Importance of the Bicycle Helmet Design and Material for the Outcome in Bicycle Accidents

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ABSTRACT

In Sweden the most common traffic group that needs to be hospitalized due to injury is cyclists where head injuries are the most common severe injuries. According to current standards, the performance of a helmet is only tested against radial impact which is not commonly seen in real accidents. Some studies about helmet design have been published but those helmets have been tested for only a few loading conditions. Therefore, the purpose of this study was to use finite element models to evaluate the effect of the helmet's design on the head in some more loading conditions.

A detailed head model was used to evaluate three different helmet designs as well as non-helmet situations. The first helmet (Baseline Helmet) was an ordinary helmet available on the market. The two other helmet designs were a modification of the Baseline helmet with either a lower density of the EPS liner (Helmet 1) or a sliding layer between the scalp and the EPS liner (Helmet 2). Four different impact locations combined with four different impact directions were tested.

The study showed that using a helmet can reduce the peak linear acceleration (85%), peak angular acceleration (87%), peak angular velocity (77%) and peak strain in the brain tissue (77%). The reduction of the strain level was dependent on the loading conditions. Moreover, in thirteen of the sixteen loading conditions Helmet 2 gave lowest peak strain.

The alteration of the helmet design showed that more can be done to improve the protective effect of the helmet. This study highlighted the need of a modification of current helmet standard test which can lead to helmets with even better protective properties as well as some challenges in implementing new test standards.

Keywords: bicycle, helmet design, head injuries, finite element analysis.

1 INTRODUCTION

Cyclists are one of the road user group that are least protected in road traffic. In Sweden, the cyclists are the largest group of the severely injured in traffic accidents [1]. Several studies [2]–[4] have shown that injuries to the extremities and head are most common in bicycle accidents. Bicycle helmet is one of the protections that cyclists can use. In countries with compulsory helmet laws, such as Australia and New Zealand, the percentage of helmet usage is high, above 75% [5], [6]. At the meantime some other high income countries have a very low helmet use rate, e.g. Belgium with 3% [7]. This despite the fact that several epidemiological, experimental and computational studies have shown the protective effect of bicycle helmets [5], [8]–[14].

Today's bicycle helmets in the European Union are certified according to EN1078:2012 [15] that includes shock absorption tests. In the shock absorption test an aluminium headform and the helmet is dropped vertically at a speed of 5.4 m/s against both a flat and a kerbstone-shaped anvil. In the test, the linear acceleration is measured and to pass the test the resultant linear acceleration should not exceed 250g [15].

As only the linear acceleration is measured, in the current test method, there is a risk that the helmets are mainly optimized to reduce the risk of skull fracture and not brain injuries. This since several studies have shown that the brain is more sensitive to angular motion [16]–[18]. The few computational studies [19]–[22] that have evaluated head impacts in bicycle accidents have shown that pure radial impacts are rare. Bourdet et al. [20] presented reconstructions using the rigid body simulation tool MADYMO and showed that the average head impact velocity computed from 24 real accidents with head Abbreviated Injury Scale (AIS) ranging from 0-5 was 6.8 m/s at an angle of 60 degrees against a passenger car. Verschueren [21] showed the results of 22 accident reconstructions where the average impact velocity was 6.9 m/s and an impact angle of 45 degrees to the ground or the passenger car. Otte and Haasper [23] have studied the German In-Depth Accident Study (GIDAS) database and showed that out of 2979 bicycle accidents, 65 % had impacted a passenger car and only 14% were defined as falls. A similar study was performed in France [8] where 13 797 bicycle accidents were retrieved in the Rhône road trauma registry from 1998 to 2008 where 21% of them involved a motor vehicle and 65% were defined as single accidents. An even higher proportion, 77% single accidents, was found when evaluating the hospital records in Sweden between 2003 and 2012 [2].

A few finite element (FE) studies have been performed to evaluate different types of helmet designs. Forero Rueda and Gilchrist [24] have evaluated the influence of helmet shell and geometrical factors in equestrian helmets. They have shown that the shell stiffness could give a large difference in peak linear acceleration. In a drop test with a velocity of 7.7 m/s, the peak linear acceleration was 316g for a 50 GPa material and 1996g for a 2 GPa material but with a drop velocity of 4.4 m/s the peak was highest in the 50 GPa material (200g) and lowest in the 2 GPa material (134g). Mills and Gilchrist [25] have tested different bicycle helmet designs and concluded that increased thickness of the EPS liner together with lower density could increase the protection effect of the helmet. Hansen et al. [26] have evaluated a new material for the liner in bicycle helmets and found a reduction in peak linear acceleration with 14% and angular acceleration with 34%. Asiminei et al. [27] found similar results with a reduction in head linear acceleration by more than 80% with an anisotropic foam. In these previous studies the helmet design has been evaluated with just a few loading conditions but a bicycle accident can results in many different impact conditions.

Despite all studies mentioned above, the overall protection properties are not fully understood. In order to understand the protective potential of a helmet all possible impact situations should be investigated. The purpose of this study was to investigate a few more possible impact situations than studied before. Both the potential of the helmet and the effect of different helmet designs were evaluated. The tool was an advance FE model of the human head in order to analyse the energy transferred to the brain.

2 METHODS

2.1 The Head Model

A detailed FE head model developed at KTH Royal Institute of Technology was used in this study. The model consists of the scalp, skull, brain, meninges, cerebrospinal fluid, eleven pairs of the largest parasagittal bridging veins and a simplified neck including an extension of the brain stem into the spinal cord. More details about the model can be found in previous publications [28], [29]. The model has both been compared to cadaver experiments [30]–[32] and real accidents [28], [33]–[36].

2.2 The Helmet Models

In this study, three different helmet designs were tested. A FE model of a helmet available on the market was built in a previous study [14]. A detailed description of the helmet model and the validation can be found in the previous publication. A brief summary of the helmet model is described here for clarity.

The helmet was modelled with an outer shell consisted of elastic shell elements, a EPS liner modelled with hexahedral elements and material model MAT_MODIFIED_HONEYCOMB in LS Dyna [37], and a neck retention system made of elastic shell elements. This helmet model is henceforth referred to as the Baseline helmet.

The two other models were developed from the Baseline model. The first altered design, called Helmet 1, was created by reducing the density of the EPS liner. This was done by reducing the compression and shear stress by 25% and increasing the normal strain with 25% in the material curves, see Figure 1. The material constants were adjusted as shown in Table 1. In the second altered helmet design, called Helmet 2, the Baseline helmet had an additional low-friction layer (friction coefficient equal to 0.2) between the inner surface of the EPS liner and the scalp. The extra layer was modelled with shell elements with a thickness of 0.8 mm and an elastic material model (density 1160 kg/m^3, Young's modulus 1.6 GPa and Poisson's ratio 0.45).

