FIDELITY OF ANTHROPOMETRIC TEST DUMMY NECKS IN ROLLOVER ACCIDENTS

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ABSTRACT

Rollover accidents pose a serious cost to society, while they account for 10% of all passenger car accidents they cause about 20% of Harm (Harm is defined as the sum of all injured people each weighted in proportion to the outcome, as represented by the cost of the person's most severe injury). This is primarily due to serious head and neck injuries resulting in permanent disability or death. The literature regarding neck injury in rollovers can be divided into two groups, one that attributes neck injury to significant roof crush and the vertical excursion due to the vehicle belt system. The other who subscribes to the "diving injury" and "torso augmentation" theory and conclude that roof crush is inconsequential. The Hybrid-III test dummy has been used in rollover and drop tests to support this conclusion.

This paper will show that the Hybrid-III head and neck complex is significantly stiffer and less flexible than the human neck. This lack of biofidelity means that the Hybrid-III neck cannot provide meaningful data in rollover environments.

INTRODUCTION

The human spine is a complex mechanical structure composed of bony vertebrae, ligaments and intervertebral discs. Its primary function is to protect the spinal cord and nerve roots while carrying loads and allowing physical motion. Inherent in the spinal structure are natural curvatures throughout the spine in each of the major spinal sections: cervical, thoracic, lumbar and sacral. The spine's curvature provides flexibility and shock absorption.

The Hybrid-III neck is designed to mimic the cervical portion of the human spine and has a rigid attachment to the torso and the head. The neck is a one piece column made of butyl elastomers separated with aluminum discs. Through the center of this column runs a single steel cable. There is no inherent curvature to the Hybrid-III neck column.

The Hybrid-III dummy is the most widely accepted dummy for automotive crash testing in the world. Its response resembles that of a human more closely than any previous anthropometric test dummy (ATD) before it. The Hybrid-III neck was designed to meet flexion and extension criteria established by Mertz[1] in a series of sled tests involving volunteer and cadaveric subjects. Other neck response modes were secondary considerations in the development of this neck. The neck response corridors are a relation between the moment about the occipital condyles and the angular position of the head relative to the torso. These requirements are not adequate to properly reproduce human motions. Mertz et. al. [2]and Melvin et al.[3] have suggested neck requirements that relate to the head trajectory as well. Further refinements in response requirements and neck design are required in order to achieve more biofidelic and useful test dummies.

LITERATURE

The Hybrid-III was designed to perform primarily in frontal impacts. However, the Hybrid-III head and neck complex displays significant differences from human volunteers even in this mode. Seemann, et al. [4] conducted tests at the Naval Biodynamics Lab that studied the responses of human volunteers and the Hybrid-III in various deceleration directions. Comparing the head trajectories for the two test groups, it became apparent that the downward travel and timing of the human head in 15g frontal sled tests differed dramatically from Hybrid-III. The human head traveled farther downward and over a longer period of time, while the Hybrid-III head rebounded faster after translating downward a smaller distance. It was noted that the Hybrid-III head vertical displacement was less than half the displacement of one volunteer. Similarly, when subjected to 12g vertical accelerations, the Hybrid-III neck was too stiff. Seemann concluded that the Hybrid-III neck is too stiff and is not biofidelic in flexion.

Early studies to determine the strength of the human cervical column were conducted with vertical drops and conducted by Nusholtz and Huelke[5] who dropped intact cadavers both constrained and unconstrained from drop heights up to 1.8 meters. Various cervical fractures were observed, however those with potential spinal cord injury involved fractures of vertebral bodies C-4 and C-5, fracture of dens and subluxation of lower cervical elements. This study indicates that 1.1 to 1.8 meter drop heights are required to produce serious cervical injury.

Vertical drop studies of intact cadavers by Yoganandan and Sances[6] were conducted with restrained upper cervical columns and unrestrained preparations. The force of impacts on the head was recorded along with force from gauges implanted in the cervical vertebral spaces and accelerometers along the spine. Photo targets at various vertebral bodies and the occipital were also used. The cervical fractures resulting in probable clinical injury included anterior subluxation of C-5 and C-6 with locked facets, Jefferson fracture of C-1, a C-3 fracture and a C-2 vertebral body lamina fracture. These injuries occurred at drop heights between 0.91 and 1.5 m. Cervical compression of 2 to 4 cm were observed. A force of 2600 N was recorded in the cervical column gauge with a C3 fracture. This supports the drop height range required for cervical injury discussed by Nusholtz and Huelke[5].

