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Advanced Cycle Helmet Testing Protocols: Effects of Headform Type on Cycle Helmet Safety Performance during Oblique Impacts

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1 Introduction

Cycling is increasing in popularity, both as a mode of transport and as a recreational activity, with around 30% of the 28 European Union Member States' (EU28) population and over 20% of the United States' (US) population reported to cycle each year (TNS Opinion & Social, 2013;Statista, 2017). Cyclists are a particularly vulnerable road user group, however, with 2,131 cyclist fatalities reported across the EU28 and 729 cyclist fatalities reported on US roads during 2014 alone (ERSO, 2017;NHTSA, 2017).

Traumatic brain injuries (TBIs) are associated with around three-quarters of cyclist fatalities and one-third of cyclist hospital admissions (Thompson *et al.*, 2000;Macpherson A, 2008;Olivier and Creighton, 2016). Cycle helmets are a form of personal protective equipment that aim to mitigate the severity of TBIs through managing the impact energies transferred to the head (Hynd *et al.*, 2009). It has widely been recognised that, despite being a critical item of personal protective equipment, the impact safety performance of cycle helmets varies considerably between models (Stigson and Kullgren, 2015;DeMarco *et al.*, 2016;Stigson, 2017). Although current international cycle helmet certification standards establish minimum requirements for evaluating impact safety performance, no independent and freely available information is provided to support cyclists at the point of sale with assessing the differences in safety performance between helmets (BSI, 2012).

Whilst current cycle helmet certification standards assess impact safety performance during linear helmeted headform drop tests, no standard assesses the rotational impact safety performance of helmets during oblique impacts against angled surfaces. Oblique impacts are widely recognised as being more representative of real-world cyclist falls and collisions, as the majority of cyclist falls and collisions include an element of translational motion, whilst the rotational velocities and accelerations transferred to the head during oblique impacts have been found to be highly associated with diffuse brain injuries (Bourdet *et al.*, 2012;Peng *et al.*, 2014). With cyclists typically travelling at speeds between 4.2-43.2 km/h (mean 18.4 km/h), a significant proportion of cyclist collisions and falls include a tangential velocity vector (Boufous *et al.*, 2018). Thus, it remains important to assess the safety performance of helmets when tangentially loaded.

Furthermore, current standards use rigid headforms to assess if minimum protective requirements have been met, with these headforms regarded as being unbiofidelic in their design (Willinger *et al.*, 2015). The Hybrid III headform range, originally designed for automotive crash testing, has therefore been suggested for advanced testing protocols, as it specifies a more biofidelic headform with representative head circumferences, inertial properties and scalp mechanical properties. The influence of the Hybrid III headform on the assessment of injury risk has, however, only been established for linear impacts, whilst its influence on the rotational kinematics of the headform during oblique impacts is still yet to be established (Stuart *et al.*, 2013).

This novel research study therefore aims to quantify the differences in head injury risk between the EN 960:2006 (here on: EN 960) specified and Hybrid III headforms during oblique impacts against angled anvils for two different helmet models. This research will guide future advanced cycle helmet testing protocols by identifying any differences in headform kinematics, investigating whether these differences remain consistent across both cycle helmet models and providing recommendations as to which may be the preferred headform for future advanced testing and assessment protocols.

2 Methods

No ethical approval was required for this experimental study, as no human subjects were recruited for participation.

An EN 1078:2012+A1:2012 (here on: EN 1078) compliant rail guided headform drop test assembly, with a modified “horseshoe” drop carriage design, was used for all drop tests (BSI, 2012). The horseshoe drop carriage was modified to allow the helmeted headform to deflect laterally during drop test impacts against the angled anvil without interference from the drop carriage restraint ring. Helmeted headform drop tests used either a full, EN 960 compliant, 575 mm circumference magnesium headform (4.82 kg combined mass for both the headform and accelerometer array) or a 50th percentile Hybrid III headform (4.54 kg combined mass for headform and accelerometer array) (First Technology Safety Solutions, MI, USA) (BSI., 2006). Helmeted headforms were dropped onto a flat steel anvil angled at 45° to the horizontal plane, with fresh 80 g.m^{-2} sandpaper attached securely to the anvil face for each test.

