VALIDATION OF THE CERVICAL SPINE MODEL IN THUMS

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ABSTRACT

THUMS (Total human model for safety) [Watanabe et al[1]] is a finite element model of human body developed to study various injury mechanisms and for use as a substitute for crash test dummies. The development team of Toyota Central R&D Labs (TCRDL) has validated different parts of this model against experimental data available in literature. Neck response data for different impact conditions is available in Mertz and Patrick[2],[3] and McElhaney et.al[4,5]. A preliminary validation of the neck model in Thums, against some of these tests, has been presented by the TCRDL group [Oshita et.al[6]] but no extensive validation has been reported for the variety of test conditions reported in literature. Typically, frontal and rear end impacts are of interest and these cause bending, axial as well as torsional loading on the cervical spine. A computational model can be expected to validate against multiple boundary conditions and initial conditions. Therefore, validation of a computational model (THUMS) in varying test conditions is of significance. Thus the objective of the current work is to independently investigate the fidelity of the neck model of THUMS under varying impact conditions.

From the initial seating position the Thums model has been modified to match the initial position in the tests. The impact test conditions used in the experiments have been then recreated in PAMCRASH™ and simulations have been carried out to validate the neck model. The models and the material properties have then been iterated and the performance of the Thums model has been investigated vis-à-vis the experimental results.

INTRODUCTION

Injuries to the neck, or cervical region, are very important since there is a potential risk of damage to the spinal cord. High-speed transportation have increased the number of serious neck injuries and made us increasingly aware of its consequences. The incidence data from the injury surveillance program at the Swedish National Board of Health and Welfare, [Karrin[7]] is indicative of this..

The lower cervical spine is the most frequently observed location for spinal trauma. It has been shown that cervical spine injuries are more often connected with spinal cord injuries than the lower spinal regions, Pintar and Narayan[8]. There is also a strong association between head and face trauma and neck injuries. Hence a neck injury in automobile crashes is a problem that needs to be addressed with new preventive strategies.

FE Models of the human body are now being developed to aid in development of new protection devices for vehicles. These models include realistic anatomical geometry of the human body and their physical properties, to predict kinematics, kinetics, and internal stresses and strains inside the human body. THUMS is one such human body model[1].

The THUMS model represents a 50 percentile American male in seating position. The model has been developed by Toyota Central R&D Labs. Inc, Toyota System Research Inc., and Toyota Motor Company in conjunctions with the Wayne State University[1,9,10]. The model contains about 60,000 nodes and 80,000 elements. Each bone consists of cancellous zone modeled using solid elements and cortical zone modeled using shell elements. In the joints of THUMS model, ligaments that connect the bones are modeled using shell / beam elements and sliding interfaces are defined on the contacting surfaces of these bones. Skin and muscles that cover the bone are modeled with solid elements.

The purpose of THUMS is to simulate responses of human body sustaining impact loads. However these FE models need to be validated before they can be used effectively. Various studies for the validation of different parts of the THUMS model have been reported by the Toyota group (6, 9, 10 to name a few).

In the present work Human Cervical Spine (Neck) Model of THUMS has been validated for different impact conditions (Frontal, Rear and Torsion). Simulations have been developed for these impact configurations and compared against experimental data already available in literature. We first briefly mention the experimental data used in this work and then describe our simulations and comparisons.

EXPERIMENTAL DATA

Considerable work has been done in the area of measurement of the response and tolerance of the human neck in impact environment. Some of these papers include the information of the actual test condition and boundary condition imposed on human cadavers and volunteers. In this section we describe some of this data that we have used for validations.

Mertz and Patrick[2], conducted several tests on cadavers for investigating the kinematics and kinetics of whiplash. The work also proposes mathematical modeling of dynamics of human head for different impacts. Later, Mertz and Patrick[3] conducted test for neck response envelopes for the extension and flexion of the neck. They report motion of the head relative to the torso in the segittal plane and the static and dynamic strength of the neck in flexion and extension.

McElhaney etal[4] investigated the lateral, anterior and posterior passive bending responses of the human cervical spine from cadavers. Results include moment angle curves, relaxation modulii and the effect of cyclic conditioning on bending stiffness of cervical spine. Later, McElhaney etal[5] have investigated the responses of the unembalmed cadaver cervical spine to axial rotations of the head about a vertical axis. Thunnissen and Philippines[11] investigated the head-neck response; the neck loads and the sustained injuries obtained from human cadaver experiments in the frontal, lateral and rear-end collisions. Ono and Koji[12] analyzed the motion of the cervical vertebrae under varying conditions. They investigated head and neck responses in low speed rear-end impact conditions and have focused on the head kinematics using sled tests with post mortem human subjects. Rizzetti et al. [13] reported skull, brain and cervical spine injuries through direct head impacts. Fourteen head impacts (frontal, lateral or occipital) with cadavers were performed.