Table 1. Material constants for the EPS liner of Helmet 1.

	Density [kg/m^3]	Young's modulus [MPa]	Poisson's ratio [-]	Yield stress for fully compacted honeycomb [MPa]	Young's modulus unloading [MPa]	Shear modulus unloading [MPa]
Baseline Helmet/Helmet 2	86	38	0.05	10	40	25
Helmet 1	65	38	0.05	10	30	19

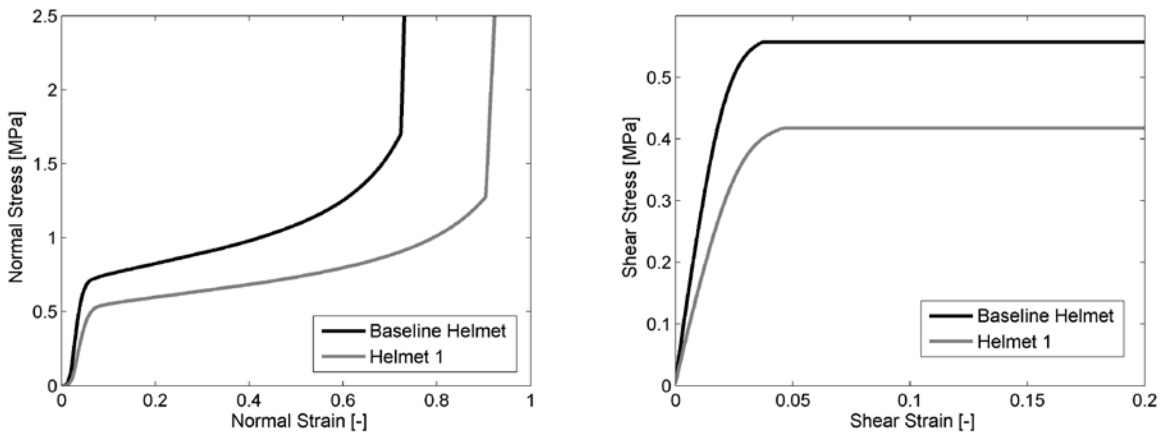

Figure 1. Material curves for Baseline/Helmet 2 and Helmet 1.

2.3 The Loading Conditions

Four different impact locations on the helmet were tested in the present study: Crown, Front, Rear and Side (Figure 2). For all impact locations, four different impact directions were applied. Table 2 shows the different loading directions related to Figure 2. A resultant velocity of 6.4 m/s and an impact angle of 45 degrees were chosen except for a strictly vertical impact. The velocity was applied strictly vertical as in the current helmet test standards or a vertical component and a horizontal direction in either x- or y-direction to account for different impact situations. The resultant velocity and impact angle were chosen to be within the range of previous reconstruction studies that have investigated head impact velocity in bicycle accidents [19]–[21].

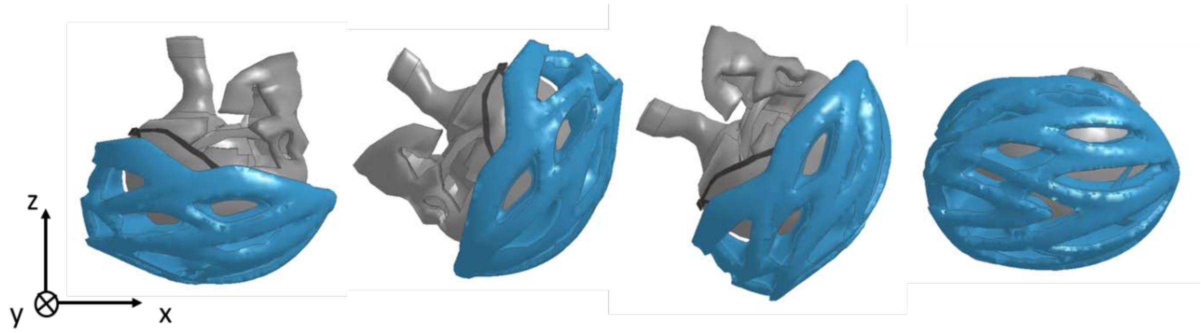

Figure 2. The four impact locations, from left to right: Crown, Front, Rear and Side.

Table 2. The different loading directions.

Impact Direction	V_x [m/s]	V_y [m/s]	V_z [m/s]	Impact Angle [degrees]
1	-4.5	0	-4.5	45
2	4.5	0	-4.5	45
3	0	-4.5	-4.5	45
4	0	0	-6.4	90

In addition to the comparison between the different helmet designs, all the simulations were also performed without a helmet. A total of 64 impact scenarios were simulated. For all simulation the LS Dyna software (version 971 revision 5.1.1) was used. The friction coefficient between the ground and the helmet/scalp as well as between the scalp and helmet was set to 0.5. The impacted surface was a hard surface, such as concrete, and was assumed to be rigid.

The effect on the head was evaluated with the 1st principal Green-Lagrange peak strain of the elements within the brain together with the peak values for the resultant linear acceleration, resultant angular acceleration as well as the resultant angular velocity. All kinematics was filtered with a SAE 180 filter.

A sensitivity study of impact location was also performed for the impact location ”Side”. The helmet and head model were rotated +15, -15, -30 and -45 degrees from the initial position around the head’s vertical axis (Figure 3) and impact direction “2” was applied.

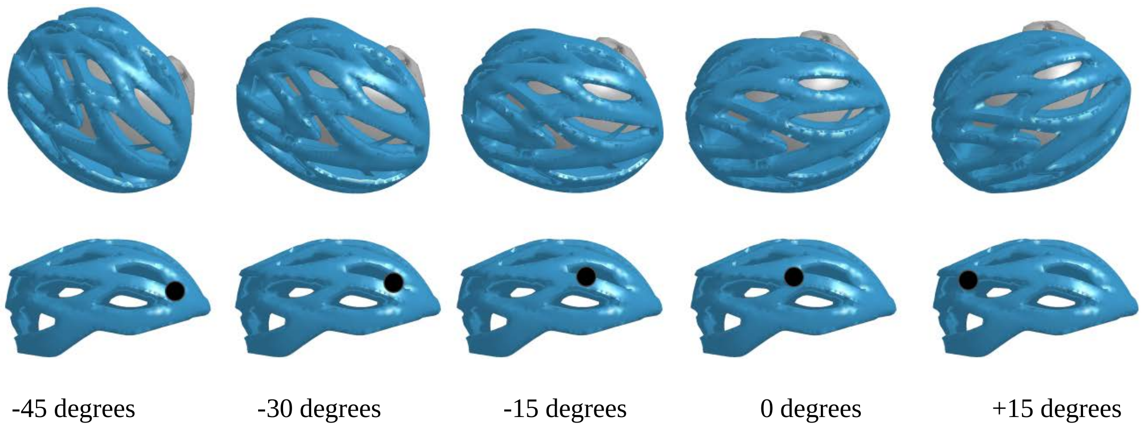

Figure 3. The initial position of the helmet and initial impact point in the sensitivity study.

