HELMETS: A NEW LOOK AT DESIGN AND POSSIBLE PROTECTION

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ABSTRACT

The paper sets out to show that by considering the requirements of a motorcycle crash helmet from first principles short comings in existing designs become apparent. Results were obtained from 150 drop tests of instrumented head forms in helmets under controlled conditions. The performance of practical helmets is compared with the protection that would be provided if the available space could be fully utilised for energy absorption as predicted by applying equations of motion to an accepted injury criterion. H.I.C. is used as the initial criterion but alternative methods of predicting brain injury are discussed and their underlying principles examined. Choice of present helmet materials and the current British Standard test procedure are examined.

INTRODUCTION

This paper reviews acceleration-based methods of assessing head injury and their history, and reanalyses from first principles the amount of protection which might be offered by an ideal motorcycle crash helmet. A number of helmets constructed from shells and liners of different stiffnesses were drop tested and their performance assessed on the basis of H.I.C. and other acceleration-based criteria. Results from these tests are compared with the theoretical ideal performance, which shows that a 60% improvement could be achieved if more appropriate materials were used. Recommendations to improve the British Standard are suggested and discussed, as are the merits of different injury criteria which could be used and incorporated into the Standard.

BACKGROUND AND REVIEW

The use of crash helmets has reduced the number of fatal head injuries, but 700 motorcyclist fatalities still occur in the U.K. each year. The mode of injury to a helmeted head tends to be different from that to the unprotected head. The injury to the unprotected head is caused mainly by skull fractures: These are often fatal but not necessarily so and if the fracture is not depressed no remedial action is required. By contrast, even the most crude design of helmet gives nearly complete protection from fracture. The efficiency of helmets in preventing fracture is demonstrated by Chamouard and Tarriere (1) who performed a series of drop tests on cadavers with and without helmets. In the 14 tests from a drop height of 1.8 m with the head helmeted there were no instances of skull fracture, whereas in the 8 tests from 1.2 m on unprotected heads, 4 instances of fractures were reported. However, with helmeted heads a new mode of injury becomes apparent. Accident surveys reveal (2) that there are many fatalities without fracture, generally termed "acceleration induced".

Acceleration need not be injurious: any rate of acceleration can be tolerated provided it is evenly applied (ie. a uniform field eg. gravity). This is because injury is caused only when body parts are strained beyond their elastic limit, and in a uniform field all parts move together so that no strain is induced. However impact acceleration is the result of large externally applied forces which are not uniformly applied and which set up internal strains in the body. These can be very difficult to measure. Average acceleration of the whole head is a convenient parameter to measure, but it can only be a crude, aggregate indicator of likely injury. The surrogate head forms used in crash testing are purposely fairly rigid in order that an accelerometer placed near the centre of gravity will accurately indicate the applied force. This situation is analogous to the use of strain gauges where stress, which is internal, is the required parameter but the physical or external change recorded by the gauge is strain. The two are linked of course by Young's Modulus (E). When a surrogate head is subjected to a force its acceleration is recorded by the instrumentation and the dynamics can be calculated. This has led to the many "acceleration" based criteria for injury, of which the best known is the Head Injury Criterion (H.I.C.).

H.I.C.

H.I.C. was evolved from the experimentally-derived Wayne State concussion tolerance curve based on observations on volunteers and animals (3). Its rather complex form is a result of attempts (using log.log graph paper) to produce a simple algorithm:

$$\text{H.I.C.} = \left[\frac{1}{(t2 - t1)} \int_{t1}^{t2} a(t)dt\right]^{2.5} (t2-t1)$$

provides a mathematical "best fit" to a set of experimental data. Here, a(t) is the acceleration at time t measured in g; t1 and t2 are times (in seconds) of the beginning and end of the contact.

Gilchrist (4) and Newman (5) have recently strongly criticised H.I.C.; Newman doubts that the dynamic process which gives rise to brain injury can be consistently correlated by an average kinematic parameter such as H.I.C. Nevertheless for lack of any generally accepted alternative the predictions and conclusions of this paper depend largely on an H.I.C. of 1000 being the biomechanical limit for fatal injury. In contrast to Newman and in support of the predictions a defence is made as follows. Newman objects to the use of a single number to give a threshold of injury. It is true that biological systems have a well known probabilistic dose effect curve, but for practical purposes it is still valid to set a single level at which response is likely. The head is delicate and there is really no "safe" limit, so a H.I.C. of 1000 should be regarded merely as being near the threshold of injury for most people. Newman (5) gives as an example two cadaver tests where H.I.C. values of 1063 and 1073 were recorded yet injury was noted for only the lower value. This supports the view that H.I.C. 1000 is probably as sensible threshold and lies at the lower end of a likely human tolerance range, though clearly the distributions of H.I.C value corresponding to injury or non injury will overlap over a wide range of values. Kessler (6) has produced evidence based on pedestrian accidents to show that H.I.C. 1000 is a 10% and H.I.C. 2000 is a 50% probability of death. Federal Regulation MVSS 218 (Motorcycle helmets) state that accelerations shall not exceed 200g for 2msec or 150g for 4msec: these approximate to a H.I.C. of 1000.