Panjabi et al.[14] reported the current understanding of the injury tolerance of the human cervical spine and characterization of the mechanical properties and injury criterion of the cervical spine. They also documented the state-of-the-art by which surrogate devices and models may be used to mimic the mechanical behavior of the human neck.

In this work we present validation of the Thums model against frontal impact tests of Mertz etal [2,3], rear end impact tests of Ono etal[12] and torsion tests of Myers etal[15].

VALIDATION METHODOLOGY AND MODEL DEVELOPMENT

The nominal posture of the Thums model is a sitting position. In order to validate the neck model of Thums we had first modified the FE mesh of THUMS, to bring its position identical to that used in the experiments. This turned out to be a non-trivial exercise for human body models. The dummy had been positioned in the correct posture by running successive simulations for altering the dummy position. The deformed / positioned dummy obtained from these simulations were used as an input mesh in the next stage, and iteratively the initial condition for the meshes is obtained.

The following sub-sections describe how the model has been developed for the three tests simulated in this work.

Simulation model for frontal impact

While conducting test on human cadavers for dynamic hyper-flexion, the subjects were restrained on a rigid chair mounted on an impact sled[3]. The sled is accelerated pneumatically over a distance of 6 ft to the prescribed velocity. The head position was set to a vertical, upright position and backrest at 15 degree from the vertical. The sled was then brought to rest rapidly by a hydraulic cylinder to generate the deceleration pulse. The model of sled in PAM-GENERIS using shell elements, defined as rigid is shown in Figure 1

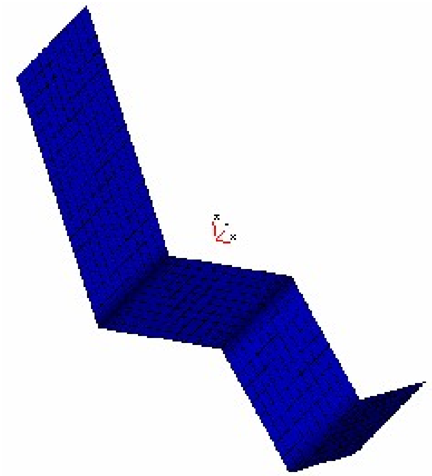


Figure 1. Sled Model using Shell Element

The THUMS model had then been brought into a sitting position on the sled i.e. in the same position as the human subjects were at the time of the experiment. This was achieved by dynamically simulating the sitting process in Pamcrash by successive single axis rotations. The initial and the final position of the THUMS model have been shown in the Figure 2 and Figure 3 respectively.

The restraint system used in the experiment consisted of a lap belt and two individual shoulder harnesses that crossed at the mid-sternum. In addition, the subject's feet were fastened to the foot support.

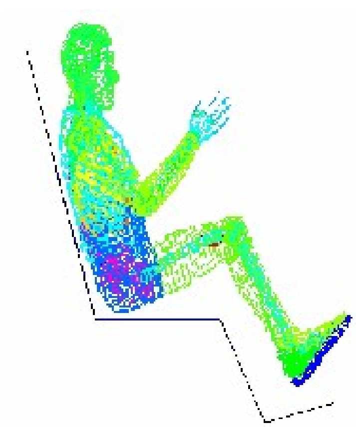


Figure 2. Initial Position of THUMS.

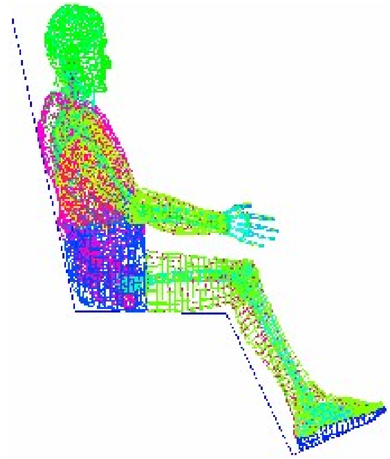


Figure 3. Final Sitting Position of THUMS for frontal Impact.

Belts have been modeled using multiple beam elements, and are assigned a material model 205 in PAM GENERIS^{TM}which is a non-linear ension-only bar element meant for modeling of seat belts.

Simulation models for low speed rear impact simulations

The experimental responses for low speed rear impact has been reported by Ono et al[12]. In these tests, the head position was set in a vertical upright position, backrest is at 20 degree from the vertical and sitting base is at 10 degree from the horizontal. The sled has been modeled in PAM-GENERIS using shell elements and has been designated as a rigid body.