Four helmet positioning angles were used in this study to impact the crown, frontal, occipital and left temporal regions of the helmet (Figure 1). The crown impact region was specified such that the basic plane was located parallel to the anvil face and the frontal aspect of the helmet was facing the top edge of the anvil, the frontal impact region was specified such that the basic plane was located at 60° to the anvil face and the frontal aspect of the helmet was facing towards the anvil, the occipital impact region was specified such that the basic plane was located at 60° to the anvil face and the frontal aspect of the helmet was facing away from the anvil and, finally, the left temporal impact region was specified such that the left side of the helmet struck the anvil first and the coronal plane was perpendicular to, in addition to the basic plane being angled at 60° to, the anvil face.

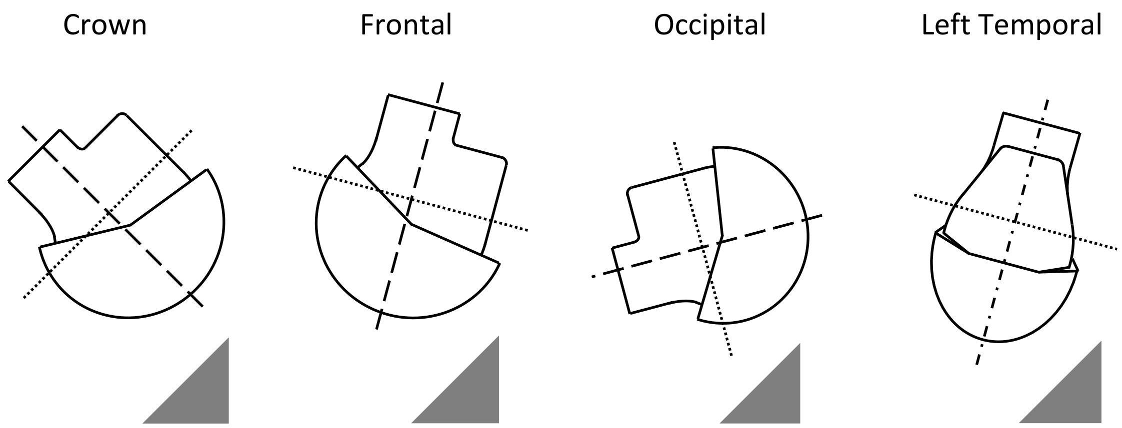

Figure 1: Schematic of helmet positioning angles for impacts to the crown, frontal, occipital and left temporal regions of the helmet

Two different cycle helmet models were used for testing within this research; a size medium (54-59 cm) Trax Mistral Bike Helmet (Model 1) and a ‘universal fit’ (54-61 cm) Bell Draft MIPS Helmet 2016 (Model 2). Twenty-four helmets for each model were tested in this study, with all helmets tested in the condition they were offered for sale, including shell apertures, accessory attachments and comfort padding, and no pre-conditioning performed.

Each helmeted headform was impacted once from a drop height of 3 m (representing cyclist falls from a head height of 1.5 m, occurring whilst cycling at approximately 20 km/h). Three repeat tests were performed for each helmet model, impact location and headform. To perform these tests, each helmet was mounted and securely fastened to the headform through its restraint system, before positioning the helmeted headform. The drop test assembly was then raised to the drop height, before being dropped onto the angled anvil.

The linear accelerations experienced at the centre of gravity of the headform were recorded via three uniaxial accelerometers (9264B, Piezoresistive Accelerometer, Endevco Meggitt, CA, USA), whilst the rotational velocities of the headform were recorded by three uniaxial angular rate sensors (ARS PRO-1500, Diversified Technical Systems (DTS), CA, USA). All instrument data channels were sampled at a rate of 20,000 Hz, before being zeroed and filtered based on ISO 6487 recommendations. Data capture was synchronised using a contact trigger.

Results presented for each test include the peak resultant linear accelerations, angular velocities and angular accelerations of the helmeted headform. The mean differences in safety performance between the headforms used are compared to assess the influence of the headform on each outcome. Results were statistically compared using two-tailed independent samples Student t-tests, with statistical significance considered at $p<0.05$ for all tests.

