Journal of the National Academy of Forensic Engineers®

http://www.nafe.org
ISSN: 2379-3252

Vol. XXII No. 1 June 2005

Forensic Engineering Analysis of Head Impacts within a Vehicle Subject to Side Impact

by Laura L. Liptai, Ph.D. (NAFE C339)

Abstract

Asymmetric design of seatbelts does not limit potentially injurious contact with vehicle interior in opposite side motor vehicle collisions. In opposite side impacts, of approximately 65 to 70 degrees, the lap and shoulder restraint functions less effectively. At these angles, with a change in velocity of greater than or equal to 20 mph, the Head Injury Criterion (HIC) may exceed tolerance when the occupant's head impacts unyielding structural surfaces within the vehicle. BioMedical engineering analysis should be conducted to determine the likelihood of sustaining brain trauma even when using a seatbelt in these scenarios.

Introduction

The objective of this work most generally is to study side impact head mechanics of impulsive impact to the cranium within the automotive environment. *Mechanics* is defined as the science that describes the actions or effects of forces and couples—two parallel, equal, noncolinear and oppositely directed forces—on solid or fluid objects and systems[1].

This research was undertaken to determine if a head injury would be sustained by a restrained driver in an opposite side impact of 10 and 20 mph changes in velocity with a principal direction of force of 65 to 70 degrees. Principal direction of force of zero degrees is defined as straight ahead while seated in the impacted vehicle.

To provide a broader perspective, the magnitude of the head injury problem is enormous: in the United States alone, upward of 50 million people are injured yearly, with automobiles accounting for about 5% of this number that led to 41,471 fatalities (Federal Highway Administration, 1998) with about 70% of these involving the head. One million Americans are annually treated and released for brain injury in emergency rooms: 230,000 are hospitalized and survive; 50,000 die; and 80,000-90,000 experience long-term disability (Center for

Laura L. Liptai, Ph.D., BioMedical Engineer and Mechanical Engineer,
Biomedical Forensics, 1660 School Street, Suite 103, Moraga, CA 94556

Disease Control, 1996)[1]. Here the research is further focused on non-penetrating brain injuries in motor vehicle collisions. These are generally produced by impact, by dynamic (impulsive) loading of the cranium due to forced motion.

According to a study by Malliaris[2], 80% of trauma traces to interior impact. In the instance of opposite side impacts, the occupant restraints are increasingly less effective in preventing interior contact as the principal direction of force increases or as the impact becomes more lateral. This is due in large part to the asymmetry of the lap and shoulder restraint system that functions less effectively in lateral collisions.

Lap and shoulder seat belts are designed to optimally protect occupants in frontal collisions. The frontal impact occupant loads are distributed through the seat belt webbing to the chest, iliac spines of the pelvis and the lower abdomen. In the frontal collision, the restraint system helps limit occupant excursion which reduces potentially harmful contact with the vehicle interior. In the same side impact, the trauma is sustained in greatest part to the occupant striking the side of the vehicle interior and/or excursion out through the side window. Side impact rule making, Federal Motor Vehicle Safety Standard (FMVSS) 214 focuses on occupant protection in same side impacts. This research work focuses on opposite side impacts. Due to the geometry of the lap and shoulder harness system, it is not uncommon for the thorax of lap and shoulder restrained occupants to come out of the seat belt in opposite side motor vehicle collisions. The seat belt is not optimally designed for protection in these types of collisions. Work by Herbert, *et al*. (1976) and Horsch (1980) showed that the limit of retention of the torso was when the direction of the crash force was at 45 degrees from straight ahead[3]. With sufficient force, occupants commonly sustain trauma when their heads contact the vehicle interior when the thorax comes out of the shoulder portion of the seat belt. Specifically, opposite side occupants with head injuries of AIS ≤ 2, come out of the shoulder section of the seat belt 35% of the time[3].

This studies the head accelerations of the opposite side occupant when contact with the adjacent seat occurs at vehicular changes in velocity of 10 and 20 mph in an effort to quantitatively assess the potential for head trauma in opposite side impacts. Principal directions of force of 65 to 70 degrees from straight ahead were utilized. At these side impact angles, the shoulder harness does not remove significant energy from the upper part of the body by the seat belt before the thorax slips out of the restraint[3]. Therefore, the human system was modeled by an inverted pendulum subsystem test design of different heights to generate the vehicular changes in velocity.