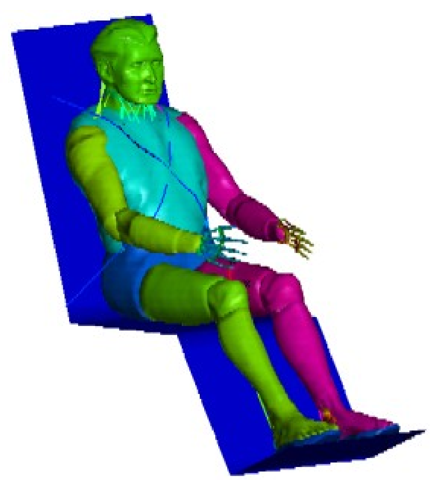


Figure 4. THUMS with chest restraint system

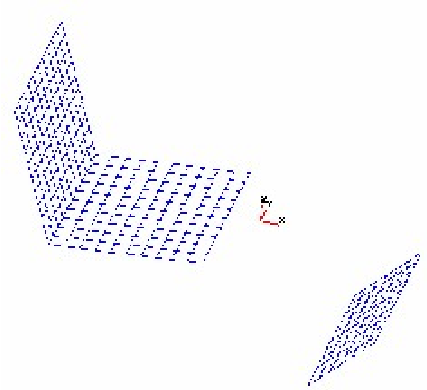


Figure 5. Modeled Sled for rear impact

THUMS mode had been modified so as to have the same initial position as reported in Ono et al[12] (Figure 6).

Simulation models for the neck in torsion

Myers etal[15] reported experimental response for the cervical spine in torsion. The experimental cervical spine specimen included the base of the skull, approximately two centimeters around the foramen magnum and the first thoracic vertebrae at the caudal end, with all the ligaments structures kept intact. It is found after experimentation that all failures were confined to the atlanto-axial joint. A similar model of neck has been prepared for simulation by eliminating the structures other than C2 to T1.

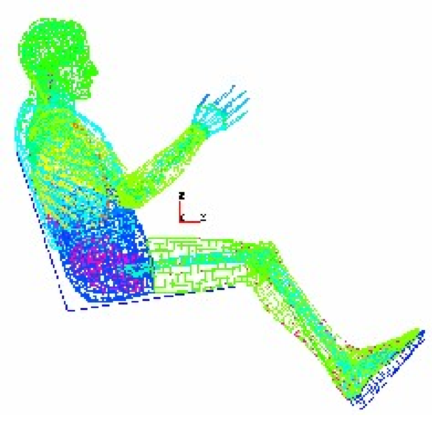


Figure 6. Final sitting position of THUMS for low speed rear impact simulations

The thoracic vertebra was kept fixed and the axis was given various input rotational velocities. Specimens were cast into aluminum cups so that the ends were parallel. The axial movement of the neck has been permitted. Same boundary conditions have been incorporated in the simulation models shown in Figure 7.

Models for the three test conditions, viz, frontal impact, rear impact and torsion, had thus been duplicated to reproduce the geometry and end-fixity conditions. Subsequently the THUMS neck muscle material model and the associated material properties were tuned to match with the experimental results available. Hill material Model was implemented in all the neck muscles, and its properties have been iterated to match the results. The next section describes the results of the simulations and their comparisons with experimental data.

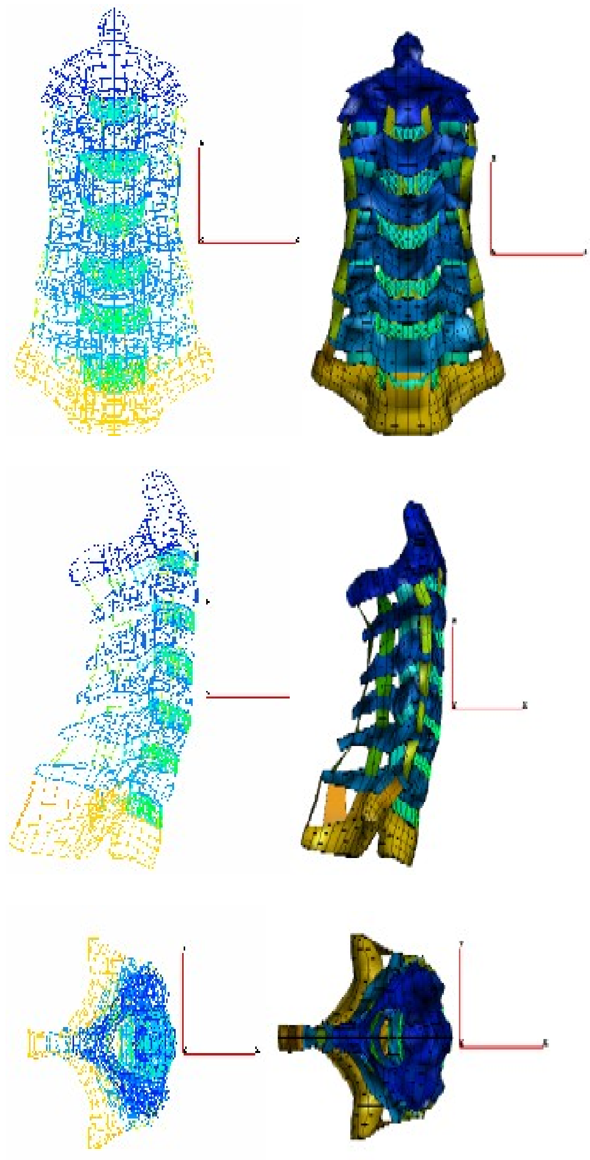


Figure 7. Front, side and Top view of the neck model prepared for simulating Human cervical spine to torsion.

