Head Injury Criterion and the ATB

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Introduction

Injuries to the head are responsible for 50,000 deaths and nearly one million hospitalizations per year in the United States. Motor vehicle crashes are responsible for nearly half of these head injuries[1]. Head injury continues to be a leading cause of death and disability.

Parameter studies using mathematical modeling of motor vehicle occupants have been used to complement mechanical testing using anthropomorphic test dummies (ATD), animal, cadaver and live human test subjects. This has considerably advanced the understanding of head injury mechanisms.

One of the mathematical models used to complement full scale mechanical testing is the Articulated Total Body (ATB) model[2]. The Articulated Total Body (ATB) Model is a public domain computer program that is used to simulate the dynamic motion of jointed systems of rigid bodies. The ATB includes the ability to calculate for any given simulation the Head Severity Index (HSI) and the Head Injury Criteria (HIC).

This paper presents background on head injuries, head injury criterion and the use of the ATB in research and litigation to simulate occupant motions and calculate the head injury criterion.

Background on Head Injuries

The human head is a complex system[3]. The human head consists of three components:

1. The bony skull
    - Cranial and facial bones
2. The skin and other soft tissue covering the skull
    - Which consists of layers known as the SCALP (**S**kin, **C**onnective Tissue, **A**poneurosis (Galea), **L**oose connective tissue and **P**eriosteum)
3. The contents of the skull
    - Most notably the brain, but also including the brain's protective membranes (meninges) and numerous blood vessels (see Figure 1).

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[1] Bandak, F. A., Eppinger, R. H. and Ommaya A. K.(eds.) (1996) **Traumatic brain injury: Bioscience and Mechanics**. Mary Ann Liebert Publishers.

[2] Cheng, Rizer, Obergefel, "Articulated Total Body Model Version V – User's Manual", Report AFRL-HE-TR-1998-0015, February 1998

[3] Portions from Pike, J. A., Automotive Safety: Anatomy, Injury, Testing and Regulation, SAE, 1990

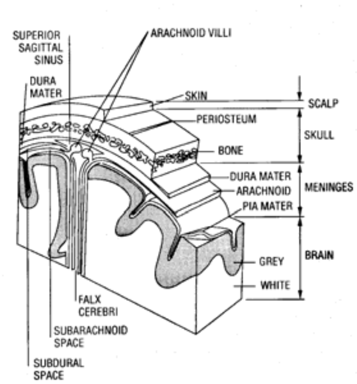

Figure 1 The Scalp, Skull, Meninges and Brain (Figure 2.1 from Reference 3)

Injuries to the skin may be categorized as superficial or deep, and include contusion (bruise), laceration (cut), and abrasion (scrape). Injuries to the skull may break one or more of the bones of the skull in which case the skull is said to have been fractured (broken). Two aspects of a skull fracture are whether it is open or depressed.

Injuries to the brain and associated soft tissue are the result of either head impact or abrupt head movement (e.g., deceleration injury) or some combination of the two. Injuries may be due to the skull fracturing and being pushed inward (a depressed fracture), or from the brain impacting the interior of the skull, or from internal stressing of the brain (i.e., shear, tension and/or compression). The complexities of the head and brain system are reflected in the rather bewildering array of head injury consequences.

Three various methods are used to categorize brain injuries:

1. The cause of injury, either contact vs non-contact
2. The type of injury, either primary vs secondary
    a. Primary in which the injury occurs at time of initial injury producing event
    b. Secondary in which the injury results from some injury producing event but does not develop until somewhat later (through an intermediate process such as a metabolic effect)
3. The type of injury, either focal vs diffuse
    a. Focal (i.e. fairly localized)
    b. Diffuse (rather distributed)

In injury producing events, there are generally 3 collisions which occur:

1. The "first collision" is that in which the injury producing event occurs, e.g. the vehicle strikes another car or object and as a result the vehicle is rapidly decelerated and/or rotated.
2. The "second collision" is the movement of the occupants in the vehicle and their subsequent contact with the vehicle interior.
3. The "third collision" is when the internal organs of the occupant collide and/or move within the occupant.

The human head is a complex system.

Background on Head Injury Criteria

Attempting to categorize the possible injuries to the human head is a complex process[4]. Brain injury assessment functions are based on the observed impact responses of cadavers, animals, volunteers, or accident victims. There are limitations of each of these sources of data.