Finally, these results compare the safety performance of the helmeted headforms against a combination of current legislative performance criteria and published head injury thresholds. When considering linear head accelerations, Newman (1980) established a scale relating linear acceleration thresholds to Abbreviated Injury Scale (AIS) scores (Newman, 1980). This concluded that peak linear head accelerations of >250 *g* are associated with an AIS5+ head injury severity, whilst peak accelerations of >100 *g* correlate with an AIS2+ severity. When considering rotational head velocities and accelerations, *in-vivo* American Football data has previously been used to estimate 50% probability injury thresholds for AIS2+ concussions (Rowson *et al.*, 2012). This research concluded that peak rotational velocities of 28.3 $rad \cdot s^{-1}$ and peak rotational accelerations of 6,383 $rad \cdot s^{-2}$ were associated with a 50% probability of a concussive injury.

3 Results

The peak linear accelerations (Figure 2), rotational velocities (Figure 3) and rotational accelerations (Figure 4) experienced by each headform are illustrated for each helmet model and impact location, alongside key legislative performance criteria and published head injury criteria thresholds.

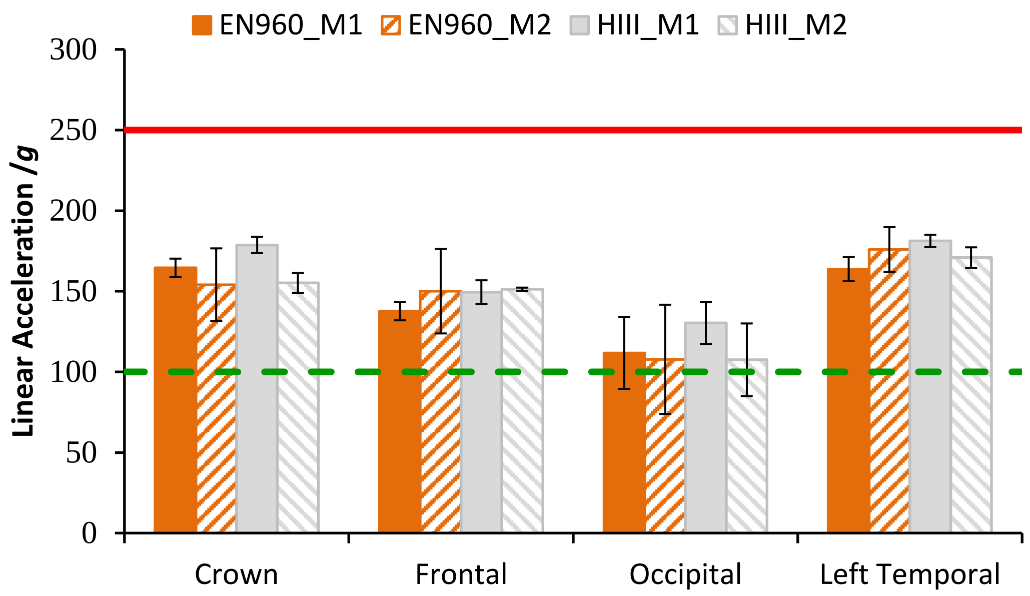

Figure 2: Peak linear accelerations experienced by the EN 960 and Hybrid III (HIII) headforms when tested using two different helmet models (M1, M2) at four different impact locations (crown, frontal occipital and left temporal regions). Data are presented as mean values with 95% confidence intervals. Peak linear acceleration thresholds include the 250 g EN 1078 pass/fail criteria (solid line) and 100 g AIS2+ injury criteria (dashed line) (BSI, 2012;Newman, 1980).

