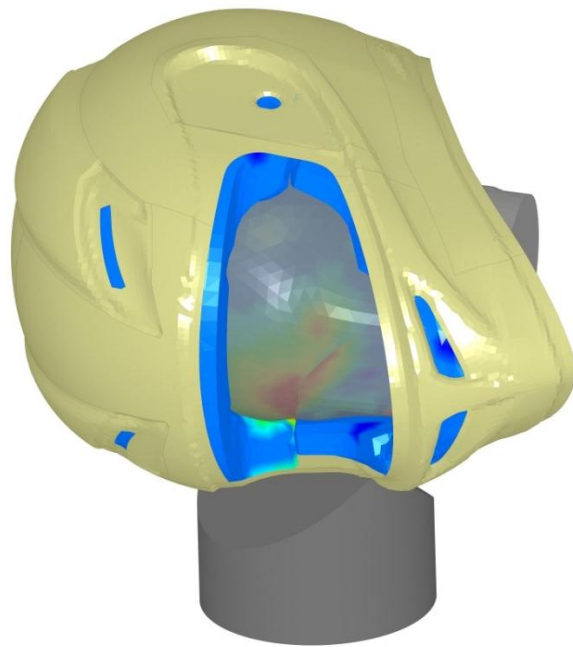


Helmet Performance and Design



Editors: Peter RN Childs, Anthony Bull, Mazdak Ghajari

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Contents

Preface – Anthony Bull, Peter Childs, Mazdak Ghajari

Papers

1. The development of next generation test standards for helmets.
Peter Halldin and Sven Kleiven
2. The impact attenuation test of motorcycle helmet standards.
Mazdak Ghajari, Gaetano Davide Caserta and Ugo Galvanetto
3. Model based head injury criteria for head protection optimization.
Rémy Willinger and Caroline Deck
4. A review of blast induced traumatic brain injury research.
Sami Dabbagh, Imogen Keane, Richard Pangonis and Holly Wilson
5. Speculation on the future of military helmet technology.
Alexander R. Haley
6. Evaluation of blunt impact protection in a military helmet designed to offer blunt & ballistic impact protection.
Peter Halldin, Daniel Lanner, Richard Coomber and Sven Kleiven
7. Finite element modelling of a honeycomb reinforced helmet.
Gaetano Davide Caserta, Mazdak Ghajari, Ugo Galvanetto and Lorenzo Iannucci
8. A comparative study of turbulence models performance for the study of air flow in helmets.
BS Shishodia, S. Sanghi, P Mahajan.
9. Helmet research in the WP3 of the MYMOSA project.
Ugo Galvanetto, David Hailoua Blanco, Gaetano Davide Caserta, Mazdak Ghajari and Alessandro Cernicchi
10. The influence of velocity on the performance range of American football helmets.
Andrew Post, Anna Oeur, T. Blaine Hoshizaki and Michael D. Gilchrist
11. Efficiency of head protection equipment for two mainstream sports – a comparison.
Daniel J. Plant, Timothy R. Hault, Joseph Townsend, James Pedder and P. Shaun J. Crofton
12. Application of an effects database in idea generation approach for helmet design.
Zhihua Wang, Han Kak Lee, Dan McGlaughlin and Peter Childs
13. Applying problem Structuring methods to the design process for safety helmets.
Bruce Garvey and Peter Childs

Abstracts for Presentations without Publication

- A1 An examination of headform dynamic response for concussive and traumatic brain injuries.
Anna Oeur, Clara Karton, Andrew Post, Philippe Rousseau, Blaine Hoshizaki, Shawn Marshall,
Susan Brien, Aynsley Smith and Michael Cusimano
- A2 Impact studies on motor-cycle helmets with different shells.
Puneet Mahajan, Arun Baby and Sanjeev Sanghi
- A3 The Assessment of Inbound Mass Variation on the Distribution of Brain Tissue Deformation.
Clara Karton, Andrew Post, T. Blaine Hoshizaki and Michael D. Gilchrist
- A4 The Patent Landscape for Protective Headgear Technologies.
Robin Walton, Benoit Geurts
- A5 Finite Element Analysis of Helmeted Impact and Head Injury Assessment of a Commercial
Motorcycle Helmet.
Fábio A.O. Fernandes, Ricardo J. Alves de Sousa and Rémy Willinger

Chapter interleaf images courtesy of Hankak Lee, Royal College of Art/Imperial College London

Preface

Helmet performance and design are intimately related, where the technical functional aspects are for protection and safety, yet non-technically functional aspects determine usability and commercial viability. These proceedings from the Helmet Performance and Design conference held at Imperial College London on the 15th February 2013 reflect the various aspects of research that is key in the field. As organisers we are grateful for the contributions from leaders in the discipline who have travelled far and wide to come together to present the state of the art and to debate the future of the field. Presenters have come from North America, Europe and Asia, bringing world-leading expertise in brain injury, biomechanics, forensic analysis, computational mechanics, materials science, and testing and design. The breadth of participants' expertise and background demonstrates that this is a truly inter-disciplinary field with wide applicability in sports, motorcycle and bicycle equipment, military helmets, and other fields including clinical treatment.

Helmet performance cannot be separated from an understanding of head, neck and brain injury due to impact, penetration, or shock, yet designers are also focused on other technical aspects such as aerodynamics and reliability. It is clear that these various aspects result in optimised structures that are suitably tested using standard testing procedures and equipment that is, as yet, still not validated fully with detailed, fidelic, clinical, morphological and physiological data. Therefore, we are happy to see papers with subject matter ranging from this detailed clinical data to developing new standards for physical testing, whilst also encompassing computational testing and validation.

It is apparent that there are certain aspects of performance not yet appropriately codified in standards; it may or may not be appropriate to do so. Examples of these include thermoregulation parameters and parameters of performance related to vision in the military context. All of these are considered and included in these proceedings. The influence of consumer choice and coercion is deliberated in the context of reducing injury, yet how do we also consider the influence of comfort, aesthetics, weight, and thermal characteristics? This final question is addressed by a paper on structured decision making in determining parameters and their relative importance in helmet design.

The Imperial College mission statement places at its heart the application of science and engineering to industry, commerce and healthcare in the context of multidisciplinary working. It is our hope as organisers of this conference that this collection of long and short papers will act as a catalyst to improved helmet design bringing societal benefits in terms of injury reduction, thus fulfilling that mission described above, and fulfilling a wider mission for engineers, medics and scientists to work together for the greater good.

Anthony Bull, Peter Childs, Mazdak Ghajari, February 2013

The Development of Next Generation Test Standards for Helmets.

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ABSTRACT

Injury statistics show that accidents with a head impact often happen with an angle to the impacting object. An angled impact will result in a rotation of the head if the friction is high enough. It is also known that the head is more sensitive to rotation than pure linear motion of the head. CEN has initiated the work to design a new helmet test oblique or angled impact test method a helmet test method that can measure the rotational energy absorption in a helmet during an angled impact. This paper presents a short summary of possibilities and limitations on how to build a helmet test method that can measure the rotational energy absorption in a helmet during an angled impact.

Keywords: *helmet, impact, oblique, test method*

NOMENCLATURE

TBI	Traumatic Brain Injury
DAI	Diffuse Axonal Injury
SDH	Subdural hematoma
MIPS	Multi-directional Impact Protection System

I INTRODUCTION

The most common injuries in motor and sport activities are injuries to the head. The best way to protect the head is to wear a helmet. Sports and automotive helmets are today tested only for pure radial impacts to the helmet, except for the BS 6658 and EN 22.05 oblique impact test for MC helmets. A radial impact is however not the most common impact situation according to injury statistics and accident reconstructions, which show that an oblique impact is more frequent (Aare et al. 2003,

Otte et al. 1997, Verschueren 2009, Bourdet et al. 2012, Mellor and Chin 2006). The number of epidemiological studies including the direction of impact, speed and location on the helmet is few. The studies mentioned above do only give a first estimation of the impact speed and direction of impact to motorcycle, equestrian and bike accidents. As the head is more sensitive to angular motion than translational motion it is important to investigate if a test method can include a tangential component (Holbourn, 1943, Gennarelli et al., 1987; Kleiven, 2006).

A recent summary report from a sport helmet symposium presented the synthesis of information and opinion from a range of presenters and disciplines (McIntosh et al. 2013). It was concluded that there is a need to develop new test methods for helmet including an oblique impact test. McIntosh concluded that there are a number of parameters that need to be evaluated before a new oblique test could be defined. The performance criteria mentioned by McIntosh et al. was: biofidelity of the head (size, shape, mass, inertia, helmet fit and restraint fit), repeatability, robustness, reliability and validity (use of appropriate injury criteria considering combinations of angular and linear kinematics, impact force, direction and location).

There are several publications on how to design a method to measure the energy absorption in an oblique impact with a significant tangential force acting on the helmet. Aldman et al. (1978) presented a method with a spinning concrete wheel that was used to drop helmets on. Halldin et al. (2001) and Mills and Gilchrist (2008) presented methods where the head was dropped on a sliding steel plate in order to result in an oblique impact.

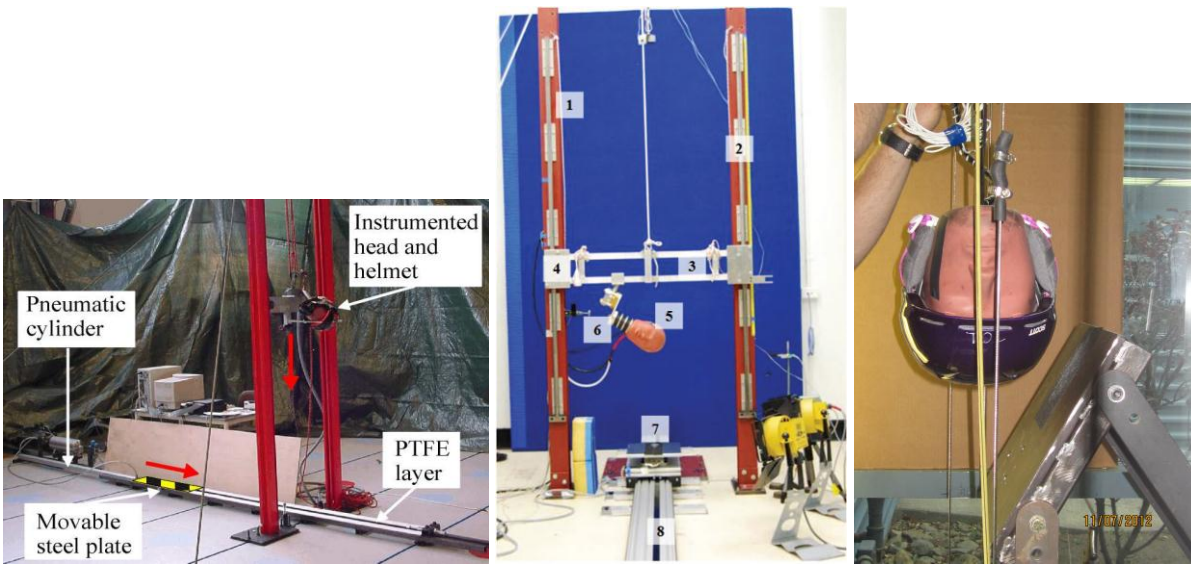


Figure 1: Different oblique test methods. Aare et al 2003, Pang et al. 2011 and photo of angled impact surface.

Pang et al. 2012 presented a method similar to the method by Halldin but with the addition of a HIII neck and also the possibility to measure the force on the plate. Other ways to test helmets for oblique impacts are to drop the helmet to an angled surface (Finan et al. 2008, Deck et al. 2012).

The results from different experimental studies including an angled impact show that it is possible to measure the energy absorption and differentiate helmets that will absorb the rotational forces better from helmets that show less good energy absorption (Halldin et al. 2001, Aare et al. 2003, Finan et al. 2008, Phillips 2013). Figure 2 shows results from a benchmark study on ski helmets performed in the test rig by Aare in Sweden. The helmets were dropped from 0.7 m. This results in a vertical speed of approximately 3.7 m/s. The horizontal speed is set to 6.4 m/s, resulting in an impact speed of 7.4 m/s and an impact angle of approximately 30 degrees. Six ski helmets from the market called A, B, C, D, E and F was compared to three helmets A, B and C also from the market but with the MIPS technology installed. The Multi-directional Impact Protection System (MIPS) was inspired by the human head and allows the outer helmet shell to move relative to the liner in the interior. This is just to exemplify that: 1; There is A wide spread in the measured data from different helmets and 2; that there are potential to increase the energy absorption in an oblique impact. The MIPS helmet presented here should be seen as one example on how to reduce the energy transmitted

to the brain. There are other examples of technologies that can reduce the rotational acceleration (Phillips 2013).

The results from these studies have also raised questions about how to design a test method that should be robust, inexpensive and reproducible. There are discussions on how to fixate the head to the helmet in an oblique impact. How hard or loose should the helmet be fixated to the head. Another question is whether the neck as a boundary condition to the head is needed in the test.

The helmet manufacturer will aim to produce the helmet that consumers want. The helmets sold to the market today are sold more or less on design, weight and comfort. Safety is not a real argument. One reason is that the approval tests are not really discussed and understood by the dealer or the end consumer. It is therefore also important to use test methods that are realistic and gives the helmet manufacturers new goals to achieve in the struggle to improve the energy absorption in the only safety barrier that is between the brain and the obstacle. It should however be stressed on that the most important is that people wear a helmet. A new test method should not result in too expensive helmets.

Within the European Committee for Standardization (CEN) TC158 (Head protection) the work has initiated to design a new test method for helmets in general. CEN TC158 has been working on this topic in the past without any concrete results like a new standard. One reason for that the work has started again is that the knowledge about head injury biomechanics has come to a new level,

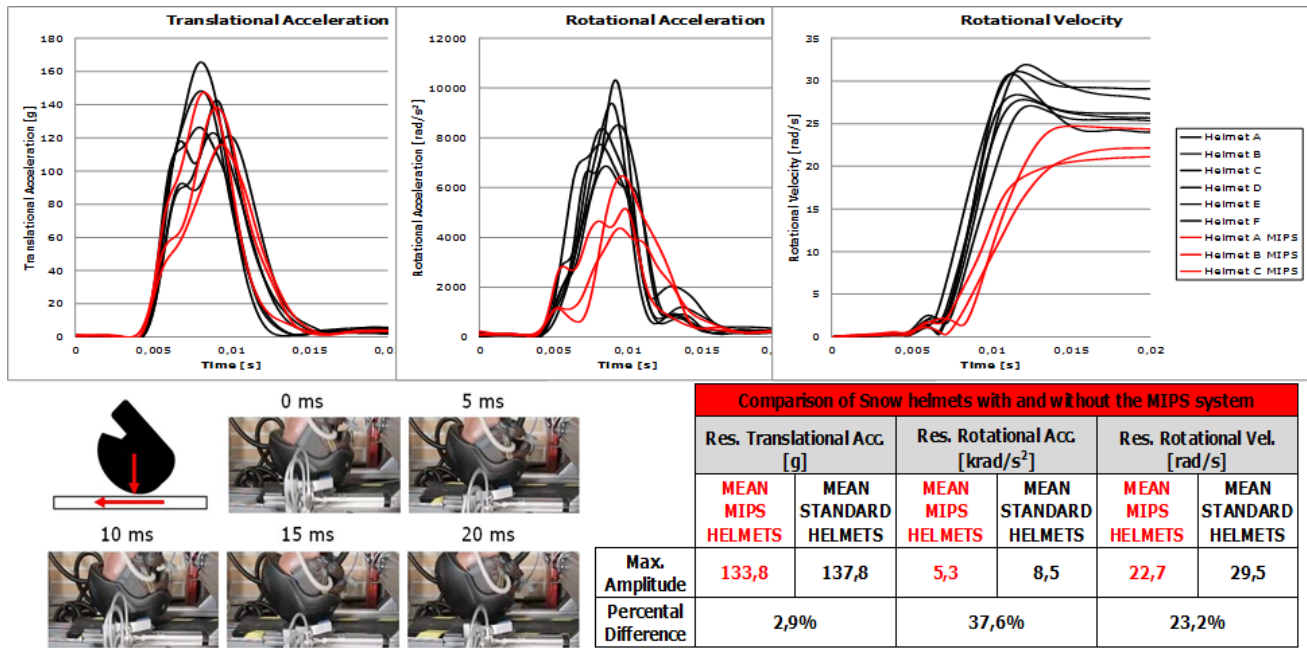


Figure 2: Benchmark Ski-helmets

very much depending on more sophisticated experimental and numerical simulations of real impact scenarios.

This paper presents a short summary of the initial work within CEN TC158 and also the possibilities and limitations on how to build a test that can measure the rotational energy absorption in a helmet during an angled impact.

II REQUIREMENTS FOR AN OBLIQUE HELMET IMPACT TEST

In this section the fundamentals will be defined for a new test method to measure rotational energy absorption in impacts including a tangential force. In order to design a new test method for homologation tests of helmets used worldwide the following basic requirements need to be fulfilled:

- Simple, robust and cost effective.
- Impact conditions based on real accident data.
- Adjustable for several helmet segments.

CEN TC158 (Working group 11) has the following subtasks defined in order to design the new test method, Figure 3. The subtasks are not fixed and more tasks will probably be added. Below are most of the tasks addressed and discussed. No final suggestions are made, but it is important to spread information regarding the work within WG11 in order make the best possible test method in place.

A. IMPACT SPEED AND ANGLE

The goal is that the test should be designed for each helmet segment. The typical speed, angle and impact surface can vary within each sport and activity in unlimited ways. There are anyway a number of studies published that could give estimation for different impact situations for each helmet segment. Table 1 presents accident reconstruction studies for bike, MC, equestrian and ski (Super-G and Down-hill).

A normal drop tower has limitations regarding the height which makes it difficult to test helmets above 10 m/s. To design a test method that can be used for all helmet segments might be difficult as the impact speed for some helmet segments would need a drop tower higher than 10 m.

The impact angles presented in Table 1 are between 20-60 degrees (0 degree is if you are lying on the floor and 90 degrees is how helmets are normally tested today). The question is if the impact angle should be chosen based on accident reconstruction studies or if the impact angle should be chosen in order to introduce as much tangential force as possible. Here it is not meant that a test should be designed with the goal of a high tangential force in a sport where it is not evident. The test should measure the rotational energy absorption in the helmet. If the angle is too steep the helmet will just slide on the impact surface and that will not evaluate the helmets possibility to absorb the rotational energy, (Mills

et al. 2009, Ghajari et al. 2012). The angle should probably be between 30-45 degrees in order to result in a normal force between the helmet and the ground large enough to avoid slippage. As Mills present the slippage is very much dependent on the normal force component, the coefficient of friction between the head/helmet and helmet/plate and the total inertia of the head and the helmet.

TABLE I. IMPACT SPEED AND ANGLE FOR MC, EQUESTRIAN, BIKE AND SKI HELMET FROM ACCIDENT RECONSTRUCTION STUDIES.

Helmet	Reference	Speed (m/s)	Angle (Degree)	Surface
MC	Otte et al. 1999	12	<30	Side of a car or the road
Equestrian	Mellor and Chinn 2006	9	37	Hard grass
Bike	Verschueren 2009	5,3	40	road
	Bourdet et al. 2012	6,8	60	car
Ski (DH and Super-G)	Inhouse report.	19	21	Hard snow/ice

The impact surface is another subject that needs to be evaluated both for the stiffness and the coefficient of friction. The impact surface for MC and bike helmets should probably be hard, and a steel plate covered with grinding paper is probably a robust and simple design.

In a sport like equestrian the impact surface should mimic hard grass or turf (Forero 2009). However, an

impacting surface that is deformable might be difficult to control or expensive. The final solution might be to use a stiff surface and reduce the impact speed to get a shell deformation that is realistic for an equestrian accident.

B. TEST METHOD DESIGN

There are many ways to design a test method for an oblique/angled impact as shown in Fig. 1. There are two existing test methods as presented in (UN ECE reg. 22-05, Methods A) for MC helmets. Test method A is designed in order to measure the tangential force between the helmet and the impacting plate that is angled 15 degrees. The idea of dropping the helmet at an angle is tempting as it is simple with just one part moving, the helmeted head. The simplicity of measuring the tangential and the normal force in the plate is interesting as it is a cheap solution instead of having a number of accelerometers and/or rotational transducers. However, it has not been shown that the tangential force in the plate can measure the energy absorption in the helmet as accelerometers in the head form can do (Mills et al. 2009). A possible improvement of the test used in ECE 22-05 is to change head form used and install accelerometers or a combination of translational accelerometers and rotational transducers. Deck et al. 2012 presented a proposal for a new test method for Bike helmets where the helmet should be dropped onto a 45 degree angle. Deck proposed to use the HIII neck in the test.

One benefit of a test method using a vertical drop onto an angled surface is that it can be installed in most test

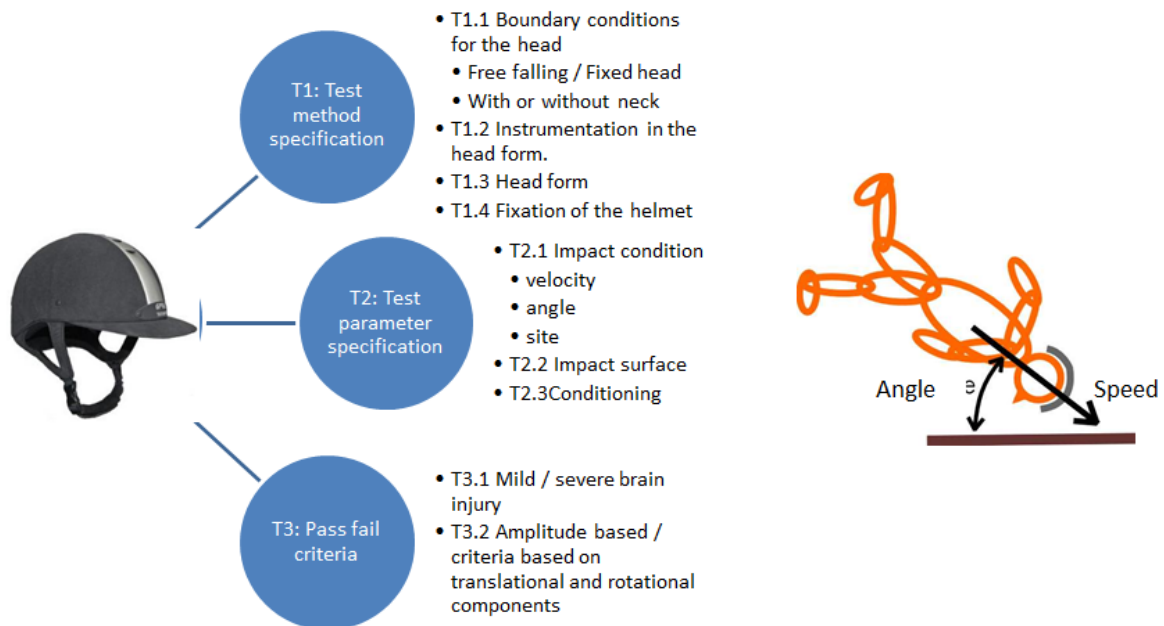


Figure 3: Shows the tasks that need to be addressed when designing a new oblique test method

institutes with minor changes. The existing drop tower can be used if the drop height is below 5 m.

Another method is as presented by Halldin et al. (2001), Mills and Gilchrist (2009) and Pang et al. (2011) to drop the helmet against a plate that is accelerated to a controlled speed. This design has its benefits as well as limitations. The benefit is that it is easy to set different combinations of impact speeds and impact angles. One limitation is that the test is more complex and therefore more expensive compared to a drop against an angled surface.

A third test method is the one developed by NOCSAE where a linear impactor is accelerated by a pneumatic cylinder to hit the centre of gravity in the dummy head (NOCSAE 2006). The dummy head is positioned on a Hybrid III dummy neck. The impactor is equipped with curved plastic surface to mimic a helmet to helmet hit (Designed for American football or ice hockey helmets). The test designed by NOCSAE is currently modified adding measurements for the rotational acceleration and also an initial off-set to introduce more rotation in the test than in the test set up designed by NOCSAE (Rousseau et al. 2011). The test set-up proposed by Rousseau is designed for ice hockey helmets and results in an impact with a minor tangential component which makes the test less effective to analyze the rotational energy absorption in a helmet. It is however possible to design the NOCSEA so that a larger tangential component is introduced.

So, there are several methods to introduce tangential force to the helmet and also measure the energy absorption in the head. The question is which method will reach the demands on robustness and low cost.

But before deciding on which test method to be used the following questions need to be addressed:

- Boundary condition for the head
 - Do we need the neck as a boundary for the head?
 - How to control the fixation of the helmet on the head?
- How to control the impact location?
- Injury thresholds or pass/fail criteria.

C. BOUNDARY CONDITION FOR THE HEAD

In current test methods the head is either falling unrestrained onto the impact surface (European test standards) or constrained to a monorail through a rigid arm attached to the head (US test standards). This can be

said to be two extremes. Between these extremes is the normal situation where the head is constrained by the human neck. In order to design an oblique test method there are questions if the neck will affect the measured translational and rotational accelerations in the dummy head. It is clear that the head is restraint by the neck and at some time will rotate around a point in the neck or even lower down in the thoracic region. Earlier studies like the COST 327 study has shown that the amplitude of the rotational acceleration is affected by the neck (COST327 2001). Helmeted full body Hybrid III dummies were dropped on an angled surface and compared to free falling helmeted head forms. The results showed that the rotational acceleration differed in amplitude by about 20%. Ghajari et al. 2012 showed that the rotational acceleration components could differ as much as 40% comparing a helmet impact with the full body and the head only. In this study Ghajari used the THUMS model and simulated an oblique impact on the lateral portion of the helmet. Ghajari proposed to change the inertial properties of the head in order compensate for the neck and the body if using the head only in an oblique impact test.

Forero 2009 reconstructed 12 jockey accidents using MADYMO. Two of these were studied in detail simulating with and without the body in a helmet to racetrack turf. The rotational acceleration was increased from 6462 krad/s² to 10104 krad/s² in one case and from 5141 krad/s² to 6444 krad/s² in the second case comparing the simulation with a complete body and a simulation with the head only. It was mentioned in this study that the MADYMO human body model has an unrealistic representation of the flexibility in the vertebral joint representation that could have resulted in this large difference.

Verschueren et al. 2009 performed reconstruction of 22 bike accidents using MADYMO. Nine of the accidents were simulated with the head only and also with the complete body. The result from this study showed that the correlation between the rotational acceleration between the head only and the simulation with the complete body correlated *well* for four of nine reconstructions. The correlation was defined as *medium* for three and two out of nine were defined as *bad* with a difference of about 30% for one example which was defined as *bad*.

The duration of impact time is different in the jockey accident against the racetrack turf (8-20 ms) and the bike accidents against a hard road (5 ms). If a test should be designed with a surface mimicking a racetrack turf for Jockey helmets a neck might be demanded. Forero also

mentioned that absence of the neck and the body might result in that the direction of the acceleration is altered.

It is therefore possible that there are impacts against harder surfaces where the neck does not have time to affect the head during the time of impact.

The conclusion that can be made here is that the neck in general affects the motion of the head. It can also be argued that a test method could be defined with impact angles where the effect of the neck is less during the short time (5 ms) when the helmet has contact to the impacting surface.

The main reason to define a test method without a neck is to make the test simpler and less expensive. If this is the case and impact directions are chosen where the neck affect the rotational acceleration this need to be taken into account in the test either by:

- The proposal by Ghajari et al. (2012) where the head inertia is scaled to take the boundary forces from the neck into account.
- To scale the pass/fail criteria.

One reason to include the neck like the HIII neck is that it makes the fixation of the head easier and more controlled as proposed by Pang et al. 2011. The HIII neck is on the other hand known to be too stiff and not validated to volunteer or cadaver experiments except for pure frontal impacts at 11 m/s.

The other boundary condition that needs taken into account is fixation of the helmet to the head. Mills and Gilchrist (2008) performed oblique tests on bicycle helmets using a HII head equipped with an acrylic wig to mimic the hair and scalp. Aare and Halldin (2003) also performed tests using an artificial scalp. The effect of these artificial hair or scalp models did affect the measured rotational acceleration. The fixation of the helmet on the head is important and needs to be controlled. Most helmets today are using a head restraint system that can be adjusted by a screw or air pump systems. The amount of adjustment must be defined in a test standard.

It can be concluded that the influence of the neck and the body on the head accelerations needs to be investigated further. Also the fixation of the helmet to the head needs to be specified.

D. IMPACT LOCATION ON THE HELMET

The impact location on the helmet should if possible be chosen from accident statistics like COST 327, McIntosh et al. 1998. The impact location could either be

defined with impact point or a region/area. There are benefits of defining just an impact point on the helmet as well as defining a region on the helmet. It is of course appreciated of the test engineer in the test institute can define a spot within a defined area on the helmet, as he or she will have the skill to locate the weakest point on the helmet. The limitation with defining a point on the helmet could make the helmet perform well for just that point. Defining a region on the other hand can, if the region is too large, result in a large variation in the measured rotational acceleration depending on which point is chosen within the region. Fig. 3 shows an example where a HIII head equipped with an FE model of a motorcycle helmet is impacted in the front region. The helmet initial position is altered 10 degrees from a baseline position. The computed rotational acceleration in this case differed around 15%.

E. PASS/FAIL CRITERIA

It is important to decide if the helmet should protect for concussion or more severe brain injuries like DAI and SDH. No generally accepted thresholds exist for rotationally induced brain injuries but the tolerance curves for DAI by Margulies and Thibault (1992) of around 8000 rad/s^2 and 70 rad/s could be a starting value for the onset of severe brain injuries like DAI. However, these values need to be reduced when adding the translational acceleration to the impact pulse, (DiMasi et al.1995, Kleiven, 2007). It is also likely that the thresholds will need to be different for different impact directions or include the head kinematics for all degrees of freedom of the head (Kleiven, 2003, 2006).

It is possible to use a detailed FE model to derive a test specific pass/fail criteria based on the translational and rotational components as proposed by Aare and Kleiven 2004. Another proposal by Deck et al. 2012 is to use a detailed FE model of the human head and brain as a black box and compute the stress or the strain in the brain

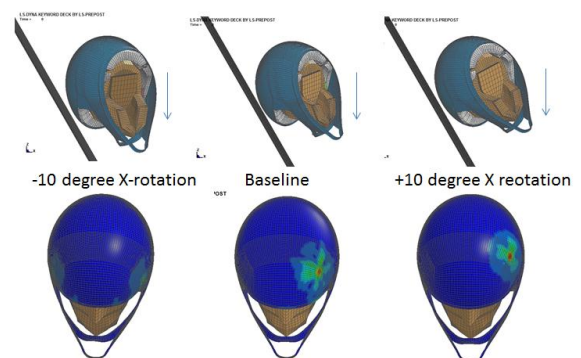


Figure 4: Example of different impact points on the helmet.

by applying the kinematics from the specific test of interest.

III CONCLUSIONS

Several different research groups in Europe, the US and Australia have defined the importance of complementing the current test methods with an oblique helmet test. The final solution for such a test is not yet defined. The challenges are primarily to:

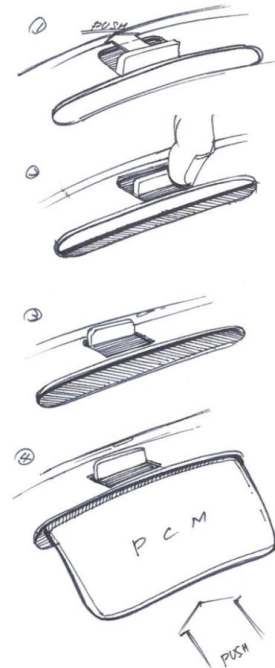
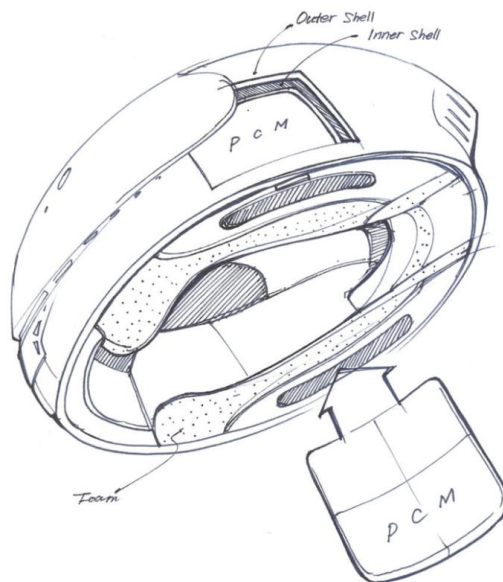
1. Quantify the effect of the boundary conditions to the head in all impact situations.
2. Define simple pass/fail criteria.
3. Design a test that is easy to use, cheap and robust.

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Helmet Performance and Design



The Impact Attenuation Test of Motorcycle Helmet Standards

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ABSTRACT

In this paper, the methods of the European (UNECE22.05), American (FMVSS218), British (BS6658), Australia and New Zealand (AS/NZS1698) and Snell (M2010) standards for evaluating the impact absorption performance of motorcycle helmets are described and compared. The compared features are the test apparatus, impact initial conditions, impact points, impact output and the approval limit. This comparison reveals that these standards adopt the same method for evaluating the impact performance of helmets, which is positioning the helmet on a metal headform and dropping them onto a rigid anvil. During impact, the linear acceleration of the centre of gravity of the headform is measured; the approval criterion is based on this acceleration.

Several studies on the relevance of this test method to real-life accidents are reviewed and their main findings are summarised. The review includes studies on the interaction between the head and neck during helmeted head impacts and those on assessing the performance of helmets during oblique impacts by using rotational acceleration, along with linear acceleration. It appears that in both areas, more research needs to be carried out to be able to influence current standards.

Keywords: *helmet; motorcycle; impact; standard*

I. INTRODUCTION

Although the number of fatalities in motorcycle accidents is high in comparison with motorcycle use [1], the almost only equipment that prevents motorcyclists from fatal injuries is the helmet. In order to evaluate the protective performance of helmets during accidents, they are tested according to a standard method. Almost all standards follow the same concepts for evaluating the effectiveness of helmets during accidents, which are:

- the helmet shall be able to absorb impact energy,
- it shall remain on the head during the accident, and
- it shall resist penetration.

However, details of procedures in force in various countries are different. Hence, it is probable that a helmet satisfying the requirements of one standard will not comply with all requirements of another standard.

In this paper, the impact absorption test of the European standard (UNECE 22.05 [2]) is described and compared with the method prescribed by four other standards. In addition, some studies and criticisms on different aspects of the impact absorption test method are reviewed.

II. THE IMPACT ABSORPTION TEST OF UNECE 22.05

The United Nation's regulation on the construction of motorcycle helmets in the United Nations Economic Commission for Europe (UNECE) is the regulation No. 22 "uniform provisions concerning the approval of protective helmets and their visors for drivers and passengers of motor cycles and mopeds". Any helmet manufacturer who intends to sell their products in the countries that have agreed to adopt this regulation into their legislations (contracting parties) should obtain a type approval. This regulation is adopted by over 50 countries worldwide [3] and is probably the most widely accepted set of requirements for manufacturing helmets and visors in the world. The latest amendments entered into force were series 05 [4]. This version of the regulation is referred to as the UNECE 22.05 standard throughout this paper.

In this standard, the impact absorption capacity of a helmet is determined by "recording against time the acceleration imparted to a headform fitted with the helmet, when dropped in guided free fall at a specific impact velocity upon a fixed steel anvil" [2]. Impacts shall be carried out on specific points on the helmet (Figure 1), which are point B in the frontal area, point X in either left or right lateral area, point R in the rear area, point P round the vertex of the helmet and S in the lower face cover area (if the helmet is closed-face).

The test apparatus should have the following tools (Figure 2):

1. Base: it shall be made of steel, concrete or both and weigh at least 500 kg. Natural frequencies of the base or its parts shall not influence the impact results.
2. Anvils: two anvils are used in impact tests; flat and kerbstone. The flat anvil shall have a circular impact area with a diameter of 130 mm. The kerbstone anvil shall have two sides forming an angle of 105° , each of them with a slope of 52.5° towards the vertical and meeting along a striking edge with a radius of 12 mm. The height must be at least 50 mm and the length not less than 125 mm. The orientation is 45° to the sagittal plane for impacts at points B, P, and R, and 45° to the reference plane for impacts at point X (front low, back up).
3. Mobile system and the guide: the mobile system shall provide a free fall for helmeted headform and the guide shall be such that the impact

velocity is not less than 95% of the theoretical velocity.

4. Accelerometers.

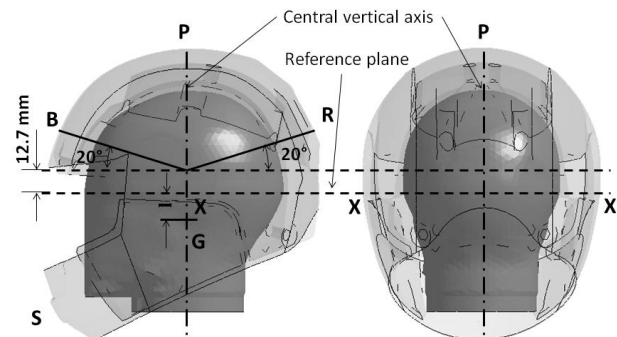


Figure 1: Definition of impact points in UNECE 22.05

Test headforms shall be made of metal and their resonance frequency shall not be less than 3000 Hz. General characteristics of headforms are presented in Table 1. Size in this table is the circumference of the headform at its reference plane (Figure 1). The centre of gravity (c.g.) of the headform shall be near the point G on the central vertical axis, shown in Figure 1 (dimension l is defined in [2]), where there should be a housing for a set of three orthogonal accelerometers.

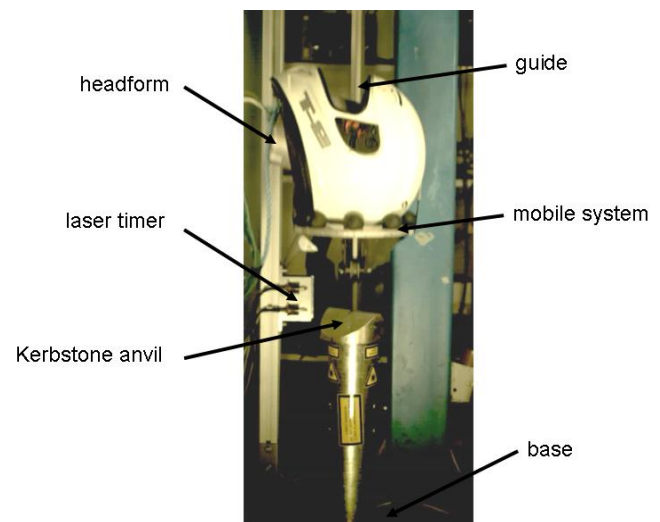


Figure 2: Test apparatus of the UNECE 22.05 standard

TABLE 1: GENERAL CHARACTERISTICS OF TEST HEADFORMS

symbols	size (cm)	mass (kg)
A	50	3.1±0.1
E	54	4.1±0.1
J	57	4.7±0.1
M	60	5.6±0.1
O	62	6.1±0.1

The drop height shall be equal to that required to achieve an impact speed of 7.5 m/s for both flat and kerbstone anvils and 5.5 m/s for tests at point S. During impacts, linear acceleration of the headform at its c.g. is recorded. The absorption efficiency is considered sufficient if the resultant linear acceleration of the headform ($|a(t)|$) does not exceed 275 g, and *HIC* does not exceed 2400. *HIC* stands for the head injury criterion and is defined as

$$HIC = \max \left\{ (t_2 - t_1) \left(\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} |a(t)| dt \right)^{2.5} \right\} \quad (1)$$

where t_1 and t_2 are, respectively, any starting and ending time in impact pulse duration.

III. COMPARISON BETWEEN HELMET STANDARDS

In this section, the impact absorption test method of the UNECE 22.05 standard is compared to the methods of four other standards, namely: the British standard (BS6658) [5], the U.S. Department of Transportation's Federal Motor Vehicle Safety Standard No. 218 (FMVSS218) [6], the Snell Memorial Foundation's standard (M2010) [7] and the Australia and New Zealand standard (AS/NZS 1698) [8].

A. Test apparatus

These standards require a guided fall. Their test apparatus should have all four components that were mentioned for the UNECE 22.05 standard, i.e. a base, anvils, a mobile system and guide, and one uniaxial accelerometer. According to Snell M2010, AS/NZS 1698, FMVSS218 and BS6658, the headform shall be attached to the mobile system through a ball joint. This joint allows for rotation and vertical translation, but constraints horizontal translations. Therefore, only one accelerometer is needed to record the headform linear acceleration. However, when testing in accordance with UNECE 22.05, the headform shall fall freely with no constraint, and thus three uniaxial accelerometers are required to measure $|a(t)|$.

Some experts believe that constraining the headform provides better repeatability than using a free motion headform [9]. Mellor et al. [10] found a coefficient of variation of 0.9% for acceleration of a guided headform as compared to 2.3% for the free motion headform. In addition, the guided headform acceleration was approximately 4 g higher, which was attributed to the restricted rotation of the headform. Thom et al. [11] also showed that when helmets were tested according to the American standard, which constraints the headform, the resultant linear acceleration was larger than when the same helmets were tested in accordance with the UNECE 22.04 standard. Nonetheless, Mills [12] believes that in motorcycle accidents within short impact duration of 10 ms, the neck provides very little resistance to rotation and therefore, the method of the UNECE 22.05 is more realistic. The influence of the neck on head acceleration will be further discussed in section IV.B.

The mass of the drop assembly, including the masses of the supporting arm, ball socket stem and headform, varies in the standards. In UNECE 22.05, BS6658, AS/NZS 1698 and Snell M2010, it depends on the headform size, as indicated in Table 1. The previous version of the Snell standard [13] specified one mass (5 kg) to test different helmet size. In FMVSS218, the mass of the drop assembly can have three values: 3.5 kg (small headform), 5.0 kg (medium headform) and 6.1 kg (large headform). It seems that designers of helmet standards have assumed that the mass of the human head increases with its size. The circumference and mass of the "J" size headform, 57 cm and 4.7 kg, are within the range of the circumference and mass of the 50th percentile human head, 572±12 mm and 4.54±0.31 kg [14], respectively. The mass of the drop assembly of the medium headform of the American standard is close to this range.

The geometry of headforms used in UNECE 22.05, BS6658, AS/NZS 1698 and Snell M2010 comply with the specifications of the ISO DIS 6220 [15] standard. This standard specifies a 5 kg mass regardless of headform size. However, the above mentioned helmet standards use different masses for different headform size. The source of the geometrical specifications of FMVSS218 headforms is uncertain [16].

Anvils used by different standards are described in TABLE 2. All standards use a flat anvil in their impact absorption test. Flat shape objects were the second frequent opposite objects (9%), in the COST 327 database [1], after round objects (79%). This database,

however, did not report the range of curvatures of round objects. There are some criticisms about using a hemispherical anvil in some standard test methods. Gilchrist et al. [17] argued that hemispherical anvils should be replaced with kerbstone anvils, because statistics show that accidents involving a hemispherical object are rare. COST 327 reported that edge shape objects, such as kerbstones, had a frequency of 4% but most serious injuries occurred for edge struck objects; 40% of all collisions to edge objects resulted in head injury with AIS 5. This evidence justifies employing kerbstone and edge anvils by some standards.

TABLE 2: ANVILS OF THE IMPACT ABSORPTION TEST OF STANDARDS

Anvil	Flat	Hemi-spherical	Kerbstone	Edge
UNECE 22.05	D ¹ = 130	-	$\chi = 105$ H $\geq$ 50 r = 12	-
Snell M2010	D $\geq$ 127	R = 48	-	L = 180 W = 6.3 H = 35
AS/NZS 1698	D $\geq$ 127	R = 48	$\chi = 90$ H = 85 r $\leq$ 0.5	-
BS 6658	D = 130	R = 50	-	-
FMVSS 218	D = 127	R = 48	-	-

¹ D: diameter (mm), R: radius (mm), χ : vertex angle (°), H: height (mm), r: fillet radius (mm), L: length (mm), W: width (mm).

B. Impact initial conditions

Prescribed initial conditions for impact tests are different in the standards. ECE 22.05, BS 6658, Snell M2010 and FMVSS 218 define impact velocities but AS/NZS 1698 defines drop heights (Table 3). Snell M2010 and BS6658 require a second impact at the same site, but at lower impact velocities. AS/NZS 1698 and FMVSS218 also require a second impact at the same site, but initial conditions are the same as the first impact. Gilchrist et al. [17] believe that requiring a second impact prevents the liner foam's density to be optimised for the first impact and leads to using stiffer foams. They argue that the major impact damages about 5% of the whole protecting area of the helmet, so

there is a small probability that the second possible impact occurs within this area. The only standard that does not require a second impact is UNECE 22.05.