A relationship between the acceleration level and impulse duration with respect to head injury was first presented by a series of six data points that indicated a decreasing tolerable level of acceleration as duration increased. The relationship became known as the Wayne State Tolerance Curve (WSTC), named for the affiliation of the researchers, and has become the foundation upon which most currently accepted indexes of head-injury tolerance are based. The original data only covered a time duration range of 1 to 6 milliseconds and only addressed the production of linear skull fractures in embalmed cadaver heads. The curve was later extended to durations above 6 msec with comparative animal and cadaver impact data and with human volunteer restraint system sled test data.

The WSTC has been criticized on various grounds since its inception: the limited number of data points, possible questionable instrumentation techniques, a lack of documentation regarding the scaling of animal data used in its extension to longer durations, and the uncertainty of definition of the acceleration levels.

From a biomechanical standpoint the main criticism of the WTSC is that there have been no direct demonstrations of functional brain damage in an experiment in which biomechanical parameters sufficient to determine a failure mechanism in the tissue were measured (this is also applicable for those advocating a rotational acceleration-induced mechanism of brain injury). The assumption behind the WTSC work is that the translational acceleration produces pressure gradients in the region of the brainstem that result in shear-strain-induced injury. The extension to the human brain remains to be verified.

The WSTC data was plotted by Gadd[5] on log paper and an approx straight line function was developed for the weighted impulse criterion that eventually became known as the Gadd Severity Index (GSI). In ATB, the program calculates the Head Severity Index (HSI) which is the same as the GSI and is defined as the time integration of the function a(t), the acceleration time history[6] :

$$HSI = \int \left[a(t)\right]^{2.5} dt \qquad (1)$$

In response to a study of the analysis of the relationship between the Wayne State Tolerance Curve and the Gadd Severity Index by Versace in 1971[7], a new parameter, the Head Injury Criterion (HIC), was defined by NHTSA in 1972. The HIC is currently used to assess head injury potential in automobile crash test dummies. It is based on the ***resultant*** translational acceleration rather than the frontal axis acceleration of the original WSTC.

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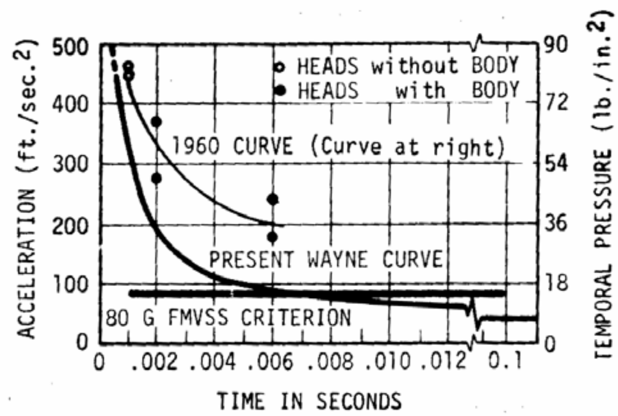
[4] Portions from Nahaum & Melvin, Review of Accidental Injury Biomechanics and Prevention, 2nd edition, 2002, Springer-Verlag New York
[5] Gadd, CW "Criteria for injury potential. Impact Acceleration Stress Symposium, National Research Council publication no 977, National Academy of Sciences, Washington DC pp141-144, 1961
[6] King, W.F., Mertz, H.J., Human Impact Reponse, Measurement and Simulation, Proceedings of the Symposium on Human Impact Response, GM Laboratories, Warren, Michigan, Oct 1972,
[7] Versace, J. "A Review of the Severity Index", Proceedings 15th Stapp Car Crash Conference, SAE paper 710881, pp 771-796



The original curve indicated the relationship between acceleration and time required to produce fracture in cadaver heads falling freely and heads falling with the body. Values 1-5 represent peak acceleration values, while 6 (180 G) is given as an average value (peak acceleration for this point was apparently 557 G), although not plotted. No date points have been published in the later version shown, but the new Wayne Curve is intended to show impact tolerance for the human brain in forehead impacts against plane, unyielding surfaces. Superimposed on this curve is the present FMVSS 80 G 3 ms performance criterion (224)

Fig. 10 - A comparison of the 1960 Wayne Curve (225, p. 8, Fig. 3) and the present Wayne Curve (223, p. 170, Fig. 1).

