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International Cycling Safety Conference 2017: Cycle Helmet Workshop Report

Cycle helmet safety: Global harmonisation of consumer information rating schemes

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Report details

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1 Introduction

Cycle helmets are a critical item of personal protective equipment that aim to reduce both the occurrence and severity of head injuries by providing adequate head protection during collisions. The safety performance of a cycle helmet is fundamental to protecting cyclists during a fall or collision; however, very little is known about the relative protective qualities of different cycle helmet models.

To address this issue, a number of research institutes have begun to develop cycle helmet testing and assessment programs to rate the relative safety performance of cycle helmets. These institutes are truly international, with the UK, US, Sweden, Germany and France all beginning to develop such schemes. To maximise the positive impact of such schemes, the global harmonisation of these approaches at an early stage may be beneficial.

The Cycle Helmet Safety Workshop aimed to provide an opportunity for global experts to present the latest outcomes of their research, discuss current best practices for testing and assessing cycle helmet safety performance, and provide a forum for debating the global harmonisation of the various approaches currently being researched across the World. The Workshop also involved the creation of a roadmap for achieving a global cycle helmet safety consumer information scheme.

2 Workshop Attendees

Chairperson

David Hynd	Transport Research Laboratory

Organiser

Phil Martin	Transport Research Laboratory

Presenters

Randy Swart	Bicycle Helmet Safety Institute
Helena Stigson	Folksam
Siobhan O'Connell	Transport Research Laboratory
Megan Bland	Virginia Tech
Steve Rowson	Virginia Tech
Remy Willinger	University of Strasbourg

Participants

Ed Becker	Snell
Emily Bliven	Apex Biomedical
Zouzias Dimitris	Lazer
Narelle Haworth	Queensland University of Technology
Silas Klug	Bosch
Dave Krzeminski	University of Southern Mississippi
Valeria La Saponera	A_2_Z
Loretta Moore	City of Davis
Jake Olivier	University of New South Wales
Michelle Tsai	Go Pro
Tony White	Trek
Hong Zhang	Snell

3 Workshop Approach

The Workshop comprised of three presentation sessions followed by an interactive session. Each presenter was required to give a 15 minute presentation, which was followed by five minutes of questions from the Workshop attendees.

The first presentation session involved a keynote presentation by Randy Swart (Executive Director of the Bicycle Helmet Safety Institute and Co-Vice Chair of the ASTM F8.53 helmet standards subcommittee) which outlined suggestions for improving current helmet testing standards. The session focused on the strengths and challenges facing the current cycle helmet testing landscape and the requirements for ensuring such a scheme remains focused on the needs of the consumer. The presentation abstract may be found in Appendix C and the respective presentation slides may be found in Appendix D.

The second session focused on global approaches toward testing the safety performance of cycle helmets and required presenters to comment on current best practices used by each research institute and the effects that these approaches had on outcomes. Three presenters gave presentations during this session including: Helena Stigson (Senior Researcher, Folksam Insurance Group), Siobhan O'Connell (Graduate Researcher, Transport Research Laboratory) and Megan Bland (PhD Student, Virginia Tech). Abstracts for all three presentations may be found in Appendix C, whilst the respective presentation slides may be found in Appendix D. The range of key focus topics, scoped for discussion within the second presentation session, are included in Table 1 below.

Table 1: Key focus topics for second presentation session on global approaches toward testing cycle helmet safety performance

Key Focus Topic	Areas Discussed
Testing Philosophy	Linear impacts, oblique/rotational impacts
Drop Assembly	Free-fall carriage, support arm, wire/rail guided
Headform	ISO, ASTM, NOCSAE, HIII, mass, inertial properties
Neck	None, solid, HIII, attached to guide carriage/support arm, other designs
Measurement Approach	9-axis accelerometer array, 6 DoF sensor (ARS+Accel)
Impact Location	Single/multiple points, single/multiple regions, impacts
Head Orientation	Multiple headform orientations possible per impact location
Anvil Angle	Flat, acute angles, 45°, obtuse angles
Impact Energies	High energies, low energies, normal/tangential energies
Anvil Shape	Flat, kerbstone, hemispherical
Anvil Surface	Planar, sandpaper, concrete, rigid bars
Performance Differentiation	Accurate differentiation between helmet models
Repeatability	Variation between helmet tests performed using the exact same process
Reproducibility	Variation between helmet tests performed between laboratories

The third presentation session focussed on global approaches toward assessing cycle helmet safety performance and centred around the current assessment philosophies used by each research institute and the effects that these approaches had on outcomes. Two presenters gave presentations during this session including: Steve Rowson (Assistant Professor, Virginia Tech) and Rémy Willinger (Professor, Strasbourg University). Abstracts for the presentations may be found in Appendix C, with the respective presentation slides found in Appendix D. The range of key focus topics, scoped for discussion within the third presentation session, are included in Table 2 below.

Table 2: Key focus topics for second presentation session on global approaches toward assessing cycle helmet safety performance

Key Focus Topic	Areas Discussed
Linear Kinematic Injury Criteria	Peak Linear Acceleration (PLA), Head Injury Criterion (HIC), Gadd Severity Index (GSI), Skull Fracture Criterion (SFC)
Rotational Kinematic Injury Criteria	Peak Rotational Acceleration (PRA), Peak Rotational Velocity (PRV), Rotational Injury Criterion (RIC), Brain Rotational Injury Criterion (BrIC), Power Rotational Head Injury Criterion (PRHIC)
Combined Kinematic Injury Criteria	Generalized Acceleration Model for Brain Injury Threshold (GAMBIT), Head Impact Power (HIP), Combined Probability (CP), Principal Component Score (PCS), Brain Injury Threshold Surface (BITS)
Finite Element Analysis Models	Boundary conditions, material properties, validation and verification, differences between models
Finite Element Analysis Injury Criteria	Intracranial Pressure (ICP), Brain von Mises Stress, Strain, Strain Rate
Rating Schemes	Approach, weightings, relevance to accident/injury risk data, costs

The interactive Workshop aimed to develop a three-year plan for achieving an evidence-based, successful and sustainable cycle helmet safety consumer information scheme and provided a platform for discussion around the various benefits/disbenefits surrounding key issues for harmonisation. A group discussion activity (Section 4.1) was used to generate a wall chart illustrating the Workshop participants' vision for what a successful scheme may look like in three years' time. Workshop participants were then split up into three groups to create roadmaps that set out the current state-of-the-art, their three-year vision for a global rating scheme and the steps that would need to be implemented to ensure this three-year vision is achieved (Section 4.2).

4 Outcomes of the Interactive Workshop

4.1 Group Discussion Activity

The group discussion activity was focused on four key topics surrounding the challenges of harmonising the consumer information rating scheme. These included discussions around the specifications for the oblique impact tests, kinematic and finite element injury criteria, the relative weighting of star ratings, and developing a cost-effective scheme. There was also a general discussion about how the future scheme may look and how it could be executed. The outputs of this discussion may be found in Appendix A.

4.2 Creating a Roadmap towards Global Harmonisation

Where are we now?

Current standards are not felt to be stringent enough; cyclists are still suffering serious and fatal head injuries whilst wearing helmets. Several institutions and groups across the World are developing rating schemes for cycle helmets (including Folksam, University of Strasbourg, Virginia Tech, CEN WG11 & TRL) (Figure 1). Each institution, however, tests using different conditions and assesses safety performance using different rating systems. It is important at this time to consider which approaches should be used in any future harmonised scheme in regards to impact conditions, headforms, impact locations, velocities and injury criteria.

Year 0 — Various standards and rating schemes

Send out group of helmets and assess agreement between test methodologies
Compare kinematic responses of setups using bare headform drops onto MEP?

Draft test conditions/protocols
- Input from current research and criteria (α and ω)
- STAR rating criteria best kinematic parameters
- Maybe based on FE accident analysis

Year 1

Raise funding / awareness from insurance + non-profits
Decide on a cost scheme and financial plan for testing
Compare helmets across all standards and rating schemes (raw data)

Agree on impact conditions and injury criteria
What are the scenarios we want to base conditions on? i.e. prevalence of impact scenario vs. stressing differing helmet properties (compression vs. shearing properties)
Establish limit in number of conditions?
Head Trauma database

Year 2 — Actual testing process determined - round robin tests could help spread determination process
What criteria do you have to meet to test the helmets or be eligible to give a 5-star rating?
Build up a collision database

Year 3 — Global agreement on rotational setup/ requirements
Global agreement on evidence based weighted assessment criteria for rating helmet safety

Figure 1: Three year timeline for the roadmap to a globally harmonised cycle helmet safety testing and assessment protocol

Where do we want to be?

Individual roadmaps for each group may be found in Appendix B, with an overall roadmap shown in Figure 1. In three years' time the Workshop participants would like to have agreement on a weighted consumer information scheme. Similarly to NCAP testing, the assessment would be valid for a set number of years with updates made every 3-5 years to continually push the safety of cycle helmets forward. The assessment scheme would produce a 5-star rating through a variety of test conditions based on likelihood of incidents and injuries. The assessment may also include helmet fit and retention tests and randomly generate test point locations to avoid helmets being designed to pass specific tests. In the short term the assessment scheme may be based on injury metrics, with a longer term goal of introducing FEA models. Finally, reproducibility tests should be undertaken via worldwide round robin testing, using consistent testing conditions and ratings between groups.

4.3 Key Outcomes and Recommendations

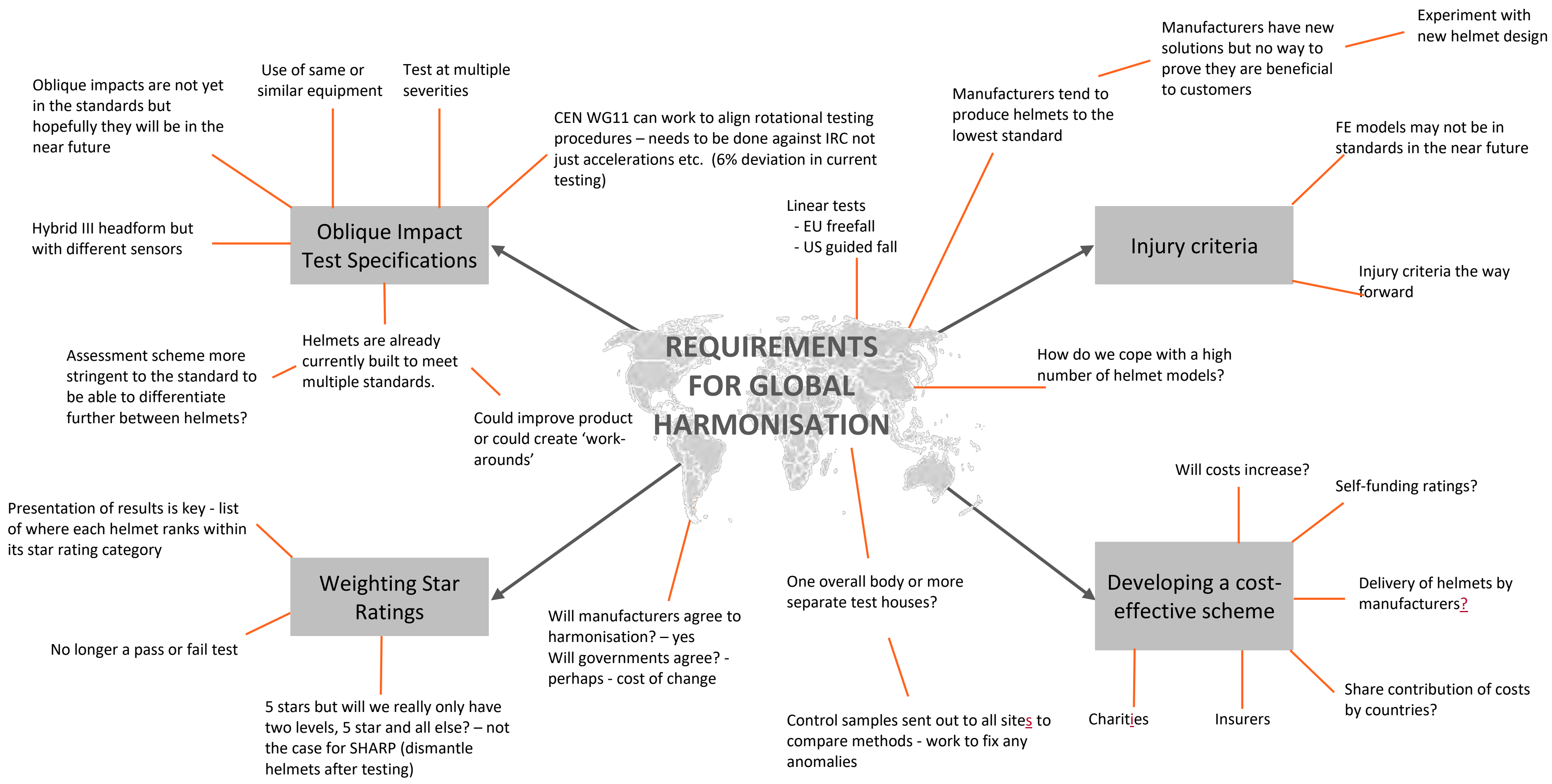
The outcomes of the Workshop resulted in a number of generally agreed conclusions:

