A NEW BIOMECHANICAL ASSESSMENT OF MILD TRAUMATIC BRAIN INJURY PART 2 – RESULTS AND CONCLUSIONS

James Newman, Cameron Barr, Marc Beusenberg, Ed Fournier, Nicholas Shewchenko, Eric Welbourne, Christopher Withnall
Biokinetics and Associates Ltd., Ottawa, Ontario, Canada

ABSTRACT

This paper follows Part I that was presented at the 1999 IRCOBI Conference. A methodology, described in Part 1, has been developed that permits the reconstruction of certain incidents that occur in American football. Twenty-four cases of helmeted head impact, for which concussion was diagnosed in 9 cases, have been replicated with Hybrid III ATDs. Rigid body translational and rotational head accelerations have been measured in each case. Correlations between head injury and head kinematics have been sought. Peak translational and peak rotational acceleration and velocity, HIC and the Gadd SI, as well as the GAMBIT have all been considered. A new approach employing the maximum value of the global rate of energy dissipation has proven to provide the best correlation between concussion probability and head kinematics. This new relationship provides a basis for a new head injury criterion function, the HEAD IMPACT POWER.

KEY WORDS
Biomechanics, brains, helmets, injury criteria, injury probability

HEAD INJURY OCCURS IN MANY WALKS OF LIFE including athletic events such as football. The biomechanics of minor traumatic brain injury MTBI, or concussion, has been the subject of extensive research for the past several years. In North America, certain professional football players have been engaged in a program whereby athletes who undergo significant head impact, have that event reconstructed using instrumented anthropomorphic mannequins.

Concussion during the sport of American football is actually a rather rare occurrence. This is in no doubt partly due to the use of quite highly effective helmets. These helmets are required to meet the performance specifications of the standard of the National Operating Committee on Sports and Athletic Equipment NOCSAE first published nearly thirty years ago, recently re-issued in 1997 (National Operating Committee on Standards for Athletic Equipment (NOCSAE), 1997).

One of the principal objectives of the present study is to provide new insight into the nature of head impacts in football and to thereby provide guidelines for improved helmet standards. An equally important goal is to use the data gained from the study to augment our basic knowledge about the biomechanics of concussion and to perhaps generate more appropriate head injury indices and biomechanical assessment functions.

CURRENT KINEMATIC HEAD INJURY ASSESSMENT FUNCTIONS

Several kinematic head injury assessment functions have evolved over the past forty years. They include:

- Maximum linear acceleration, used for many years and continues to be used in many helmet standards. (Snell 1995, CSA, 1985).

$$a_m < N$$

- Maximum linear acceleration with dwell times, employed by the NHTSA for the US motorcycle helmet standard. (NHTSA Standard 218, 1997).

$$a_m < 400G$$
$$\text{time at } 200G < 2\text{msec}$$
$$\text{time at } 150G < 4 \text{ msec}$$

- Average acceleration with time duration. (Gurdjian et al, 1964)

$$\bar{a}^{2.5}T < 1{,}000$$

It has never actually been used in any performance test but is the basis for

- The Severity Index SI. (Gadd, 1966)

$$\int_T a^{2.5}dt < N$$

where a is the resultant linear acceleration of the center of gravity of the head. The SI is set to a limiting value of 1200 in the current NOCSAE standard.

- The Head Injury Criterion HIC (NHTSA, 1974)

$$[1/(t_2 - t_1)\int_{t_1}^{t_2} a(t)dt]^{2.5}(t_2 - t_1) < 1{,}000$$

It was first employed as part of FMVSS 218, and is now the most widely referenced head injury assessment function.

- Angular acceleration combined with angular velocity change (Ommaya, et al, 1971)

AIS	Acceleration, rad/s^2	Velocity Change, rad/s
0	<4500	<30
1	<1700	>30

- Angular and linear acceleration GAMBIT (Newman, 1986). This requires establishing the maximum value of the following function.

$$G(t) = \left[\left(\frac{a(t)}{250}\right)^2 + \left(\frac{\boldsymbol{\alpha(t)}}{25000}\right)^2\right]^{\frac{1}{2}}$$

G = 1 was set to correspond to a 50% probability of AIS>3.

