# Consumer Testing of Bicycle Helmets

Helena Stigson, Matteo Rizzi, Anders Ydenius, Emma Engström, Anders Kullgren

***Abstract***

Current bicycle helmet standards do not include angular acceleration, for certification even though it is known that it is the dominant cause of brain injury. The objective of this study was to develop an improved test method, including oblique impacts, to evaluate helmets sold on the European market. Four physical tests were conducted, shock absorption with straight perpendicular impact and three oblique impact tests. Computer simulations were made to evaluate injury risk. In total, 17 conventional helmets and one airbag helmet were included. All helmets except five showed a linear acceleration lower than 180 g, which corresponds to a low risk of skull fracture. The airbag helmet performed three times better than the conventional helmets (48 g vs. an average of 175 g). The simulations indicated that the strain in the grey matter of the brain during oblique impacts varied between helmets from 6% to 44%, where 26% corresponds to 50% risk for a concussion. The lowest strain was measured in the brain when the airbag helmet was tested. Helmets equipped with Multi-directional Impact Protection System (MIPS) performed better than the others. However, all helmets need to reduce rotational acceleration more effectively. A helmet that meets the current standards does not necessarily prevent concussion.

***Keywords*** Angular acceleration, Bicycles, Head injury, Helmets, Oblique impact tests

## I. 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

Associate Professor H. Stigson (email: helena.stigson@folksam.se tel: +46 70 831 62 04,), M. Rizzi PhD, A. Ydenius PhD, E. Engström PhD and Professor A. Kullgren are all the authors are affiliated with Folksam Research, Sweden. Stigson, Rizzi and Kullgren are also affiliated with the Department of Applied Mechanics at Chalmers University of Technology, Sweden. Stigson is also associated with the Division of Insurance Medicine, Department of Clinical Neuroscience, Karolinska Institutet, Sweden.

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.

## II. 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.

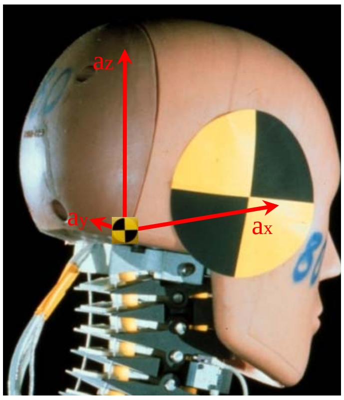


Fig. 2. Translational acceleration. Note that there was no neck in the tests.

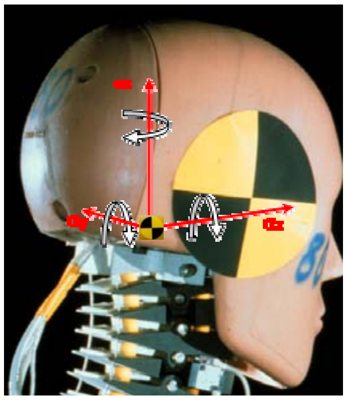

Fig. 3. Rotational acceleration.

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

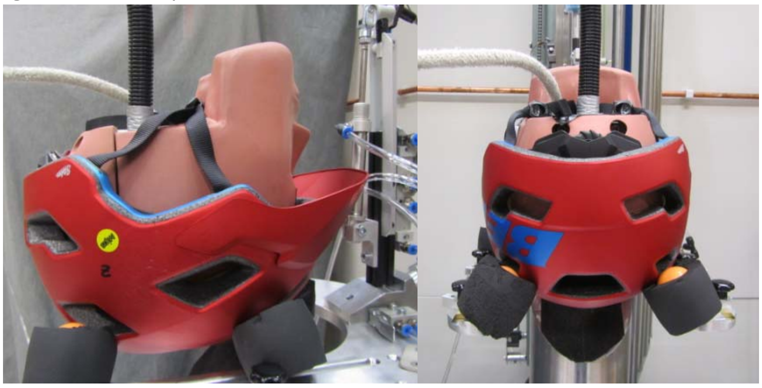

Fig. 5. The oblique test B with rotation around Y-axis.

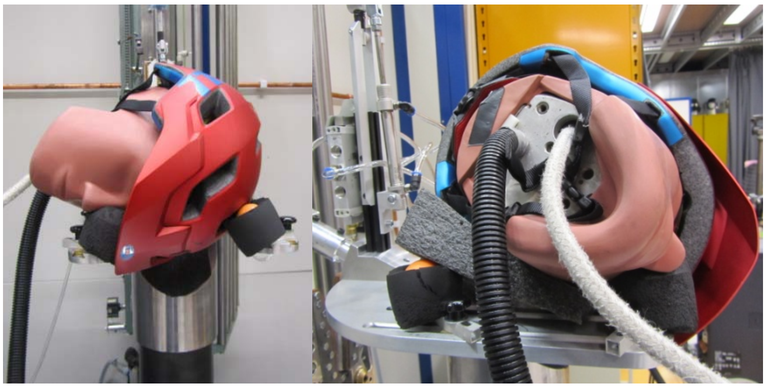

Fig. 6. The oblique test C with rotation around Z-axis (pre-positioned 20° in X and 35° in Z).

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].

## III. 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 |

## IV. 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/s$^2$) to 44% (18.1 krad/s$^2$), 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.

## V. 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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