash situation. The various physical parameters used in the evaluation of head injury are translational and rotational acceleration levels of head impact, impact force, velocity and kinetic energy, impulse and impulse duration, and many more.

$$\mathrm{HIC} = \left\{ (t_2 - t_1)\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)\,dt\right]^{2.5} \right\}_{\max}$$

Where t_1 and t_2 are the initial and final times (expressed in seconds) of the interval during which the HIC attains a maximum value and a(t) is the resultant acceleration (expressed in G) measured at the head center of gravity. The time duration $(t_2 - t_1)$ used in the calculation should be taken as the contact time for the impact; however, this is often very difficult to ascertain in physical evaluations using crash test dummies. In using HIC for assessing the potential of concussion then a maximum time duration of 15 milli seconds should be used, which was the maximum time duration for which the original tolerance curve was developed. Longer contact time durations can be used to predict skull fracture. The highest acceleration, independent of location or direction, should be used in the Head Injury Criteria, which will therefore be the resultant acceleration measured at the heads centre of gravity. The final formulation for the HIC is one in which the time interval from t_1 to t_2 is that which maximizes the value of the expression in equation.

The regulations of Federal Motor Vehicle Safety Standards (FMVSS) and Federal Aviation Administration (FAA) states that HIC is a method for defining an acceptable limit i.e. the maximum value of the HIC should not exceed 1000. If this index exceeds 1000, the situation is considered to be dangerous and the occupants are expected to overcome serious injuries or even death. The index is less than 1000, the situation is considered not to be life threatening. The time interval greatly affects HIC calculation. The maximum time interval (t_2-t_1) which is considered to give appropriate HIC values was set to 36 ms by automotive industry. In last few years, time interval has been gradually replaced by a 15 ms in order to restrict the use of HIC to hard contact impacts.

The recommended critical HIC levels for the various occupant sizes can be tabulated as follows:

Injury Criteria	Hybrid III Mid-Sized Male	Hybrid III Small Female	Hybrid III 6-Year-Old Child	Hybrid III 3-Year-Old Child	12-Month-Old Infant (CRABI)
Head Injury criteria (HIC_{36})	1000	1000	1000	900	660
Head Injury criteria (HIC_{15})	700	700	700	570	390

Table 1 Critical HIC levels for various occupant sizes [12]

II. OBJECTIVE AND PROBLEM STATEMENT

A. Objective of the Study

The objective of study is to develop simple as well as anatomically and dynamically discrete parameter mathematical model of human head-neck-torso system. The models are developed using the multi-body methodology, where the elements of the head-neck-torso system are considered as a linkage of rigid bodies. The multi-body model corresponds to a 50^{th} % adult male. In order to study the dynamic response and the injury mechanisms of the head and the cervical region, the models are subjected to various acceleration/force pulses that depict the real crash situations. The loads are related to the cervical vertebrae to their experimentally verified load limits and also correlate the dynamic responses of cervical vertebrae of the model. The test can be applied to the development of improved design standards to reduce injuries in actual vehicle crash situations.

B. Problem Statement

Develop anatomically and dynamically discrete parameter mathematical model of human head-neck-torso system that corresponds to a 50^{th} % adult male. The model is designed and analyzed using the experimental values. The model is being subjected to the constraints such as rotational/torsional spring dampers, and the stiffness and damping coefficients. The results such as position, velocity, acceleration, moment, are being compared with the experimental results. The extension responses of the model are being noted and then compared with the experimental data. Another important criterion that is to be calculated and compared with the experimental values is Head Injury Criteria. After analyzing all the outputs of the model with the respective experimental values, the conclusion of whether the criteria of the model are in the safer zone or in the life threatening zone.