- More needs to be done to improve cycle helmet safety over and above that required by current standards.
- Multiple institutions around the world are currently working on rating schemes and have adopted slightly differing test and assessment methods.
- A single rating scheme will be more beneficial to both manufacturers and consumers than multiple ones which may provide conflicting ratings.
- It is important for Institutes to exchange work regularly and discuss successes and challenges to reduce the amount of work each group is carrying out.
- Future funding of joint research programs should be considered
- Oblique testing and assessment protocols are required to help protect against the effects of rotational accelerations.
- Safety performance should be based on the real-world risks of injury occurring at a particular helmet impact location.

The outcomes of the Workshop also highlighted areas where further evidence is required to be able to reach a consensus on the way forward:

- The use of a headform with a neck should be investigated.
- Greater understanding required for the implications of using either kinematic injury criteria or numerical modelling approaches to assess injury risk.
- Greater understanding required for the implications of using combined linear and rotational kinematic injury risk functions for assessing injury risk.
- Debated as to whether each test house should use the same equipment/methods or use relative ratings to provide consistent ratings with differing approaches.
- Should research efforts be split between institutions or combined to carry out the same initial tests and compare results?
- Further research into bicycle collisions is key to ensuring the test methods are representative of real world collision scenarios – in-depth collision investigations of injury mechanisms should therefore be considered in any future research proposals.

Appendix A Group Discussion Activity Outcomes



Appendix B Group Roadmaps

WHERE ARE WE NOW?

- Who is currently making ratings?
  - Folksam, Virginia Tech, UniStra?
- What are the questions to assess?
  - Impact conditions? Head, neck, location, velocity
  - Injury criteria

Agree on impact conditions and injury criteria.

- What are the scenarios we want to base conditions on? i.e. prevalence of impact scenario vs. stressing differing helmet properties (compression vs. shearing properties)
- Establish limit in number of conditions?

Head Trauma database

Send out group of helmets and assess agreement between test methodologies

Compare kinematic responses of setups using bare headform drops onto MEP?

WHERE DO WE WANT TO BE?

- Constant rescaling?
- Set FEA model or injury metric for set number of years
- Same impact setup OR consistent relative ratings between groups

WHERE ARE WE NOW?

- Government standards do not provide consumers with information about differences in helmet safety
- Various institutions testing under different conditions and different rating systems/ metrics of analysis

Year 0

Year 1

- Draft testing conditions/protocols
- Input from current research and criteria (α and ω)
- STAR rating criteria best kinematic parameters
- Maybe based on FE accident analysis
- Raise funding / awareness from insurance + non profits
- Decide on a cost scheme and financial plan for testing

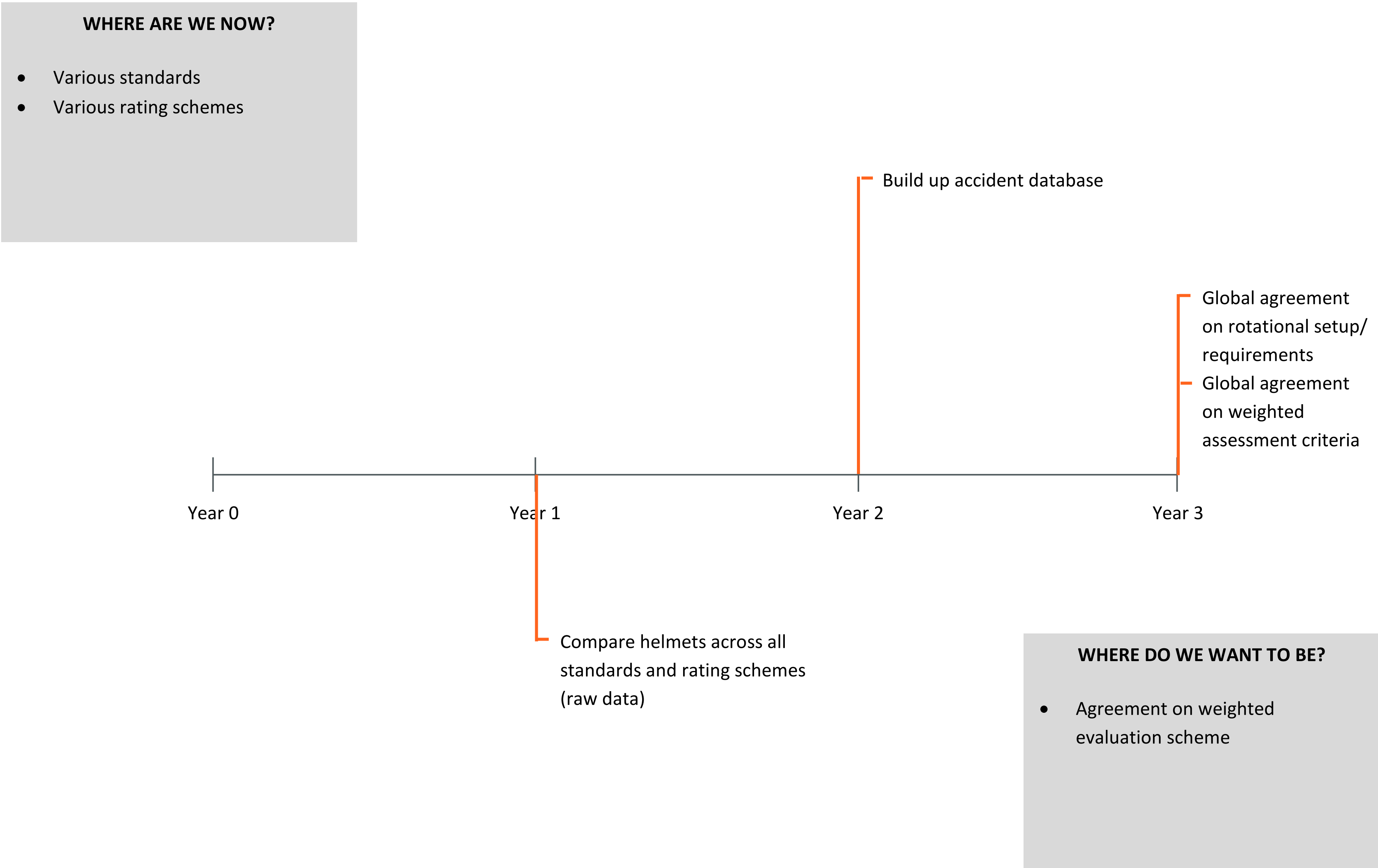
Year 2

- Actual testing process determined.
  - Round robin tests could help to spread determination process
- What criteria do you have to meet in order to test the helmets or be eligible to give a 5-star rating?

Year 3

WHERE DO WE WANT TO BE?

- A work in progress, to be updated
- 5 star system, worldwide + date (updates every 3-5 years)
- Randomly generated test point location?
- A variety of testing conditions based on likelihood of incidents and injuries
- Round robin testing worldwide with the same testing conditions
- Helmet retention and fit testing
- No FE analysis of results unless validated experimentally – eventually use FE analysis


WHERE ARE WE NOW?
Various standards
Various rating schemes
Build up accident database
Global agreement on rotational setup/ requirements
Global agreement on weighted assessment criteria
Year 0
Year 1
Year 2
Year 3
Compare helmets across all standards and rating schemes (raw data)
WHERE DO WE WANT TO BE?
Agreement on weighted evaluation scheme

Appendix C Presentation Abstracts

C.1 Randy Swart (Executive Director, Bicycle Helmet Safety Institute)

Presentation to the Sixth International Conference on Cycling Safety Workshop on global harmonization of consumer information rating schemes

Randy Swart

Executive Director

Bicycle Helmet Safety Institute – helmets.org

INTRODUCTION

I am pleased to present to this Workshop. I regard your work as the most significant opportunity for progress in improving bicycle helmets in the world today.

The traditional standards organizations are not well suited to promoting improvements in helmet performance. They have so much inertia that they are slow to improve their standards, and are frustrated by inability to justify perfect new levels for testing.

Without web presence to explain performance levels, they are stuck with calling out a single level of performance to be printed on the helmet sticker. That level becomes the designers' target. Lawyers and corporate representatives in the room agonize over the legal repercussions of every change and making current models obsolete. No manufacturer can promote a "safer" model for fear of lawsuits involving other models.

There are few sources of test data for consumers. The one consumer magazine here in the US tests only a few models, leaves out the mass-merchant models that most people actually buy, and provides opaque and oversimplified testing reports.

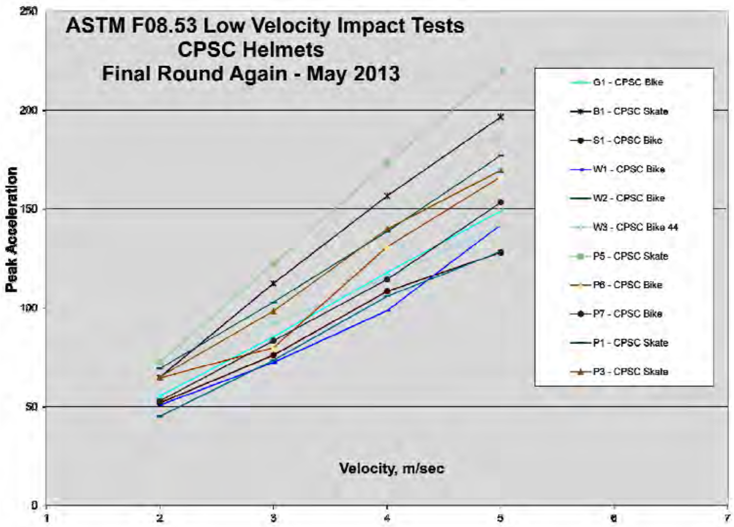
This group can do better. I have a few suggestions for you.

LOW IMPACT TEST CRITERIA

Our ASTM F08.53 helmet subcommittee has been testing helmets in low velocity impacts. As this graph illustrates, there are significant differences in their ability to manage the energy of the impact, although they all met the high impact requirements.

The data points represent the average peak g of four impacts on four helmets on the flat anvil at ambient temperature. The helmets mostly came from the marketplace.

Based on that testing, a group of us tried to add a low velocity drop to ASTM bicycle helmet standards. With current research showing there is no threshold of concussion, we settled arbitrarily on 3.0 m/s (about .5m drop height).



You will note from the graphic above that it would pass the majority of helmets tested but would flunk about a third of them. The best helmets came in at about 60g and the worst were about double that. We were voted down with many different rationales.

