HEAD AND NECK DYNAMICS IN HELMETED HYBRID III IMPACTS

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ABSTRACT

A rig was developed to permit impacts on a Hybrid III 50th-percentile male anthropomorphic test device (ATD) at selected velocities and head impact orientations in guided free-fall. This is to investigate the ability of a bicycle helmet to reduce angular head acceleration and subsequent brain injury in crashes, as well as the effects of helmets on neck injury. Helmets were able to reduce head linear acceleration significantly and angular accelerations in lateral impacts. There were no significant reductions in biomechanical predictors of neck injury risk in helmeted impacts as compared to unhelmeted impacts. At worst bicycle helmets do not appear to exacerbate head injury risks arising from angular acceleration or neck injury risks.

Keywords: BICYCLES, DROP TESTS, HELMETS, HYBRID III, INJURY CRITERIA

THE CAPABILITY OF BICYCLE HELMETS to reduce angular head acceleration and subsequent brain injury in crashes, as well as the effects of helmets on neck loading and injury, are two topics that have generated debate (Mills and Gilchrist, 2008; Curnow, 2003). There is currently no helmet standard test method that assesses angular acceleration, in part because this requires a biofidelic head and neck, as well as injury assessment reference values for angular acceleration.

It has been reported that bicycle helmets offered little benefit in reducing injuries at the junction of the head and neck (Hitosugi et al., 1999). Some researchers have also questioned whether the extent of head coverage currently offered by helmets is sufficient (McIntosh et al., 1998, Depreitere et. al., 2007; Ching, et. al., 1997). The base of skull and upper neck are also exposed to direct impact without protection.

This project's main objective was to investigate neck loading and angular head acceleration in helmeted head impacts. A broader objective is to compare helmet performance in energy and velocity equivalent impacts between a full Hybrid III ATD, Hybrid III headform ± neck and rigid headform.

DUMMY DROP TEST

A test rig was constructed to allow a free falling instrumented 50th-percentile male Hybrid III Anthropomorphic Test Device (ATD) to impact a flat rigid surface (Fig. 1). Three impact locations were selected: frontal, temporal-parietal and crown impact (Fig. 2). The dummy was restrained in the required position with safety harnesses. The test configuration is presented in Table 1.

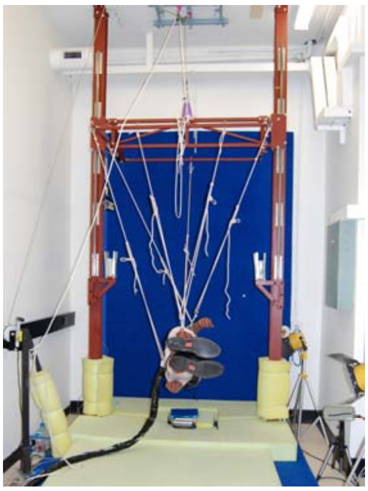

Fig. 1 Dummy drop test rig

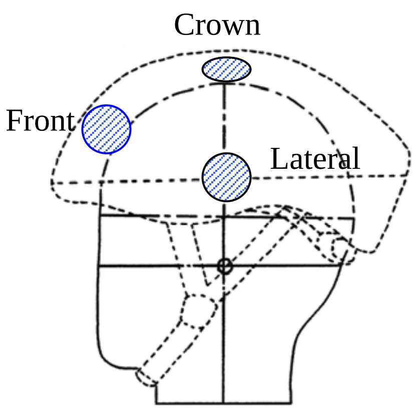

Fig. 2 Head impact sites

The test procedure and range of drop heights were determined after considering the robustness of the ATD. Helmeted and unhelmeted tests were conducted. The bicycle helmets used in the tests had an ABS shell and expanded polystyrene foam liner. The helmets had been certified by the Australian/New Zealand Standard AS/NZS 2063 *Pedal Cycle Helmets*. ATD data were utilised to calculate the Head Injury Criterion (HIC) and the N_{ij} neck injury criterion. The ATD was instrumented with a three-axis angular velocity sensor at the head centre of mass. Angular velocity was used to derive angular displacement and acceleration. The ATD data were acquired in accordance with SAEJ211. Differences between sample means were assessed using a t-test (α=0.05).