RESULTS AND VALIDATIONS

Results of frontal impact, rear impact and torsion response of THUMS neck are now discussed in this section.

Validation for frontal sled impact test

For the frontal sled impact simulation using THUMS conditions corresponding to cadaver 1538[3] have been simulated. Additional weight of 1.36 kg has been put at the center of gravity of head of cadaver, and an initial velocity of 5.88 m/s has been given. The sled has been made to stop within a distance of 0.254 m with a deceleration pulse (Figure 8) having a plateau deceleration of 66.7 m/s^2.

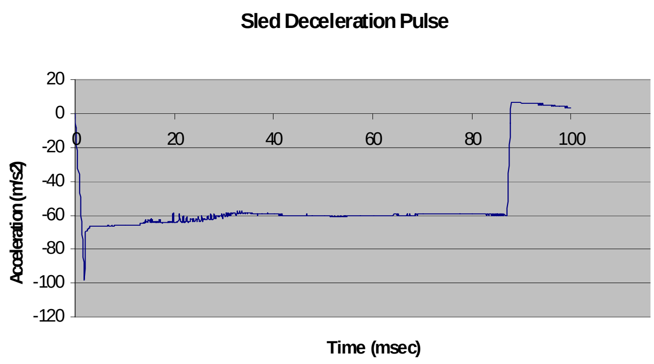


Figure 8. Deceleration pulse used in Rum 200 with cadaver 1538

With the test conditions as stated above, is the model was simulated for 100 msec termination time.. Figure 9 shows the movements in THUMS after 0.025, 0.05, 0.075 and 0.092 msec.

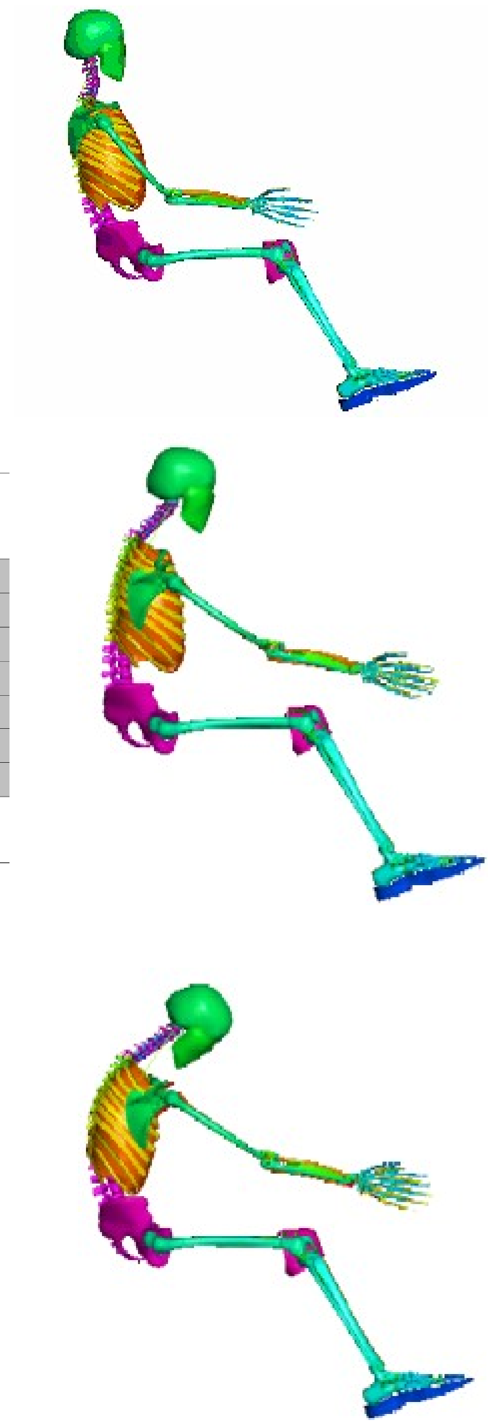


Figure 11 plots the equivalent moment about the occipital condyles as a function of angular rotation of head relative to torso, for experiment and simulation using unmodified, elastic and Hills model for muscles. This curve is of primary interest for validating THUMS neck behavior.

Figure 9. Movements in THUMS after 0.025, 0.05, 0.075 and 0.092 msec.