TABLE 3: IMPACT INITIAL CONDITIONS OF DIFFERENT STANDARDS

Anvil	Flat		Hemispheric al		Kerb-stone
Impact	1 st	2 nd	1 st	2 nd	
UNECE 22.05	7.5 m/s	-	-	-	7.5 m/s
Snell M2010	7.75 m/s	A-E: 7.09 m/s J: 6.78 m/s M: 5.73 m/s O: 5.02 m/s	The same as flat anvil.		7.75 m/s
AS/NZS 1698	1830 mm	1830 mm	1385 mm	1385 mm	-
BS 6658 (Type B ¹)	6.5 m/s	4.6 m/s	6 m/s	4.3 m/s	-
FMVSS 218	6 m/s	6 m/s	5.2 m/s	5.2 m/s	-

¹ BS 6658 has two types of assessment: Type A which is for users who demand an especially high degree of protection and Type B which is suitable for ordinary motorcycle riders on public roads.

In the COST project [1], the impact velocity of the rider's head was estimated using the impact speed of the motorcycle, kinematics of the motorcyclist during accident and position of the body with respect to the struck object before the impact. It was found that head injury severity increased when the head impact velocity increased (Figure 3). The median speed (50% cumulative speed) was 18 km/h (5 m/s) for AIS 1, 50 km/h (13.9 m/s) for AIS 2-4 and 57 km/h (15.8 m/s) for AIS 5/6. In general, the median speed was 44 km/h (12.2 m/s). The impact speed of the UNECE 22.05 standard corresponds to 20% cumulative speed for AIS 2-4 and 15% cumulative speed for AIS 5/6. An increase in the 20% cumulative speed from 7.5 m/s to 9.5 m/s changes the head injury severity from AIS 2-4 to AIS 5/6, which is equivalent to saving of about 1000 lives per year in Europe [1]. However, this increase in the impact speed is equal to a 60% increase in the

kinetic energy. The kinetic energy of impact absorption tests, which defines the severity of impact tests, determines the thickness of the helmet liner and thus its external dimensions. Motorcyclists often refuse wearing large helmets because they are not aesthetically pleasing. In the final proposal of COST for an improved test method, an impact speed of 8.5 m/s was specified.

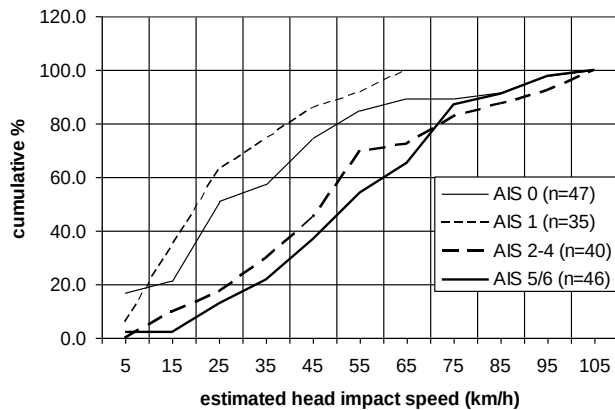


Figure 3: Head AIS vs. head impact speed [1]

C. Impact points

In contrast to UNECE 22.05, which defines impact points, the other four standards do not define specific impact points. In Snell M2010, impact points should be on or above a test line and at least 120 mm apart. AS/NZS 1698 and FMVSS218 also define a test line and require the impact points to be on or above it. According to the BS 6658 standard, a helmet should be tested at three impact points. The points should be located at the rear or side, front and any other site above a defined line. These standards vis-à-vis UNECE 22.05 leave some discretion to the helmet tester regarding the impact point selection. Hence, the tester can investigate the potential weaknesses of helmets.

Among important outputs of the accident reconstruction program of the COST project were body impact angle and head impact angles, shown in Figure 4. The body impact angle is the angle between the anatomical axis of the body and the tangent line of the opponent object, e.g. the road surface. The location of the impact point on the head is defined by the XY and XZ impact angles.

Distributions of the body and head impact angles among 95 motorcyclists suffered head injuries with AIS 2+ are shown in Figure 5. This figure indicates that most of severe head injuries occurred in shallow body impact angles. In addition, 26% of impacts were

perpendicular to the head axis. The XY impact angle was -45° to $+45^\circ$ for 64% of impacts. As can be seen, the head was impacted at different sites even though the frequency of impacts at some sites was considerably higher. It seems that standards have chosen impact sites so that the majority of the helmet area is subjected to impact.

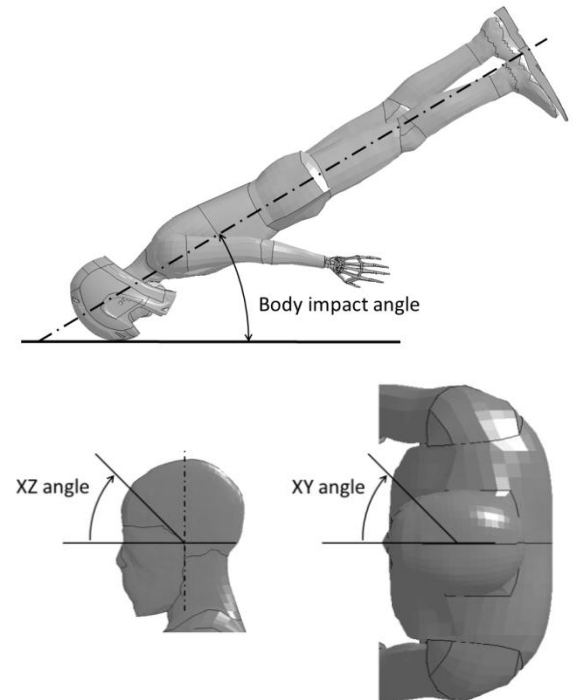


Figure 4: Body impact angle and head impact angles.

Another output of COST was the location of the impact on the helmet, which was found from damaged areas on the helmets. The frequency of the damaged locations were 26.9% lateral right, 26.3% lateral left, 23.6% frontal and 21.0% rear. The least frequent impact area was crown with 2.2% occurrence. One of the criticisms about UNECE 22.05 is that it specifies an impact point at the crown site (point P), while the frequency of impacts at this point is very low.

D. Impact output and approval limit

The studied standards employ the same impact output in their impact absorption test procedure, which is resultant linear acceleration of the headform or the support assembly versus time. However, their criteria and relevant limits are different, as can be seen in Table 4. Some experts believe that adopting a higher limit for peak linear acceleration will result in stiffer helmets, which may prevent fatal injuries but foster more common but less severe injuries [9]. The dwell time at an acceleration level defined in the AS/NZS 1698 and

FMVSS218 standards reflects the concept of the Wayne State University curve: the tolerance of the human head to linear acceleration decreases at longer dwell times. Despite the fact that HIC is based on this curve, there is significant debate about its suitability for helmet standards [9]. In addition, it has a high limit in the UNECE 22.05 standard, which lets currently available helmets pass the test. The HIC limit for AIS 3 head injuries was found to be 1500 [1].

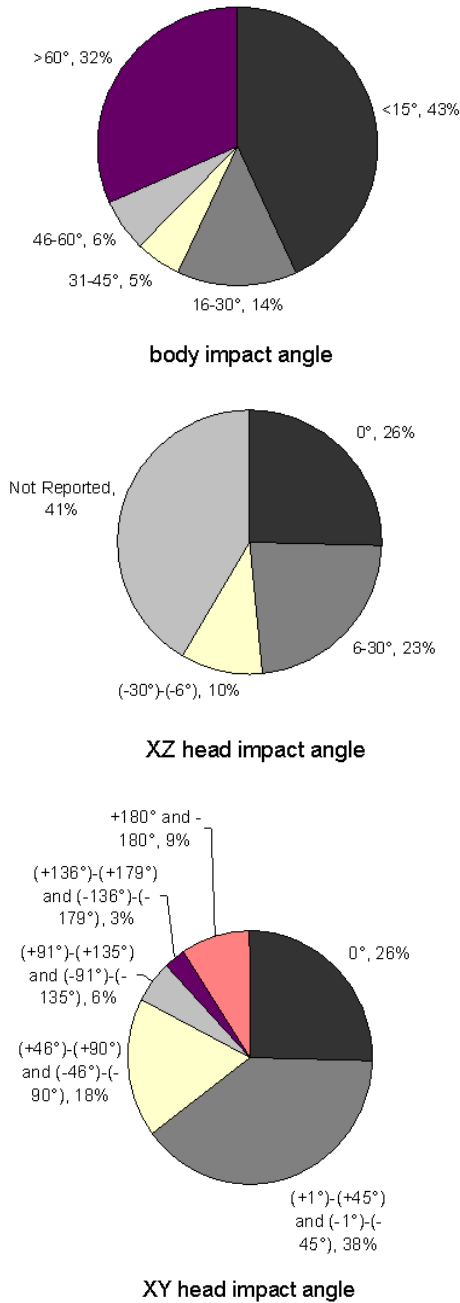


Figure 5: Distribution of body impact angle and head impact angles of AIS 2+ in 95 motorcycle accidents. Figures were generated based on data presented in [1].

IV. STUDIES ON THE IMPACT ABSORPTION TEST METHOD

Comparison between five helmet standards in the previous section reveals that these standards adopt the same method for assessing the impact absorption capability of helmets, i.e. dropping a helmeted headform onto an anvil and measuring linear acceleration of the headform. This test method was devised more than forty years ago [16]. Studies on its various aspects have shown that this test method can be further improved in order to define better guidelines to helmet designers and subsequently mitigate accident injuries [1]. However, only a few studies of this type were found in literature, which can explain why standards have not adopted their suggestions yet. In this paper, some of these studies are reviewed.

TABLE 4: TEST CRITERIA AND THEIR LIMITS OF SOME HELMET STANDARDS

Criterion No.	1	2	3
UNECE 22.05	$PLA^1 \leq 275 \text{ g}$	$HIC \leq 2400$	-
Snell M2010	A-J: $PLA \leq 275 \text{ g}$ M: $PLA \leq 264 \text{ g}$ O: $PLA \leq 243 \text{ g}$	-	-
AS/NZS 1698	$PLA \leq 300 \text{ g}$	3 ms at 200 g	6 ms at 150 g
BS6658	$PLA \leq 300 \text{ g}$	-	-
FMVSS 218	$PLA \leq 400 \text{ g}$	2 ms at 200 g	4 ms at 150g

¹ PLA: Peak Linear Acceleration

A. Rotational acceleration during oblique impacts

The COST study showed that in more than 50% of motorcycle accidents, the body impact angle was less than 30° (Figure 5). In addition, the median impact speed, 12.2 m/s, was much larger than the speed of a free fall from a typical rider's head height (1.5 m), i.e. 5.4 m/s. These observations point to the fact that in majority of motorcycle accidents the impact velocity

has a significant tangential component. These impacts are called oblique impacts.

Only the UNECE22.05 and BS6658 standards prescribe oblique impacts; the helmeted headform is dropped onto a flat anvil inclined at 15° to the vertical and covered with an abrasive paper or projections. These standards have another method, which is dropping the helmeted headform on a base that moves in the horizontal direction. The tangential force is measured during the oblique impacts. Halldin et al. [18] believe that the main purpose of this standard test is to insure that the tangential force on the helmet shell, when it impacts a rough flat surface, is not larger than the shear resistance capability of typical shell materials used in 1985 (the year of including the test in the British standard).

In addition to linear acceleration of the head [19, 20], its rotational acceleration has been given special attention as a cause of brain injury. Holbourn [21] believed that rotational acceleration applied to the head, with or without direct impacts, results in shear and tensile strains in the brain and bridging veins, which causes haematoma and diffuse axonal injury (DAI). Gennarelli [22] stated that the most frequent head traumas in motor vehicle accidents that results in either fatality or need for long-term rehabilitation are subdural haematoma (SDH) and DAI. He concluded that SDH was mainly due to short duration and high amplitude rotational acceleration, while DAI was mainly due to long duration and low amplitude rotational acceleration. Margulies and Thibault [23] found that the onset of DAI was a combination of a 10 krad/s^2 rotational acceleration and a 100 rad/s maximum change in rotational velocity. However, Ueno and Melvin [24] showed that if translational and rotational motions were combined, the above limit had to be decreased.

Several studies have been conducted addressing the need for oblique impact testing of helmets and recording rotational acceleration during impacts, along with linear acceleration. Halldin et al. [18] constructed a test rig which was a modified version of the oblique impact test apparatus of the British standard. A helmeted headform was dropped, in a guided free fall, on a moving plate covered with an abrasive paper. Linear acceleration of the head and one component of the rotational velocity were recorded. With this test set up, they investigated the effect of inserting a low friction layer between the foam and the liner of a helmet on the angular acceleration transferred to the head. Aare [25] used this test rig to subject helmets to

oblique impacts. A modified head of the Hybrid III dummy was equipped with enough accelerometers to measure its linear and rotational accelerations. However, similar to Halldin et al. [18], he used this test facility to study new ideas for helmet design rather than investigating impact absorption test of standards.

In the COST study [1], rotational acceleration of the head during oblique impacts was monitored through dropping helmeted Hybrid II headforms onto a flat anvil inclined at 15° to the vertical and covered with an abrasive paper. The aim was to find a possible correlation between the rotational acceleration of the headform and the tangential force applied on the helmet. Four types of helmets, with different shell materials (thermoplastic and composite), liner densities and masses, were tested at impact speeds ranging from 6 to 12 m/s. Linear and rotational accelerations of the head and the tangential force on the anvil were measured. The mean values of peak rotational acceleration varied between 2.5 krad/s^2 and 8.5 krad/s^2 and those of peak tangential forces varied between 0.8 kN and 2.5 kN. Linear regression analysis showed a strong correlation between the peak values of rotational acceleration and the tangential force.

Mills et al. [26], however, argued that this correlation is only valid when a single site of the helmet is impacted and the normal component of the impact velocity is low ($V_N \leq 2.5 \text{ m/s}$). Based on the results of helmet oblique impact simulations, they showed that for more severe oblique impacts, with $V_N \geq 5 \text{ m/s}$, and at a range of impact sites, the correlation between the peak headform rotational acceleration and the tangential force is poor.

By plotting the peak linear acceleration of the headform vs. V_N , collected from not only COST oblique impact study but also [27, 28], Mills [12] demonstrated that the peak resultant linear acceleration of the headform is a linear function of V_N , which agrees with the findings of other studies [29, 30]. This was attributed to the liner crushing distance increasing linearly with V_N . The normal force is also a function of the liner crushing distance [17]. Mills and Gilchrist [31] showed that the normal force has a significant contribution to the rotational acceleration of the headform. Therefore, helmet designs that reduce the normal force, thus linear acceleration, would probably reduce the rotational acceleration.

B. Effects of the presence of the body

In real-life motorcycle accidents, the body can interact with the head during the impact. In the impact

absorption test of helmet standards, however, the possible effects of this interaction are not taken into account. It should be noted that the mass of the headform, or drop assembly, adopted by standards is close to the mass of the human head. Mills [12] argued that given the short duration of helmet impacts, nearly 10 ms, the neck/head interaction is not significant and thus the head can be assumed to be isolated. However, the results of drop testing cadaver head-neck on a thick layer of foam [32-34] do not confirm this opinion. The experiments showed a nearly 7 ms delay in the onset of the lower neck load with respect to the head load. A numerical study [30], using a biofidelic model of the human body and a commercially available helmet, showed that this delay can be significantly shorter when using foams that are often used in helmets.

In [1, 35-39], the effects of the body on the dynamic response of the helmeted head have been investigated by employing dummies. In the COST study [1], the Hybrid III headform and its detached head were employed. Helmet impacts were performed by dropping the helmeted dummy (full-body impact) and the helmeted head (isolated-head impact) onto flat anvils. The highest impact velocity was 6 m/s. It was found that the linear acceleration of the head was smaller in full-body impacts.

Ghajari et al. [38] used a model of the Hybrid III dummy and a commercially available helmet to investigate the effects of the body during impacts with initial velocities of 6 m/s and 7.5 m/s. The model was validated against impact experiments (Figure 6), with respect to the head linear and rotational accelerations, upper neck forces and moments and anvil force [40]. When the liner foam was not compressed beyond its plateau regime, i.e. at 6 m/s, the head linear acceleration in full-body impacts was smaller than that in isolated-head impacts. However, at 7.5 m/s, the liner bottomed out during the full-body impact, which resulted in very high contact force and head accelerations.

Based on the solutions to an analytical model of the helmeted head impact [38], increasing the mass of the headform was suggested as a practical method for taking into account the effect of the body in isolated-head impacts. The analytical model revealed that the presence of the body and the added mass have the same effects on the dynamic responses of the head and helmet; they increase the liner crushing distance and the normal force but decrease linear acceleration of the head when the liner foam is not loaded beyond its plateau regime.

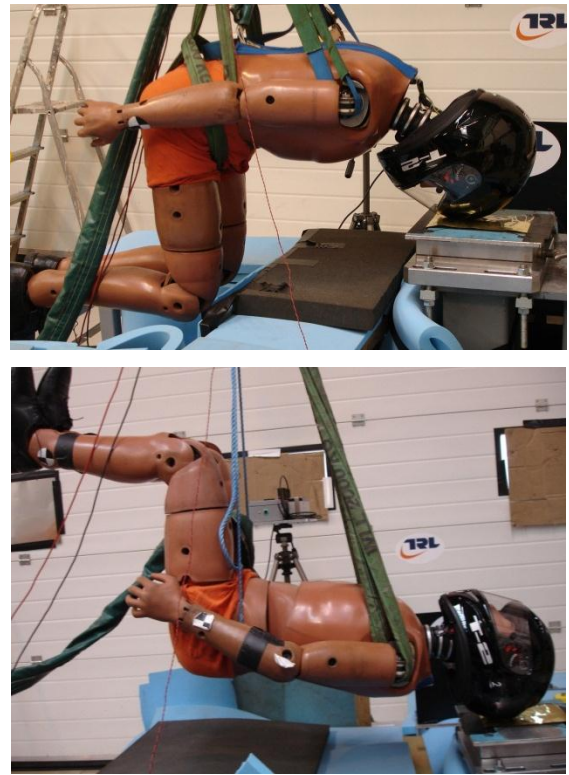


Figure 6: Helmeted Hybrid III dummy impact test set-up

One limitation of these studies was using a Hybrid III dummy as the human body surrogate. This dummy has a very stiff neck under axial compression loading, as compared to the human neck [41-43]. To evaluate the added mass, Ghajari et al. [44] employed a biofidelic model of the human body, THUMS, featuring a very detailed neck model. The model was validated against cadaver experiments of [45] with respect to the upper neck forces. The model of a commercially available helmet was coupled with the head of THUMS and impacts at front, rear and side sites and various body impact angles were simulated. The added mass was determined for these impact configurations. The results showed that the added mass increased linearly with the body impact angle and it was nearly independent of the impact site. At a body impact angle of 0°, the added mass was approximately 10% of the original mass of the headform. This percentage was 20% and 40% for body impact angles of 30° and 90° respectively.

It should be noted that only increasing the mass of the headform would cause helmet designers to use stiffer liners, i.e. liners with larger plateau stress. This change can increase the level of acceleration suffered by the rider's head during an accident. To avoid such

designs, two solutions were proposed [44]: a) decreasing the acceptance limit of head acceleration, b) prescribing impacts with two sets of headforms: one with the original mass and another with the increased mass.

V. CONCLUSIONS

The impact absorption test of the UNECE 22.05 standard was described and compared with four other standards. It has been shown that helmet standards prescribe the same method for assessing the impact absorption capability of helmets; a helmet positioned onto a headform is dropped onto an anvil and linear acceleration of the headform versus time is measured. However, their details are different, which can affect the design of helmets and the level of safety that they offer.

There are some common important features in the helmet standards. Among them is employing an isolated headform, whose mass is in the range of human head's mass. It seems that designers of helmet standards have presumed that the influence of the body on the impact response of the head and helmet is negligible. However, studies, using dummies and biofidelic models of the human body, have shown that the neck force exerted on the head during an impact can significantly increase the crushing distance of the helmet liner. In severe conditions, the liner may bottom out resulting in very high head accelerations. More studies appear to be needed to further investigate this important issue.

The studied standards have adopted pass/fail criteria that are based on linear acceleration of the head. However, brain injury can be better predicted by knowing the complete kinematics of the head, which includes its rotational acceleration as well as linear acceleration. Linear and rotational accelerations of the head can be used to drive detailed models of the human head and obtain information about different types of tissue-level head injury, such as SDH and DAI.

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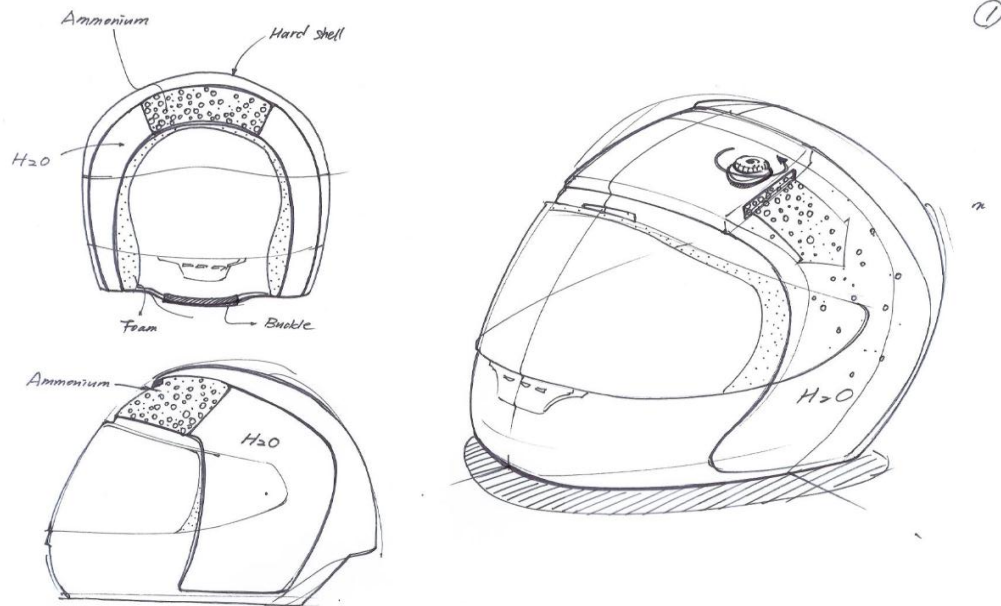
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Helmet Performance and Design



Model Based Head Injury Criteria for Head Protection Optimization

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ABSTRACT

This paper presents an original numerical human head FE models followed by its modal and temporal validation against human head vibration analysis in vivo and cadaver impact tests from the literature. The human head FE model developed presents two particularities : one at the brain-skull interface level where fluid-structure interaction is taken into account, the other at the skull modelling level by integrating the bone fracture simulation. Validation shows that the model correlated well with a number of experimental cadaver tests including skull deformation and rupture, intra-cranial pressure and brain deformation. This improved numerical human head surrogates has then been used for numerical real world accident simulation. By correlating head injury type and location with intra-cerebral mechanical field parameters, it was possible to derive new injury risk curves for injuries as different as subdural haematoma and neurological injuries. Illustration of how this new head injury prediction tool can participate to the head protection system optimisation is also provided.

Keywords: *head modelling; head injury criteria; head protection*

I. INTRODUCTION

The head and more specifically the brain is among the most vital organs of the human body.

Over the past forty years, a slant has been put by the biomechanical research on the understanding of the head injury mechanisms. Nevertheless, an injury is always a consequence of an exceeded tissue tolerance to a specific loading. Even if local tissue tolerance has very early

been investigated, the global acceleration of the impacted head and the impact duration are usually being used as impact severity descriptors. The *Wayne State University Tolerance Curve* has therefore been proposed since the early Sixties thanks to several works by Lissner *et al.* (1960) [1] and Gurdjian *et al.* (1958) [2]. This curve shows the link between the impact of the head described by the head acceleration and the impact duration and, on the other hand the head injury risk. Hence, after the work of Gadd (1966) [3], the *National Highway Traffic Safety Administration* (NHTSA) proposed the *Head Injury Criterion* (HIC) in 1972. This is the tool used nowadays in safety standards for the head protection systems using headforms. Since it is based solely on the global linear resultant acceleration of a one mass head model, some limitations of this empiric criterion are well-known, such as the fact that it is not specific to direction of impact and that it neglects the angular accelerations.

A proposed alternative method for assessing head injury risk is to use a human head Finite Element Model (FEM), which can enable the investigation of the intra-cranial response under real world head impact conditions. This method is well known since 1975 when one of the first three dimensional model was developed by Ward *et al* [4]. This method thereby leads to added useful mechanical observables which should be closer to the description of known injury mechanisms. Hence, new injury criteria can be proposed. In the last decades, more than ten different three dimensional finite element head models (FEHM) have been reported in the literature by Ward *et al.* (1980) [4], Shugar *et al.* (1977) [5], Hosey *et al.* (1980) [6], Di Masi *et al.* (1991) [7], Mendis *et al.* (1992) [8], Ruan *et al.* (1991) [9], Bandak

et al. (1994) [10], Zhou *et al.* (1995) [11], Al-Bsharat *et al.* (1999) [12], Willinger *et al.* (1999) [13], Zhang *et al.* (2001) [14]. Fully documented head impact cases can be simulated in order to compute the mechanical loadings sustained by the head tissues and to compare it to the real injuries described in the medical reports. It has for example been shown in Zhou *et al.* (1996) [15], Kang *et al.* (1997) [16] and more recently in King *et al.* (2003) [17], Kleiven *et al.* (2007) [18] and Deck *et al.* (2008) [19] that the brain shear stress and strain rates predicted by their FEHM agree approximately with the location and the severity of the axonal injuries described in the medical report. Since these finite element head models exist, new injury prediction tools based on the computed intracranial loadings should become available.

In order to undertake a statistical approach to injury mechanisms, more accident cases were introduced in Marjoux *et al.* (2007) [20] and Deck *et al.* (2008) [19] and a first attempt of injury criteria to specific mechanisms was proposed. Another FEHM presented in Takhounts *et al.* (2003) [21] is very suitable for this kind of study due to the very short computing duration: the *Simulated Injury Monitor* or SIMon. A number of scaled animal model loading conditions lead the authors to propose as well injury mechanisms and related injury criteria based on animal experiments

In this context, the objective of the present study is to present a validated human head model and to investigate, on a set of real world head trauma, the injury prediction capability of the provided injury mechanisms related criteria. In a final section this novel head injury prediction tool is used in the context of an improved helmet test method and applied to the head protection optimization via the coupling of the head model to helmet models.

II. STRASBOURG UNIVERSIT FE HEAD MODEL

A. Meshing aspects

Kang *et al.*, in 1997 [16], has developed the Strasbourg University Finite Element Head Model (SUFEHM) under Radioss software.. The main anatomical features modelled were the skull, falx, tentorium, subarachnoid space, scalp, cerebrum, cerebellum, and the brainstem. Globally, SUFEHM model consists of 13208 elements. Its total mass is 4.7 kg, and a representation is given in Figure 1.

STRASBOURG UNIVERSITY FINITE ELEMENT HEAD MODEL

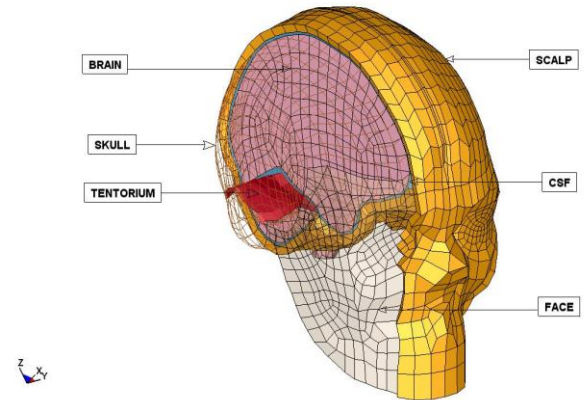


Figure 1: Section through the Strasbourg University Finite Element Head Model (SUFEHM).

B. Mechanical properties

Material properties are all isotropic, homogenous and elastic, with mechanical properties came from (Willinger *et al.*, 1995 [22]). Table 1 summarizes mechanical properties.

TABLE 1: MECHANICAL PROPERTIES OF SUFEHM

Part/Material property	Material parameter	Value
Face/Elastic	Density	2500 kg.m ⁻³
	Young modulus	5.0E+03 MPa
	Poisson's ratio	0.23
Scalp/Elastic	Density	1.0E+03 kg.m ⁻³
	Young modulus	1.67E+01 MPa
	Poisson's ratio	0.42
CSF/Elastic	Density	1040 kg.m ⁻³
	Young modulus	0.12E-01 MPa
	Poisson's ratio	0.49
Falx/Elastic	Density	1140 kg.m ⁻³
	Young modulus	3.15E+01 MPa
	Poisson's ratio	0.45
Tentor./Elastic	Density	1140 kg.m ⁻³
	Young modulus	3.15E+01 MPa
	Poisson's ratio	0.45

The brain is assumed to be visco-elastic. This model allows the modelling of visco-elastic behaviour for beams, shells and solids. The shear relaxation behaviour is described by:

$$G(t) = G_{\infty} + (G_0 - G_{\infty}) \exp(-\beta t)$$

With G_0 short-time shear modulus, G_{∞} Long-time shear modulus and β Decay constant. Values of the

parameters are. $G_0=4.9E^{-02}$ MPa, $G_{\infty}=1.62E^{-02}$ MPa and $\beta=145\text{ s}^{-1}$.

The skull was modelled by a three layered composite shell representing the inner table, the diploe and the external table of human cranial bone. The material model has three failure criteria expressions for four different types of in-plane damage mechanisms. Each of them predicts failure of one or more plies in a laminate. The expressions accommodate four in-plane failure modes: matrix cracking, matrix compression, fiber-matrix shearing and fiber breakage. Skull mechanical parameters are presented in Table 2.

TABLE 2: SKULL MECHANICAL PARAMETERS

	<i>Cortical bone</i>	<i>Diploe bone</i>
Mass density [kg/m ³]	1900	1500
Young modulus [MPa]	15000	4665
Poisson's ratio	0.21	0.05
Shear stress parameter	-0.5	-0.5
Longitudinal and transverse compressive strength [MPa]	145	24.8
Longit. and transverse tensile strength [MPa]	90	34.8

III. STRASBOURG UNIVERSITY FINITE ELEMENT HEAD MODEL (SUFEHM) VALIDATION

A. Model validation

The experimental data used in order to validate brain behaviour were published by Nahum et al.(1977) [23] for a frontal blow to the head of a seated human cadaver. as shown in figure 2. Intracranial pressures were recorded in this test, at five well defined intra-cranial areas.

In order to reproduce the experimental impact conditions, the anatomical plane of the SUFEHM was inclined about 45°like in the Nahum's experiment. For modelling a direct head impact, the model was frontally impacted by a 5.6 kg rigid cylindrical impactor launched freely with an initial velocity of 6.3 m/s.

A good agreement for the impact force was found as the time duration of impact and the amplitudes were well respected. The comparison of pressure time histories between numerical and experimental data for, five intracranial pressures matched the experimental data

very well. The maximum difference of pressure peak is under 10 %.

Experimental tests carried out by Yoganandan et al. in 1994 has been used in order to validate the ability of the human head finite element model to predict a skull fracture. The surface of the impactor was modelled by a 96 mm diameter rigid sphere. Initial conditions were similar to the experimental ones i.e. a mass of 1.213 kg with an initial speed of 7.1 m/s. The base of the skull was embedded as in the experiment. For the model validation, the contact force and the deflection of the skull at the impact site, were calculated.

In order to validate material and section definition of the skull, Yoganandan's experiment was simulated. The numerical force-deflection curves are compared to the average dynamical response of experimental data. The dynamical model responses agree well with the experimental results, both the fracture force and the stiffness level. The model indicates fracture located around the impact point which complies with experimentall observations.Maintaining the Integrity of the Specifications.

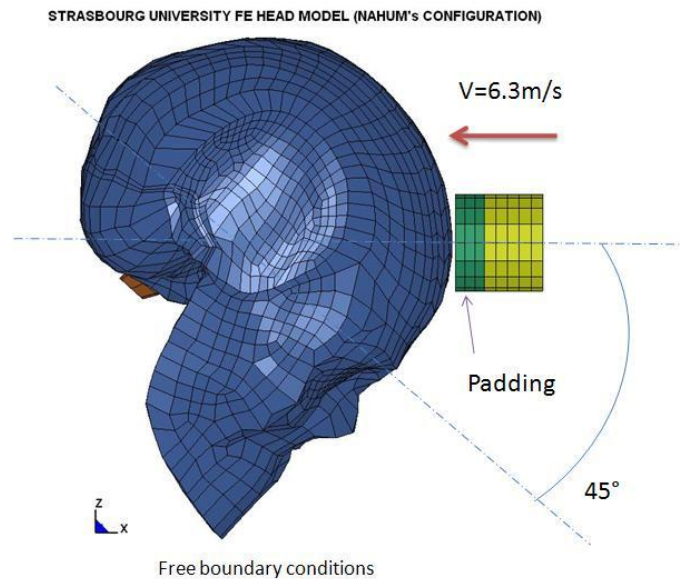


Figure 2: Head model under validation impact (Nahum1977)

IV. MODEL BASED HEAD INJURY CRITERIA

A Methodology

SUFEHM tolerance limits to specific injury mechanisms are available under Radioss code and published by Deck et al. (2008) [19]. The objective here is to propose tolerance limits under Ls-dyna code. For this, 59 head impact conditions that occurred in

motorcyclist, American football and pedestrian accidents were reconstructed with the SUFEHM under Ls-Dyna code.

The reconstructions involved applying the motion of the head from the accidents to the rigid skull of the SUFEHM. Same methodology (statistical analysis) than methodology used by Deck et al. (2008) [19] has been undertaken.

For the statistical analysis the injuries for the accident data were categorised into the following types and levels based on the details of the medical report from each accident case:

- Diffuse axonal injuries (DAI): DAI cases covered all incidences in which neurological injuries occurred and covered concussion, unconsciousness and coma. Incidences of DAI were broken down into mild and severe levels according to coma duration (<24H for moderate DAI and >24H for severe DAI)
- Subdural Haematomas (SDH): This category of injuries covered all incidences in which vascular injuries with bleeding were observed between the brain and the skull of which there were six cases.

B SUFEHM tolerance limits to specific injury mechanism

Results computed with the SUFEHM under Ls-Dyna code are reported in terms of correlation coefficients (Nagelkerke R-Squared values) in order to express their injury prediction capability. Based on SPSS method it appears that DAI are well correlated with intra-cerebral Von Mises stress. Maximal principal strain as well as Von Mises strain presents also an acceptable correlation. Coming to maximum R² values, the maximum Von Mises stress conducts to 0.6 and 0.39 for respectively moderate and severe neurological injury.

The threshold values for this parameter are reported in Table 5 and via the injury risk curves in figure 3.

Concerning the SDH injuries two mechanical parameters, i.e. CSF minimum pressure and CSF strain energy were considered. With the SUFEHM it was shown (Table 4) that the best correlation with SDH was the maximum strain energy within the CSF, with a R² value of 0.465 and a threshold value of about 4950 mJ.

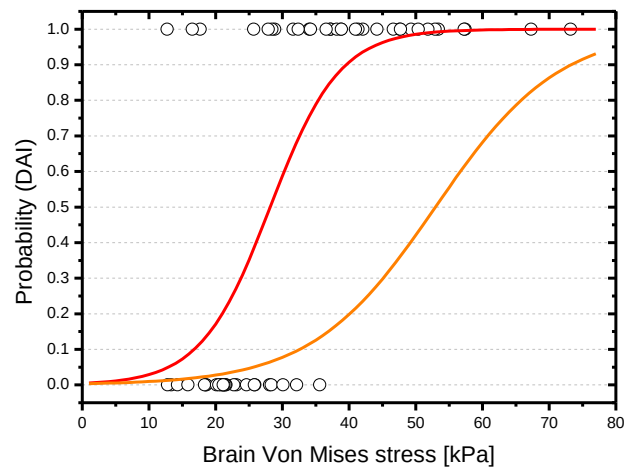


Figure 3: Example of model based head injury tolerance curves corresponding to moderate and severe brain injury

After the analysis of regression correlation method Table 4 and Table 5 report the tolerance limits and the injury risk curves obtained with the SUFEHM for each of the injury types with an injury risk of 50%.

TABLE 3: NAGELKERKE R-SQUARED VALUE FOR THE LOGISTICAL REGRESSIONS BETWEEN THE INJURY PREDICTORS COMPUTED WITH SUFEHM AND THE INJURY DATA.

Injury Predictors	DAI severe (mild)	SDH
CSF minimum pressure		0.367
CSF strain energy		0.465
Peak brain Von Mises stress	0.6 (0.39)	
Peak brain first principal strain	0.43 (0.35)	
Peak brain Von Mises strain	0.43 (0.35)	

TABLE 4: TOLERANCE LIMITS CALCULATED FOR DAI

	Mild DAI	Severe DAI
Brain Von Mises stress [kPa]	28	53
Brain Von Mises strain [%]	30	57
Brain First principal strain [%]	33	67

TABLE 5: TOLERANCE LIMITS CALCULATED FOR SDH

	SDH
Minimum of CSF pressure [kPa]	290
CSF strain energy [mJ]	4950

V. TOWARDS HELMET OPTIMIZATION

A New helmet test method

Based on the previous head injury criteria, a new helmet evaluation and optimization method has been suggested by Deck et al. 2012 [25]. In the proposed approach the experimental linear and rotational head acceleration constitutes the inputs which will drive the head FE model, in charge of the latter to compute the injury parameters related to skull fracture, sub dural haematoma and neurological injury. By this methodology it will be possible to predict head injury risk means a coupled experimental versus virtual evaluation and optimisation procedure as illustrated in figure 4.

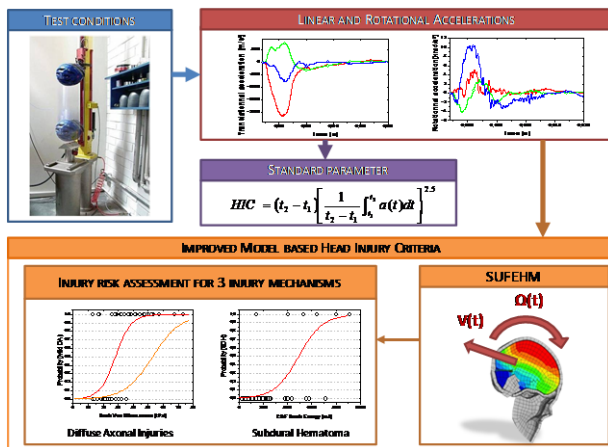


Figure 4: Illustration of the coupled experimental versus virtual helmet test method

B Bicycle helmet performance

In order to evaluate bicycle helmet performance against model based head injury criteria, a finite-element model of a brand new bicycle helmet was developed, implemented and validated under the LS-DYNA® explicit crash code by Milne et al 2012a [26]. The numerical simulation of 90 experimental normative impact tests, under three environmental conditionings and two anvils, was performed by coupling the helmet FEM to a rigid 5.7 kg ISO headform complying with the EN 960 standard. Results in terms of headform acceleration time-history and peak acceleration values between experiment and simulation were in good agreement for most of the impact points thus validating the helmet model. A helmet model validation under tangential impact conditions was conducted as well and reported by Milne et al 2012b [27]. Once validated, this helmet FE model was coupled to the Strasbourg University FE Head Model as shown in figure 5 in order to assess the head injury risks of both DAI and SDH injuries. Results show that normative impacts on flat anvil are more critical than impacts against kerbstone anvil. Results also show that the computed injury risk is acceptable for most of the impact points as shown in figure 6. This work is therefore a step towards both helmet optimisation against biomechanical head injury criteria and the consideration of model based head injury criteria into future helmet standards.

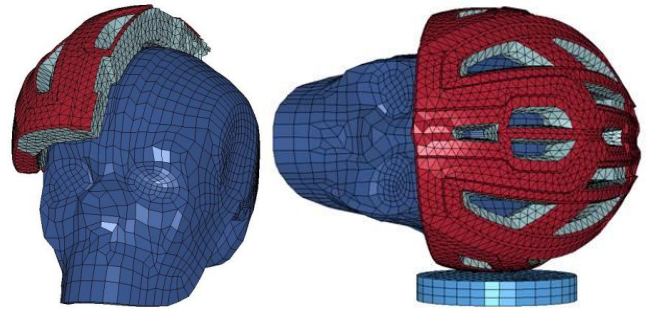


Figure 5: Coupling of Helmet and head models (left) and Positioning of the coupled system for a normative impact on flat anvil (right)

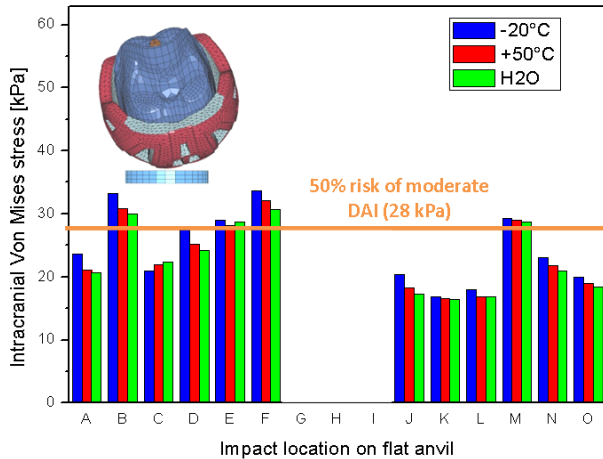


Figure 6: Risks of sustaining moderate neurological injuries during impacts on kerbstone and flat anvils under standard bicycle helmet test conditions.

C Improvment of motorcycle helmet

The above presented model based head injury prediction tool was used also for the development of a new method for enhancing motorcycle helmet performances during an impac by Tinard at al 2012 [28]. In a first step an approved composite helmet finite element model is coupled with the anatomical head finite element model evaluated in terms of injury risks (risks of neurological injuries or subdural haematoma) under normative impact conditions (ECE 22.05 standard). Figure 7 shows that the risk of moderate brain injuries is very high, especially for the lateral impact (point X) even if the considered helmet passes the standard.

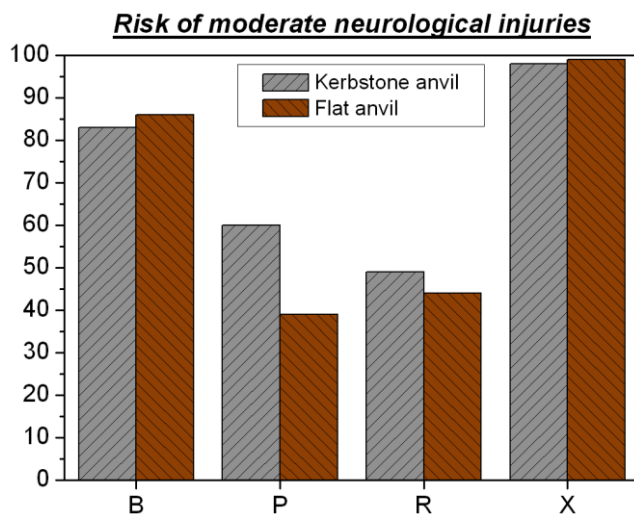


Figure 7: Risks of sustaining moderate neurological injuries during impacts on kerbstone and flat anvils under standard motorcycle helmet test conditions.

A final step illustrated in figure 8 consists in proposing a new method for improving the helmet behaviour in case of impact by focusing on the outer shell characteristics and by assessing the head injury risk with the human head finite element model. A modal analysis of the entire helmet model is performed and makes it possible to define areas of the outer shell to be modified. Under standard impact conditions this new virtual helmet conduces to a very significant decrease of the head injury risk, both in terms of neurological injuries as well as in terms of subdural haematoma.

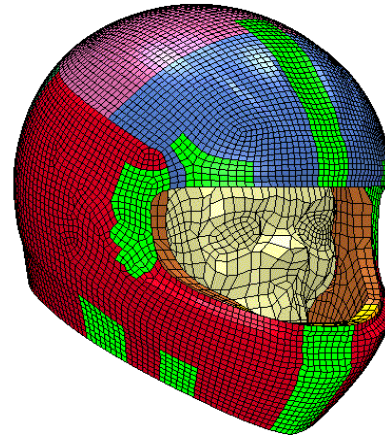


Figure 8: Illustration of the coupled head motorcycle helmet model for optimization purposes.

VI. CONCLUSION

In this study the Strasbourg University Finite Element Head Model (SUFEHM) has been presented and validated..

In an attempt to develop model based head injury criteria a total of 59 real world head trauma that occurred in motorcyclist, American football and pedestrian accidents were reconstructed with SUFEHM.

Two tolerance limits to specific injury have been computed for a 50%injury risk:

- A maximum Von Mises stress value of 28 kPa for moderate DAI and 53 kPa for severe DAI.
- A maximum CSF strain energy of 4950 mJ for SDH.