Table 1. Dummy drop test configuration

No. of Test	Velocity (m/s) [SD]	Impact angle (°)	Head impact point	Helmet
5	3.18 [0.02]	30 from horizontal	Frontal	Bicycle
2	3.16 [0.02]	30 from horizontal	Frontal	None
4	3.07 [0.21]	30 from horizontal	Temporal-parietal	Bicycle
2	3.14 [0.17]	30 from horizontal	Temporal-parietal	None
2	1.58 [0.03]	0 from vertical	Crown	Bicycle
2	1.61 [0.02]	0 from vertical	Crown	None

RESULTS

ANALYSIS OF HEAD LOADS

The use of a bicycle helmet significantly reduced the head linear accelerations and HIC for all of the impact orientations (Fig 3a and b) compared with no helmet. HIC values were up to 70% lower for helmeted impacts; 135 (helmeted) compared with 457 (unhelmeted). Compared to unhelmeted cases, helmeted impacts resulted in less angular acceleration; especially in lateral impacts where a 35% decrease was observed. There was no statistically significant effect of the helmet on angular acceleration for front and crown impacts (Fig. 3c). Head rotation was found to be slightly greater in helmeted impacts compared with unhelmeted cases (Fig. 3d).

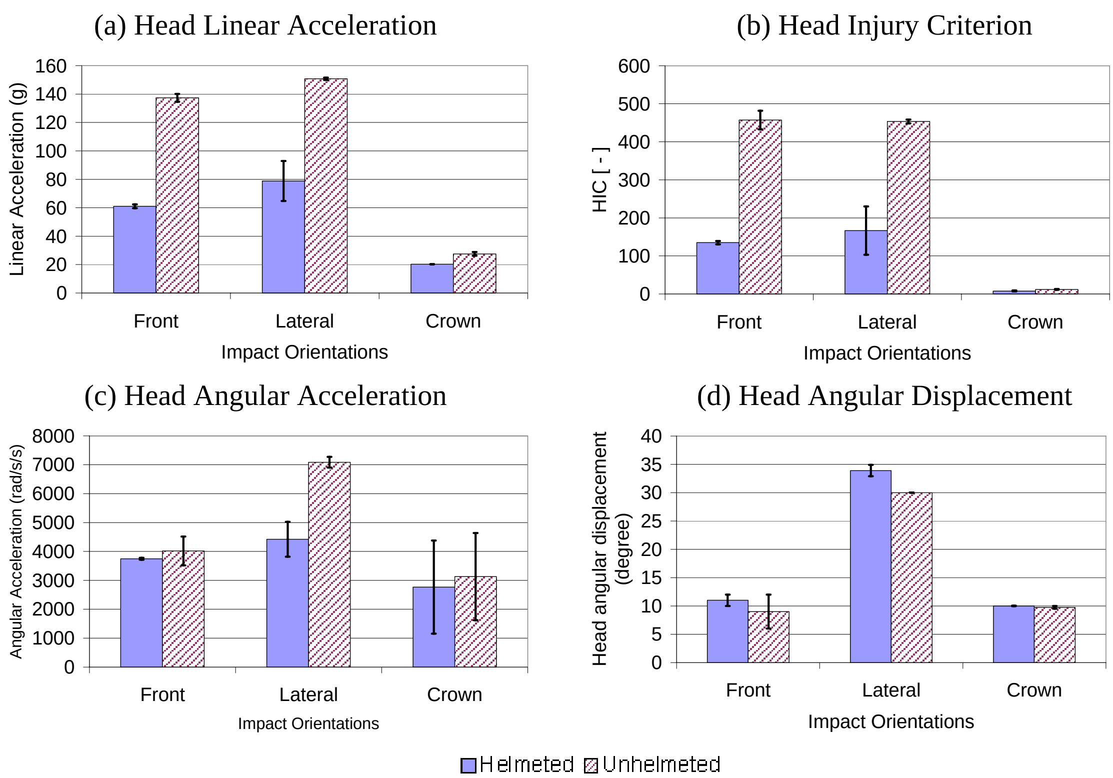

Fig. 3 Head responses versus impact orientations: a) Peak linear accelerations, b) Head injury criterion, c) Peak angular accelerations, and d) Angular displacement. Lateral and frontal impacts at approx. 2.48 to 2.65 m/s and crown at 1.05-1.29 m/s. (Mean ± one standard deviation)