Chamouard and Tarriere (1), who performed a series of drop tests on cadavers with and without helmets, concluded that there was no correlation between H.I.C. and injury. It was found that when skull fractures occurred the H.I.C.

was lower and the injuries sustained were higher than when there was no fracture. However, when the skull collapses the resulting injury is likely to be far more serious than when it does not, so that any correlation between H.I.C. and injury severity will be invalid once skull collapse occurs. Chamouard and Tarriere's apparently anomalous result can be explained because their accelerometer was fitted to the side of the skull opposite to the impact, and therefore recorded a low acceleration as the skull collapsed. However it is stated in the original specification (3) that an H.I.C. is valid only if calculated from accelerations recorded at the centre of gravity of an anthropomorphic test device ie. a non-collapsible headform. This suggests that H.I.C. is useful for comparing energy absorbing safety devices in impacts where death but not extensive skull fracture is likely to have occurred. Accident studies show that death frequently occurs without skull collapse.

OTHER CRITERIA

Another criterion for predicting likely injury is that 80g should not be exceeded for longer than 3ms. This was originally stated as a requirement that acceleration shall not exceed 80g, but pulses of less than 3ms shall be ignored, when it was introduced by the United States G.S.A. (General Services Administration) in 1965, and revised in 1966 to form draft J885a. This was adopted by industry and equates to a H.I.C. of 177. However, since the much higher H.I.C. value of 1000 is regarded as a working limit by many, and surveys suggest that speeds of survivable collisions are often far higher than that at which survival would be possible if this lower limit were valid, it is not considered further here.

S.I. (Severity Index) is a method of assessing injury by weighted integration of the acceleration (though other parameters may be used) against time, developed by Gadd (7). S.I., when used with the recommended weightings, yields (for simple pulse shapes) values similar to H.I.C. Both the 80g 3ms exceedance and S.I. were derived from Wayne State data, and H.I.C. is a development of S.I. but with different limits to the integration. Versace (8) and Newman (9) have demonstrated that simpler expressions than H.I.C., for example

$$a^{(t + 0.0015)} = 0.7$$

can describe the Wayne State data over the "time range of interest", but there are no reports of the use of such indicators.

Some researchers recommend the use of peak acceleration without any limits of duration as a standard. Some crash helmet test standards specify peak acceleration as the criteria. British Standard 6658 permits a linear acceleration of up to 300g. Curiously, a higher acceleration is permitted for cycle crash hats. Newman (5) suggests that only combined rotational and linear peak accelerations need be considered when predicting injury. In the process of validating "Gambit", Newman's injury prediction model, loci are presented to show experimental data filtered at different frequencies and compared with data predicted by "Gambit". Filtering at 100 hz gives the closest agreement. This implies that not only is duration unimportant but that peaks which occur at frequencies above 100 Hz are also unimportant. This is a conclusion which contrasts markedly with other research.

Newman has recommended the use of a simple linear injury scale based on peak resultant acceleration as follows:-
<50g = A.I.S. 0, 50-100g = A.I.S. 1, 100-150g = A.I.S. 2, 150-200g = A.I.S. 3, 200-250g = A.I.S. 4, 250-300G = A.I.S. 5 and >300g = A.I.S. 6.

It is clear that biological materials are not brittle, so a test based on peak acceleration is difficult to interpret.

Unfortunately there is no general agreement on the degree of viscous elasticity. Gadd (7) has demonstrated a simple method to describe the behaviour of materials which are neither entirely brittle nor viscous elastic: his reasoning is sound and the coefficients and factors suggested fit independently obtained data. On this basis it is reasonable to include a degree of time dependence in standard tests. In the B.S. tests velocity is stated so the maximum pulse length is implicit, but the worst case H.I.C. value is very large (7000). Kessler (6) has investigated sufficient cases for a statistical analysis, and his conclusion that a H.I.C. value of 1000 corresponds to a 10% probability of death, with a H.I.C. of 2000 representing a 50% probability, is likely to be more valid than most.

ANALYTICAL METHODS

The above methods of injury prediction are empiric rather than analytic. Increased computing power now makes an analytical approach practical. Predictive models are available, for example the M.S.C. model developed by Stalnaker (13). This is a two mass mass spring-damped system which allows the head to be modelled from first principles, but its predictive ability is uncertain, since direct verification is not possible.

For a modelling approach to be useful the properties of the materials involved must be known. Thibault and Gennarelli (12) showed that 5 to 10% strain produces recoverable injury to the axonal membrane, but above 25% strain the injury is irrecoverable. However the specimen used was the axon of the giant squid and it is not known how this compares with a human brain. Livers (14) in his work on side impacts on motorcycles using the M.S.C. model suggests that as little as 0.6% strain in the brain may cause injury. This criterion when presented in terms of acceleration versus time (fig.1), is much more conservative than a H.I.C. of 1000.

ROTATION

Gilchrist (4) suggests that the maximum angular acceleration that can be tolerated is 4500 rad/sec 2 but a pulse length is not specified. Newman (5) proposes a criteria based upon a combination of peak rotational and linear acceleration (Gambit) and suggests that it is the resultant stress that causes injury. Newman states that maximum rotational and linear acceleration frequently coincide and never appear at distinctly different times. This implies that the injurious rotation is associated with the reaction generated on impact and not with whiplash associated with general body movement as suggested by Bothwell (15).