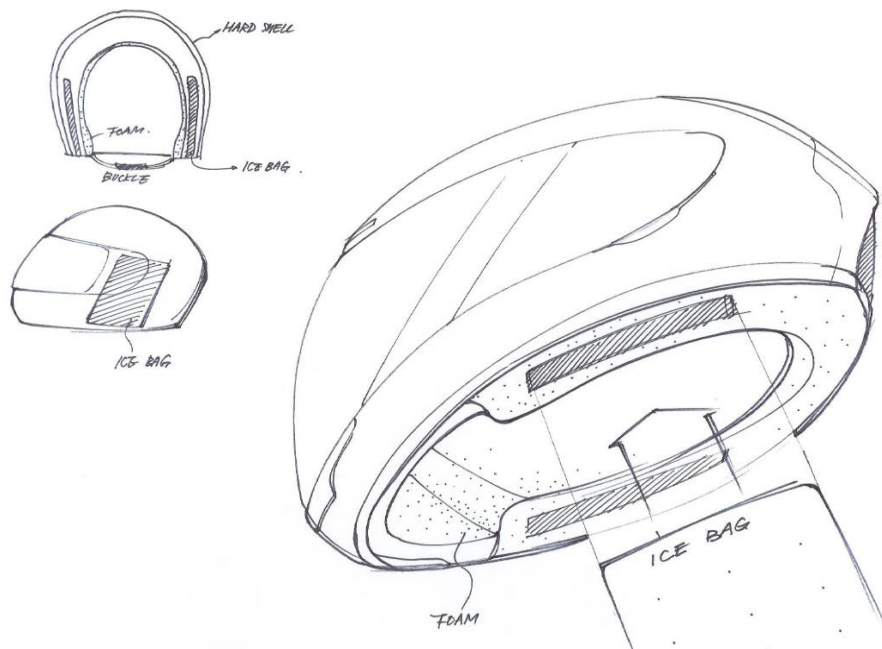
Finally the proposed model based head injury criteria have been applied in the context of an attempt of experimental and numerical head helmet evaluation and optimization.

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Helmet Performance and Design



2

A Review of Blast Induced Traumatic Brain Injury Research

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ABSTRACT

This paper presents an overview of the state of scientific knowledge of blast induced Traumatic Brain Injury (TBI), as well as a review of the current helmet standards and various studies into physically modelling an impact to the head. Past research conducted on TBI present key gaps concerning shockwave induced traumas and the resulting injuries. Kinematics, the study of relative motion between the different entities of the brain, is an area of weak but important research. This, along with identifying the mechanisms of blast transfer to the brain is imperative such to define the injuries that arise. Through comparison and evaluation of the Economic Commission of Europe (ECE) 22.05, the Department of Transport (DOT) and the Snell Foundation testing standards for helmet safety performance, several key limitations have been identified. The first being the wide use of a non-deformable headform, leading to an unrealistic impact response and injury prediction. Furthermore, current head injury criteria used in these testing standards are based only on one injury parameter, translational acceleration, increasing the likelihood of an inaccurate TBI prediction [1].

Key words: *Traumatic Brain Injury; Blasts; Shockwaves; Headforms; Testing Standards; Head Injury Criterion; Military Helmet; Primary Blast Injury*

I. INTRODUCTION

Traumatic Brain Injury (TBI) is often referred to as the signature injury of modern war, and with good

reason. Terrorist blast and landmine injuries have become more and more common in recent decades, particularly in the recent Iraq conflict due to the widespread use of the improvised explosive device (IED). Since 2003 approximately half of the US and coalition forces casualties can be attributed to these devices [2] while 52% of soldiers suffering from blast related injuries suffer some degree of TBI [3]. However preventative measures are limited by a lack of knowledge of these complex events.

Much of the research available is focused on musculoskeletal injuries and blast effects on the lung as opposed to TBI, what is available is mostly concerned with investigating blunt impact or bullet penetration. Overall there is a clear lack of research into the effects of primary blast injury (injuries caused by the pressure wave) on the head. Much of what is available uses Finite Element modelling [4] [5] but very few of the models have been validated against cadaver data, which makes their accuracy questionable and there is currently a lack of physical modelling of blast impacts on the head using cadavers or headforms. Military helmet testing standards in the US and Britain currently do not take primary blast injury into account, despite emerging proof that it is a major cause of TBI therefore further studies are needed to increase understanding of the blast injury physics and aid in the development of preventative measures [1] [6].

II. TESTING STANDARDS

Many studies have been carried out investigating the testing standards used in safety performance of helmets and ways to make them more effective. The investigations can be categorised into two sections; those

that wish to improve helmet performance through the use of the current standards and those that wish to improve the current standards in terms of the realism of the impact/headform and the injury criteria used, leading to the improvement of helmet performance as an effect. Therefore, it is necessary to evaluate and compare the current standards and determine the areas in which they could be developed. Current testing standards are based around a minimum performance requirement. Helmets have to pass a set of standard tests before they can go on the market, meaning manufacturers design helmets to meet minimum established requirements for safety, reliability and function rather than aiming to prevent the maximum level of injury. When discussing testing standards we need to take into consideration two main areas; Military and Motorcycle helmets.

In the UK, the National Institute of Justice (NIJ) testing standards are used for testing the safety performance of military helmets [7] producing a classification covering the ballistic threat of a bullet by its composition, shape, calibre, mass and impact velocity. The NIJ standards created by the Law Enforcement Standards Laboratory (LESL) of the National Bureau of Standards are continually reviewed and are subjected to change at the discretion of the Program Manager at NIJ. They are based on penetration testing preventing injury from gunfire rather than looking at preventing brain trauma caused in a blast shockwave. Penetration is defined in the standards report as being 'perforation of a witness plate by any part of the test specimen or test bullet, as determined by the passage of light when the witness plate is held up to a 60-W light bulb.' The NIJ standards also carry out a ballistic impact attenuation test in which an accelerometer is placed at the centre of gravity of a headform that is mounted in such a way that it is free to move in the direction of the bullet (Figure 1). Four rounds of bullets are then fired at the helmeted headform; one to the front, one to the back and one to each side with the velocity of each hit and corresponding headform acceleration being recorded. If the acceleration of the headform does not exceed a threshold value of 320G where G is the standard acceleration of free fall, then the helmet passes the safety standards. This approach does not give a sufficient or realistic view of potential TBI, and simply sets a minimal level of safety performance for the Mark 7A ballistic helmet.

There are a number of testing standards used around the world for measuring Motorcycle Helmet safety; the two predominant being the Department of Transportation (DOT) and the Economic Commission for Europe (ECE)

22.05. The DOT standards are based upon the Federal Motor Vehicle Safety Standard 218 (FMVSS 218) and are required by law in the U.S.A. The Economic Commission for Europe (ECE) 22.05 are the leading standards in Europe and required by law in over 50 countries. Both of these standards are based around a drop impact test; the dropping of a mounted helmeted headform onto a range of fixed steel anvils. An accelerometer is fixed at the centre of gravity of the headform with the acceleration of the headform and peak acceleration being recorded for the duration of impact. From this a graph of acceleration (in terms of G) versus time is plotted with criterion set that the helmet must meet; for DOT and ECE these include peak acceleration and duration of acceleration limit. There is an allowed peak acceleration of 250G for DOT and 275G for ECE from which the tester can determine whether the helmet is fit for market. However, there are key differences seen when comparing the two standards (Table 1); the DOT standard also requires a penetration test whereas the ECE one is based solely on the impact test thus mimicking more than one aspect of crash incidents and ensuring a wider range of helmet safety performance. Secondly, the headforms that are used vary widely, with the ECE more closely replicating a human head and the DOT being wider and flatter. As described throughout this paper the mechanical properties and anthropometrics of the headform have a great impact on the results obtained in testing and therefore these standards would produce varying results for the same impact parameters. Overall, the ECE 22.05 is the standard favoured internationally due to its mandatory batch testing, thus giving customers the assurance of quality through the eradication of the random testing performed by the other standards.

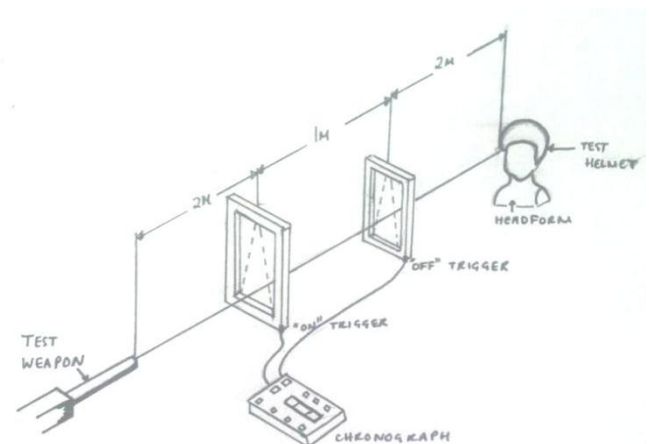


Figure 1: Test setup for NIJ Ballistic Impact Attenuation Test [7]

Alongside the law enforced testing standards, the Snell Memorial Foundation has produced a set of voluntary standards based around scientific research and education which manufactures can choose to meet. The Snell testing standards revolve around four main elements [8] critical for the protective aspect of helmets: Impact Management, Helmet Positional Stability, Retention System Strength and Extent of Protection. They use a wider range of anvils in their testing to simulate different incidents of impact, as well as a chin bar test and a penetration test. This means that helmets that have passed the Snell standards have been tested for varying locations and intensities of impact, and thus have proven effective for a wider range of incidents. However, even though the Snell Testing standards offer some improvement to current required standards, the report COST 327: Motorcycle Helmet safety [9] identified improved testing standards as the main factor in increasing the effectiveness of helmets. The current standards all have the same main limitations; the use of a non-deformable headform, hence unrealistic injury response and the use of outdated injury criteria. In the DOT, ECE 22.05 and NIJ testing standards base injury criteria solely on linear acceleration to determine the level of injury, thus, not taking into account rotational acceleration. Studies including those in the COST 327 paper have shown that both the linear and rotational acceleration occur together causing TBI. Furthermore, in various studies rotational acceleration is identified as the main cause for specific TBI such as diffuse axonal injury (DAI) [10]. Therefore, further work needs to be carried out in order to determine a threshold for rotational acceleration and incorporation of this into the testing standards. [11]

III. HEAD INJURY CRITERIA

There are several methods for quantifying head injury in literature and this paper will focus on those that can predict closed traumatic brain injury (without skull fracture), although there are others that use parameters like skull deformation and fracture as a measure of injury. The criterion that apply to closed TBI can be broadly divided into two categories, acceleration based head injury criteria and stress/strain based head injury criteria

A. ACCELERATION BASED HEAD INJURY CRITERIA

1) Maximum resultant head acceleration (MRHA)

Often used because of its simplicity, the maximum resultant head acceleration is perhaps the most basic of the standards; injury thresholds vary and are defined by application. The ECE.R-22 criterion (the standard in over 50 European countries) uses a maximum

acceleration of 275g (measured from a headforms centre of gravity) [12] as a failure value, although fatal injuries have been estimated at values as low as 200g, values which are consistent with data published by Newman [13]. It should however be noted that ECE.R-22 uses other criterion such as HIC (below) in conjunction with maximum acceleration. A major deficiency in the MRHA criterion is that it doesn't take acceleration duration into account, a deficit that was criticised by Lissner [14] as 'not even valid for rigid engineering materials'. The human head has a higher resonant frequency than the rest of the body and so cannot tolerate as high a change in velocity under impact acceleration conditions [15] which suggests that acceleration duration is also a key factor. An expansion of the criterion that is sometimes used adds an additional parameter saying that there are several threshold acceleration values that should not be exceeded for more than a certain duration.

2) Head Injury Criterion (HIC)

The HIC is the most widely used measure of head injury, appearing in many product specifications and automotive regulations. The HIC is based off the Wayne State Tolerance Curve (WTSC) and the Gadd Severity Index (GSI). It uses a magnitude-duration acceleration curve (like the WTSC) but takes the integral of the curve with respect to time. This means it can be used to model injury for complex impact functions and multiple impacts as it takes acceleration history and duration into account.

The HIC is defined as

$$HIC = \left(\frac{\int_{t_1}^{t_2} a(t) dt}{t_2 - t_1} \right)^{2.5} \times (t_2 - t_1) \quad (1)$$

Where t_1 and t_2 are two time values in the experiment selected so as to maximise the HIC value.

The HIC has been shown as a reasonable measure of head injury severity [9], however it has been criticized because it does not accurately show the effect of the impact direction or account for the effects of rotational acceleration. Also the HIC was created based on skull fracture (although it can be applied to closed TBI), but it is now well established that serious TBI can occur without skull fracture.

Standard	Year	Drop Test Apparatus	Headforms	Headform Sizes	Drop Weight Assembly	Anvils	Impact Criteria	Number of Impacts	Failure Criteria
FMVSS No. 218 (Basis for D.O.T)	1988	Monorail	D.O.T.	Small Medium Large	3.5 kg 5.0 kg 6.1 kg	Flat Hemisphere	Velocity Flat: 6.0 m/s Hemisphere: 5.2 m/s	4 different sites, 2 impacts at each	≤ 400 g peak acceleration < 2.0 ms at 200 g < 4.0 ms at 150 g
ANSI Z90.1	1992	Monorail or Guide-Wire	D.O.T or ISO	Small Medium Large or A, E, J, M	5.0 kg ²	Flat Hemisphere	Velocity Flat: 6.9 m/s Hemisphere: 6.0 m/s	4 different sites, 2 impacts at each	≤ 300 g peak acceleration
AS 1698	1988	'Guided Fall' ¹	Magnesium AS 2512.1 (D.O.T)	A, B, C, D	3.5 kg 4.0 kg 5.0 kg 6.0 kg	Flat Hemisphere	Drop Height Flat: 1830 mm Hemisphere: 1385 mm	4 different sites, 2 impacts at each	≤ 300 g peak acceleration < 3.0 ms at 200 g < 6.0 ms at 150 g
BS 6658	1985	'Guided Fall' ¹	ISO	A, E, J, M	5.0 kg	Flat Hemisphere (same for both types A and B)	Velocity Type A Flat: 7.5 m/s then 5.3 m/s Hemisphere: 7.0 m/s then 5.0 m/s Type B Flat: 6.5 m/s then 4.6 m/s Hemisphere: 6.0 m/s then 4.3 m/s	3 different sites, 2 impacts at each (the same anvil has to be used for each drop)	≤ 300 g (the multi part shells must remain intact)
CAN3-D230	1985	'Guided Fall' ¹	ISO	A, E, J, M	5.0 kg	Flat Hemisphere	Velocity Flat: 5.1 m/s then 7.2 m/s Hemisphere: 4.3 m/s then 6.1 m/s	4 different sites, 2 impacts at each	Lower velocity: ≤ 200 g peak acceleration Higher velocity: ≤ 300 g peak acceleration
Snell M-95	1995	Monorail or Guide-Wire	ISO	A, E, J, M	≥ 5.0 kg ≤ 6.5 kg	Flat Hemisphere Edge	Energy Flat and Hemisphere: 150J then 110J Edge: 150J	Flat and Hemisphere: 4 different sites, 2 impacts at each Edge: 1 impact at 1 site	≤ 300 g
ECE 22.4	1995	Unrestrained Headform with Tri-axial accelerometer at centre of gravity	ISO	A, E, J, M, O	3.1 kg 4.1 kg 4.7 kg 5.6 kg 6.0 kg	Flat Curb	Velocity 7.5 m/s for both anvils	4 sites in sequence with a 5 th test at 4 m/s (flat) or 8.5 m/s (curb)	Resultant: ≤ 275 g HIC not to exceed 2400

1. Apparatus not specified
2. Small and Large D.O.T headforms currently not available in 5 kg

TABLE 1: TABLE OF COMPARISON FOR DIFFERENT TESTING STANDARDS [39]

Despite criticism, in a report by the European Commission for Energy and Transport [9] the HIC was shown to be a more accurate injury indicator during standard drop tests and for estimations based on existing accident data than individual parameters such as peak linear acceleration, peak rotational acceleration, rotational velocity and impact velocity as well as outperforming the competing criterion GAMBIT. The report suggested a HIC value of 1000 as an injury threshold.

A major limitation of all of the above measures of head injury is that none of them take rotational acceleration into account, however many investigations consider shear strains resulting from rotational accelerations as the primary cause of a concussion [16] [17] [18] [19] [20] [9] as well as several other types of TBI, particularly acute subdural haematoma and diffuse brain injury [21]. There are 2 additional common acceleration based criterion that do take rotational acceleration into account.

3) Generalised Acceleration Model for Brain Injury Threshold (GAMBIT)

The GAMBIT assumes translational and rotational acceleration equally and independently contribute to TBI, and combine the two into an inequality that must be satisfied for the injury risk to be below acceptable levels.

$$G = \frac{|a_m|}{a_c} + \frac{|\dot{a}_m|}{\ddot{a}_c} \leq 1 \quad (2)$$

Where a_m is the mean value for translational acceleration

$\dot{a}_m$ is the mean value for rotational acceleration.

a_c is the maximum translational acceleration (250 g)

$\ddot{a}_c$ is the maximum rotational acceleration (10,000 rad/s²)

While GAMBIT has been around for a while (first published by Newman 1986) it has never been validated to a high degree as a head injury criterion

4) Head Injury Power (HIP)

Newman also theorised [22] that the rate of change of translational and rotational energy (power) could be a good measure of head injury and developed empirical expression for head injury power for each degree of freedom.

$$\begin{aligned} HIP = & 4.5a_x \int a_x dt + 4.5a_y \int a_y dt \\ & + 4.5a_z \int a_z dt + \\ & 0.016 \alpha_x \int \alpha_x dt + 0.024 \alpha_y \int \alpha_y dt \\ & + 0.022 \alpha_z \int \alpha_z dt \end{aligned} \quad (3)$$

There are six terms in the HIP equation, corresponding to linier acceleration in the x, y and z directions, and angular acceleration around the x, y and z axis, all of which sum to give an absolute value for HIP. The coefficients in the equation ideally represent the heads directional sensitivity to damage, however values for this are currently not well researched so the ones in the above equation represent the average mass and moments of inertia of the human head, in the above equation a denotes translational acceleration and α donates rotational acceleration. As of yet the HIP is only validated for mild TBI. [1]

B. STRESS/STRAIN INJURY BASED CRITERIA

Much of the head injury research carried out has been focused on the effects of acceleration, however more recently it has been theorised that internal stresses and strains in the brain can provide a good indicator of TBI. Stress and strain based models show promise as indicators of blast induced TBI since the transient pressure wave that characterises primary blast injury creates internal stresses and strains in the brain that lead to injury, particularly Diffuse Axonal Shearing (DAI) [23] which has been identified as one of the major injuries characteristic of blast exposure [24]. These methods are only applicable to Finite Element models as there are currently no validated commercially available headforms measuring stress, strain or pressure, although there are some prototypes being developed [25]. However no FE models have been extensively validated against cadaver data for all the mechanical properties they are expected to simulate and most are just the head in isolation, failing to model the effect of the neck and thorax.

All the stress-strain based injury criteria are qualitative techniques rather than established protocols, as such the thresholds used need to be defined by the experimenter and if possible validated against cadaver data, either in literature or from other experiments.

1) Cumulative Strain Damage Measure (CSDM)

The Cumulative Strain Damage Measure (CSDM) sums the strain across the whole brain to give an injury

value. It works on the assumption that diffuse axonal injury (DAI) is directly related to the cumulative volume fraction of the brain experiencing strains in excess of 15%. It was found that if 5% of the brain mass exceeded this stress, that is a CSDM level of 5 corresponds to mild DAI and a CSDM level of 22 (so 22%) corresponds to moderate DAI [23]. Modified forms of the CSDM where the Cumulative Strain Damage are summed across regions of the brain rather than the whole brain have been used by [26] to model impact TBI in a human FE model and [27] in a rat FE model.

2) *Dilatation Damage Measure (DDM)*

This works off the cumulative volume fraction of the brain experiencing negative pressure levels greater than a specified amount (Bandak et al [23] suggested 5% of brain volume at -14.7 psi) as an injury threshold. This is potentially applicable to the modelling of primary blast injuries as the injury mechanism for them is primarily a transient pressure wave.

3) *Relative Motion Damage Measure (RMDM)*

This measure the motion of the brain surface caused by both translational and rotational accelerations of the head as this motion is suspected to cause subdural haematomas [10]. It is a good criterion for measuring the likelihood of contusions caused by the brain impacting on the inside of the skull.

4) *Intracranial pressure*

In cadaver experiments (reviewed by Horgan in 2005) it was shown that a pressure gradient is set up in the cranial cavity under impact loading [28] which suggests it can be used as an indicator for head injury. While not extensively tested, threshold values have been proposed for different degrees of injury in [9] [29] [30] and are summarised in [28]

5) *Von Mises stress and strain energy*

Suggested by Willinger and Baumgartner (2003) it was found that the internal energy levels in the cerebrospinal fluid layer were a valid predictor for a Sub-Dural Hemorrhage. However this criterion has not been extensively validated or tested as a predictor for other forms of brain injury [31] [32] [28].

IV. BRAIN INJURY

Traumatic Brain Injury is caused by a severe impact to the head which has devastating effects on brain function. With the brains complexity, the injuries that arise are not black and white, as a range of injuries arise from different impacts.

TBI can be categorised into four main groups: Primary, Secondary, Tertiary and Quaternary. Primary

TBIs are a consequence of subjection to overpressure waves or shockwaves. They also fall into the category of closed TBIs where by the skull remains intact. The resultant features of this kind may not be obvious at first. The injuries caused are predominantly internal caused by a rapid pressure increase and acceleration of the head when subjected to a blast. Secondary TBI arises from impact of fragmentation resulting in penetrating wounds causing external and internal bleeding. From the blast, solids and liquids are rapidly converted into a highly pressurised and heated gas. This is called a blast wind and it is able to launch objects a considerable distance, leading to Tertiary TBI. Quaternary TBI incorporates all other injuries which fall outside the other groups such as burns [33].

A common feature found in TBI patients is a Coup-Contrecoup contusion. This type of injury is associated with a force being applied directly to the skull causing a cerebral contusion and bruising of the brain. When moving objects come into contact with a stationary head, Coup contusions transpire. Contrecoup contusions arise when a moving head hits a stationary object, the head stops abruptly and the brain collides with the inside of the skull causing bruising. When talking about Coup-Contrecoup contusions, Coup contusions happen at the site of impact and as a result the brain 'bounces' off the wall at the back of skull leading to Contrecoup contusions as shown in Figure 2. These contusions result in two major problems.

The first involves shearing of axons and blood vessels whereby acceleration of the head induces a vast amount of force that damages such components and jeopardises their function. If blood vessels are damaged, Haematomas and Oedema can arise. Following this, if blood rich with oxygen and nutrients cannot get to certain areas of brain tissue, those areas will become completely depleted of substances, leading to tissue death. Tissue death in the brain has detrimental effects whereby the individual's cognitive and physical ability is affected. In addition to shear forces, it is worth noting that there are also rotational forces that account for major injuries to the cerebral white matter and brain stem structure. However, in past research, the focus has been brought upon linear acceleration, disregarding the effects of rotational acceleration. This has skewed the data accumulated from such research as it has been recently noted that the angle at which the head is impacted and hence the rotational acceleration it goes through, is imperative for TBI.

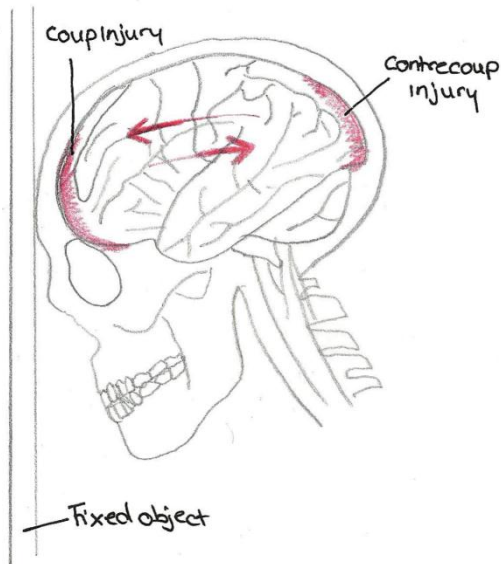


Figure 2: Coup Contrecoup contusion

The second issue concerns the increase pressure gradient within the skull. From the blast, internal bleeding is fairly common. The blood itself acts as an irritant and causes inflammation. The swelling increases the pressure and as there is a limited space within the skull, the cortex is pushed against the skull. Acceleration of the head also induces lacerations of the frontal and temporal lobes. This again, causes shearing of axons and blood vessels, damaging their function, leading to further brain damage.

There have been several experiments undertaken in order to classify the different mechanisms by which the blast transpires to the brain. The three main mechanisms explored are as followed:

- *Thoracic Mechanism:* Cripps and Cooper [34] conducted an experiment using pigs to find a relationship between lung injury and the peak acceleration of the lateral thoracic wall. This was implemented by directly coupling the incident shockwave into the thoracic wall and measuring the resultant acceleration. The results found that the shockwave induced a large pressure force upon the thorax causing blood to rush into the brain, inducing an increased intra cranial pressure.
- *Cranial Mechanism:* Chavko et al [35] subjected rats to a shockwave of 40kPa, enough to cause TBI. Pressure sensors were placed in the third cerebral ventricle of the brain, which detected

short pressure waves of magnitude large enough to cause nervous tissue damage.

- *Head Acceleration:* Experiments conducted by Zhang [36] and Krave [37] have produced a link between rotational and translational acceleration and TBI. Stuhmiller et al. [38] used finite element modelling to form a headform and expose it to blast waves. Upon analyse of the simulation, it provided a likely relationship between acceleration of the head and primary TBI [39].

There are key gaps that have been identified within the research that has taken place. The study of Kinematics of the brain is a weak area, where few have managed to identify the relationship of motion between numerous entities of the head i.e. Skin, Skull, Dura and Brain, and the relative injuries. Another major gap highlighted earlier is the rotational acceleration, which the head undergoes when subjected to a blast wave. When evaluating resulting head injuries the linear rotational acceleration is the predominant focus, excluding rotational. However, it has been found that the rotational acceleration is a key mechanism to brain injury [6] [28].

V. HEADFORMS

The success of reducing and preventing traumatic brain injury revolves around the unearthing of a biofidelic, deformable head form that can be used for testing and improvement of testing standards. The first phase of headform development is reviewing the current, available designs, which fall into the following categories:

- Finite Element Modelling (FEM)
- Lumped Parameter Models
- Deformable Models

With the most important models being the Hybrid series (II and III), the Magnesium K1A and the Bimass 150.

Finite Element Modelling (FEM) is the dominant discretization technique in structural mechanics [40]. The basic concept involves taking a complex mathematical problem and subdividing it into disjointed components (finite elements) and response of each element is expressed in terms of a finite number of degrees of freedom and can be solved in relation to one another. FE modelling of head forms takes advantage of the Eulerian and Lagrangian models; with the Eulerian

representing the blast waves and air whilst Lagrangian models represent the solid head form and helmet where applicable. The advantage of this type of modelling is the fact that it is the only model that can predict intracerebral parameters such as pressure, principle stresses/strains and relative displacement of principle head components [9]. A report by Mills and Gilchrist [41] highlighted the fact that FEM took into account the variation of skull thickness at different sites, which allows for a higher precision and use of virtual prototyping when developing head forms. The two main limitations are the lack of material characteristics and the lack of validation against accident injury mechanisms whilst the limited biofidelity in produced head forms is a cause for concern. The report by Strasbourg University [42] identified a key way of modelling the skull by digitising the inner and outer profile of the human skull before using brick elements to simulate the cerebrospinal fluid whilst the successful calibration against the Nahum-Cadaver data in which the FE model was shown to give accurate predictions of all five sites within the brain as examined by Nahum emphasises the precise nature of this type of modelling.

A second type of modelling is the Lumped Parameters model, where the components are broken down into discrete, linear entities. This simplifies the behaviour of the spatially distributed physical systems that approximate the behaviour of the distributed system under certain assumptions. R. Willinger et al [42] stated that Lumped parameter models can be used to identify the parameters which may affect the performance of the helmets in a simple manner which does not call for high computing costs. However they then go on to say that lumped parameters are “not suitable for investigating the geometric aspects of helmets or the stress level in the continuum of materials of which it is made” and that to combat this, the aforementioned finite element model was being preferred. The main downfall of the lumped models is the assumption that the components act as rigid bodies, and their interaction is only through further discrete components such as dampers and springs. The four lumped models that are most prominent in literature are: Wooden Headforms, Aluminium Headforms, Hybrid II and Hybrid III.

The first type of lumped models are completely rigid and fixed, with the wooden headform falling into this category. It's main use is for shock absorption and penetration tests with a fixed head form and helmet assembly. The rigid nature of this set-up results in major limitations in the results, as a rigid head does not take into account the nature of the skin, brain tissue and

different masses associated with the head. Therefore, acceleration of any kind cannot be measured and the accuracy is compromised. Another headform that fits this category is the Aluminum headform, which is mainly used for drop impact tests and is also a rigid object. Fundamentally, the Aluminum headform is used in the classification of the testing standards, and does not focus on the effects on the head, skull and brain. Furthermore, the headform can vary in size and weight from 500-620mm circumference and from 3.1 ± 0.1 - 6.1 ± 0.18 kg, which arguably gives a greater adaptability than the aforementioned Wooden Headform.

The Hybrid models of headforms gives a more complete evaluation of the effects of protrusion on the body and can be seen as more like anthropometric test devices [43], they act as much better mechanical surrogates to the human body than the previous headforms discussed. The Hybrid II dummy is therefore mostly used in crash-tests as a full representation of the human form, although the most applicable part of this would be the head and neck area. Much like the Aluminum Headform, it is mainly used to assess the nature of protective devices as opposed to the fundamental cranial injuries associated. The Hybrid III dummy is more biofidelically faithful, with the emphasis purely on the head and the neck. It is also commonly found within the 50th percentile, however it can come in a range of sizes, something which is a limitation in the Hybrid II (5th percentile- smaller female and 95th percentile- larger male). J.E. Manning et al [44] commented that the Hybrid III has “the correct shape of the head” however the acceleration in the X plane occurs slightly earlier than anticipated with regard to the dummy position on a seat. We plan to thoroughly test the Hybrid III during our implementation plan with respect to the coupling of the head and neck, and how faithful the results are with real published data on human head trauma in blasts.

The final category of headforms that are prominent are the so called ‘Deformable’ Headforms. The BiMass 150 is a more biofaithful dummy head [42]. It evaluates the characteristics of the helmets with regard to the involved specific cranio-encephalic lesion mechanisms and has the advantage of distinguishing between the mass of the brain and that of the skull. There is therefore a decoupling of the brain in relation to the skull, which enables more prominent features of cranial trauma to be explored, as opposed to merely the protective measures. The Hybrid dummy forms are clearly more suited for repeatability, whereas the Biomass headform gives the most realistic representation of injury, although it does

only come in one size. The Bimass headform can also be represented as a lumped parameter model, as seen below in Figure 3.

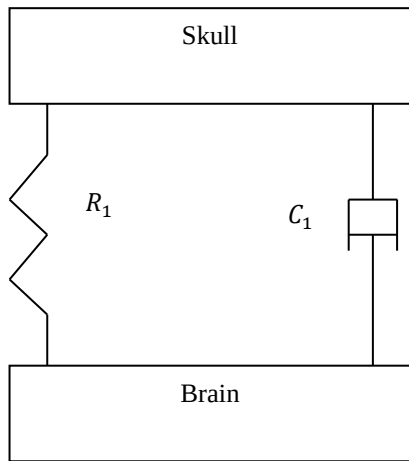


Figure 3: Bimass Headform Component Circuit

The Magnesium K1A head form complies to ISO/DIS 6220 & EN 960 Standards, and is used for uniaxial impact attenuation testing. The headform is available in a number of sizes, and is deformable in the sense that it can undergo a number of tests before failure, unlike many of the headforms dedicated to impact testing. The full headform magnesium K1a is built with a five axis numeric system, which gives the most accurate dimensions, and only one accelerometer is needed for each headform.

Two further headforms that are in development are the JHU and DERAMAN Headform. The JHU Headform is currently being developed by the John Hopkins University. It is composed of a head and flexible neck with its main aim to measure pressure and acceleration under blast loading. No data has been published yet, with an initial prototype having only just been built. The DERAMAN headform is classed as one of the most intelligent headforms; piezoelectric sensors are imbedded within the skull and the brain to record the pressure variations at different positions. The plastic skull incases a polyurethane brain and is covered with a polymer flesh. The headform is currently commercially available but predominantly used for crash tests. However detailed geometric and material properties data has not been released for the DERAMAN, so its accuracy cannot be externally validated and it has yet to be thoroughly tested against crash data from literature. [45] [46] [25]

Due to the increased threat of blast injury from IEDs research is being carried out in several universities into Primary Blast Traumatic Brain Injury (BTBI) with several experimental prototypes for headforms specifically for blast incidents being manufactured and

tested. The most advanced of these studies is being carried out by the Blast Simulation Laboratory of University of Nebraska Lincoln (UNL). The research can be split up into two parts; the first being the design and manufacture of a realistic headform ideal for use in shock wave testing, known as the ‘RED head’ – Realistic Explosive Dummy Headform. It consists of a polyurethane one-piece skull, a PDMS skin simulant and a silicone gel brain model with fibre-optic and PDVF pressure sensors encapsulated within. Due to its material properties and anthropometric accuracy, the headform can be used in realistic experiments for blast shock wave testing. However, the headform has not been externally validated and there have been several suggested modifications to the current design in order to improve biofidelity. The second area of research is the use of the ‘RED head’ in experimental set ups using the UNL shock tube which is capable of reproducing shockwaves produced by an IED blast to monitor and predict the TBI associated with blast incidents. From this they will then be able to research the effects of helmets in the propagation of shock waves throughout the head and design modifications to prevent injury [25].

VI. CONCLUSION

In this paper, the impact of blast induced traumatic brain injury was explored in relation to current testing and helmet standards. An often neglected area of expertise, traumatic brain injury accounts for over half blast injuries sustained, and testing in this area is not sufficient enough to establish a new design of helmet to reduce these injuries. The variation in head injury criteria (from acceleration based to stress/strain based) ultimately makes it difficult to standardize a procedure of design and development of protective headgear to reduce and prevent the effects of blast injury.

Furthermore, the ambiguity in the current headform market is disconcerting for the development of helmet standards to meet blast injury criterion. The lack of biofidelity in the current models used in testing exhibit a lack of faithfulness to the material properties of the skull, brain and surrounding tissue, lowering the degree of accuracy to which they can be tested. This has an impact on the prevention of head injury and consequently, there has been no change in the trends of injuries sustained and resultant fatalities. The main issue is the current lack of deformability, which not only lengthens the testing process, but gives a blurred idea of how injury can be prevented. There is also an omission on the effects of the rest of the body on head injury, with the majority of available literature focusing on a detachable headform without regards to the rest of the body. This again

highlights potential caveats with current standards, with the impact of the rest of the body on absorption and impact unknown to a high extent.

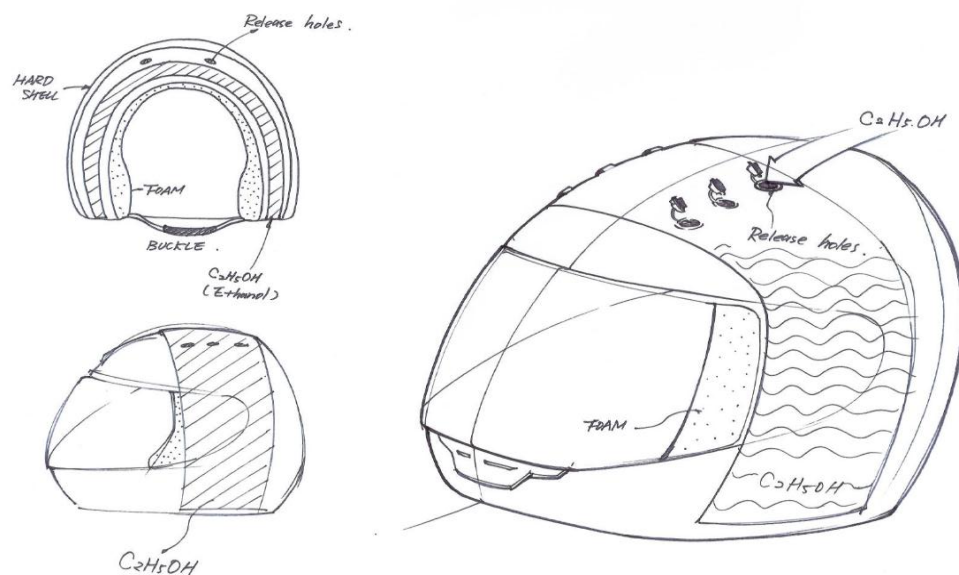
Ultimately, there needs to be a transgression from the available standards if any development will be made regarding the prevention of traumatic brain injury and the design of protective headgear. Willinger et al [42] have begun to explore further by noting the effects of the internal stresses and strains on the brain during blasts, hypothesizing that these effects are arguably more important than head acceleration (ECE.R-22 criterion). [47]

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Helmet Performance and Design



Speculation on the Future of Military Helmet Technology

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ABSTRACT

This article aims to provide a forward-looking technological review and offer informed speculation on the future use of technology to improve the protection offered by ballistic helmets. Also examined is the role of helmets as a platform for augmentation technologies.

Changing protection requirements along with new technologies and novel solutions to current helmet limitations are going to have a dramatic impact on the next generation of helmet designs.

Keywords: *Ballistic; helmet; head-injury, Augmented Reality*

I. INTRODUCTION

To understand why modern ballistic helmets take their current form they must be viewed in context of their development history and previous attitudes to combat helmet design.

Helmets have been a core part of the protective equipment used by soldiers since the First World War [1]. They were reintroduced with the aim to reduce the large number of casualties that had occurred from head injuries such as those sustained from the iconic ‘shrapnel’ shell bursts that were one of the defining weapons of the time.

These primitive helmets comprised of a steel shell with suspension webbing and a chinstrap. They were fundamentally similar in form to medieval helmets such as the ‘sallet’, but offered better protection due to improved metallurgy and industrial metalworking abilities.

By the 1980’s it was clear that steel shell designs were not suitable for the modern battlefield. Previously effective against aerial fragments such as shrapnel (a WWI shell containing shot, not to be confused with popular slang for general fragments), steel helmets provided very little protection against bullets. In 1986 both the USA and Britain made the transition to composite shell helmets with the introduction of the PASGT helmet for the US and the model ‘Mk.6’ for the British.

Ballistic composite shells provided a notable improvement in penetration resistance against fragments and projectiles compared with steel. Since then, improvements in the composite materials used and manufacturing capabilities have reduced the overall weight of helmets. However, there has been little improvement in protection offered.

With the current design of composite helmets there is little room for significant improvements. Consequently, the next generation of helmets will represent a radical change on current designs. It is also likely that some new lateral technologies will be incorporated that will expand the function and form of helmets to be used both on the battlefield and elsewhere such as law-enforcement and security.

II. CURRENT HELMET TECHNOLOGY AND LIMITATIONS

Ballistic helmets are currently comprised of a hard penetration resistant ‘ballistic composite’ outer shell, typically Kevlar or ballistic-weight Nylon. Suspension of the shell on the head varies in form, but commonly is a simple layer of foam padding such as the foam pads used

in the Advanced Combat Helmet (ACH). A chinstrap is present to maintain the helmet in position.

One of the significant problems with current helmets that cannot be overcome by material science or geometry changes alone is the ineffective heat removal from the head in hot ambient conditions. Unhindered, the head dissipates a high proportion of heat and plays a core role in thermoregulation of the body.

Helmets represent an effective barrier to heat loss. Passive air-cooling is grossly ineffective in mild climates, and is all but useless in hotter conditions. In other types of helmets such as those used by motorcyclists with cooling vents, the cooling effect of ventilation yields values in the order of 5W in mild ambient temperatures and moderate wind speeds [2]. The brain is highly thermogenic, typically generating around 20W of heat in an adult. Clearly, there exists a significant deficit in the removal of brain-generated heat from a helmeted head. During physical exertion the head acts as a powerful heat sink for the rest of the body when a helmet is not being worn. . This heat dissipation can exceed a hundred Watts under optimal ambient conditions [3].

Overheating is a serious problem; small increases in body temperature have a notable detrimental impact on performance. Large increases in body temperature lead to thermal fatigue that can result in incapacitation. Thermal fatigue occurs once internal temperatures reach 40.1-40.2°C for oesophageal measurements and 40.7-40.9°C for muscular readings [4].

Based on previous research into helmet cooling, active air-cooling (forced convection and/or chilled air inflow) performs only marginally better than passive airflow and should be discounted. The most viable option identified was found in the combination of compact water cooling for the interface between head and cooling system and Peltier based heat pumps for the transfer of heat from system to the environment.

The need for active cooling highlights one of the most profound changes coming to helmet designs. New helmets with active elements will require a source of power. At first glance this presents a major problem in the added weight and logistical challenges associated with a power supply. This is however not the case; recent technological advances make it probable that the future users will carry a fuel cell. There is also the probability that exoskeleton technology will provide weight-supporting functionality.

There are a set of concussive brain injuries categorised loosely as ‘traumatic brain injury’ (TBI) that are currently the focus of major research efforts. The need to address TBI is rapidly developing into one of the critical military health challenges of the day. Evidence is emerging that TBI and especially repeated exposure causes long-term neurological impairment [5].

The suspension layer between the helmet shell and skull is important in protection against concussive injuries. The role of the shock-absorbing layer in brain injury prevention is well established in motorcycle helmets. Increased padding for military helmets is currently not possible as it would exacerbate the overheating problem to intolerable levels.

It is not uncommon for soldiers to fit custom padding to their helmets, many opting for slimmed down, lighter-weight padding in an effort to make their helmets more comfortable. These soldiers fail to realise is that they are potentially reducing the protection against concussive and blast injuries, however, concussive mechanisms of injury with helmets are not currently well understood. An example of this kind of aftermarket helmet liner system is the ‘ACH Occ-Dial Liner Kit’ by Ops-Core [6].

One possible cause of helmet related concussive injuries arises from the way in which the composite shell deforms in response to impact from a projectile. Deformation of the helmet shell occurs during impact with high-energy fragments.

This deformation can cause the inside of the helmet shell to make direct contact with skull of the wearer and results in transfer of kinetic energy. While full understanding of the mechanisms involved is being researched, it is likely that this mechanism of direct contact is playing a contributing role in concussive head trauma. While too soon to prove definitively. In body armour, this phenomenon is known as ‘behind armour blunt trauma’.

It is likely that not only will we see improvements in the protective capabilities of ballistic helmets, but also in the way we test and evaluate helmet models. Current testing protocol was established during the time of steel shell helmets and focuses entirely on the ability of the helmet shell to resist penetration against projectile impacts. One of the most commonly used is the American National Institute of Justice (NIJ) set of testing standards for ballistic helmets [7].

Given that penetration resistance alone is not sufficient to safeguard against death and injury, more

rigorous testing standards accounting for back-face deformation in helmets would encourage developers to begin to address concussive injuries. This is an important move towards the reduction of concussive injury rates among armed forces personnel.

While the standards remain focussed solely on penetration there is little incentive for manufacturers to invest resources in protecting against other mechanisms of injury associated with combat helmets.

The primary risk that helmets were designed to protect against was falling fragments. This situation has changed to predominantly ground and low-level originating threats. Helmet manufacturers have flirted with the idea of extending protection to cover some of the jaw and more of the face, but so far none of these have made it past trial stages. A full-face helmet will exacerbate the over-heating effect of the helmet.

Facial injuries from high-energy fragments, when not fatal, do not lend themselves to reconstruction and often result in major loss of sensory function such as loss of sight [8, 9] Previous and current designs leave the face totally exposed in favour of unhindered sight and hearing. This exposure to harm is a significant flaw. Once eyesight is impaired through trauma there is often little that can be done to restore it.

There have been recent efforts to introduce ballistic face protection to work with helmets via purely mechanical designs. Examples include the ‘ballistic face mask’ by Inter-American Security Products Inc. [10] and detachable jaw pieces trialled by some manufacturers such as the Ops-Core ‘Moto Mandible’ [11]. There are a number of problems with both of these.

The Kevlar facemask may be fundamentally flawed. While able to stop the penetration of impacting small arms rounds, there is little gap or shock-absorbing material between the composite and the face. There is potentially nowhere for the kinetic energy of the incoming fragment to go but into the tissue and bone structures of the face. Based on the behaviour of similar devices, this might lead to injury analogous to behind armour blunt trauma.

III. HELMET AUGMENTATION TECHNOLOGIES

There is a growing trend to attach many discreet pieces of hardware to helmets in order to provide secondary augmentation functions such as low-light vision and communications. This will continue as new technologies are developed. From a practical viewpoint, there is a finite amount of space and weight per helmet available.

One prevalent design myth is the ‘need’ for a direct line of sight between the soldier and the firearm while shooting. Rifle sights are progressing to superior digital technologies rather than traditional passive glass optics. Combining this with digital display technology in the helmet could allow the soldier to aim the weapon without exposing the head to incoming fire and ballistic threats. Next generation helmets are almost certainly going to have digital display technology incorporated, and so this is a feasible step. This is just one of the ways in which seemingly abstract design changes can have dramatic impacts on soldier behaviour and associated injury risks.