3 RESULTS

Figure 4 shows an example of the resultant linear acceleration, resultant angular acceleration, resultant angular velocity and maximum 1st principal Green-Lagrange strain in the brain over time for impact location “Front” and impact direction “3”. The impact without a helmet resulted in considerably higher acceleration amplitudes. However, the duration was longer for the impacts including a helmet.

Figure 5 to Figure 8 summarize the maximum amplitude values for all 64 impact scenarios. The peak linear acceleration ranged between 428-576g for the non-helmet impacts, 71-268g for the Baseline helmet, 69-245g for Helmet 1 and 68-281g for Helmet 2. The highest peak value was found for impact direction “4” (strictly vertical velocity) as seen in Figure 5.

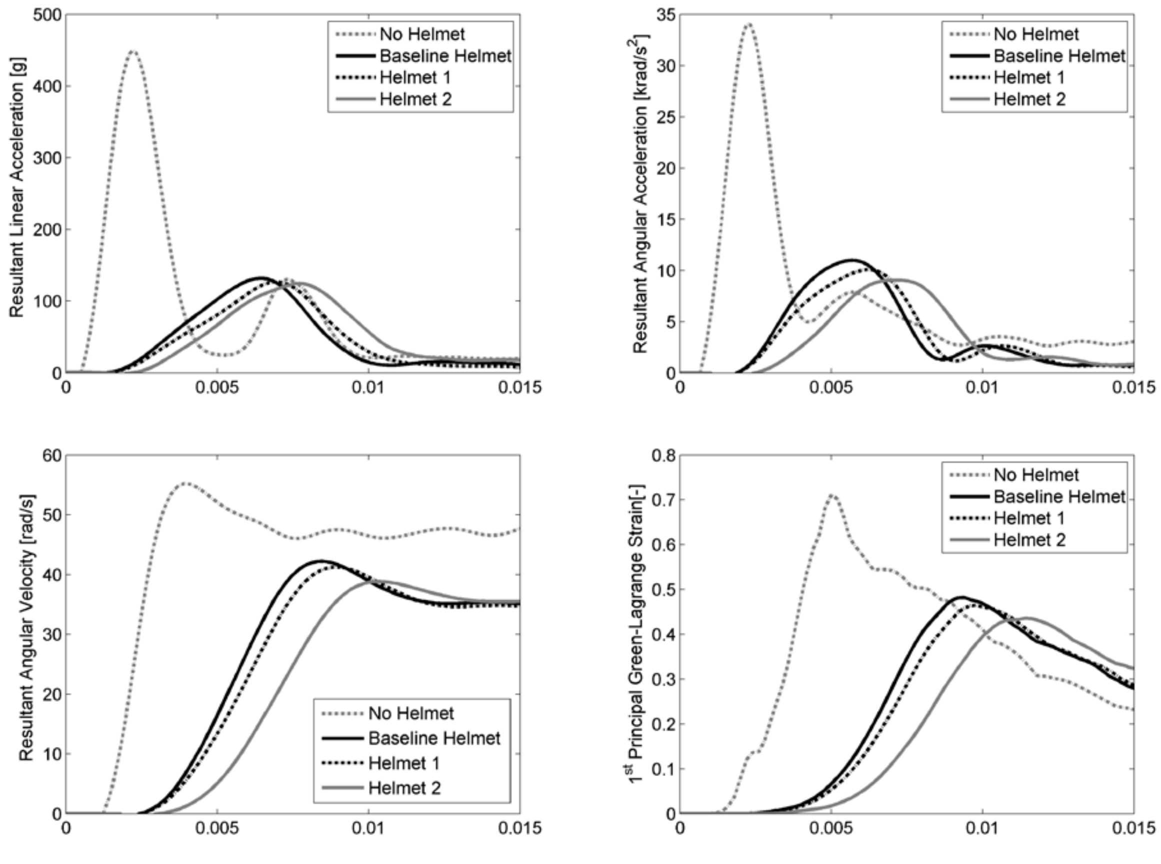

Figure 4. Example of the kinematics and strain over time for “Front 3”.