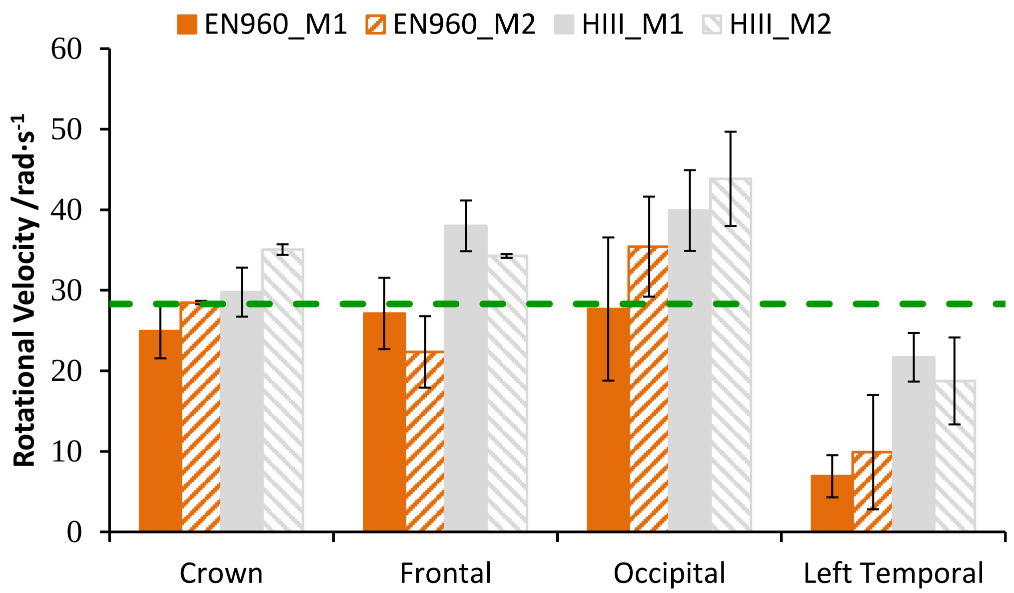

Figure 3: Peak rotational velocities experienced by the EN 960 and Hybrid III (HIII) headforms when tested using two different helmet models (M1, M2) at four different impact locations (crown, frontal occipital and left temporal regions). Data are presented as mean values with 95% confidence intervals. Peak rotational velocity threshold includes the 28.3 rad·s^{-1} criteria representing the 50% probability of a concussive injury (dashed line).

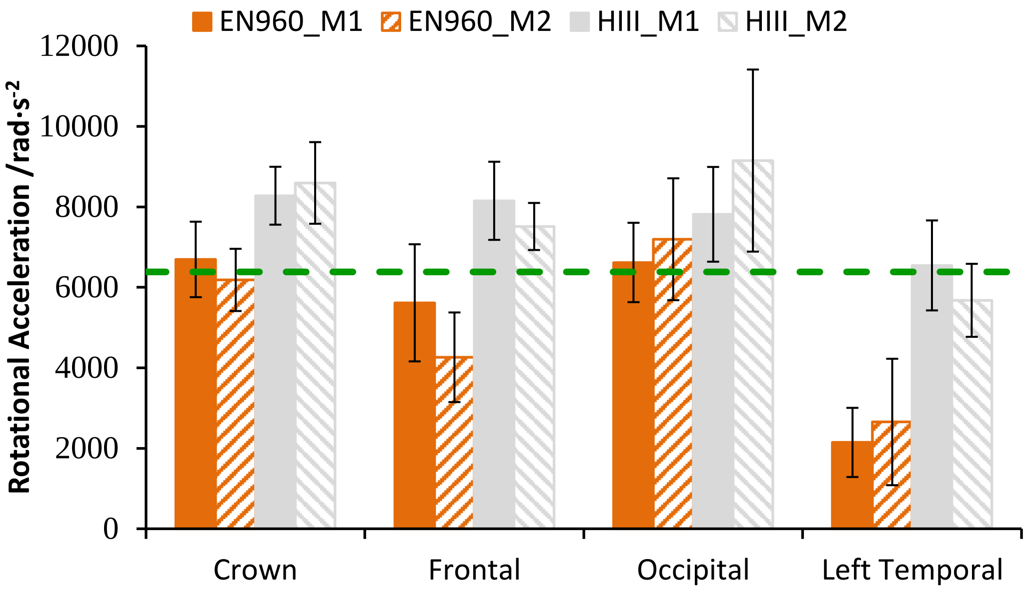

Figure 4: Peak rotational accelerations experienced by the EN 960 and Hybrid III (HIII) headforms when tested using two different helmet models (M1, M2) at four different impact locations (crown, frontal occipital and left temporal regions). Data are presented as mean values with 95% confidence intervals. Peak rotational acceleration threshold includes the 6383 rad·s^{-2} criteria representing the 50% probability of a concussive injury (dashed line).