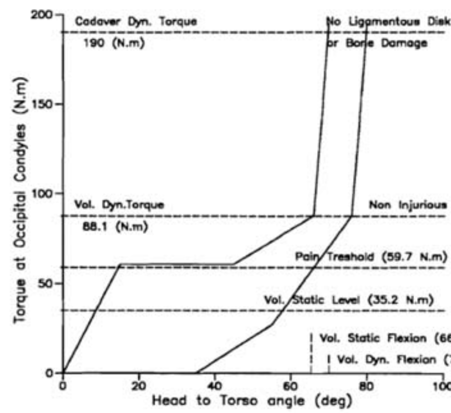

Figure 1 Neck flexion corridor

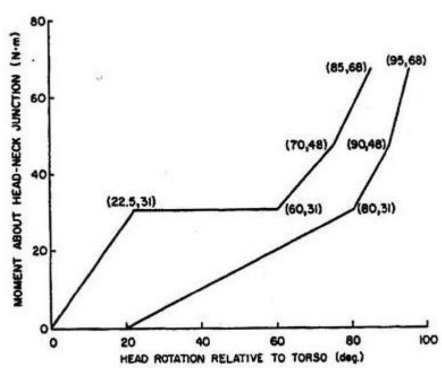

Figure 2 Neck extension corridor

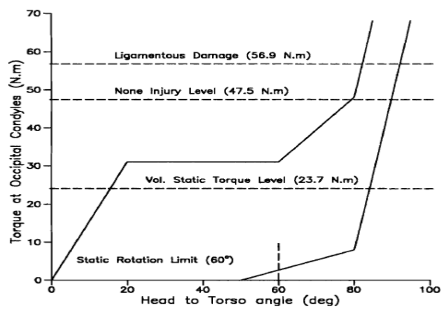

Figure 3 Flexion response for the loading phase

Head-neck response envelops in flexion for the loading phase according to Mertz and Patrick:

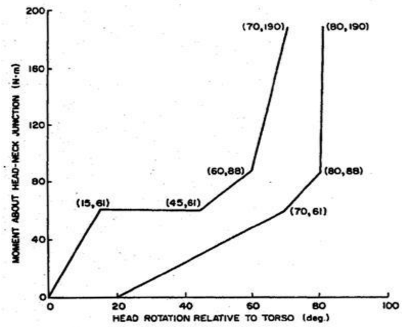

Figure 4 Extension response for the loading phase

III. METHODOLOGY

A. Modeling of Human Head-Neck-Torso System

The head-neck-torso system is modeled as nine discrete bodies, the head, the seven cervical vertebrae elements C1 through C7, and the upper torso T1 as a rigid base. We also include the ground. The cervical vertebrae elements are attached to the upper thoracic region. This model has nine degrees of freedom. Each of these bodies is modeled as rigid bodies. All the bodies have reasonably correct proportions, and inertial properties. The system is modeled as a series of rigid bodies. All these bodies are linked with a set of joints between each body which constraint their motion relative to each other. We use two types of kinematic constraints in this model. They are translational and revolute joints. Revolute joints are used between the bodies that impersonate the kinematic constraints in the upper spinal column. They allow rotation about the sagittal plane (or say xy - plane). Translational joint is used between the torso and the ground, which allows the motion only along sagittal plane (xy - plane), constraining the rotation in the same plane. We also use the nonlinear rotational springs and dampers that show the effects of muscles, intervertebral discs, cartilage, ligaments, cerebrospinal fluids, and other tissues.

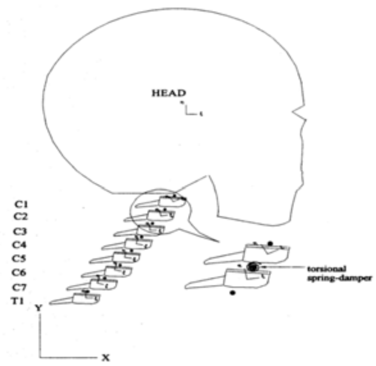

Figure 5 Dynamic model of two-dimensional flexion-extension head-neck-torso system [13]

The mass and mass moment of inertia of each component is very important properties in the simulation of the model. The total neck mass from C1 through C7 used is 1.568 kg which is equal to the neck mass of a 50th % adult male. The mass and mass moment of inertia of body T1 are large compared to the rest of the system, this is because the input forces are to be large relative to the reaction forces produced by the motion of the rest of the system.

The range of motion angles measured on living humans are defined according to the motion limits for voluntary forcing joint stops.

A rotational spring, in Fig. 6, applies pure moments on the bodies, equal in magnitude and opposite in direction. The moment is found as

$$n^{(r-s)} = k\,(\theta - \theta^0)$$

Where k is the spring stiffness, θ is the deformed angle of the spring, and θ^0 is the undeformed angle.

Table 2 Geometrical and mechanical properties of a 50th % adult male head-neck model

Body Number	Mass m (kg)	Moment of Inertia μ (kg.m^2×10^{-4})	Coordinates of the center of mass x (m)	y (m)	ϕ (deg)
1 Head	3.53	6.25	0.05	0.195	0.0
2 C1	0.156	5.2	0.037	0.137	13
3 C2	0.156	5.2	0.05	0.110	12
4 C3	0.156	5.2	0.06	0.820	9
5 C4	0.205	5.9	0.063	0.063	3
6 C5	0.269	6.6	0.065	0.047	-8
7 C6	0.226	9	0.064	0.03	-20
8 C7	0.40	33	0.06	0.018	-29
9 Torso	50.0	12430	0.058	0.006	0.0
10 Ground	0.0	0.0	0.0	0.0	0.0