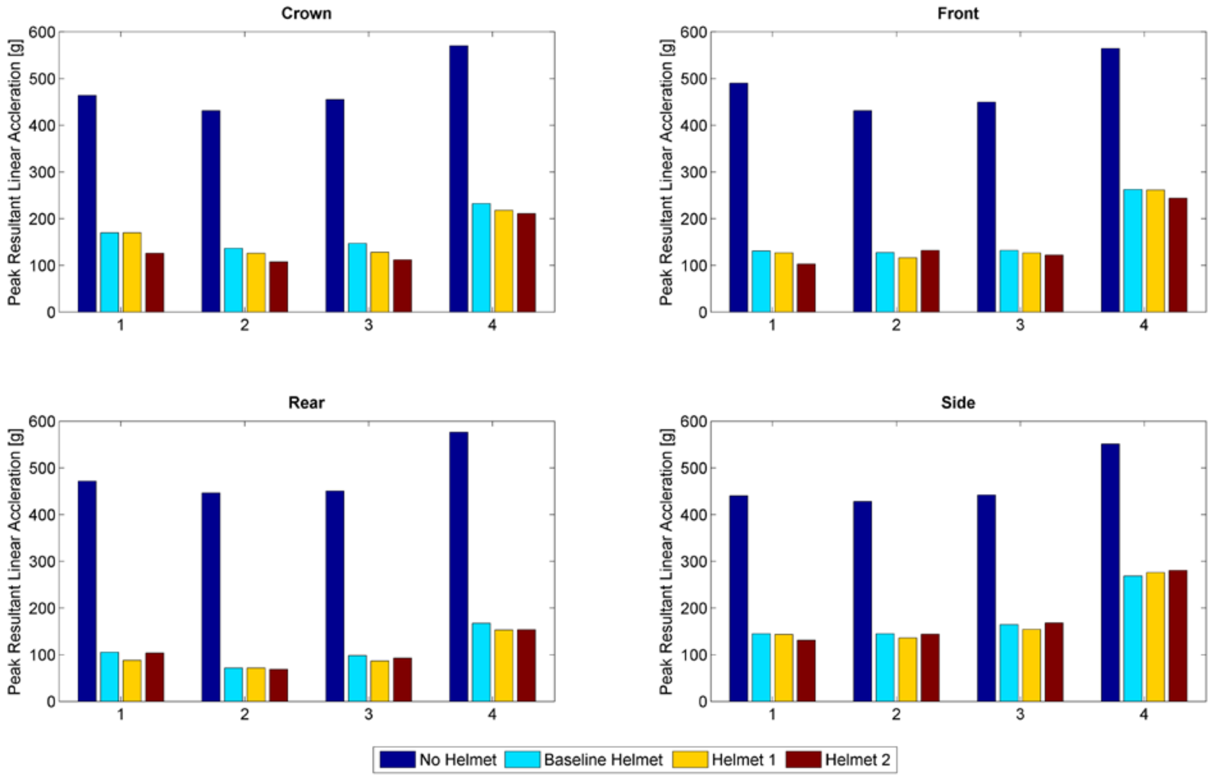

Figure 5. Peak resultant linear acceleration for the different loading conditions.

The peak angular acceleration ranged between 5.1-35.5 krad/s^2 for the non-helmet impacts, 2.0-12.5 krad/s^2 for the Baseline helmet, 1.8-11.3 rad/s^2 for Helmet 1 and 1.9-12.4 krad/s^2 for Helmet 2 (Figure 6). For the non-helmet and all three helmet designs the impact location "Side" and impact direction "2" gave highest peak angular acceleration.

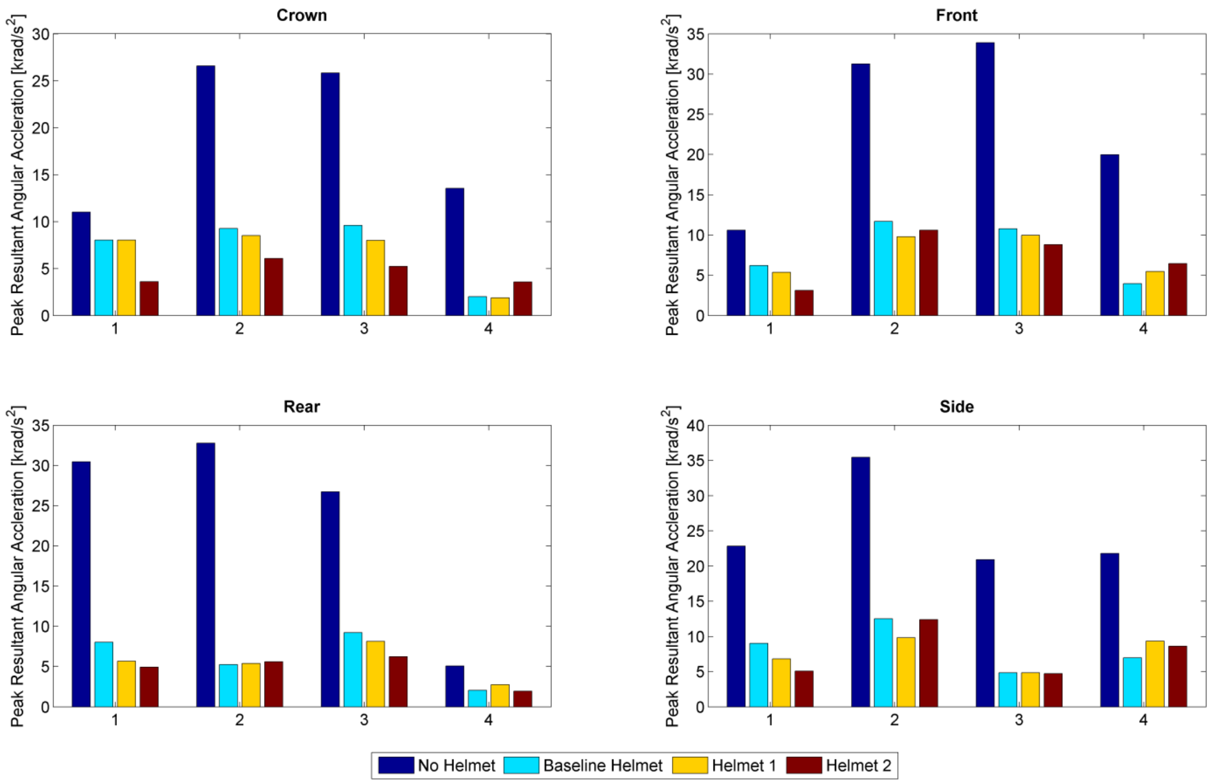

Figure 6. Peak resultant angular acceleration for the different loading conditions.

The peak angular velocity varied between 7.9-64.0 rad/s for non-helmet situations, 4.1-44.6 rad/s for Baseline helmet, 3.6-43.3 rad/s for Helmet 1 and 1.8-38.8 rad/s for Helmet 2 (Figure 7). The highest peak angular velocity was found for impact location "Side" and impact direction "2" for all helmet designs except Helmet 2 where the highest peak value was found for impact location "Front" and impact direction "3".

For all impact location and impact direction the strain varied between 0.15-0.78 for the non-helmet situations, 0.06-0.53 for Baseline helmet, 0.05-0.49 for Helmet 1 and 0.05-0.49 for Helmet 2 (Figure 8). The highest strains were found for the "Side" and "Front" impact location for the non-helmet cases. For the Baseline helmet the "Front" close followed by the "Rear" and "Side" impact location had the highest peak strain. For both Helmet 1 and Helmet 2 the highest peak strain was found for the "Side" impact location. As seen in Figure 8 the strain had also a variation within the different impact locations.

The peak strain for the different impact locations in the "Side" impact is shown in Figure 9. The peak value ranged from 0.30-0.78 for non-helmet, 0.21-0.59 for Baseline helmet, 0.25-0.53 for Helmet 1 and 0.17-0.54 for Helmet 2.