McElhaney[7] studied injuries from swimming pool accidents and attempted to replicate the drop heights for cervical injury from trajectory and velocity profiles of test subjects. He extrapolated cervical injuries from his simulations ranging from 3.3 m/s to 7 m/s for edge of pool dives. This study was however conditioned upon the history of the accidents and based on the analogy of injuries in swimming pools in contrast to non-aquatic situations.

Pintar[8] produced clinical type cervical injuries in the vertically oriented human cadaver head/neck preparation with the head stabilized with a spring in the posterior area and dead weights in the anterior area. The study described the need for preflexion (the removal of the resting lordosis) to create vertebral compression injuries. The preparations were impacted with a high speed piston at the top of the head with the cervical column aligned vertically and forces recorded in the region of T-1. Fractures produced were of compression or compression-flexion type and also included partial subluxation of C-5 on C-6. Burst fractures and anterior wedge body fractures were produced. This study included parallel investigations with the Hybrid-III head and neck. The Hybrid-III head and neck complex transmits 3 times as much force from the head to the distal point at T-1. Approximately 80 to 90 percent of the force applied to the head of the Hybrid-III was measured at the distal plate at T-1, while forces in the region of 25 to 30 percent of those applied to the head of the human preparation from the piston were measured in the distal force plate at the T-1. These experiments indicate that the Hybrid-III transmits force markedly different than the human/neck complex because of its stiffness. The quasi-static biomechanical comparison of the axial compressive characteristics of the human cadaveric neck and the Hybrid-III conducted by Yoganandan and Sances[9] indicate that the Hybrid-III neck is 3 to 5 times stiffer at loading rates up to 0.25 m/s.

Subsequent studies conducted by Pintar and Yoganandan[10] determined the dynamic characteristics of the human cervical column based on twenty cadaveric specimen loadings under axial compression. The corridors indicate an initial soft response of the human head/neck preparation which commences to stiffen following approximately 10 to 15 millimeters of deformation reaching an average stiffness of approximately 555 N/mm at speeds of 2.5 to 8 m/s with a mean force of 3326 N at failure. Recent studies by Sances and Voo[11] found that the Hybrid-III head/neck preparation was at least 2 times stiffer compared to the human under dynamic axial compressive loading. Dynamic forces and curves in the Hybrid-III necks in comparison to the cadavers of Pintar are shown in Figure 1. Typically the stiffness of the Hybrid-III was 958 N/mm at 2 m/s, 1111 N/mm at 4 m/s and 2160 N/mm at 7.6 m/s.

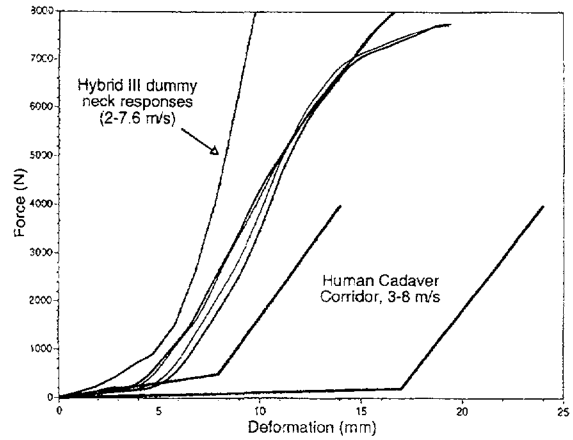

Figure 1. Dynamic loading of Hybrid-III and human cadaveric necks

A recent study by Nightingale and McElhaney[12] showed fractures of the cervical column of the human cadaver with a 32 pound weight attached to the torso with vertical drops of the head/neck complex onto a force plate. The experiments demonstrated cervical fractures at impact speeds of approximately 3.2 m/s. However, this study does not account for the flexibility of the thoracic spine which will decrease the overall stiffness of the preparation. Larger drop heights will be expected to receive similar cervical fractures due to substantial thoracic telescoping and decrease in overall stiffness of the spine. Marked thoracic bending was also observed in high speed photographs of subjects dropped vertically.