Figure 1
Rotation of the occupant that causes impact with the opposite seat during opposite side impact collisions at ΔV = 15 mph ± 5 mph and PDOF 65 – 70 degrees in an exemplar van

Experimental and Theoretical Methods

The history of the force, its direction and location of application is determined. The 10 and 20 mph change in velocity (ΔV) opposite side impacts with a principal direction of force (PDOF) 65-70 degrees rotate the body and cause the head to contact the base of the seat in the seating configuration of the late 1990's American manufactured van. A kinematic study with a driver moving at 65-70 degrees illustrates seat contact in Fig. 1.

First a subsystem test is designed to collect the accelerations of the center of the head while impacting the seat. Then Lagrangian dynamics calculates the impact speed of the head at the point of contact with the seat at different vehicular changes in velocity. Then, the height of the inverted pendulum is determined such that the impact speed of the head to the seat is obtained for 10 and 20 mph vehicular changes in velocity. The experiments are conducted, and the data acquired, analog to digital converted, filtered and analyzed. The Head Injury Criteria is calculated and compared to tolerance.

The subsystem design and apparatus, digital data acquisition, instrumentation, acquisition and analysis are discussed as methodology.

A. Subsystem Design and Apparatus

The 11-foot and 21-foot inverted pendulums correspond to changes in velocity of 10 and 20 mph. **Figure 2** illustrates the how the 11-foot inverted pendulum comes into contact with the seat to match the contact shown in **Figure 1**. The 11

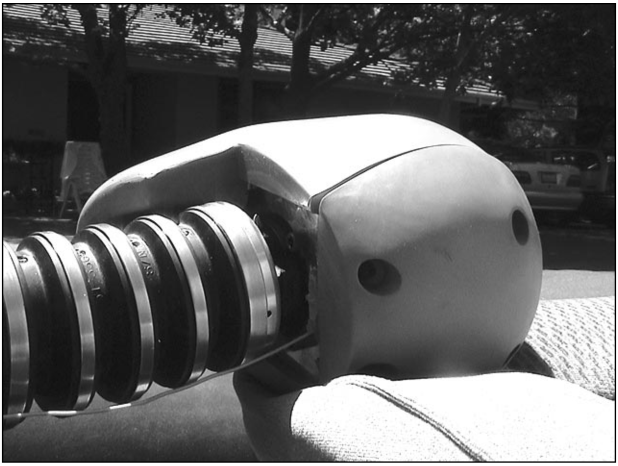

Figure 2
Hybrid III head attached to inverted pendulum impacts the seat at the same location as that shown in Figure 1.

foot pendulum replicates10 mph change in vehicular velocity. This inverted pendulum length is calculated by utilizing the formula shown in **Eq. (2)**. The 10 and 20 mph changes in velocities can be generated by the inverted pendulum subsystem test. The theoretical basis for this is detailed in Section B.

B. Theoretical Lagrangian Mathematical Modeling Analysis

Figures 3 and **4** are excerpts from the *Mathematica* code that was used in the Lagrangian mathematical analysis. **Figure 3** outlines the variables used in the analysis and Figure 4 is the set-up for the kinetic-co energy and potential energy equations. Equation 1 is the Lagrangian equation solved to determine acceleration.

$$\frac{\partial}{\partial t}\left(\frac{\partial T}{\partial \Theta'}\right) - \frac{\partial T}{\partial \Theta} + \frac{\partial V}{\partial \Theta} = 0 \qquad (1)$$

Equation 1
The Lagrangian equation solved to determine acceleration

Utilizing the acceleration equation from the Lagrangian analysis, this result can be integrated to determine the speed following a 90 degree rotation; or more specifically, the impact speed of the head into the base of the passenger seat. The head impact speed varies with the increasing change in velocity, or impact speed from opposing vehicle.

■ **Variable Declaration**

L = Length from buttocks to top of head (feet)
d = Chest depth (feet)
m = Mass of torso, neck, and head (slugs)
J_x = Mass moment of inertia
ω = Rotational velocity (rad/sec)
g = Gravity (ft/sec^2)
θ = Angle between torso relative to gravity (rad)
Speed = Initial velocity (ft/sec) due to impact

Figure 3
Variables used in the Lagrangian mathematical analysis

$$\omega = \theta'[t];$$
$$J = \frac{m}{12}\left(4\,L^2 + d^2\right);$$
$$T = J\,\frac{\omega^2}{2} + m\,\frac{Speed^2}{2}$$
$$V = m\,g\,L\,(1 - Cos[\theta[t]])$$
$$Coords = \{\theta[t]\};$$
$$Q = \{0\};$$
$$\frac{m\,Speed^2}{2} + \frac{1}{24}\,(d^2 + 4\,L^2)\,m\,\theta'[t]^2$$
$$g\,L\,m\,(1 - Cos[\theta[t]])$$