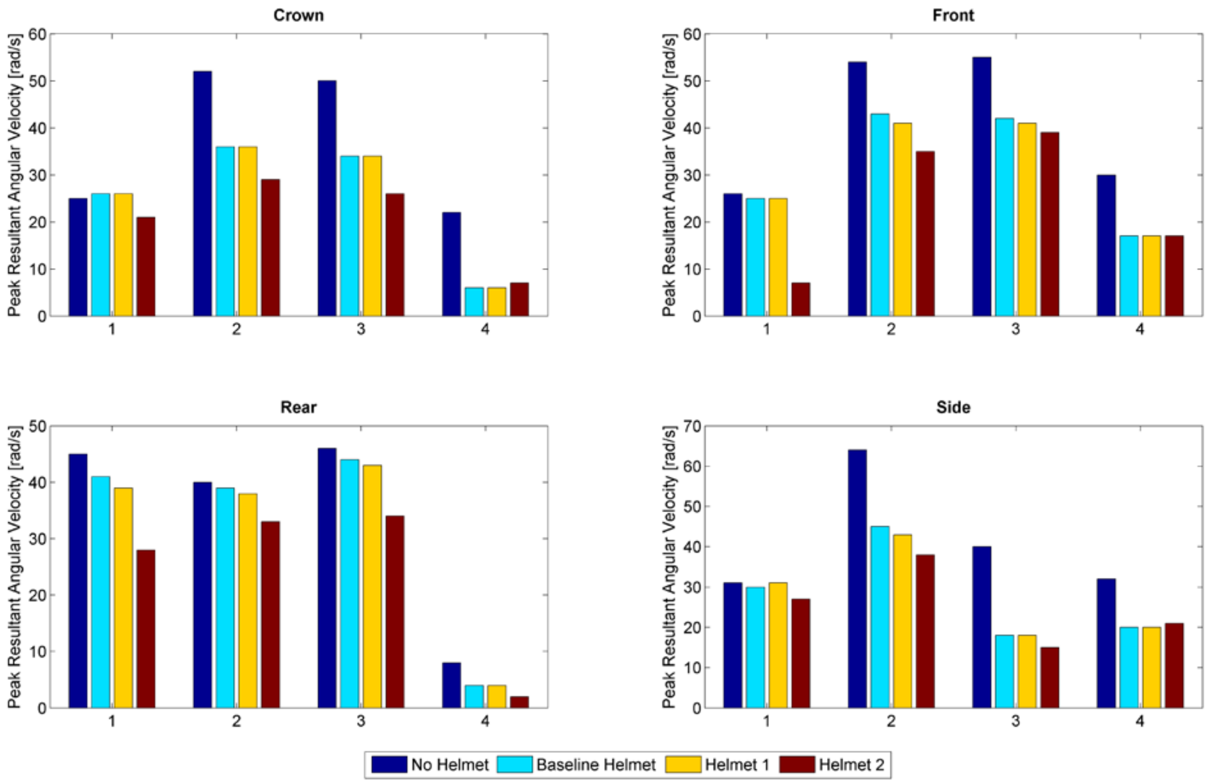

Figure 7. Peak resultant angular velocity for the different loading conditions.

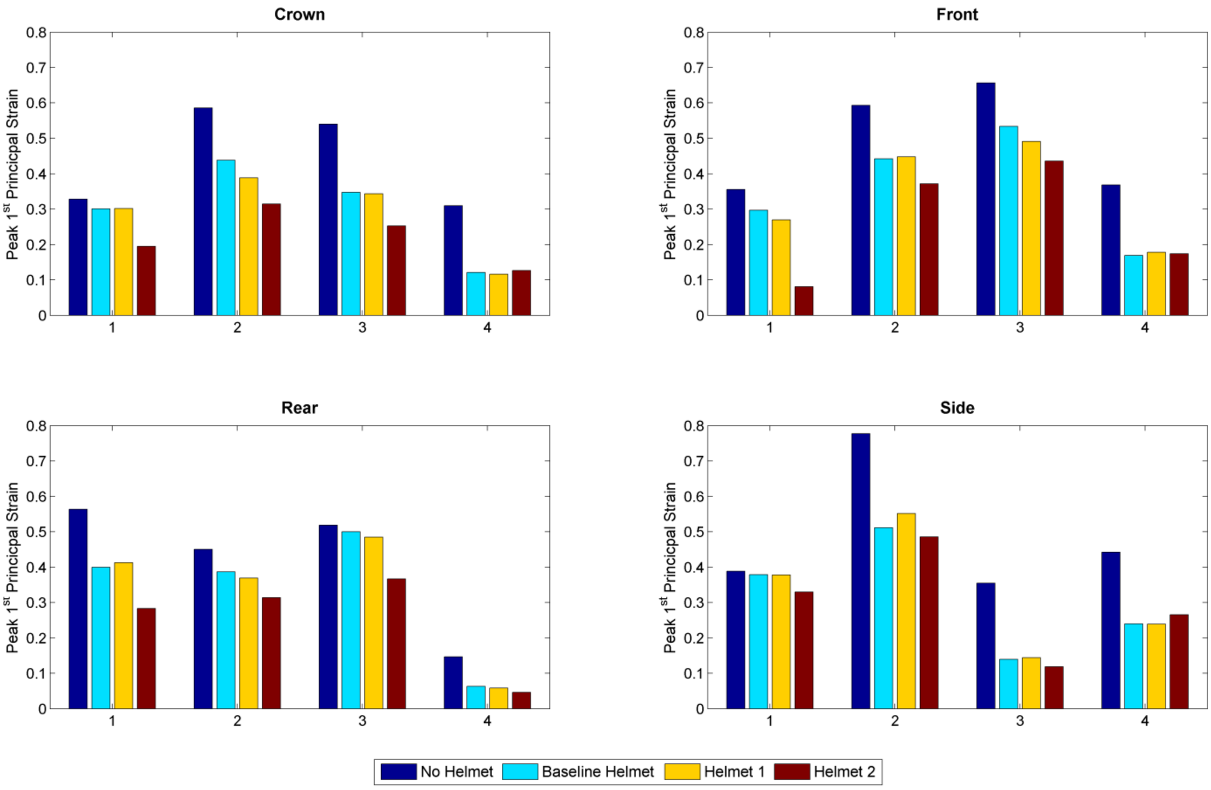

Figure 8. Peak 1st principal strain for the different loading conditions.

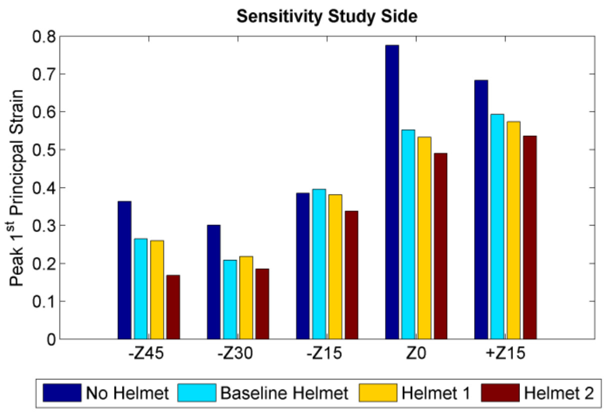

Figure 9. The peak 1st principal strain in the sensitivity study.