When considering the linear impact safety performance of the cycle helmets during oblique tests, no headform, helmet model or impact location combination was observed to exceed the 250 *g* performance criterion specified by EN 1078 for linear accelerations. The 100 *g* AIS2+ linear acceleration injury criterion was, however, exceeded during almost all impact tests, with only a single impact to the occipital region resulting in a linear acceleration value of <100 *g*.

For the rotational velocities and accelerations experienced by the headforms during the oblique tests, it is clear that a large proportion of the helmeted headform drop tests exceeded the 50% probability of AIS2+ concussion safety performance thresholds. In total 25 (48%) helmet drop tests exceeded the 28.3 rad·s^{-1} rotational velocity injury threshold and 28 (54%) helmet drop tests exceeded the 6,383 rad·s^{-2} rotational acceleration injury threshold, with 22 (42%) tests exceeding both injury thresholds.

The differences in safety performance metrics between the Hybrid III and EN 960 headforms for both helmet models are shown in Table 1. It can be seen that there was a significant increase in the rotational velocity and acceleration experienced by the Hybrid III headform when compared to the EN 960 headform, regardless of the helmet model used and impact location. A significant increase in linear headform accelerations was observed across all impact locations for Model 1 only, whilst for Model 2 no significant difference was observed between the linear accelerations experienced by the headforms.

Table 1: Difference in safety performance metrics between the Hybrid III (HIII) and EN 960 headforms for the Trax Mistral Bike Helmet (Model 1) and Bell Draft MIPS Helmet 2016 (Model 2) helmets

Impact Location		Mean Difference (HIII less EN 960)*					
		Linear Acceleration /*g*		Rotational Velocity /rad·s^{-1}		Rotational Acceleration /rad·s^{-2}	
		Diff.	*p* value	Diff.	*p* value	Diff.	*p* value
Model 1	Crown	14 [9, 19]	***.001***	5 [2, 8]	***.010***	1583 [820, 2346]	***.005***
	Frontal	12 [6, 18]	***.006***	11 [7, 14]	***.001***	2536 [1407, 3665]	***.003***
	Occipital	19 [2, 35]	***.037***	12 [6, 19]	***.007***	1197 [206, 2188]	***.029***
	Left Temporal	17 [9, 26]	***.003***	15 [11, 18]	***<.001***	4399 [3288, 5511]	***<.001***
Model 2	Crown	1 [-14, 16]	.854	7 [6, 7]	***<.001***	2410 [1587, 3233]	***.001***
	Frontal	1 [-16, 18]	.865	12 [9, 15]	***<.001***	3250 [2438, 4062]	***<.001***
	Occipital	0 [-27, 26]	.976	8 [3, 14]	***.013***	1955 [198, 3712]	***.037***
	Left Temporal	-5 [-22, 11]	.479	9 [0, 17]	***.046***	3024 [1139, 4908]	***.008***

* Differences provided as mean difference [95% confidence interval]. P-values calculated using two-tailed independent samples Student's t-tests.

4 Discussion

This research is the first to investigate the differences in head injury risks between the EN 960 and Hybrid III headforms during oblique impacts against angled anvils. The results from this research demonstrated that greater peak rotational velocities and accelerations are experienced by Hybrid III headforms during standardised oblique impacts, when compared to those experienced by EN 960 headforms, regardless of helmet model or impact location. The results further showed that, for helmet Model 1, greater peak linear accelerations were experienced by the Hybrid III headform during oblique impacts. Finally, this research established that, for all impact conditions modelled by this study, almost every helmeted headform impact exceeded the linear acceleration AIS2+ head injury criterion, whilst approximately 50% of impacts exceeded the rotational velocity and acceleration thresholds for AIS2+ concussions.