A NEW HEAD INJURY ASSESSMENT FUNCTION

A recent review of the above head injury indices suggests that the rate of change of linear and rotational kinetic energy, i.e. power, could be a viable biomechanical MTBI assessment function. An empirical expression relating a measure of power to concussion would be of the form

$$HIP = Aa_x\int a_x dt + Ba_y\int a_y dt + Ca_z\int a_z \mathbf{dt} + \eta\alpha_x\int\alpha_x dt + \beta\alpha_y\int\alpha_y dt + \chi\alpha_z\int\alpha_z dt$$

where the coefficients denote the relative sensitivity for each of the six degrees of freedom of the head. It is hypothesized that should the numerical value of this expression exceed some limiting value, during the impact event, an MTBI is probable. The development of this function is fully described in Newman et al (2000). In the current absence of information regarding directional sensitivity, the coefficients in the above equation are simply set to reflect the mass and mass moments of inertia of the Hybrid III headform.

THE DATABASE

During the 1995 to 1999 playing season, several hundred concussions were documented. Of these, videotape and medical records of one hundred incidents, were examined. From these, 12 incidents involving 24 players who collided with one another were subject to full-scale laboratory reconstruction. These cases all involved helmeted head to head strikes generally between players of opposing teams.

The distribution of MTBI is shown in Table 1.

Table 1: Distribution of MTBI

Case Number	Striking Player Concussed	Struck Player Concussed
07	No	No
38	No	Yes
39	No	Yes
48	No	No
57	No	Yes
59	No	No
69	No	Yes
71	Yes	No
77	No	Yes
84	No	Yes
92	No	Yes
98	No	Yes

It will be noted that all reported concussions were affirmatively diagnosed. However, it is possible that the non-MTBI cases could include players who sustained a concussion but, because they did not appear injured, were not clinically examined.

RE-ENACTMENT METHODOLOGY

The details by which any incident is reconstructed is discussed in Part 1 of this study (Newman, et al 1999) and the interested reader is invited to refer thereto.

Laboratory-based reconstruction of football head impacts is achieved using Hybrid III ATD's. On the football field, MTBI occurs from head contact with many surfaces, including the ground, knees, elbows and other heads. The primary focus of this study is head-to-head collisions, for two reasons. First, the impacting surfaces, being certified helmets, are well defined and characterized. Secondly, in head-to-head collisions, two data sets can be collected for each reconstruction: the injured and non-injured players.

From the kinematic analysis of game video, the relative velocity of one player's head to the other is determined. A common thread that was noted among most of such cases was that the injured (usually the struck) head was impacted laterally, and the non-injured (usually the striking) head was impacted vertically. In this fashion, the body mass of the struck player did not factor in the collision, but the body mass of the striking player was indeed a factor.

The injured player headform is mounted on a standard Hybrid III neck, which is connected to a carriage on a vertical track. The carriage has provision to adjust the orientation of the head and neck to match that of the player at collision. The non-injured headform is mounted on a full Hybrid III test dummy neck and torso, which is held in static suspension by four spring-loaded tethers at the hips and shoulders. The arms and legs are removed. The carriage is raised to a height that upon release in free-fall yields the intended impact speed. In some cases where the impact speed was higher than possible by gravity alone, elastic shock cords were used to boost the carriage. Upon impact, the stationary dummy is free to rebound, thus preserving the momentum of the collision. An illustration of the set-up is shown in Figure 1.