Table 3 Revolute joint data (global coordinates) for a 50th % head-neck model

Joint No.	Bodies	x (m)	y (m)
R1	1,2	0.0313	0.147
R2	2,3	0.0423	0.127
R3	3,4	0.0572	0.093
R4	4,5	0.061	0.071
R5	5,6	0.0658	0.054
R6	6,7	0.0648	0.04
R7	7,8	0.064	0.024
R8	8,9	0.06	0.126

Table 4 Rotational Spring-Damper data for a 50th % head-neck model

Spring-damper	Rotational Stiffness k (N.m/rad)	Rotational Damping coefficient d (N.m.s/rad)	Initial Difference Angle β (deg)	Flexion (deg)	Extension (deg)
Head-C1	271.2	2.226	13	10	-25
C1-C2	271.2	0.911	-1	11	0
C2-C3	271.2	0.911	-3	7	-2
C3-C4	271.2	0.911	-6	10	-4
C4-C5	271.2	0.911	-11	13	-9
C5-C6	271.2	0.911	-12	15	-3
C6-C7	271.2	0.911	-9	9	-10
C7-T1	271.2	0.911	29	4	-6

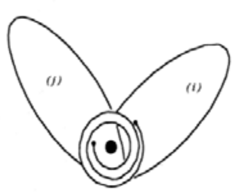

Figure 6 Rotational spring-damper between any two bodies of the cervical spine

B. *Designing the Model*

A three-dimensional mathematical model of the human head-neck-torso musculoskeletal system is developed. This model studies the motion responses of the human cervical spine. The model is developed within a rigid body dynamic simulation program, Visual Nastran 4D. We use two types of kinematic constraints in this model. They are translational and revolute joints. Revolute joints are used between the bodies of the upper spinal column since they allow rotation about the sagittal plane (or say xy – plane). Translational joint is used between the torso and the ground, allowing the motion only along sagittal plane (xy – plane), constraining the rotation in the same plane. We use the nonlinear rotational springs and dampers that help in showing the effects of muscles, cartilage, and other tissues.

Visualization of the mathematical model in Visual Nastran 4D:

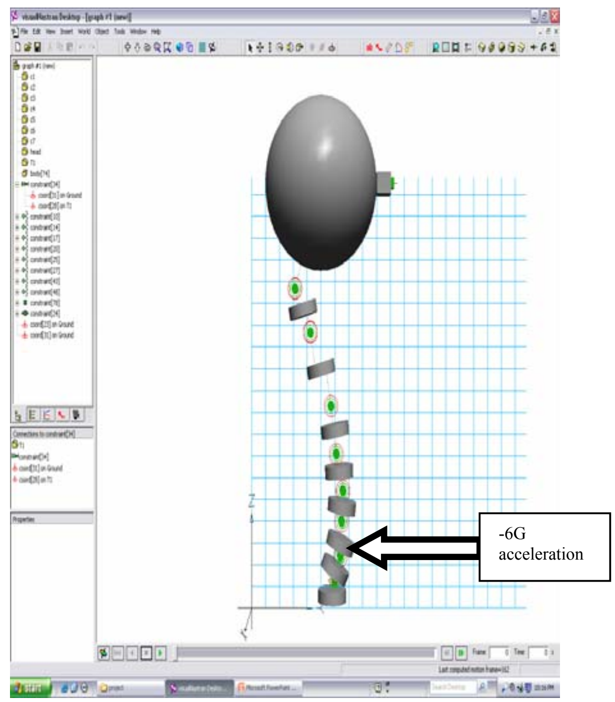

Figure 7 The mathematical model in Visual Nastran 4D

C. *Model Validation for –G_X (frontal impact)*

For the purpose of the model validation, the input accelerations of the model were taken from Naval Aerospace Medical Research Laboratory (NAMRL) sled tests. These sled tests were conducted at 6G in -G_x. These sled tests were carried out using human volunteers and were designed specifically for the study of head and neck dynamics. The experimental data from these tests provide basic information for validating and improving mathematical models and anthropomorphic dummies for impact simulation studies.

D. *Input Accelerations*

The T1 accelerations (output accelerations from torso) resulting from the experiments carried out using human volunteers, were averaged and used as input to the model. The input accelerations were applied in only horizontal direction at the base of the neck (or say T1). Since the translational joint used between the torso and the ground allows the motion only in along xy-plane or the sagittal plane, no input accelerations in vertical directions.

