COMMENTARY

Assessment of current bicycle helmets for the potential to cause rotational injury

V J M St Clair, B P Chinn. TRL Project Report PPR213, 2007, ISBN: 978-1-84608-846-9

Original paper available for purchase External Link

Summary of paper (from authors' abstract)

Concern has been expressed that current bicycle helmets may increase the risk of brain injuries from rotational motion. A range of child and youth bicycle helmets have therefore been tested to evaluate their linear and oblique impact performance. This data was used to assess the propensity of the helmet to influence rotational motion and was considered against post-mortem human surrogate data to allow comparison of the risk of injury to that of an unhelmeted head.

Un-helmeted post-mortem human surrogate data indicates that a simple skull fracture for an unhelmeted head (injury rated as AIS 2) may occur at 5kN - 6kN which corresponds to between 100g and 150g for a head mass of between 4kg and 5kg. Assuming that the response of the unhelmeted head is similar to the helmeted head during an oblique impact at 8.5m/s at 15º, this may generate between 7500rad/s² and 12000rad/s² of rotational acceleration. This is potentially more severe than the 3000rad/s² to 8500rad/s² measured during abrasive and projection oblique tests with size 54cm (E) helmeted headforms. However, for the most severe cases using a size 57cm (J) headform, rotational acceleration was typically greater than 10,000rad/s² and increased to levels of 20,000rad/s², a level at which a 35% - 50% risk of serious AIS3+ injuries is anticipated.

Overall, it was concluded that for the majority of cases considered, the helmet can provide life saving protection during typical linear impacts and, in addition, the typical level of rotational acceleration observed using a helmeted headform would generally be no more injurious than expected for a bare human head. However, in both low speed linear impacts and the most severe oblique cases, linear and rotational accelerations may increase to levels corresponding to injury severities as high as AIS 2 or 3, at which a marginal increase (up to 1 AIS interval) in injury outcome may be expected for a helmeted head.

The true response of the bare human head to oblique, glancing blows is not known and these observations could not be concluded with certainty, but may be indicative of possible trends. A greater understanding is therefore needed to allow an accurate assessment of injury tolerance in oblique impacts. Linear impact performance, head inertia and helmet fit were identified as important contributory factors to the level of induced rotational motion and injury potential. The design of helmets to include a broad range of sizes was also concluded to be detrimental to helmet safety, in terms of both reduced linear and rotational impact performance. The introduction into EN1078 of an oblique impact test could ensure that helmets do not provide an excessive risk of rotational head injury.

BHRF Commentary

Summary of comments

Rotation of the head was identified as a potential mechanism for, and a significant cause of, brain injury in the literature review of TRL Report PPR213. Tests using disembodied headforms showed that, for a 15 degree oblique impact at 19.6 miles/hr (vertical speed 4.7 miles/hr, horizontal speed 18.4 miles/hr) rotations could have a greater injury potential than direct impact forces.

For the 57 cm headform, which fitted more snugly than the 54 cm headform, rotational accelerations from the oblique impact test averaged 13,500 rad/s², a level expected to produce a 35-50% risk of serious AIS3+ brain injury, higher than predicted for a non-helmeted head. Report PPR213 adds: *"For rotational accelerations the research shows that concussion, AIS 1-2, can occur at 5,000 rad/s² and fatal injury AIS 5-6 at 10,000 rad/s². This*

correlates with data from the same research that indicates there is a 35% risk of a brain injury of AIS 3-6 at 10,000 rad/s².”

Tests using crash dummies produced even more horrific results. Dummies in bicycle helmets had average rotational accelerations of 58,000 rad/s² (six times the threshold for potentially fatal brain damage) in tests simulating going over the handlebars at 45 km/hr. Motorbike helmets performed much better, producing only half the rotational accelerations measured for dummies wearing bicycle helmets (Corner, Whitney, O'Rourke and Morgan, 1987).

Corner, Whitney, O'Rourke and Morgan, 1987 cited research showing that that even motorcycle helmets would not slide on a smooth surface in tests with forward velocity and a drop height of 1.4 m, presumably because of the high normal force acting during the impact. A test for sliding impact friction was therefore recommended for both motorbike and bicycle helmets (Corner, Whitney, O'Rourke and Morgan, 1987) but, unfortunately, never adopted for bicycle helmets. In contrast, the standard for motorbike helmets (Reg 22.05) now includes a test for tangential force in oblique impacts (which has a similar function to a sliding test). The TRL Report recommends a similar test for bike helmets.