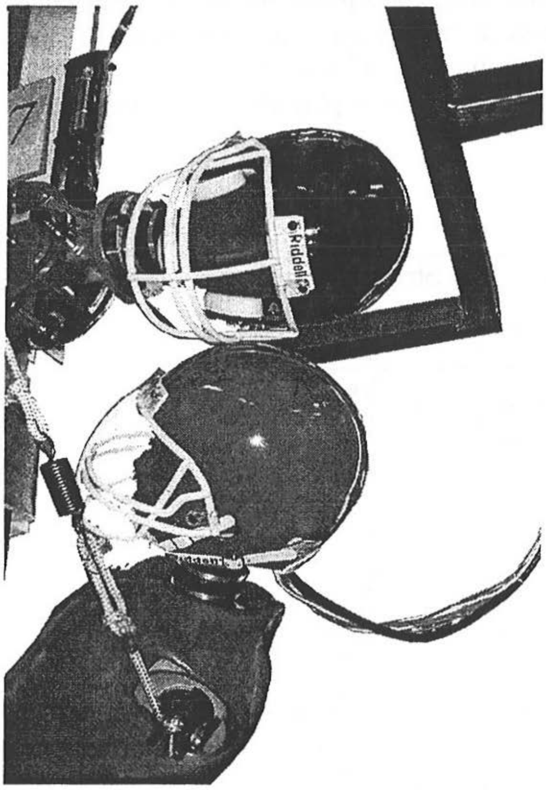


Figure 1: Laboratory re-enactment set-up

Alignment of the headforms is achieved through careful analysis of the game video, and the calculated respective camera positions. The direction of the carriage travel represents that of the calculated relative velocity vector of collision. High-speed video cameras, capturing 500 frames per second, are positioned at the same relative angles to the point of collision as the game cameras. Then the headforms are rotated and aligned to achieve the same orientations in the lab cameras as that seen in the game video. When the set-up appears correct from multiple views, the case is documented and the test is run. It should be noted that game video is captured at only 30 frames per second, and that there is rarely a video frame of the actual point of impact, only one before and one after. In these cases, some subjective interpolation is required for the set-up.

An example game video snapshot from a selected case is shown in Figure 2, and the corresponding lab video of the reconstruction set-up shown in Figure 3. This view was taken from high up along the carriage rail, looking almost in line with the velocity vector. The correctness of the test set-up is verified by proper rebound kinematics. If the headforms do not move in the same way the players' heads did, the set-up is adjusted and the test repeated.

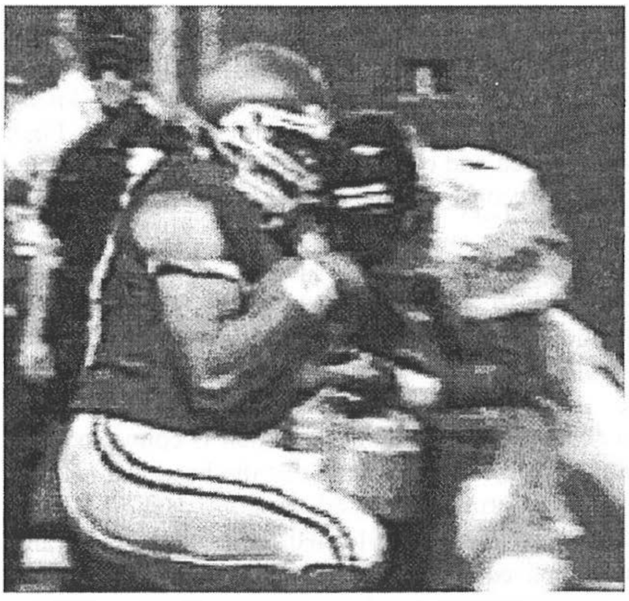


Figure 2: Sample snapshot from game video (case 48)