The research methods adopted by this study were limited by a number of necessary assumptions and simplifications. Whilst the biomechanical response of the headform was the key variable investigated throughout this research, the response of the headform may still not accurately represent the response of the head during impact due to the lack of anchorage to a flexible neck and the rigidity of the skull structures (Ghajari *et al.*, 2013). Although the injury thresholds used for these results are founded on the best available evidence base, there has been international debate for many years over the use of appropriate head/brain injury criteria. Whilst kinematic injury criteria and finite element analysis (FEA) approaches are both seen as fundamental to the field, this study specifically compares the results against a combination of kinematic legislative performance criteria and published head injury thresholds (Yoganandan *et al.*, 2014;BSI, 2012;Newman, 1980;Rowson *et al.*, 2012). Finally, it is recognised that the impact conditions modelled by this study do not represent all collision or fall characteristics that may be experienced by cyclists. Cyclists may therefore experience collisions that occur at different speeds, whilst wearing different helmet models and that impact a different part of the helmet, with all of these likely to affect the response during impact.

Differences between the kinematics of the EN 960 and Hybrid III headforms during oblique impacts were observed in this research, with these differences most apparent for the peak rotational velocities and accelerations. The use of the Hybrid III headform during the oblique impact tests resulted in greater peak rotational velocities and accelerations, regardless of helmet model or impact location. These significant differences are primarily due to the differences in the physical and material characteristics of the Hybrid III headform, when compared to the EN 960 headform. Whilst the rotational inertia properties of the EN 960 headform are not controlled within current standards, and so may vary widely, the rotational inertia properties of the Hybrid III headform are closely specified. Furthermore, the Hybrid III headform implements a headform skin, which is highly nonlinear and viscoelastic in its mechanical response, whilst the EN 960 headform is typically cast from magnesium and so is typically considered a rigid body (Wood *et al.*, 2010). Finally, the combined mass of the EN 960 headform and accelerometer array was approximately 0.28 kg heavier than the combined mass of the Hybrid III headform and accelerometer array.

Importantly, this study also finds that, for the range of impact conditions modelled by this study, all helmeted headform impacts exceeded at least one published criteria for AIS2+ head injuries. Linear accelerations were observed to exceed the 100 *g* injury criterion in all but one test, whilst the rotational velocities and accelerations of the headform exceeded the 28.3 rad·s^{-1} and 6,383 rad·s^{-2} injury criteria in approximately 50% of tests. As the helmet drop tests were performed against an anvil angle and from a drop height to best represent a typical cyclist fall occurring whilst travelling at 20 km/h, it may be hypothesised that these headform kinematics reflect those experienced by the head during such collisions.

Several studies have assessed the biomechanics of head injury risks during oblique helmeted headform drop tests which may be compared to the results established by this study. Mills and Gilchrist (2008) investigated the linear accelerations and rotational accelerations experienced during oblique helmeted headform drop test impacts (3.6 m·s^{-1} tangential and 4.5 m·s^{-1} normal velocities) to the frontal and lateral aspects of several cycle helmets (Mills and Gilchrist, 2008). These impacts resulted in peak linear accelerations of 105-117 *g* and 109-129 *g* and peak rotational accelerations of 1,000-1,500 rad·s^{-2} and 4,800-7,500 rad·s^{-2} for each impact location. This was supported by McIntosh et al. (2013), who found that, during drop tests to the lateral and occipital aspect of the helmet using 1.0 m and 1.5 m drop heights and impact plate tangential speeds of 4.2 m·s^{-1} and 6.9 m·s^{-1} (the most similar parameters to those used by this study), peak linear accelerations ranged from 100-149 *g*, whilst peak rotational accelerations ranged from 8,060-12,146 rad·s^{-2} (McIntosh *et al.*, 2013).