In linear impacts the energy absorber (liner) of a helmet has limited travel, so that if it is optimised for one set of conditions it may run out of crush in more violent impacts. Rotations do not have this constraint and a design optimised for modest angular accelerations is likely to remain effective at high values. Injurious rotation could therefore be kept to a minimum by ensuring that the frictional coefficient is kept as low as possible. Ways of reducing the friction to a value even lower than that required to pass B.S. 6658 are being investigated and will be reported on in a later paper.

THE IDEAL HELMET

The foregoing discussion establishes that head injuries are complex and that no single indicator is likely to be satisfactory. At the moment,, there is no generally accepted alternative to the H.I.C., and in this paper therefore it has been used to estimate the protection that may be afforded by a normal sized helmet.

It is generally considered that for a given set of conditions a uniform acceleration is the least injurious. A uniform acceleration gives the lowest possible value for a given change of speed, and an energy absorber that achieves this provides the "ideal" helmet. (N.B. all estimates of H.I.C. given here are based on a square wave form. Some researchers use a triangular wave form enclosing the pulse, which gives much larger H.I.C. values.)

Calculations assuming uniform, acceleration indicate that if H.I.C. 1000 is a valid fatality threshold, then even with an "ideal" helmet protection at impact speeds above 7m/sec (25km/h) normal to the helmet surface is not possible with existing sized helmets, which provides a thickness of 20 mm total crush. It will be shown later that current helmets are far from the ideal and offer protection only up to impact speeds of about 5 m/sec. Even so accident studies have shown that wearing a helmet greatly reduces the risk of a fatal injury. From this it can be deduced either that in many impacts the "normal" velocity is below 5 m/sec, or it may also be the case that a H.I.C. value of 1000 is a conservative estimate for the fatality threshold for motorcyclists. As noted previously Kessler's research (6) supports this latter supposition, suggesting that a H.I.C. of 2000 may be a more realistic figure, but even using this higher threshold the "normal" velocity that an ideal helmet will protect against rises to only 8.25 m/sec (fig 2 shows that a small change of velocity produces a large change of H.I.C.). On the other hand, if the 80g for 3 msec criterion corresponding to a H.I.C. of 177 is used this velocity is only 4.4 m/sec, suggesting that this is a very conservative criterion indeed.

Accident studies have shown that about 75% of accidents occur at a motorcycle impact speed of 48 km/h (30 mile/h) or less and over 90% occur at 64 km/h (40 mile/h) or below. The normal component of head velocity on head impact will in most cases be a good deal less than this (many head impacts occur at glancing angles, or into yielding surfaces), and many of the fatalities are occurring at a velocity which current-sized helmets could be made to protect against. A series of tests were performed on a sample of current helmets to determine how close they come to the predicted ideal.

TESTS

1. APPARATUS

The drop-test apparatus consists of two parallel vertical taut wire guides with a P.T.F.E. ferrule on each. The helmet and headform are suspended from the ferrules which are released simultaneously by a solenoid, allowing the helmet to fall freely with the ferrules. These are extremely light and have no effect on the impact. The complete system can be raised to any height up to 7.9m (26ft), which provides a maximum impact speed of 12.5m/s (28 mile/h). The helmets tended to be resilient and bounce upwards, and to prevent a second impact a catching device was constructed. This consists of a conical net with a hole through which the helmet passes, first on the drop and then on the rebound. The hole is then drawn closed catching the helmet (see plate 1).

A solid wooden headform of B.S.I. type 6489 and mass 5kg has been modified by the insertion of a tapered hollow steel cylinder. A tri-axial accelerometer (Endevco type 7267A) is fixed to the base of the cylinder, which is at the centre of gravity of the headform. This sub-assembly is drawn into the headform by a bolt which passes through the crown and is recessed into a counter bore. When correctly assembled there is negligible spurious response and so filtering is not required. This is important as displacement is derived by double integration. The impacts are normal to a rigid piezo electric transducer (Kistler type 9293) mounted on a 1000 kg anvil. The ouput data from the transducers is captured on a 12 bit digital recording system sampling at 100khz, and although frequency filtering is not required for the analogue signal, to avoid aliasing in the digital recordings, the recorder is preceded by an analogue low pass filter with a 68db per octave attenuation at 4khz. Digital filtering to S.A.E. J211B is available, although in most cases the data was not filtered, as the case for the use of filters to mimic biological response is unproven and their use in research is probably best avoided (Searle (10) Hodgson (11).

An example of the output is shown in (fig.3) where resultant acceleration, velocity, displacement and force are plotted against time. Force is calculated by integrating the acceleration and by applying F = ma. Also plotted is the load as seen by the force transducer, and two force displacement curves. The velocity at impact is calculated from the drop height, and the position of impact is calculated from the output of the tri-axial accelerometer. H.I.C. exceedence of 80g and total velocity change are also calculated and recorded.

IMPACTS

Each helmet was dropped onto five sites, the crown, the front, each side and the rear at 45 deg to the vertical. Three stiffnesses of glass fibre shell were tested, standard, stiff and very stiff of relative stiffness 1.0, 1.5 and 1.8. Four densities of polystyrene liner were used 25 g/l, 32 g/l, 44 g/l and 55 g/l and all possible shell/linear combinations were tested.