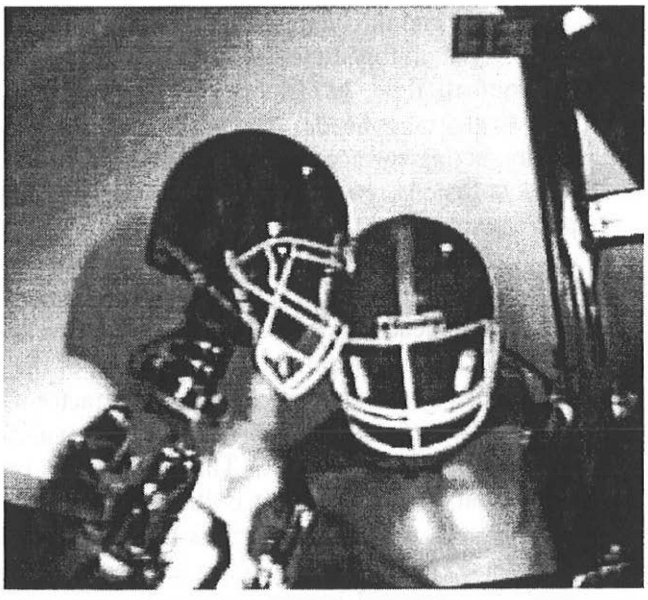


Figure 3: Sample lab high-speed video, same camera angle (case 48)

Each headform is instrumented with nine linear accelerometers arranged to allow the calculation of triaxial linear and rotational accelerations following the NHTSA protocol (DiMasi, 1995). The stationary headform was additionally instrumented with a six-axis upper neck load cell, in case of possible neck loading investigation in the future. All data were collected at 10kHz following SAE J211 protocol. Acceleration data were pre-processed according to CFC 1000 requirements, and then later re-filtered digitally at CFC 180. This secondary filtering was found to remove spurious noise from the rotational acceleration data without unduly affecting the overall signal.

RESULTS

IMPACT VELOCITY: The computed relative velocity of the two colliding players for each case is shown in Table 2.

Table 2: Relative Velocity

Case	Concussion Yes/No	Relative Velocity (m/s)
7	No	6.8
38	Yes	9.7
39	Yes	11.1
48	No	9.2
57	Yes	8.4
59	No	5.2
69	Yes	10.1
71	Yes	10.5
77	Yes	10.1
84	Yes	9.6
92	Yes	10.6
98	Yes	9.4

It will be observed that there is no direct correlation between MTBI and relative velocity. Concussion will occur at speeds as low as 8.4m/s. It is not how fast the players run with respect to each other when they collide but how their heads interact upon impact. This is measured by examining the kinematic response of each head.

HEADFORM RESPONSE: The acceleration response data for each of the impacting heads has been processed for several published engineering injury indices. The results are shown in Table 3.

Table 3: Headform Response

Case No. 1 = tackler 2 = tackled	Reported MTBI 0=no 1=yes	a_m (m/s^2)	α_m (rad/s^2)	SI	HIC	GAMBIT	HIP (kW)
07-2	0	596	6265	121	93	0.35	6.7
38-2	1	1162	9678	743	554	0.60	23.3
39-2	1	1263	5729	663	521	0.55	19.8
48-2	0	562	5855	157	130	0.32	9.7
57-2	1	758	5786	255	207	0.38	12.1
59-2	0	807	5035	207	138	0.38	8.0
69-2	1	595	4168	181	130	0.25	9.0
71-2	1	1211	5434	655	510	0.52	24.0
77-2	1	788	5128	272	185	0.37	13.2
84-2	1	804	9244	317	225	0.49	17.6
92-2	1	1054	8877	706	508	0.48	21.6
98-2	1	893	7548	366	301	0.46	18.3
07-1	0	489	2832	65	51	0.23	3.4
38-1	0	588	5205	158	127	0.32	6.6
39-1	0	431	4184	61	43	0.18	3.3
48-1	0	310	2817	45	37	0.17	2.6
57-1	0	317	3937	51	37	0.20	4.0
59-1	0	314	1950	32	28	0.14	1.8
69-1	0	371	2593	83	50	0.17	3.6
71-1	0	1005	5555	519	433	0.45	19.3
77-1	0	342	2563	68	53	0.17	4.4
84-1	0	442	3036	98	77	0.22	4.6
92-1	0	586	6070	218	164	0.33	8.3
98-1	0	827	4487	245	187	0.38	10.4

LOGISTIC REGRESSION: Univariate logistic regressions for the 24 cases were performed for each of the above six independent variables identified as possible predictors of concussion. Concussion probability curves based upon these regression analyses are provided in the Appendix. Each of these curves provides a measure of the probability of MTBI based on this data set of 24 cases. From these functions one can extract discreet probabilities and examine the specific corresponding value of each assessment function. This has been done for 3 such probabilities as shown in Table 4.