In comparison, this research found that, across all oblique drop tests performed in this research, peak linear headform accelerations ranged between 97-191 *g*, whilst peak rotational accelerations ranged from 1,427-10,195 rad·s^{-2}. These results are, broadly speaking, similar to those from the literature, although there are some differences in outcomes. These differences exist due to the differences in experimental procedures used, including the helmet models used, the locations impacted and the impact velocity vectors (with helmets in this study impacting the anvil at velocities 5.42 $m \cdot s^{-1}$ tangential and 5.42 $m \cdot s^{-1}$ normal to the anvil face). These differences were perhaps most apparent for the left temporal region, where the linear accelerations were greater and rotational accelerations lower than those found across the literature. This implies that the headforms in this study were not subject to the same level of translational forces than those across the literature, with the reasons for this being due to any combination of the previously mentioned differences in experimental procedures.

This research quantifies how the assessment of cycle helmet safety performance for oblique impacts may vary in outcome according to the headform implemented for the test procedure. With future test and assessment protocols proposing the use of oblique impacts to determine the rotational impact safety performance of cycle helmets, this study established the existence of clear differences between the kinematic responses of the two most popular headform options (the EN 960 and Hybrid III headforms) during oblique impacts (Willinger *et al.*, 2015). Due to the consensus expert opinion that the Hybrid III headform is more biofidelic in its design and the importance of using the most biofidelic headform during drop testing, it is recommended that a 50th percentile Hybrid III headform is considered further for advanced cycle helmet test protocols. The more biofidelic helmeted headform response, if found to be repeatable, could be a key modification to the array of tests to be used for differentiating between the protective qualities of different helmet models.

By impacting helmeted Hybrid III headforms onto an angled anvil from representative drop heights, this study also establishes a method for testing and assessing cycle helmet safety performance during oblique impacts. The choice of anvil angle (45°) and drop height (3 m) is based on current estimates of average cyclist speed (approximately 20 km/h) and an assumption that the cyclist head will strike the ground from a height of 1.5 m (as assumed by current EN 1078 testing standards) (Boufous *et al.*, 2018;BSI, 2012). Although other collision and fall characteristics may be represented using different anvil angles and drop heights, the combinations used in this study are likely to best represent the majority of real-world collisions and falls. Such an approach may therefore be adopted by future advanced cycle helmet testing protocols which implement oblique impact tests to measure the rotational impact safety performance of the cycle helmet. In a similar fashion, should future standards wish to assess cycle helmet safety performance at different impact energies (i.e. representing different cyclist speeds), then the effects of alternative anvil angle and drop height combinations should be further explored.

Finally, comparisons against cyclist collision data are required to evaluate the relative importance of each safety performance metric within these proposed oblique impact test and assessment protocols and with respect to other established linear impact test and assessment protocols. Weightings should be developed to ensure that the outcomes of each test or safety performance metric are given a proportional weighting that is based on the relative real-world importance of each impact characteristic.

5 References

Boufous S, Hatfield J and Grzebieta R (2018). The impact of environmental factors on cycling speed on shared paths. *Accident Analysis & Prevention., 110*, 171-176.

Bourdet N, Deck C, Carreira R and Willinger R (2012). Head impact conditions in the case of cyclist falls. Proceedings of the Institution of Mechanical Engineers. *Journal of Sports Engineering and Technology., 226*(3-4), 282-289.

BSI (2012). *BS EN 1078:2012+A1:2012: Helmets for pedal cyclists and for users of skateboards and roller skates.* British Standards Institution (BSI), London.

BSI. (2006). *BS EN 960:2006: Headforms for use in the testing of protective helmets.* British Standards Institution (BSI), London.

DeMarco A, Chimich D, Gardiner J and Siegmund G (2016). The impact response of traditional and BMX-style bicycle helmets at different impact severitieS. *Accident Analysis & Prevention, 92*, 175-183.

ERSO (2017). *Annual Accident Report 2016*. European Road Safety Observatory (ERSO).

Ghajari M, Peldschus S, Galvanetto U and Iannucci L (2013). Effects of the presence of the body in helmet oblique impacts. *Accid Anal Prev, 50*, 263-271.

Hynd D, Cuerden R, Reid S and Adams S (2009). *The Potential for Cycle Helmets to Prevent Injury a Review of the Evidence,* (Published Project Report PPR 446). Transport Research Laboratory, London.