The step towards integration is going to be the single most important design progression required before new technologies can be added to helmets. This is required in order to overcome the current approach whereby each discreet unit monopolises either the eyes or ears and excludes further technological augmentations of that sense. If you combine these devices into an integrated unit you can have additive rather than exclusive choices for secondary functionality.

There have been significant advances in the ability to generate real-time battlefield intelligence through the use of unmanned drones and other devices. The way in which this information is delivered to the combat units ‘on the ground’ will determine the degree to which it provides a combative advantage. One only has to look to civilian smartphones and tablets to realise that it is possible to deliver information dynamically and preconditioned for maximum effectiveness. With the move to integration comes the ability to stream battlefield intelligence directly to soldiers, with the potential to enhance soldier performance.

Up until now helmet were designed with the aim of minimising sensory hindrance to the user. We are now able to enhance and augment senses such as vision and hearing thus providing a significant advantage.

This new technology is forming the basis for new video gaming and military funding is helping this development. Many of these novel technologies are currently being developed for use in other industries. The potential of augmented reality has been identified by both gaming and industry in the last year or so.

The videogames industry has seen renewed efforts to develop virtual reality headsets with many advances as a result. At least one developer has products that are nearing commercial release with the notable example of the Oculus Rift virtual reality headset.

Once helmets have an integrated display and auditory hardware the scope of possible applications is diverse. This technology might include a broader field of view and real time updates of enemy units. Maps and

navigational data updated and connections to remote cameras facilitating better surveillance. Real-time remote diagnostics is another active area of interest for military and civilian industries. Increasingly small, portable, and efficient sensors mean that it is now possible to mount these kinds of devices unobtrusively into equipment like helmets. It is probably not going to be long before we see such applications as helmets fitted with a 'black box' able to report the magnitude of the blast the soldier experienced or the remote monitoring of soldiers' health from afar.

IV. CONCLUSIONS

We have reached the limits of the protection that can be offered with the simple composite shell designs. The current approach has limited effectiveness against penetration by small arms, but is woefully inadequate against blast threats and higher energy fragments. Critical parts of the head such as the face are unguarded. Novel improvements in protection against concussive injuries may arise once the overheating problem is addressed. A number of new technological advances will see helmets take a more active role during use.

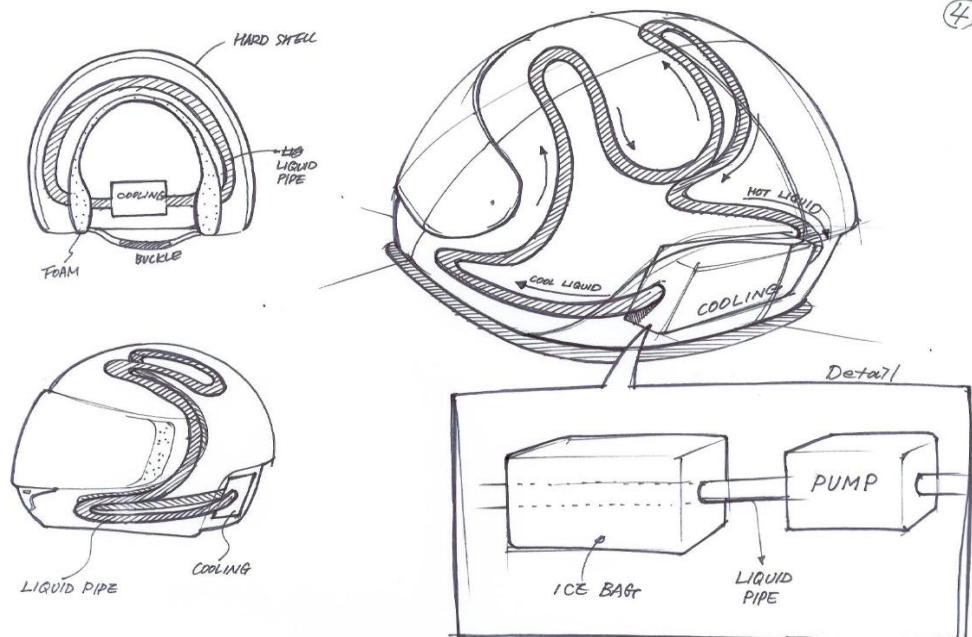
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Helmet Performance and Design



Evaluation of Blunt Impact Protection in a Military Helmet Designed to Offer Blunt & Ballistic Impact Protection

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ABSTRACT

This paper describes both a numerical and an experimental approach to measuring the ballistic and blunt impact protection offered by military helmets. The primary purpose of military helmets is to protect users from ballistic impact but modern military helmets contain a liner that protects against blunt force as well. Altering ballistic shell stiffness, lining the shell with material of different density, even separating the liner from the shell so that they can move independently, all affect the transfer of stress to the head and the resulting strain experienced by the brain. The results of this study suggest that there is potential for a helmet that protects the user from both blunt and ballistic impact that can be further improved by implementing an energy absorbing sliding layer, such as the MIPS system, between the shell and the liner to mitigate the effect of oblique impacts.

Keywords: *military helmet, impact, ballistic, blunt, oblique, ACH, PASGT*

NOMENCLATURE

ACH Advanced Combat Helmet
CSF Cerebral Spinal Fluid
DAI Diffuse Axonal Injury
FE Finite Element

MIPS Multi-directional Impact Protection System
PASGT Personal Armor System for Ground Troops
SDH Subdural hematoma
TBI Traumatic Brain Injury

I. INTRODUCTION

Military helmets consist of a composite shell that provides ballistic protection and a liner separating the shell from the head that provides blunt impact protection and comfort. Even with this protection, military personnel experience injuries to both the skull and brain. US today reported the following:

Last summer, battlefield doctors in Afghanistan diagnosed more than 300 service members per month with concussions or mild traumatic brain injuries and smaller numbers of service members with more moderate or severe head wounds. Concussions are a common wound among troops knocked about inside armored vehicles or flung to the ground while on foot patrols by an explosion from a roadside bomb [1].

The number of diagnosed Traumatic Brain Injuries (TBI) experienced by military personnel is increasing. The United States Department of Defence reports an

increase from 10,936 in the year 2000 to 35,591 in 2011 with approximately 76% of these classified as concussions or moderate TBI [2].

A military helmet shell primarily protects the head from ballistic impacts; today's ballistic helmets can stop handgun bullets and even some rifle rounds from penetrating the outer shell. If the bullet has enough kinetic energy, it can generate a large deformation of the composite shell, causing delamination of the inner layer, potentially resulting in contact between the inside of the helmet shell and the skull. This contact can cause blunt force injury to the skull and brain [3, 4].

Blunt impact protection resulting from contact with hard surfaces is another potential source of injury. A study by Mertz et al [5] estimated a 5% risk of skull fractures for an impact resulting in peak accelerations of 180 gravities (g) and a 40% risk of fractures for 250 g.

Military helmet blunt impact protection is therefore essential. Helmet shells and the lining that separates them from the head must absorb impact energy to protect the soldier's head and help him remain battle ready.

In the US Army Advanced Combat Helmet (ACH) Specification CO/PD-05-04:2007, helmets are dropped straight down on a hemispherical rigid surface with an initial speed of 3m/s (10 fps), equal to a drop from 0.5 m. Helmets that reach peak accelerations less than 150 g are said to pass. Recently military helmet manufacturers have come to understand the need for improved blunt impact protection and are working to develop helmet liners that experience peak accelerations below 150 g when dropped at 4.3 m/s (14.4 fps).

The current U.S. Military helmet, the ACH, is unable to meet this requirement. A study from 2005 investigated the blunt protection offered by two U.S. Military helmets, the Personal Armour System for Ground Troops helmet (PASGT) and the Advanced Combat Helmet (ACH) [6]. Researchers dropped the helmets at 3 m/s (10 fps) and 4.3 m/s (14.4 fps). The result was that the ACH helmet experienced peak accelerations below 150 g at 3 m/s (10 fps) but not 4.3 m/s (14.4 fps). The energy absorption material between the headform and the helmet shell was unable to absorb enough energy.

Blunt impact in a purely radial direction will cause linear acceleration of the head while a purely tangential impact around the head's centre of gravity will cause angular acceleration of the head. In reality it is more likely that an oblique impact will occur, causing both linear and rotational head acceleration. The human brain is sensitive to this rotational motion [7, 8, 9].

The human head has its own safety system to help protect against rotational impacts. During an oblique

impact to the head, the brain slides relative to the skull in the cerebrospinal fluid thereby reducing the rotational acceleration experienced. There are helmets on the market that are designed to reduce the rotational energy transmitted to the brain. The PHPS anti-rotation helmet has a lubricated flexible membrane on the exterior of the helmet shell that decreases the rotational force caused by impact [10]. The Multi-directional Impact Protection System (MIPS) was inspired by the human head and allows the outer ballistic shell to move relative to the liner in the interior [11]. The MIPS system has shown that it reduces stress and strain during oblique impacts. It will be investigated as an example of a concept that has potential to reduce the rotational energy imparted to the head during a blunt impact.

As a result of military specifications and testing methods, current helmets are optimized to reduce linear acceleration of the head and related injuries, such as skull fractures [12]. Rotational motion is not included in any current helmet testing standard and it is not known to what extent current military helmets reduce rotational acceleration during head impact.

Brain tissue can be considered a fluid because its primary mode of deformation is shear and its bulk modulus is roughly 10^5 times greater than its shear modulus [13]. Rotational acceleration may be a better indicator of TBI risk than linear acceleration because common severe injuries, such as subdural haemorrhage and diffuse axonal injury, are more easily caused by severe rotational head motion [14, 15].

The purpose of this paper is to describe numerical and experimental methods of analyzing both perpendicular and oblique ballistic and blunt impact protection properties of military helmets and liners. A liner with the potential of reducing the rotational forces transmitted to the head as a result of an oblique impact will be tested using these methods.

II. METHODOLOGY

A. The Numerical Study

A numerical simulation was used to investigate the interaction between helmet shell and liner during both ballistic and blunt impacts.

1) *Head*: A detailed Finite Element (FE) model of the human head was used to compute the strain in the brain and the stresses in the skull. The head model used in this study was developed at the Royal Institute of Technology in Stockholm [16, 17]. The head model includes the scalp, the skull, the brain, the meninges, the cerebrospinal fluid (CSF) and eleven pairs of the largest parasagittal bridging veins (Fig. 1).

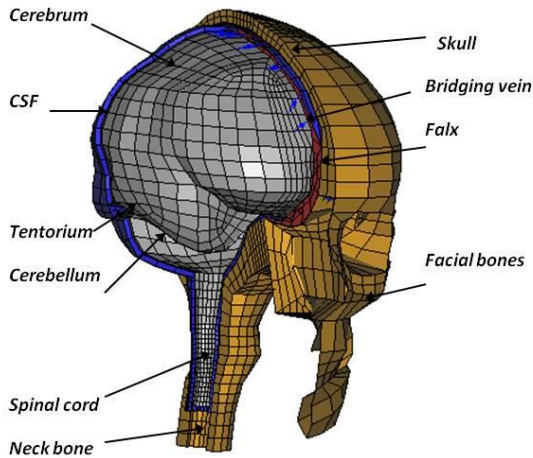


Figure 1: Detailed finite element (FE) model of the human head (Kleiven, 2002, 2007)

2) *Helmet shell & liner:* An FE model of a PASGT helmet was validated and used in an earlier study to analyze the effect of ballistic impact to the helmeted head (Fig. 2) [18]. The outer shell, consisting of reinforced aramide fibers, was modelled using a Chang–Chang Composite Failure model (material type 22 in LS-DYNA). Three failure criteria, namely, matrix cracking, compression failure and fiber breakage were used in this model. Two different shell stiffnesses were used in order to represent the range of material found in military helmets from different manufacturers [18].

- Soft - 35 mm maximum deflection resulting from ballistic impact
- Hard - 25 mm maximum deflection resulting from ballistic impact

Three different helmet liner materials were studied.

- Air – The space between the head and the helmet shell is mainly filled with air, the shell is fixed to the head using an interior system of straps as in the PASGT helmet (Fig. 2)
- EPP24 - Expanded Poly Propylene with a density of 24 kg/m³ is an example of a less stiff foam used in helmets.
- EPP70 - Expanded Poly Propylene with a density of 70 kg/m³ is an example of typical foam stiffness used in bicycle, motorcycle and equestrian helmets.

The LSDYNA material model MAT_CRUSHABLE_FOAM was used to model the EPP liners (Fig. 3).

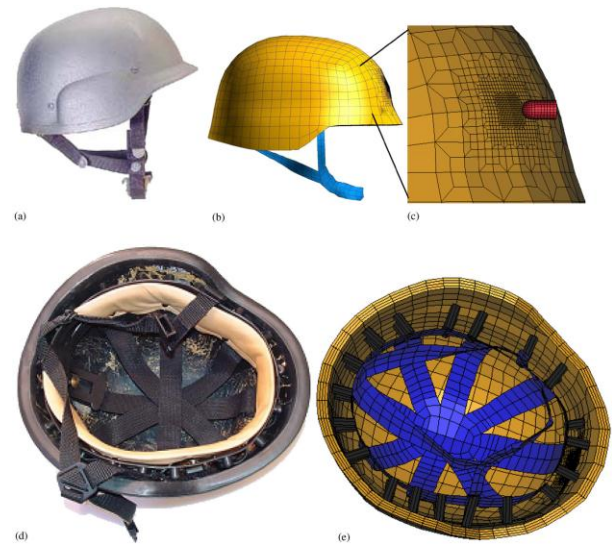


Figure 2: Real ballistic helmet (a), Finite Element model of a ballistic helmet (b & c), and finite element model of the interior (e) (Aare and Kleiven, 2007)

3) *Impact configuration set up:* The numerical simulation set up for both ballistic and blunt impact can be seen in Fig. 4 and 5. In the ballistic impact (Fig. 4), an 8 gram projectile with a speed of 427 m/s was used. Straight (90 degree) and 45 degree oblique impacts at 4m/s and 7m/s were studied. The helmets were dropped towards a rigid steel plate with a coefficient of friction of 0.5 between the plate and the helmet. Table 2 shows the ballistic simulations that were conducted while Table 3 shows the blunt force simulations.

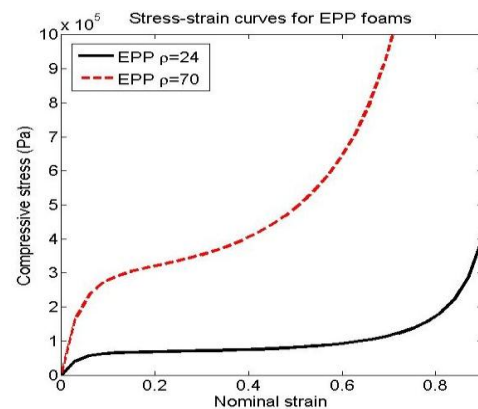


Figure 1: The stress-strain curve for EPP24 and EPP70 used in finite element simulations (reconstructed from the empirical function and data published by Avalle et al, 2007)

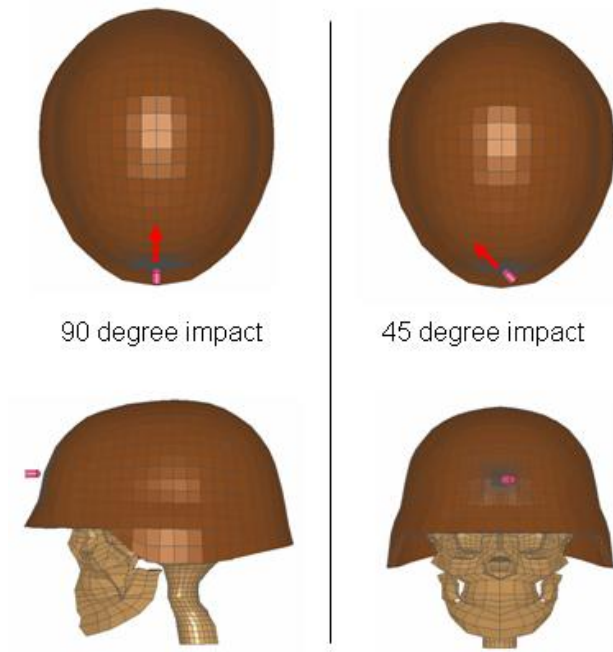


Figure 2: Ballistic impact configurations simulated in numerical study

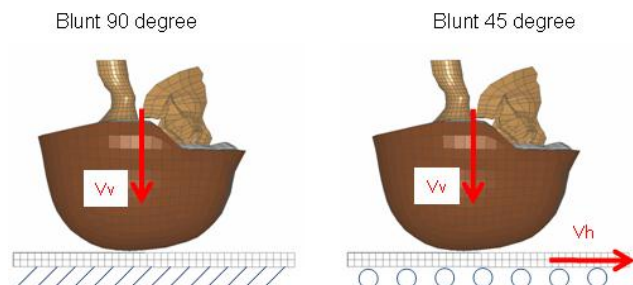


Figure 3: Blunt impact configuration simulated in numerical study

4) *Biomechanical analysis:* In this study a human head model was used to compute the stress in the skull and the strain in the brain tissue. Key elements determined during the simulation included shell elements in the compact bone of the skull with the largest von-Mises stress and the elements in the brain with the highest maximum principal strain. The stress data was filtered using a SAE filter (1000 Hz). The post-processing was done using the commercial software LS-PrePost (LS-PrePost – 3.1, Livermore Software Technology Corp). The strain level in the brain model was compared to real accidents including both mild and severe traumatic brain injuries. A strain level of about 20% in the FE model of the brain is associated with a risk for concussion and a

stress in the skull bone above 90 MPa is related to risk for fracture [17, 19, 20].

5) *Evaluation of MIPS concept:* The helmet FE model was altered to evaluate a helmet liner concept that allowed the outer helmet shell to move relative to interior liner. This was simulated using an EPP70 liner in the hard helmet shell. The EPP70 liner was 1 mm from the outer edge of the helmet, a coefficient of friction of 0.14 was assumed between these surfaces and a blunt impact at 45 degrees with a speed of 7 m/s was used to simulate an oblique impact with the MIPS concept.

B. The Experimental Study

Several ACH military helmets were blunt impacted to investigate whether the MIPS liner, designed to absorb rotational energy, had an effect on accelerations experienced by a headform compared with an EPP50 liner.

1) *Set up of blunt impact tests:* A test rig designed to impact helmets with both translational and rotation acceleration was used [21]. In this test, a free falling instrumented head form impacted a horizontally moving steel plate. The vertical hit to the moving plate resulted in an oblique impact that included both translation and rotational forces. A 50th-percentile Hybrid III dummy head form was fixed into an ACH military helmet. The head form was instrumented with a 9 (3-2-2-2) head accelerometer array designed to measure both translational and rotational acceleration around all axes [22].

2) *Blunt impact configurations tested:* A total of 4 blunt impact configurations (Table 1) were tested. Configurations A and B were oblique tests with both a vertical and horizontal component. In configurations C and D, the plate did not move, resulting in a purely vertical impact.

3) *Modification of the original helmet to improve rotational acceleration energy absorption:* An ACH with a MIPS liner designed to reduce the rotational energy transmitted to the brain as well as an EPP50 liner were tested. The MIPS helmet liner was fixed in an ACH shell by making minor changes to the shell and an EPP50 liner, Fig 6. The rubber edge trim on the shell edge was removed. A low friction liner piece of polycarbonate was attached to an EPP50 liner. This assembly was installed in an ACH ballistic shell using Velcro®.

III. RESULTS

A. The Numerical Study

The 1st principal strain and the von Mises stress for each simulation can be found in Table 2 and 3.

- i. *Translational acceleration:* The simulated translational acceleration was compared between the three liners, Air, EPP24 and EPP70 in Fig. 7a for the 90 degree blunt impact; this corresponded to the maximum principal strain in Fig. 7b. Though accelerations were high for all liners, simulations with the Air Liner showed the highest acceleration and highest strain in the brain.

TABLE I. DETAILS OF THE BLUNT IMPACT SIMULATIONS CONDUCTED

Blunt Impact Tests on ACH						
	Impact angle	Liner type	Number tested	Impact velocity m/s (ft/s)	Vertical speed of head & helmet m/s (ft/s)	Horizontal speed of plate m/s (ft/s)
A	45	MIPS	3	3.0 (10.0)	2.2 (7.1)	2.2 (7.1)
B	45	MIPS	3	4.4 (14.4)	3.1 (10.2)	3.1 (10.2)
C	90	MIPS	3	3.0 (10.0)	3.0 (10.0)	0
D	90	MIPS	3	4.4 (14.1)	4.4 (14.4)	0
A	45	EPP50	3	3.0 (10.0)	2.2 (7.1)	2.2 (7.1)
B	45	EPP50	3	4.4 (14.4)	3.1 (10.2)	3.1 (10.2)
C	90	EPP50	3	3.0 (10.0)	3.0 (10.0)	0
D	90	EPP50	3	4.4 (14.1)	4.4 (14.4)	0

- ii. *Strain distribution in a sagittal cross-section with time:* The strain distribution in a sagittal cross-section of the brain at different times for both the air liner and the EPP70 liner can be found in Fig. 8 and 9 (Simulations 1 and 2). In these simulations, the penetration of the projectile caused the inner surface of the outer shell to come in contact with the scalp of the head. The strain distribution followed the same pattern for both helmet liners (Air and EPP70). Fig. 10 and 11 show the corresponding strain distribution for a 45 degree blunt impact. Force was transmitted

to the head later with the Air Liner (Fig. 10) than the EPP70 liner (Fig. 11).

- iii. *Stress & strain as a function of time:* An example of stress and strain as a function of time for elements with the highest computed skull stress and brain strain can be seen in Fig. 12. The helmet with the EPP70 liner experienced maximum strain and reached this maximum before the helmet with the air liner. Fig. 12b shows two distinct spikes in the stress experienced by the headform in the helmet with the air liner. This double peak was also seen when an EPP liner was present.
- iv. *Maximum stress & strain:* Tables 2 & 3 show the elements with the highest principal strain and the maximum von Mises stress for all ballistic and blunt impacts. Strain levels greater than 20% were noted for several ballistic impacts (Table 3) These were especially high for blunt oblique impacts at 45 degrees in hard shells.
- v. *Ballistic and Shell hardness:* In simulations where the shell impacted the skull, the helmet with the hard shell showed lower stress and strain compared to the helmet with softer shell (Fig. 13 & 14).
- vi. *Ballistic and liner stiffness:* Comparing the results between helmets with Air, EPP24 and EPP70 liners it can be seen that the EPP24 increased the computed strain compared to the helmet with Air, while the EPP70 foam resulted in slightly lower strain (Fig. 13).

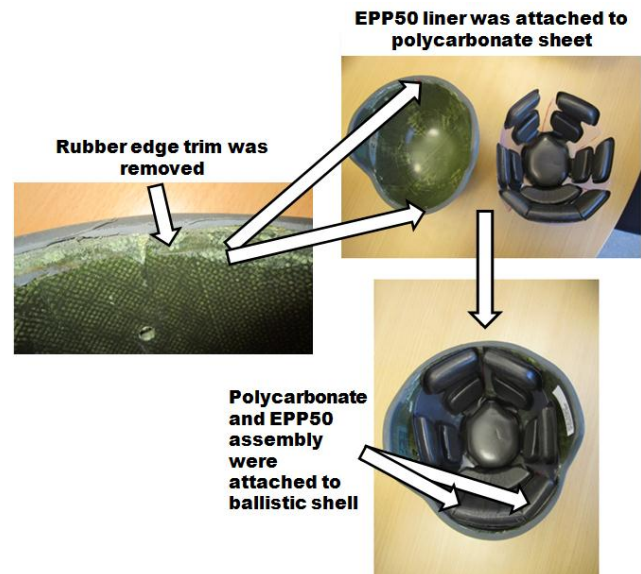


Figure 4: Installation of the MIPS liner in a ACH

- vii. *Blunt impact and liner stiffness:* Liner stiffness played a large role in the resulting stress and strain. For a 90 degree blunt impact (Fig. 14) the EPP24

liner showed a lower strain than the Air Liner. For a 45 degree impact the results were reversed, the Air Liner showed the lowest strain because it reduced the transmission of rotational forces. As speed is increased, the benefit of the liner increased. A helmet with the EPP70 liner reduced strain in the brain more than the Air Liner.

- viii. *Liner vs impact angle:* Comparing the strain in the brain for each helmet liner and shell stiffness, a 45 degree oblique impact resulted in the largest strain for all helmet configurations. Considering stress, the EPP24 and EPP70 liner reduce the stress experienced by the headform.

TABLE II. DETAILS OF THE BALLISTIC IMPACT SIMUALTLIONS CONDUCTED

Ballistic Impact Simulations					
Projectile weight: 8 g Projectile speed: 427 m/s					
Sim ulation	Im pact angle	Liner type	Shell stiffne ss	1 st princip al strain (%)	Von Mises Stress (MPa)
1	90	Air	Hard	5.9	28.1
2	90	EPP7 0	Hard	4.1	25.7
3	90	EPP2 4	Hard	8.4	47.8
4	90	Air	Soft	12.2	58.8
5	90	EPP7 0	Soft	7.8	43.8
6	90	EPP2 4	Soft	12.7	66.9
7	45	Air	Hard	4.9	12.6
8	45	EPP7 0	Hard	4.2	12.3
9	45	EPP2 4	Hard	5.2	19.2
10	45	Air	Soft	8.8	41.1
11	45	EPP7 0	Soft	4.7	22.8
12	45	EPP2 4	Soft	8.3	45.4

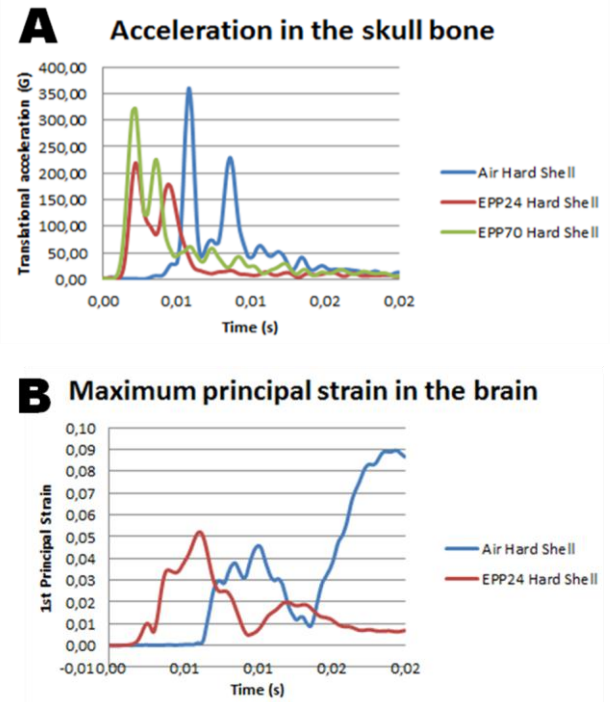


Figure 5: Simulated translational acceleration compared between the three liners, air, EPP24 and EPP70 in a hard shell (a) and the corresponding strain (b)

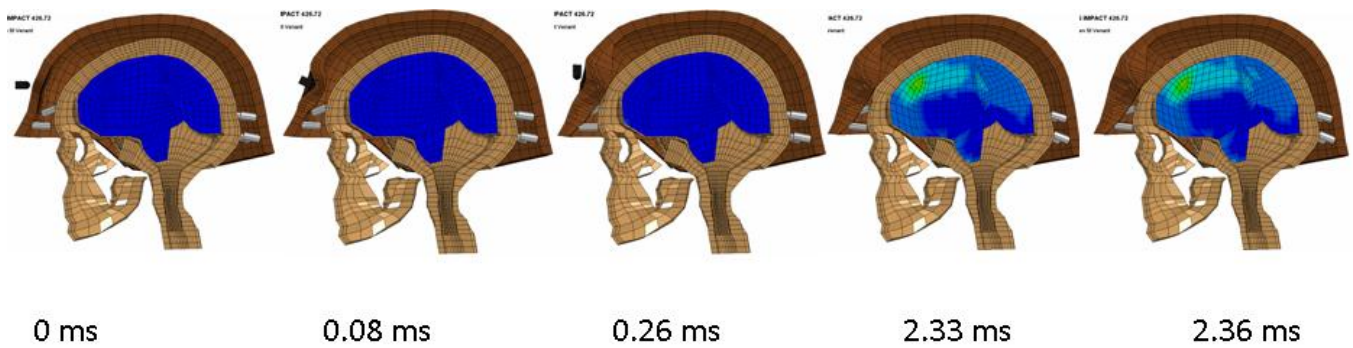


Figure 6: Animation sequence for a 90 degree ballistic impact with a PSGT helmet and Air Liner in a hard shell (Simulation 1).

BLUE = 0% strain, RED = 10% strain

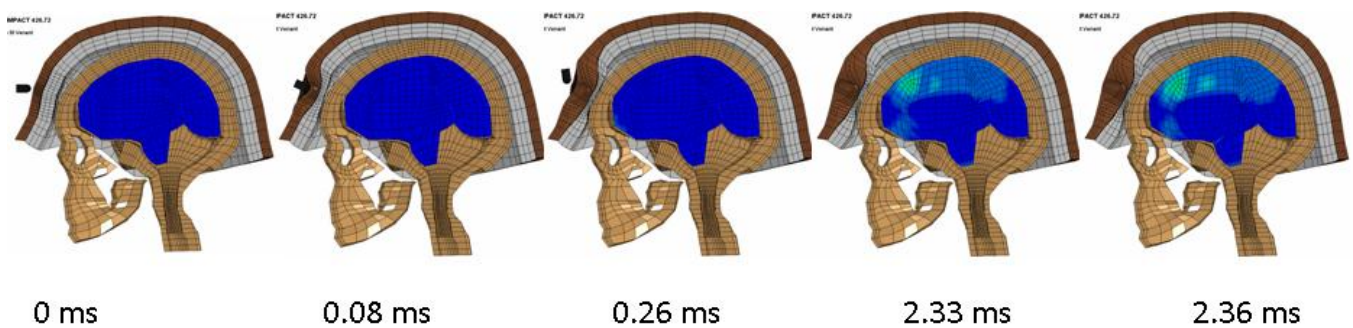


Figure 7: Animation sequence for a 90 degree ballistic impact with a PSGT helmet and EPP70 liner in a hard shell (Simulation 2).

BLUE = 0% strain, RED = 10% strain

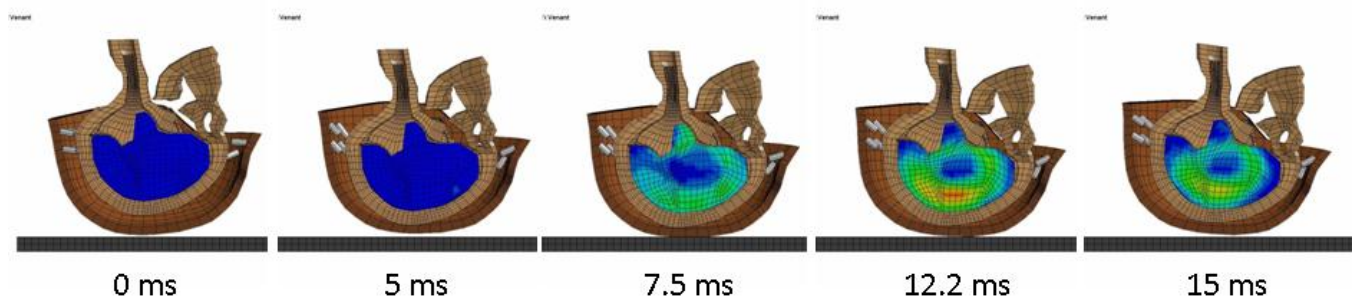


Figure 8: Animation sequence for a 45 degree, 4 m/s blunt impact with a PSGT helmet and EPP70 liner in a hard shell (Simulation 15).

BLUE = 0% strain, RED = 20% strain

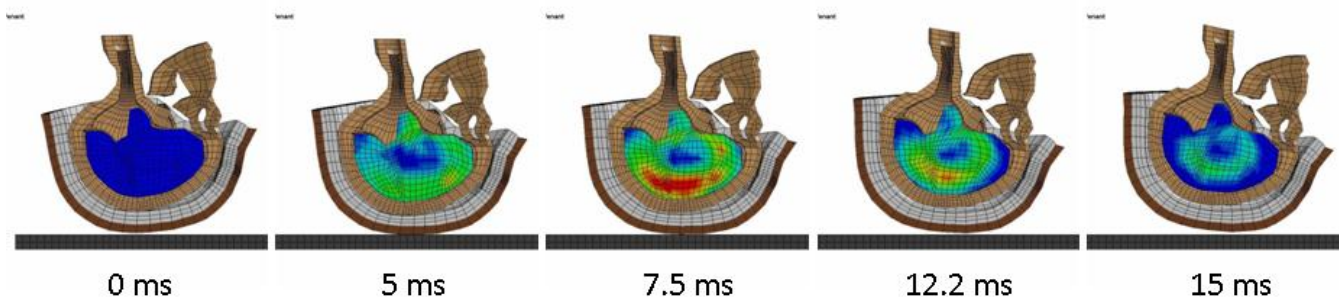


Figure 9: Animation sequence for a 45 degree, 4 m/s blunt impact with a PSGT helmet and EPP70 liner in a hard shell (Simulation 16).

BLUE = 0% strain, RED = 20% strain

TABLE III. DETAILS OF THE BLUNT IMPACT SIMULATIONS CONDUCTED

Blunt Impact Simulations						
Simulation	Impact angle	Liner type	Shell stiffness	Impact velocity (m/s)	1 st principal strain (%)	Von Mises Stress (MPa)
13	90	Air	Hard	4	9.0	78.5
14	90	EPP70	Hard	4	7.8	17.0
15	90	EPP24	Hard	4	5.2	12.2
16	90	Air	Soft	4	8.5	60.5
17	90	EPP70	Soft	4	7.5	17.0
18	90	EPP24	Soft	4	5.2	11.9
19	90	Air	Hard	7	16.7	97.3
20	90	EPP70	Hard	7	14.9	27.1
21	45	Air	Hard	4	22.0	28.2
22	45	EPP70	Hard	4	23.3	12.9
23	45	EPP24	Hard	4	21.0	9.4
24	45	Air	Soft	4	22.3	27.5
25	45	EPP70	Soft	4	23.8	12.8
26	45	EPP24	Soft	4	21.7	9.4
27	45	Air	Hard	7	50.9	78.7
28	45	EPP70	Hard	7	41.3	19.7

15 depicts the strain results. In this simulation, the liner has moved approximately 10 mm with respect to the shell.

b. The Experimental Study

For oblique blunt impacts the MIPS liner reduced the rotational acceleration and rotational velocity compared to the EPP70 fixed liner. For purely vertical impacts, the MIPS liner also reduced the translational and rotational acceleration but not to the same degree. The translation & rotational acceleration and the rotational velocity as a function of time for configurations B & D can be found in Fig. 16 & 17.

IV. DISCUSSION

Using numerical and experimental methods, perpendicular and oblique ballistic and blunt impacts were investigated.

i. *Comparing Air and EPP Liners:* The strain distribution in helmets with an Air Liner and and EPP Liner are similar, though the peak strain occurs earlier when an EPP liner is present. Helmets equipped with a liner distribute the external load over a larger area of the scalp.

ix. *Evaluation of the MIPS concept:* The simulation using the MIPS liner concept that allowed the helmet shell to move relative to the EPP70 liner within, this system reduced the strain from 37% to 25% for a blunt impact at 45 degrees with a speed of 7 m/s. Fig.

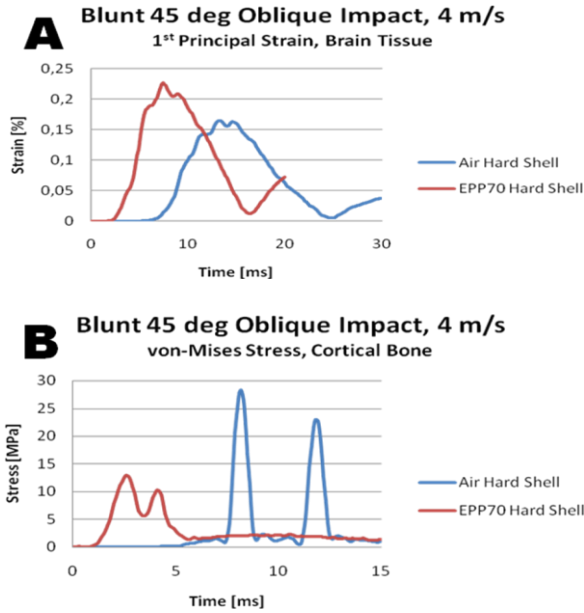


Figure 10: First principal strain in the element with the highest peak strain (a) von-Mises stress in the shell element with the highest peak value (Simulation 21 and 22)

- ii. *Finite Element model limitations:* The FE model of the helmet used a sphere of EPP, not separate cushions such as those that are found in actual helmet liners (Fig. 6). This may have increased the peak accelerations, such as those seen in Fig. 12, determined using the numerical simulation compared to an actual test of the complete system. A blunt 45 degree oblique impact resulted in the largest strain for all helmet configurations due to the effect of rotational forces on the brain. The FE model of the PASGT helmet with Air Liner lacks elements connecting the shell to the head that would transmit rotational forces during a blunt 45 degree oblique impact. Subsequently this simulation under predicted strain values compared to a head in a helmet equipped with a liner. Fig. 12 shows two distinct spikes in the stress experienced by the headform in the helmet. For the helmet with the Air Liner, this resulted because the stiff helmet bounced between the impact plate and the headform. For the helmets with EPP liners, a similar double spike was seen. In this case, the bounce was due to the nonideal fit of the helmet within the liner and shell.
- iii. *Limitations of experimental blunt impact testing:* The blunt impact tests performed herein involved a headform without a neck. However similar test set-up

with the HIII neck attached to the head has been performed at the Biokinetics lab in Canada showing similar reduction (Table 4).

- iv. *Shell hardness:* The harder helmet shell showed lower stress and strain compared to the softer shell.
- v. *Reducing head accelerations during an oblique impact:* Results presented here show that the MIPS system reduces the measured head accelerations for most impact situations, including pure vertical drops and oblique impacts. These results are comparable to other blunt impact tests [6].

V. CONCLUSION

The limited sample size of this study suggests that there is potential for design of a helmet that protects from both blunt and ballistic impact. Blunt protection could be further improved by implementing an energy absorbing sliding layer such as the MIPS system between the shell and the liner to mitigate the effect of oblique impacts.

TABLE IV. BLUNT IMPACT TEST RESULTS FROM BIOKINETICS. HIII HEAD AND NECK.

Blunt Impact Tests using MIPS system					
Impact angle	Liner type	Impact velocity m/s (ft/s)	Peak linear acceleration (g)	Peak angular acceleration (rad/s ²)	Peak angular velocity (rad/s)
90	Foam	3 (10)	79	3392	6
90	MIPS	3 (10)	69	2704	6
45	Foam	3 (10)	27	4027	24
45	MIPS	3 (10)	22	1823	14
90	Foam	4 (14)	121	4683	11
90	MIPS	4 (14)	115	3882	14
45	Foam	4 (14)	43	6607	32
45	MIPS	4 (14)	36	2745	22

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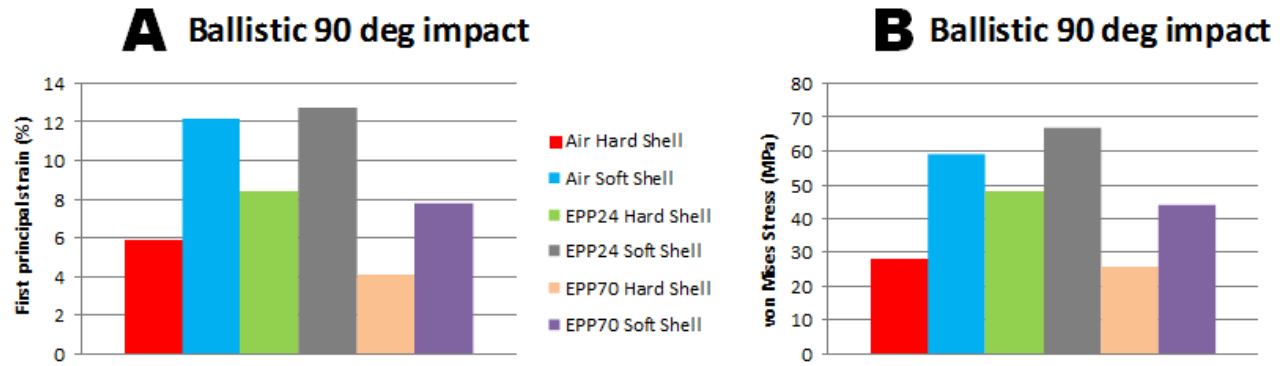


Figure 13: Ballistic impact at 90 degrees (Simulations 1-6) showing the maximum 1st principal strain in the brain (a) and showing the maximum von-Mises stress in the skull (b)

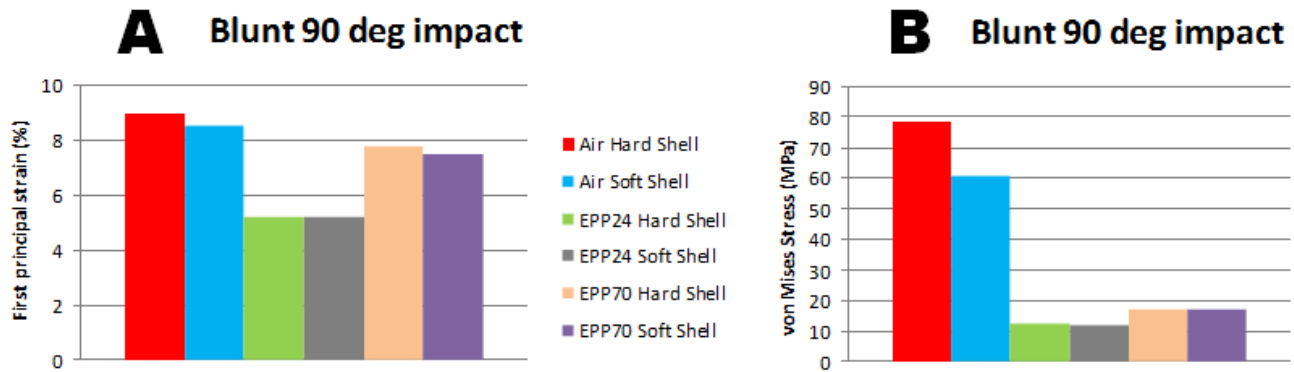
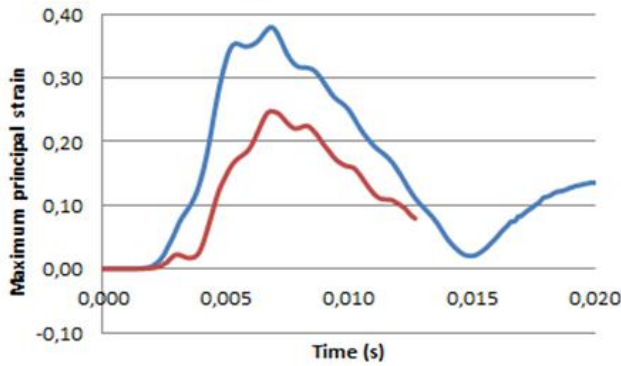


Figure 14: Blunt impact at 90 degrees and 4 m/s (Simulations 13-18) showing the maximum 1st principal strain in the brain (a) and showing the maximum von-Mises stress in the skull (b)

A Blunt 45deg, EPP70, Hard Shell, 7m/s



B

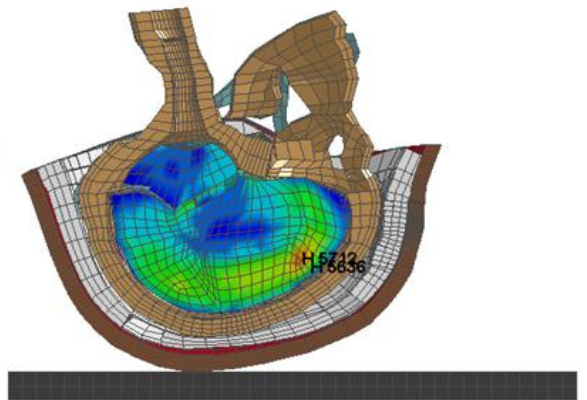


Figure 15: The MIPS helmet liner experiencing a blunt impact at 45 degrees in a hard shell at 7 m/s 1st principal strain in the element with the highest peak value (a) strain distribution in the brain (b). Showing a cut through the sagittal plane in the FE model of the head and MIPS helmet.