Each combination of shell and liner was tested at 6.7m/s (15 mile/h) on each of the five sites. Standard helmets and 25 g/l liners were tested at different velocities, and a sample of each liner was tested at 6.7 m/s without the shell. A purely experimental helmet consisting of an 18 S.W.G. aluminium shell and polyurethane liner was impacted at 6.7 m/s. Two types of cycle helmet were tested, one similar to a motorcycle helmet and the other of the traditional padded bar type.

RESULTS

Table 1 gives a summary of the tests of standard helmets and shows helmet type, H.I.C. and 80g exceedence. Examination of the results shows that, for conventional helmets, the highest H.I.C. from a 6.7 m/s impact was 3914 and was recorded from a crown drop of a very stiff shell fitted with a high density 55 g/l liner. The lowest value 1353 was obtained with a standard shell and a 25 g/l low density liner. Overall, the lowest H.I.C. was 587 with a 25 g/l liner and no shell, seen in the summary of results for "experimental" helmet forms in table 2.

The trend was for the H.I.C. to increase as the stiffness of the shell and liner density increased (fig.4). This was accompanied by an increase in rebound velocity which was large for all the tests on the conventional helmets and was typically 0.6 of the initial velocity. This high rebound was largely a function of the shell design, since the liner alone gave typically only 0.3 of the initial velocity. An experimental helmet consisting of an aluminium alloy shell and 29 g/l polyurethane liner gave an H.I.C. of 602 at an impact velocity of 6.7 m/s (see fig.5), with a rebound velocity of 1.69 m/s. Tests on a standard helmet at different velocities are plotted in fig.6. They show that the percentage of energy absorbed increases with velocity, but it reaches high levels only at speed beyond those at which survival is likely, about 5 m/s.

For comparison and interest a range of cycle helmets (hard and traditional) were tested. These gave some unexpected results. The hard helmet gave an H.I.C. in excess of 5000 from 6.7 m/s (the recorder overloaded). The traditional helmet could not be tested at this velocity because of the risk of damage to the transducers. Tests at lower velocities produced overloads when the absorber bottomed out. This prevented calculation of H.I.C. and rebound velocity though the latter was observed to be modest. Nevertheless, although the protection afforded by these helmets is limited to very low velocities examination of the plots revealed excellent ride down until the helmet ran out of travel, so that within its limitations the energy absorbing mechanism is good.

DISCUSSION

PROTECTION RELATED TO H.I.C.

Our test results and H.I.C. values are similar to those measured by Grandel and Schaper (17) who concluded that the liner space is poorly utilised and that polystyrene will crush satisfactorily to only half its depth. Helmet dynamics do not seem to be universally understood. A well known cycle helmet bore the warning "insufficient strength for motor vehicle use", yet when tested at 6.7 m/sec a H.I.C. greater than 5000 was recorded because the very dense liner had crushed very little. The helmet was, in fact too strong at any speed. Conversely, a soft type cycle helmet which is often viewed as offering little protection did indeed give a very high H.I.C. at an impact speed of 4.4 m/sec, but inspection of the curves showed that up to the limit of crush the energy had been absorbed in a near perfect way. Up to this point, corresponding to an impact velocity of 3 m/sec, a H.I.C. of 167 was calculated. This helmet was constructed with bars fore and aft over the top of the head and it is estimated (making allowance for the low test velocity) that with bars of twice the diameter this helmet would have out-performed the best motorcycle helmet. The energy absorber was cross-linked polyethylene and the potentially superior performance was probably the result of high stress in the material rather than its properties. Small bars, however apparently effective when tested, may induce loads which the skull cannot sustain and may be unsuitable. Biomechanical data for concentrated skull loads is not available.

Resilience is a problem with both the shell and the liner. Polystyrene is used universally for liners and has a typical rebound of -0.3. However when tested alone it sustains an H.I.C. value very much lower than a complete helmet. A standard liner in a non-resilient shell (aluminium alloy) is a significant improvement and substituting polyurethane for polystyrene gives a result close to the "theoretical" best.

The current British Standard 6658 permits a linear acceleration of up to 300g from an impact velocity of 7.5m/sec, but the pulse length is not specified. It follows from this that a helmet with typical rebound characteristics, ie. where the total velocity change is 1.6 times the initial, could give a clearly fatal H.I.C. of 6800 yet still pass B.S. 6658. It seems therefore that the standard is inappropriate . As time dependence is real it should be included in a standard test. H.I.C. was derived for car impact tests where there are multiple events and unpredictable pulse shapes. The calculation of H.I.C. requires a two-dimensional search of the data to find the worst case this requires about twenty minutes on a "micro", which may be unsuitable for routine testing. However, for simple pulse shapes the simple weighted integration of the S.I. calculation yields a similar value.

It is interesting to note that H.I.C. values recorded by Chinn (16) at T.R.R.L. during experimental collisions between moving cars and motorcycles to develop leg protectors were much lower than anticipated. Though head impact velocities of up to 18.3 m/sec were recorded it was rare for the H.I.C. to exceed 1000. By contrast Table 1 shows that test drops at 6.7 m/sec of helmets which pass British Standard 6658 produce H.I.C. values well in excess of 1000. It seems that the car is a better energy absorber than the helmet!