In another investigation, McElhaney, et al. (1983)[13] conducted tests on Rhesus monkeys and unembalmed cadaver cervical spines. They concluded that there is a significant deformation rate dependence and that small eccentricities (±1cm.) in the load axis could change the buckling mode from posterior to anterior. A 1 cm. difference in alignment created flexion, compression or extension injuries. McElhaney[14] in 1988 reported that the Hybrid-III dummy neck was significantly stiffer than the human cadaveric cervical spine which had quasi-static bending loads applied. The study also included compression flexion, tension flexion and compression lateral bending. The bending stiffness was significantly influenced by direction of the bending moment and the type of end restraint. These experiments indicate that when the loading is eccentric (as it almost always is), the primary deformation mode is bending.

Myers, et al.[15] reported the Hybrid-III neck is substantially stiffer than the human cadaver with various end conditions. Myers concluded that the risk of cervical injury may be strongly dependent on the degree of head restraint imposed by the contact surface. He suggests that injury environments should be designed to minimize this constraint. In the absence of significant constraint, the stiffness of the structure was low, the spine was able to flex significantly and no injuries were observed.

Myers concluded that the Hybrid-III neckform was 10 to 50 fold stiffer than the human cadaver cervical spine depending on the imposed end condition (10 times for fully constrained; 12 times for rotationally constrained, 50 times for unconstrained). Myers further concludes that the availability of a high biofidelity head-neck for compression would be a great benefit to the design and evaluation of safety equipment. Also, since the imposed end condition reliably alters the failure mechanism of the cervical spine, safety equipment and injury environments should be designed to minimize the degree of imposed constraint on the head.

The risk of cervical injury can be avoided if the loading is not transmitted to the cervical spine secondary to escape of the head from the loading surface. This was observed in the vertical drop studies by Yoganandan and Sances and others.[16, 17] The studies of Pintar and Nightingale have determined an approximate tolerance level for the human spine in the region of approximately 4 kN in compression. For vertical drops the combined effects of the head-neck plus torso tissues will undoubtedly reduce the transmitted force to the cervical spinal column when the additional displacement of the thoracic column is included in the tested preparation. Consequently, one would expect substantially greater drop heights for injury similar to those observed by Nusholtz[18] and Yoganandan[19] for an intact preparation. In rollovers there is a high probability of bilateral locked facets which usually occurs in the absence of bony damage and produces serious reductions in the spinal canal with concomitant spinal cord trauma. These injuries most likely occur secondary to force imposed upon the region of the head posterior to the vertex during flexion of the head-neck complex. Flexion of the head is a likely protective mechanism invoked by an occupant during a rollover event.

One can demonstrate that the intact human can substantially increase the head to roof clearance by flexing the head-neck complex and/or rotating the torso. This increase in head to roof clearance has substantial potential for injury reduction. (See Figure 2)

Figure 2.

Cadaveric studies should be weighted because of the disproportionate strength of the older cervical elements. Yamada[20] has shown that a 40 to 50 percent decrease in the strength of vertebral bodies in the geriatric population compared to a 26 year old. Also ligaments decrease in strength with age. Therefore greater drop heights would be required for fractures of vertebral bodies preparation in the younger population than the 60-80 year old preparations often used for cadaveric studies.

Early human studies by Ewing and Thomas[21] have demonstrated that the head lags the body in frontal or rear collisions. The Hybrid-III does not exhibit this behavior. Various studies have been done to show that head lag occurs with human studies during rear sled impacts. Cadaveric preparations of the head and neck complex demonstrated that the head lag produced an S-curve in the cervical column for either frontal or rear impacts. This characteristic is not replicated by the Hybrid-III head and neck complex.