I hope that members of this group will consider incorporating a test to verify the performance of bicycle helmets in lower level impacts in their rating schemes. No standard in the world now provides consumers with that rating. Given our current focus on concussion, all of us want to know how our helmet performs in a lower level impact.

HIGH IMPACT PERFORMANCE

There are gradations of high level impact performance that are not being revealed by any current standard in the world. Legal constraints keep manufacturers from advertising that any model outperforms their others. The result is that designers are aiming for the performance required in the standards plus a slim margin to cover production variations. Our testing of cheap and expensive models showed amazingly uniform performance across the spectrum [1]. Another high impact question is what helmet protection will best serve the various types of ebike users.

ROTATIONAL PERFORMANCE

Europe has begun work on a rotational performance standard. In the US we are just beginning to think about one. Relating it to actual injury in the field is difficult.

AREA OF COVERAGE AND TEST LINE

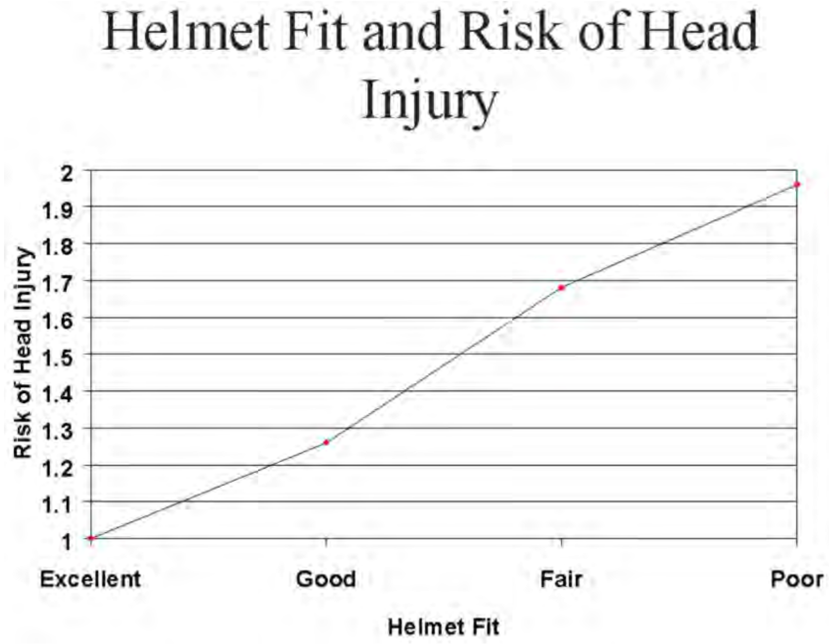
Many current helmets use the shape of the "trail" model to suggest additional rear coverage. In fact, when the helmet is properly positioned at the front, the extra rear coverage is hiked up and disappears. Consumers can evaluate that in a shop but they are not trained to do that. Online buyers have no opportunity to do that. But the area of coverage is critical, since research with damaged helmets shows that many impacts occur on or below the test line [2].

OPTIMAL EXTERIOR SHAPE

We continue to believe that rounder helmets are superior for crashing. Lab tests showed long ago that a helmet that adheres to the impact surface increases the user's neck strain and g's to the brain [3]. There is no standard for that, although the exterior shape is most likely the first line of defense against excess rotational energy in a crash.

FIT

Although fit is perhaps the most important variable for a helmet, no standard in the world has an adequate protocol for testing the ability of a helmet to be fitted correctly. Most have no test for the ability of the helmet to maintain that initial fit over time. And no standard rates the need for undue time and fiddling. This graphic from a Snell report emphasizes the importance of good fit.



Most of the variation occurs because of impacts along the edges, or along the area where the edge of the helmet should have been. If the helmet does not fit well, there may be nothing between the consumer's head and the hard object at the time of impact.

We consumer advocates are looking forward to seeing the results of your research and ratings.

REFERENCES

[1] http://www.helmets.org/testbycost.htm

[2] Circumstances and Severity of Bicycle Injuries, Snell Memorial Foundation, 1996 found at

http://www.smf.org/docs/articles/report#A9

[3] Voigt Hodgson, Wayne State University, Skid Tests on a Select Group of Bicycle Helmets to Determine Their Head-Neck Protective Characteristics found at http://www.helmets.org/hodgstud.htm

Note: We at the ASTM F08.53 Subcommittee are grateful to Dave Thom of Collision and Injury Dynamics/ACT Labs, who has conducted pro bono all of our low level impact testing. Dave is a pioneer in helmet testing and standards, and runs a righteous lab. Web: ci-dynamics.com

C.2 Helena Stigson (Associate Professor, Folksam)

Consumer Testing of Bicycle Helmets

H. Stigson, M Rizzi, A Ydenius, E. Engström, A. Kullgren

Keywords: Angular acceleration, Bicycles, Head injury, Helmets, Oblique impact tests.

INTRODUCTION

Data from real-word crashes show that bicycle helmets are effective reducing injuries. Two out of three head injuries from bicycle accidents could have been avoided if the cyclist had worn a helmet [1]. In the event of more severe head injuries the protective effect is even higher [2]. Reconstructions of real-world bicycle accidents clearly show that a bicycle helmet can decrease the risk for skull fracture and brain injuries [3]. Other reconstructions have shown that oblique impacts are the most common impact scenarios [4].

In the current European certification tests, however, only the energy absorption in a perpendicular impact is evaluated, with the helmet being dropped straight onto a flat anvil and onto a kerbstone anvil. An approved helmet should comply with the 250 g limit [5], a threshold focused on avoiding skull fractures. A peak acceleration of 250 g is associated with a 40% risk of skull fracture [6]. This threshold is thus involved with a significant risk of head injury, even after the helmet has cushioned the impact.

Concussion, or what is known as Mild Traumatic Brain Injury (MTBI), with or without loss of consciousness, occurs in many activities, often as a result of the brain being subjected to rotational forces in the event of either direct or indirect forces against the head [7]. In general, 8% of concussions result in long-term or permanent symptoms, such as memory disorders, headaches and other neurological symptoms, which clearly shows the importance of preventing these injuries. According to Zhang et al [8] concussion with or without loss of consciousness can occur at approximately 60–100 g. Researchers [9][10] have also shown that the brain is much more sensitive to rotational acceleration than to linear acceleration. The risk of concussion or more serious brain injuries, such as diffuse axonal injury (DAI), hematoma or contusion, are not connected to the translational acceleration but rather to the rotational acceleration and the rotational velocity [11-13]. Despite this, translational acceleration is mainly used today to optimize helmets and protective equipment in the automotive industry, for example. The bicycle helmet standards do not include angular acceleration for certification, even though it is known that angular acceleration is the dominant cause of brain injuries [14]. The initial objective of the helmet standards was to prevent life-threatening injuries, but with the knowledge of today it is important to also prevent injuries resulting in long-term consequences. The objective of this study, therefore, was to develop an improved test method that included rotational acceleration in order to evaluate helmets sold on the European market.

METHODS

In total, 17 conventional helmets and one airbag helmet (Hövding 2.0) were selected from the Swedish market. To ensure that a commonly used representative sample was chosen, the range helmets available in bicycle/sports shops and in online shops were all considered. In addition, some helmets with special protective features were selected. Seven of the conventional helmets were equipped with an extra protection, called MIPS (Multi-directional Impact Protection System), which is aimed at lowering rotational acceleration in the event of an oblique impact. One helmet (Smith Forefront) was selected because it is claimed to be extremely light and impact-resistant as the material in the helmet is partly made up of a honeycomb structure. Another helmet (Yakkay) was because since it is sold with a cover as additional equipment. The intention was to evaluate the effect of the different features on the test results. The Hövding 2.0 was selected because it had three to four times better shock absorption than conventional helmets in a previous test [15]. It has a head protector that is inflated during an accident situation and acts as an airbag for the head. However, the Hövding 2.0 had never before been tested for oblique impacts. The test set-up used in the present study corresponds to a proposal from the CEN Working Group's 11 "Rotational test methods" [16][17]. In total, four separate tests were conducted (Table 1). A finite element (FE) model of the brain was used to estimate the risk of brain tissue damage during the three oblique impact tests.

TABLE I
INCLUDED TESTS

TEST	VELOCITY	ANGLE	DESCRIPTION
Shock absorption test	5.6 m/s	0°	The helmet was dropped from a height of 1.5 m to a horizontal surface correlated to the regulation EN 1078.
Oblique impact A. Contact point on the upper part of the helmet.	6.0 m/s	45°	A test that simulates an actual cyclist-vehicle-crash or a single bicycle crash. Rotation around X-axis.
Oblique impact B. Contact point on the side of the helmet.	6.0 m/s	45°	A test that simulates an actual cyclist-vehicle-crash or a single bicycle crash. Rotation around Y-axis.
Oblique impact C. Contact point on the side of the helmet.	6.0 m/s	45°	A test that simulates an actual cyclist-vehicle-crash or a single bicycle crash. Rotation around Z-axis.
Computer simulations	-	-	As input into the FE model, the measured rotational and translational accelerations from the HIII head in the three tests above were used.

Shock absorption test

The helmet was dropped from a height of 1.5 m to a horizontal surface according to the European standard which sets a maximum acceleration of 250 g [5] (Fig. 1). The shock absorption test is included in the test standard for helmets (EN 1078), in contrast to the oblique tests. The ISO head form was used and the test was performed with an impact speed of 5.42m/s. The helmets were tested in a temperature of 18°C. The test was performed by Research Institutes of Sweden (RISE) which is accredited for testing and certification in accordance with the bicycle helmet standard EN 1078.

Fig. 1. The method used in shock absorption test.

Oblique Tests

In three oblique tests the ISO headform was replaced by the Hybrid III 50th percentile Male Dummy head. The reason for this choice was that the Hybrid III 50th percentile male dummy head has much more realistic inertia properties and it allows for measurements of the linear and rotational velocity and acceleration. A system of nine accelerometers was mounted inside the Hybrid III test head according to the 3-2-2-2 method described by Padgaonkar et al. [18]. Using this method it is possible to measure the linear accelerations in all directions and the rotational accelerations around all the three axis X, Y and Z, as illustrated in Figure 2 and Figure 3. The accelerometer samples were obtained at a frequency of 20 kHz and all the collected data were filtered using an IOtechDBK4 12-pole Butterworth low-pass filter. This is further described by Aare and Halldin [19]. The helmeted head was dropped against a 45° inclined anvil with friction similar to asphalt (grinding paper Bosch quality 40). The impact speed was 6.0m/s. The Hybrid III dummy head was used without an attached neck.

The impact to the side of the helmet was located at parietal level. The impact was applied in the frontal plane, resulting in rotation around the X axis. The headform was dropped 90° horizontally angled to the right, resulting in a contact point on the side of the head (Fig. 4). The impact to the upper part of the helmet resulted in rotation around the Y axis (Fig. 5). This impact simulates a crash with oblique impact to the front of the head. The third impact was located at parietal level and was applied in the frontal plane, resulting in a rotation around the Z axis. The head was angled to the side, which gave a contact point on the side of the head (Fig. 6). All three oblique tests simulated a single bike crash or a bike-to-car crash with oblique impact to the head. The tests were performed by RISE and the test method was developed specifically for the current test.

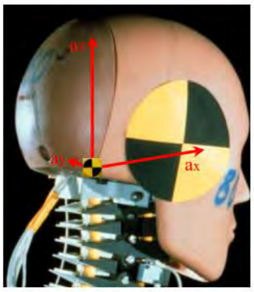


Fig. 2. Translational acceleration. Note that there was no neck in the tests.

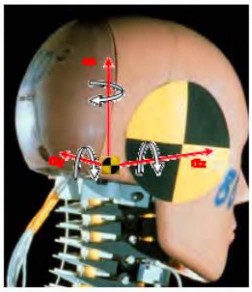


Fig. 3. Rotational acceleration.