Figure 2 Comparison of Wayne State Tolerance Curve from 1960 and 1970 (From Reference 8)[8]

$$HIC = \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a dt \right]^{2.5} (t_2 - t_1) \qquad (2)$$

Where

a is a resultant head acceleration

t_2-t_2 $\leq$36 ms

t_2, t_1 selected so as to maximize HIC (2)

Some questions which arise with respect to HIC are: should HIC be applied to head accelerations which do not result from impact? For a glancing blow to the head for how substantial an impact should HIC be considered? The 36 millisecond interval is meant to encompass the maximum loading for impact waveforms which last longer than 36 milliseconds. Relatively short duration waveforms tend to be associated with head contact whereas longer duration HIC intervals tend to be associated with head deceleration without impact.

For example: for a 36 ms HIC interval, a HIC of 1000 corresponds to a constant head acceleration of approximately 60 g (that is, $A^{2.5}$(.036)=1000, or A = 60g (approximately). Are values above 1000 HIC meaningful? If so, what injury does it correspond to? Is a 2000 HIC twice as 'bad' as a 1000 HIC? Likewise, for lower values, is it proportionately 'better'?

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[8] Snyder, "State-of-the-Art Human Impact Tolerance, SAE 700398

The HIC is unique among FMVSS 208 injury criteria in that the HIC limit of 1000 was NOT based on tests where HIC was measured and injuries observed. Head acceleration was measured (as a function of time) and a single parameter (initially HSI, then HIC) was derived. The HIC has no specific meaning in terms of injury mechanism.

Recent research by NHTSA related to Improved Injury Criteria[9] have included reviewing the existing regulations which specify a HIC for the 50th percentile male ATD. As of 2000[10], the NHTSA final rule adopts limits which reduce the maximum time for calculating the HIC to 15 milliseconds (HIC_{15}) vs. the prior HIC_{36} and revising the limits for different sizes of dummies as shown below in Table S1-1. The head structure for the whole dummy family used in FMVSS 208 is essentially a padded rigid aluminum shell that does not deform as the human skull does deform under loading. The amount and type of deformation in the human skull, for a particular loading, varies significantly with age, and with marked difference between very young children and adults [**9**].

Table S1-1: Head Injury Criterion for Various Dummy Sizes

Dummy Type	Large Sized Male*	Mid-Sized Male	Small Sized Female	6-Year-Old Child	3-Year-Old Child	1-Year-Old Infant
HIC_{15} Limit	700	700	700	700	570	390

* The Large Male (95th percentile Hybrid III) is not included in the final rule, but the performance limits are listed here for informational purposes.

Figure 3 NHTSA FMVSS 208 revised Head Injury Criterion (HIC_{15}) 2000

Another recent report included the proposal of a new head injury criterion entitled the Head Impact Power (HIP). The HIP was proposed to consider not only kinematics of the head (rigid body motion of the skull) but also the change of kinetic energy of the skull which might relate to the deformation of and injury to the non-rigid brain matter. The Head Impact Power (HIP)[11] is based on the general rate of change of the translational and rotational kinetic energy. The HIP is an extension of previously suggested "Viscous Criterion" first proposed by Lau and Viano in 1986 [12], which states that a certain level or probability of injury will occur to a viscous organ if the product of its compression C and the rate of compression V exceeds some limiting value.

Categorizing the possible injuries to the human head is a complex process. There is currently not any consensus opinion on a standardized procedure or predictor.

[9] Eppinger, et al, "Development of Improved Injury Criteria for the Assessment of Advanced Automotive Restraint Systems – II", NHTSA, Nov 1999

[10] Eppinger, et. al, "Supplement: Development of Improved Injury Criteria for the Assessment of Advanced Automotive Restraint Systems – II", NHTSA, March 2000

[11] Newman, Shewchenko, Welbourne "A Proposed New Biomechanical Head Injury Assessment Function - The Maximum Power Index", SAE paper 2000-01-SC16, Stapp Car Crash Journal, Vol. 44

[12] Lau, I.V., and Viano, D.C., "The Viscous Criterion: Bases and Applications of an Injury Severity Index for Soft Tissues", Proceedings of 30th Stapp Car Crash Conference (P-189), pp. 123-142, SAE Technical Paper No. 861882, Society of Automotive Engineers,
1986.