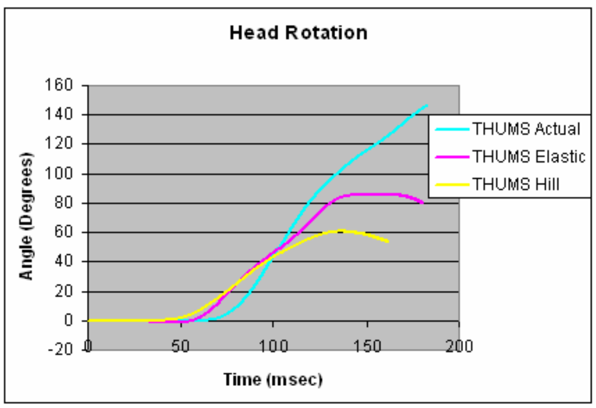


Figure 10. Angular rotation of Head with Torso from simulation.

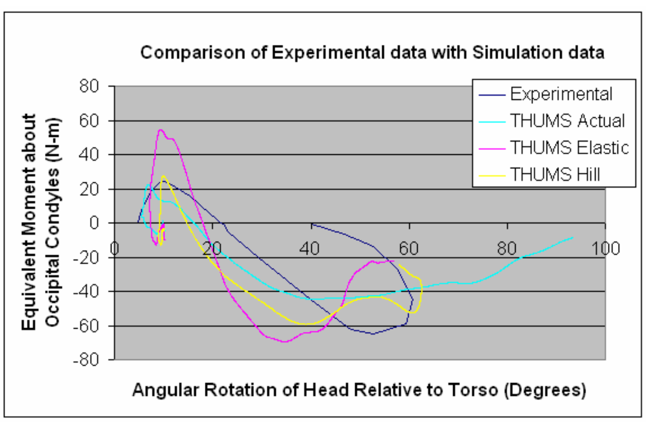


Figure 11. Moment as a function of the angular position of head, under hyper flexion.

Figure 10 shows that the THUMS model had larger relative head angulations, 140 deg as compared to 73 for cadaver 1538. This large difference can be due to difference in neck stiffness and muscular behavior.

For cadaver experiment, maximum equivalent moment of 27 Nm was observed at relative head rotation of 20 deg. For THUMS this was 40 Nm at 19 degree relative head rotation. Peak resisting moment for cadaver was 62 Nm at 50 deg of relative head rotation. For THUMS this was 60 Nm at 50-60 deg of relative head rotation which is similar to the experiment.

After the peak of resisting equivalent moment is achieved, THUMS head was not coming back to its initial position due to inadequate muscular forces and chin chest reactions. Rather the angular rotation of head increased unto 140 deg of relative head rotation. This could be because of cadaver 1538 having a neck stiffer than the neck of THUMS . This magnitude of relative head rotation was observed in cadaver 1404 neck response which had the most flexible neck among the cadavers, with maximum relative head rotation of 100 deg for same test conditions.

The result showed considerable improvements in the model behavior when Hill material model was incorporated in the neck muscles of THUMS. The nature of the overall equivalent moment with head rotation response of the model shows a good agreement with the experimental corridor. Peak values matched but the area under the response curve deviated from the corridor.

The peak value of head rotation had improved to be 60 degree which is as reported by Mertz et al (1971). The head also whipped back after reaching a rotation limit

Low speed rear impact simulation

Speed selected for simulating test conditions was 4 km/h which wass the same as in the experiments.

The parameters that had been tracked in simulation were sled acceleration, head acceleration; thoracic spine acceleration, frontal chest acceleration and cervical vertebrae motion analysis. The motions of entire cervical vertebrae were represented by the changes in the relative rotational angle and translation of the third cervical vertebra from the sixth cervical vertebra.

The deceleration pulse given as an input to the sled was generated based on the experimental response data reported by Ono et al[12] (Figure 12).

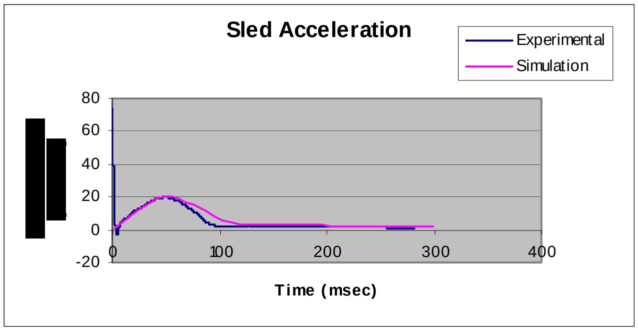


Figure 12. Acceleration Pulse given to the sled from low speed rear impact simulation, Koshiro Ono (1997).

With the test conditions above, THUMS was simulated for 300 msec termination time. Sixty stages have been created at increment of 5 msec per stage. Figure 13 shows the movements in THUMS after 0, 100, 150, 200, 250 msec respectively.