ANALYSIS OF NECK LOADS

There were only small differences in neck loads between the helmeted and unhelmeted impacts. For all three impact orientations, when the head was constrained on the impact surface, the translation movement of the neck-torso with respect to the constrained head caused high shear forces at the upper cervical spine junction (Fig. 4a). Besides shear forces, the neck column was also subjected to a variety of bending loads in association with axial loads, i.e. compression and tension. In the lateral and crown impacts, the mean upper neck bending moment was reduced in the presence of a helmet (Fig. 4b). Large neck compression forces were observed up to 4.6 kN in crown impacts. A small tensile load was observed because the neck was stretched when the head or helmeted head slid across the anvil surface (Fig. 4c and d). There were no significant differences in axial and bending loads associated with the use of a helmet in crown impacts.

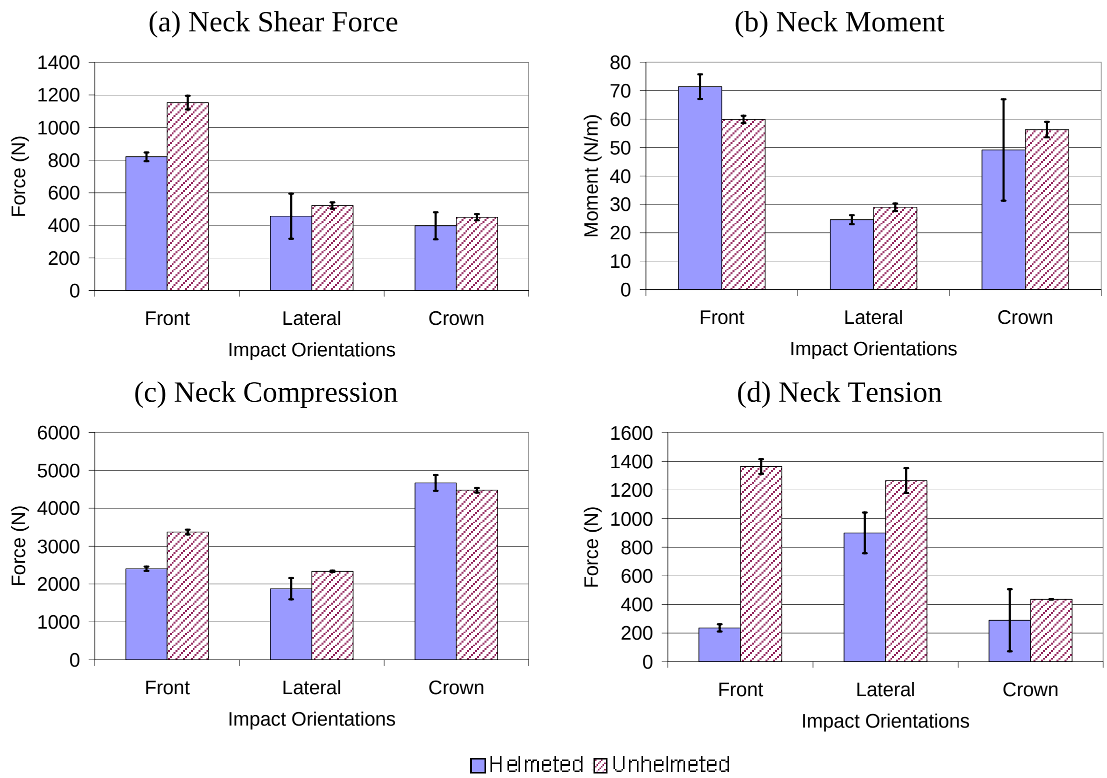

Fig. 4 Peak neck loads in relation to impact orientations: a) Neck shear forces, b) Neck bending moments, c) Neck compression forces, and d) Neck tension forces. Lateral and frontal impacts at approx. 2.48 to 2.65 m/s and crown at 1.05-1.29 m/s. (Mean ± one standard deviation)