Macpherson A SA (2008). Cochrane review: Bicycle helmet legislation for the uptake of helmet use and prevention of head injuries. *Evidence Based Child Health: A Cochrane Review Journal, 3*(1), 16-32.

McIntosh A, Lai A and Schilter E (2013). Bicycle helmets: head impact dynamics in helmeted and unhelmeted oblique impact tests. *Traffic Inj Prev., 14*(5), 501-508.

Mills N and Gilchrist A (2008). Oblique impact testing of bicycle helmets. *International Journal of Impact Engineering., 35*(9), 1075-1086.

Newman J (1980). Head injury criteria in automotive crash testing. *SAE Technical Paper 0148-7191*.

Newman J (1980). Head injury criteria in automotive crash testing. *SAE Technical Paper;, 0148-7191*.

NHTSA (2017). *Traffic Safety Facts 2015 Data: Bicyclists and Other Cyclists*. National Highway Traffic Safety Administration (NHTSA).

Olivier J and Creighton P (2016). Bicycle injuries and helmet use: a systematic review and meta-analysis. *International journal of epidemiology., 46*(1), 278-292.

Peng Y, Yang J, Deck C, Otte D and Willinger R (2014). Development of head injury risk functions based on real-world accident reconstruction. *International Journal of Crashworthiness., 19*(2), 105-114.

Rowson S, Duma S and Beckwith J (2012). Rotational Head Kinematics in Football Impacts: An Injury Risk Function for Concussion. *Annals of Biomedical Engineering, 40*(1), 1-13.

Statista (2017). *Number of cyclists/bike riders within the last 12 months in the United States from spring 2008 to spring 2017 (in millions)*.

Stigson H and Kullgren A (2015). *Folksam's Bicycle Helmet Test 2015*.

Stigson H (2017). *Bicycle Helmets 2017 Tested by Folksam*. Folksam: Stokholm.

Stuart C, Cripton P, Dressler D, Dennison C and Richards D (2013). *Bicycle Helmet Efficacy Using Hybrid III and Magnesium Headforms.*

Thompson D, Rivara F and Thompson R (2000). Helmets for preventing head and facial injuries in bicyclists. *Nursing times, 97*(43), 41.

TNS Opinion & Social (2013). *Special Eurobarometer 406: Attitudes of Europeans towards urban mobility*. Directorate-General for Mobility and Transport (DG MOVE).

Willinger R, Halldin P, Bogerd C, Deck C and Fahlstedt M (2015). *Final report of Working Group 3: Impact engineering. A COST Action TU1101/HOPE collaboration.*

Wood G, Panzer M, Bass C and Myers B (2010). Viscoelastic properties of Hybrid III head skin. *SAE International Journal of Materials and Manufacturing., 3*, 186-193.

Yoganandan N, Nahum A and Melvin J (2014). *Accidental Injury: Biomechanics and Prevention.*, Springer, New York.

Advanced Cycle Helmet Testing Protocols: Effects of Headform Type on Cycle Helmet Safety Performance during Oblique Impacts

Current certification standards establish minimum impact performance requirements for cycle helmets. These standards, however, do not use a biofidelic headform available to assess impact performance and do not assess impact performance for oblique impacts against angled surfaces, which better represent real-world collisions. The effect of these characteristics on head injury risk therefore requires further research to inform future advances in the biofidelity of cycle helmet testing and assessment protocols.

The differences in oblique impact performance between two different helmet models and two different headforms (EN 960 and the more biofidelic Hybrid III) were evaluated for four impact locations (crown, frontal, occipital and temporal regions). Helmets were mounted to each headform and impacted against a 45° angled anvil from a drop height of 3 m at each location.

Increased rotational accelerations and velocities were observed for the Hybrid III headform, when compared to the EN 960 headform, across both helmet models and all impact locations. Increased linear accelerations were observed across all impact locations for one helmet model only.

Advanced cycle helmet testing protocols should consider adopting the Hybrid III as their headform and ensure the effects of impact location are considered when evaluating cycle helmet impact performance.

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