Although the TRL report claims helmets can provide lifesaving protection during typical linear impacts, it is not clear if tests using disembodied headforms provide a realistic representation of the injuries produced in a crash. Dutch research on non-helmeted dummies shows that the head often impacts after the arms and shoulders. A helmet, if worn, it is likely to impact sooner and with greater force than a non-helmeted head.

TRL report PPR213 also presents some interesting and useful results showing that helmets actually provide very limited protection against linear accelerations. For side impacts at 12.1 miles/hour, half the helmets had peak linear accelerations in excess of 200 g, corresponding, according to Table 4.7 of the report, to severe, critical or unsurvivable head injuries

If more people knew and understood these results, were aware that the vast majority of seriously debilitating head injuries are caused by bike/motor vehicle collisions often involving forces outside the design limit for bike helmets, and knew about the Dutch research showing that a reduction in vehicle impact speed from 40 to 30 km/hr halves maximum head acceleration (Janssen and Wismans, 1985), there might be more emphasis on measures to prevent high impact car/bike crashes (e.g. by education, enforcement of traffic laws and random breath tests) rather than on secondary measures such as cycle helmets, for which the benefits are much less certain.

Literature Review

TRL report PPR213 contains a review of recent literature including research suggesting that rotations of 10,000 rads/s² have a 35% probability of causing a head injury of severity AIS 3-6. The report adds: *“For rotational accelerations the research shows that concussion, AIS 1-2, can occur at 5,000 rad/s² and fatal injury AIS 5-6 at 10,000 rad/s². This correlates with data from the same research that indicates there is a 35% risk of a brain injury of AIS 3-6 at 10,000 rad/s².”*

These results demonstrate the importance of Australian research (not mentioned in this review) simulating going over the handlebars at 45 km/hr and hitting a smooth surface (Fig 1). Measured rotational accelerations of dummies wearing bicycle helmets (averaging 58,000 rads/s², nearly 6 times greater than the 10,000 rad/s² noted above to represent a potentially fatal injury) were described as “*enormous*”. Corner, Whitney, O'Rourke and Morgan, 1987 compared these values with tolerances of 1,800 rad/s² for concussion and 4,500 rad/s² for onset of vein rupture (which leads to bleeds in the brain) . Rotational accelerations in dummies wearing motorbike helmets averaged 30,000 rads/s², half those of

dummies wearing bicycle helmets.

Rotations cause serious brain damage

The research of Corner, Whitney, O'Rourke and Morgan, 1987 as well as that reported in PPR213 are of great concern because it has long been known that rotational accelerations are a significant cause of brain damage. Rotations can shear the brain's neuronal connections, which are a vital part of our thought processes. This condition is known as diffuse axonal injury. Unlike a focal injury which occurs at a specific site in the brain (e.g. the site of impact), diffuse injuries can happen throughout the brain.

One of the most interesting experiments, conducted by Gennarelli et al, 1982, subjected 12 squirrel monkeys to linear accelerations with peak levels 665-1230 g and 13 other monkeys primarily to rotational accelerations from 348 to 1025 g. Contact phenomena were minimised by the design of the apparatus. None of the monkeys receiving linear acceleration was concussed, but all 13 receiving rotational acceleration suffered concussion, and the group had a high incidence of brain injuries such as subdural haematoma, subarachnoid haemorrhage and intracerebral petechial haemorrhage (bleeding and bruising of the brain).

The Australian NHMRC (National Health and Medical Research Council) discussed the possibility that football helmets might increase rotational injuries: "*The use of helmets increases the size and mass of the head. This may result in an increase in brain injury by a number of mechanisms. Blows that would have been glancing become more solid and thus transmit increased rotational force to the brain. These forces result in shearing stresses on neurones which may result in concussion and other forms of brain injury.*" (NHMRC, 1994)

Anecdotal evidence lends some support to this hypothesis: For example, public health physician, Dr Ashley Bloomfield wrote (Bloomfield, 2000): "*The earliest murmurings that I heard against helmets ...[were from] ... a neurosurgeon whom I worked for in 1994. He claimed that cycle helmets were turning what would have been focal head injuries, perhaps with an associated skull fracture, into much more debilitating global head injuries. We had a couple of examples on the ward at the time*".

Experiments using helmeted headforms

Instead of more realistic tests using dummies, the TRL research tested helmeted headforms impacting an abrasive anvil at an angle of 15 degrees and a nominal speed of 8.5 m/s (19 miles/hr, 30.6 km/hr). Measured accelerations of the 54 cm helmeted headform ranged from 3,000 to 8,500 rads/s². However, those of a 57 cm helmeted headform typically exceeded 10,000 rads/s² and increased to levels of 20,000 rad/s². Report PPR213 explains that this is a level at which a 35% - 50% risk of serious AIS3+ injuries would be expected.