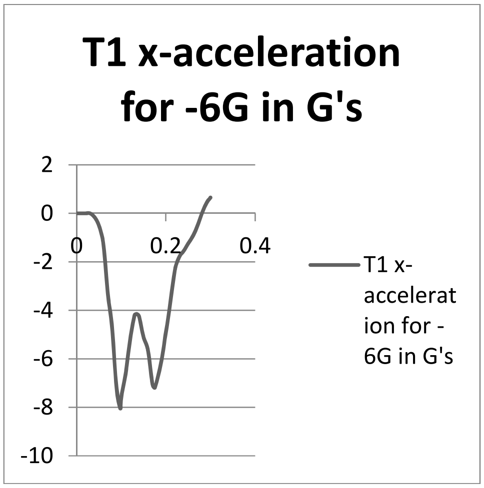

Figure 8 T1 acceleration along x-axis from -6G experimental sled test

IV. SIMULATION RESPONSE FOR $-G_X$ (FRONTAL IMPACT)

A. Simulation Results

The best simulation results such as head resultant acceleration, head angular acceleration, head angular velocity and head angular position are being noted, and then compared with the NAMRL experimental results.

The simulation curve profiles match well with the experimental results but the curves for head resultant acceleration, head angular acceleration and head angular velocity are slightly higher than that of the experimental curves and for head angular position are slightly lower than that of the experimental curves. The slight lag observed in the curves of experimental and simulated responses of head angular accelerations, head angular velocity and head angular position are due to the following reasons:

- The rigidity of the model compared to the flexibility of the human subject.
- NAMRL results are the average of five subjects, for the purpose of obtaining the exact values of input for T1.

Also shown are the head linear displacement, head linear velocity and moment at the occipital condyle for -6G sled tests respectively.

Results for -6G:

Time (sec) vs Head Resultant Acceleraion (G's):

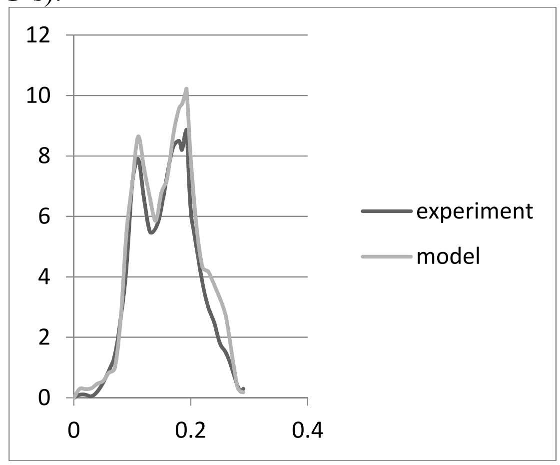

Figure 9 Resultant acceleration of the head from -6G sled test

Time (sec) vs Head Angular Acceleration (rad/sec^2):

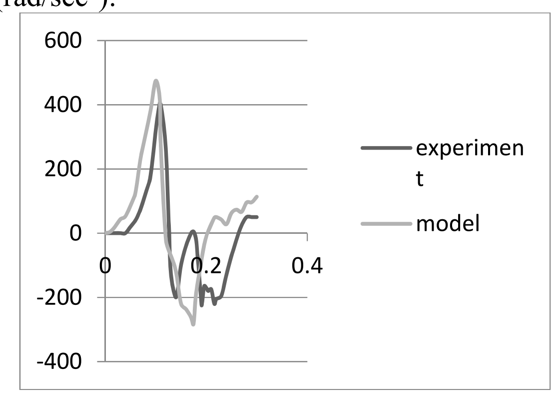

Figure 10 Angular acceleration of the head from -6G sled test

Time (sec) vs Head Angular Velocity (rad/sec):

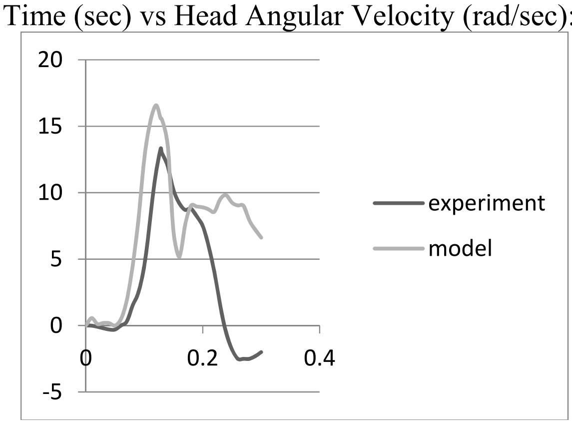

Figure 11 Angular velocity of the head from -6G sled test

Time (sec) vs Head Angular Position (rad):