Table 4

Probablity (%)	a_m (m/s^2)	α_m rad/s^2)	SI	HIC_{15}	GAMBIT	HIP (kW)
5	392.2	3377	23.50	[<0]	0.2231	4.700
50	761.5	6322	291.2	239.8	0.3935	12.79
95	1131.0	9267	558.9	485.2	0.5638	20.88

However, not each of these head injury assessment functions is as reliable as the other.

Two measures of the significance of the various predictions is provided by the regression analysis and the results are provided in Table 5.

Table 5: Results of Logistic Regression Analyses

	a_m	α_m	SI	HIC_{15}	GAMBIT	HIP
Significance P-value	0.011	0.029	0.024	0.020	0.013	0.008
-2LLR	18.059	20.676	18.195	19.347	18.031	14.826

The Significance (p-value) is often used as a screening tool in regression analyses, with a suggested threshold of $p \leq 0.25$ for the inclusion of an independent variable in the model. In common with –2LLR, lower numerical values are associated with higher significance. The –2 Log Likelihood Ratio (-2LLR) provides a means of assessing whether adding an independent variable to the constant has improved the significance of the model. A smaller numerical value of –2LLR denotes a result of higher significance. An exact fit of the regression model to the data is associated with a zero value of –2LLR.

Examination of Table 5 shows that the different injury indices are of different levels of significance. The least significant, not unexpectedly, is the peak resultant angular acceleration. It alone is not capable of providing a good prediction of the likelihood of concussion. HIC, SI and a_m are all of modest significance when considering both the p value and -2LLR. The two best predictors are the two that incorporate both linear and rotational motion, GAMBIT and HIP. GAMBIT itself however is only marginally better than a_m alone. HIP is considerably more significant than all the other parameters. It can thus be concluded, at least from this set of data, that the head impact power HIP is a better kinematic head injury assessment function for the prediction of mild traumatic brain injury MTBI, than other published head injury assessment functions.

DISCUSSION

The implications of these results may be of value not just in American style football of course but perhaps in many situations where a head impact injury may occur. In American football nevertheless, the implications are of some special interest as they relate to helmet performance specifications.

As pointed out above, the NOCSAE standard for football helmets employs a failure criterion of SI = 1200. The SI is derived largely from frontal drop tests of cadavers and other A-P inertial loading environments. It is based solely on the measured linear acceleration of the head. In this particular data set, concussions occur primarily to players whose helmeted head undergoes a lateral impact. In the current data set, a SI over 600 almost assures that a concussion will occur. Thus, for these kinds of impacts, the current standard failure level may be too high. As has been suggested in the literature from time to time (McIntosh, et al, 1996), the tolerance of the head to injury may well be lower for lateral impacts than for other directions. In addition, the injuries of the current data set are associated with high-speed contact between the heads of opposing players, not contact with some relatively stiff ground surface as is presumed in the current standard test method. It is also likely that the contribution of rotational acceleration, which is not taken into account by the current SI, plays a significant role in the production of MTBI.

Of some additional interest are the results for HIC and a_m. The predicted HIC value for a 50% probability of a MTBI is nearly 240 while the comparable value for a_m is 78G (762m/sec/sec). The latter numbers are not far from what auto designers try to achieve when designing systems that do not have a high probability of serious head injury. However the current NHTSA HIC=1000, which allegedly corresponds to <20% risk of AIS4 (Mertz et al, 1997), would comport to a very high probability of MTBI. On the other hand, Transport Canada has previously proposed a failure limit of 80G head acceleration for advanced automotive restraint systems (Welbourne, 1994). This would be a more demanding requirement and comports to only a 50-50 chance of a concussion.