A concerted effort is being made by the department of Transportation to develop an improved dummy neck in concert with others because of the lack of biofidelity of the Hybrid-III neck both in flexion/extension and axially. The National Highway Traffic Safety Administration (NHTSA) has set out to develop a more biofidelic frontal impact dummy. NHTSA[22] concluded, after conducted tests at its Vehicle Research and Test Center, that the Hybrid-III neck was stiff when compared with the Naval Biodynamics Laboratory volunteer data. The most important missing characteristic was the lack of head lag exhibited by the volunteers when the neck went into flexion. The location and timing of the head motion relative to the vehicle interior is important to vehicle impacts. Their solution for a new neck was to create a neck that was less stiff than the Hybrid-III and a spring/cable exterior to the neck was needed to produce the proper head lag. Researchers added design goals to included biofidelity in frontal and lateral flexion, extension and axial compression. One prototype was designed and tested by GESAC. Also a computer simulation using DYNAMAN was created for testing and development . The prototype neck's axial stiffness was 400 N/mm versus 560N/mm for the Hybrid-III and 159N/mm for cadavers.

In order to understand the effectiveness of head restraints, TNO Crash-Safety Centre[23] has started to develop the Rear Impact Dummy (RID). TNO determined current crash test dummies are not biofidelic for low severity rear impacts.

A lack of biofidelity of the Hybrid-III is attributed to excessive high resistance to bending of the neck and the torso in low severity impacts. Minimum requirements for the RID neck are an accurate representation of the human head rotations and neck bending, otherwise the head might impact the headrest in a different location or even will have no contact with the headrest or interior at all. Since the Hybrid-III lower spine is rigid, any new retrofitted neck must incorporate the flexibility of the thoracic spine as well as the cervical portion it physically represents. The Hybrid-III neck was concluded to be too stiff for the low severity rear impact environment since a majority of occupants do not anticipate rear impacts and therefore have not tensed their neck muscles. TNO has developed two prototype necks that retrofit to the Hybrid-III that appear to be first steps toward developing a true rear impact dummy.

DeSantis [24] discusses the development of a prototype multi-directional neck (MDN) by Transportation Research Center under contract to NHTSA which targets biofidelic motion in both frontal and lateral impacts. The MDN is designed with two ball-and-socket joints which connect the head and thorax with a rigid link. The neck is designed to retrofit onto the Hybrid-III and the Advanced Dummy. A series of tests were conducted on the prototype neck and compared to the Hybrid-III neck and the volunteer data from the Naval Biodynamics Laboratory. It is apparent that from this papers that, especially at low speeds, that the Hybrid-III neck remains on the stiffer side of the corridors and sometimes outside of them relative to human motion. The MDN neck demonstrated improved biofidelity in both frontal and lateral impacts when compared to volunteer data.

REAL WORLD INJURY PATTERN

As part of the work done by the Liability Research Group, we have studied the injuries received by occupants in motor vehicle accidents. Over a period of 6 years, 1992 to 1998, we have studied in depth, 124 cases of serious rollover incidents. Of these, 57 people suffered major neck injuries. In this sample, the medical examinations revealed 29 flexion type injuries, 11 combination flexion-compression injuries, 3 compression injuries, 1 rotation type injury and the remaining 13 cases no determination was made (some due to lack of comprehensive autopsy). The majority of these cases, 40 of 44 cases (91%), exhibited a flexion type or flexion-compression injury pattern,. The dummy studies conducted by Moffatt, et al.[25], [26] are inconsistent with this real world injury data.

One explanation for this discrepancy lies in the lack of biofidelity of the Hybrid-III neck in a rollover accident mode. In a rollover accident, a human neck will usually rotate itself into a non-aligned arrangement to relieve pressure. Injury occurs when the head and neck is pushed beyond its limits or in rare cases when the cervical spine is aligned without the natural curvature and compressed by approximately 15-25 mm. For the Hybrid-III neck, the force is transmitted straight through the neck and the neck load cells record high axial loads with small deflections and prior to any significant bending.

Friedman et al.[27] performed a set of experiments that showed that with a pre-flexed head in contact with a rigid roof, the human volunteer was uninjured at drop heights of 15 cm. A Hybrid-III neck would likely have registered injurious loads. Friedman and Friedman[28] have conducted a series of drop tests where a Hybrid-III dummy in an inverted rollcaged vehicle recorded injurious levels of load in the neck (similar to the studies conducted by Moffatt et. al) while a human volunteer, in the identical environment, was unhurt.