Figure 4
Equations used in the mathematical analysis to solve for the kinetic co-energy and potential energy and their outputs (non-bold text)

Using the change in velocity, the appropriate inverted pendulum length can be determined by utilizing Equation (2). In this equation, the inverted pendulum length is represented by the term, h. So by applying the changes in velocity of 10 and 20 mph, the pendulum lengths of 11 and 21 feet, respectively, can be computed.

$$\frac{1}{2} I\omega^2 = mgh \tag{2}$$

Equation 2
Equation utilized equating the potential energy of the inverted pendulum height to the kinetic rotational energy of the head initiated by the change in velocity of the vehicle

C. Experimental Digital Data Acquisition

The technique utilized to acquire the head acceleration at the center of gravity is the conversion of analog signals into digital representations or digital data acquisition. The ability to extract information from the digital numbers which accurately characterized the acceleration analog signal relies upon the fidelity of the digital data[4]. Sampling rate was selected at 10,000 samples per second for each of the x, y and z directions to control adequate digital data fidelity.

D. Experimental Instrumentation

The instrumentation consists of five parts: 1) accelerometer positioned at the center of gravity of the Hybrid III head to collect the data (rated at 500 g's), 2) Sensor Signal Conditioner that amplifies the signal, 3) Data Acquisition Board/Interface with Card to function as the analog to digital converter/computer interface, 4) computer to collect and store the digital data, and 5) filters to satisfy the sampling theorem and to eliminate aliasing effects on the data. Figure 5 illustrates the instrumentation utilized in the digital data and acquisition.

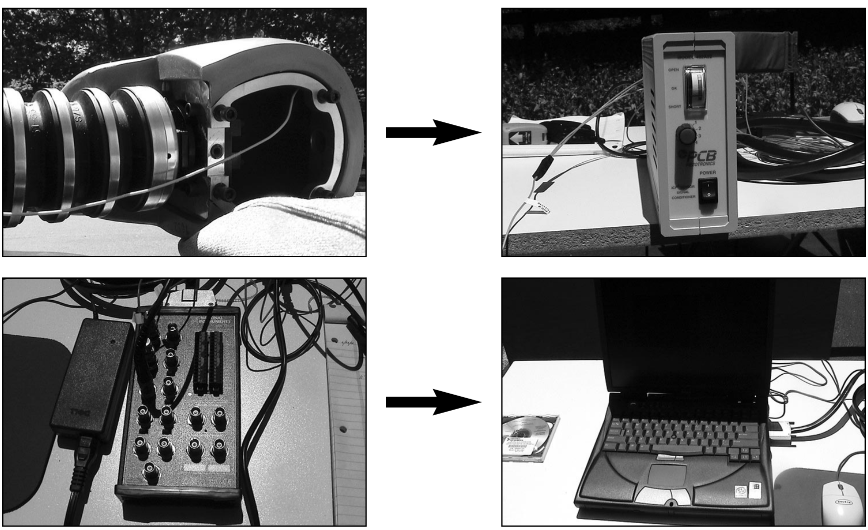

Figure 5
Instrumentation flow chart illustrated

E. Digital Data Acquisition and Filtering

The data acquisition and filtering flow diagram is illustrated as Figure 6. As illustrated, the x, y and z directions are collected and filtered individually. The resultant acceleration is calculated, and the HIC determined from this resultant value.

Criteria for Head Injury Measurement

Since the brain and blood tissue is so complex and inhomogeneous, failure limits for specific regions of the brain have not been established. According to Nahum and Melvin, brain failure can not clearly be delineated because physiologic dysfunction can occur at levels below those which mechanical disruption of neural tissues can occur[5].