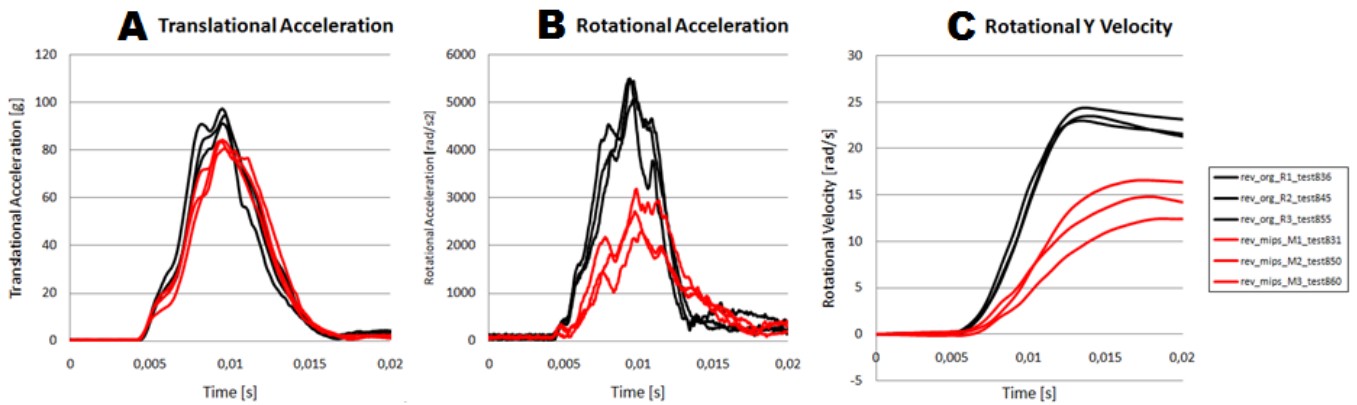


Figure 16: Test configuration B with oblique impact at 4.4 m/s (14.4 ft/s) showing the translational & rotational acceleration as well as the rotation velocity as a function of time **Black** = EPP50 liner **RED** = MIPS

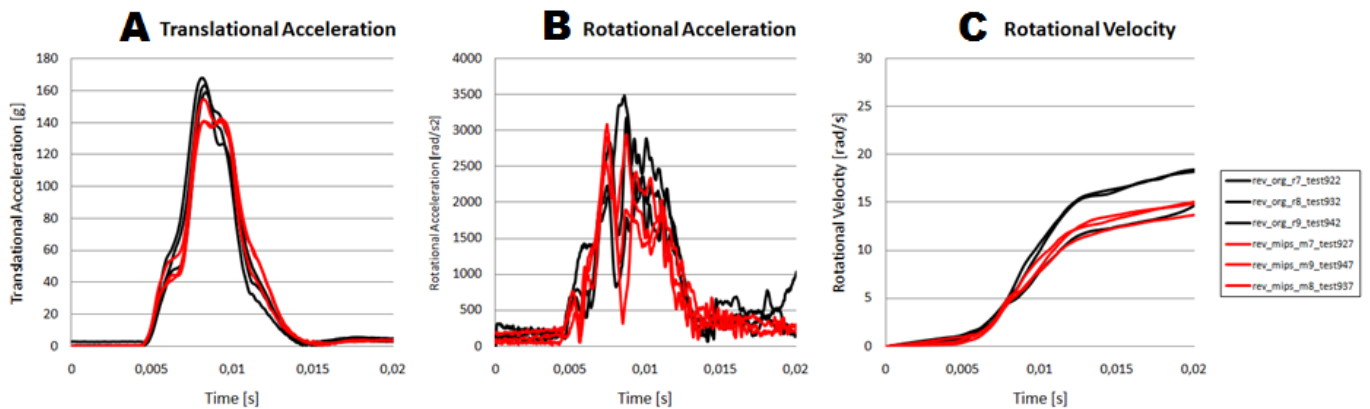
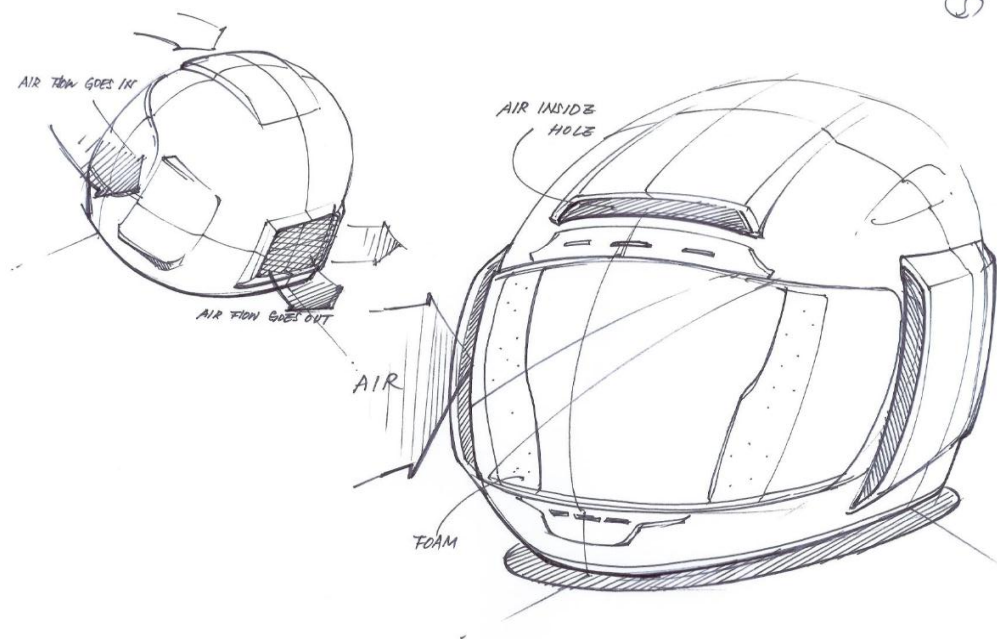


Figure 17: Test configuration D with vertical impact at 4.4 m/s (14.4 ft/s) showing the translational & rotational acceleration as well as the rotation velocity as a function of time **Black** = EPP50 liner **RED** = MIPS

Helmet Performance and Design



Finite Element Modelling of a Honeycomb Reinforced Helmet

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ABSTRACT

The improvement of the protection offered by motorcycle helmets through use of non-conventional energy absorbing materials could significantly reduce the number of motorcyclists' fatalities. The use of Finite Element Analyses is of considerable importance for the design of innovative helmets and prediction of their shock absorption properties under a multitude of loading conditions. This paper investigates the modeling of a honeycomb reinforced helmet in Ls-Dyna environment. The ECE 22.05 standard impact tests are simulated in the front, top and rear regions of the helmet, and the numerical outcomes are compared to experimental results.

Overall, the model realistically reproduces the impact response of the helmets. Particularly good agreement with experimental results is observed from impacts in the front and rear regions, against the kerbstone anvil. However, finite element results obtained from impacts in the crown region highlight the limitation of the strategy adopted in the present research, suggesting that further work is needed to improve the modeling of the helmet

Keywords: *Honeycomb Helmet, Finite Element Analyses, Impact, Ls-Dyna*

NOMENCLATURE

A	Material parameter
B	Material parameter
C	Material parameter
D	Material parameter
d	Honeycomb cell size
E	Young's modulus
m	Material parameter
p	Material parameter
P_0	Internal pressure
R	Foam relative density
s_{ij}	$i = x, y, z$, deviatoric stress
ε	Engineering strain
$\dot{\varepsilon}$	Uniaxial strain rate
ε_y	Yield strain
ε_D	Densification strain
ν	Poisson's ratio
ρ	Density
σ	Engineering stress
σ	Yield stress

I. INTRODUCTION

The improvement of the protection offered by motorcycle helmets through use of non-conventional energy absorbing materials could significantly reduce the number of motorcyclists' fatalities. Currently, most

of safety helmets feature a foam liner as the main energy absorbing component. Some foam materials have a nearly linear compressive stress-strain curve, which is an ideal behavior for energy absorption systems. Nonetheless, researchers are still investigating use of other materials in order to improve the impact absorption capacity of helmets. Pinnoji et al. [1] investigated the impact response of a helmet featuring an outer shell made of aluminium foam, and compared to the one offered by a commercial helmet, whose outer shell was made of a thermoplastic material. Results obtained in [1] showed an improvement of the protective function provided by the innovative helmet, in comparison to its commercial counterpart. In another study [2], a prototype helmet, where an Acrylonitrile-butadiene-styrene (ABS) lamina shaped as an array of deformable cones was used as energy absorbing liner, was investigated via use of Finite Element Analyses (FEA). The ECE 22.05 impact tests were simulated and the numerical response was compared to the one achieved on simulations of a commercial helmet produced at Dainese s.p.a, of same geometry and weight. It was found that the prototype helmet could provide better protection from impacts in the front and rear areas, with respect to its commercial counterpart.

In a previous phase of the research presented in this paper [3], the impact behaviour of a modified version of a commercial helmet, where aluminium honeycombs were introduced in the front, top and rear region of the energy absorbing liner, was assessed following UNECE 22_05 standards [4]. It was concluded that the use of aluminium honeycombs as reinforcement material for the energy absorbing liner can lead to an improvement of the safety levels provided by current commercial helmets, although some results highlighted the need of an optimisation of the prototype helmets to be performed.

The use of Finite Element Analyses is of considerable importance for the design and optimization of helmet shock absorption properties, under a multitude of loading conditions. Among the first FE models of helmets was the model of an open-face helmet developed by Yettram et al. [5], which was used to study the influence of the material properties of the foam and shell on the head acceleration. Later, a more advanced model was developed to study the fit effect of helmets [6] and the protective capability of the chin bar of full-face helmets [7].

Composite materials have a significant number of failure modes, which probably is the reason that composite shells are able to absorb a greater portion of the impact energy compared to conventional thermoplastic shells. Hence, a suitable material model should be used for them in FE models. In a paper by Kostopoulos et al. [8], the impact response of a composite shell helmet dropped onto a hemispherical anvil at the crown site was studied by using the FE method. One important feature of their model was simulating delamination. This model, however, was not validated against experiments. Another attempt to model a composite shell helmet was made by Aiello et al. [9]. They used the FE method to model a commercially available helmet and validated it against experimental drop test results. Nonetheless, the failure modes assumed for the shell were not described properly. Cernicchi et al. [10] investigated the mesh sensitivity issue but they used an elastic-plastic material model for the shell. They developed an FE model of a commercially available composite shell helmet. The model gave acceptable predictions of experimental drop test results. Ghajari et al. [11, 12] modeled a composite-shell helmet and simulated both normal and oblique impacts. Good agreement was found between numerical and experimental results with respect to the linear and rotational accelerations of the head.

In this paper, the development and validation of a Finite element model representative of a honeycomb reinforced helmet prototype, produced and tested during the initial phase of this research, is discussed. Major focuses of this study are the material characterisation of the helmet parts, and the comparison of the acceleration histories obtained from numerical simulations, with the experimental counterparts [3]. The explicit solver Ls-Dyna 971 was used to model the helmet prototype and simulate the experimental tests.

II. THE FINITE ELEMENT MODEL

The helmet model comprised the outer shell, the inner liner, the top layer, the lateral cheek pads, the honeycomb layers, the chin strap and the rigid headform. Previous studies [10, 13] showed that the visor and the comfort pad do not provide significant contribution to the shock absorption capabilities of the helmets, excluding impacts on the chin area. Therefore, such parts were not included in the model.

A. The headform

For the generation of the headform model (Figure 1), a faithful digital representation of the ISO 62 rigid headform prescribed by ECE 22.05 standards [4], was reconstructed using a three dimensional digitizer laser scanner over the real headform adopted for the tests [1]. The digital version of the headform was then converted into an .IGES file and imported in Hypermesh 9.0 software [14] for mesh generation. As result of the process, 10,961 solid tetrahedral elements [15]. The typical length of these elements was 16mm.

The headform was modeled as infinitely rigid material, in line with previous researchers on FE modeling of helmets [10, 13, 16]. The Ls-Dyna material algorithm *20_rigid* [17] was employed. In order to avoid contact instabilities, realistic material properties must be assigned to this particular algorithm. For the purposes of this research, the material properties of the Magnesium K1A alloy ($\rho = 1740 \text{ kg/m}^3$, $E = 38 \text{ GPa}$, $\nu = 0.34$) were defined, being this material commonly used for the production of ISO standard headform [18].

The simulated weight of the headform was 6.1 kg, in accordance to standard recommendations.

B. The inner liner, cheek pads and top layers

The inner liner was made of expanded polystyrene (EPS) foam with a density of 50 kg/m^3 density and its thickness ranged from 35 to 40 mm throughout the surface, except for the crown region, where the thickness was lower to accommodate a lighter layer of EPS foam (35 kg/m^3). [19] stated that the use of lighter foams in the top area compensates the excessive rigidity of the shell in the crown, attributed to the local pronounced double curvature and lack of free edges in proximity [20], resulting in a better protection of the head. The cheek pads were made of EPS foam with 70 kg/m^3 density. The thickness of the lateral cheek pad ranged from 20 to 35 mm, while the thickness of the chin pad varied from 15 to 20mm. Due to the fact that preliminary analyses showed that the contribution of the chin pad to the helmet impact performances were negligible, such component was not included in the final model, to reduce computational costs. Solid tetrahedral elements [15] were used for the generation of the polymeric components of the helmet. Mesh size (average length equal to 16 mm), was assigned on the base of the results obtained from a mesh convergence study carried out during the present research [21]. In addition, previous similar analyses showed that the use

of such mesh size realistically predicts the impact response of helmet liners [10]. As result of the meshing process, 20,550 elements were used to generate the helmet main liner, 6,444 elements were used to

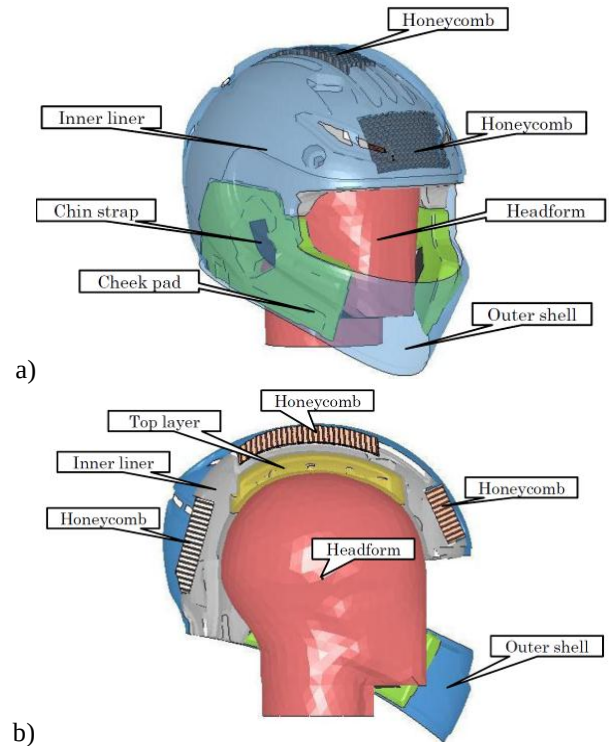


Figure 1: Prototype finite element model. a) Perspective view; b) Section view

generate the top layer and 5,950 elements were adopted to generate the cheek pads. Polystyrene foams belong to the category of closed cell foams [22] and when subjected to compressive loadings, they deform exhibiting three characteristic deformation regimes, which occur with the following sequence: linear elastic (I), plateau (II) and densification (III). Linear elastic regime is short compared to the other two deformation regimes, and ends at relatively low strain values (typically 3-5 %). It consists of a nearly linear increase of stress values with strain. For increasing values of the strain, the foam cells start collapsing plastically at an approximately constant stress (plateau regime). The densification regime occurs at large compressive strains, when the foam cell walls completely crush and the constituent material is compressed. Such phenomenon in the stress versus strain curve is represented by a steep increase of the stress values with strain.

For the purposes of the present research, the foam components were modeled as isotropic elastic-plastic

materials. For the purposes of the present research, the material algorithm *63_crushable foam* [17], was adopted. Under compressive loadings, the complete range of deformation stages of the foams (i.e. linear, plateau and densification regimes) is simulated through the introduction of user defined Young's modulus and stress versus volumetric strain curves. Lateral deformation is also considered through use of a user defined Poisson's ratio. However, previous studies showed that EPS foams subjected to mono-axial compressive loadings do not exhibit significant lateral deformations [23, 24] and that their average Poisson's ratio is of the order of 0.01. Thus, such value was introduced in the material card. For tensile loadings, the material is modeled as linear elastic and failure is considered when a user defined cut-off tensile stress is reached. In the present investigation, due to the fact that no significant tensile stress fields were expected to develop during the analyses, the tensile cut-off strength was assumed equal to the yield stress of the modeled foam (Table I). This assumption did not cause instabilities during the analyses. The foam material properties were obtained via the use of the semi-empirical approach proposed by Gibson and Ashby [22] for the modeling of the compressive behavior of isotropic closed cell foams, such as EPS foams.

According to [22], the three compressive deformation regimes can be adequately represented by the following equations:

$$\begin{aligned}
 I) \quad \sigma &= E\varepsilon + \frac{P_0\varepsilon}{1-\varepsilon-R} ; \quad 0 < \varepsilon < \varepsilon_y \\
 II) \quad \sigma &= \sigma_y + \frac{P_0\varepsilon}{1-\varepsilon-R} ; \quad \varepsilon_y < \varepsilon < \varepsilon_D \left(1 - \frac{1}{D}\right) \\
 III) \quad \sigma &= \sigma_y \frac{1}{D} \left(\frac{\varepsilon_D}{\varepsilon_D - \varepsilon}\right)^m + \frac{P_0\varepsilon}{1-\varepsilon-R} ; \quad \varepsilon > \varepsilon_D \left(1 - \frac{1}{D}\right)
 \end{aligned} \tag{1}$$

where σ and ε are the engineering stress and strain, E is the Young's modulus, σ_y is the compressive yield stress, ε_y is the strain value corresponding to the yield stress, ε_D is the full densification strain, P_0 is the internal initial pressure (equal to the atmospheric pressure 0.1 MPa), and R is the foam relative density defined as the ratio between the density of the foam and the density of the solid polymer with which the foam is made. D and m are constants equal to 2.3 and 1 [10, 22]. In addition to this model, the authors provided definitions of the Young's modulus, yield stress and full densification strain in function of the relative density R as follows:

$$E = AR^2 + BR \tag{2}$$

$$\sigma_y = CR^{1.5} \tag{3}$$

$$\varepsilon_D = 1 - 1.4R \tag{4}$$

where A , B and C are material constants.

TABLE I. EPS FOAM MATERIAL PROPERTIES

Helmet component	Material Properties			
	ρ [kg/m ³]	E [MPa]	σ_y [MPa]	ν
Top layer	35	11.8	0.29	0.01
Main energy absorbing liner	50	27.1	0.54	0.01
Cheek pads	70	46.4	0.9	0.01

In the present study, the foam material constants were attained through curve fitting of experimental data collected during a material characterisation study carried out at the initial phase of the present research [21, 25]. Assuming the density of the bulk polystyrene equal to 1050 kg/m³ [26], it was obtained approximately $A = 7 \times 10^9$ MPa, $B = 2.7 \times 10^8$ MPa, $C = 4.73 \times 10^8$ MPa. With regard to the coefficient D (Eq. 1), it was found that a reduction of 35% with respect to the value suggested in literature ($D = 2.3$) could adequately model the onset of the densification regime in the outcomes obtained from experimental tests on EPS foams.

C. The honeycombs

The honeycombs used for the assembly of the helmet prototypes were the hexagonal 5.2 Al 3003 cores, produced by expansion of glued aluminium sheets at Cellbond Composites Ltd. The honeycomb cell size, defined as the distance between two parallel faces in a single hexagonal cell, was $d = 6.35$ mm. The honeycomb layers were modelled as three-dimensional hexagonal cell structures using two-dimensional shell elements. Four-noded square shell elements [15] were adopted for the generation of the mesh. Particular care was given to the modeling process, to ensure that the geometry and dimensions of the models reproduced faithfully the honeycombs used during the experiments [25]. The materials were oriented in a way that all the cell walls with doubled thickness were parallel to the symmetry plane of the helmet (see Figure 2).

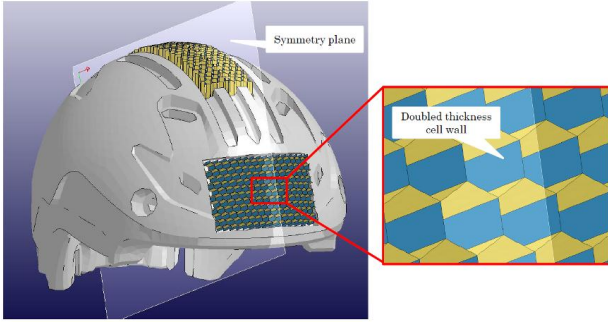


Figure 2: Orientation of the honeycombs with respect to the symmetry plane of the prototype liner

The cell wall thickness was assigned using Ls-Dyna algorithm *section_shell* [17] and was set equal to the thickness of the honeycombs used for the experiments ($t = 30 \mu\text{m}$). Five through-thickness integration points were assigned to each element of the honeycomb model, to capture the complex stress field which is generated in the honeycomb cell walls during the plastic collapse of the materials. Each element side was equal to 0.3mm, in accordance to the outcomes of a convergence study carried out during the present research [21]. As result of the meshing process, the front honeycomb layer consisted of 155,271 elements, while the top and the rear layers consisted of 117,418 elements and 254,833 elements respectively. The material properties of the Al 3003 H18 alloy ($\rho = 2730 \text{ kg/m}^3$, $E = 68.9 \text{ GPa}$, $\sigma_y = 186 \text{ MPa}$, $\nu = 0.33$) were implemented using the Ls-Dyna algorithm *24_piecewise_linear_isotropic_plasticity* [15], being this alloy the one used for the manufacturing of the honeycombs.

This material model allows the definition of arbitrary stress versus strain curve and strain rate dependency. Different stress versus volumetric strain curves for various strain rates can be introduced. Strain rate dependency is taken in to account through interpolation between curves. When stress versus strain curves are not available, it is possible to introduce in the material model arbitrary values of the yield stress σ_y , Young's modulus E and Poisson's ratio ν . The yield surface is defined through the Von Mises flow rule [27]:

$$\phi = \frac{1}{2} s_{ij} s_{ij} - \frac{\sigma_y^2}{3} \leq 0 \quad (5)$$

where s_{ij} is the deviatoric stress and σ_y is the yield stress. At each time step, the update of the deviatoric stresses is assumed as linear and the yield function is

checked. If the Von Mises rule is satisfied, then the deviatoric stresses are accepted. If the yield function is not satisfied, then the overcoming stresses are scaled back to the yielding surface. When experimental data are not available, strain rate dependency is treated through the use of a mathematical model proposed by Cowper and Symonds [28]. According to the authors, the strain rate sensitivity of metallic alloys can be adequately represented by the following equation:

$$\sigma = \sigma_0 \left[1 + \left(\frac{\dot{\epsilon}}{C} \right)^{\frac{1}{p}} \right] \quad (6)$$

where σ is the dynamic stress at uniaxial strain rate $\dot{\epsilon}$, σ_0 is the material yield stress measured at strain rate 1s^{-1} , C and p are parameters that can be obtained from experimental tensile tests. Previous studies on the strain rate sensitivity of steel alloys [29], have shown that such model can accurately reproduce rate sensitivity at both low (10^{-4} s^{-1}) and high strain rates (1000 s^{-1}).

It is known that aluminium strain rate effects are also dependent on alloy [30]. For aluminium 3003 alloys, it was found that $C = 2.5 \times 10^5 \text{ s}^{-1}$ and $p = 8$ [31]. In the present investigation, the formulation proposed by Cowper and Symonds was used, and the coefficients proposed by [31] were adopted.

D. The outer shell

The outer shell was a two layered composite laminate. The upper layer was a woven hybrid material, made with threads of Kevlar, carbon and glass fibres. The bottom layer was a woven fabric made of Kevlar fibres. A third additional layer, a short fibre glass composite with random oriented threads 30 mm long, was added on the external front surface of the helmet, right above the upper edge of the visor. All the composite materials were impregnated with an epoxy resin solution. The whole component was modeled using 10,524 four-noded shell elements. The length of the shell elements side (3mm) were chosen on the base of the outcomes of a preliminary FE study conducted during the present research [13]. Ghajari [13], simulated UNECE 22.05 standard impact tests on a FE model of an AGV full face helmet produced at Dainese s.p.a. The resultant acceleration of the centre of gravity of the headform was considered as evaluation criteria, and the mesh sensitivity of the shell was investigated by using four-noded elements presenting three different average lengths: 3mm, 6mm

and 10mm. It was observed that use of 6mm and 3mm elements resulted in very similar acceleration histories, both in terms of magnitude and duration, while a significantly different dynamical response was observed from use of 10mm elements. These results were found to be in line with previous finite element convergence studies [10, 26]. Cernicchi et al. [10, 26], simulated impact on the front surface of a commercial helmet against the kerbstone anvil prescribed by UNECE 22.05 regulations. Six mesh densities were adopted, where the element average dimension ranged from 2mm to 15mm. The force experienced by the anvil was plotted over time and numerical outcomes suggested that convergence between results was obtained only for meshes where the average side ranges from 2 mm to 5 mm

In the present investigation, the shell constituent materials were modeled as orthotropic materials, through use of the Ls-Dyna material algorithm *58_laminated_composite* [17]. The stress-strain mechanical response of the singular ply is modeled through user defined in-plane compressive and tensile Young's moduli, shear moduli and Poisson's ratios. Damage is simulated through degradation of the in-plane stiffness matrix components [15]. Four failure modes are considered for each lamina:

- Tensile fibre failure (fibre rupture);
- Compressive fibre failure (fibre buckling);
- Matrix cracking under transverse tensile and shear loading
- Matrix cracking under compressive and shear loading.

The maximum strengths in tension, compression and shear must be also defined with their correlated strain values. The stacking ply sequence of the composite shell was simulated using the Ls-Dyna algorithm *part_composite* [17]. Each layer was identified by an integration point, to which Ls-Dyna users can assign thickness, material properties and orientation.

For the purposes of this research, all the material properties were kindly provided by the helmet manufacturer, Dainese, s.p.a. Such parameters were obtained from quasi-static compressive, tensile and shear tests prescribed by ASTM standard regulations [32-34], performed at MAVET s.r.l., on flat coupons made with the composite materials used for the production of the Gp-Tech. For confidentiality reasons, such properties cannot be reported in this paper.

E. The chin strap

The chin strap, a 300 mm long, 25 mm wide and 1.5 mm thick polyethylene terephthalate (PET) woven band, was modelled using four-noded shell elements. The initial shape created, passed through the holes of the cheek mouldings and closely fitted the headform chin. A hundred 4-noded shell elements were used for the generation of the chin strap shape. The material card MAT24 [17] was used to model the chin strap mechanical properties ($\rho = 800 \text{ kg/m}^3$, $E = 1.83 \text{ GPa}$, $\nu = 0.2$, $\sigma_y = 47 \text{ MPa}$). A preliminary FE simulation was carried out to pull the ends of the chin strap through the cheek pad holes until the shape conformed to the chin of the headform. To achieve this aim, a force equal to 10 N was applied at both the ends of the chin strap model and directed towards the top of the model (Figure 3). Then, the deformed mesh of the chin strap was introduced in the prototype model with no pre-stress conditions, in accordance to previous studies on the modelling of the chin strap in crash helmet simulations [13, 35]. The link between the retention system and the shell of the Gp-Tech was simulated by constraining the nodes at the ends of the chin strap model to the surface of the outer shell.

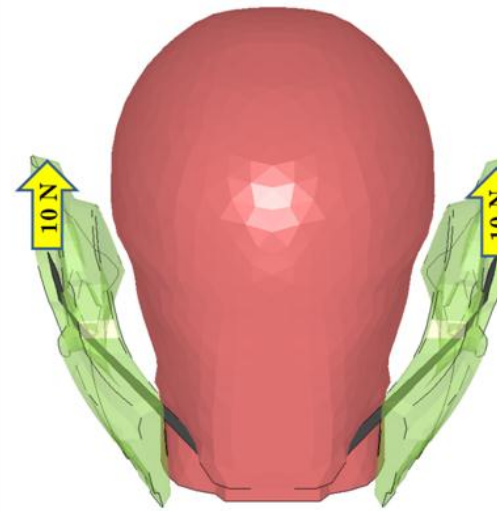


Figure 3: Virtual tightening of the chin strap (the complete model is not shown).
Front view

F. The anvils

The flat and kerbstone anvil prescribed by standards, and used in the present investigation, were created through use of pre-built Ls-Dyna rigid wall algorithms. A cylindrical surface (with 130mm diameter and 50mm thickness) was used to simulate the

flat anvil, while a combination of a cylindrical surface (with 30mm diameter and 125mm length) and two flat surfaces (125mm length x 80mm width) was used to generate the kerbstone shape (Figure 4)

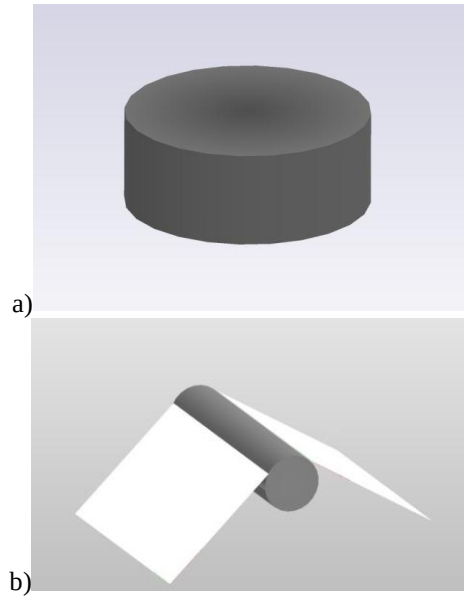


Figure 4: Simulated anvil shapes.
a) flat anvil; b) kerbstone anvil

G. Contact

Three typologies of penalty stiffness contact algorithm [15] were used to model contact between the helmet parts:

- *Automatic_surface_to_surface*, defined at the interfaces shell/liner, shell/honeycombs, honeycombs/liner, liner/headform and all the interfaces between the chin strap and the headform, cheek pads and shell;
- *Automatic_nodes_to_surface*, defined at the top and bottom nodes of each honeycomb layer model, to avoid penetration of the honeycomb edges in the surface of the shell and the liner during the simulations.
- *Automatic_single_surface*, defined for each honeycomb layer, to prevent self-penetration of the honeycomb cell walls during the buckling of the honeycombs.

Static and dynamic friction at the interfaces between the helmet parts was modelled through Ls-Dyna built-in functions, which are based on the Coulomb friction model. For the contact between the polystyrene components, the static coefficient of friction was set equal to 0.5 [10, 26]. For contact between foams and

metallic parts (i.e. at the interfaces liner/honeycomb and liner/headform), due to lack of available data in literature, the coefficient of friction between polystyrene and steel (1.05) was taken as reference. For contact between the shell and the other parts of the helmet, and between the chin strap and the cheek pads and headform, a unique coefficient of friction equal to 0.3 was adopted, in line with existing FE studies [13]. In all the interfaces, the dynamic coefficient of friction was set equal to one third of the value of their static counterparts.

H. Simulations

Impacts were simulated in the front (B), top (P) and rear (R) region of the helmet against both the kerbstone and flat anvil.

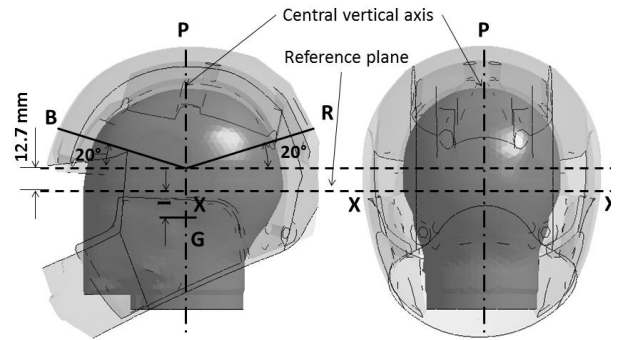


Figure 5: Location of impact points on the headform

In each impact configuration, the kerbstone anvil was inclined by a 45 degree angle with respect to the plan of symmetry of the helmet, in accordance to standard prescriptions. Prior to simulation, the virtual helmet was positioned in a way such as the impact point was aligned to the centre of the surface of the anvil. Impact speed was simulated by assigning initial velocity equal to 7.5 m/s to all the nodes of the model, by using the LS-Dyna algorithm *initial_velocity* [17]. For the headform, mass properties and initial velocity were defined through the algorithm *part_inertia*.

Acceleration histories of the centre of gravity of the headform (point G in Figure 5) were recorded and processed using the software LS-PrePost [36]. A digital filter was applied to the numerical acceleration signals to remove undesired oscillations. The filtering frequency was equal to the one adopted during the experiments (1.7 kHz) for the removal of undesired numerical oscillations caused by contact instabilities. In each simulation, the solving time step was calculated on the size of the honeycomb elements and it was of

the order of 10^{-9} s. Simulations were performed on a high performance cluster using 8 CPUs and 6GB RAM. Due to the high number of elements employed for the modelling of the helmet, the wall-time for a 12 ms impact simulation was 72 hours.

III. RESULTS AND DISCUSSION

In this section, the Finite Element model of the helmet prototype is validated against the experimental results obtained from tests carried out on honeycomb reinforced helmets, performed during the initial phase of this research [3]. In each graph, the numerical resultant acceleration histories of the centre of gravity of the headform are compared to their experimental counterparts. The peak linear acceleration, defined as the maximum acceleration in magnitude during the impacts, is also considered as validation criteria and compared to the average values recorded during the experiments (Table II).

TABLE II. AVERAGE PEAK LINEAR ACCELERATIONS. COMPARISON BETWEEN NUMERICAL AND EXPERIMENTAL RESULTS

Impact point	Flat anvil		Kerbstone anvil	
	FE prediction [g]	Experimental [g]	FE prediction [g]	Experimental [g]
B	199 (+ 9.4 %)	182	150 (+ 6.0 %)	141
P	203 (-1.5%)	206	156 (- 16.0 %)	184
R	194 (- 4.0 %)	202	140 (+ 2.9 %)	136

A. Rear Impacts

The acceleration histories of the centre of gravity of the headform for impacts in the rear area are depicted in Figure 6 a and b. In general, the model provided very good agreement with the trends observed experimentally, both in terms of shape of the acceleration histories and magnitudes. Comparing the peak linear accelerations, it can be observed that the simulated values were also very close to the average recorded from experiments.

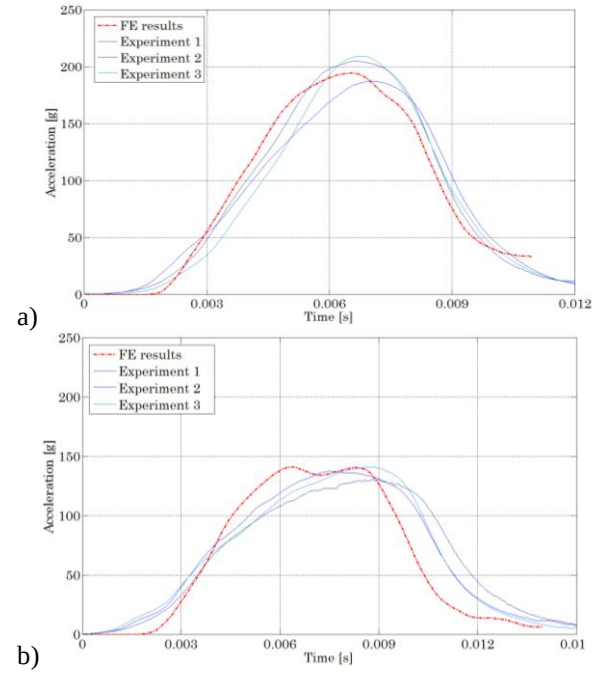


Figure 6: Acceleration histories from impacts at $v = 7.5$ m/s in the rear area. a) impacts against the

B. Front impacts

Figure 7 a and b show the acceleration histories recorded from impacts in the front area, against the flat and kerbstone anvil. The model provided good agreement with experimental results. Comparing the peak linear accelerations (Table II.), it can be also observed that the numerical predictions are relatively higher (9.4% for the impacts against the flat anvil and 6% for impacts against the kerbstone anvil) than the average obtained from experimental results. These discrepancies were attributed to the difference between the simulated material properties of the helmet parts and the actual material properties of the components used for the manufacturing of the prototypes. For example, it is known that environmental factors such as temperature and humidity might have a significant degrading effect on the mechanical properties of polystyrene foams [22, 37]. With particular reference to the effect of humidity on the compressive properties of EPS foams, in an existing study available in literature [37], it was concluded that the plateau stress of EPS foams compressed in normal conditions (25 °C and relative humidity 30%) decreased by approximately 20% when the same materials were tested at relative humidity equal to 85%. It is therefore possible that because of not ideal storing conditions, the foams tested in the present research [25] and used for the characterisation of the FE liner might have weakened

due to exposure to humidity, compared to the foams used for the manufacturing of the prototypes, contributing to the difference between numerical and experimental outcomes. It must be also stressed that the material characterisation of the shell was attained through quasi-static tests on flat material coupons, a methodology commonly adopted in literature for the FE modelling of the outer shell [8-10], but not actually representative of real helmet loading conditions. In addition to this, because of local curvature and imperfection in the manufacturing processes, the mechanical response of composite shells significantly varies from the one offered by same materials in a flat form [19].

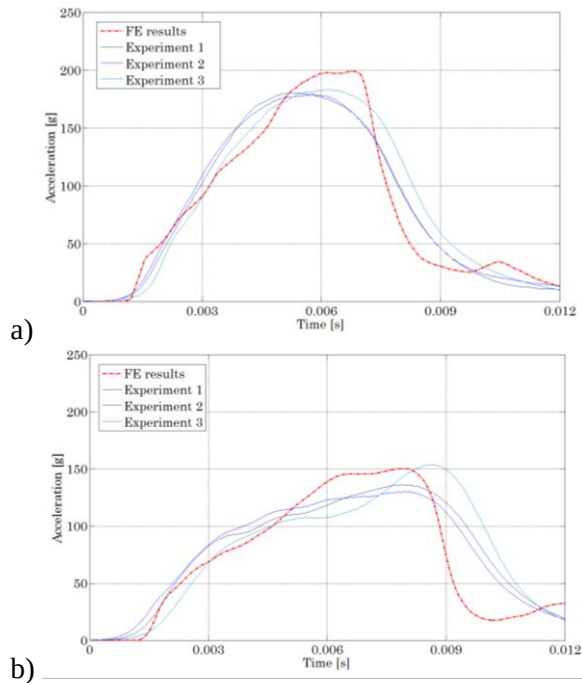


Figure 7 - Acceleration histories from impacts at $v = 7.5$ m/s in the front area. a) impacts against the flat anvil; b) impacts against the kerbstone anvil

Another major discrepancy consists in the duration of the numerical accelerations, which is in general shorter than the one observed experimentally. Evident scatter between the curves can be observed in the unloading region (i.e. the region of the curve after the maximum peak acceleration), where numerical resultant acceleration traces drop following a steeper pattern compared to the experimental counterparts. In a preliminary finite element investigation conducted during the present research [13], such behaviour was attributed to the modelling of the unloading of the foams in material model *MAT_63_crushable foam*

[15]. When using such material card, the unloading of the foam in the stress versus strain curve is assumed to follow a straight line, whose slope is by default equal to the user defined foam Young's modulus [15]. However, due to the fact that the foam densification region exhibits higher slopes than the one typical of the elastic region, Ls-Dyna automatically adjusts the value of E in a way that the slope of the unloading curve is higher than the steepest slope present in the user input curve. In all the simulations performed in the present investigation, the value of E was automatically increased by two orders of magnitude compared to the value defined as input.

C. Top impacts

The results obtained from FE impacts in the crown area are shown in Figure 8 a and b, for the two evaluated impact surfaces. With regard to the impact against the flat anvil (Fig. 8a), the model could reasonably reproduce the shape of the experimental accelerations, and provided exceptional agreement in terms of peak linear accelerations (Table II). However, it can be noted that the discrepancies observed from impacts in the front region are here more pronounced. Such effect was attributed to the more pronounced doubled curvature of the shell in the crown area, which might have further altered the simulated mechanical response of the outer shell. Regarding impacts against the kerbstone anvil (Fig. 8b), while the experimental outcomes showed a characteristic double peak shape, FE accelerations were characterised by a single peak followed by oscillations around a nearly constant value, until unloading occurred.

In our preliminary FE results [13], analogous phenomena were observed from simulations of impacts against the kerbstone anvil on the rear area of a commercial helmet, whose shell was made of similar materials to the ones of the shell presented in this research. The model was compared to experimental observations, and such behaviour was linked to higher amounts of energy dissipated by the shell in finite element analyses, compared to the one dissipated by the shell during experiments. Such conclusion was confirmed from comparison of the sequence of numerical deformation of the helmet with experimental counterparts. In FEA, the rebound of the helmet occurred slightly later than in experiments because of the more pronounced deformation of the shell, which also had a stabilising effect on the maximum accelerations transmitted to the head. On the other hand, the earlier rebound of the helmet in experiments suggested that most of the energy stored by the outer

shell during the impact was released as kinetic energy during the unloading phase, resulting also in a peak of the acceleration values. In the present investigation, similar conclusions are assumed to justify the discrepancies observed in Fig. 8b.

IV. CONCLUSIONS

An FE model of an innovative helmet, where aluminium honeycomb is used as reinforcement material, was generated in Ls-Dyna environment. The

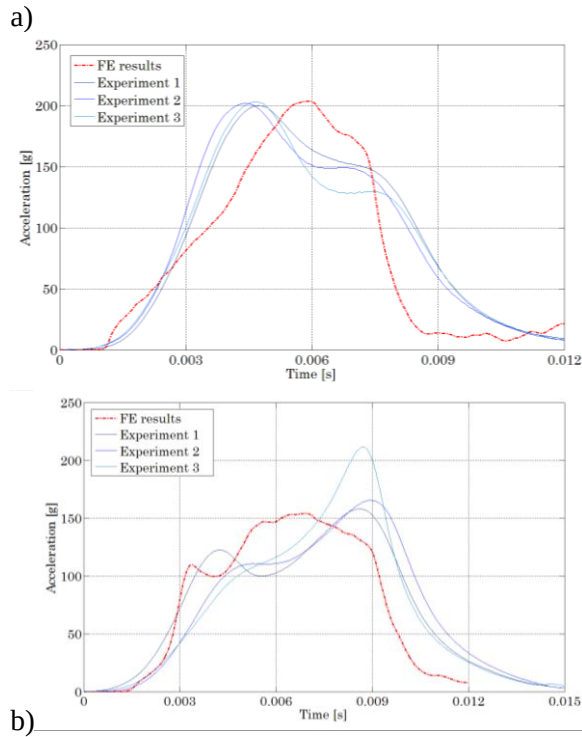


Figure 8: Acceleration histories from impacts at $v = 7.5$ m/s in the crown area. a) impacts against the flat anvil; b) impacts against the kerbstone anvil

UNECE 22.05 standard impact tests in the front (B), top (P) and rear (R) region of the helmet were simulated, and numerical outcomes were compared to experimental results attained during the present investigation [3]. The present study is similar to recent published investigations on the FE modelling of motorbike helmets [10, 38]. The mechanical behaviour of the outer shell was modelled through use of an algorithm based on a continuum damage mechanics model. The dimensions of the shell elements were chosen on the base of a mesh convergence study carried out during the present investigation [13]. Material properties of the shell components were obtained from tests on representative flat coupons and provided by the helmet manufacturer. Analogously to existing FE researches [6, 8, 10], the polymeric liner

parts of the helmet were modelled as isotropic materials, and their mechanical behaviour was modelled through use of the semi-empirical equations proposed by Gibson and Ashby [22]. These equations were calibrated with experimental results obtained in the present analysis from compressive tests on expanded polystyrene samples. The foam model provided good agreement with experimental observations.

The mechanical response of the honeycomb layers was approximated through use of a material algorithm based on piecewise linear elasto-plasticity principles. The honeycomb alloy material properties were retrieved from available data in literature, and the FE model was validated against experimental tests performed in the present investigation, on aluminium honeycomb samples [25]. Good agreement was observed between numerical and experimental outcomes.

Overall, the model could realistically reproduce the impact response of the prototype helmets tested in this investigation, for the three evaluated loading sites and the two anvils used. Particular good agreement with experimental results was observed from impacts on the front and rear region, against the kerbstone anvil.

However, FE results related to impact in the crown region highlighted the limitation of the strategy adopted in the present research, and although the prediction of the maximum accelerations falls well within the range of values recorded experimentally, further work is needed to improve the modelling of the helmet. The discrepancies were attributed to the use of composite materials properties obtained from tests on flat coupons for the modelling of the shell, which are known to alter the actual shell load spreading capabilities. The validation of the shell model over tests performed on doubled curvature composite materials could improve the accuracy provided by the helmet model. In addition to this, failure due to delamination and sensitivity of the composite materials to strain rate are not included in the material model used in this investigation. This is believed to have further contributed to increase the differences between numerical and experimental outcomes. However, the modelling of delamination would have resulted in excessive computational time costs, and previous studies on the FE modelling of motorbike helmets [8] showed that in carbon, Kevlar and glass fibre epoxy composites, delamination failure takes approximately only 10% of the total impact energy absorption share. With regard to the strain rate sensitivity of laminate composites, no specific material

algorithms are currently available in Ls-Dyna, although some energy based material models including strain rate effect are under development [39]. The model described in the present paper could set up the framework for future research, where optimisation of the honeycomb reinforced helmet is carried out under a wider set of loading conditions.