4 DISCUSSION

This study has used a validated FE model of the human head and bicycle helmets to investigate how different structural designs can affect the protective ability of a helmet. Several studies have shown that a helmet considerably has reduced the risk of skull and brain injuries [8], [9], [11], [13], [14]. Some previous studies have also studied the influence of helmet design but only with a few loading conditions whereas in this study 20 different loading conditions were evaluated. In the present study, the results of the computed global kinematics showed a considerable reduction of the peak linear and angular accelerations when comparing impacts with and without a helmet. The linear accelerations were reduced between 49% and 85% for the different impact conditions. The angular acceleration was reduced between 27% and 87%.

When evaluating the peak linear acceleration for the different helmet configurations, smaller differences were seen compared to peak angular acceleration and peak angular velocity. The linear acceleration had a variation up to 26% for the different helmet designs within the same impact conditions. While angular acceleration had a variation up to 55% and 71% for angular velocity. Hansen et al. [26] also found a smaller reduction in linear acceleration (14%) compared to angular acceleration (34%) when comparing a traditional helmet against a helmet with a modified liner material. The linear acceleration is the only measurement the helmets are evaluated against in today's helmet test standard.

To analyse the protective properties of a helmet the risk of skull and brain injury need to be evaluated. Mertz et al. [38] estimated the risk of skull fracture with peak linear acceleration, 40% risk of fractures for 250g. This threshold was exceeded in all impacts without a helmet with minimum peak linear acceleration of 428g. While for the impacts with a helmet only two simulated impacts had a higher linear acceleration than 250g (impact location "Front" and "Side" at impact condition "4"). The risk for a brain injury like concussion or DAI has earlier been shown to correlate well to the maximum principal strain in experiments and FE simulations [28], [39]–[41]. A significant reduction of peak strain was seen in most loading conditions for all three helmet designs in this study. However, there are a few loading conditions where the computed strain in the FE model of the head showed just a few percentage of reduction using a helmet. In these simulations it could be seen that in the impact situations that the non-helmeted head more or less bounced without much rotation. The reason for this is that the rotation caused by the tangential force in these impact conditions were eliminated by the gravity force acting in the center of gravity of the head. However, in these impacts where the helmet did not reduce the strain and the risk of a brain injury, it considerably reduced the linear accelerations and therefore the risk of a skull fracture.

In thirteen of the sixteen loading conditions Helmet 2 had the lowest peak strain. The loading conditions where Helmet 2 did not have highest reduction of peak strain were strictly vertical for "Crown", "Front" and "Side". The reduction in peak strain was between 15-77% for Helmet 2 compared to Baseline Helmet 2-61% and Helmet 1 2-65%. The peak strain in the Baseline helmet and Helmet 1 was relatively close for the majority of loading conditions where the maximum difference in percentage was 12% compared to that between the Baseline Helmet and Helmet 2 which was 72%.

In the present study, only the Baseline helmet has been evaluated against experimental tests. The evaluation showed that the peak linear acceleration was within 15% for four different loading conditions, within 14% for the angular acceleration and 17% for the angular velocity [14]. The density of Helmet 1 was decreased by altering the material data with 25% which gave a density of around 66 kg/m^3. For Helmet 1 the material data was just reduced 25% as mentioned above. This could be improved by using data from material test of an EPS liner with lower density. Helmet 2 had an extra sliding layer between the EPS liner and the scalp which is an example on how the helmet can be design to reduce rotational forces. There are also other solutions for this [26], [27], [42].

The resultant linear acceleration exceeded the threshold for the pass criteria for European helmet standards of 250g in some of the simulations with impact direction "4". This is due to the fact that the impact direction "4" has a higher impact velocity than found in the standard tests, 6.4 m/s compared to 5.4 m/s.

This is a very limited design parameter study. The goal was not to show the best possible helmet design in this study, but to show that there is a potential to improve the helmet if the impact condition is not only pure vertical as in the current helmet test standards. One improvement could be to include other impact directions. Another improvement could be to include rotational components, such as the angular acceleration or velocity, as several studies [16], [17] have shown that the brain is more sensitive to rotation than linear motion.

In this study only one geometrical helmet design was evaluated. The helmet geometry in the current model has a relatively irregular geometry due to the many ventilation holes. The irregular geometry could make the helmet more sensitive to impact location. In this study the impact locations were kept constant for three of the four different impact points. The sensitivity study of the "Side" impact location with a positive x-velocity and negative vertical velocity showed a large spread in the computed peak strain for the different impact locations, e.g. 0.17-0.53 for Helmet 2. This implies that the impact location is of great importance and must be considered when testing helmets. Several studies have evaluated impact marks on the helmet after bicycle accidents and found that a high percentage of the impacts were to the front and side [13], [43]–[46].

The impact velocity, like the impact location, can vary with a large range in bicycle accidents. In the present study only one resultant velocity of 6.4 m/s was evaluated but with four different loading situations. Verschueren [21] have performed reconstructions of bicycle accident either involving an impacted object or a single fall and found that the resultant velocity varied between 3 to 15 m/s. Bourdet et al. [19] have performed a parameter study of bicycle accident in either 20 or 40 km/h. They found an average resultant impact velocity of 6.9 km/h and 6.4 km/h dependent on accident situation (impacting a curb or skidding) in bicycle velocity 20 km/h and for 40 km/h either 11.3 m/s or 9.1 m/s. The impact angle was between 34 and 58 degrees.

There are some limitations concerning the helmet-head interaction because no chin strap or comfort foam was modelled. When studying the simulations and the peak values then it seems like the chin strap should give minor changes to most of the results since the chin loose contact with the strap at impact. However, the comfort foam and the fit of the helmet could affect the results. McIntosh et al. [11] found that the angular acceleration in the coronal plane had an average of 13.9 rad/s^2 and 9.9 rad/s^2 for a loose retention and a tight retention respectively.

Another limitation with this study was that the simulations were performed with a head model not including the rest of the body. Some previous studies [21], [47]–[49] have evaluated the difference between including the rest of the body or not, but no general conclusion can be drawn. Ghajari et al. [47] evaluated one impact situation with a motorcycle helmet with and without the body. They found that the peak angular acceleration was altered up to 40% and both the linear acceleration and crushing distance of the liner was altered. Forero Rueda [49] performed jockey accident reconstructions with MADYMO and found that the head acceleration pulses was too long in this accident situations to ignore the body but that should be investigated further with a more detailed model of the neck. Verschueren [21] evaluated the influence of the neck and the rest of the body in 9 bicycle accident situations with the MADYMO model. He found that the neck and the body influenced the impact force and angular acceleration in some of the cases.