Fig. 4. The oblique test A with rotation around X-axis.

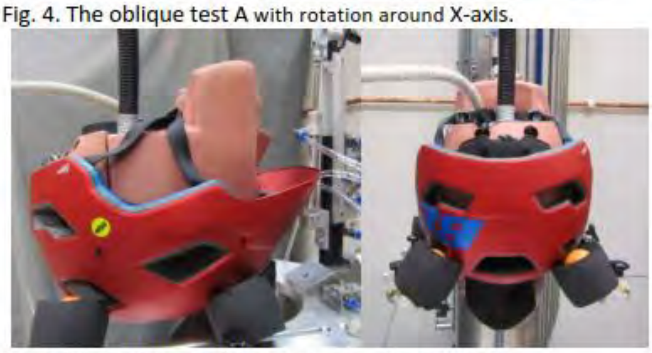


Fig. 5. The oblique test B with rotation around Y-axis.

Fig. 6. The oblique test C with rotation around Z-axis (pre-positioned 20° in X and 35° in Z).

When testing the Hövding 2.0, similar principles were applied as in the standard EN 1078, 5.1 Shock Absorption. However, in both the shock absorption test and in the three oblique tests, an anvil with larger dimensions was used (Fig. 7). The reason was that if Hövding 2.0 had been tested against the anvil used for a conventional helmet, there was a risk it could get in contact with the sharp edges of the anvil. The Hövding 2.0 was pre-inflated and had a pressure of 0.55 bar.

Fig. 7. The Hövding 2.0 and the larger anvil.

FE Model of the brain – Computer simulations

Computer simulations were carried out for all oblique impact tests. The simulations were conducted by KTH (Royal Institute of Technology) in Stockholm, Sweden, using an FE model that has been validated against cadaver experiments [20][21] and against real-world accidents [10][22]. It has been shown that a strain above 26% corresponds to a 50% risk for concussion [21]. As input into the FE model, X, Y and Z rotation and translational acceleration data from the HIII head were used. The FE model of the brain used in the tests is described by Kleiven [10][23].

RESULTS

Shock Absorption Test

All helmets scored lower than 250 g in resultant acceleration in the shock absorption test (Table 2). All except five helmets (Abus S-Force Peak, Carrera Foldable, Giro Sutton MIPS, Occano Urban Helmet and Yakkay) showed a linear acceleration lower than 180 g, which corresponds to a low risk of skull fracture. The Hövding 2.0 helmet performed in average almost three times better than the conventional helmets (48 g vs. an average of 175 g for helmets tested). The POC Octal (135 g) performed best of the conventional helmets, and Yakkay (242 g) preformed worst of the conventional helmets.

TABLE II

SHOCK ABSORPTION - LINEAR ACCELERATION

HELMET	TRANSLATIONAL ACCELERATION (G)
Abus S-Force Peak	202
Bell Stoker MIPS	155
Biltema	189
Carrera Foldable	225
Casco Active-TC	170
Giro Savant MIPS	153
Giro Sutton MIPS	212
Hövding 2.0	48
Limar Ultralight	169
Melon Urban Active	173
Occano U MIPS	178
Occano Urban	192
POC Octal AVIP MIPS	140
POC Octal	135
Scott Stego MIPS	166
Smith Forefront	231
Spectra Urbana MIPS	168
YAKKAY without cover	242
Mean/Median	**175/172**

Oblique Tests

Table III shows the tests that reflect the helmet's protective performance in a bicycle crash with oblique impact to the head (rotation around the X-axis, Y-axis and Z-axis). The mean value of the rotational accelerations varied between the three tests and the lowest strain was measured in the oblique test with an impact to the side of the helmet (rotation around X-axis). The simulations indicated that the strain in the grey matter of the brain during oblique impacts would vary between helmets, from 6% to 22% in the test with rotation around X-axis, 7% to 35% in rotation around Y-axis and 19% to 44% in rotation around Z-axis. Thereby, the threshold for 50% risk of concussion was not exceeded when the impact caused a rotation around the X-axis. When impacting the upper part of the helmet (rotation around the Y-axis) the threshold was exceeded in four of 19 tests, and only one helmet did not give results that exceeded the threshold for a 50% risk of concussion during the impact with rotation around Z-axis. In total, the lowest strain was measured when the airbag helmet was tested. Helmets equipped with MIPS performed, in general, better than the others. The mean values for MIPS were lower in all tests: 13% rotation around X-axis, 21% Y-axis and 32% Z-axis compared to 18% rotation around X-axis, 27% Y-axis and 35% Z-axis.

TABLE III

OBLIQUE TESTS (ROTATION AROUND THE X, Y AND Z-AXIS)

	OBLIQUE IMPACT A (X-AXIS)				OBLIQUE IMPACT B (Y-AXIS)				OBLIQUE IMPACT C (Z-AXIS)			
HELMET	T. ACC. [G]	R. ACC. [KRAD /S^2]	R. V [RAD /S]	STRAIN (%)	T. ACC. [G]	R. ACC. [KRAD /S^2]	R. V [RAD /S]	STRAIN (%)	T. ACC. [G]	R. ACC. [KRAD /S^2]	R. V [RAD /S]	STRAIN (%)
Abus S-Force Peak	175	7.5	29.9	16	131	7.2	33.4	23	145	14.4	39.5	33
Bell Stoker MIPS	112	4.2	23.2	11	100	6.2	31.8	23	114	10.5	39.1	31
Biltema helmet	124	5.1	29.3	15	115	7.1	36.4	25	126	13.8	42.1	34
Carrera Foldable	180	7.9	27.6	16	147	7.8	32.9	25	157	15.5	42.3	35
Casco Active-TC	123	7.8	31.3	20	116	8.0	38.8	29	76	10.1	44.4	35
Giro Savant MIPS	120	5.3	24.7	12	100	4.2	28.3	17	103	9.5	38.7	31
Giro Sutton MIPS	124	4.5	23.8	11	116	6.1	34.2	23	139	13.7	41.0	33
Hövding 2.0	42	1.5	26.9	6	37	1.7	28.6	7	27	2.8	37.1	19
Limar Ultralight	132	8.5	31.5	18	121	7.0	36.9	26	111	12.4	43.2	34
Melon Urban Active	138	6.1	29.2	16	131	8.2	34.5	26	128	12.6	40.5	33
Occano U MIPS	121	5.1	23.9	12	126	5.3	29.1	20	156	14.7	39.5	32
Occano Urban	161	7.5	31.6	17	121	7.6	35.8	27	131	13.9	42.6	35
POC Octal	102	7.6	32.2	19	90	6.2	35.6	24	77	9.2	43.5	33
POC Octal AVIP MIPS	95	5.3	23.2	12	87	4.5	30.5	19	74	10.2	42.1	33
Scott Stego MIPS	94	6.8	35.4	19	103	6.7	32.7	24	86	9.4	42.8	33
Smith Forefront	136	8.1	31.5	18	166	10.0	38.9	30	149	13.5	40.0	33
Spectra Urbana MIPS	155	6.1	21.9	12	115	5.8	27.2	19	109	10.5	40.2	32
YAKKAY with cover	150	5.8	19.2	14	156	5.1	24.3	16	167	14.1	36.0	30
YAKKAY without cover	174	10.8	33.8	22	174	12.9	39.4	35	142	18.1	46.6	44
Mean	129	6.4	27.5	15	119	6.7	33.1	23	117	12.0	40.9	33
Median	124	6.1	29.2	16	116	6.7	33.4	24	126	12.6	41.0	33

DISCUSSION

There was a large variation in the results of the test that simulate the helmets' capacity to absorb impact energy. The measured linear acceleration varied from 48 g to 242 g. The best conventional helmet, POC Octal, reduced the kinetic energy exposing the head form to 135 g, which is nearly half the level in the test standard (250 g). This was, however, almost three times higher than the corresponding value for Hövding 2.0 (48g), a head protector that is inflated during an accident situation and acts as an airbag for the head. A similar result was shown by Kurt et al [24] when comparing a conventional helmet with Hövding 2.0. All of the evaluated bicycle helmets comply with the legal requirements in Sweden. It has previously been shown that conventional helmets often have a good protective effect of reducing of head injury (see, for example, [1][2][25][26]). The effectiveness is higher for skull fractures than for brain injuries [26]. However, the limit for linear acceleration of 250 g in the test standard is relatively high, mainly with a focus on avoiding skull fractures. Research has shown that the risk of skull fractures could be dramatically reduced (from 40% to 5% risk) if the translational acceleration was reduced from 250g to 180g [6]. In the present study, all helmets except five showed a linear acceleration lower than 180 g. It would be interesting to study if the current limit could be reduced and thereby increase the helmet's effectiveness for skull fractures. Notably, the results showed that a conventional helmet that meets today's standards would not prevent a concussion in case of a head impact. A concussion could result in permanent symptoms such as memory loss and it has been reported as a common injury resulting from head impacts in bicycle crashes [27]. To prevent concussions, all of the studied helmets would need to absorb energy more effectively. Reconstructions of head to head impacts in American football indicate that concussions start to occur at 60-100g [8]. Brain injury is primarily caused by rotational movement rather than linear forces [9-13]. Despite this, the bicycle helmet standards do not include angular acceleration for certification. The present helmet standard (Shock absorption included EN 1078) have therefore been criticised, since there is a risk that the effect is that helmets today are mainly designed to reduce the risk of skull fracture and not brain injury [16][17]. Oblique impacts will probably be included in future standards, but before that consumer tests can play an important role. The present study has implemented rotational acceleration in consumer tests for bicycle helmets and evaluated the variation in performance. Few helmets provide good protection against oblique impacts (rotational velocity combined with translational acceleration), which is the most common accident scenario for a bicycle accident with a head impact [28]. Some of the included helmets (Bell Stoker MIPS, Giro Savant MIPS, Giro Sutton MIPS, Occano U MIPS, POC Octal AVIP MIPS, Scott Stego MIPS and Spectra Urbana MIPS) are designed to absorb rotational forces. These helmets generally perform well in the rotational tests. The Hövding 2.0 also obtained very good results in the rotational tests. The simulated strain in the brain during impact with rotation around the z-axis varied from 19% (rotational acceleration 2.8 krad/s2) to 44% (18.1 krad/s2), where 26% corresponds to 50% risk for a concussion. The Hövding 2.0 was the only helmet that did not give results that exceeded the threshold for 50% risk of concussion. Thus, the Hövding 2.0 is the recommended choice considering both shock absorption and oblique impacts. This assumes, however, that it inflates properly in the case of an accident. The above examples clearly demonstrate that there are several ways to design a helmet to absorb rotational acceleration. However, the results show that an oblique impact to the head involves a high risk of severe injury, such as concussion with a loss of consciousness or DAI. To prevent these injuries in the future, oblique impact tests similar to those conducted in the current study must be included as part of the legal standard requirements for bicycle helmets. Based on the results, the computer simulations seem to be more relevant to use for comparing and rating the helmets. The peak values do not capture the influence of the duration. Therefore it is recommended to use computer simulations in future tests.