PROTECTION AT VELOCITIES ABOVE 8m/sec

Gilchrist (4) suggests that the ranking order of helmets B.S.2495 and 6658 will be reversed at "higher" impact energies. The B.S. 2495 and 6658 specify peak accelerations of 400 and 300g respectively at 7.5 m/sec. A helmet which meets the lower requirement will bottom out more readily at higher speeds. The results described in this paper suggest that impacts above 160 J(impact velocity 8.0 m/sec) will cause death, even with an "ideal" helmet so ranking of helmets above this level is purely academic. It is confirmed however that helmets to B.S. 2495 appear to give better protection at high energies ie. above 160 J (8.0 m/sec).

With present helmet design, the calculation described previously suggests that protection much above 8.0 m/sec is not possible with existing sized helmets. If this is correct, then the possibility of protection at higher speeds can only be considered if the principle of energy absorption at frequencies above those to which the brain will respond is considered. There is anecdotal evidence (18) to suggest that a pulse of less than 0.6ms is not injurious, in which case a sacrificial helmet shell which shatters and reduces the energy to a level which could be absorbed by the liner may provide a possible solution.

CONCLUSIONS

1. It can be demonstrated using theory that the performance of existing sized helmets could be significantly better in linear impacts if shell resilience were lower and better use was made of the available liner crushing space. It is estimated (using H.I.C.= 1000 as the criterion) that the

velocity up to which existing helmets would be effective could be increased by up to 60%. The present design of helmet absorbs energy efficiently only at velocities at which survival is highly unlikely.

2. The results from the experimental helmet show that if the crush depth is fully utilised a significantly lower H.I.C. value can be obtained at an impact from 6.7 m/sec compared with a standard helmet. However once the available crush is fully utilized the helmet can offer no further protection and the accleration seen by the head will rise extremely rapidly. It is very important therefore that if potential improvement in performance at low velocity is to be realised, the human tolerance must be known with more certainty. Researchers have suggested limits ranging from H.I.C. values of 176 to 2000. The generally accepted figure is 1000, which implies that protection can be provided up to a maximum "normal" impact speed of 7 m/sec. There is evidence to suggest that a H.I.C. value of 2000 might be a more realistic criterion: this represents a level of injury causing roughly 50% fatality, and a maximum impact speed of 8.25 m/sec.

3. The criteria by which helmets are judged should be based on a weighted integration of acceleration against time. The use of a peak value of acceleration without limit on duration can permit helmets to pass current tests and yet have an unacceptably high H.I.C. (up to 6800 in the case of B.S. 6658). S.I. (equivalent to H.I.C. for simple pulses) would provide a simple method of calculation.

4. The investigation reported in this paper has been concerned with linear impact velocities and the protection afforded by a crushable liner. Rotational acceleration is also an important cause of injury, and needs further examination. Problems over bottoming out of a crushable energy absorber do not apply to rotational acceleration, so devices which provide protection at low angular accelerations should not hinder the performance at high ones.

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TABLE 1

RESULTS FROM TESTS OF STANDARD TYPE HELMETS

IMPACTS AT 6.7 M/SEC (15 MPH)

SHELL	LINER DENSITY		SITE 1	2	3	4	5
LAMINATED	g/l						
STANDARD	25	HIC	1825	1367	1606	1353	1494
		TIME	5.2	5.6	5.4	5.9	5.6
STANDARD	32	HIC	1721*	1507	2481	1132	2056
		TIME	4.0	5.4	4.5	4.4	4.9
STANDARD	44	HIC	3383	1851	2421	2001	2480
		TIME	4.1	4.8	4.4	4.5	4.5
STANDARD	55	HIC	3351	2511	2511	2250	2870
		TIME	3.9	4.4	3.9	4.1	4.5
STIFF	25	HIC	2469	1517	2076	1726	1862
		TIME	5.4	5.9	5.2	5.3	5.6
STIFF	32	HIC	2711	2065	2275	2445	2089
		TIME	5.1	5.3	5.2	5.2	5.1
STIFF	44	HIC	3599	2232	2219	2348	2524
		TIME	4.5	4.8	4.2	4.4	4.7
STIFF	55	HIC	1990*	1708*	3231	2365	2626
		TIME	4.3	4.7	4.0	4.0	4.6
V STIFF	25	HIC	2241	1613	1965	1881	NA
		TIME	5.3	5.7	5.4	5.4	NA
V STIFF	32	HIC	2984	1963	2437	2505	2331
		TIME	4.8	5.3	4.9	5.1	5.1
V STIFF	44	HIC	3709	3112	2799	2523	3119
		TIME	4.4	4.6	4.2	4.2	4.6
V STIFF	55	HIC	3914	2644	3044	3137	3516
		TIME	4.2	4.6	4.5	4.6	4.4

RELATIVE STIFFNESS

STIFF = 1.5 * STANDARD   V STIFF = 1.8 * STANDARD

SITE 1 = CROWN = FOREHEAD   3 = R SIDE   4 = L SIDE   5 = NAP

HIC = HEAD INJURY CRITERION

TIME = TIME IN MILLISECONDS FOR WHICH 80g IS EXCEEDED.