Background on ATB and HIC

For research purposes mathematical models have been used to complement full-scale testing of cadavers, animals, volunteers, and used in the development and refinement of ATDs. The ATB has been used to complement full-scale mechanical testing for over 30 years. When used to interpolate and/or extrapolate full-scale testing, the ATB has been demonstrated to provide a useful tool to assist in the understanding of occupant responses in a variety of environments. In controlled laboratory correlation experiments, wherein the detailed setup, starting conditions and acceleration environment are know beforehand, sensitivity studies can be performed on unknown or immeasurable variables to improve understanding and correlation. The ATB includes the ability to calculate the HIS and HIC for a simulation run. The HIC and HIS calculations provide a crude benchmark for comparison in research and development with full-scale testing and parametric studies.

Every simulation model, particularly occupant simulation models like the ATB, is "an approximation of the physical system it represents and, of necessity, is based on simplifying assumptions concerning the various aspects of the nature and behavior of the real system. For this reason differences between predicted and observed responses of an actual system are to be expected, the disparities depending in part on the adequacy of the assumptions and approximations for the particular operating environment of the system"[13].

Parametric studies of ATB[14] for a limited number of model parameters have been performed in which some parameters were altered from their baseline values to investigate the effects on the occupant kinematics. They demonstrated the sensitivity of various responses of ATB to some parameters. Parameters investigated included: shoulder belt stiffness, lap belt stiffness, knee bolster stiffness, the guide loop location, the guide loop coefficient of friction. It should be noted that for an accident reconstruction, there are few if any known baseline values and therefore any conclusions or results may be subject to variations of 50% or greater (see **Figure 4**).

Recent studies comparing the responses of the latest Rear Impact Dummy (BioRID I) with volunteer human testing[15] demonstrate improvements in correlation with human responses to that of the Hybrid III. However, the responses are still lacking in some areas including producing a human-like retraction motion. Other recent testing comparing the responses of different sized Adult Hybrid III Dummies[16] demonstrated that rear impact kinematics are affected by parameters such as backset, head restraint height, seat-head restraint interaction, and anthropometry. The research concluded that rear impact responses are dictated not only by vehicular component characteristics (backset and head restraint) but also by factors such as gender, occupant size and active muscles.

[13] Validation of CVS, V2 Engineering Manual, Part 2, Validation effort, Aug 82, P178 Concluding remarks:
[14] Yih-Charng Deng, "An Improved Belt Model in CAL3D and its Application", SAE paper 900549

[15] Linder, et al, Design and Validation of the Neck for a Rear Impact Dummy (BioRID I), Traffic Injury Prevention, 3:167-174,2002, Taylor and Francis
[16] Derosia, et al "Rear Impact Responses of Different Sized Adult Hybrid III Dummies", Traffic Injury Prevention, 5:50-55, 2004, Taylor and Francis

Table 1. Derived system parameters.

Response Sensitivity

(Change in Percentage)

		HIC	Head Accel.	Head Displ.	Spine Accel.	Chest Def.	Pelvis Accel.	Abdomen Def.	L. Femr Load	R. Femr Load	S Belt Force	L Belt Force
Test		430	54	38.8	41.1	4.34	56.9	-	290	260	800	620
Model		404.6	47.1	38.3	43.1	4.19	52.9	3.71	245.2	206.7	814.1	626.8
S Blt	X 0.5	+44.8	+21.9	+29.2	- 0.2	- 0.2	+ 4.3	+16.2	+ 1.3	+ 4.5	-18.2	+ 6.8
Stiff	X 2.0	-26.7	-17.2	-23.0	- 7.2	- 4.5	- 4.0	-13.5	- 0.3	- 2.5	+16.7	- 6.4
Lp Blt	X 0.5	+ 6.4	+ 1.7	- 1.6	+ 2.3	+ 7.2	- 1.7	-21.3	+24.9	+39.9	+ 5.3	-31.1
Stiff	X 2.0	+ 1.4	- 1.5	+ 1.8	- 3.9	- 4.8	- 4.7	+14.8	-29.7	-34.2	- 5.0	+27.6
Kn B	X 0.5	+ 2.9	- 0.8	- 0.5	- 2.3	+ 1.7	- 6.8	+ 3.8	-32.6	-25.9	+ 0.4	+ 0.8
Stiff	X 2.0	- 0.4	- 2.3	+ 0.8	+ 5.3	- 0.2	+13.2	- 4.6	+68.9	+52.4	- 0.2	- 0.5
GL Dx	+ 20.	+35.8	+12.7	+11.2	+ 6.0	+ 5.5	+ 2.3	+ 4.8	- 9.0	+ 0.6	+11.7	+ 1.5
GL Frc	X 5.0	- 7.9	- 4.0	- 2.9	- 0.5	- 2.4	- 0.8	- 1.1	+ 1.1	- 0.5	+ 2.2	- 0.5

Table 2. Parametric study results.