LIMITATIONS

In the shock absorption test in the current ISO headform was used. It could be argued that this headform is quite far from the human skull [16]. The applied approach was nevertheless consistent with the current European standard. The helmets were impacted at the crown to be as close as possible to the centre of mass and to minimise the effect of helmet design. In the oblique impact tests a Hybrid III dummy head was used. This head has much more realistic inertia properties and could easily be fitted with rotational accelerometers that allow measurements of the rotational velocity and rotational acceleration. The oblique impact locations were chosen to correspond with the most frequent head impact locations. One possible explanation for the variation in the results for both translational and oblique impacts is the difference in geometric helmet design. In addition, the variation may also be caused by the fact that the helmet was not fitted equally firmly on the headform. The helmets were fitted on the head form with the intention that the neck adjustment system should be adjusted as consistently as possible, using the same procedure as in the certification tests. Furthermore, the tests were conducted under specific laboratory settings, which might not reflect real-world conditions. For example, the helmets fitted properly with the headform and they were strapped on correctly. The airbag helmet Hövding 2.0 was pre-inflated before all of the test impacts, even though it needs a real-world accident scenario, e.g. with high acceleration or rotation, to be activated and inflated properly. The performance regarding activation of Hövding 2.0 was not part of the test series. Today, helmets are tested using a free falling headform excluding the rest of the body. The test set-up used in the present study corresponds to either the current helmet standard or the proposal from the CEN Working groups 11 "Rotational test methods" and both set-ups exclude the rest of the body. Previous studies have examined the influence of the neck and the body on the helmet performance. It has been shown that the body influence the kinematics and it effect the brain tissue strain [29]. Furthermore, in the oblique tests the Hybrid III dummy head was tested without an attached neck. In-house test with Hövding 2.0 and a conventional helmet tested with and without an attached neck showed that the conventional helmet was more sensitive for variations [30]. The influence of the body and neck on head kinematics needs to be further addressed. Each of the four included tests were only conducted once. Variations within helmets of the same model could occur as an effect of small differences between the helmets or due to small variations in the test setup. They might be more important for the oblique tests. It is thus recommended that future studies address the effect of such variations and that at least two helmets are tested for each test configuration. However, the CEN/TC 158 Working Group 11 have initiated a Round-robin tests that showed that the measured results, within a test lab and between test labs, can be controlled [31]. Several FE models exist that appear to be appropriate to use for simulating the risk of brain injuries. In the present study a model developed by KTH [10][23] was used, which has been validated against experiments and real-world data [10][20-23]. However, future studies could be useful to compare outcomes from various models.

CONCLUSIONS

The current European certification test standard do not cover the helmets' capacity to reduce the rotational acceleration, i.e., when the head is exposed to rotation due to the impact. The present study provides evidence of the relevance of including rotational acceleration in consumer tests and legal requirements. The results have shown that rotational acceleration after impact varies widely among helmets in the Swedish market. They also indicate that there is a link between rotational energy and strain in the grey matter of the brain. In the future, legal bicycle helmet requirements should therefore ensure a good performance for rotational forces as well. Before this happens, consumer tests play an important role in informing and guiding consumers in their choice of helmets. The initial objective of the helmet standards was to prevent life threatening injuries but with the knowledge of today a helmet should preferably also prevent brain injuries resulting in long-term consequences. Helmets should be designed to reduce the translational acceleration as well as rotational energy. A conventional helmet that meets current standards does not prevent a cyclist from getting a concussion in case of a head impact. Helmets need to absorb energy more effectively.

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Advanced Cycle Helmet Testing Protocols: Effects of Testing Procedure on the Outcomes of Bicycle Helmet Safety Tests

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Keywords: helmet, head injury, impact, cycling.

INTRODUCTION

Cyclists are a particularly vulnerable road user with the lack of adequate protection during collisions increasing the risks of serious trauma [1]. In Great Britain, 3,327 cyclists were killed or seriously injured in 2015 alone [2]. During this period, cyclists were observed to be the second most vulnerable road user (VRU) in Great Britain, experiencing a casualty rate of 3,327 casualties per billion cycled kilometres. Cycle helmets are, however, a vital item of personal protective equipment that aim to reduce head injury severity by providing wearers with adequate protection during a collision.

Currently, there is no freely available and independent information provided to consumers at the point of sale to allow them to assess the relative safety performance of cycle helmets. One key reason for this is the need to understand the fundamental science underpinning the development of such protocols. Four test packages were undertaken which investigated the effects of impact energy, compound impacts (where a single helmet location is impacted multiple times), impact anvil angle, impact location and the suitability and repeatability of different headforms.

METHODS

Compound Impacts

This study aimed to ***quantify the effects of impact energy and compound impacts, for both flat and kerbstone impact anvil designs, on head injury risks for a single helmet model***. Wire-guided linear drop tests, following CPSC – 16 CFR Part 1203 protocols, were performed to assess the effects of impact energy and compound impacts on head injury risk. Cycle helmets were securely mounted to EN 960:2006 specified three-quarter headforms, before impacting EN 1078:2012+A1:2012 specified flat and kerbstone shaped anvils at predefined impact locations within the left and right temporal regions of the cycle helmet (Figure 1). Only one helmet was model was selected for use in this study (Trax Mistral Bike Helmet).

Two consecutive drop tests of each helmet were performed for each impact location. The first test was performed across a range of heights ranging from 1-3 m in 0.5 m increments, whilst the second test was performed from a height of 1 m only. Helmets were dropped on to either a flat or a kerbstone anvil based upon testing requirements. Various metrics were recorded for each helmet impact and compared to a range of current state-of-the-art head injury criteria; here outcomes are presented for peak linear accelerations only.

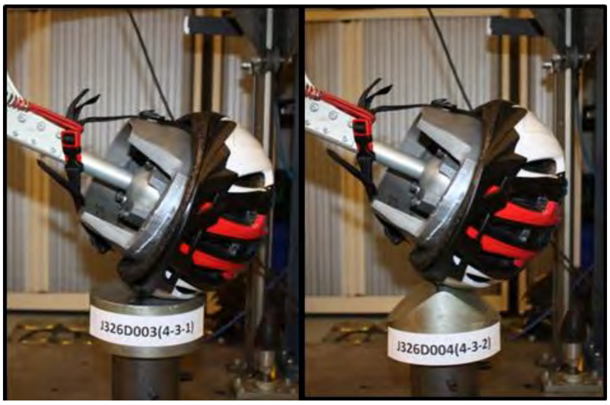

Figure 1: Wire-guided linear headform drop test set-up impacting the right temporal region on the flat and kerbstone anvils.

Influence of Anvil Angle

This study aimed to ***quantify the effects of the impact anvil angle across a range of cycle helmet models and establish the repeatability of the oblique impact testing approach***. Free fall drop carriage tests, which adapted EN 1078:2012+A1:2012 protocols to perform oblique impact tests, evaluated the effects of anvil angle on head injury risk. Helmets were securely mounted to a full sized EN 960:2006 specified headform, before being positioned on a modified “horseshoe” drop carriage design to ensure the left temporal region of the helmet was impacted. Four different helmet models were selected for this study:

- Model 1: Trax Mistral Bike Helmet
- Model 2a: Bell Draft MIPS Helmet 2016
- Model 2b: Bell Draft MIPS Helmet 2016 – with MIPS structure removed
- Model 3: Mongoose Urban Helmets

Drop tests were performed by impacting helmeted headforms against a flat steel angled anvil, with fresh 80 gsm sandpaper attached securely to the anvil face for each test. A range of anvil angles was investigated in 5° increments between 30-60° to the horizontal, with impact velocities calculated specifically for each anvil angle to represent collisions occurring at different cyclist speeds. Outcomes are presented for peak linear accelerations, rotational velocities and rotational accelerations and compared to a range of head injury criteria. To assess repeatability, 20 additional helmet drop tests (five repeat helmet drop tests for each helmet model) were also performed using a 45° anvil angle with a 3 m drop height.

Influence of Headform Type and Effects of Impact Location

This study aimed to ***quantify the differences in the kinematics of the head between the EN 960:2006 and Hybrid III headforms*** during oblique cycle helmet impacts. Using an updated approach, based on the lessons learnt from the previous study, a free fall drop carriage test was performed by impacting helmeted headforms against an angled anvil to assess head injury risk. Helmeted headform drop tests used either a full sized, EN 960:2006 compliant, 575 mm circumference magnesium headform (4.82 kg) or 50^{th} percentile Hybrid III headform (4.54 kg). Two different helmet models were selected for use in this study:

- Model 1: Trax Mistral Bike Helmet
- Model 2a: Bell Draft MIPS Helmet 2016

Helmets were securely mounted to the specified headform, before being positioned on the modified “horseshoe” drop carriage to impact the helmeted headform across four different specified helmet impact locations including: the crown, frontal, occipital and left temporal regions. Each helmeted headform was dropped from a height of 3 m onto a flat steel anvil angled at 45° to the horizontal plane, with fresh 80 gsm sandpaper attached securely to the anvil face. Each helmeted headform was impacted once, with three repeat tests performed for each helmet model, impact location and headform.

Outcomes were calculated for peak linear accelerations, rotational velocities and rotational accelerations and compared to a range of head injury criteria. Mean differences in safety performance between the headforms used were compared to evaluate the influence of the headform on each outcome.

Repeatable Differentiation of Performance

This study aimed to explore the repeatability of the oblique impact testing protocols and establish, by simulating an idealised helmet slip plane, whether these protocols may be used to differentiate between the rotational impact safety performance of different cycle helmet models. In order to simulate the idealised helmet slip plane, the 80 gsm sandpaper (which would normally be securely attached to the anvil face) was strategically cut to leave ≤5 mm of material supporting its attachment to the anvil. This study then compared the differences in outcomes between tests performed with the idealised slip plane and with the sandpaper securely attached to the anvil face.

Free fall drop carriage tests were performed by impacting helmeted headforms against a 45° angled anvil from a drop height of 3 m to assess head injury risk. Helmeted headform drop tests used a 50th percentile Hybrid III headform (4.54 kg), whilst only one helmet model was selected for use in this study (Trax Mistral Bike Helmet). Helmets were securely mounted to the specified headform, before being positioned on the modified “horseshoe” drop carriage to impact the helmeted headform across four different specified helmet impact locations including: the crown, frontal, occipital and left temporal regions. Each helmeted headform was impacted once, with five repeat tests performed at each impact location and for each experimental slip plane case.

Outcomes were calculated for peak linear accelerations, rotational velocities and rotational accelerations and compared to a range of head injury criteria. Mean differences in safety performance between the “no slip” (i.e. fixed 80 gsm sandpaper) and the idealised slip plane cases were compared to evaluate whether the proposed oblique impact safety performance protocol will be able to establish any difference in performance between helmet models.

RESULTS

Compound Impacts

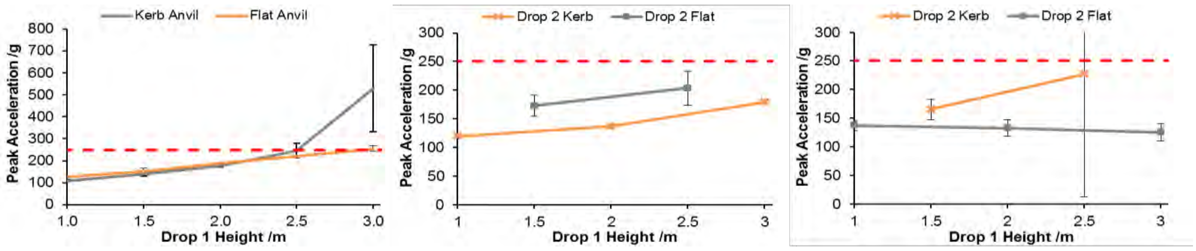

Figure 2: Mean peak linear headform accelerations for (a) the first impact (Drop 1) against both the flat and kerbstone anvils and the compound impact (Drop 2) against both the flat and kerbstone anvils when compared to the drop height of the first impact (Drop 1) against the (b) flat and (c) kerbstone anvils

Higher impact energies resulted in greater peak head accelerations, regardless of impact partner shape. Impacts against a kerbstone anvil carry a greater injury risk than flat anvil impacts at drop heights of ≥2.5 m. Compound impact injury risks were affected by the proportion of undamaged helmet engaged by the second impact.