* = RECORDED BUT UNRELIABLE (INCORRECT ASSEMBLY)

TABLE 2

EXPERIMENTAL HELMETS

CROWN IMPACTS AT 6.7 METRES PER SECOND

DESCRIPTION	H.I.C.	TIME 60g	TIME 80g	PEAL g
SHELL LINER		ms	ms	
STD BS 6655	3351	4.3	3.9	305
ALLOY CORED STYR	2753	4.2	3.9	300
ALLOY OASIS	1887	3.5	2.6	318
ALLOY STYR	1237	5.8	4.7	205
ALLOY URTH	602	7.9	5.7	10
ALLOY *	1052	7.4	6.4	145
STD #	864	9.3	7.9	111
STD *	1721	6.9	5.9	199
STC STYR 25g/1	1825	5.7	5.2	189
LINER ONLY 25g/1	587	8.1	5.9	97
CYCLE HARD	14000	1.7	1.6	600
CYCLE (4.4 m/s)	14000	1.5	1.3	600
CYCLE +	1280	5.3	4.8	182
INTEGRAL MOPED	2481	4.9	4.1	283
FRENCH SPEC				

* = POLYURETHANW CAST ROUND HEAD IN STANDARD SHELL

\# = LOW DENSITY POLYURETHANE CAST IN STANDARD SHELL

\+ = TWO SOFT CYCLE HELMETS TAPED TOGETHER

OASIS = PROPRIETRY FOAM

TIME xg = TIME ms FOR WHICH ACCELERATION EXCEEDED xg

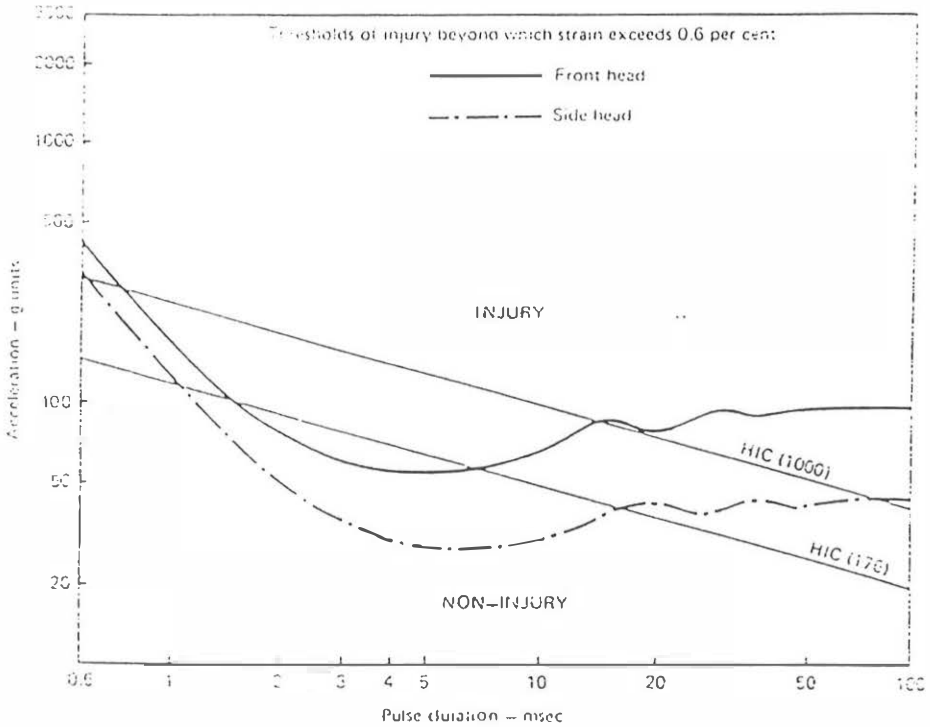

Fig 1 Maximum strain criterion for humans (Ref. 14)