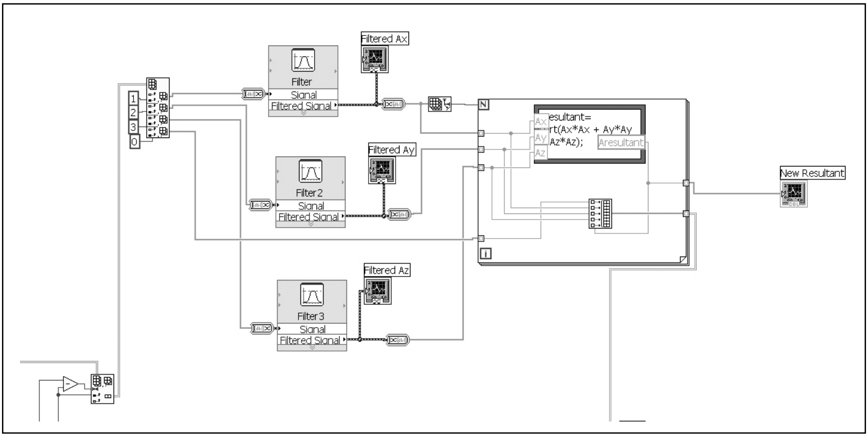

Figure 6
Digital data acquisition flow diagram

This is taken into consideration in quantifying brain injury tolerance. Brain injury levels are specified as the magnitude over time of a mechanical parameter considered to be a key indicator of cerebral trauma.

The Head Injury Criteria, or HIC, was utilized to quantify the concussion hazard. Developed by Versace in 1971, HIC takes into consideration a time-averaged weighted acceleration, based on a fit of the acceleration versus impact duration of the Wayne State Tolerance Curve. The HIC is calculated by acceleration, a(t), in g's measured at the center of mass of the head. If the duration of acceleration is less than 15 milliseconds, t_1 and t_2 are the initial and final times in seconds between which the HIC is evaluated as a maximum. The HIC calculation is stated below in Equation (3).

$$\left\{ \frac{1}{(t_2-t_1)} \int_{t_1}^{t_2} a\,\mathrm{d}t \right\}^{2.5} (t_2-t_1) < 1000 \tag{3}$$

Equation 3
Head Injury Criteria or HIC Calculation

The HIC was approved for use in federal motor vehicle safety standards in 1972. It is measured by linear acceleration, and uses primate, canine and cadaver test data as well as human subject data at lower accelerations levels to quantify human tolerance. Since the HIC has been adopted as the standard for automobile safety testing, it is used for this research.

Results of Lagrangian Modeling Input and Subsystem Experimentation

Figure 7 is a sample output from the head impact test: acceleration verses time datum collected. The first peak measures 230 g's with two definitive subsequent lesser impacts consistent with the head's bounce before it comes to rest. The bounces of the head also substantially increased the duration of the impact from less than 15 milliseconds to roughly 25 milliseconds.

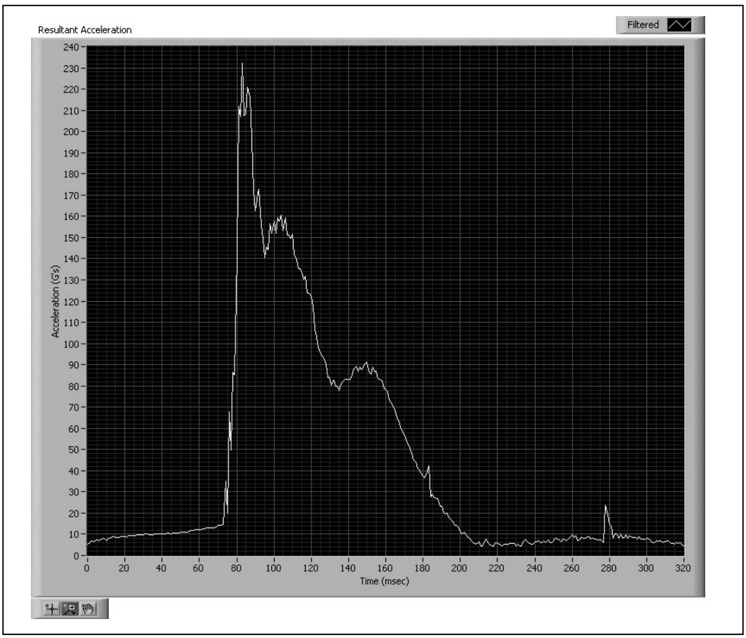

Figure 7
Example of head impact acceleration (g's) verses time (milliseconds) results

Utilizing Equation 3, the Head Injury Criterion (HIC) is calculated from the acceleration verses time datum for each of 10 trials. There are 5 trials for the 11 foot pole (10 mph ΔV) and 5 trials for the 21 foot pole (20 mph ΔV) as listed in Table 1. Note that the 11 foot pole impacts are less variable than the 21 foot pole impacts tracing in great part to the variability in impact location that occurs with a longer pole length. In some cases head impact occurred in the center of the spring portion of the seat, greatly reducing the HIC, compared to the frontal unyielding structural portion of the seat as shown in Figure 8.