The numbers for the second most significant predictor GAMBIT, do not appear to be out of line with expectations. It will be recalled that G=l corresponds to a 50-50 chance of AIS=3. Thus a 50-50 chance of an AIS=l at G=0.4, appears reasonable.
The injury assessment function HIP is the most significant variable, however the 5, 50 and 95% probability numbers cannot be commented upon yet. The utility of this new function awaits further validation and it must be tested against new and existing data sets. Clearly however this new "head injury criterion" may find application in many situations where head impact injury may occur. In particular, it is hoped that at the very least, it will be considered as a more accurate, more appropriate function for assessing the capacity of a helmet to protect against concussion. In time it may be shown to be a suitable measure of the probability of concussion and of more serious forms of inertially-induced brain injury in a variety of situations.

ACKNOWLEDGEMENTS

The authors would like to express their sincere appreciation to Dr. Elliot Pellman, Chairman of the NFL Subcommittee on Mild Traumatic Brain Injury and to Ronnie Barnes, Dean Blandino, Jay Brunetti, Dr. Ira Casson, Dr. Henry Feuer, Dr. Mark Lovell, Dorothy Mitchell, Dr. John Powell, Dr. Doug Robertson, Dr. Andrew Tucker, Dr. David Viano and to Dr. Joe Waeckerle for their valuable contributions to this effort.

REFERENCES

Canadian Standards Association Standard CAN3-D230-M85, Protective Headgear in Motor Vehicle Applications, 1985.

DiMasi, F.P., "Transformation of Nine-Accelerometer-Package (NAP) Data for Replicating Head Part Kinematics and Dynamic Loading", U.S. Department of Transportation, National Highway Traffic Safety Administration, Technical report DOT HS 808 282, National Technical Information Service, Virginia, 1995.

Eppinger, R., Sun, E., Bandak, F., Haffner, M., Khaewpong, N., Maltese, M., "Development of Improved Injury Criteria for the Assessment of Advanced Automotive Restraint Systems – II", National Highway Traffic Safety Administration, Technical Report, U.S. Department of Transportation, Washington, DC, 1999.

Gadd, C.W., "Use of a Weighted-Impulse Criterion for Estimating Injury Hazard", Proceedings of the Tenth Stapp Car Crash Conference, pg. 164, 1966.

Gurdjian, E.E., Roberts, V.L., Thomas, L.M., "Tolerance Curves of Acceleration and Intracranial Pressure and Protective Index in Experimental Head Injury", Journal of Trauma, pg 600, 1964.

McIntosh, A.S., Svensson, N.L., Mattern, R., Krabbel, G., Ikels, K., "Head Impact Tolerance In Side Impacts", Fifteenth International Technical Conference on the Enhanced Safety of Vehicles, National Highway Traffic Safety Administration, Washington, D.C., USA, 1996.

Mertz, H.J., Prasad, P., Irwin, L.A., "Injury Risk Curves for Children and Adults in Frontal and Rear Collisions", Proceedings of the 41st Stapp Car Crash Conference, Society of Automotive Engineers, pp. 13-30, 1997.

National Highway Traffic Safety Administration , Department of Transportation Standard No. 218, Motorcycle Helmets, 1988.

National Operating Committee on Standards for Athletic Equipment (NOCSAE), doc .002-96, Standard Performance Specifiication for Newly Manufactured Football Helmets, January 1997.

Newman, J.A., "A Generalized Model for Brain Injury Threshold (GAMBIT)", International Conference on the Biomechanics of Impact (IRCOBI), 1986.

Newman, J.A., Beusenberg, M., Fournier, E., Shewchenko, N., Withnall, C., King, A., Yang, K., Zhang, L., McElhaney, J., Thibault, L., McGinnis, G., "A New Biomechanical Assessment of Mild Traumatic Brain Injury, Part 1 – Methodology", International Conference on Biomechanics of Impacts (IRCOBI), 1999.