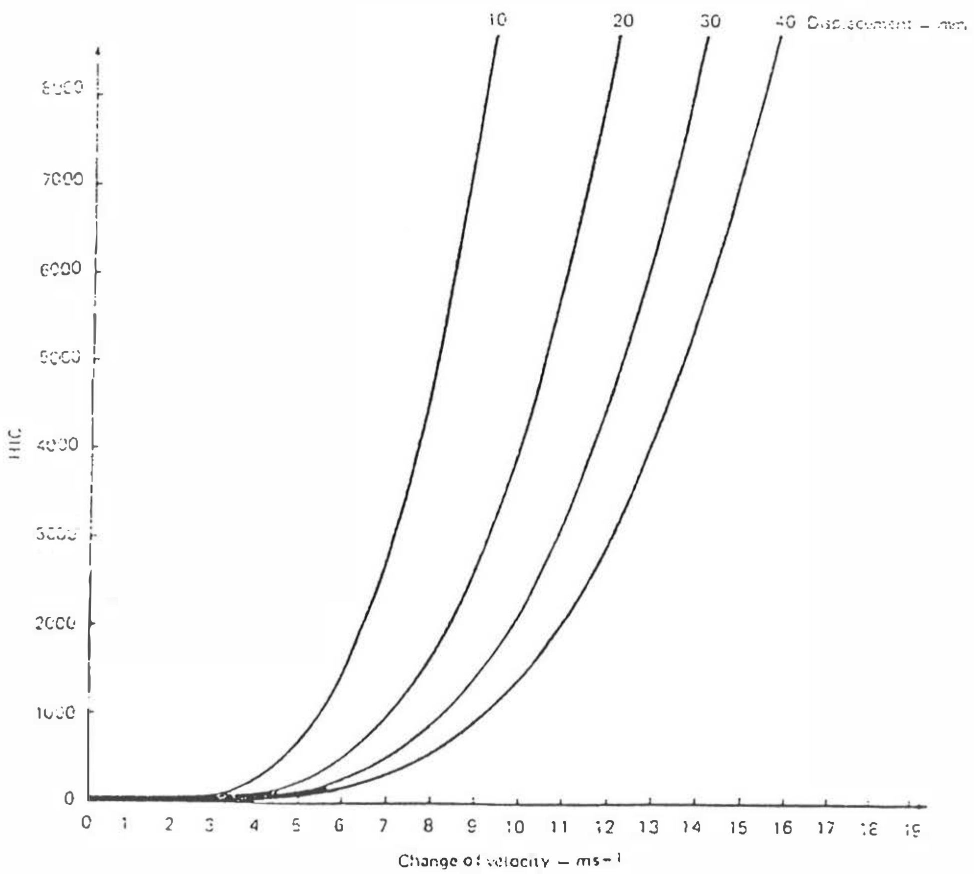

Fig. 2 Plots to predict HIC from stopping distance and change of velocity

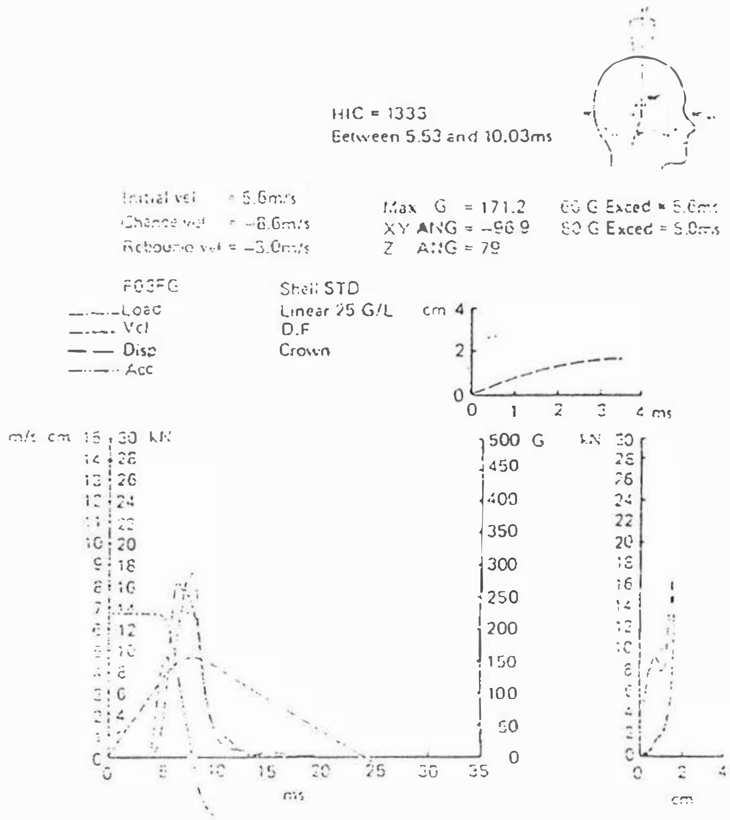

Fig. 3 Plots of Acceleration, Displacement, Force and Velocity

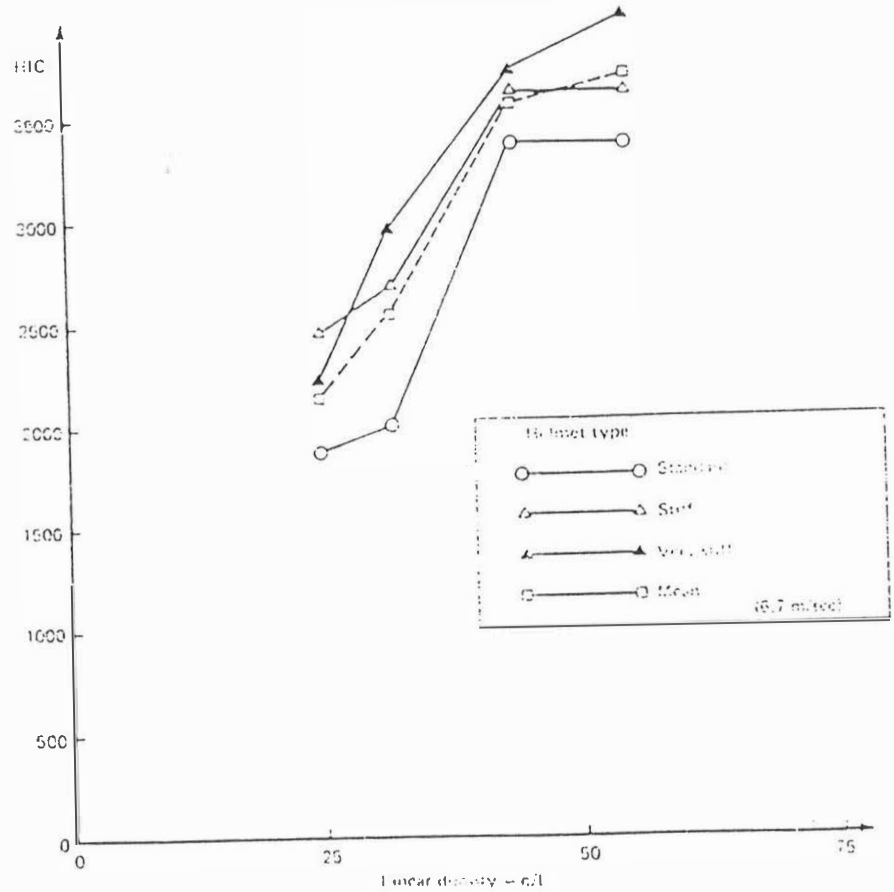

Fig. 4 HIC Versus Linear Density for Different Helmet Types