The head rotation curve shown in Figure 16 indicates that THUMS underwent larger relative head angulations, 32 deg as compared to 20 deg for cadaver. After the peak of head rotation is attained, in THUMS the head is not restored to its initial position but angular rotation of head kept on increasing unto 40 deg of relative head rotation. This could be because of less stiff neck of THUMS or improper muscle model in the neck.

In Figure, the time history of equivalent moment for THUMS and that from the experiments has been compared. The peak values of both positive and resisting moments are much higher than the experimental data. For experimental run, maximum equivalent moment of 8 Nm is observed while for THUMS this is coming out to be 25 Nm. Peak resisting moment in experimental data is 3 Nm and for THUMS this is 14 Nm.

In Figure 14, acceleration response of head of THUMS has been compared with that of experimental data.. Peak experimental value achieved is 22m/sec2 whereas from simulation this is coming out to be 28 m/sec2. Rotational angle of C3 in crash condition has been compared for experimental and simulation results in Figure 16. In general, large variations can be seen between the experimental and simulation results. One of the main reasons for this, we feel, is that the neck muscles have not been modeled completely in Thums.

From the head rotation curve obtained for THUMS™ with Hill model, the peak value of head rotation is 17 degree, which is close to the value of 22 degree for volunteers. Another significant change that can be observed from the head rotation curve is the coming back of head after attaining the peak value of 17 degree.

Comparison has been made in head acceleration experimental data and simulation data. Nature of the both the curves are same and the peaks values are quiet same with 24 m/sec^2 for THUMS and 23 m/sec^2 for volunteers.

The time history of equivalent moment for THUMS and that from the experiments has been compared. The peak value of positive moment is 7.5 N-m for experimental data and 9 N-m for THUMS. The peak resisting neck moment value for THUMS neck is 4 N-m whereas for volunteer it is 2 N-m. The nature of the curve is same as that of experimental data.

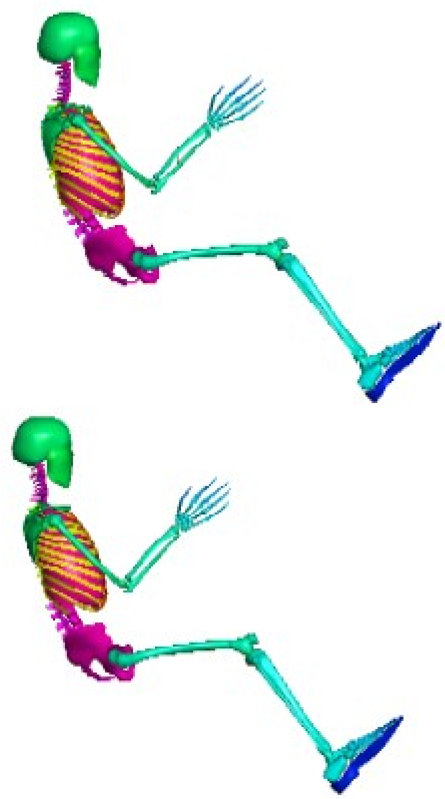


Figure 13. Movements in THUMS after 0 msec, 100 msec, 150 msec, 200 msec and 250 msec respectively.

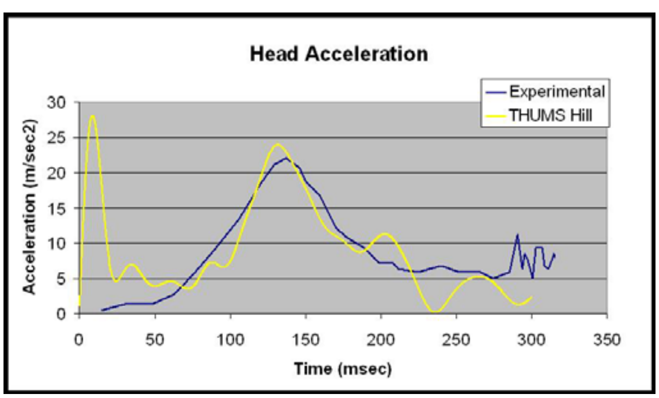


Figure 14. Acceleration response of head of THUMS from simulation.

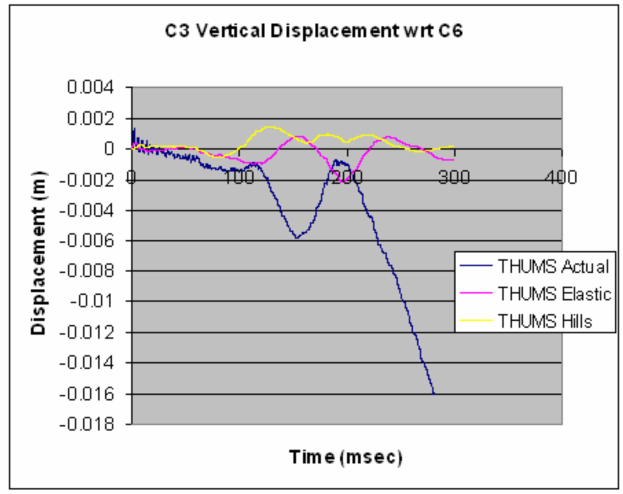


Figure 15. C3 motion relative to C6 – Vertical translation.