The prototype helmets tested for the validation of the model were produced following a non-industrialised process, because of time and budget constraints. Hence, adequate information regarding the manufacturing costs of the proposed helmet design could not be provided. Future studies should determine such costs, and compare these to the ones associated to the manufacturing of commercial helmet designs. In addition, the transmission of rotational accelerations to the head, known to cause severe head injuries [40], was not assessed in the present study. Therefore future work should also include follow-up work design to evaluate whether the honeycomb reinforced helmets provide adequate protection against transmission of rotational acceleration, and whether the protection offered is superior to the one currently offered by their commercial counterparts.

ACKNOWLEDGMENT

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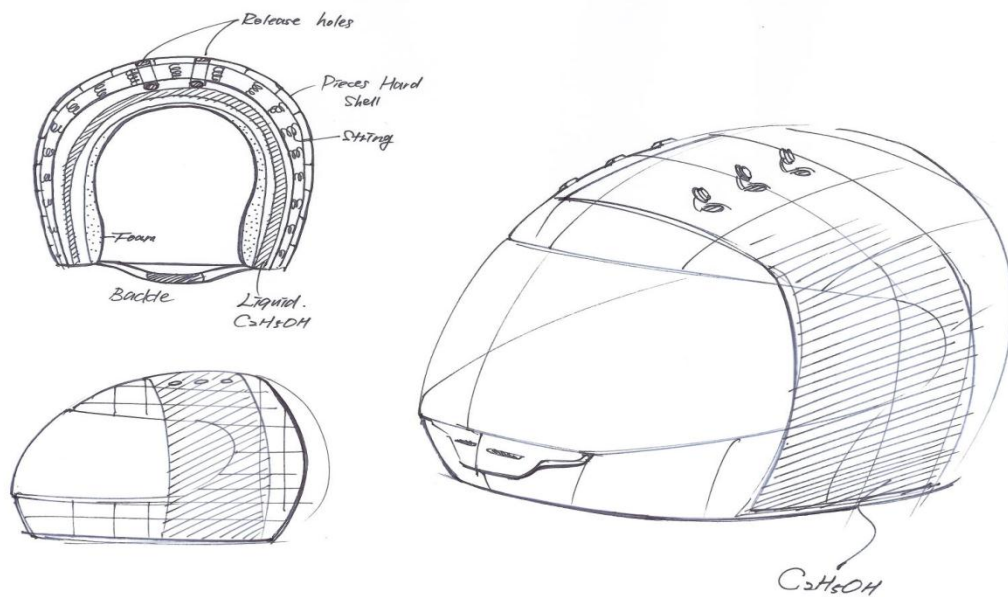
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Helmet Performance and Design



A Comparative Study of Turbulence Models Performance for the Study of Air Flow in Helmets

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ABSTRACT

Flow prediction using CFD in a flow geometry becomes a complex issue when the flow is in the transition zone. The motivation for this study is to find the best turbulence model for predicting air flow in the gap between the head and the helmet. In CFD if the flow is known to be turbulent then any standard turbulence model such as the $k-\epsilon$ model does a reasonable job of predicting the mean flow quantities. As a first step this study aims to find the optimum turbulence model for near transition flows. Using simple flows such as pipe and lid driven cavity we show that even in these simple cases if the flow is laminar and a turbulence model is used in the CFD simulations, the results with most turbulence models are erroneous. If the model were to perform well under a laminar condition it will predict a laminar profile and nearly zero eddy viscosity. We also show that for pipe flows the one equation Spalart Allmaras (S-A) model shows these trends. Its relevance to helmets is because when we carry out CFD for a helmet the flow domain is large and most of it includes the region outside the helmet where the flow is known to be turbulent. The boundary conditions are prescribed on this outer domain. The gap between the head and the helmet is thin and since we do not know the velocity at the inlet of this region we do not know apriori if the flow is laminar or turbulent or in the transition zone. Thus it becomes important to work with a turbulence model which will perform well in laminar as well as turbulent conditions. The numerical experiment on simple pipe flow shows that S-A model performs better than the standard two equation models when the flow is in the laminar or transition regime and performs almost the same as the other two equation models in the turbulent regime. The flow between the head and the helmet is not a standard geometry for which the Reynolds number at which the

flow becomes turbulent is well established and hence the need for a proper model performing well in all regimes. Having established this, we then try to match the results of the S-A model with experimental results and found that for 3D flows the S-A model does a better job than the two equation $k-\epsilon$ models.

Keywords: *Spalart-Allmaras turbulence model; CFD; flow; helmet*

NOMENCLATURE

d, y	Distance from wall
ρ	Density
μ	Dynamics Viscosity
μ_t	Frictional Velocity
C_{b1} , etc	Empirical Constants in turbulence models
f_{v1} , etc	Empirical function in the turbulence model
b	Gravity vector,
g, r, S	Intermediate variables
$\chi \equiv \nu / \nu$	Intermediate variable
ν_t	Kinematic turbulent viscosity
ν	Kinematic viscosity
S	Measure of deformation tensor
U	Mean velocity in x-direction
P	Pressure
W_{ij}	Strain rate tensor
k	Turbulent kinetic energy
$\dot{\nu}$	Time derivative of velocity
ϵ	Turbulent Dissipation rate
t	Time
σ	Turbulent Prandtl number
K	Von Karman constant

∇	Vector Gradient Operator
τ_{ij}	Viscous Shear Stress
$\mathbf{v}$	Velocity
ν	Working variable of turbulence model

I. INTRODUCTION

The aim of this work is to compare the various turbulence models for the study of near wall internal flows. Computational analysis is performed to assess the performance of the turbulence models. The motivation for this study is to find the best turbulence model for predicting air flow in a helmet. Flow prediction using CFD in a new flow geometry becomes a complex issue when the flow is in the transition zone. If the flow is known to be turbulent then any standard turbulence model such as the k- ϵ model does a reasonable job of predicting the mean flow quantities. The eventual aim of this study is to carry out the CFD simulations for predicting the flow of air in the air gap between head and helmet. Since the flow takes place at low speeds in a thin gap it cannot be established apriori whether the flow has become turbulent or not. As a first step this study aims to find the optimum turbulence model for near transition flows. In such situations this is achieved by means of CFD simulation of various benchmark problems such as pipe flow and lid driven cavity. The best model for predicting the turbulence in each problem is determined after comparing the results with the standard data available. This model is then used for predicting air flow in the gap between the head and the helmet surface. The results indicate that as the Reynolds number decreases, the one equation S-A model predicts the flow quantities better than two equation k- ϵ models for internal flows driven by a stream wise pressure gradient. However in case of shear driven flows almost all the models show a similar performance with the Realizable k- ϵ model showing the best results. Simulations are performed on a 2-dimensional hemispherical head-helmet model and results are compared with experimental data. The results of the simulations indicate that the S-A model shows an acceptable agreement with pressure data at holes on central plane for different inlet velocities. Simulations for a 3-dimensional hemispherical head helmet model show that the S-A model predicts the flow better than two equation k- ϵ models.

II. GOVERNING EQUATIONS FOR FLUID FLOW IN HELMET

The governing equations for fluid flow in the air gap of helmet is described by the Navier-Stokes equations given below. The Navier-Stokes equations form a second

order, coupled system of nonlinear PDE's involving variables as pressure as p and velocity as $\mathbf{v}$, describing the conservation of mass and momentum for a fluid flow.

$$\frac{\partial \rho}{\partial t} + \rho(\nabla \cdot \mathbf{v}) = 0 \quad (1)$$

$$\rho \frac{D\mathbf{v}}{Dt} = \rho \mathbf{g} - \nabla p + \nabla \cdot \boldsymbol{\tau} \quad (2)$$

The Total Time Derivative is defined as:

$$\frac{D\mathbf{v}}{Dt} = (\mathbf{v} \cdot \nabla) \mathbf{v} + \frac{\partial \mathbf{v}}{\partial t} \quad (3)$$

For Incompressible Newtonian viscous flows the generalized equations for three dimensional viscous stresses can be written as

$$\tau_{xx} = 2\mu \frac{\partial u}{\partial x}; \tau_{yy} = 2\mu \frac{\partial v}{\partial y}; \tau_{zz} = 2\mu \frac{\partial w}{\partial z} \quad (4)$$

$$\tau_{xy} = \tau_{yx} = \mu \left(\frac{\partial u}{\partial y} + \frac{\partial v}{\partial x} \right) \quad (5.a)$$

$$\tau_{xz} = \tau_{zx} = \mu \left(\frac{\partial w}{\partial x} + \frac{\partial u}{\partial z} \right) \quad (5.b)$$

$$\tau_{yz} = \tau_{zy} = \mu \left(\frac{\partial v}{\partial z} + \frac{\partial w}{\partial y} \right) \quad (5.c)$$

Replacing the expressions of viscous shear stresses from equations (4) and (5) in Equation (2) we get the Navier-Stokes equations for incompressible Newtonian fluid -

$$-\nabla p + \mu(\nabla^2 \mathbf{v}) + \rho \mathbf{b} = \rho \dot{\mathbf{v}} \quad (6)$$

The most popular turbulence models are the one equation S-A model, or two equation standard k- ϵ model, Realizable k- ϵ model and RNG k- ϵ model. These models are based on the Boussinesq assumption that relates the apparent turbulent shearing stresses to the rate of mean strain through an apparent scalar turbulent or "eddy" viscosity. Consequently, the relation between the Reynolds stresses and the velocity is linear (Samy *et al.*, 2009).

III. TURBULENCE MODELING

Virtually all fluid engineering applications are turbulent and therefore requires a turbulence model to predict the flow. Turbulence modeling is commonly considered to have deviations from experimental data in predicting flow through the tested cases. The selection of a proper turbulence model for simulation of a particular flow problem is therefore a key issue in CFD. A large family of turbulence models exists but no pretense has

been made that any of these models can be applied to all turbulent flows: as such a ‘universal’ model does not exist. Each turbulence model has its advantages, disadvantages, limitations and appropriate flow regimes. In the literature a large number of turbulence models have been discussed which is far too expansive to be reviewed here but since this work is focused on the comparison of turbulence models a brief description of few models especially the S-A, standard $k-\varepsilon$, Realizable $k-\varepsilon$ and RNG $k-\varepsilon$ models is inevitable. Some of the discussion below follows from Dewan (2011) and Sanghi (2001).

A. One Equation-S-A Model

The Spalart–Allmaras model is a one equation model for the turbulent viscosity. It solves the Reynolds – averaged Navier-Stokes equations and a transport equation for eddy viscosity. The Reynolds stresses are given by $\overline{u_i u_j} = 2\nu_t S_{ij}$. The eddy viscosity is given by

$$\nu_t = \nu f_{v1} \quad (7)$$

A transport equation is solved for ν which may be referred to as the Spalart–Allmaras variable. The one-equation model is given by the following equation (Spalart et al, 1994).

$$\begin{aligned} \frac{D\tilde{\nu}}{Dt} = & C_{b1}[1-f_{t2}]\tilde{S}\nu + \frac{1}{\sigma}\left[\nabla \cdot ((\nu + \tilde{\nu})\nabla \nu) + C_{b2}(\nabla \nu)^2\right] \\ & - \left[C_{w1}f_w - \frac{C_{b1}}{k^2}f_{t2}\right]\left[\frac{\tilde{\nu}}{d}\right]^2 + f_{t1}\Delta U \end{aligned} \quad (8)$$

where ν obeys the above transport equation also

$$f_{v1} = \frac{\chi^3}{\chi^3 + C_{v1}^3} \quad (9)$$

$$\chi \equiv \frac{\nu}{\nu} \quad (10)$$

$$\tilde{S} \equiv S + \frac{\nu}{k^2 d^2} f_{v2} \quad \text{Where, } S = \sqrt{2W_{ij}W_{ij}} \quad (11)$$

$$f_{v2} = 1 - \frac{\chi}{1 + \chi f_{v1}} \quad \text{and} \quad f_w = g \left[\frac{1 + C_{w3}^6}{g^6 + C_{w3}^6} \right]^{1/6} \quad (12)$$

$$g = r + C_{w2}(r^6 - r) \quad (13)$$

$$r \equiv \min \left[\frac{\nu}{Sk^2 d^2}, 10 \right] \quad (14)$$

$$f_{t2} = C_{t3} \exp(-C_{t4} \chi^2) \quad (15)$$

$$W_{ij} = \frac{1}{2} \left(\frac{\partial u_i}{\partial x_j} - \frac{\partial u_j}{\partial x_i} \right) \quad (16)$$

The wall boundary conditions are:

$$\nu_{wall} = 0 ; \nu_{farfield} = 3\nu_{\infty} : to : 5\nu_{\infty}$$

The constants are:

$$\begin{aligned} C_{b1} &= 0.1355 ; \quad \sigma = 2/3 ; \quad C_{b2} = 0.622 ; \quad K = 0.41 \\ C_{w2} &= 0.3 ; \quad C_{w3} = 2 ; \quad C_{v1} = 7.1 ; \quad C_{t3} = 1.2 ; \quad C_{t4} = 0.5 \\ C_{w1} &= \frac{C_{b1}}{K^2} + \frac{1 + C_{b2}}{\sigma} \end{aligned}$$

B. Two Equation $k-\varepsilon$ models

There are several two-equation models. Three of the more popular, accurate and widely used models are the various variants of the $k-\varepsilon$ model. All the three models can be used for a range of flow problems with good accuracy. In the $k-\varepsilon$ model transport equations are solved for two quantities. The first variable is turbulent kinetic energy (k) and the second variable is the turbulent dissipation rate (ε). The eddy viscosity is determined by the relation

$$\nu_t = C_{\mu} \frac{k^2}{\varepsilon} \quad (17)$$

B1. Standard $k-\varepsilon$ model

The standard $k-\varepsilon$ model is derived by assuming that the flow is fully turbulent and the effects of molecular viscosity are negligible. For locations near walls, the standard $k-\varepsilon$ model, therefore, demands an additional model, which comprises the effects of molecular viscosity. In this situation, wall functions based on semi-empirical formulas and functions are employed.

B2. RNG $k-\varepsilon$ model

In the standard $k-\varepsilon$ model the eddy viscosity is determined from a single turbulence length scale, so the calculated turbulent diffusion is that which occurs only at the specified scale, whereas in reality all scales of motion will contribute to the turbulent diffusion. The RNG model was developed using Re-Normalization Group (RNG) methods to renormalize the Navier-Stokes equations, to account for the effects of smaller scales of motion.

B3. Realizable $k-\varepsilon$ model

The term "realizable" means that the model satisfies certain mathematical constraints on the normal stresses, consistent with the physics of turbulent flows. The realizable $k-\varepsilon$ model was intended to address the deficiencies of traditional $k-\varepsilon$ models by adopting the following:

- a new eddy-viscosity formula involving a variable C_μ originally proposed by
- a new model equation for dissipation (ε) based on the dynamic equation of the mean-square vorticity fluctuation

IV. PIPE FLOW PROBLEM

For a known laminar case, the velocity profile should become parabolic and match with the laminar simulation. The effectiveness of the turbulence model at low Reynolds number ranges is judged by the ability of the model to reproduce laminar flow solution even with the turbulence model present. For this to happen, the eddy viscosity in such cases should reduce to almost zero. To compare the various turbulence methods pipe flow problem was first analyzed. A pipe geometry was created in GAMBIT with Pipe Diameter of 2 m, Pipe length of 160 m with structured meshing as shown in Figure 1. Grid Independence was checked for Reynolds number 500, grid independence was achieved at cell count of 3750.

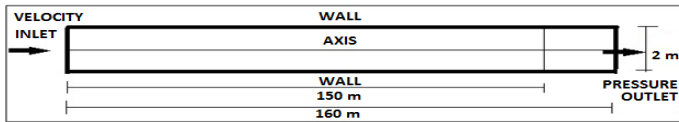


Figure 1: Dimensions of pipe and boundary conditions

Simulations were performed at different Reynolds numbers ranging from 500 to 10000. Graphs were plotted for different simulations and results were analyzed for two different parameters

- Velocity profile at different Reynolds numbers (Figure 2).
- Turbulent viscosity ratio as a function of the radial distance (Figure 3).

Figure 2, shows the velocity profile (fully developed) at different values of Reynolds number. It is observed from Figure 2 that for $Re < 1000$, the profile observed by S-A model is quite close to the laminar profile. The other models predict a turbulent profile. This is further verified when turbulent viscosity ratio (V_t/ν) is observed for laminar case. While the $k-\varepsilon$ models predict a value of

V_t/ν in the range of 3 to 4 for $Re=500$, the S-A model shows the maximum value of V_t/ν as 0.3.

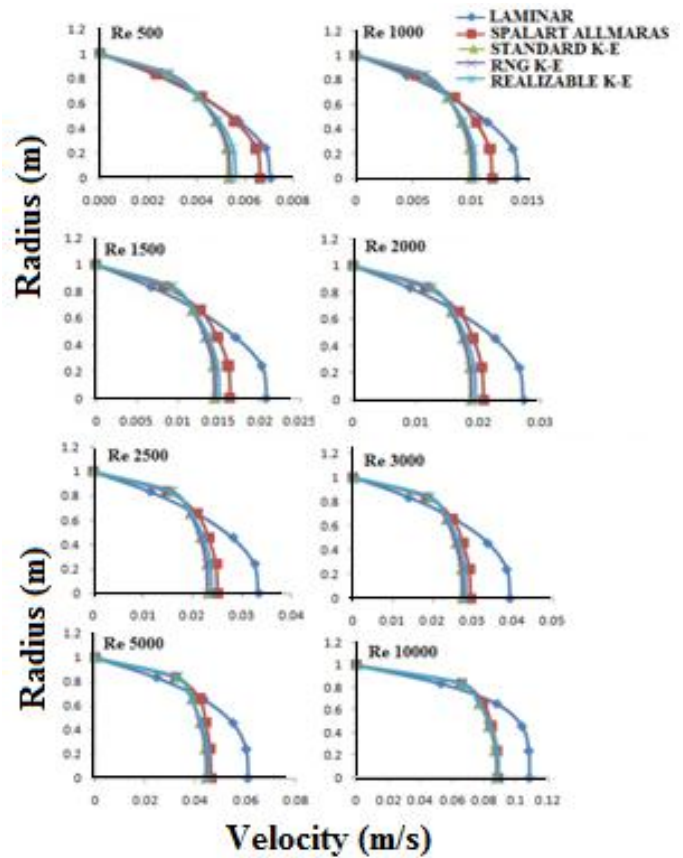


Figure 2: Velocity profile at different Reynolds numbers

Thus the above results clearly indicate that as Reynolds number of flow decreases in a pipe, the one equation S-A model predicts the flow better than two equation $k-\varepsilon$ models.

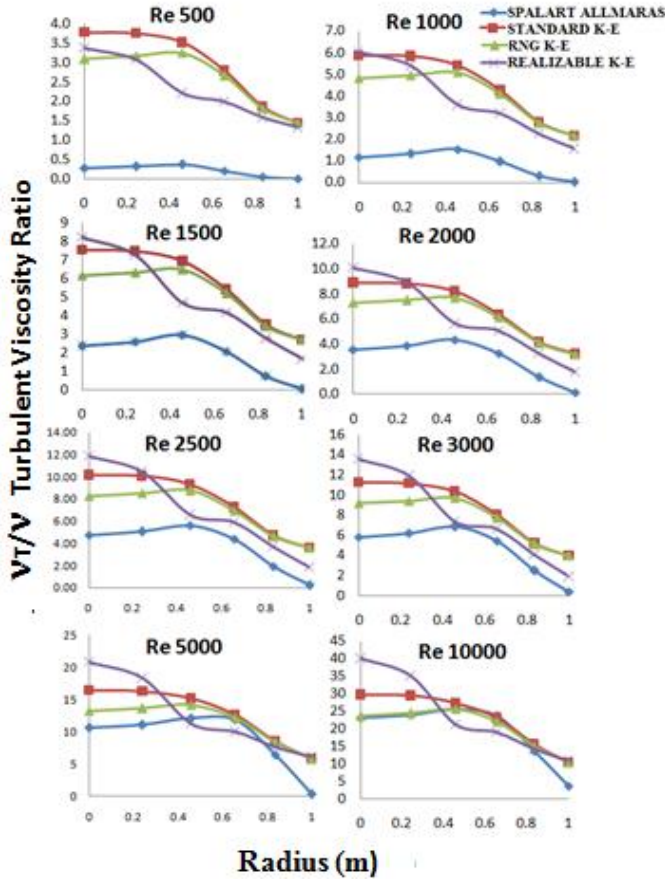


Figure 3: Turbulent viscosity ratio as a function of the radial distance

V. LID DRIVEN CAVITY FLOW PROBLEM

From the above analysis it is evident that the S-A model at low Reynolds numbers, in the near transition region for pressure driven internal flows gives better results than $k-\epsilon$ models. The next step is to enquire whether the S-A model can predict the flow in shear driven flows better than $k-\epsilon$ models or not. The lid driven cavity flow problem is selected for our investigation, as it is one of the most investigated problems in CFD and benchmark results are available for it. We compare the results with (Ghia *et al*, 1982). The geometry for the lid driven cavity flow problem was created in GAMBIT with structured meshing having a mesh size of 129 x 129. Following Ghia *et al*, (1982), uniform mesh refinement as shown in Figure 4 was used. The simulation of flow through the 2-D lid driven cavity flow problem was performed on FLUENT 6.3.26 at different Reynolds numbers ranging from 100 to 3200. Figure 5 shows the variation of the x- component of velocity along a vertical plane passing through the centre for different Reynolds numbers.

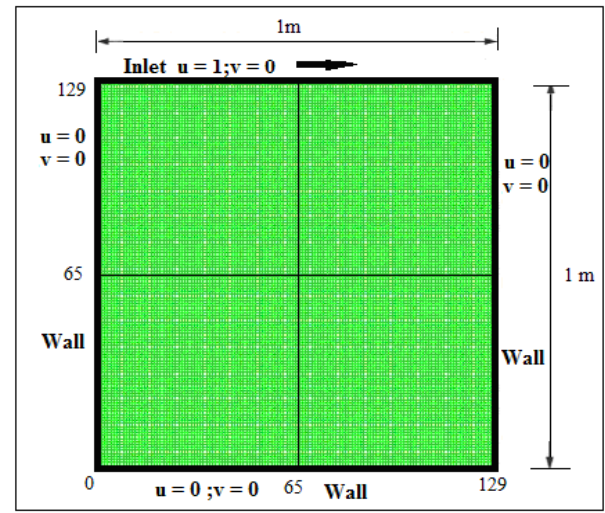


Figure 4: Lid driven cavity geometry and Boundary condition

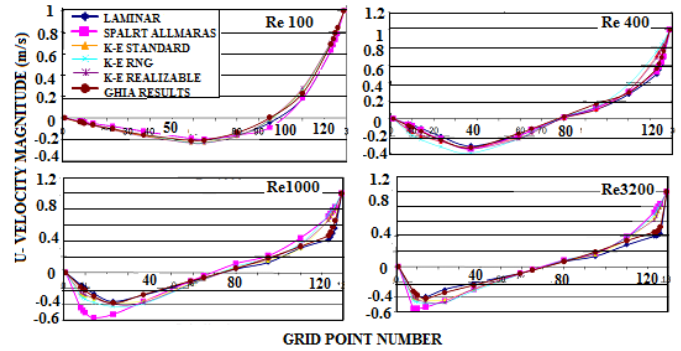


Figure 5: U- Velocity at a vertical line through the centre for different Reynolds numbers

From the above results it can be concluded that in this class of flows all the models perform similarly. The velocity profile as predicted by all the models is almost identical to the results obtained by Ghia *et al*, (1982).

VI. EXPERIMENTS ON THE CYLINDRICAL MODEL OF HELMET

Initially, the computational simulations are carried out for a 2- dimensional case. This is done to study the flow and its characteristics, i.e. whether the flow is predominantly 2-dimensional or 3-dimensional. The experiments were conducted in the wind tunnel with cross section of 450 mm x 750 mm with different inlet velocities (Yadav, 2006). The wind tunnel with experimental model is shown in Figure 6. In order to measure the flow velocity in the top of air gap of helmet, a hole was made in the helmet model, and a 3-hole probe was inserted from the top to measure velocities at different distances from the head in the helmet and for measuring the inlet velocity impinging on to the model, a

Pitot tube was placed in the front of the model, at a distance of 15 cm from top of wind tunnel, in the middle plane of the wind tunnel. The Pitot tube was connected to the Betz micrometer. Pressure readings were also taken at the back of the model assembly, by mounting a Pitot tube.

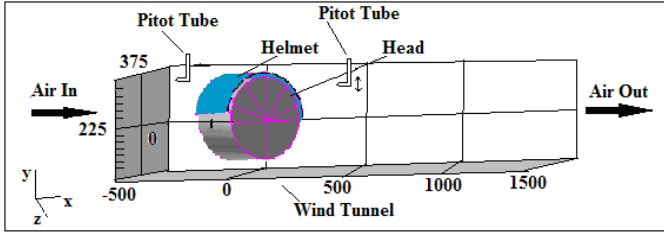


Figure 6: Wind tunnel with experimental setup

The cylindrical model of helmet and head with holes for pressure measuring tab locations on central plane is shown in Figure 7. The geometrical structure of the setup was an elongated cylinder, which was substituted for the human head, suggesting the flow to be 2-dimensional as the velocity profile and distribution along any cross section of the geometry will be similar, as long as the end effects are negligible. It consisted of a cylinder with 75 mm radius and the helmet as a cylindrical shell subtending an angle of 180 degrees from the center. The experiment was done with a 7 mm air gap. The material of the setup was chosen to be Perspex due to its availability and applicability.

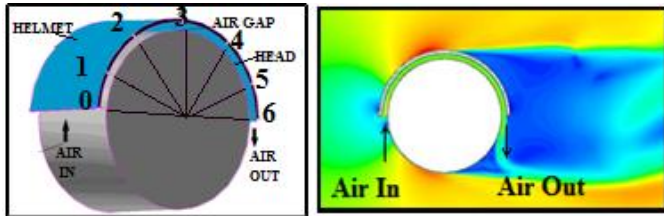


Figure 7: Cylindrical model of helmet and contour of U-velocity

A. Observations

The mesh in the vicinity of the model was made fine, and it was made coarser as distance from the center increases radially. Grids were made as fine as possible in the vicinity of helmet. Meshing in the air gap is done separately with structured uniform mesh. Results of simulation indicate that in the air gap the velocity decreases close to the walls but it increases with the distance from the wall and it is maximum near the centre of the gap. The x- component of velocity is maximum near the centre of the air gap as shown in Figure 8.

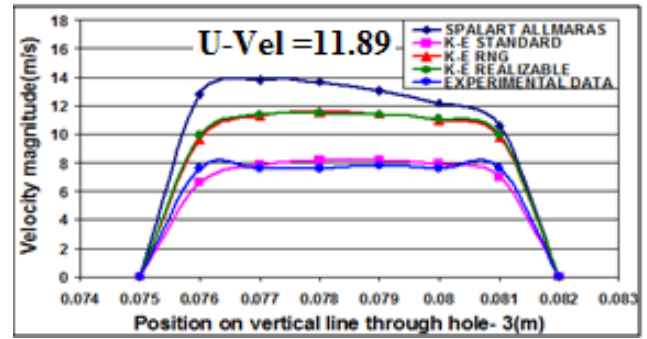


Figure 8: Comparison of velocities in helmet air gap

A low pressure region is formed towards the downstream side of the air gap and helmet. The results of the simulation were checked, and it was found that pressure drop readings from the CFD simulations did not show a close match with the experimental results. Figure 9 shows the variation of pressure at the centre of the air gap as a function of the downstream location.

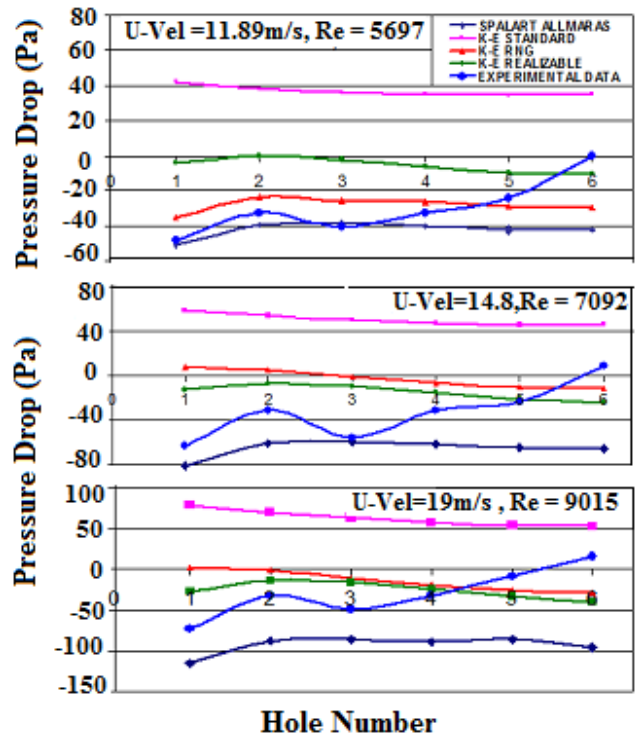


Figure 9: Variation of pressure at the centre of the gap as a function of the downstream location

The results presented above indicate that the experimental results do not match very well with the numerical results for all turbulence models. This is probably because in the simulation a 2-D model has been used which blocks the flow and forces the air in the gap between the head and the helmet.

VII. 3- DIMENSIONAL HEMISPHERICAL MODEL OF HELMET

The experimental set up was made similar to the 2-D model analysis. The hemispherical helmet model for the experiments was cut from a plastic ball of 222 mm diameter which was then fixed to a hollow wooden cylinder as shown in Figure 10.

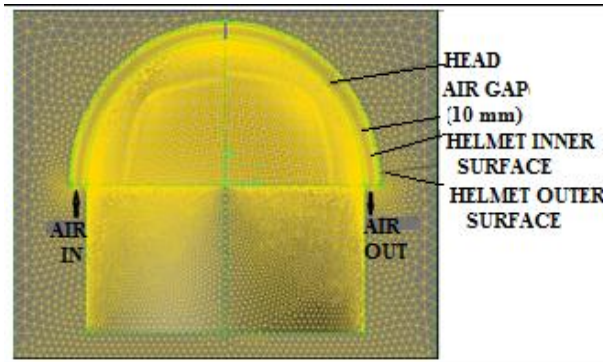
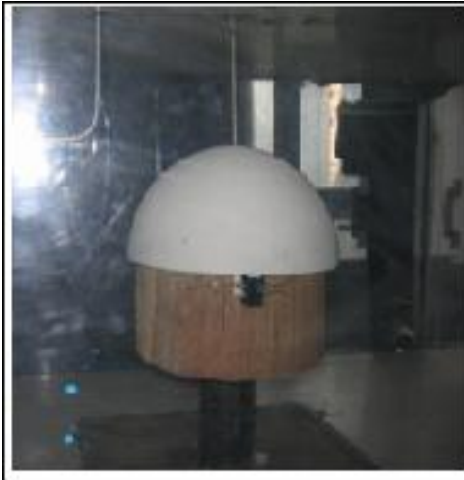


Figure 10: Head-helmet arrangement (Yadav, 2006) and mesh in air gap of the 3-dimensional hemispherical model of the helmet.

The experiment was performed with inlet velocity of 18 m/s and backpressure of 354 Pa (Yadav, 2006). A three hole probe was used to measure the velocity component in the direction of flow at eight points on the top of head in the air gap. The readings of the Pitot tubes were also taken at the downstream side.

A. CFD Simulation on 3-Dimensional hemispherical model of Helmet

The simulations of the experimental model were carried out using Fluent 6.3.26. The geometry was constructed in GAMBIT and meshing of head and helmet was done with a quadrangle mesh. The domain very near to head and helmet was meshed with a triangular mesh

and the remaining domain was meshed with Hex/Wedge mesh as shown in Figure 11.

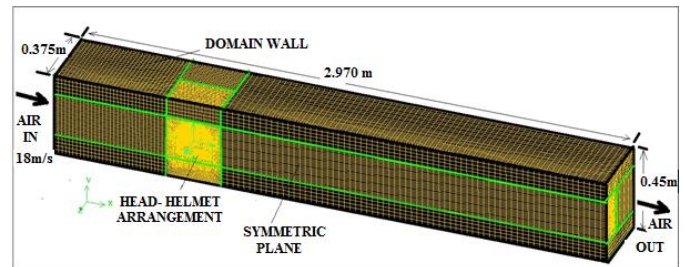


Figure 11: Mesh for the entire domain along boundary conditions for the 3-dimensional hemispherical model of helmet

The head is considered as a hemisphere of radius 99 mm and the helmet inner surface radius was 109 mm, so that the air gap between head and helmet was 10 mm. The thickness of helmet is 2 mm. Since the geometry is symmetrical about the central XY plane, CFD simulations are performed only on half the section with the symmetry boundary condition applied on the central plane. Grid independence was achieved with mesh size of 1318019, when there were 20 mesh elements in the air gap. The meshed head-helmet arrangement and the air gap between head and helmet are shown in Figure 11. The S-A and k- ϵ models were used to carry out simulations. Air is admitted in the flow domain with velocity of 18 m/s and at a back pressure of 354 Pa. The velocity of air at the top of the head is compared with the experimental data in Figure 12.

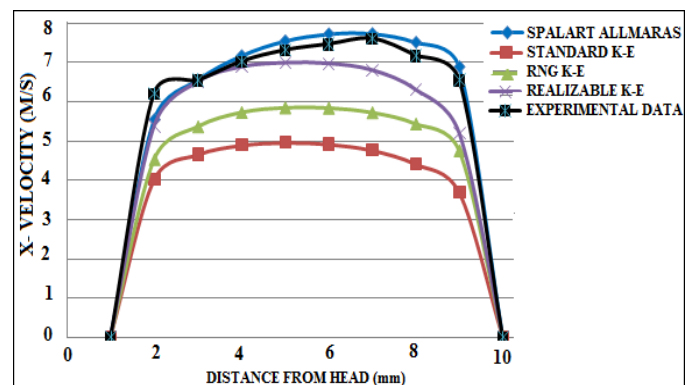


Figure 12: Comparison of x-velocities on a vertical line at the centre of the air gap at the top of head.

The average value for the root mean square deviation of velocity obtained from a particular model and the experimental data (averaged over 8 non wall points) at the top of head is presented in table 1.

TABLE 1: AVERAGE ROOT MEAN SQUARE DEVIATIONS OF VELOCITY FOR DIFFERENT TURBULENCE MODELS.

	Spalart Allmaras	Standard k- ϵ	RNG k- ϵ	Realizable k- ϵ
Average RMS Deviation	0.3262	2.4711	1.5835	0.71409

The results indicate that the S-A model captures turbulence better than the $k-\epsilon$ models for 3-dimensional hemispherical model of the helmet and the head.

The S-A model performs better than the $k-\epsilon$ standard models at low Reynolds numbers in the near transition region for internal pressure driven flows possibly due to following reasons:

- 1) In the boundary layer the blocking effect of a wall is felt at a distance through the pressure term, which acts as the main destruction term for the Reynolds shear stress. This suggests that there should be a destruction term in the transport equation for eddy viscosity (Spalart *et al*, 1994). In the S-A model the transport equation for eddy viscosity contains a *destruction term*

$$C_{w1} = \frac{C_{b1}}{k^2} + (1 + C_{b2})/\sigma \quad (18)$$

This destruction term establishes an equilibrium between the production and diffusion term (all positive) in the log layer, but it decays very slowly in the outer region of the boundary layer. To address this deficiency, a non-dimensional *decay function* (f_w) is included in the transport equation. In $k-\epsilon$ standard models no such destruction and decay terms are present so they may not give good results when used for simulations of flows near wall.

- 2) In the S-A model turbulent viscosity is derived from modified turbulent viscosity ν and the near wall damping function f_{v1} i.e. $\nu_t = \nu f_{v1}$

The eddy viscosity ν_t equal to $ky\mu_t$ in the log layer but not in the buffer layer and viscous sublayer. To overcome this deficiency in the S-A model a transport quantity ν also known as modified turbulent viscosity is defined, such that ν is equal to $ky\mu_t$ all the way to the walls. ν is multiplied by the near wall *damping function* f_{v1} . This near wall damping function is constructed in such a way that ν maintains its log – layer behavior all the way to the walls. At low

Reynolds numbers due to the damping effect of f_{v1} the flow is closer to the laminar profile with the S-A model solving the flow for modified turbulent viscosity ν closer to zero.

- 3) The S-A model solves a transport equation for the eddy viscosity directly and the destruction terms account for the near wall effects. However in the $k-\epsilon$ models the transport equations are solved for both k and ϵ and the eddy viscosity is calculated as a ratio of k^2 and ϵ . Thus, when the flow is laminar unless k^2 goes to zero faster than ϵ , the eddy viscosity will not decay to zero, which should be the case for laminar flows.

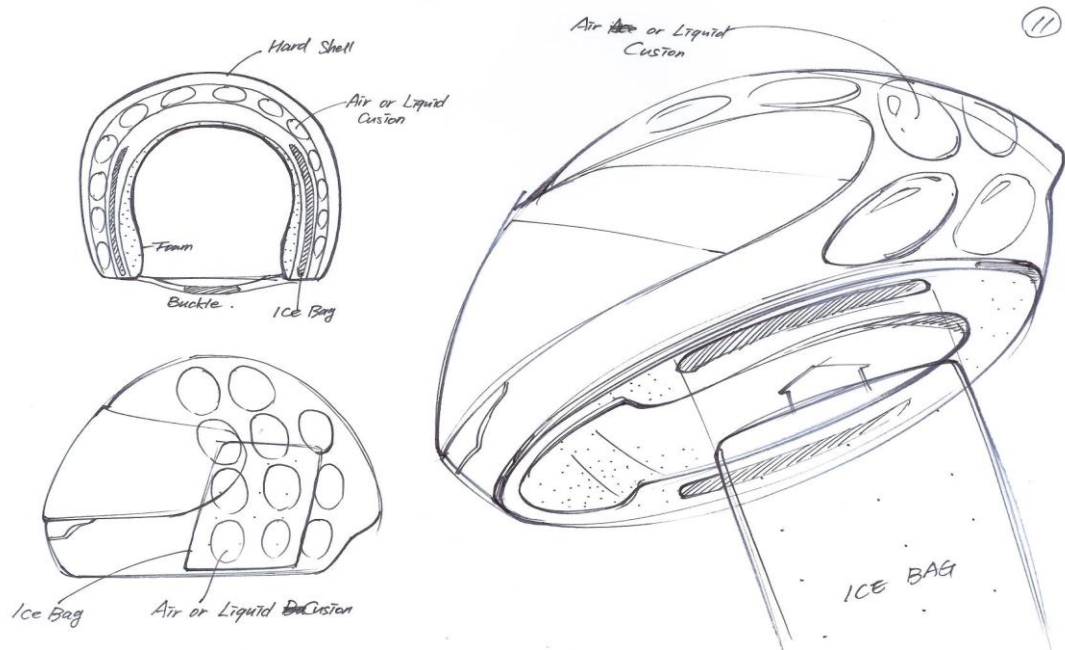
VIII. CONCLUSIONS

The above results support our finding that the one equation S-A model predicts the flow better than two equation $k-\epsilon$ models in the near transition region for pressure driven near wall internal flows such as pipe flows. In the fully turbulent region, the performance of this model is almost identical to that of the standard $k-\epsilon$ models. This is a significant finding because the S-A model was initially developed for open flows past a body (i.e. a semi-infinite domain) with one end being the wall and the other ‘end’ being open. The S-A model is found to provide a better matching with the experimental results as compared to the $k-\epsilon$ models in 2-D cylindrical and 3-D spherical helmet geometries.

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Helmet Performance and Design



Helmet Research in the WP3 of the MYMOSA Project

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ABSTRACT

The Research Training Network MYMOSA was a project funded by the European Union to train young researchers in motorcycle safety. One of its work-packages, number 3, was about personal protective equipment and was almost only devoted to helmets. Three early stage Researchers were employed and worked on two main topics: a critical revision of the safety helmet standard currently adopted in the EU and the examination of innovative materials/structures to be used in the manufacturing of novel energy absorbing liners. The main findings of the research activities carried out within WP3 are summarized in the present paper.

Keywords: *helmets; standard; energy absorbing liner, passive safety.*

NOMENCLATURE

ESR: early stage researcher.
MCRTN: Marie Curie Research Training Network.
HIC: head injury criterion.
ABS: acrylonitrile butadiene styrene.
EPS: expanded polystyrene.
PC: polycarbonate.

I. INTRODUCTION

The acronym MYMOSA stands for Motorcyclist and Motorcycle Safety and was chosen as the name of a Marie Curie Research Training Network funded by the European Union. Motorcyclists and moped drivers are road users with a particularly high accident risk since motorcycle accidents are severe in nature, due to the relative lack of protection of motorcyclists. It is well known that in Europe riders represent only 6-8% of road users but 16-18% of road fatalities [1]. Furthermore, given the young age of many victims, these accidents often result in a high loss of life expectancy for fatalities and high social-economic costs for severely injured motorcyclists. The ambition of this project was to provide a significant contribution to the education of new experts in the field of road safety with a particular emphasis on powered two wheelers.

The prime objectives of MYMOSA were:

- to educate several Early Stage Researchers (ESRs) in the partially unexplored field of Powered Two Wheelers' and riders' safety
- to facilitate the development of R&D abilities and the formation of a European network of personal relationships in an early stage of the careers of the researchers (many years benefiting their careers/specialization)

- to stimulate co-operation between researchers of 5 universities, 3 research centres and 6 industries (2 SMEs) through visits, secondments and training.

The research work in the network was organized in four interacting work-packages, as shown in Figure 1:

- WP1, accident dynamics,
- WP2, integrated safety,
- WP3, personal protective equipment and
- WP4, biomechanics.

The present paper will summarize the main work carried out within the WP3 of the MYMOSA project.

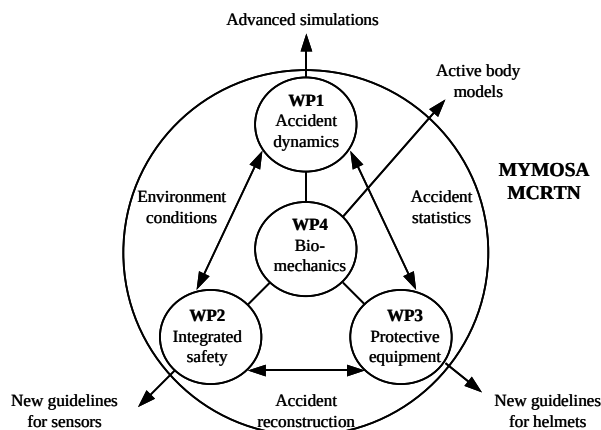


Figure 1: Sketch of work-packages and expected outputs of MYMOSA

II. WP3 - PERSONAL PROTECTIVE EQUIPMENT

Three post-graduate researchers were hired within the project to work on personal protective equipment, they were called, according to EU jargon, ESRs (early stage researchers). The first ESR (36 man-months) working on personal protective equipment examined the helmet standard currently adopted within the EU and suggested further investigations on how to make it more relevant to real life accidents. Two other ESRs (for a total of 50 man-months) worked on new concepts for improved motorcycle helmets. Two ESRs were based at Imperial College London and the third one at Dainese SpA, Italy. Helpful support was provided as well by the Biomechanical Strasbourg team at Université Louis Pasteur, LMU Munich, DEKRA (D), TRL (UK) and Cellbond Ltd (UK). After a brief revision of the standards for safety helmets currently in use in the EU, a summary of the main findings of the three early stage researchers will be given.

A. Helmet standards currently adopted in the EU

The adopted standard has a crucial importance on the safety of riders because it provides the criteria according to which helmets will be evaluated, often with limited direct reference to the mechanics and the biomechanics of real life accidents. The main tests according to the ECE 22.05 Helmet Safety Standard [2] are impact tests based on the use of equipment such as that shown in Figure 2. The impact points are B, P, X, R of the helmet, as shown in Figure 3, two types of anvils, flat and kerbstone (see Figure 4), are used. Impact tests have to take place at given impact speeds and at prescribed temperatures in order to consider the variation of mechanical properties of the materials in the different seasons of the year. For all prescribed impact conditions the linear acceleration measured at the center of mass of the headform during the impact must always be below a given limit. Moreover another parameter called Head Injury Criterion (HIC), more related to the duration of the impact, must be below a fixed threshold.