Pendulum Height_Trial # (height in feet)	Peak Accelerations (g's)	Head Injury Criterion (HIC)
11_1	144.3	765.4
11_2	143.5	813.8
11_3	147.8	720
11_4	145.1	767.4
11_5	146.7	746.2
21_1	224.7	1238.4
21_2	192.5	1424.6
21_3	180	917.5
21_4	339.1	4550
21_5	315.8	702.9

Table 1
Final acceleration and HIC results for each impact test

Discussion

As anticipated, the HIC tolerance was not exceeded at the vehicular change in velocity of 10 mph level (11 foot pole length). However, the HIC tolerance was exceeded at the vehicular change in velocity of 20 mph level (21 foot pole length) in three of five trials. Due to the visually extensive nature of the padding, it was not anticipated that the human tolerance of the HIC could be met and exceeded by impacting the Hybrid III head and neck with the thickly padded seat bottom cushion. The HIC results at a vehicular change in velocity of 20 mph were highly variable tracing to the variability in the surface itself.

Experimental results revealed that the specific location of head impact with the seat substantially affected the results. This traced to the construction differences in seat construction by area: the anterior portion had a rigid base with a steel reinforcement across the superior anterior aspect (top front) under 5 inches of padding, while the center was comprised of a spring mattress-like support under 5 inches of padding. These constructions are pictured in Figure 8.

Figure 8
Construction of the seat bottom base substantially affected the magnitudes of accelerations by impact location

Variability in the 20 mph experiments traced to the dramatically different design in the seat bottom as a function of the specific location of head impact onto the seat. If the head impacted the seat on the anterior portion, with steel reinforcement under the 5 inches of padding, the peak accelerations were as high as 339 g's, and the HIC values were as high as 4550, respectively. If the head impacted the seat in the center, with a spring mattress-like support under the 5 inches of padding, the peak accelerations were as low as 180 g's and the HIC values were as low as 702. To conduct the 20 mph tests, the inverted pendulum test assembly was 21 feet in height. With the subsystem test at 21 feet, the impact angle could be maintained at 65 to 70 degrees but the impact site could vary due to the 21 foot length of the inverted pendulum. In the actual human system, the seated height is 35 inches; this introduces 0.15 foot variability in head to seat impact location which would limit contact to the anterior portion. Therefore, every effort was made, without slowing the inverted pendulum with guides or wires, to impact the front portion of the seat since that is what would occur if the occupant sustained a principal direction of force of 65-70 degrees.

Conclusion

The asymmetric design of the seatbelt does not prevent potentially injurious high acceleration contact with the vehicle interior in opposite side motor vehicle collisions. In opposite side impacts of 65 to 70 degrees the lap and shoulder restraint function less effectively. At these angles with a change in vehicular velocity of greater than or equal to 20 mph, the HIC may exceed tolerance when the occupant's head impacts the anterior portion of the adjacent seat bottom. Even with a 5 inch seat cushion over the structural seat, the padding "bottomed out," and the Head Injury Criteria (HIC) tolerance was exceeded in three of five impacts. Even when using a seatbelt in opposite side impacts, the occupant may sustain brain trauma that exceeds human tolerance per the HIC. BioMedical engineering testing and analysis should be conducted on case specific contact surfaces to determine the likelihood of sustaining brain trauma even when using a seatbelt in these scenarios.

References

1. Goldsmith, W., 2001, "The State of Head Injury Biomechanics: Past, Present and Future: Part I," *Critical Reviews in Biomedical Engineering,* 29, pp. 441-600.
2. Malliaris, AC, *et al.*, 1982, "Search for Priorities in Crash Protection," Society of Automotive Engineers, #820242.
3. Mackay, G. M., 1991, "Restrained Occupants on the Non-Struck Side in Lateral Collisions," 35th Annual proceedings Association for the Advancement of Automotive Medicine, Toronto, Canada, page 120.
4. Hull, M. L., 1980, "Principles of Digital Data Acquisition," Instrumentation, Systems, and Automation Transactions, 19(3), pp. 25-36.
5. Nahum, A.M., and Melvin, J., 1985, The Biomechanics of Trauma, Appleton-Century-Crofts, East Norwalk, CT.

Dedication and Appreciation

This is dedicated to the life and work of Werner Goldsmith, Ph.D., one of the grandfather's of head impact research in the United States.

The author wishes to acknowledge the generous co-operation of Dennis K. Lieu, Ph.D., University of California at Berkeley.

This is also to thank Mr. Richard C. Watters of Fresno, California for granting the funds for this research and the willingness to pursue fact.

Keywords

Head Injury, Head Trauma, Brain Injury, Injury While Restrained, Opposite Side Impact, Simulation, Head Impact Experimentation.