This evidence led the authors to conclude that injuries from this type of impact are more likely to be due to rotations than linear impacts. In other words, for an impact speed of 19 miles/hr, Report PPR213 shows that helmet wearers may have a significant risk (35-50%) of serious head injuries due to rotational acceleration.

The authors also comment on the effect of helmet sizes. For 3 of the 8 models tested, it was thought that the size range specified for the helmet had been purposely chosen to be less than the range of head sizes the helmet would fit, in order to avoid tests with the larger headform. The 57 cm headform has greater mass and so may represent a more severe test due to the increased impact energy. When tested with this headform and impact speed of 5.42 m/s (12.1 miles/hour), one helmet produced a peak linear acceleration of 285 g, failing the EN1078 standard of at most 250 g. The report comments "*The 285 g result is a consequence of the helmet's inability to absorb some, or any, of the additional impact energy associated with the larger headform size (72.3 J for size 57 cm compared with 63.8 J for size 54 cm).*

Consequently, the probability of fatal injury is approximately 55% and a critical injury would be highly likely”.

It was also speculated that helmets might rotate around the smaller headform, reducing the amount of rotational acceleration transmitted to the headform. This could explain why the smaller headform had lower rotational accelerations than the larger headform with a more snug (but still comfortable) fit.

Fig 4.10 of the report shows the relationship between peak linear acceleration of a helmeted headform and peak rotational acceleration. The authors speculate that, if the same relationship held for a bare head, a similar impact to the oblique test might correspond to 7,500rad/s² and 12,000rad/s² of rotational acceleration, substantially less than the 57 cm headform but greater than the 54 cm headform (see the abstract of the report). However, it is not clear why the same relationship would hold for a bare head, which has a much smaller diameter and hence a smaller turning moment, so much lower rotational accelerations might be expected

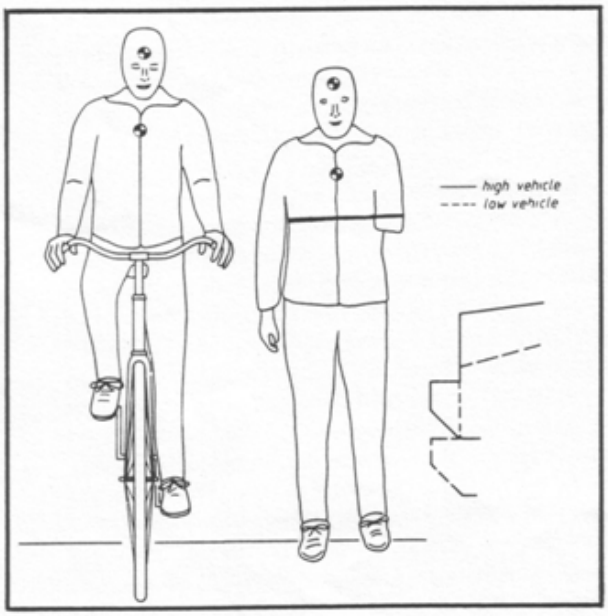
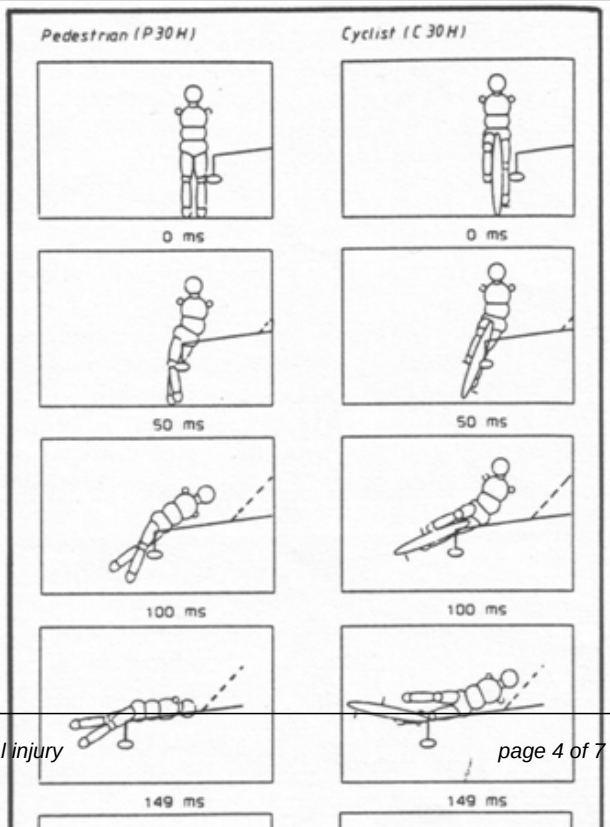


Fig 2 (from Janssen and Wismans, 1985).
Simulation of pedestrians and cyclists being hit by vehicles – initial set-up.