Influence of Anvil Angle

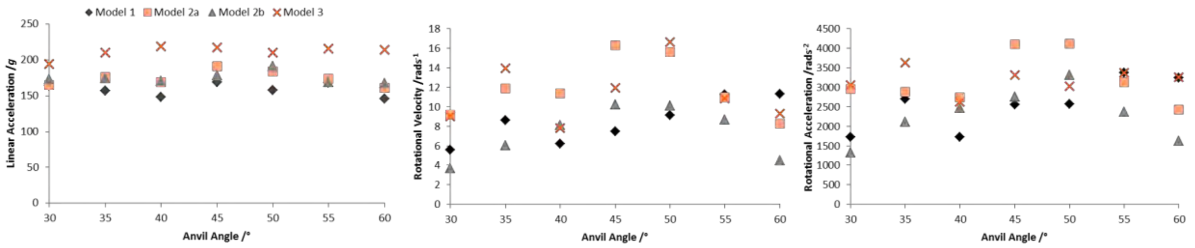

Figure 3: Peak (a) linear accelerations, (b) rotational velocities and (c) rotational accelerations experienced by four different helmet models when impacted against a range of different anvil angles

No particular angle seems to consistently provide a “worst-case” angle across helmet models tested; 45° perhaps the most appropriate test angle as it was a peak value for both rotational velocity and acceleration.

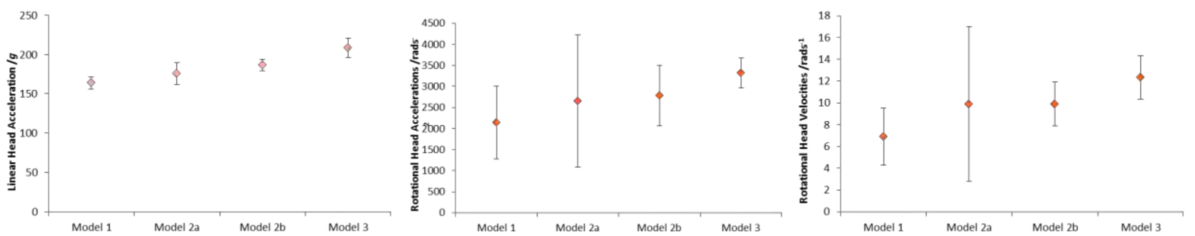

Figure 4: Repeatability of oblique impact tests for four different helmet models

Fair repeatability for linear acceleration response, but very poor for rotational velocity and acceleration was identified, so the helmet mounting process was reviewed and updated.

Influence of Headform Type and Effects of Impact Location

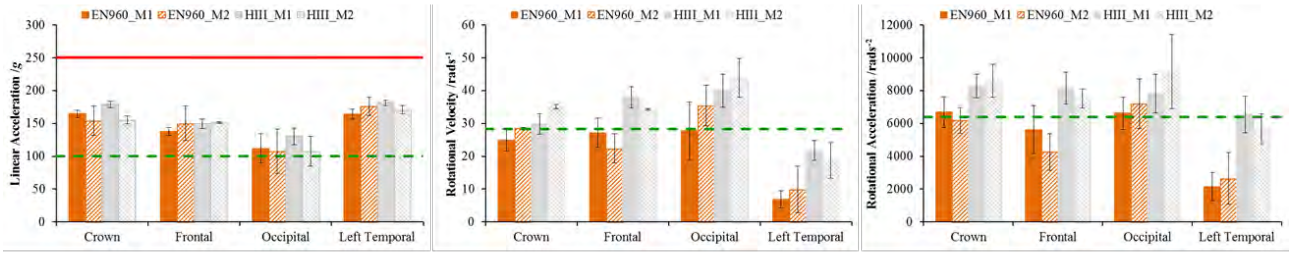

Figure 5: Mean peak (a) linear accelerations, (b) rotational velocities and (c) rotational accelerations experienced by the EN 960:2006 (EN960) and Hybrid III (HIII) headforms when testing two different helmet models (M1, M2) at four different impact locations

A significant increase in rotational velocities and accelerations for the Hybrid III headform was found 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 no significant difference was observed for Model 2.

Repeatability and Differentiation

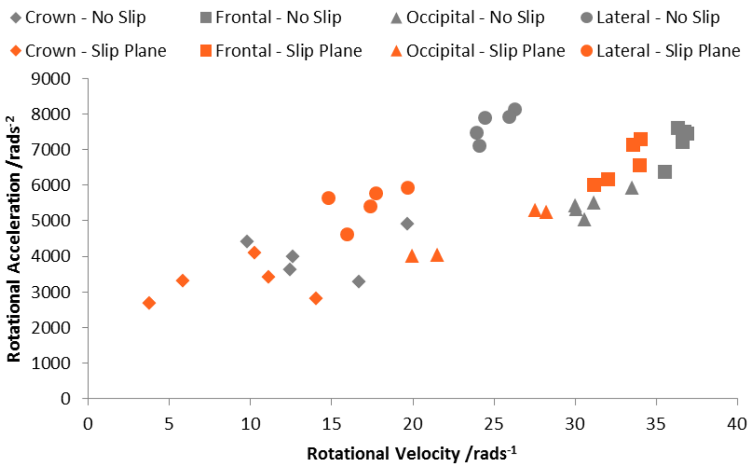

Figure 2: Comparison of rotational velocity and rotational accelerations for each impact location and each slip plane case

The repeatability of the final procedure across the five repeat tests for each impact location and slip plane case was found to be acceptable. Linear accelerations were found to have a ~3% coefficient of variation (CoV), rotational accelerations had a ~10% CoV and rotational velocities had a ~10% CoV. The “no slip” helmeted headform drop tests experienced greater rotational velocities across all impact points and greater rotational accelerations for the frontal, occipital and temporal regions. For the linear accelerations only the temporal region experienced any differences in peak linear acceleration.

DISCUSSION

Compound Impacts

Impact energies, impact partner shapes and compound impacts were all shown to affect the safety performance of cycle helmets (Figure 2). Higher impact energies were observed to result in greater peak linear headform accelerations. Although a considerable increase in headform accelerations was caused by the kerbstone anvil for drop heights of 2.5 m or greater, high energy impacts onto the flat anvil only exceeded legislative safety performance criteria when impacted from a 3.0 m drop height (when compared to drop heights of 1.5 m in current test standards). Compound impacts were primarily affected by the proportion of undamaged EPS material engaged by the compound impact (Figure 7). Compound impact tests which had the greatest area of impact with non-damaged helmet material (in this case the kerbstone-flat combination) had the best impact safety performance. It was therefore recommended that advanced testing protocols should recognise and assess the relative safety performance of cycle helmets against these various variables.

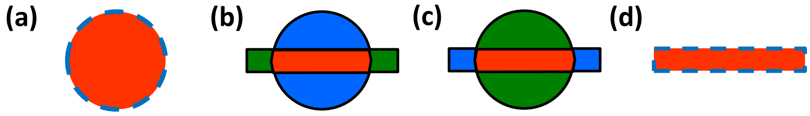

Figure 7: Schematic of impact partner shape overlap areas for the following impact sequences (a) flat-flat, (b) flat-kerbstone, (c) kerbstone-flat and (d) kerbstone-kerbstone. The initial impacts are illustrated in blue, overlapping compound impact areas are red and non-overlapping compound impact areas are green.

Influence of Anvil Angle

Differences seemed to exist in cycle helmet performance when impacted at different anvil angles, with different helmet models seeming to respond differently to different anvil angles. No specific angle, however, seemed to consistently provide a "worst-case" angle across all helmet models tested. An anvil angle of 45° to the horizontal and impacted from a drop height of 3 m was, however, perhaps the most appropriate combination to use as peak values for the rotational velocities and accelerations seemed to be located approximately at this point. Furthermore, this combination represented a cyclist fall occurring whilst cycling at 20 km/h, which is approximately the average speed for a cyclist.[3]

Differentiation between the oblique safety performances of the helmet models, particularly for the rotational headform velocities and accelerations, was unachievable due to the poor repeatability of the oblique impact test methods adopted. This was a key outcome of this study and identified a necessity to highly control several of the key testing variables. These key variables included the impact location, strength of the helmeted headform anchorage, helmet position on the headform and adequate adjustment of the retention system.

Influence of Headform Type and Effects of Impact Location

Although no legislative performance criteria were exceeded, at least one AIS2+ head injury criterion was exceeded during each oblique impact test. The 100 g AIS2+ linear acceleration injury criterion was exceeded during all but one impact test, whilst 48% of helmet drop tests exceeded the 28.3 rads-1 AIS2+ rotational velocity injury threshold and 54% exceeded the 6,383 rads-2 AIS2+ rotational acceleration injury threshold.

A significant increase in rotational velocities and accelerations for the Hybrid III headform was found 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 no significant difference was observed for Model 2. Given these differences, and the consensus expert opinion that the Hybrid III headform is more biofidelic in its design, it was recommended that future advanced cycle helmet testing protocols consider the use of the Hybrid III range as the test headform.

Repeatability and Differentiation

When considering differentiating between the oblique impact safety performances of the two slip plane cases, it is clear to see that impact safety performance was more sensitive to helmet impact location than differences in helmet designs. When oblique impact safety performance is compared at equivalent impact test locations, however, it is clear that safety performance may be differentiated between the two slip plane cases. This difference, although distinct, remains only marginal for certain impact locations (e.g. frontal), whilst is much larger for other impact locations (e.g. lateral). This implies that, should the idealised slip plane assumption of the "slip plane" cases hold true, there may be very little real-world benefit to be gained by introducing a slip-plane at the impact locations with a marginal difference in safety performance. This outcome does, however, direct future advanced cycle helmet test and assessment protocols towards ensuring that multiple impact locations are assessed. It is important to explore whether these outcomes are transferrable between other models and especially those that claim improved performance during oblique impacts.

FUTURE RESEARCH REQUIREMENTS

Although this project makes significant progress towards advancing the state-of-the-art in advanced cycle helmet impact safety performance test and assessment protocols, a number of topics still require further research. These can be split into two key sections, topics that require further research before finalising the linear and oblique impact test protocols and topics that require further research to develop future test and assessment protocols.

When considering the current linear and oblique test and assessment protocols proposed within this project, further research is required prior to being able to finalise the protocols. Firstly, and perhaps most importantly, a case-by-case investigation into cyclist collisions and falls may be required to reconstruct the cases and better understand the magnitude and angle of impact forces experienced by cycle helmets during typical impact scenarios. This may be used to prioritise the test points and headform orientations for use in the protocols. It is also important to better establish the influence of both the impact angle and drop height during oblique impact tests. As the repeatability of the test methods were improved in the later studies, an improved analysis, that significantly reduces the variation between results, may now be performed for a range of cycle helmets to more robustly understand the effects of impact angle and drop height on outcomes. Finally, the reproducibility of the proposed test and assessment protocols between laboratories should also be established and improvements made to the reproducibility of the protocols.

Future research is also required for the development of new test and assessment protocols. This research can be split into key topics that aim to develop the impact testing, comfort testing and safety performance assessment aspects of future rating schemes. Impact safety performance test protocols may be developed to assess the retention system strength and stability, high/low energy oblique impacts and the helmet coverage area and performance. Furthermore, the influence of different neck forms during impact also requires investigation. Comfort rating test protocols may also be developed to evaluate the fit, field of view, mass, visor fogging, waterproof, acoustic emission, aerodynamic and ventilation performance of cycle helmets. Finally it is important to establish the benefits of using finite element analysis (FEA) approaches to determine head injury risk for use in the assessment protocols. This will remove the need to derive separate injury criteria for the linear and oblique impact testing protocols and promote the use of a combined cycle helmet safety performance criterion.

CONCLUSIONS

This project has contributed to advancing the state-of-the-art in the testing and assessment of cycle helmet impact safety performance. The impact performance of cycle helmets during higher energy linear impacts against flat and kerbstone anvils were characterised, alongside the safety performance of cycle helmets during linear compound impacts. Oblique impacts to the helmet were also investigated, with the effects on the outcomes of both the angle of the anvil and the headforms used during testing established. The repeatability of the impact safety performance testing and assessment protocols was then analysed to evaluate the suitability of these protocols for the advanced cycle helmet test and assessment protocols. Whilst the proposed test and assessment protocols apply current best practices, further research is required before finalising the protocols.