CONCLUSIONS

The aforementioned studies and continued evolution of dummy necks presents clear evidence that the Hybrid-III neck does not provide adequate biofidelity. Despite the overwhelming evidence, a recent publication interprets the works of other authors and suggests that the Hybrid-III is biofidelic.[29] Under compressive loading, the Hybrid-III neck compares closest to the human neck only once the human spine has the lordosis removed, the load path is within 1 cm of the spine's vertical axis and with the spine constrained. Even in these conditions, it remains many times stiffer than the human spine. The human spine's stiffness and load bearing characteristics change dramatically when natural or induced spinal curvature is included, the load path is off center more than 1 cm or the flexibilty of the thoracic spine is considered. The Hybrid-III neck becomes more than an order of magnitude stiffer than human spine under these conditions.

A properly biofidelic neck must represent, as accurately as possible, a human neck in all directions. This will insure accurate motion and loading to the dummy. Due to excessive stiffness in all directions, the Hybrid-III neck cannot provide meaningful data in environments in which accelerations and loadings come from multiple directions such as those seen in rollovers. Previous studies[30, 31] have used the Hybrid-III neck to draw the conclusion that in rollovers, roof strength is not casual to neck injury. Injury rates occurring in these rollover tests were two orders of magnitude more frequent than that seen in the real world accidents. This can be directly attributed to the lack of biofidelity of the Hybrid-III neck, and its tendency to over-represent axial compression injuries. We believe that if the Malibu studies were conducted with a biofidelic neck, the results would support a relationship between roof crush and injury. Recent drop tests[28] have reinforced that the Hybrid-III dummy overpredicts injury in humans.

Important considerations for future dummy necks are:

- reproducing head trajectories comparable to humans;
- incorporate geometry representative of the human curvatures and structure;
- representation of the thoracic spine characteristics;
- replicate rate sensitivity, load transmission, stiffness and bending characteristics.

REFERENCES

[1] Mertz, H.J., Patrick, L.M. *Strength and Response of the Human Neck*, SAE 710855, 1971.

[2] Mertz, H.J., Neathery, R.F. and Culver, C.C. *Performance Requirements and Characteristics of Mechanical Necks*, Human Impact Response, 1973.

[3] Melvin, J.W., McElhaney, J.H., Roberts, V.L., *Evaluation of Dummy Neck Performance,* Human Impact Response, 1973.

[4] Seemann, M.R., Muzzy III, W.H., Lustick, L.S. *Comparison of Human and Hybrid-III Head and Neck Dynamic Response*, SAE 861892, 1986.

[5] Nusholtz G.S., Huelke D.E., Lux P., Alem P., Montalvo F. *Cervical Spine Injury Mechanisms.* Proceedings of the 27th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale, PA, 1983, pp. 179-198.

[6] Yoganandan N., Sances A. Jr., Maiman D., Myklebust J., Pech P., Larson S. *Experimental Spinal Injuries with Vertical Impact.* Spine, 11(9): 855-860.

[7] McElhaney J., Snyder R., States J., Gabrielsen M. *Biomechanical Analysis of Swimming Pool Neck Injuries.* SAE/SP-79/438, 1979.

[8] Pintar F.A., Sances A. Jr., Yoganandan N., et al. *Biodynamics of the Total Human Cadaver Cervical Spine.* Proceedings of the 34th Stapp Car Crash Conference, Orlando, FL, 1990, pp. 55-72.

[9] Yoganandan N., Sances A. Jr., Pintar F.A. *Biomechanical Evaluation of the Axial Compressive Responses of the Human Cadaveric and Manikin Necks.* Journal of Biomechanical Engineering, III (3): 250-255, 1989.

[10] Pintar F.A., Yoganandan N., Voo L.M., Cusick J.F., Maiman D.J., Sances A. Jr. *Dynamic Characteristics of the Human Cervical Spine.* SAE Transactions 104 (6): 3087-3094, 1995.

[11] Sances A. Jr., Voo L.M. *Biofidelity of the Hybrid III Neck for Spinal Trauma Assessment.* ASME Advanced Bioengineering. BED-Vol. 36: 249-250, 1997.

[12] Nightingale R., McElhaney J., Camacho D., Kleinberger M., Winkelstein B., Myers B. *The Dynamic Responses of the Cervical Spine: Buckling, End Conditions, and Tolerance in Compressive Impacts.* SAE 41st Stapp Car Crash Conference, 1997 Proceedings, SAE 973344.