Fig. 5 shows that compression combined with flexion of the cervical spine was the most prominent injury mechanism in the simulated dummy drop test. A helmet provided some protection for the upper neck from the compression/flexion loading during both the frontal and temporo-parietal impacts. Tension/extension was significantly higher for helmeted impacts compared to non-helmeted impact. The helmet did not reduce neck loads during a crown impact.

DISCUSSION

Laboratory investigations have shown that the bicycle helmet appeared to be most effective in reducing peak linear accelerations and hence HIC values in all the impact orientations. This reflects the helmet's primary function and its assessment. There was a significant reduction in head angular acceleration associated with a helmet in lateral head impacts, but not frontal and crown impacts. This shows that at worst helmets do not exacerbate angular head acceleration and may in fact reduce it.

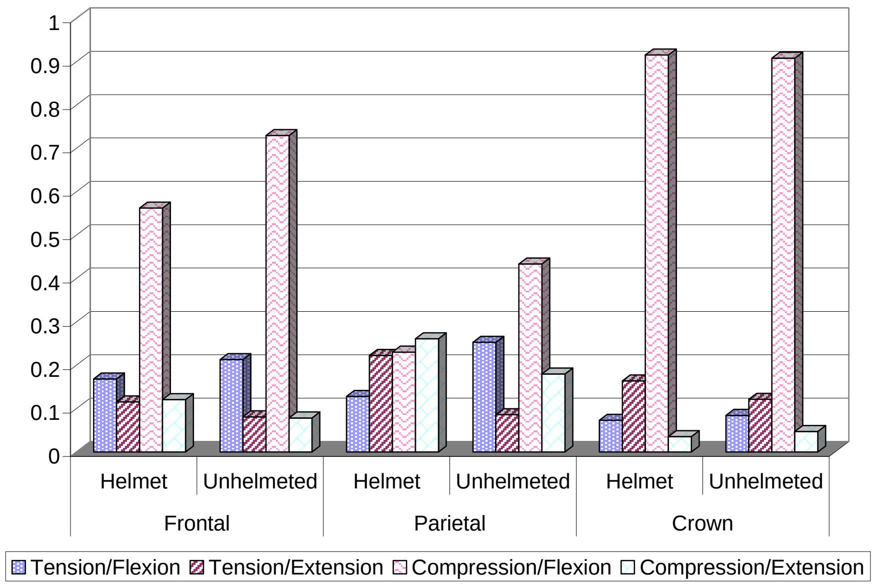

Fig. 5 A comparison of neck injury criteria (N_{ij}) for helmeted and unhelmeted impacts

Compression/Flexion was found to be the dominant mechanism of upper cervical spine loading in all impact orientations. It appears that when the helmeted- or unhelmeted-head impacted the anvil, it stopped moving, while the torso continued to push down through the cervical spine against the head. This is reflected by the upper neck compression force measurements. At the same time the head or the helmet acts as a fulcrum which induces a flexion bending moment on the upper neck. Wearing a helmet can reduce the compression force at the upper neck, but the magnitude may still be high.

The results of this study indicated that another possible injury mechanism is the relative displacement of the head with respect to the neck-torso. The relative displacement caused high shear forces applied to the upper cervical spine or occipital condyle region. The differences of shear forces measured between the helmeted and unhelmeted were, however, small and not significant. The helmet appears to offer little protection against this type of loading.

There were some methodological limitations. Due to restrictions in equipment capabilities, this test series was limited in impact speed, energy, body impact angle, and impact type. The Hybrid III was not designed specifically for inverted impacts, and its neck was found to be significantly stiffer than human cadaver tissue (Viano, and Davidsson, 2002).

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