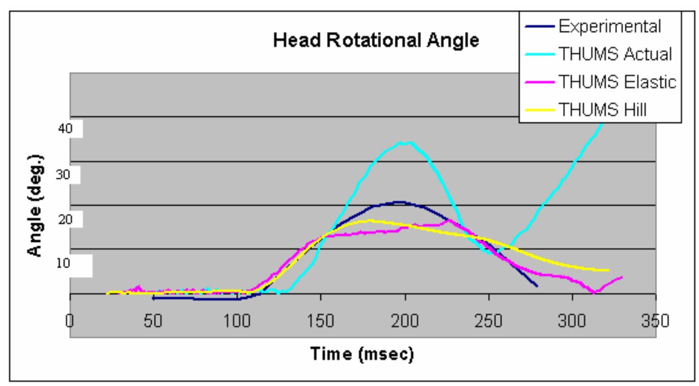


Figure 16. C3 motion relative to C6 – Rotational angle.

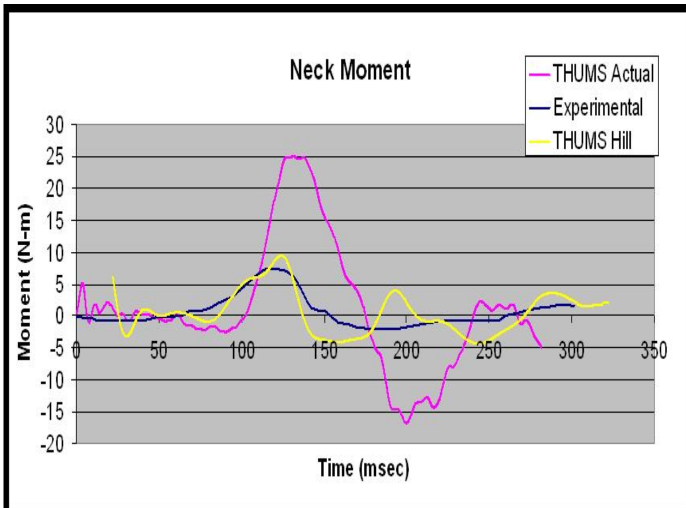
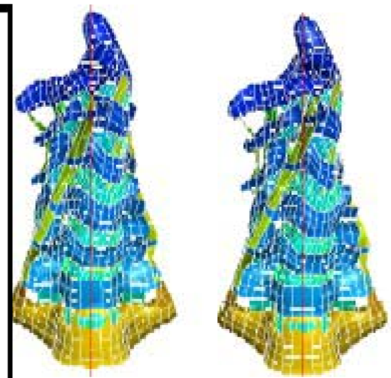


Figure17. Time history of neck moment.



ure 18. Movements in THUMS neck after 0 ec, 100msec, 200 msec and 300 msec respectively.

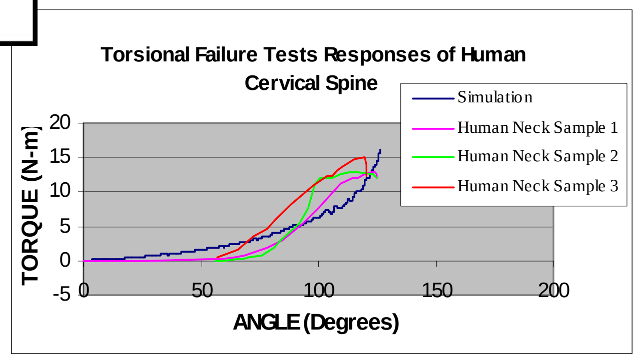
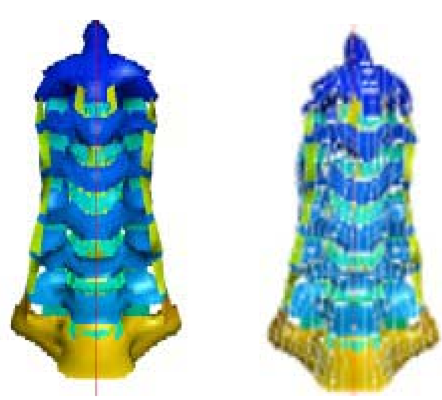


Figure 19. Torsion Failure Test Responses of Human Cervical Spine.