This study demonstrated the limitations with the present helmet test standard where only the radial impacts are tested and only linear acceleration is measured. A new helmet test standard could lead to even better helmets. But this study also highlights some challenges in implementing a new test standard such as choosing the impact location and direction.

5 CONCLUSION

The conclusions from this numerical study of 20 different impact configurations for a non-helmet head compared to three possible helmets designs are:

- The helmet considerably absorbs energy for all impact configurations and reduces the risk of injuries to the skull and the brain.
- Impacts with an angled impact direction gave a higher risk of brain injury compared to a purely radial impact for most impact locations on the helmet.
- There are larger differences in head kinematics and strain levels between helmets with different structural designs in the angled impacts than in pure vertical impacts.
- The helmet design could be improved from today's helmet design to protect the head more.
- As well as the helmet test methods could be improved in order to distinguish between different helmet designs.

CONFLICT OF INTREST

Peter Halldin has interest in the helmet company MIPS AB, Sweden.

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REFERENCES

[1] Trafikanalys, “Persons hospitalized due to road traffic accidents 2011.” (2013)

[2] M. Rizzi, H. Stigson, and M. Krafft, “Cyclist Injuries Leading to Permanent Medical Impairment in Sweden and the Effect of Bicycle Helmets,” in Proceedings, *International Research Council on Biomechanics of Injury (IRCOBI) Conference*, Gothenburg, Sweden, 2013, 412–423.

[3] R. Fredriksson and E. Rosén, “Priorities for Bicyclist Protection in Car Impacts – a Real life Study of Severe Injuries and Car Sources,” in Proceedings, *International Research Council on Biomechanics of Injury (IRCOBI) Conference*, Dublin, Ireland, 2012, 779–786.

[4] W. S. Chen, R. Y. Dunn, A. J. Chen, and J. G. Linakis, “Epidemiology of Nonfatal Bicycle Injuries Presenting to United States Emergency Departments, 2001-2008” *Acad. Emerg. Med.* **20** (2013) pp. 570–575.

[5] L. J. Povey, W. J. Frith, and P. G. Graham, “Cycle Helmet Effectiveness in New Zealand” *Accid. Anal. Prev.* **31** (1999) pp. 763–70.

[6] M. H. Cameron, A. P. Vulcan, C. F. Finch, and S. V Newstead, “Mandatory Bicycle Helmet Use Following a Decade of Helmet Promotion in Victoria, Australia - an Evaluation” *Accid. Anal. Prev.* **26** (1994) pp. 325–337.

[7] K. S. Klein, D. Thompson, P. C. Scheidt, M. D. Overpeck, and L. a Gross, “Factors associated with bicycle helmet use among young adolescents in a multinational sample.” *Inj. Prev.* **11** (2005) pp. 288–93.

[8] E. Amoros, M. Chiron, J.-L. Martin, B. Thélot, and B. Laumon, “Bicycle Helmet Wearing and the Risk of Head, Face, and Neck Injury: a French Case–Control Study Based on a Road Trauma Registry” *Inj. Prev.* **18** (2012) pp. 27–32.

[9] D. Thompson, F. Rivara, and R. Thompson, “Helmets for Preventing Head and Facial Injuries in Bicyclists (Review)” *Cochrane Database Syst. Rev.* (1999).

[10] R. G. Attewell, K. Glase, and M. McFadden, “Bicycle Helmet Efficacy: a Meta-Analysis.” *Accid. Anal. Prev.* **33** (2001) pp. 345–52.

[11] A. S. McIntosh, A. Lai, and E. Schilter, “Bicycle Helmets: Head Impact Dynamics in Helmeted and Unhelmeted Oblique Impact Tests” *Traffic Inj. Prev.* **14** (2013) pp. 501–508.

[12] P. A. Cripton, D. M. Dressler, C. A. Stuart, C. R. Dennison, and D. Richards, “Bicycle helmets are highly effective at preventing head injury during head impact: Head-form accelerations and injury criteria for helmeted and unhelmeted impacts” *Accid. Anal. Prev.* **70** (2014) pp. 1–7.

[13] D. Otte and B. Wiese, “Influences on the Risk of Injury of Bicyclists’ Heads and Benefits of Bicycle Helmets in Terms of Injury Avoidance and Reduction of Injury Severity” *SAE Int. J. Transp. Saf.* **2** (2014) pp. 257–267.

[14] M. Fahlstedt, P. Halldin, and S. Kleiven, “The Protective Effect of a Bicycle Helmet – A Finite Element Study” *Manuscript* (2014).

[15] EN1078, “European standard EN1078:2012. Helmets for pedal and for users of skateboards and roller skates.” 2012.

[16] T. A. Gennarelli, L. E. Thibault, and A. K. Ommaya, “Pathophysiologic Responses to Rotational and Translational Acclerations of the Head,” in Proceedings, *Stapp Car Crash Conference*, 1972.

[17] A. H. S. Holbourn, “Mechanics of Head Injuries” *Lancet* (1943) pp. 438–441.

[18] T. A. Gennarelli, L. E. Thibault, G. Tomei, R. Wiser, D. I. Raham, and J. Adams, “Directional dependence of axonal brain injury due to centroidal and non-centroidal acceleration,” in Proceedings, *Proc. 31st Stapp Car Crash Conference*, 1987.

[19] N. Bourdet, C. Deck, R. P. Carreira, and R. Willinger, “Head Impact Conditions in the Case of Cyclist Falls” *Proc. Inst. Mech. Eng. Part P J. Sport. Eng. Technol.* **226** (2012) pp. 282–289.

[20] N. Bourdet, C. Deck, T. Serre, C. Perrin, M. Llari, and R. Willinger, “In-Depth Real-World Bicycle Accident Reconstructions” *Int. J. Crashworthiness* **19** (2014) pp. 222–232.

[21] P. Verschueren, “Biomechanical Analysis of Head Injuries Related to Bicycle Accidents and a New Bicycle Helmet Concept”, PhD thesis, Katholieke Universiteit Leuven, 2009.

[22] D. S. McNally and S. Whitehead, “A Computational Simulation Study of the Influence of Helmet Wearing on Head Injury Risk in Adult Cyclists” *Accid. Anal. Prev.* **60** (2013) pp. 15–23.