[13] McElhaney, J.H., Paver, Jacqueline,G.P., McCrackin, H.J., Maxwell, G.M., *Cervical Spine Compression Responses,* SAE 831615, 1983.

[14] McElhaney J.H., Doherty B.J., Paver J.G., Myers B.S., et al: *Combined Bending and Axial Loading Responses of the Human Cervical Spine.* Proceedings of the 32nd Stapp Car Crash Conference, Atlanta, GA, 1988, pp 21-28.

[15] Myers B.S., McElhaney J.H., Richardson W.J., et al: *The Influence of End Condition on Human Cervical Spine Injury Mechanism.* Proceedings of the 35th Stapp Car Crash Conference, San Diego, CA, 1991, pp 391-399.

[16] Yoganandan N., Sances A. Jr., Maiman D., Myklebust J., Pech P., Larson S. *Experimental Spinal Injuries with Vertical Impact.* Spine, 11(9): 855-860.

[17] Nightingale R., McElhaney J., Camacho D., Kleinberger M., Winkelstein B., Myers B. *The Dynamic Responses of the Cervical Spine: Buckling, End Conditions, and Tolerance in Compressive Impacts.* SAE 41st Stapp Car Crash Conference, 1997 Proceedings, SAE 973344.

[18] Nusholtz G.S., Huelke D.E., Lux P., Alem P., Montalvo F. *Cervical Spine Injury Mechanisms.* Proceedings of the 27th Stapp Car Crash Conference, Society of Automotive Engineers, Warrendale, PA, 1983, pp. 179-198.

[19] Yoganandan N., Sances A. Jr., Maiman D., Myklebust J., Pech P., Larson S. *Experimental Spinal Injuries with Vertical Impact.* Spine, 11(9): 855-860.

[20] Yamada H. *Strength of Biological Materials.* Williams Wilkins, July 1969, p. 297.

[21] Ewing C.L., Thomas D., Lustick L., Muzzy W.H., et al. *The Effect of Duration, Rate of Onset and Peak Sled Acceleration on the Dynamic Response of the Human Head and Neck.* Proceedings of the 20th Stapp Car Crash Conference, Dearborn, MI, Society of Automotive Engineers, Inc., 1976.

[22] White, R.P. Jr., Zhao, Y., Rangarajan, N., Haffner, M., Eppinger, R., Kleinberger, M. *Development of an Instrumented Biofidelic Neck for the NHTSA Advanced Frontal Test Dummy.* ESV, 96-S10-W-19, 1996.

[23] Thunnissen, J.G.M., Van Ratingen, M.R., Janseen, E.G. *A Dummy Neck for Low Severity Rear Impacts.* ESV, 96-S10-O-12, 1996.

[24] DeSantis, Kathleen, *Development of an Improved Multi-Directional Neck Prototype.* SAE 912918, 1991.

[25] Moffatt, E., Orlowski, K.F., Stocke, J.E., Bahling, G.S., Bundorf, R.T., and Kaspzyk. *Rollover and Drop Tests - The Influence of Roof Strength on Injury Mechanics Using Belted Dummies.* SAE 902314, 1990.

[26] Orlowski, K.F., Bundorf, R.T., Moffatt, E.A. *Rollover Crash Tests- The Influence of Roof Strength on Injury Mechanics.* SAE 851734, 1985.

[27] Friedman, D. and Chng, D. *Human Subject Experiments in Occupant Response to Rollover with Reduced Headroom.* SAE 980212, 1998.

[28] Friedman, K. and Friedman, D. *Upper Interior Head, Face and Neck Injury Experiments.* ESV 1998, Paper Number 98-S8-P-11.

[29] Piziali, R., Hopper, R., Girvan, D. and Merala, R. *Injury Causation in Rollover Accidents and the Biofidelity of Hybrid III Data in Rollover Tests.* SAE 980362, 1998.

[30] Moffatt, E., Orlowski, K.F., Stocke, J.E., Bahling, G.S., Bundorf, R.T., and Kaspzyk. *Rollover and Drop Tests - The Influence of Roof Strength on Injury Mechanics Using Belted Dummies.* SAE 902314, 1990.

[31] Orlowski, K.F., Bundorf, R.T., Moffatt, E.A. *Rollover Crash Tests- The Influence of Roof Strength on Injury Mechanics.* SAE 851734, 1985.