ACKNOWLEDGEMENTS

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C.4 Megan Bland (PhD Student, Virginia Tech)

Influence of Headform and Neck During Bicycle Helmet Testing

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INTRODUCTION

Although head injuries in American football have received copious media attention and research efforts in recent years, injury surveillance systems have shown that cycling actually accounts for more head injuries treated in U.S. emergency rooms annually than any other helmeted recreational activity or sport [1]. Outside the U.S., cycling is also a common sport, recreational activity, and mode of transportation in many other countries, and as a result, head injuries sustained from cycling is an issue of global magnitude.

Bicycle helmets have been shown to reduce risk of head injury, and must pass safety standards mandating that a helmet limit peak linear acceleration (PLA) of an ISO half-headform in drop tests normal to the impact surface. However, this type of impact is not reflective of real-world cyclist accidents, which are oblique to the impact surface [2] and induce rotational acceleration as well as linear, a key factor in diffuse brain injury [3]. The choice of headform and its rigid coupling to the drop rig also do not reflect human properties. Testing helmets in oblique impacts using a biofidelic headform and neck would enable a more realistic assessment of helmet effectiveness, but to-date there is a lack of agreement concerning the boundary conditions of these methods. Specifically, it has been debated whether the type of biofidelic headform used yields differing impact response [4-5], as well as how the presence of a neck and/or effective mass affects the dynamic response during the impact event [6].

The purpose of this study was to investigate effects of varying headform and neck configurations on dynamic response in bicycle-helmeted, oblique impacts.

METHODS

Impact tests were conducted on a linear drop tower using two human-like headforms common in helmet testing: a 50th percentile male Hybrid III (HIII) and a National Operating Committee on Standards for Athletic Equipment (NOCSAE) headform. Three neck conditions were considered for each headform: no neck (headform alone), free neck, in which a 50th percentile male HIII neck was connected to the headform but not attached to the drop tower (meant to simulate an effective mass of the neck), and guided neck, in which the HIII neck was connected to the headform and attached to an effective torso mass (16 kg) connected to the drop tower (Fig. 1). In the no neck and free neck conditions, the head/neck was not constrained to the drop tower and was free to rotate off the anvil upon impact. The NOCSAE headform was modified to accommodate the HIII headform according to previously published studies [4], and the mass of the modified NOCSAE was similar to that of the HIII headform.

Headforms were fitted with a bicycle helmet and impacted at a frontal and parietal location at 6 m/s against a sandpaper-covered 45° anvil (simulating oblique road impacts). The impact velocity, locations, and angle were selected to reflect common cyclist head impacts [8]. Consistency in hitting these precise impact locations for the no neck and free neck conditions was ensured through the use of a dual-axis inclinometer.

The two headforms, three neck conditions, and two impact locations yielded 12 configurations, each of which was tested five times. Three linear accelerometers and three angular rate sensors at the headform center of gravity (CG) captured kinematics for all tests. Differences in impact duration, time to PLA, PLA, peak rotational velocity (PRV), time to PRA and peak rotational acceleration (PRA) were evaluated for all tests. Radius of rotation, defined as the distance of the center of rotation from the head/neck CG and calculated as PLA/PRA, was also computed for all tests. All comparisons were made using 3-way ANOVA and Tukey's HSD post hoc tests.

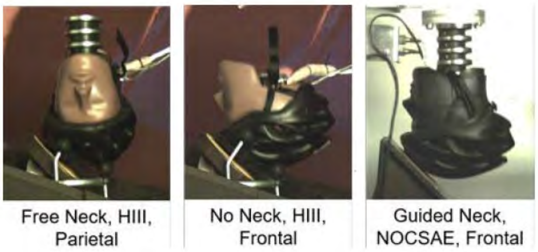

Figure 1: Example configurations demonstrating the two headforms, three neck conditions, and two locations.

RESULTS

Each of the 12 impact configurations produced distinct kinematic responses (Table 1). Time- and linear-based metrics were similar between the HIII and NOCSAE, with average differences less than 6%, while rotation-based metrics were more substantial, with the HIII producing 23% and 33% greater PRV and PRA, respectively. Trends were similar across impact location. Differences were significant between headforms for all metrics except PLA and time to PRA, and average coefficients of variance (CV) were 5 and 7% for the HIII and NOCSAE.

Table 1: Metric averages ± standard deviations for each headform and neck condition. Percent differences from the first column of the headform/neck condition and significance level are given in the second rows.

	Headform		**Neck**		
	HIII	**NOCSAE**	**Guided Neck**	**Free Neck**	**No Neck**
Duration [ms]	10.3 ± 0.7	9.8 ± 1.0 -4.6% ($p<0.01$)	10.3 ± 1.0	10.3 ± 0.7 -0.6% ($p=0.06$)	9.5 ± 0.7 -7.6% ($p<0.01$)
Time to PLA [ms]	4.5 ± 0.6	4.3 ± 0.6 -5.3% ($p<0.01$)	4.3 ± 0.6	4.5 ± 0.7 3.1% ($p=0.36$)	4.4 ± 0.4 0.1% ($p=0.99$)
PLA [g]	135.9 ± 17.2	136.5 ± 20.1 0.5% ($p=0.66$)	114.8 ± 10.2	138.8 ± 6.9 20.9% ($p<0.01$)	155.1 ± 7.4 35.1% ($p<0.01$)
PRV [rad/s]	29.2 ± 6.9	22.6 ± 5.1 -22.8% ($p<0.01$)	26.3 ± 6.1	20.7 ± 5.7 -21.1% ($p<0.01$)	30.7 ± 5.1 16.9% ($p<0.01$)
Time to PRA [ms]	3.5 ± 1.1	3.4 ± 1.4 -3.0% ($p=0.58$)	3.7 ± 1.6	2.7 ± 1.0 -26.2% ($p<0.01$)	4.0 ± 0.5 10.7% ($p=0.21$)
PRA [rad/s^2]	7168 ± 1333	4827 ± 1424 -32.7% ($p<0.01$)	5901 ± 2121	4977 ± 1450 -15.7% ($p<0.01$)	7114 ± 1057 20.5% ($p<0.01$)
Rotation Radius [cm]	19.1 ± 3.6	29.9 ± 8.9 56.5% ($p<0.01$)	21.6 ± 8.2	29.9 ± 10.0 38.5% ($p<0.01$)	22.0 ± 4.4 1.8% ($p=0.80$)

The neck condition produced considerable differences in all variables except duration and time to PLA. PLA was largest for no, then free, then guided necks, while PRV and PRA were largest for no, then guided, then free necks. The PLA-PRA relationship was similar for the no and guided neck conditions, generating nearly identical radii of rotation. Trends were similar across location, and average CV were 4, 6, and 8% for no, free, and guided necks.

DISCUSSION

The different headforms and neck conditions evaluated herein produced clear differences in kinematic response. The main similarity between configurations was the linear response between headforms, which may be attributable to the fact that both headforms were initially validated based on linear acceleration responses in cadaver drop tests [4]. Rotational responses were markedly different between headforms, however, and point to differing inertial properties. The HIII produced higher rotational velocities and accelerations and a smaller radius of rotation, suggesting this headform may have a smaller moment of inertia (MOI) about than the NOCSAE and/or that the resultant force vector was closer to its CG. The HIII MOI about various axes were designed based on cadaver data, while the NOCSAE MOI are not well-documented [5]. While both headforms produced low variance and are suitable candidates for impact testing, further testing is needed to determine MOI characteristics. The two headforms also contain different surface friction, the effects of which should be evaluated in dynamic testing.

Neck condition produced considerable differences in both linear and rotational kinematic response. No neck produced the highest PLA, then free, then guided. This suggests that the neck and torso masses contributed to overall effective mass in these impact scenarios, lowering PLA. The addition of a neck and torso also affected inertial properties, generating substantial differences in rotational metrics. These differences did not follow the same trend as PLA, with free neck producing the lowest values rather than guided. No neck and guided neck conditions produced the same radii of rotation. Incidentally, this relationship between PLA and PRA is similar to those seen in concussive impacts [8], although under differing impact scenarios. A major limitation to the guided neck system, however, is that it inherently involves loading of the surrogate neck, which is known to be unrealistic in certain configurations. No neck may be superior as it avoids this complication and produced the lowest variance. Testing additional velocities, helmets, locations, and anvil angles would further elucidate these head/neck effects.

CONCLUSIONS

This study demonstrates that choice of headform and neck are key factors affecting dynamic response in oblique impacts. Optimizing the test setup to replicate real-world cyclist impacts can stimulate improved helmet safety.

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C.5 Steven Rowson (Assistant Professor, Virginia Tech)

Summarizing Helmet Performance: Injury Risk and Rating Schemes

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INTRODUCTION

Owing to increased public awareness of sport and recreational-related brain injuries such as concussion, growing priority has been placed on identifying, understanding, and preventing these injuries [1]. Helmets are a primary preventative measure in many activities, and are currently designed around passing standards that aim to minimize risk of skull fracture [2]. However, helmet effectiveness in reducing risk of other types of head or brain injury is not assessed. In order to evaluate a helmet's ability to mitigate these risks as well as stimulate improved helmet design, laboratory testing needs to accurately model real-world impact scenarios, and resulting kinematics need to be properly interpreted in terms of their relation to injury risk. Consideration of a helmet's ability to reduce injury risk provides a basis for issuing helmet ratings in order to differentiate performance among various models.

Kinematic parameters measured from laboratory impact testing of helmets are commonly related to brain injury risk, as they are thought to be indicative of the inertial response of the brain inside the skull [3]. Specifically, two types of head motion are often cited as resulting in two separate modes of injury: linear motion and rotational motion. Linear motion has been shown to be correlated to transient intracranial pressure gradients and results in more focal injury, while rotational motion is correlated to relative brain motion and strain and results in more diffuse injury [4]. These two injury modes have traditionally been studied separately, although real-world impacts nearly always involve both linear and rotational motion, and thus both likely factor into injury [3]. Understanding the mechanisms of brain injury allows researchers to relate kinematic parameters to injury risk through the development of brain injury criteria, which can then be used to predict likelihood of injury from a given impact.

BRAIN INJURY CRITERIA

Review of Criteria

Early investigation of the relationship between kinematic parameters and head injury led to the development of the Wayne State Tolerance Curve, which related linear acceleration and duration of impact to injury tolerance. Various injury metrics were subsequently developed from this curve – namely the Gadd Severity Index (GSI) and Head Injury Criterion (HIC), which now serve as the basis for head injury safety standards in the automotive and helmet industries. These criteria are primarily correlated to skull fracture, but are thought to also correlate with severe brain injury. However, the criteria do not evaluate rotational motion, which is now known to be a key factor in brain injury [4]. Despite this, implementation of these criteria into standards have been effective in reducing injury incidence, as they ensure that energy input to the head is limited through appropriate countermeasures.

More recent efforts to relate kinematic parameters to injury have produced newer brain injury criteria that assess rotational motion alone, such as the Kinematic Rotational Brain Injury Criterion (BRIC) [5], or a combination of linear and rotational motion, such as Head Impact Power or the Concussion Correlate [6-7]. The underlying kinematic data informing these criteria stem from laboratory reconstructions of head impacts in football, advanced animal testing, and on-field measurements of football head impacts through helmet instrumentation. These newer criteria enhance injury classification capabilities, especially for mild traumatic brain injuries (mTBI).