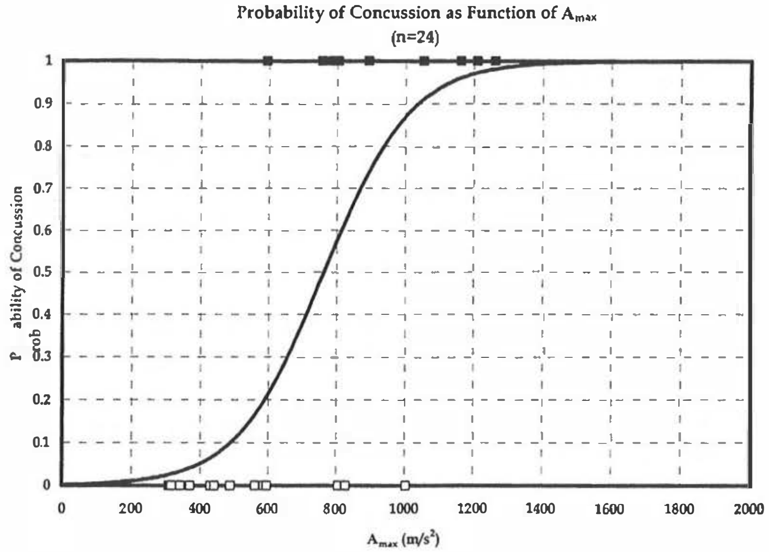
Newman, J.A., Shewchenko, N., Welbourne, E., "A New Biomechanical Head Injury Assessment Function – The Maximum Power Index", to be presented at the 44th Stapp Car Crash Conference, 2000.

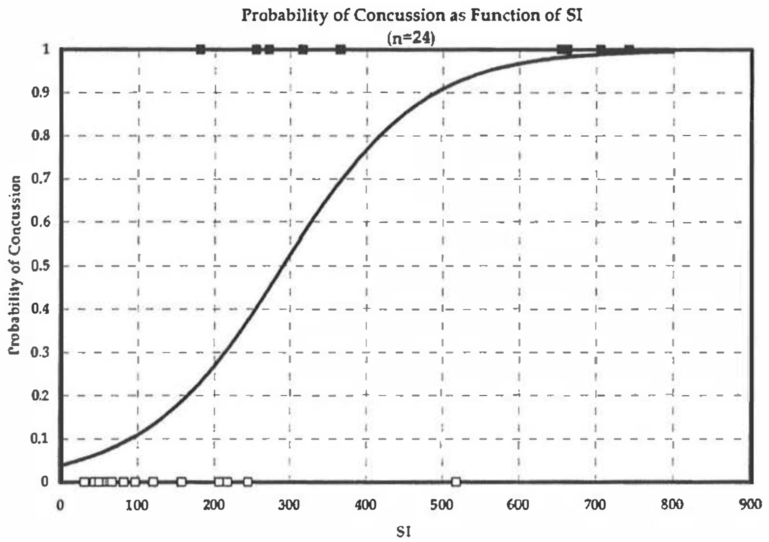
Ommaya, A.K., Hirsch, A.E., "Tolerances for Cerebral Concussion from Head Impact and Whiplash in Primates", Journal of Biomechanics, Volume 4, No. 13, 1971.