Evidence from dummies hit by ‘vehicles’

A significant weakness of this study is that there were no measurements on non-helmeted headforms, let alone measurements of helmeted and non-helmeted dummies. Consequently, the estimates for non-helmet wearers are a matter of speculation. In contrast, Dutch researchers (Janssen and Wismans, 1985) measured impact accelerations in non-helmeted dummy pedestrians and cyclists hit by an apparatus simulating the front of a vehicle (Fig 2). The results were used to create a mathematical simulation of the kinematics. Fig 3 shows that primary impacts sites are the legs, hips, shoulders and finally the head. Because of its greater size and mass, a helmet will probably impact sooner and with greater force and greater turning moment than a bare head.

The Dutch research considered two impact speeds – 40 and 30 km/hr. Maximum head accelerations averaged 2033 and 1026 m/s² at 40 and 30 km/hr respectively; head injury coefficients (HIC) averaged 2556 and 548. This research clearly shows the benefit of reducing

headforms.

Helmet standards

The authors of the TRL report recommend *"the introduction into helmet standard EN 1078 of a test for tangential force in oblique impacts"*. Corner, Whitney, O'Rourke and Morgan, 1987 also recommended a test for sliding impact friction (which would serve a similar purpose) citing Aldman (1984) that, in motorbike helmet tests involving forward motion and a drop height of 1.4 m, no sliding occurred even on a smooth surface presumably because of the high normal force (i.e. force at 90 degrees to the surface) acting during the impact. Unfortunately this recommendation has not been adopted for bike helmets.

Corner, Whitney, O'Rourke and Morgan, 1987 also made several other important recommendations that were never adopted, including replacing the magnesium headform with a headform designed to react more like a person's head, such as that developed by Wayne State University. One consequence of using a more realistic headform (which would deform on impact similar to the way a head deforms) is that a stiff helmet liner would deform less, and so absorb less energy. Corner, Whitney, O'Rourke and Morgan, 1987 added *"The indications from experimental work are that current helmet liners are too stiff and a linear foam density of about 30 kg/m³ should be used rather than 50 kg/m³ "*.

The TRL report provides some interesting data to help interpret the linear acceleration measurements reported in helmet tests. For example, Table 4.7 provides the following information (which is repeated in Table 4.12)

Peak linear acceleration	AIS head injury severity	Injury interpretation	Approximate probability of fatality
< 50 g	0	No injury	0.0%
50 - 100 g	1	'Minor' injury	0.0%
100 - 150 g	2	'Moderate' injury	0.1 – 0.4%
150 – 200 g	3	'Serious' injury	0.8 – 2.1%
200 - 250 g	4	'Severe' injury	7.9 – 10.6%
250 – 300 g	5	'Critical'	53.1 – 58.4%
> 300 g	6	'Unsurvivable' (Maximum)	> 58.4%

The source of these values is not fully explained, nor why peak linear accelerations of 200–250 g have a 7.9–10.6% probability of fatality, with a huge jump to 53.1 – 58.4% probability for peak linear accelerations of 250–300 g.

The data in Tables 4.5/4.6 of the TRL report show that for side impacts at 12.1 miles/hour, half the helmets tested had peak linear accelerations in excess of 200 g, corresponding, according to the above table, to severe, critical or unsurvivable head injuries. Helmets would be expected to provide even less protection against the much higher impact forces often encountered in bike/motor vehicle crashes.

A UK study (Maimaris, Summers, Browning and Palmer, 1994) of emergency room presentations found that bike/motor vehicle crashes were responsible for only 25% of non-head injuries, but half of all head injuries. A US study (Kraus, Fife and Conroy, 1987) considered all head injuries to cyclists for an entire year in one county in Florida. Although a significant proportion resulted from bike only crashes, all fatal or seriously debilitating head injuries were caused by collisions with motor vehicles.