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Figure 4 Sample response Sensitivity of ATB to model parameters

When the ATB is used in accident reconstruction and/or applied to individual accidents, there are inherent limitations on the predictability and veracity of the results of ATB. The following lists some of limitations of the ATB for individual accident reconstruction[17],[18],

1. Limitations of any mathematical model of occupant kinematics.
    - No two humans are alike in their physical properties or lifetime experiences.
        - The physical condition and inherent response of individuals (reflexes) affect what occurs during any potential injury producing event.
    - The GEBOD program is generally used to create occupant inputs for ATB
        - GEBOD creates a "percentile" representation of an occupant which only represent the actual occupant in but a few measurements or properties.
    - The ATB occupant is a "passive" occupant.
        - Active muscles and reflexes can play a substantial role in the responses of the occupant, particularly at lower impact speeds
    - The initial position and posture of occupants are not known and minor changes can have a dramatic affect on the ATB results.

2. Limitations of the ATB model for HIC
    - The ATB model of the head is a simplified elliptical representation of the skull.
    - Validations of predications of HIC for ATB are based on ATDs which have non-deformable heads. Human heads deform.
    - The human head is connected to the neck.
        - The movement and response of the head to acceleration are influenced and limited by the modeling of the neck and torso.

[17] Also see: James, Nordhagen ,et al, "Limitations of ATB/CVS as an Accident reconstruction Tool", SAE Paper 97-0945

[18] Also see: McHenry, B "Occupant Kinematics in Forensics: Evaluating the Appropriateness and Applicability of an ATB Application", 2002 ATB Users' Group Conference

    - The human neck and spine is a structure composed of 7 cervical, 12 thoracis, 5 lumbar, 5 sacral and 4 coccygeal vertebrae. Each vertebrae is composed of a cylindrical vertebral body connected to a series of bony elements collectively referred to as the posterior elements.
    - The modeling of the head/neck/vertebrae system in the ATB is a simplification of the human vertebrae to three joints which connect the upper, middle and lower torso.

3. Limitations of the HIC
   - The HIC calculation is not an injury predictor; it is a pass/fail baseline measure used for ATDs in full-scale testing.
   - There is no direct correlation of and/or values for HIC other than the pass/fail value prescribed by NHTSA (1000 for HIC_{36}, 700 for HIC_{15}, etc.)

4. Limitations of the detailed properties of the vehicle compartment
   - There is no standardized procedure for measurement of interior force-deflection properties for use with ATB.
   - The ATB requires consideration of the part of the occupant that interacts with the interior component
     - e.q., the head striking a roof pillar has a different effective force-deflection function than an arm, etc.
   - Many of the components of the vehicle may move or deform during the impact and/or during the interaction of the occupant with the component.
     - The rate and amount of movement of vehicle components are not known.
     - Arbitrarily moving vehicle components can have a profound effect on the resulting forces acting on the occupant.

5. Limitations of the technique used to reconstruct the injury producing event
   - All accident reconstruction techniques require assumptions and have limitations.
   - The ATB can accept various forms of input for the acceleration pulse (positions, velocities, accelerations) and each one has inherent limitations and sensitivities.

Conclusions

The human head is a complex system. Attempting to categorize the possible injuries to the human head is a complex process. The ATB has been used to complement full-scale mechanical testing for over 30 years. When used to interpolate and/or extrapolate full-scale testing, the ATB has been demonstrated to provide a useful tool to assist in the understanding of occupant responses in a variety of environments.

The ATB should be used in forensics and accident reconstruction only as a tool to assist in understanding gross occupant kinematics. Any results or conclusions drawn from an ATB application related to HIC calculations or detailed occupant kinematics involve so many approximations, estimates, and assumptions that they must be recognized as not being compatible with sound engineering practices and principles and, therefore, not scientifically supportable.