[23] D. Otte and C. Haasper, “Effectiveness of the Helmet for Bicyclists on Injury Reduction in German Road Accident Situations – State of Affairs on GIDAS” *Int. J. Crashworthiness* **15** (2010) pp. 211–221.

[24] M. A. Forero Rueda and M. D. Gilchrist, “Computational analysis and design of components of protective helmets” *Proc. Inst. Mech. Eng. Part P J. Sport. Eng. Technol.* **226** (2012) pp. 208–219.

[25] N. J. Mills and A. Gilchrist, “Bicycle helmet design” *Proc. Inst. Mech. Eng. Part L J. Mater. Des. Appl.* **220** (2006) pp. 167–180.

[26] K. Hansen, N. Dau, F. Feist, C. Deck, R. Willinger, S. M. Madey, and M. Bottlang, “Angular Impact Mitigation System for Bicycle Helmets to Reduce Head Acceleration and Risk of Traumatic Brain Injury” *Accid. Anal. Prev.* **59** (2013) pp. 109–17.

[27] A. Asiminei, G. Van Der Perre, I. Verpoest, and J. Goffin, “A Transient Finite Element Study Reveals the Importance of the Bicycle Helmet Material Properties on Head Protection during an Impact,” in Proceedings, *International Research Council on Biomechanics of Injury (IRCOBI) Conference*, York, UK, 2009, 357–360.

[28] S. Kleiven, “Predictors for Traumatic Brain Injuries Evaluated through Accident Reconstructions” *Stapp Car Crash J.* **51** (2007) pp. 81–114.

[29] M. Fahlstedt, P. Halldin, S. Kleiven, B. Depreitere, and J. Vander Sloten, “Biomechanical Reconstructions of Traumatic Brain Injuries in Bicycle Accidents – Correlation between Injury Pattern and FEA” *Manuscript* (2014).

[30] S. Kleiven, “Evaluation of head injury criteria using a finite element model validated against experiments on localized brain motion, intracerebral acceleration, and intracranial pressure” *Int. J. Crashworthiness* **11** (2006) pp. 65–79.

[31] S. Kleiven and W. N. Hardy, "Correlation of an FE Model of the Human Head with Local Brain Motion-Consequences for Injury Prediction." *Stapp Car Crash J.* **46** (2002) pp. 123–144.

[32] S. Kleiven and H. Von Holst, "Consequences of Head Size Following Trauma to the Human Head" *J. Biomech.* **35** (2002) pp. 153–160.

[33] D. Patton, A. S. McIntosh, and S. Kleiven, "The Biomechanical Determinants of Concussion: Finite Element Simulations to Investigate Brain Tissue Deformations during Sporting Impacts to the Unprotected Head" *J. Appl. Biomech.* **29** (2013) pp. 721–730.

[34] H. von Holst and X. Li, "Numerical impact simulation of gradually increased kinetic energy transfer has the potential to break up folded protein structures resulting in cytotoxic brain tissue edema" *J. Neurotrauma* **30** (2013) pp. 1192–9.

[35] J. Mordaka, S. Kleiven, M. V. S. Nooij, and R. De Lange, "THE IMPORTANCE OF ROTATIONAL KINEMATICS IN PEDESTRIAN HEAD TO WINDSHIELD IMPACTS .," in Proceedings, *International Research Council on Biomechanics of Injury (IRCOBI) Conference*, 2007, 83–94.

[36] M. Fahlstedt, K. Baeck, P. Halldin, J. Vander Sloten, J. Goffin, B. Depreitere, and S. Kleiven, "Influence of Impact Velocity and Angle in a Detailed Reconstruction of a Bicycle Accident," in Proceedings, *International Research Council on Biomechanics of Injury (IRCOBI) Conference*, Dublin, Ireland, 2012, 787–799.

[37] Livermore Software Technology Corporation, *LS DYNA KEYWORD USER'S MANUAL Volume II Material Models*, **I**. 2014.

[38] H. J. Mertz, P. Prasad, and A. L. Irwin, "Injury Risk Curves for Children and Adults in Frontal and Rear Collisions," in Proceedings, *Stapp Car Crash Conference*, 1997, 13–30.

[39] A. C. Bain and D. F. Meaney, "Tissue-Level Thresholds for Axonal Damage in an Nervous System White Matter Injury" *J. Biomech. Eng.* **122** (2000) pp. 615–622.

[40] B. Morrison, H. L. Cater, C. C. B. Wang, F. C. Thomas, C. T. Hung, G. a Ateshian, and L. E. Sundstrom, "A tissue level tolerance criterion for living brain developed with an in vitro model of traumatic mechanical loading." *Stapp Car Crash J.* **47** (2003) pp. 93–105.

[41] T. A. Gennarelli, L. E. Thibault, R. Tipperman, G. Tomei, R. Sergot, M. Brown, W. L. Maxwell, D. I. Graham, J. H. Adams, A. Irvine, L. M. Gennarelli, A. C. Duhaime, R. Boock, and J. Greenberg, "Axonal Injury in the Optic Nerve: a Model of Diffuse Axonal Injury in the Brain" *J. Neurosurg.* **71** (1989) pp. 244–253.

[42] PhillipsHelmets, "Phillips Head Protection System," 2013. [Online]. Available: http://www.phillipshelmets.com/. [Accessed: 28-Oct-2013].

[43] T. A. Smith, D. Tees, D. R. Thom, and H. H. Hurt Jr, "Evaluation and Replication of Impact Damage to Bicycle Helmets" *Accid. Anal. Prev.* **26** (1994) pp. 795–802.

[44] A. McIntosh, B. Dowdell, and N. Svensson, "Pedal Cycle Helmet Effectiveness: A Field Study of Pedal Cycle Accidents" *Accid. Anal. Prev.* **30** (1998) pp. 161–168.

[45] R. P. Ching, D. C. Thompson, R. S. Thompson, D. J. Thomas, W. C. Chilcott, and F. P. Rivara, "Damage to Bicycle Helmets Involved with Crashes" *Accid. Anal. Prev.* **29** (1997) pp. 555–562.

[46] M. Williams, "The protective performance of bicyclists' helmets in accidents" *Accid. Anal. Prev.* **23** (1991) pp. 119–131.

[47] M. Ghajari, S. Peldschus, U. Galvanetto, and L. Iannucci, “Effects of the Presence of the Body in Helmet Oblique Impacts” *Accid. Anal. Prev.* **50** (2013) pp. 263–271.

[48] COST 327, “Motorcycle Safety Helmets Final report.” (2001)

[49] M. A. Forero Rueda, “Equestrian Helmet Design : A Computational and Head Impact Biomechanics Simulation Approach”, PhD thesis, University College Dublin, 2009.