These results highlight the difficulty in setting appropriate standards for helmets. Although lower density foam might provide more protection in bike only crashes, the benefits are likely to be offset by increased risk of injury in high impact bike/motor vehicle crashes. Don Morgan, one of the authors of Corner, Whitney, O'Rourke and Morgan, 1987, described his experience when he " *went out with the traffic investigation squad to understand the accidents and to retrieve the helmets. What he discovered was bone fragments, fluid and teeth embedded into the foam but the liners showed little or no evidence of damage* ." (Morgan, 2007)

Don was so concerned that he continued to think about the problem. Almost 20 years later, he invented the Conehead liner that is expected to deform over a wider range of impact forces. Another person concerned about the serious brain damage suffered by helmeted motorcyclists was physician Dr Ken Phillips. He observed that the scalp helps protect the brain against rotational forces because it is elastic, compressible and moves around the

skull without friction. To mimic this process, he designed a helmet with an outer shell that moves independently of the inner cushion. No helmet manufacturer was interested, so Dr Phillips started his own company to make and market them (Phillips helmets).

However, until more comprehensive (and expensive) tests become mandatory, most helmets on sale will be designed to meet the current standards, and so provide no more protection than the liners inspected by Don Morgan that showed little or no evidence of damage despite bone fragments, fluid and teeth embedded into the foam.

New helmet standards could improve protection in low-impact crashes. Unfortunately, the vast majority of seriously debilitating brain injuries result from car/bike collisions and involve forces considerably greater than the absorption capacity of a lightweight helmet. It is therefore important to educate cyclists on the limitations of helmets. Many people have an exaggerated faith in the ability of helmets to offer adequate protection in most circumstances. A false sense of security is likely to increase injury rates. Education programs should therefore explain the limitations of helmets, and that impacts to the helmet at only12.1 miles/hour may produce severe or unsurvivable head injuries

SUPPLEMENTARY COMMENTARY

cyclehelmets.org has received further comments about this report which supplement the commentary above and which we commend to you.

You are invited to read the supplementary commentary here.

References

Bloomfield, 2000

Bloomfield A, 2000. Cycling: your health, the public's health and the planet's health. Making Cycling Viable symposium .
http://www.cyclehelmets.org/1146.html

Corner, Whitney, O'Rourke and Morgan, 1987

Corner JP, Whitney CW, O'Rourke N, Morgan De, 1987. Motorcycle and bicycle protective helmets: requirements resulting from a post crash study and experimental research. Federal Office of Road Safety Report CR55.

Gennarelli et al, 1982

Gennarelli TA, 1982. Diffuse axonal injury and traumatic coma in the primate. Annals of Neurology 1982;12:564-574.

Janssen and Wismans, 1985

Janssen EG, Wismans JSHM, 1985. Experimental and mathematical simulation of pedestrian-vehicle and cyclist-vehicle accidents. 10th International Tech Conference on Experimental Safety Vehicles Oxford July 1985.

Kraus, Fife and Conroy, 1987

Kraus JF, Fife D, Conroy C, 1987. Incidence, severity and outcomes of brain injuries involving bicycles. American Journal of Public Health 1987;77:76-78.. External Link
http://ajph.aphapublications.org/doi/abs/10.2105/AJPH.77.1.76

Maimaris, Summers, Browning and Palmer, 1994

Maimaris C, Summers CL, Browning C, Palmer CR, 1994. Injury patterns in cyclists attending an accident and emergency department: a comparison of helmet wearers and non-wearers. BMJ 1994 Jun 11;308(6943):1537-40.
http://www.cyclehelmets.org/1144.html

Morgan, 2007

Morgan D, . Cone head helmets. Click the 'Inspiration' tab. External Link
http://www.abc.net.au/tv/newinventors/txt/s2006698.htm

NHMRC, 1994

Football injuries of the head and neck. National Health & Medical Research Council, AGPS, Canberra .1994.

Phillips helmets

Phillips helmets. . External Link
http://www.phillipshelmets.co.uk/

The Bicycle Helmet Research Foundation (BHRF), an incorporated body with an international membership, exists to undertake, encourage and spread the scientific study of the use of bicycle helmets. Also to consider the effect of the promotion and use of helmets on the perception of cycling in terms of risk and the achievement of wider public health and societal goals.

BHRF strives to provide a resource of best-available factual information to assist the understanding of a complex subject, and one where some of the reasoning may conflict with received opinion. In particular BHRF seeks to provide access to a wider range of information than is commonly made available by those that take a strong helmet promotion stance. It is hoped that this will assist informed judgements about the pros and cons of cycle helmets.

For more information, please visit www.cyclehelmets.org.

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