<u>Simulation for the neck in torsion</u>

Simulations have been done for the torsion tests conducted by Myers etal[15]. These include visco-elastic tests using relaxation and constant strain rate conditions. The specimen is loaded to failure by applying a ramp and hold at 500 degrees/sec. Relaxation tests use ramp and hold signals with 0.25-second rise times. The deflection is then held constant for next 150 seconds. Myers etal[15] also report a failure test, which has been categorized as high velocity failure tests using ramp to failure velocity displacements. The purpose of these tests was to provide a database representing the lower bound (No muscle action) of the stiffness of the human neck in rotation.

Figure shows the load to failure response of the experiments conducted on 3 human cervical spines reported by Myers and McElhaney[15] and the simulation response of THUMS neck. The results show a fair degree of correspondence.



DISCUSSION

In frontal impact simulations, the nature of the overall equivalent moment with head rotation response of the model shows a good agreement with the experimental corridor. But primarily the positive moment path of the model response is deviating from the corridor. In the neck model of THUMS this has been observed that many muscles have been modeled using bars elements, which have been assigned a null material model. This doesn't incorporate the forces coming either in tension or compression on the model. Because of the absence of these forces there is no effect of these muscles on the movement of neck. As a consequence of this the head of THUMS is not whipping back even after a relative head rotation with respect to the torso of more than 140 deg, (Figure 11).
This can be concluded that initially THUMS neck had a very less stiffness value and because of this actual THUMS had a 150-degree of Head rotation. The main cause for this looseness of THUMS neck was identified to be the improper material model used for modeling its neck muscles. There is no contribution of the muscle forces in actual THUMS. By changing the material model of neck muscles to elastic and then to

Hill material model, THUMS neck response has improves considerably.

In the low speed rear impacts, the sled acceleration, head acceleration and cervical spine motion have been compared between the experiment and the simulations. The compressive vertical motion plays an important role in minor neck injuries. In the rear impact simulations, downward and rearward extension motion of the C3 compared to the C6 has been observed, resulting in the cervical spine getting compressed in early stage of the impact Figure 15. Similar behavior was also observed in the experiments.

The motion of C3 in terms of rotational angle (Figure 16) and vertical translation as compared to C6 (Figure 15) reveals that the rotational angle of C3 increases over the time and reaches its peak around 150 ms after impact. After a drop of 10 deg in next 50 msec, it starts increasing again. The vertical translation of C3 wrt C6, on the other hand, reaches its first peak at 150msec after the impact and after a drop in its value for next 50 msec it starts rising again. This variation is however, missing in the experiments. This is primarily because neck muscles have not been modeled completely in the Thums neck model.

In the simulations that have been run for the failure tests in torsion on THUMS neck model; primary goal was the duplication of the in vivo kinematics and dynamics at the computer simulation level, as all the future work is based onto it. The centre of rotation is one such parameter which has been successfully identified in the THUMS neck model based upon the minimum energy method theorem(Myers and McElhaney[15]). The neck moment results obtained from THUMS show a good conformity with experimental results. The mean value of THUMS neck stiffness lies in the range of 0.472 Nm/degree in the high stiffness region.

These simulations have given us a good insight into the THUMS neck model and also requirements needed from human body FE models in general. The cervical neck is an extremely complicated joint, and its FE modeling is an arduous task. We have run numerous simulations to study the importance various aspects of these simulations. On the basis of these we are now able to highlight various aspects of these models, which need further attention for a closer validation under different conditions.

THUMS neck needs to be modeled in greater detail, especially with greater care for the muscles and tendons. Also, appropriate pre-tensioning needs to be included for these elements.

No failure model is defined for any of the parts involved in available model of THUMS, though inclusion of failure model for ligaments has been reported in later versions of the THUMS model of some other body parts. In the current model, elements continue to stretch endlessly under load, without failure / rupture.

Similarly, the material model of the soft tissues as well as that of ligaments is found to be critical. Variations in geometry as well as properties from cadaver to cadaver, repeatability of the experiments and establishing appropriate corridors in these experiments are other issues, which need to be addressed. The simulation results suggest that the properties of ligaments and muscles need to be verified and implemented with greater care. This is particularly important as ligament injuries are of considerable interest in most situations.
Neck muscles can alter the head and neck kinematics during frontal and rear end impact. Reflex time, activation level, co-contraction and the initial activation of the muscles can influence the head and neck motion. Additionally, initial seating posture and head restraint position influence the global and local head neck response in a rear end impact as was observed in the simulation results. Therefore, for accurate THUMS validation exact information on muscle activation, seating posture and position of seat and head restraint is essential.

To summarize, in this paper we have verified the THUMS cervical spine model against three sets of experimental data available in literature. The model validates well in some cases but is found lacking in some others. The reasons for the same have been discussed and possible directions for improvement have been suggested. These include better material models for soft tissues, better muscle model, better failure / rupture models, better contact interfaces and inclusion of more details in the neck model, to name a few. We are currently investigating most of these issues and would have more suggestions in these areas in the months to come.

ACKNOWLEDGEMENT

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