Given the growing public awareness of brain injury, there now exist a large number of new brain injury criteria, with generally no consensus on the superiority of one over another. This leaves researchers with the rather open-ended challenge of selecting which criteria to employ when evaluating brain injury risk from impact testing. To address this, laboratory reconstructions of football concussive head impacts were performed using a pneumatic linear impactor. Six degree-of-freedom data were collected and used to assess the predictive capabilities of thirteen different brain injury criteria, including those evaluating linear motion, those evaluating rotational motion, and those evaluating both. All criteria were found to perform similarly, although those combining linear and rotational kinematics demonstrated superior predictive capabilities. As such, it is recommended that mechanistic criteria evaluating both linear and rotational aspects of head impacts be utilized when evaluating brain injury risk.

Application of Criteria

While brain injury criteria provide a basis for predicting injury risk, interpreting differences in criteria values in a clinically meaningful is not always straightforward. There is typically a non-linear relationship between criteria and risk of injury. For example, peak linear acceleration can be used to predict risk of mTBI; however, the difference in risk between a 90 and 110 g impact may be much smaller than the difference in risk between a 180 and 200 g impact (Fig. 1). This stems from the fact that a person is much more likely to experience mTBI when subjected to impacts at this more severe level than at the lower severity [7]. To account for this, injury criteria often need to undergo non-linear transformation to a risk scale in order to improve interpretation of differences.

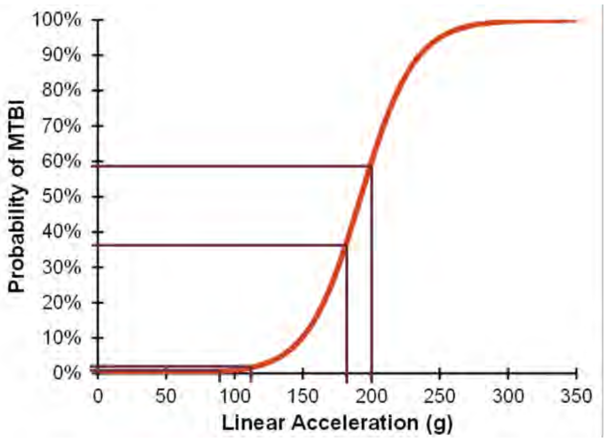

Figure 1: Risk of mTBI as a function of peak linear acceleration. Vertical lines correspond to 20 g increments, demonstrating that equal differences in injury criteria produce vast differences in injury risk.

To predict brain injury risk specific to bicycle helmet evaluation, it is recommended that a multivariate risk function be developed based linear acceleration and rotational velocity. These parameters are related to underlying injury mechanisms and are some of the best-correlated parameters to injury [5,7]. There are, however, several limitations in the ability of current risk functions to predict brain injury risk specific to cyclist head impacts. The first is that existing volunteer data consist almost entirely of football head impacts, and it is probable that these players demonstrate differing tolerances to head impact than the general population [7]. Secondly, football head impacts are not representative of all types of head impacts. A cycling-specific risk curve could be substantially enhanced through collection of cycling-specific head impact data. Additionally, this risk curve could be further refined through incorporating directionality, as it is generally accepted that injury tolerance differs as a function of impact direction [6].

EXAMPLE HELMET RATING PROTOCOL

To assess the ability of a helmet to reduce risk, one must consider the range of impact scenarios that a person might experience in such a helmet. These scenarios can be replicated through a number of laboratory test conditions, which may yield a wide range in injury risk results. The challenge thus becomes how to summarize helmet performance across tests into a single consumer rating. One possible method is the Summation of Tests for the Analysis of Risk (STAR), which has been implemented for football and hockey helmets [8]. This method evaluates helmets through a series of impact tests based on two fundamental concepts: tests are weighted according to how often they occur (exposure), and helmets that lower head kinematics reduce risk (Eq. 1). Exposure is a function of impact location and velocity and is based on distributions of real-world data, while risk is a function of desired kinematic parameters. For bicycle helmets, these may include linear acceleration (a) and rotational velocity (ω).

$$STAR = \sum_{L=1}^{n}\sum_{V=1}^{n}(Exposure(L,V)\cdot Risk(a,\omega)) \qquad (1)$$

The resulting distribution of STAR values across all helmet models evaluated can then be separated into various bins, and these bins assigned a rating in terms of number of stars. For football and hockey, the safest helmets are assigned five stars, descending to zero stars for the least safe helmets. This rating approach is similar to those employed by the New Car Assessment Program (NCAP) or Insurance Institute for Highway Safety (IIHS) for rating automobile safety, and has demonstrated the ability to stimulate improved helmet design.

CONCLUSIONS

Rating helmets based on their ability to reduce injury risk is an effective method for promoting safety in sports and recreation. These ratings should be informed by appropriate brain injury criteria. To-date, many criteria are able to predict injury with reasonable accuracy, although further collection of refined datasets may enhance relevance to particular activities such as cycling. Researchers should select criteria that are mechanistic in nature and evaluate risk based on non-linear transformation methods. To summarize helmet performance into a single rating, impact testing and subsequent weighting of risks should be based on real-world impact exposure data.

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C.6 Remy Willinger (Professor, University of Strasbourg)

Advanced linear and oblique helmet impact test method for consumer tests

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Keywords: Helmet test method, bicyclist and motorcyclist's helmet, oblique impact, brain injury criteria.

INTRODUCTION

It is well known in the scientific community that head rotational acceleration is an essential factor leading to diffuse brain injury or commotion [1]. On the other hand a number of studies demonstrated that the head impact velocity vector has a significant tangential component in addition to the normal or radial component [2, 3]. Several attempts exist in the literature in order to develop new helmet test methods that include the tangential loading of the helmet at the time of impact known as oblique impacts. Despite these efforts no standard exist today that considers the oblique helmet impact in order to assesses the helmet performance under complex linear and tangential impact. The reason may be that no multidirectional brain injury criteria exist. This presentation exposes a multidirectional coupled experimental versus numerical helmet protection assessment method and its application in the context of helmets consumer tests.

ADVANCED HELMET TEST METHOD

Helmets impact conditions

The helmet should be evaluated in order to assess its protection capability under linear and tangential impact.

For the linear impact we suggest to keep the 5.45 m/s impact against a flat anvil. However it would be important to control the pure linearity of the impact by using a HybridIII headform fitted with rotational acceleration. The 6D linear and rotational accelerations under front, occipital and lateral impact should than be recorded versus time. Several accident investigations proved that current head impact angles are between 30° and 60°. For the new tangential impact test it is suggested to submit the helmeted headform to significant tangential loading in order to evaluate its ability to dissipate this kind of energy. Therefore it is suggested to fix the anvil angle at 45° at a 6.0 m/s initial velocity and to conduct three impacts as shown in figure 1.

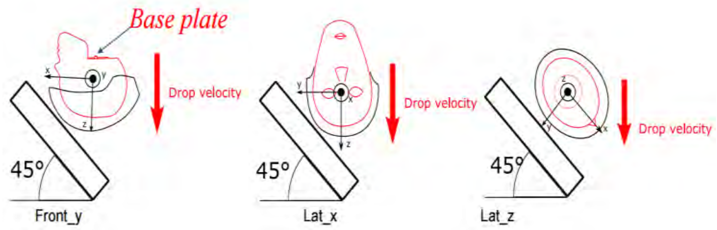

Figure 1 : Illustration of the tangential helmet test impact conditions.

Model based brain injury criteria as pass/fail criteria

For the linear impact, maximum acceleration and HIC could be an option, mainly to verify the skull fracture risk. However, this parameter is not realistically related to brain injury threshold, especially if lateral impact is concerned. It is therefore suggested to introduce the 6D linear and rotational acceleration curves into the brain FE model in order to assess the brain injury risk in a more realistic way. For the linear and the oblique test, it is suggested to introduce the 6D linear and rotational acceleration into the Strasbourg Universtity FE Head Model (SUFEHM) as shown in figure 2, in order to assess the brain injury risk in a more realistic way. The SUFEHM is an advanced brain model that includes a heretogeneous hyperviscoelastic and anisotropic brain constitutive law, which enables it to compute the maximum axon strain for a given impact. This model has been used extensively for the simulation of over 100 real world head trauma in order to derive tissue level brain injury criteria. It has been shown [3] that a moderate brain injury (AIS2) occurs for a maximum axon strain (MAS) od 15%.

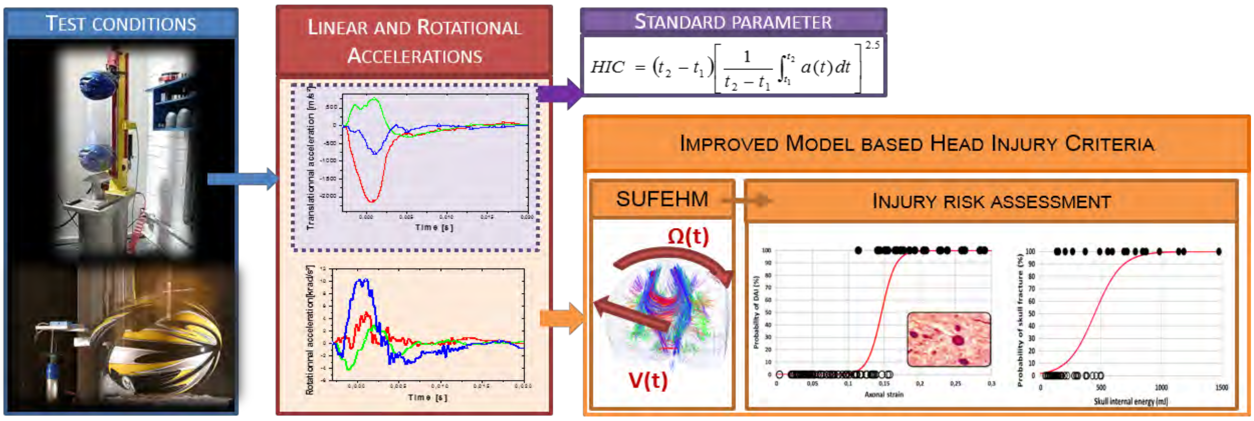

Figure 2 : Illustration of the coupled experimental versus numerical brain injury risk assessment within a new bicycle and motorcycle helmet test method.

HELMET RATING

If our main objective is to contribute to the evolution of international helmet standards, another more short term one is to evaluate helmets comparatively, in order to inform the consumer about the efficiency of the product, but also in order to assist helmet manufacturers to improve their products.

The main originality of the proposed rating system is that it is not just a matter of rating according to a parameter which ranges from the lowest value to the highest value or event to a “mean performance” of a set of helmets, but a rating system which takes into account the brain injury threshold under a multi-directional loading. Therefore the maximum grade (for example 20) is given if injury risk is under 20% and a minimum grade (for example 0) if the brain injury risk is over 100%.

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International Cycling Safety Conference 2017: Cycle Helmet Workshop Report

The safety performance of a cycle helmet is fundamental to protecting cyclists during a fall or collision; however, very little is known about the relative protective qualities of different cycle helmet models. To address this issue, several research institutes have begun to develop cycle helmet testing and assessment programs to rate the relative safety performance of cycle helmets. The Cycle Helmet Safety Workshop aimed to provide an opportunity for global experts to come together to discuss their progress and findings.

The Workshop comprised of three presentation sessions followed by an interactive session. The first presentation session involved a keynote presentation which outlined suggestions for improving current helmet testing standards. The second session focused on global approaches toward testing the safety performance of cycle helmets and required presenters to comment on current best practices used by each research institute and the effects that these approaches had on outcomes. The third presentation session focussed on global approaches toward assessing cycle helmet safety performance and centred around the current assessment philosophies used by each research institute and the effects that these approaches had on outcomes. Abstracts and associated presentation slides are provided within the document.

The interactive session aimed to develop a three-year plan for achieving an evidence-based, successful and sustainable cycle helmet safety consumer information scheme. It provided a platform for discussion around the various benefits/disbenefits surrounding key issues for harmonisation and allowed for the creation of a road map towards Global Harmonisation.

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