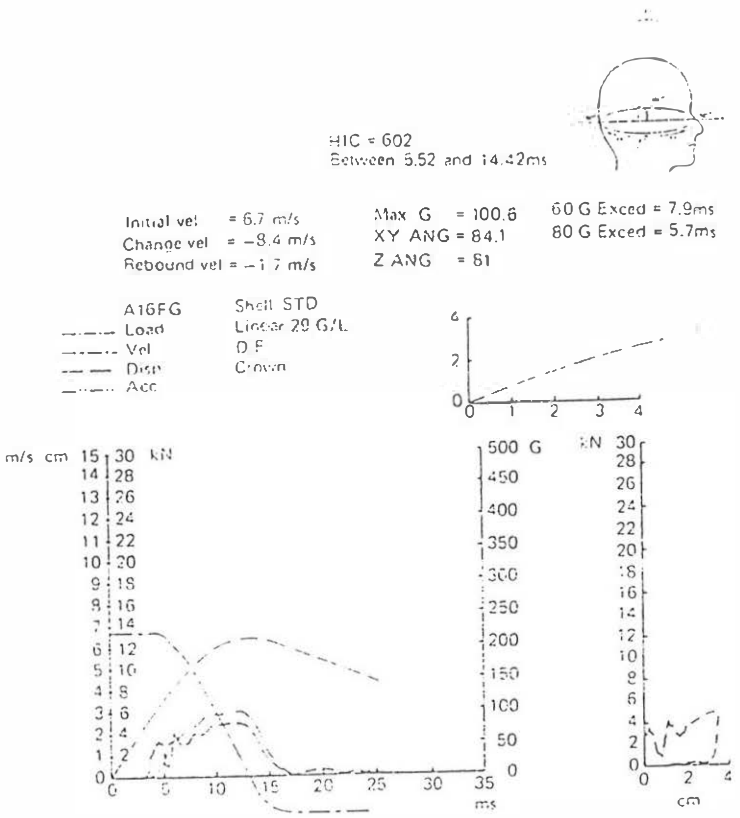

Fig. 5 Plots showing Acceleration, Displacement, Force and Velocity for the Experimental Helmet

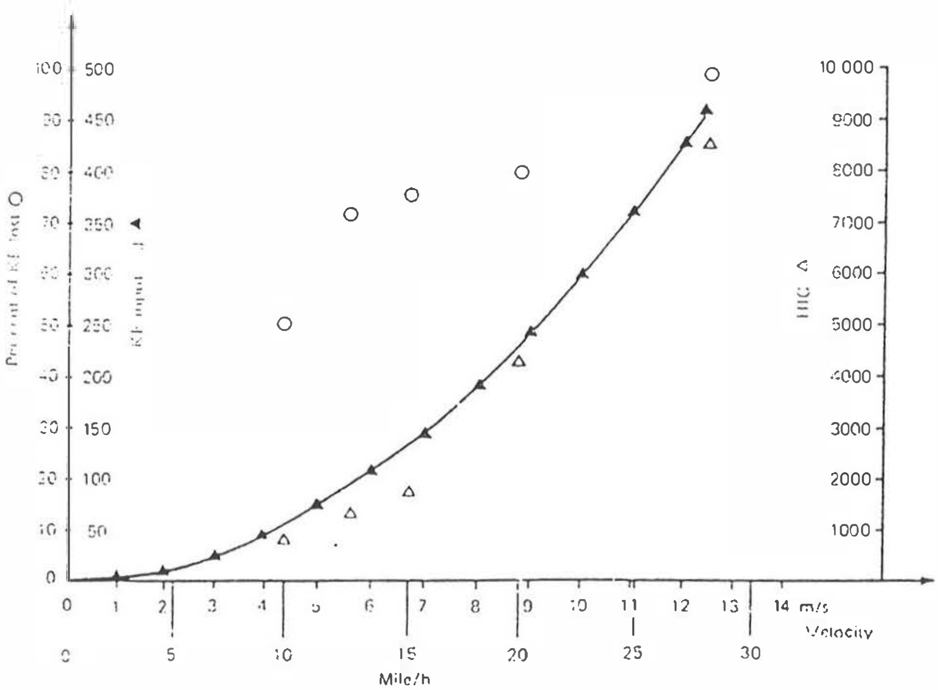

Fig.6 Plot to show loss of kinetic energy (KE) against impact velocity for Open Faced Standard Shell 25 gram/litre linear helmet

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