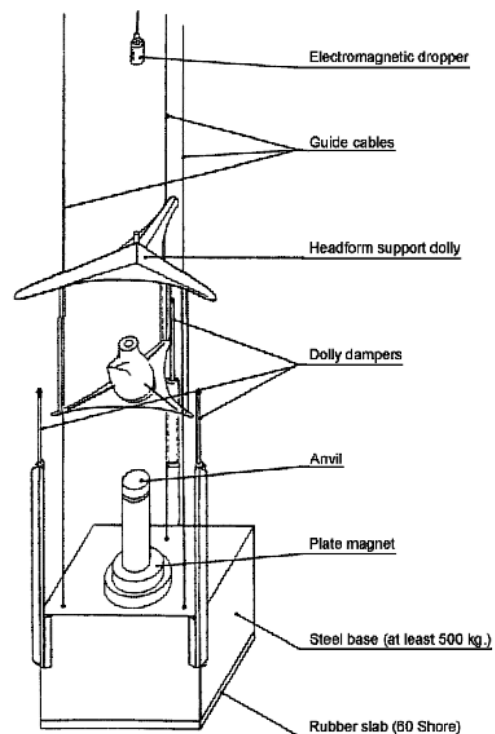


Figure 2: Sketch of the equipment used in the standard impact tests [2]

B. Proposal of modification of standard tests

It is apparent that the main difference between standard tests and real accidents is given by the fact that in real accidents the body of the rider influences the

dynamics of the head, whereas such an influence is not present in the lab, where the headform is not connected with any ‘body-form’. In order to reveal possible influences of the presence of the body on the impact response of the helmeted head, helmet drop tests using the Hybrid III dummy (full-body) were simulated and compared with simulations of drop tests in which only the detached head of the dummy was used [3]. A second step of the research involved lab experiments with the Hybrid III dummy and they validated the results of the simulations [4, 5]. The FE model of the AGV-T2 helmet was positioned on the dummy’s head and on its detached head. The simulated impacts were against a flat anvil at two impact velocities, 6 m/s and 7.5 m/s. The former had previously been used as well in the COST study [1] to perform the same comparison but experimentally. The results presented in [3-5] show that in the full-body impact, the magnitude of the acceleration $|a|$ rises sharply for impact speeds above 6 m/s and exceeds that recorded in tests in which a detached head was used. This is in contrast to the behavior exhibited for impact speeds below 6 m/s and reported in previous experimental studies (Aldman, et al., 1976, Aldman, et al., 1978a, Aldman, et al., 1978b, COST327, 2001). This phenomenon is the consequence of the bottoming out of the liner. Increasing the impact speed from 6 m/s to 7.5 m/s caused more deformation of the liner so that its maximum compressive strain in the crushed region reached 91% (using an initial thickness of 42 mm) for the dummy drop test. As a consequence the maximum value of the acceleration $|a|_{\max}$ rises considerably, much more than proportionally with respect to the value of the impact speed, and the maximum value of the helmet-anvil contact force $F_{hN,\max}$ was far larger than the skull fracture threshold, which indicate that the energy absorption capacity of the helmet was not sufficient for this impact. It was shown in [4, 5] that the numerical results of [3] represented accurately what is happening in reality and therefore that the presence of the whole body results in further crushing of the liner. Therefore the body has an important effect, which should be considered in the impact absorption tests. Since using a dummy to test helmets would have a drastic impact on their cost, other measures should be adopted. The numerical and experimental results given in [3- 5] indicate that when the liner was not loaded beyond its energy absorption capacity (at an impact speed of 6 m/s), the maximum value of the head acceleration was lower in the case of full-body impact, but the contact force between helmet and anvil and the reduction in liner thickness were greater when the effect of the body was included. It is possible to show that the only modification to the helmeted

headform impact inputs that influences the outputs in the same way is an increase of the mass of the headform.

A dimensionless parameter (γ_m) called the ‘added mass index’ has been defined, which is the ratio of the proposed increase in the headform mass to its original mass. This index quantifies the effect of the body on the impact response of the helmeted head. Using a heavier headform with the same limit of head linear acceleration can cause helmet manufacturers to use stiffer foams with higher yield stress. Consequently, a helmet designed for a heavier headform may induce higher head decelerations due to its stiffer liner as compared to a helmet approved according to the current standard test. If the mass of the headform is to be increased by γ_m , then the limit of head acceleration set in the standard should be decreased by $(1+\gamma_m)^{0.5}$ in order to avoid the design of helmets which have too stiff liners [3-5].

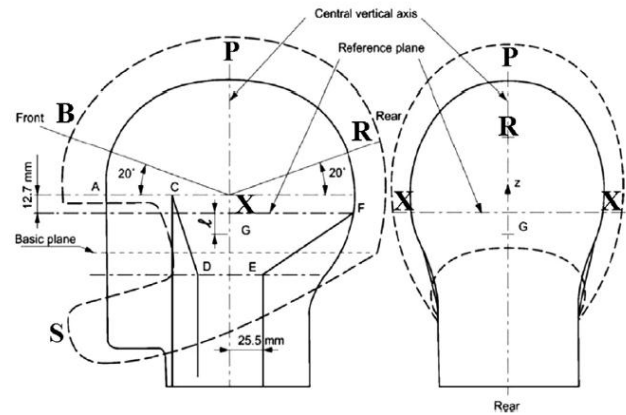


Figure 3: B-front, P-top, X-side, R-rear are the for impact points on the helmet [2]

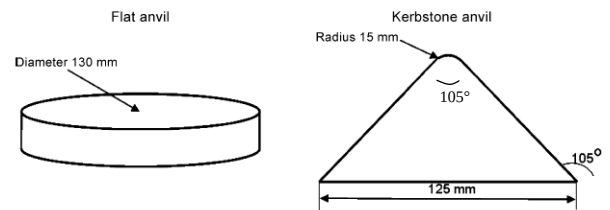


Figure 4: Flat and kerbstone anvils used in the standard tests [2]

C. New concepts for safety helmets

The structural parts of helmets responsible for impact management are basically two: the outer rigid shell and the energy absorbing liner. The outer shell is usually made of thermoplastic materials such as Acrylonitrile Butadiene Styrene (ABS) or Polycarbonate (PC), or composite materials such as Glass Reinforced Plastics. The main function of the outer shell is to spread the

impact load over a wide area of the head in order to reduce local pressure and to avoid direct contact with sharp objects.

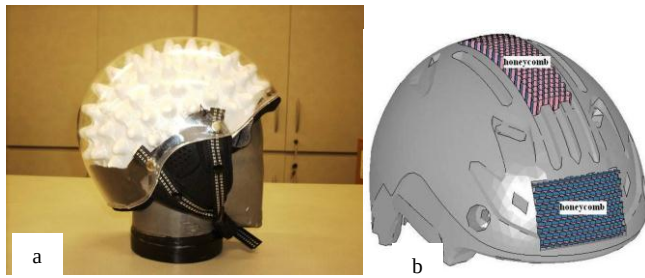


Figure 5: Prototypes produced in MYMOSA [9, 12]

The liner is the part of the helmet that absorbs the greatest portion of impact energy during crashes by providing a stopping distance and is usually made of Expanded Polystyrene (EPS). The use of EPS has some drawbacks, such as the difficulty to optimize energy absorption in different areas of the head and the excessive insulation that prevents heat evacuation. The MYMOSA ESRs carried out some research aimed at exploring the possibility of using other materials for the inner liner [9-12]. They examined two new concepts of helmet liner as shown in Figure 5. Figure 5(a) shows a helmet with (transparent shell and) an innovative liner realized in project-a: the novel liner consists of a plastic lamina shaped into deformable cones. Energy is absorbed via a combination of folding and collapsing of the cones. The main advantage that such liner may introduce over common EPS pads is that it allows a better optimization of energy absorption for different impact sites and configurations, moreover it allows for a better ventilation of the rider's head. Figure 5(b) shows an FE model of the liner realized in project-b which consists of an assembly of two different energy absorbing materials: the traditional EPS and a honeycomb structure made of aluminum. The honeycomb structure can be much more efficient than EPS and so can be used in specific locations of the liner (not necessarily those shown in the Figure) to make it lighter with no reduction in energy absorbing capabilities.

The goal of project-a [9] was to define a procedure based on the execution of Finite Element analyses and optimization routines, which is able to suggest safer ways of employing new energy absorbing materials for the manufacture of safety helmets. The project was mainly focused on the helmet energy absorbing liner, which is the component that absorbs the greatest amount of energy during an impact. The innovative liner consisted of an ABS plastic lamina with deformable cones on it. Energy was absorbed via a combination of folding and collapsing

of the cones. The work aimed at studying this new liner as alternative to the current EPS foams.

To carry out the research, multivariable optimization tools were used (Optimus [13] and LS-OPT [14]) to select the safest design configurations among all the technologically feasible possibilities. A new computerized approach including automated CAD update design, meshing and job submission to the finite element solver LS-DYNA [15] was established at Dainese S.p.a. This technique allowed for a quick parameter evaluation and subsequent liner design optimization.

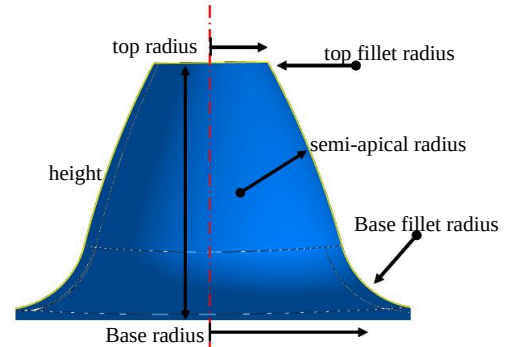


Figure 6: Main parameters defining the shape of a cone [9]

The first step of the work consisted in the examination of relative importance of various shape parameters of the cones: virtual impact tests were carried out to on a single cone to assess its energy absorption capabilities. After due considerations, seven input parameters (shown in Figure 6) and their respective ranges were fixed:

Inputs:

- Base radius: 5-10 mm
- Top radius: 2-10 mm
- Base fillet radius: 0.1-2 mm
- Top fillet radius: 0.1-2 mm
- Semi-apical radius: 20-100 mm
- Height: 15-25 mm
- Thickness: 0.5-1.5 mm

The Outputs :

- Energy absorbed (J)
- Peak Force (N)

The results obtained from this analysis showed that the thickness is the most correlated parameter to all the outputs and height, top and base radius are significant as well. Therefore, starting from seven design parameters, this first single cone study suggested that the most significant design parameters are four: top and base radius, height and specially thickness.

The second step of the project consisted in the optimization of the helmet (shown in Figures 7 and 8). A FE model of a Dainese jet helmet was created and numerical impact tests were carried out at points prescribed by the European helmet standards ECE 22.05 [2]. The liner consisted of an ABS lamina with deformable cones on it. The initial cone dimensions were set up according to some manufacturing concerns. The parameters to be optimized were reduced to two, thickness and top radius. Fixed parameters were height 35 mm (fixed by spacing available between the shell and headform), base radius = 21.5 mm (fixed by manufacturing issues), semi-apical arc radius = 60 mm, base fillet radius = 6mm and top fillet which was removed. First results indicated that kerbstone impact was critical compared to that on flat anvil for this specific type of liner. Furthermore, simulation results showed the importance of tying the cones to the shell in impact conditions. Numerical impacts on the front side put in evidence that the cones bent in the compression stage if no constrained was applied. This was attributed to the existing shear forces that appeared between the cones and shell interface. Therefore, the onset of cones bending resulted in a less efficient energy absorption folding mode as compared to the cone axial collapse. Due to geometrical reasons, bending may be more relevant in helmet regions with lower radius of curvature (front and rear).

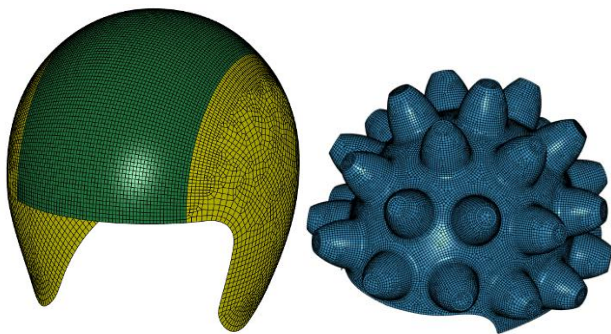


Figure 7: Helmet shell and an example of liner with a relatively small number of large cones [9]

Optimized impact results showed better helmet performance on front and rear areas compared to side and top regions. This was due to the fact that on large radius areas (flat), the shell suffered from premature buckling compared to areas with smaller radius of curvature, making more difficult to properly spread the impact load. Hence, kerbstone impact became even more critical on side and top regions compared to the front and rear areas. Problems regarding excess in stiffness were found at the crown site. Potential solutions included new cones liner

distribution to comply with the standards. According to the current design, there may be a considerable difference when the kerbstone impacts the helmet between the cones or on a cone area. The increase of the shell stiffness was proved to partially solve this problem by better spreading the impact load.

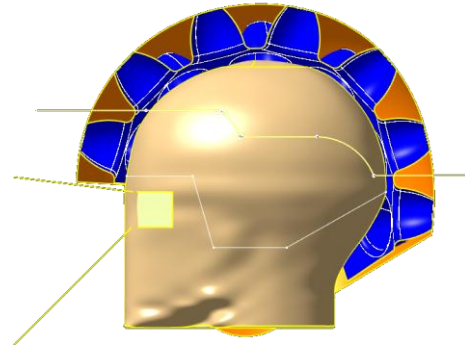


Figure 8: Cone-liner into the helmet [9]

It has to be pointed out that the choice of this design was made in accordance to a possible manufacturing design which established 21.5 mm of base radius for a height of about 35mm. Another possibility to be studied is to make smaller cones and further cover the shell area in order to better manage the kerbstone impact. The comparison between the new liner and the traditional EPS liner in impact conditions was the last step of the work and showed promising results, see Figure 9. In general, there was a reduction in peak acceleration and especially in HIC values. The results of the four impacts on kerbstone anvil are shown in Figure 10. The main advantage that this helmet may introduce to motorcycle community besides impact management is the comfort. A decrease in weight is expected and no need of special ventilations may be required as air could easily flow between the cones. Furthermore, 100 % recyclable materials would be employed for the manufacturing of such liner (engineering plastics: ABS, PC). On the other hand, the main drawbacks or technological issues to be solved are the manufacturing and gluing process of the liner.

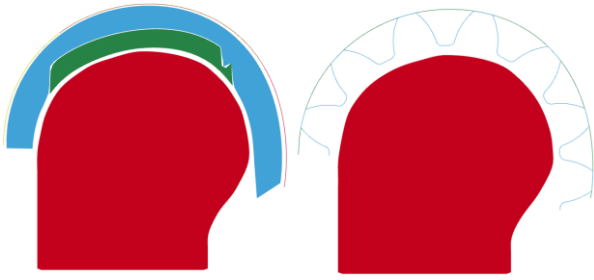


Figure 9: Meridian vertical sections of traditional helmet and new helmet [9]

The goal of project-b [10-12] was to examine the coupling of aluminum honeycombs and EPS foams for the design of an innovative helmet liner. In this case the research was mainly experimental and therefore the optimization procedure had to stop at a much earlier stage than in the case of project-a. The impact behavior of a modified version of a commercial helmet, where aluminum honeycombs were introduced in the front, top and rear region of the energy absorbing liner, was assessed following ECE 22_05 standard [2]. The modified liner is shown in Figures 5(b) and 11. Unmodified helmets, presenting same geometry and material properties (except for the honeycomb inserts), were also tested under the same conditions. Dynamical responses of the two helmets were compared and peak linear acceleration and HIC were used as evaluation criteria. Simulations of the impact tests were carried out, but they are not presented here [12]. Various complex issues had to be dealt with for the simulation of the impacts, especially for the prototype. In particular the simulation of the constitutive behavior and the definition of the contact logic governing the interaction of the various materials proved particularly challenging. Figure 12 shows an example of compressive behavior of a sample made of EPS-foam and aluminum honeycomb tested in the lab of aero-structures at Imperial College London.

Comparing the first prototypes with a commercial helmet is a very demanding approach since the comparison is carried out with the performance of a helmet which has already undergone a stringent optimization procedure. The Dainese commercial helmet easily passes all standard limits. Therefore it is clear that any improvement of its performance is rather difficult.

When comparing the results of impact tests corresponding to different impact sites and anvil types, various trends were observed for the two evaluated helmet designs.

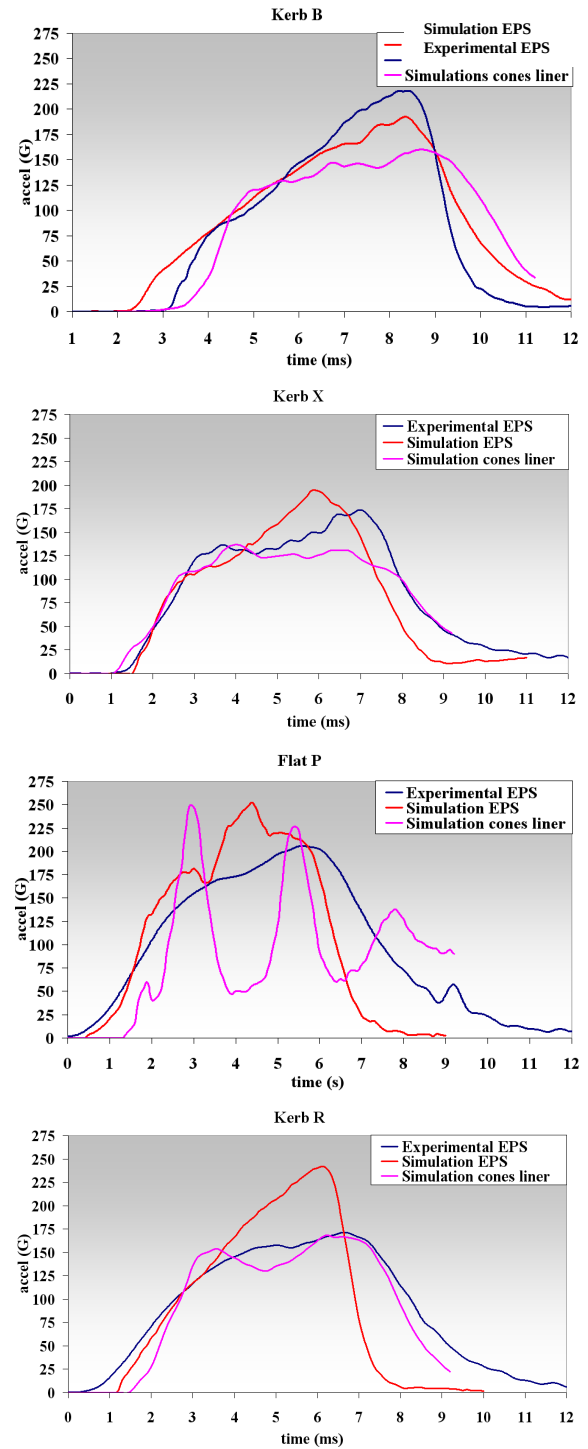


Figure 10: Comparison of impact results for helmets with innovative and traditional liners [9]

Generally, the prototype helmets provided better protection to the head from impacts against the kerbstone anvil, in particular by significantly reducing peak linear acceleration and HIC during impacts on the front and the rear surfaces. Sensitivity of results to anvil shape is frequently observed and has been already described in a

previous experimental study on the dynamic behavior of helmets [16]. Different typologies of helmets were tested against flat and hemispherical surfaces. It was observed that forces transmitted to the head are linked to the load spreading material of the shell (the stiffer the shell, the larger the load spreading area).

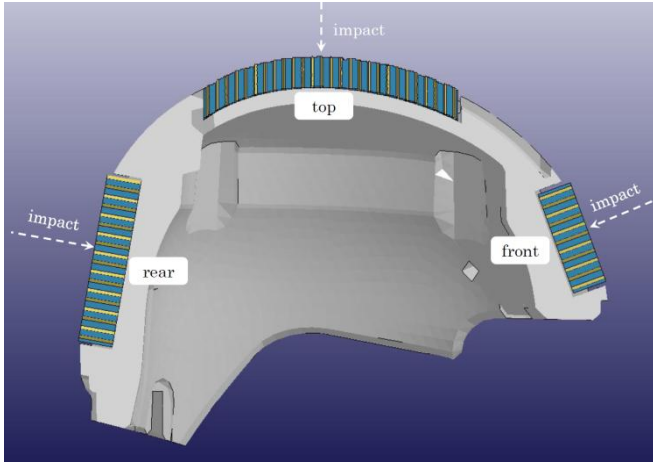


Figure 11: Vertical meridian section of the EPS-aluminum liner [12]

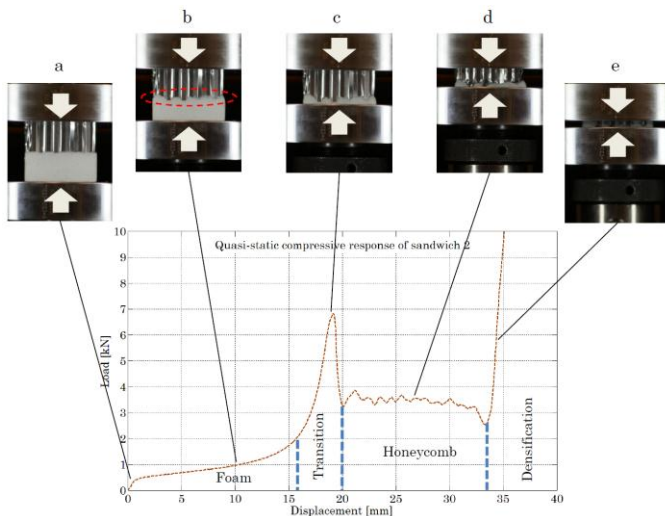


Figure 12: EPS-aluminum sample response to quasi-static load [12]

They observed that deformable shells provide better protection against flat surfaces, at expenses of protection against round surfaces. Conversely, stiff shells (such as the one used in our investigation) provide better protection against round surfaces at expenses of protection from impacts against flat ones. In addition to this, it was observed that the magnitude of the forces transmitted to the head was also dependent on the

stiffness of the underlying energy absorbing liner and the curvature of the shell in the impacted point. It was concluded that helmets cannot be optimized for all shapes of struck objects. In the research carried out in project-b, the trends observed are generally in agreement with results presented in literature. The improvements obtained were linked to the capacity of honeycombs to offer extended and constant plateau regime, which makes them capable of providing good shock absorption properties even at very high deformation stages. Some little improvements were also observed from impacts in the top region, but because of the variability of the results and the limited number of experiments carried out, it was not possible to confirm this trend. When impacts were performed against the flat anvil, the prototype top area provided best protection to the head, in terms of HIC. No significant improvements were observed from impacts on the front region, while impacts on the rear region highlighted inferior performances in comparison with the ones offered by the helmet commercial design. From observations of deformed prototype liners, it was concluded that the honeycombs in the front and rear areas did not contribute significantly to the impact energy absorption. This was attributed to a non uniform contact between the outer shell and the honeycombs during the impacts, to strain rate effects, which increased the honeycombs resistance, and to a non optimum design of the prototype liner. Surprisingly, significant reductions of the peak linear acceleration and HIC were observed from impacts on the lateral surfaces, not modified because of manufacturing difficulties, against both the anvils. It was assumed that the presence of honeycombs and the hollows in the liner might have influenced the load spreading capabilities of the helmet, and so the energy absorption. Nevertheless, observations of the damaged shell suggested that impacts did not always occur on the marked impact points, and that higher accelerations were observed when the impact occurred in proximity of the interface visor edge/shell, where the thickness of the shell was higher than in the surrounding areas. Thus, it was not possible to establish accurately the causes of such phenomenon and it is then believed that both the factors might have contributed to the difference between the prototype and commercial Gp-Tech dynamical responses.

However it must be noted that due to research time and budget restrictions, the manufacture of the prototype helmets was carried out following a non-industrialized process prone to imperfection. Moreover, such constraints did not allow for more prototypes to be made, so that there was no possibility to carry out any optimization of the prototypes.

On the basis of the results obtained in project-b it can be concluded that the use of aluminum honeycombs, as reinforcement material for the energy absorbing liner, can lead to an improvement of the safety levels provided by current commercial helmets without increasing their weight. Conversely, results from impacts against the flat anvil indicated to some extent the limitations of the strategy adopted in this research. Future work should address the optimization of honeycombs reinforced helmets for impacts against flat surfaces. Finite element analyses should be addressed to the design of prototype helmets where the gap between the outer shell and the inner liner is reduced to a minimum, especially in the rear region. Also, it would be interesting to assess of the prototype impact protection when more severe impact conditions or different standard regulations are considered. Future designs should also consider the extension of the areas covered by the honeycombs to the remaining surface of the liner, including the lateral surfaces. Most notably, this is the first study, to the knowledge of the authors, to investigate the effectiveness of helmets in which aluminum honeycombs are introduced in the liner. Results presented in this chapter could provide the framework for future research on the design of the honeycomb reinforced helmets, and to assess their performance characteristics.

III. CONCLUSIONS

MYMOSA was a successful Research Training Network that provided state of the art training to a considerable number of young researchers. The ‘Personal protective equipment’ work-package trained three ESRs who produced some interesting ideas and several international publications. The main results of the research activities carried out in the work-package are:

- A proposal to increase the mass of the headform used in safety helmet impact tests and simultaneously reduce the peak linear acceleration threshold. In this way standard tests would be more relevant for real-life accidents.
- A proposal to use inner liners made of an ABS plastic lamina with deformable cones. A preliminary investigation suggests that the novel liner would be lighter than traditional ones with better ventilation properties and no reduction in safety.
- A proposal to make the inner liners with an assembly of traditional EPS foam and aluminum honeycomb. A preliminary investigation suggests that the novel liner could provide better protection

with liner stiffness and strength adapted to the local impact point requirements.

ACKNOWLEDGMENT

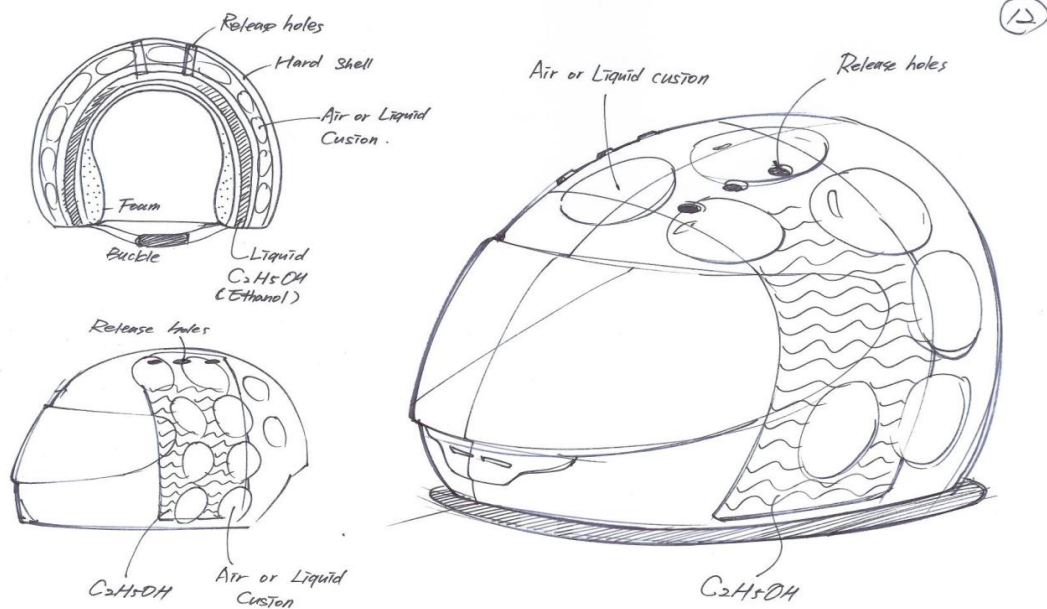
The authors want to acknowledge the financial support provided by the European Union through the Project MYMOSA, MRTN-CT-2006-035965.

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Helmet Performance and Design



The Influence of Velocity on the Performance Range of American Football Helmets

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ABSTRACT

Concussion has become a prevalent injury in the sport of American football. The nature of this injury can be influenced by the mass of the impactor, velocity, compliance, and direction of impact. As a result it is important to characterize how American football helmets perform against these impact characteristics. The purpose of this research is to examine how an American football helmet performs across velocities and impact angles which can occur in the sport of American football. The methods used a combination of Hybrid III headform impacts combined with a finite element modeling approach to find the brain deformation variables known to be associated with concussion. The results indicated that the American football helmets performed best at 5.5 and 7.5 m/s. At 9.5 m/s the brain deformation metrics showed a sharp increase in risk of concussion. Also, the region of the brain with the largest magnitude deformation shifted with differing velocities. The results indicate that current football helmet designs should expand the energy absorbing capacity of the shell and liner to accommodate these impact conditions.

Keywords: *American football; concussion; impact biomechanics; football standards; impact reconstruction*

NOMENCLATURE

m/s = meters/second

g = acceleration

rad/s² = radians/seconds squared

I. INTRODUCTION

The incidence of concussion has become an important topic in the world of contact sports. This in large part is due to research identifying multiple concussions having an additive effect over time leading to severe neurologic deficiencies later on in life [1]. Concussive injuries are characterized by symptoms ranging from headache to unconsciousness and amnesia which are represented by specific regions of brain tissue [2; 3; 4]. Research involving the causes of this injury has predominantly focused on diagnosing concussion and treatment options as well as methods to predict and therefore prevent the injury from occurring [5; 6; 7].

The brain is made up of several parts, with each part consisting of unique material properties and most likely injury thresholds [8; 9; 10]. This in part contributes to the various symptoms associated with concussion, as deformations which would injure one part may not affect another region of the brain tissue. A possible source of this variability in symptomology involves the types of loading curves generated from the impact, where the loading in response from the impact characteristics would affect one region of the brain and not another. In sport, impacts to the head are common and often result in concussive injury [11; 12]. These impacts are commonly quantified in terms of impacting mass, velocity and compliance of the impacting system. The characteristics of the resulting acceleration loading curves from these impacts are associated with changes in these three conditions. The influence of increasing velocity and mass has been documented to increase the magnitude of the dynamic response incurred from an impact [13; 14]. In addition to these independent

variables, the location and vector of the impact can also influence the dynamic response characteristics and resulting brain tissue deformations [15; 16]. Previous studies have shown that impacting ice hockey and American football helmets using a centric and non-centric impacting protocol produces significantly different linear and rotational acceleration loading curves [13; 16; 17; 18]. These changes in dynamic response are largely a result of how the head responds to impacts that are either through the centre of gravity (centric), or outside of it (non-centric). Quantifying how these independent variables contribute to the creation of large magnitude stresses and strains in regions of the brain tissue may have significance in establishing strategies for the reduction of concussion in sport.

Currently, the certification standards for American football helmets require testing at three different velocities and measure performance using linear acceleration, which is a metric best associated with traumatic brain injury. How the helmets perform across a range of velocities for impacts more suited to the sport of American football has yet to be elucidated. It is not known how velocity can influence the regions of strain distribution in the brain during an impact. A more complete understanding of velocity influences regions of brain strain may provide a more effective strategy for developing safer environments in sport through better helmet design. This study is an extension of previous work by the authors examining the evaluation and methodology surrounding the performance characteristics of American football helmets [17; 18]. The purpose of this research is to examine how American football helmet and brain deformation linked to concussion are affected by increasing velocity in a centric and non-centric impact condition.

II. METHODOLOGY

A. Test apparatus

A pneumatic linear impactor system was used to impact a Hybrid III headform fitted with a commercially available American football helmet to produce loading curves in the x, y and z axes used for finite element model simulations. The three-dimensional dynamic response of the Hybrid III head form was used as input for a finite element model of the human brain in order to predict theoretical brain tissue deformations associated with these types of impacts.

The linear impactor consisted of an impactor arm, piston and air tank. The impactor arm (length 1.28 ± 0.01 m; mass 13.1 ± 0.1 kg) and the hemispherical nylon

striker cap (diameter 0.132 ± 0.001 m; mass 0.677 ± 0.001 kg) with a vinyl nitrile 602 layer (thickness 0.0357 ± 0.0001 m) were propelled forwards using compressed air at velocities of 5.5, 7.5 and 9.5 m/s. These velocities were chosen to elicit the relationship between velocity and brain tissue deformation within the context of velocities encountered during American Football player collisions [19].

A 50th percentile adult male Hybrid III head (mass 4.54 ± 0.01 kg) and neckform (mass 1.54 ± 0.01 kg) were attached to a sliding table (mass 12.78 ± 0.01 kg) with an adjustable and lockable base that allowed the helmeted head to remain fixed in position throughout impact testing.

The headform was instrumented according to Padgaonkar et al.'s [20] orthogonal 3-2-2-2 accelerometer array that permitted the measurement of three-dimensional linear and rotational accelerations. The nine single-axis Endevco 7264C-2KTZ-2-300 accelerometers were sampled at a rate of 20 kHz and a 1000 Hz 2nd order lowpass butterworth filter was applied to the signals. The data was collected using Diversified Technical Systems TDAS Pro Lab system and TDAS software. Impact velocities were measured using an electronic time gate (width 0.2525 ± 0.0001 m) and were recorded using National Instruments VI-Logger software.

A commercially available model of an American Football helmet was used for impact testing. The helmet weighed 1991 g and was composed of a polycarbonate shell with a vinyl nitrile liner. The helmet was secured onto the Hybrid III headform as per the manufacturer's instructions and was checked between impacts to ensure proper positioning.

B. Test procedure

Two impact conditions were used to impact the football helmet (Fig. 1). One helmet was used for each impact velocity and was impacted three times per condition. There were 18 impacts in total. The impact location and angle for each condition were aligned on the headform using a laser that was mounted on the end of the impact arm prior to the attachment of the striker cap. The laser pointed to marked impact locations on the headform before the placement of the helmet to ensure accuracy and precision of impact conditions. The resulting acceleration time histories in the x, y, and z axes produced from the helmeted head form impacts were used as input into the University College Brain Trauma Model (UCDBTM) for prediction of brain tissue stress and strain.

C. Finite element model

A finite element model of the human brain was used to predict the brain tissue deformations associated with the helmeted impacts. The model used in this study was the University College Dublin Brain Trauma Model (UCDBTM) developed at the University College Dublin and was composed of the scalp, skull, pia, falx, tentorium, CSF, grey and white matter, cerebellum and brain stem [21; 22]. The geometry of the brain model was derived from CT and MRI scans of a male cadaver and the material properties are based on cadaveric anatomical research [23; 24; 25; 26; 27]. This finite element model was composed of approximately 26, 000 hexahedral elements [21; 22].

The behavior of the brain tissue was represented with a linear viscoelastic model with a large deformation theory. The brain tissue was viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus [21]. The brain was represented as elastic in nature for compression. The viscoelastic behaviour representing the shear characteristics was modeled using the following equation:

$$G(t) = G_{\infty} + (G_0 - G_{\infty})e^{-\beta t} \quad (1)$$

where G_{∞} represents the long term shear modulus, G_0 the short term shear modulus and β is the decay factor [21; 25; 28].

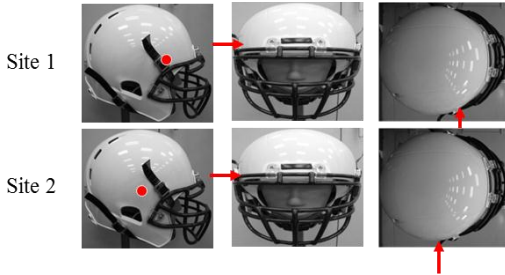


Figure 1: Impact sites

To simulate a sliding boundary condition the CSF was modeled using solid elements with a low shear modulus and a high bulk modulus [23; 25; 29; 30; 31; 32]. There was no separation for the contact interaction and the coefficient of friction was set to 0.2 [33]. The model was validated through comparisons with Nahum et al.'s [34] cadaver impact response for the cerebrum measuring intracranial pressure and Hardy et al.'s [35] relative brain and skull motion data. Further validation was done by comparing simulations to real world traumatic brain injury reconstructions [36].

The finite element model was segmented into regions relating to neurologic symptoms associated with concussive injury [37]. These functional areas chosen for

analysis only represent some of the possible regions of the brain for analysis and are not intended to be inclusive of all regions. The cerebrum was chosen for analysis because it remains the only area that has been validated using cadaver research [34; 35].

The prefrontal cortex is primarily involved in moderating social behaviours [38] whereas the dorsolateral prefrontal area regulates motor behaviour planning and memory. Complex motor actions are controlled by the motor association cortex and the primary motor cortex is involved in planning and execution of these movements. The primary somatosensory cortex is responsible for touch and proprioception. Complex sensory integration and perception of the external environment are controlled in the sensory and visual association areas [39]. Processing of visual and sound information is done in the visual cortex and auditory cortex respectively [40; 41]. Understanding and recognizing sounds is done in the auditory association area.

The brain deformation metrics used to evaluate the helmeted impacts were maximum principle strain. This measure is commonly used in finite element model research reflecting risk of concussion. All results were analyzed by ANOVA using SPSS software.

III. RESULTS

The peak dynamic response parameters used as input to the UCDBTM are found in Table I. The brain deformation output of the UCDBTM can be found in Table II. The dynamic response was similar between the two impact sites when examining linear acceleration with significant differences between the measures at 5.5 m/s ($p < 0.05$). The rotational acceleration response was higher for impact site 2 for the 5.5 and 7.5 m/s velocities ($p < 0.05$) (Table I).

A. Regional brain deformation using maximum principal strain

When examining the regions of brain strain using maximum principal strain (MPS), the highest magnitude was found at the visual association area, sensory association area, and primary somatosensory cortex at 5.5 m/s for site 1 ($p < 0.05$) (Table II). While not significant from the other brain regions except the visual cortex ($p < 0.05$), the largest peak magnitudes brain deformations shift to the dorsolateral prefrontal area at 7.5 m/s. The primary somatosensory cortex and sensory association area had the largest magnitude strains for 9.5 m/s ($p < 0.05$). At site 2, the largest magnitudes MPS are found at the dorsolateral prefrontal area, auditory cortex,

primary motor cortex, and the sensory association area for the 5.5 m/s impact condition ($p < 0.05$). At the 7.5 m/s impact condition the largest magnitudes are at the dorsolateral prefrontal area, primary motor cortex, auditory cortex, primary somatosensory cortex, and the sensory association area. The peak magnitude MPS was found only in the dorsolateral prefrontal area and primary motor cortex at 9.5 m/s ($p < 0.05$).

TABLE I. DYNAMIC RESPONSE

Site	Velocity (m/s)	Peak Acceleration	
		Linear (g)	Rotational (rad/s ²)
Site 1	5.5	55.3 (1.3)	3700 (87.71)
	7.5	78.8 (1.3)	4317 (244.7)
	9.5	119.1 (0.85)	7775 (207.2)
Site 2	5.5	61.9 (0.76)	4635 (134.1)
	7.5	80.0 (1.3)	4856 (145.6)
	9.5	117.9 (1.87)	8347 (369.0)

IV. DISCUSSION

This study was conducted to examine how impact velocity contributed to the location of peak magnitude brain deformations using two distinct centric and non-centric impact locations. The impact velocities were chosen to represent velocities of impact which would be similar to those experienced in American football and the sites were chosen to represent two possible mechanisms of injury, one through the centre of gravity (centric) and the other outside of the centre of gravity (non-centric).

The linear acceleration response of the football helmet increased with velocity for both impact sites which would be expected with an increase in the energy of the impact. The rotational acceleration response at both sites were similar at the 5.5 and 7.5 m/s impact conditions, but increased by approximately 40% for the 9.5 m/s impact condition. This phenomenon is likely indicative of the linear impactor arm coupling with the helmet shell more effectively at this velocity as well as the helmet materials reaching the end of their functional ranges which forces a larger increase in magnitude for rotation. The increased magnitude for rotation then contributed to larger brain deformations incurred as reflected in the MPS values at this velocity (Table II). When comparing the results to risk of injury literature

[6; 7], the non-centric site (site 1) showed an increase in damaging brain deformation at 7.5 m/s where site 2 (centric site) only started showing damaging strains at 9.5 m/s. This indicates that there is an interaction between site and velocity where future helmet designs may need to account for the increased risk posed by non-centric impacts.

A. Regional brain deformation

When examining the results for regions of brain deformation and how they shift across different velocity conditions interesting relationships result. When examining impact site 1, the region of largest magnitude for MPS shifts from the back of the brain (visual association area) at 5.5 m/s anteriorly to a more general strain field for 7.5 m/s. At 9.5 m/s the region of largest MPS shifts further towards the centre of the brain as represented by the primary somatosensory cortex and the sensory association area. At impact site 2, the 5.5 m/s impact condition produces large magnitude values of MPS at a very central part of the brain and stays very central across the 7.5 and 9.5 m/s velocities. When comparing the magnitude of MPS response at the varying velocities, site 1 produces larger strains at 5.5 m/s. While not significant for the 7.5 m/s and significant for 9.5 m/s ($p < 0.05$) site 1 shows considerably larger strain values. The differences between these two sites in regions of maximum principal strain are likely a result of the differing loading curve inputs generated from the different impact sites and their interactions with the different grey and white matter proportions represented in each distinct region of the brain used in this study. These results indicate that changing the impact velocity can not only influence the magnitude of the resulting brain deformation but also the region in which it incurred.

V. CONCLUSION

This study investigated the influence of impact velocity on brain deformation in different regions of the brain associated with concussive symptomology. The results demonstrated that impact velocity does influence the location in which the peak tissue deformation occurs. It was demonstrated that this relationship is also dependent on the location of the impact. These results indicate that the velocity at which an impact occurs may determine the region of brain tissue which incurs damaging deformations. The results showed that the American football helmet performed well up to 7.5 m/s but had a drop in performance at 9.5 m/s as shown by increases in rotational acceleration and brain deformations. This suggests that it may be prudent to

develop helmet technologies which can accommodate a wider range of impact velocities as 9.5 m/s is a velocity which is frequently experienced in the sport of American football [19].

A. Limitations

This study is limited to the equipment used to evaluate both the dynamic and tissue responses of the head and brain during impact testing. The Hybrid III head- and neckform are composed of a combination of steel and rubber used to approximate the geometry of a male adult head and neck however they may not be representative of a real life impact response that would include the compliant nature of these tissues. These anthropometric test devices were primarily designed for antero-posterior impacts and as such would likely produce a very stiff response in the other planes. In particular the neck was not designed for rotation and will likely influence the rotational acceleration results. The finite element model in this research imposes limitations of the methodology used to evaluate the tissue response of the brain to these impacts. The material characteristics and parameters defined in the UCDBTM govern the tissue responses to impact. It is acknowledged that the peak brain deformation results obtained in this research is specific to the model used and would produce different values as compared to another model. However, since this model was used to evaluate the complete data set, the values provide a means to compare the helmet impact conditions under the same model parameters.

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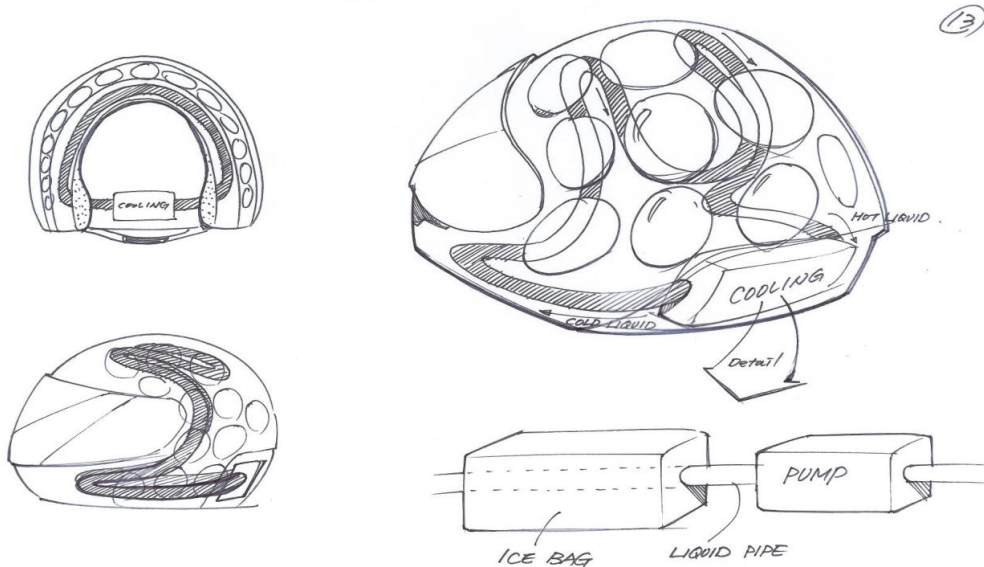
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TABLE II. Brain deformation results

		Maximum Principal Strain						
		Site 1				Site 2		
Velocity (m/s)		5.5	7.5	9.5		5.5	7.5	9.5
Prefrontal Cortex		0.121 (0.005)	0.186 (0.026)	0.256 (0.002)		0.114 (0.005)	0.117 (0.011)	0.206 (0.003)
Dorsolateral Prefrontal Area		0.146 (0.002)	0.239 (0.028)	0.310 (0.003)		0.165 (0.002)	0.176 (0.005)	0.322 (0.011)
Motor Association Cortex		0.137 (0.002)	0.221 (0.022)	0.280 (0.006)		0.148 (0.002)	0.149 (0.007)	0.249 (0.007)
Primary Motor Cortex		0.146 (0.002)	0.239 (0.028)	0.310 (0.003)		0.166 (0.002)	0.176 (0.005)	0.322 (0.011)
Primary Somatosensory Cortex		0.149 (0.006)	0.209 (0.022)	0.328 (0.003)		0.163 (0.002)	0.171 (0.007)	0.288 (0.008)
Sensory Association Area		0.149 (0.006)	0.200 (0.029)	0.328 (0.003)		0.164 (0.001)	0.171 (0.007)	0.288 (0.008)
Auditory Cortex		0.142 (0.003)	0.203 (0.022)	0.276 (0.002)		0.172 (0.001)	0.175 (0.003)	0.286 (0.006)
Visual Association Area		0.158 (0.002)	0.200 (0.018)	0.274 (0.007)		0.148 (0.007)	0.137 (0.014)	0.254 (0.007)
Visual Cortex		0.085 (0.005)	0.110 (0.008)	0.133 (0.002)		0.084 (0.003)	0.077 (0.013)	0.153 (0.004)

Helmet Performance and Design



Efficiency of Head Protection Equipment for Two Mainstream Sports – A Comparison

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ABSTRACT

The fastest growing participant sport in US College's and High Schools is 'soccer for girls'. Mindful of a large potential market FIFA issued Circular #863 in August 2003 allowing the use of soft headgear. This was the first major equipment rule change in decades. The change was brought about largely by social pressure to allow soft headgear to mitigate or ameliorate possible head injuries amongst players.