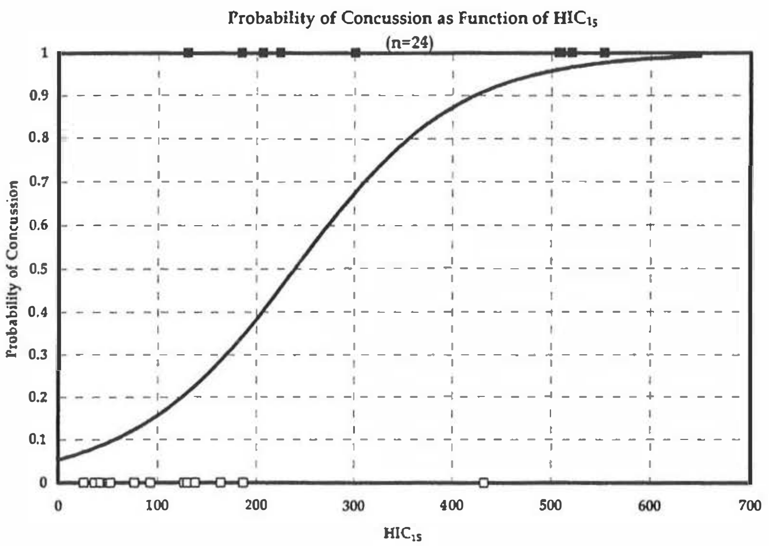
Snell Standard for Protective Headgear for Use with Bicycles, 1995.

Welbourne, E., "Use of the Head Injury Criterion as a Measure of Vehicle Occupant Protection Performance", International Conference on the Biomechanics of Impacts (IRCOBI), pp 151-162, 1994.

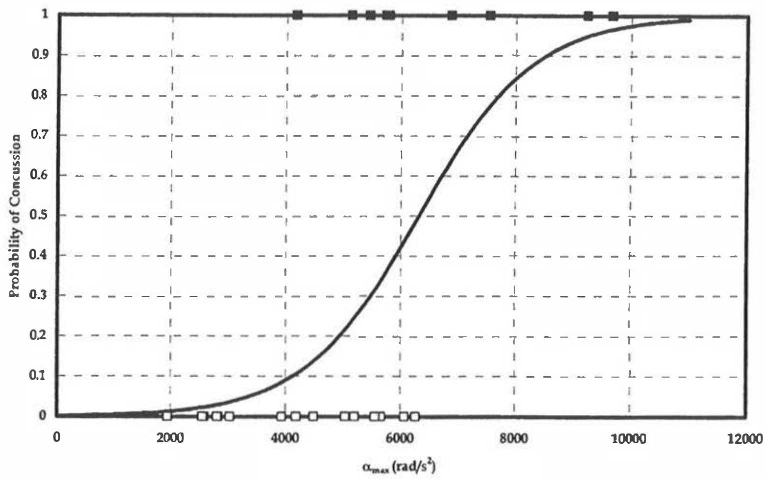
APPENDIX – MILD TRAUMATIC BRAIN INJURY RISK CURVES FOR VARIOUS KINEMATIC HEAD INJURY ASSESSMENT FUNCTIONS



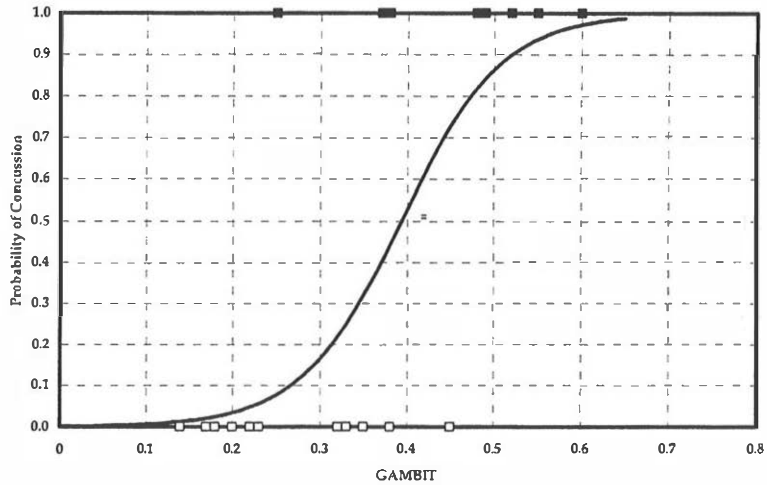




Probability of Concussion as Function of α_{max}
(n=24)



Probability of Concussion as Function of GAMBIT
(n=24)



Probability of Concussion as Function of HIP
(n=24)