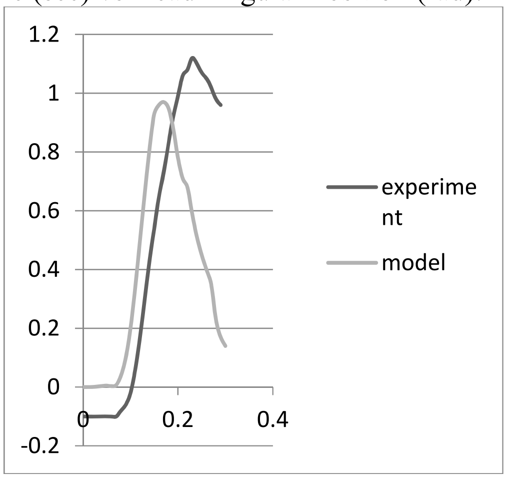

Figure 12 Angular position of the head from -6G sled test

B. Response of the System

Corridors proposed by Mertz and Patrick have been widely used for the purpose of comparison of moments generated at the occipital condyles of the head. We compare the head to torso angle with the torque response, with the Mertz and Patrick corridor. This clearly shows whether the responses fall within the corridors.

Head to Torso angle (deg) vs Torque at Occipital Condyle (N.m) compared with the flexion corridor:

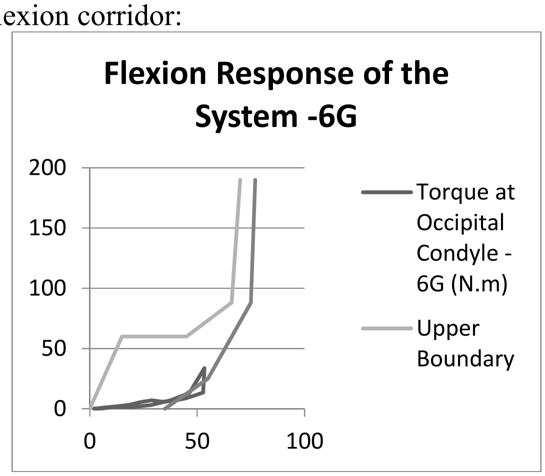

Figure 13 Flexion response of the system from -6G sled test

The above curve of head to torso angle vs torque at occipital condyle from -6G sled test when compared with the neck flexion corridor levels, we can see that the curve passes the upper boundary by slight lag. The peak value of the curve is 25 Nm, which does not cross the minimum level of 35.2 Nm. Hence the model is safe.

V. CALCULATION OF HIC

We compare the Head Injury Criteria (HIC) values for the model with the experimental values.

HIC values of the model:

Table 5 HIC values of model

Run	t_1 (msec)	t_2 (msec)	HIC
-6G	1	5	24.6

- The HIC of -6G is 24.6 which is less than 700, hence the model is safe.

When the values of HIC are observed, they are below 700. Thus the model is considered non-injurious.

Head to Torso angle (deg) vs Torque at Occipital Condyle (N.m) compared with the loading flexion corridor:

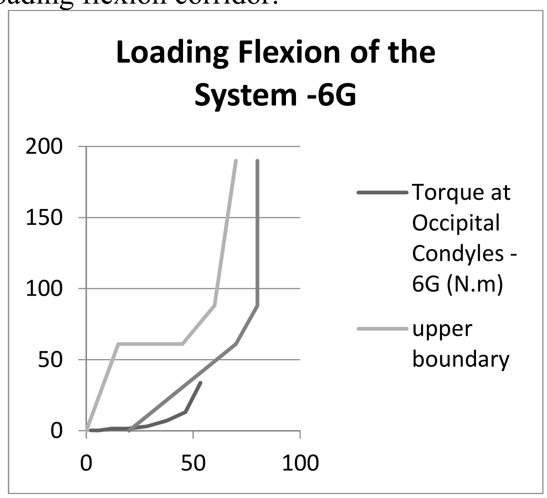

Figure 14 Loading flexion response of the system from -6G sled test

VI CONCLUSIONS

Mathematical models are a very good source that can be used to gain knowledge in the response of the occupant segments under varying conditions and to evaluate their displacements, velocities and accelerations. In this study, a mathematical model of a human head-neck-torso system were developed using rigid multibody methodology. In this methodology, the elements of the head-neck system are considered as a linkage of rigid bodies and are connected by pin or revolute joints and nonlinear rotational spring-dampers. The input accelerations are applied to the model at the torso and the output reactions of the head were noted and studied. Since all the criteria shows that they are in the limits, the model is considered to be safe and secure. We can quantify the results as follows:

- The HIC of -6G is 24.55 which is less than 700, hence the model is safe.

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