Simultaneously generations of amateur boxers have been using soft protective headgear whilst training and in competition with only minimal evidence for their effectiveness.

The work reported here used a magnesium head form to assess the effectiveness of football specific head guards, and a typical regulation boxing helmet.

To simulate the act of heading the ball a FIFA regulation ball was displaced towards the magnesium head form and during the contact the resultant accelerations of the head form were measured. Transmitted forces were also measured through the head form imparted by the football at pre-determined velocities.

These accelerations and forces were compared against the accelerations and transmitted forces that occur in an amateur boxing match using regulation equipment.

From a series of tests the average peak accelerations for a football impact were found to be $52.2g_n$ (Standard deviation (SD) 8.8). Remarkably the average accelerations were higher than those experienced for boxing at $44.8g_n$ (SD 13.2). Conversely the maximum acceleration recorded for Football was $73.9g_n$ which was less than the peak acceleration found in Boxing of $81.4g_n$.

The average of peak transmitted forces for amateur players seem to be similar in the range of 2000 N for both sports. Commercially available football helmets did little to reduce peak acceleration measurements recorded by the head form from the footballs impact. Two of the guards tested increased peak accelerations over the unguarded case. It was found that for football players the air pressure in the football had a significant effect on peak acceleration.

Keywords: *boxing, helmet, acceleration, force, football.*

NOMENCLATURE

Velocity	m/s
Force :	N
Acceleration :	g_n ($1x g_n = 9.81 \text{ m/s}$)

I. INTRODUCTION AND LITERATURE REVIEW

Football is the world's most popular team sport with over 250 Million registered players. In the UK the youngest professional domestic appearance was by a youth aged 16 years and 126 days. The fastest recorded shot is 87 mph (39 m/s). Today only a hand full of players wear a helmet or soft head protector, Cech for Chelsea and Chivu for Inter Milan. Both wear helmets although it is claimed that this is due to previous head trauma.

Whether repeated concussive or sub concussive blows cause permanent or cumulative brain injury is a complex and controversial question. Press coverage highlighted the case of Jeff Astle, a former England international football player, where the coroner ruled the cause of his death as an "industrial disease" - suggesting that repeated heading of balls during his professional career was the cause of his subsequent neurological decline [1]. This case was at odds with that of Billy MacPhail, a former

Glasgow Celtic player, who in 1998 lost a legal battle to claim benefits for dementia that he said was due to heading the old style leather footballs. Concern has been raised as to whether heading a football may be the basis for injury and cognitive impairment. In the United States this has led to calls for the use of protective headgear for soccer players, especially youngsters whose skeletal growth has not necessarily completed. The FA is conducting a 10-year prospective, closely-monitored research study, funded jointly with The Professional Footballers' Association (PFA), on the effect of playing football on brain structure and function.

In response to a BMJ article “Coroner cites football as reason for brain injury”, B. Minser [2] claimed “what happens in football when heading a leather ball occurs with greater force in the boxing ring”, this maybe the case for professional sports, but is this the case in the amateur game?

“A BMA report on boxing identified two main causes of structural damage to the brain attributed to boxing”[3]. These were cumulative effects of sustained exposure to the sport and acute effects as the direct result of a severe blow. [4]

The forces and accelerations for Olympic and professional athletes have been studied for boxing by J. Atha *et al.* [4]. The authors required Frank Bruno to punch an instrumented target mass to measure the physical properties of a punch, and T. Walilko *et al.* utilised a Hybrid III crash test dummy to record face punches of Olympic boxers [5], but less work has been done at a club or amateur level, with less trained and lower skilled participants. Both the work by J. Atha *et al.*

and T. Walilko *et al.* did not use a boxing helmet but a padded ballistic pendulum and soft tissue on a Hybrid III dummy respectively. Helmets are mandatory in the amateur and Olympic game and so have been used for this study.

Hybrid III dummies were also used for evaluation of a single brand soccer head protector by T. Smith [6], but this primarily reviewed head to head and head to post impacts which can occur from time to time during a football match.

A primary objective of this work is to review the effectiveness of the soft protectors that are currently in the market place, in conjunction with modern equipment that the club and amateur sports person may typically use. All impacts generated in these tests were intended to be typical of those that could be achieved in their respective sports by amateur club players. The data collected is compared for both average peak force, a measure of the effect of repeated impacts, and the maximum peak force measured during all the tests, equivalent to the acute and direct severe blow.

II. METHODS

To measure the force of impact from the football a regulation FIFA football was fired into the head form from the servo controlled hydraulic test machine. The head form was simply supported on a hinge to represent the neck and normal force measured with a load cell. To measure the input force for boxing a force plate was allowed to move horizontally and resisted by a spring; it was felt that the boxers could not punch a rigid body as they may damage their hands even while wearing a glove.



Figure 1: A picture of the head guards, from left to right: Kangaroo, Headblast, Full 90 Select (on headform) Full 90 Premier, Full 90 Club.

Accelerations induced in a simply supported head form were measured for both sports. The tests were conducted on the head form which was simply supported while it was impacted by a football kicked by amateur players, or punched by a boxer. In the case of the football game the head form was placed on a mat on the floor and again simply supported while the amateur players took shots with the ball. In the case of the boxers the head form was suspended as per a punch bag and then held in place by the coach. Boxers of different weight categories punched the regulation helmet attired head form using a regulation glove. An additional accelerometer was placed in the gloved hand of the boxer and the accelerations recorded.

A. Apparatus

Headform : The headform was 4 axis CNC machined from a billet of magnesium alloy as per the NIJ standard. [7], size 7¼, with a mass of 4.7 kg. Accelerometers were mounted in the X, Y, Z plane at the biofidelic center of gravity (COG), with positive X as back, Y left from front, Z up. This is a very similar setup to helmet test methods BS EN960 [8], but the head form included a face and chin. Magnesium Electron K1A grade was used as it has a higher damping coefficient.

Data Collection: Data was collected on a Compaq DAQ and NI 9233, sampled at 50 kHz. The piezo accelerometers were calibrated up to 250 g_n , and a force link piezo electric 224C PCB loadcell was used for force measurement calibrated up to 100 kN. The raw signals were recorded along with filtered signals using a CFC1000 4th order Butterworth anti-aliasing filter. The resultant vector sum of the acceleration was calculated using the method described in the BS EN960 helmet standard [8].

A fourth accelerometer was mounted on a small plate and wrapped in the boxers glove and aligned with the punch direction. This was again linked to the NI 9233 via a long BNC cable taped to the boxers arm and shoulder. The data recorder was set to permanently read data but only trigger and record above a threshold of 20 g_n , with 30% pre-trigger, this alleviated the need for an external trigger.

B. Equipment

1) Football

A new FIFA regulation ball was used for the tests, size 5. Care was taken to accurately measure the ball pressure, before and after tests.

Soft Helmet protectors for Football: For football the following commercially available protectors were utilised *Full 90*, *Kangaroo Soccer Headgear*, and *Head Blast* as shown in

Figure 1. The *Full 90* come in two types, a pro and standard version, the former was used. These were typically 6-8 mm head moulded foam produced in planar form which then wrapped into shape when placed on the head form. Hook and loop fastening allowed for accurate placement.

The *Kangaroo* guard is a much more significant 3 dimensional design, similar in construction to a martial arts soft helmet. The forehead protection is made up of two different density foams and the helmet is encapsulated by a protective layer of plastic. This build up results in a maximum thickness of 24 mm.

Head Blast is similar in design to a head band. It's typically about 4-5mm thick and has a plastic outer covering.

2) Boxing.

New regulation 18 oz Lonsdale gloves were used for all tests. Two regulation Lonsdale approved Helmets were tested, these were both used in tests, and were swapped frequently to allow for any medium term recovery.

C. Test Conditions

All tests were conducted indoors in a temperature controlled laboratory between 20-22°C and 40-55% relative humidity. All samples were maintained at those test conditions for at least 24 hours prior to testing.

III. BOXING

A. Force measurement plate for Boxing.

A padded impact plate was connected to a ballistic pendulum with a piezo electric load cell. The pendulum was resisted by a lightweight titanium spring of stiffness 47 kN/m.

This setup has been used for previous biofidelic tests to mimic the pelvic stiffness in a hip protector test by Robinovich *et al.* [9]. The assembly was mounted in such a way that the height could be adjusted to suit the size of the boxer to simulate a head punch.

The impact plate was covered by a 12 mm layer of soft foam Recticel RS 55H, to represent the soft tissue and to ensure that the boxers would not damage their fists.

Four amateur players were invited to attend the test sessions, representing four different weight categories. Bantamweight (BW), Lightweight (LW), Middleweight (MW) and Heavyweight (HW). They were accompanied by a coach and a medical practitioner was present for all tests. The boxers were keen to try to develop a force plate system that could give direct feedback on their technique and volunteered for the tests.

The boxers were encouraged to warm up and start slowly mindful of injury or pain before successive tests were logged. Tests were conducted with and without the hand accelerometer measurement. The boxers were instructed to punch the headform with a number of different styles of punch: jab, left, right, hook, the force traces were recorded, and the peak forces found.

They were also encouraged to attempt combination style punches that could typically occur in a match. The boxers continued to punch the force plate as would be typical in the ring, and did not stop between impacts, so measurements would be taken as a snapshot of time in the ring. The average and peak impact forces can be seen in Figure 2. These results cover all the different types of punch. These forces were also compared with those recorded by novices with no training where average forces are generally about 30% lower than the peak maximum forces. It is uncertain if this spread would happen in an actual match as the boxers seemed to become more accustomed to punching the rig with time. Paradoxically the experiments may have actually acted as a training aid.

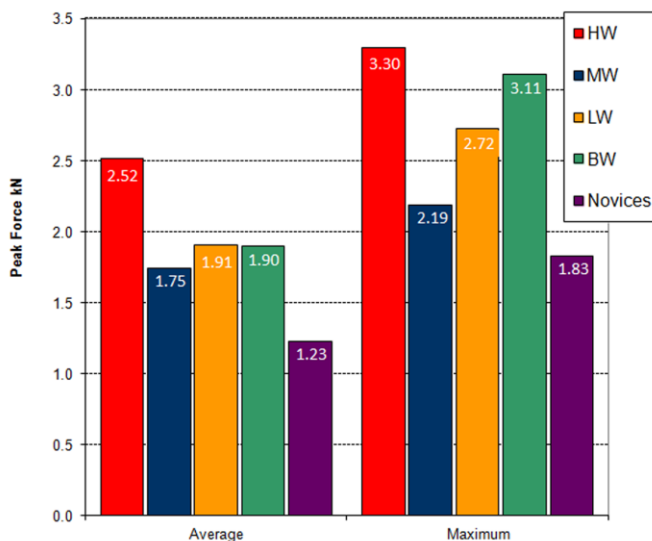


Figure 2: Average and maximum punch severity for four boxers against novices - force generated

The forces measured from Frank Bruno by J. Atha *et al.* [4] were considerably higher than those recorded here at 6320 N. The maximum recorded in these tests was only 52% in comparison. This clearly differentiates the world heavy weight from a lighter weight amateur. Our heavy weight boxer average force was 2.52 kN which again is lower than that found in the experiments performed by T. Walilko *et al.* [5] which was 4.3 kN. The average forces in the light weight class was closer to that reported by Walilko *et al.* where a hybrid III dummy was used. If we exclude the heavy weight average then the average of all other punches was below 2 kN.

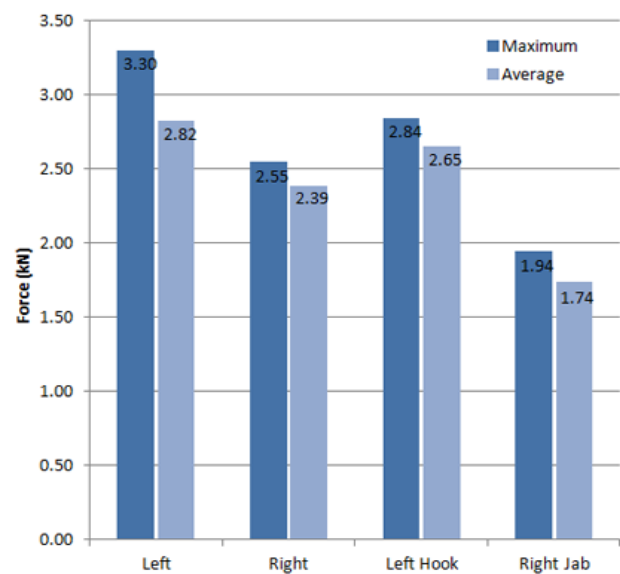


Figure 3: Average and maximum punch severity force generated

Different types of punch were also tested on the force plate; the results in Figure 3 are shown for the heavy weight. Amongst the amateur boxers there seemed to be no correlation between the type of punch: left, right, hook (preferential hand) jab (other hand) and the maximum force recorded. Two out of the four boxers acquired their highest forces with a straight punch, the other two with a hook, so it would be difficult to use this data to help in the zoned design of the boxing helmet. The technique of the amateur boxer seems to have more effect on the force, and our middleweight could consistently deliver average punches within 85% of his maximum, as opposed to the 75% found for the group as a whole.

B. Boxing Tests Headform acceleration

The head form was suspended on a tether and fitted with a regulation boxing helmet, the height of which was

adjusted to the same height as the boxer. The coach was restraining the helmet to reduce excessive movement of the head form in a similar fashion to a training punch bag. Two identical helmets were used and changed when the boxers changed.

The head form was returned to rest after each combination of punches by the coach. The accelerations were recorded and the vector average calculated as per BS EN960 [8]. The peaks were found and the average and the maximum values were recorded as shown in Figure 2. This data set is consistent with the work of Waliko *et al.* [5] conducted on a Hybrid III where average accelerations ranged from $44g_n$ to $71g_n$. Peak acceleration from the Hybrid III was recorded at $78g_n$. Our data set gave $36g_n$ to $57g_n$ for the averages, and a peak of $81g_n$. A good correlation between the magnesium headform as used in hard helmet standard tests and the Hybrid III dummy for this type of test was therefore found.

The average and peak accelerations are higher than those obtained by J Atha *et al.* [4] using the ballistic mass at $53g_n$. This difference is attributed to the different mass of the head form, 4.7 kg vs. 7 kg, as well as the restraint of the ballistic mass in a single linear direction as used by J. Atha *et al.*

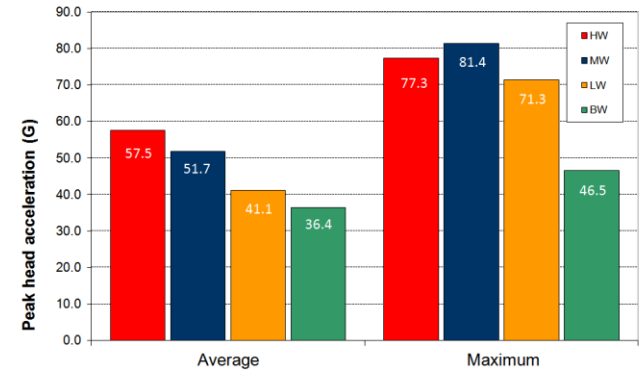


Figure 2: Average and maximum punch severity for four boxers - acceleration of headform

A typical acceleration data and force data are shown in Figure 3 for a boxer's punch.

The peak negative acceleration of the boxers hand in the glove is not at the same point in time as the peak acceleration of the head form; this suggests that the compliant surface of the glove and helmet are decoupling and dampening the impact. The X axis acceleration of the head is the highest of all the traces, but Z is higher than Y, showing that this impact was slightly off the X (front/back) axis, and slightly upwards.

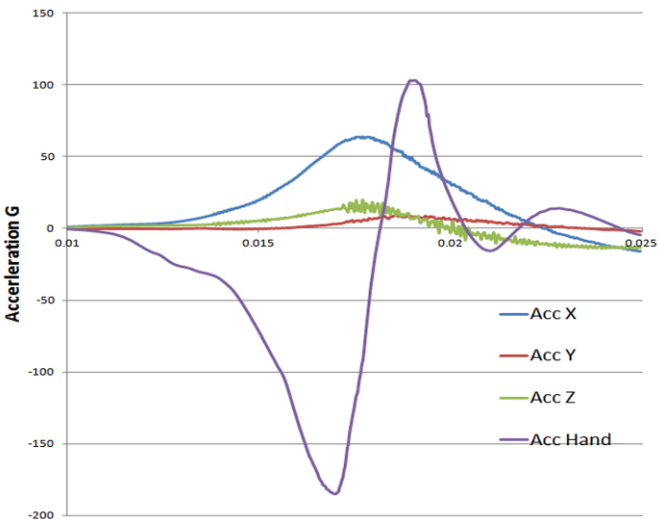


Figure 3: Acceleration data for headform and glove.

There are small differences between the X axis acceleration and the resultant vector acceleration. The Walilko study only denotes the Peak 2D in line (X axis) acceleration. Only 2/81 tests gave a peak force of over $78g_n$, the limit that has been suggested can cause concussion [6]. The length and rise time of the traces are longer than those typical in rigid helmet testing for Motorcycling and Skiing possibly because of the compliant glove and helmet.

IV. FOOTBALL

A. Football. Force Measurement

A regulation FIFA ball was fired using an Instron servo hydraulic test machine. The football was retained in 'the claw' prior to being fired by the movement of the ram. A displacement of 400 mm was used for all tests. Figure 6 shows the head form was mounted on a simple hinge to represent the neck and the transmitted loads were measured on the piezoelectric load cell.

The neck and head could be moved to change the angle of the head, but the load cell always measures the force normal to the direction of impact of the ball, as show in Figure 4.

The tests could be conducted with and without helmets. According to Withnall *et al.* [11] speed of a football is between 26.8 to 53.6 m/s have been recorded in games. Typical heading speed is about 18 m/s.

The ball pressure was measured before and after the tests. The impact events were also recorded on a high speed camera. Table 1 show the peak force measurements averaged from 5 tests.

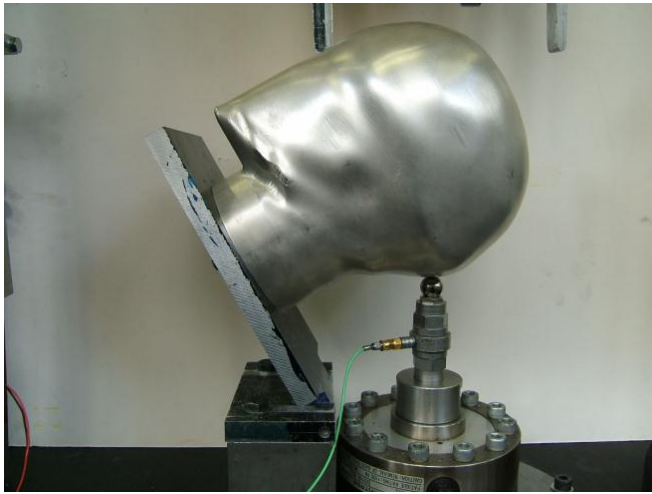


Figure 4: A view of the angle plate, head form, and load cell assembly.

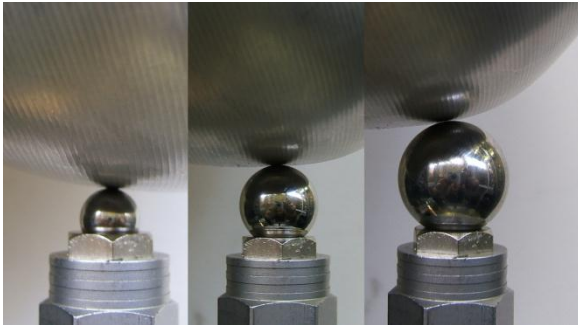


Figure 5: Pictures of the three ball bearings, situated in the top of the load cell, used to provide elevation of the headform and a different angle of impact.

TABLE 1 MAXIMUM FORCE TRANSMITTED
AVERAGED OVER FIVE TESTS

Head guard	Instron Speed (m/s)	Video Speed (m/s)	Impact Duration (ms)	Max. Force (kN)
No guard	8.9	10.036	13.012	1.32
Kangaroo	8.9	10.248	16.984	1.296
Headblast	8.9	10.185	16.108	1.335
F90 Prem	8.9	10.178	15.057	1.288
No guard	15	15.17	10.084	2.262
Kangaroo	15	15.15	14.894	2.043
Headblast	15	15.14	12.481	2.21
F90 Prem	15	15.162	10.22	1.998
No guard	25	18.047	10.644	2.824
Kangaroo	25	18.076	13.884	2.649
Headblast	25	18.107	10.551	2.701
F90 Prem	25	18.059	10.84	2.656

As the input speed of the Instron was increased the actual speed measured was recorded and verified by high speed camera. Thereafter tests were conducted at target delivery speeds of 10 m/s, 15 m/s and 18 m/s respectively. *Headblast* showed very little change in maximum transmitted force in comparison with the no guard situation. *Kangaroo* and *Full 90 Premier* showed a small change at 10 m/s and 18 m/s. However at 15 m/s the *Kangaroo* gave a 10% reduction in peak transmitted force and the *Full 90 Premier* showed a 13% reduction. Although relatively low speeds were tested, these forces are similar to those measured in boxing by the force plate.

At 15 m/s the unprotected head form transmitted 2.26kN. This is significantly lower than the findings of Broglio *et al.*, where values of 3.1 kN for an unprotected head were reported. Nearly 1 kN lower, Broglio suggests that their readings are higher than expected. The Levendusky *et al.* [12] study gives comparable force figures. The reductions in peak forces, although centered around 15 m/s are surprising as the deformation in the ball is large. Figure 8 shows the maximum deformations of the ball at 15 m/s.



Figure 6: High Speed Video still of peak deflection.

B. Acceleration Football.

A regulation football was kicked at the instrumented magnesium head form. As per the boxing experiments this was simply supported and kept from excessive movement by the coach. 21 different tests were conducted with a range of players from the 1st XI University team to one novice. This provided a range of

skills and ball velocity typical of an amateur match. The tests thereby provided impacts similar to the boxing experiment with amateur boxers.

Tests were repeated three times for each helmet at regulation ball pressure of 0.8 bar (11.6psi). The unprotected head form was then tested with the ball at inflation pressures of 0.689bar (10 psi), 0.965 bar (14 psi), and 1.241 bar (18 psi). The peak accelerations are shown in Table 2. The ball pressure was found to have a significant effect. Lower ball pressure reduced the peak accelerations recorded in the headform. This is a similar result to that reported by Naunheim [11].

TABLE 2 PEAK VECTOR ACCELERATION G AND AVERAGE

	T1	T2	T3	Average
18psi	59.74	55.94	55.59	57.09
14psi	52.35	54.87	49.22	52.14
11.6 psi	51.61	56.37	51.70	53.23
10 psi	46.83	36.28	39.37	40.83
Kangaroo	60.24	55.50	73.94	63.22
Head Blast		60.24		60.24
Full 90	52.01	59.26	50.53	53.93
Novice	47.30	64.47	38.54	50.10

The regulation pressure was used for the helmet tests. Little effect in reducing peak or average accelerations for the guards was noted. Indeed two of the guards recorded a higher acceleration in the head form. There is a ceiling on the performance benefit of the soft helmet in football. The footage from the high speed video shows the large deformation of the ball which limits how much effect a thin compliant helmet can have on a ball that deforms to this extent. The large deformations minimise the benefit for head to ball impacts.

The protective helmets show no reduction in the peak accelerations in these acceleration tests. There is no soft tissue on our headform, so these results are not directly comparable with the Hybrid III dummy tests although similar results have been reported by Withnall *et al.* [10]. In some cases the resultant peak acceleration is higher with the helmets than without.

V. DISCUSSION

In this study new data for head injury risks has been recorded by measuring accelerations from the response of a head form and a force pressure plate for punches and ball strike. Two factors differentiate this work from existing literature.

The first was that in this study amateur players were used to give a better representation of what occurs in

these amateur sports. The second was the use a magnesium head form as is typical in CE helmet certification. Concerns have been expressed that this does not fully represent the biofidelic model of the head and neck in a similar way to a Hybrid III dummy or indeed that of a human. In reality both setups have been developed for different styles of tests. The Hybrid III was originally developed in 1976 as a forwards facing automotive crash test dummy, whereas the magnesium headform has been developed for CE certification of hard motorcycle and automotive racing helmets. These latter tests involve the impact of hard anvils at considerably higher forces and accelerations than are being used for these tests with soft helmets and soft impactors (in this case the football or glove). Additionally there was no provision for a neck and no soft tissue over our headform. This may have been of particular importance while measuring force transmitted with no helmet in football, however for the boxing tests the helmet was always used as it would be used in the amateur and Olympic game.

For boxing the results measured with the force plate ranged between 0.6-3.11 kN with an average of 1.98 kN and SD of 0.45 kN. There was no clear link between weight category and peak force at the amateur level with our small set of 4 boxers. For boxing amateurs, technique played a bigger part in peak force performance than muscle mass. This is somewhat unexpected and could be an artefact of the small sample size.

Only the Heavyweight boxer was able to repeatedly deliver above 2 kN peak force. It is of interest to note that the boxers stayed on late into the evening to continue using the force plate, as they could appreciate its use as a training aid. A different study by Walilko *et al.* [5] shows similar peak force development for the middle weight class of boxers. The forces are considerably lower than those reported by Atha *et al.* [4] for a world class heavyweight.

The forces recorded for football were conducted on the headform mounted on the servo hydraulic test machine. The primary difference between the results reported in this paper and other research is the use of a magnesium head form instead of a Hybrid III dummy. The forces transmitted through the headform from the ball ranged between 1.2 to 2.8 kN, with an average of 2.0kN and an SD of 0.63 kN. These forces are in a similar range to those reported by Levendusky *et al.* [12], but typically about 1000 N less than those found by Broglio *et al.* [11]. It should be noted that the Broglio paper reviews these results as being high and ascribes this to ball acceleration. This is a difficult argument to follow since in his experiments the ball has already left the

machine and so there should only be forces acting on the ball other than windage.

The peak transmitted forces imparted by a football are reduced by two of the guards, at moderate ball speed but not significantly at higher or lower ball speeds. These findings are not mirrored when we review the peak accelerations experienced in the head form by the football. It is considered that this may be a function of the lack of a biofidelic element representing the neck. In effect the headform was rigidly mounted.

The accelerations recorded during the course of this work with amateur boxers using a regulation helmet and gloves ranged from 25-81 g_n , with an average of 44 g_n and an SD of 13.2 g_n . Different studies completed on Hybrid III dummies show a very similar range of results especially for lighter weight boxers. Waliko *et al.* [5] reports average accelerations between 44 and 71 g_n . There seems to be a good correlation between the magnesium head form and the Hybrid III for these tests. Perhaps a simpler setup using a homogenous headform could be considered as the base standard for future research as the headform is considerably cheaper and more damage tolerant than the multipart dummy.

For football a range of accelerations between 36 to 74g with an average of 52.2 g_n with a SD of 8.8 g_n were recorded. These findings are similar to those reported by Withnall *et al.* [10]. The headform was simply supported for these acceleration tests and the peak values recorded are similar to those generated from a Hybrid III dummy. This suggests that a simpler CE type testing headform could be viable in future research.

Surprisingly two guards increased the peak acceleration experienced by the headform in comparison with the unguarded headform. It is hypothesised that this is because for a frontal impact the ball is more compliant than the helmet. The protective helmet actually causes the ball to deform further around the guard than would have occurred without the protector. This hypothesis is supported by the freeze frame, Figure 7 from the high speed camera record of the impact. This is taken part way through the impact, but essentially the ball is starting to deform around the guard, well before the guard starts to measurably compress. This suggests that to protect the player from a deformable striker is more difficult than expected. Previous research has also shown that the currently available football helmets give little benefit for ball to head contacts in terms of reducing peak forces and accelerations. [11]

It is therefore important that the designers of such devices understand the relative stiffness of the helmet

when compared to the ball and the significance of this difference.

This is the first study of this type where boxing and football have been directly compared in the same work using similar apparatus. Considering the individual findings it is possible to make a further comparison between the two sports. Table 3 contrasts the results found for the two sports.

TABLE 3 RESULTS FROM AMATEUR FOOTBALL AND BOXING

Sport	Force kN		Acceleration G	
	Football	Boxing	Football	Boxing
MIN	1.1	0.7	36.3	25.1
MAX	2.9	3.1	73.9	81.4
Average	2.0	2.0	52.2	44.8
SD	0.6	0.5	8.8	13.2

The average force in the amateur game for boxing and football are of similar magnitude. The maximum ball speed used in this study was 18 m/s, that postulated by Kirkendall *et al.* [14] to be the typical speed of the headed ball. The amateur boxers were not limited to a specific speed and were asked to deliver an unhindered punch that would be typical of that in a match. The maximum forces recorded are of similar magnitude to those found in football. Only our heavy weight boxer could deliver punches over 2.9 kN, the max transmitted force recorded in football.

When reviewing the induced accelerations the averages are similar although marginally higher for football. The range for boxing is larger and shows a maximum of 81 g_n as compared to 74 g_n for football. It is worth noting that for both boxing and football a helmet was being used for those tests with the highest peak accelerations.

It is not unsurprising that the forces and accelerations in these amateur games are so similar. In mechanical terms the delivery system has a finite amount of energy and is more compliant when compared to the headform. In football the amateur player can only impart so much energy into the ball with his or her body when kicking it, and the boxer would seem to be able to deliver a similar amount of energy from his body with a full punch.

The head guards and glove combinations used in boxing do reduce the peak forces and accelerations in amateur boxing to below 78 g_n . In only 2/81 tests was an

acceleration above $78 g_n$ recorded and this was recorded by our heavy weight. Anecdotaly this is why we see very few knockouts in amateur matches from single punches when regulation gloves and helmets are worn. Combination punches and rotational effects are not reported or measured in this paper.

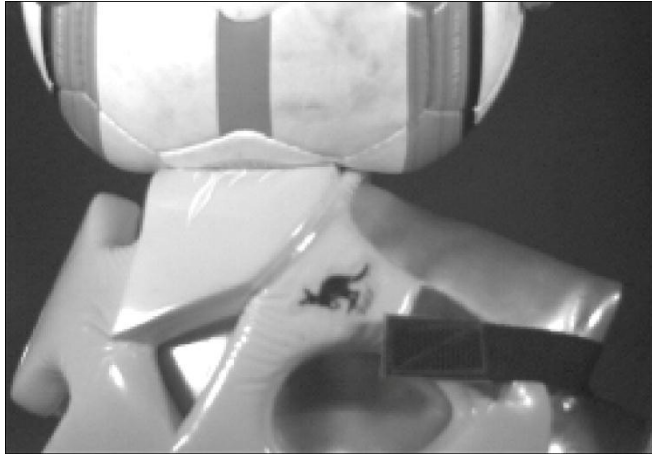


Figure 7: High Speed Camera Freeze Frame

VI. CONCLUSIONS

Acceleration and transmitted forces experienced by amateur players in both boxing and football have been considered. The results have been compared to existing research in this field. For boxing much of the work existing in the literature has been completed at the professional level but the results and values have been compared where appropriate. The differences in the apparatus used in these tests and methods used have been reviewed. There is a good correlation in results found in the open literature when using a dummy and this research when using a homogeneous head form.

The products that exist on the market place today are currently shown to be effective for boxers and the regulations for the amateur game call for standardized gloves and helmets for both amateur and Olympic matches.

Boxer technique seemed to be the biggest influence on performance as measured by peak force or induced acceleration. Indeed the equipment used for this study provided an insight into how the technique of boxers could be improved. However, there is room for an improved product that could reduce peak accelerations further, and these could be easily adopted by the regulations of the game.

The football head guard effectiveness offers little in way of reassurance for footballers. There is no significant reduction in acceleration for the head guards tested in

comparison with an unprotected head. It should be noted that the head guard manufacturers often state that the primary function of their head guard is to protect players from head-to-head collisions and head to post collisions.

There are no head guards on the market at this time that will significantly reduce forces and accelerations experienced by the players during high speed ball to head impact. The re-design of soft helmets should be considered in light of the stiffness and deformation of the ball.

Training and technique will help to reduce injuries, but using a lighter ball or smaller size ball for training will also ameliorate the effect of repeated impacts.

ACKNOWLEDGMENT

Thanks to: the sports players who all volunteered for this study. Dr G. Wilson as medical practioner present; staff of the faculty workshop for manufacture of head external form and Dr Niall McGlashan for manufacture of head internals.

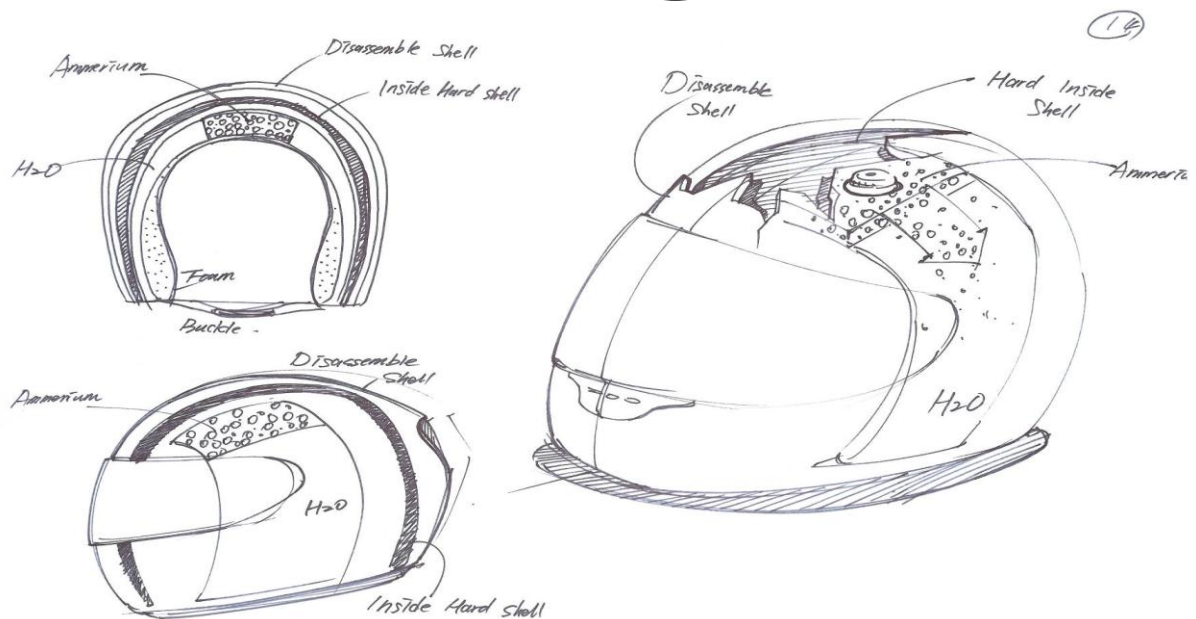
Ethical approval was not obtained for this study because the volunteers agreed to these training sessions accompanied by their professional coach. No head impacts were recorded on volunteers themselves. The forces and accelerations were those experienced by the inanimate soft targets similar or identical to those that might be used for training.

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Helmet Performance and Design



Application of an Effects Database in Idea Generation Approach for Helmet Design

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ABSTRACT

The paper presents an idea generation process for a helmet project using morphological analysis integrated with an effects database system. The objective of the project was to propose motorcycle helmet designs with beneficial heat transfer characteristics while ensuring performance functionality in aspects such as safety, comfort and aesthetics. The idea generation process was classified into three steps: keyword conclusion, related effects analysis, and idea generation. The effects database system provided application related effects to define the information scope for helmet related criteria. Based on the information gathered from the scope, data were generated to fulfil morphological analysis grids, and 18 overall solutions were generated by combing the selected means. This paper provides a description of the generalized approach and reports on one of the non-intellectual property sensitive ideas.

Keywords: *idea generation; helmet design; morphological analysis; effects database*

I. INTRODUCTION

In many countries, motorized two wheeler vehicles are a popular mode of transport. Compared with car drivers and passengers, motorcyclists have a lower degree of protection, especially of the head, a particularly vulnerable and delicate part of the human body. Many

motorcycle helmets have been developed and commercialised to protect motorcyclists' heads against impacts suffered during accidents. The UK government's authorized helmet test scheme has tested 289 helmets which were designated 'five star' with high performance in functional aspects such as safety, comfort and aesthetics [1]. Such helmets, however, normally have poor ventilation characteristics. They tend to be uncomfortable when worn in environments with high ambient temperatures or humidity. Indeed discomfort is frequently cited as a reason for not wearing a helmet by motorcyclists [2]. There have been numerous attempts to design a motorcycle helmet with a functional cooling system [e.g. 2, 3]. For example, a cooling system using a phase change material in conventional motorcycle helmets, which could cool the temperature inside the helmet to around 30°C for nearly 2 hours in tropical ambient environments, has been explored by several teams. Such cooling systems have tended to be designed in isolation, and tested on conventional motorcycle helmets. As a result the design modifications to enable effective temperature control impair other functional attributes of the helmets. To improve the heat transfer characteristics of motorcycle helmets while ensuring performance functionality in their original aspects, a helmet research partnership between Imperial College London and IIT Delhi has been formed.

The purpose of this paper is to describe some of the preliminary ideas generated by this partnership using an effects database approach in combination with morphological analysis. The effects database system provided expert guidance in defining the information scope of helmet related aspects. In section II, the helmet project is briefly introduced. The effects database system is illustrated in section III. Section IV provides a detailed description of the idea generation process for the helmet project by using morphological analysis integrated with the effects database system. An example sketch idea is presented, analysed and evaluated in section V.

II. HELMET PROJECT

In tropical countries such as India two wheeler vehicles are the major mode of transport. The study by [4] revealed that 25% to 70% of injuries or deaths in the South East Asian region were related to motorcycles. The level of injuries amongst unhelmeted riders indicates that wearing a safety helmet gives a clear benefit to motorcyclists [5, 6]. Figure 1 shows examples of helmets being used, or not used, by motorcyclists in India (images taken in January 2013 with ambient temperature ca. 5°C-12°C, in New Delhi).

Figure 1 indicates some scooters may carry unhelmeted adult or child passengers (pictures a, b and c). Some scooter drivers removed helmets to refresh their heads while they were waiting for the traffic signal (picture d). The main reason of these situations could be as follows. As the ambient temperature of tropical countries can be the range 20°C to 40°C, combined with high levels of humidity, the combination of excessive heat and sweat formation make helmets uncomfortable to wear [2]. Existing helmet standards require a helmet to have high performance in technical functional aspects such as penetration resistance and shock absorbing capacity, and reliability [7]. To be comfortably worn in the tropical countries, the helmets must also have advanced ventilation characteristics.

The helmet project described here is a cooperative research programme between Imperial College London and IIT Delhi. The major objective of this programme is to enhance effective design and optimization of ventilated motorcycle helmets to improve their heat transfer characteristics while ensuring performance functionality in terms of impact protection, brain rotation, and critically usability.



Figure 1: Examples of helmet use, and otherwise, by two wheeler drivers in New Delhi India

III. EFFECTS DATABASE SYSTEM

Sometimes, design challenges involve applications with which designers or design teams may not be

familiar. This can readily be addressed by consultations with subject experts. A challenge associated with this, however, is access to expert and effective dialogue and exchange of information. Key to dialogue between a specialist expert and designer is the framing of questions and understanding context. In order to improve access to expert information during idea generation, a database of design-related effects, named the effects database has been developed and implemented [8].

The effects database system described in [8] consists of 128 physical effects, 78 chemical effects, 28 geometric effects, 47 psychological principles and 46 design principles. The effects provided by the system aim to assist designers undertaking improvement updates on existing designs, such as integrating new functions or improving the performance of components and sub-systems as well in the formation of new concepts to deliver specific functions. For each effect or principle, a definition, book or article reference and a web reference were developed and selected. An example is shown in Table I.

TABLE I: AN EFFECT EXAMPLE

#	9
Physical principle	Thermal conduction
Definition	Energy transferred by heat. On an atomic scale, less energetic molecules in a continuum gain energy by colliding with more energetic molecules.
Book or Journal reference	Serway, R., Jewett, J. W. Jr., 2002. Principles Of Physics: A Calculus – Based Text. 3rd ed. Press: Thomson Learning.
Web reference	http://hyperphysics.phy-astr.gsu.edu/Hbase/thermo/thercond.html

The main function of the effects database was to provide problem related effects and principles to assist designers, or a team or individual involved in design, in rapidly defining the knowledge scope of design tasks. The standard working process of the system is shown in Figure 2.

The approach was developed as follows.

- *Step 1:* Based on the information and facts from previous stages in the design process, some relevant keywords are proposed. A keyword is a word related to the design problem.
- *Step 2:* Once the designer has entered a keyword into the database, the background programs search for keyword related effects or principles in

the database then list the outcomes in the results list page.

- *Step 3:* Designers briefly examine the information of each effect or principle in the results list page including its definition, book or article reference and web reference. After finishing examination of some or all of the keyword related results, designers can revisit Step 2 to enter another keyword.
- *Step 4:* Designers can refer to the book and web references of an effect or principle to explore more information from open-source knowledge-bases.
- *Step 5:* After all keywords had been entered and keyword related results examined, designers have an indication of the scope of knowledge of the problem related fields needed to explore.

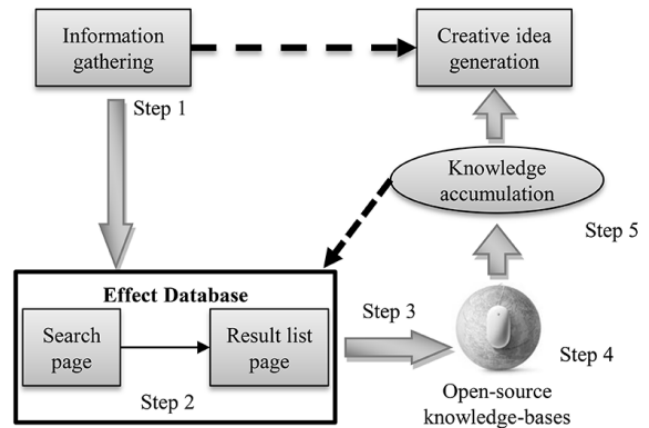


Figure 2: Using the effects database in the early stage of the idea generation process

Sometimes, the knowledge guidance from the effects database, and associated expertise, stimulates the designer to conclude new keywords or replace previous keywords by more appropriate ones. In this situation, designers were able to repeat the database searching process using the new keywords. The use of the database has been illustrated by application to a series of design tasks, indicating its suitability for promoting expert relevant suggestions [8].

IV. USING THE EFFECTS DATABASE SYSTEM IN IDEA GENERATION FOR HELMET DESIGN

A. Creativity tool

In the helmet design project, many creativity tools, including morphological analysis, have been used to produce sketch ideas for helmet design. Morphological

analysis, one particular tool used, is a combinational creativity technique that can be used to generate ideas for novel products. It involves formulation of the design space in terms of sub-functions and potential options for each of the sub-functions. The overall solution or set of possibilities is synthesized by the user's selection of sub-function options. Since morphological analysis was first developed by [9], it has been widely used and studied in diverse fields [10, 11]. Its standard working flow is as follows:

- Considering the function of a generic solution to a problem and breaking it down into a number of systems or sub-functions.
- Generating a variety of means to fulfill each of these systems or sub-functions.
- The sub-functions and potential means of fulfilling each of these sub-functions are arranged in a grid (shown in Table II).
- An overall solution is then formulated by selecting one means for each sub-function and the combination of these forms the overall solution.

TABLE II: A MORPHOLOGICAL ANALYSIS MATRIX

SUB-SYSTEM	MEANS		
Sub-system 1	Method 1 of fulfilling sub-system 1	Method 2 of fulfilling sub-system 1	Method 3 of fulfilling sub-system 1
Sub-system 2	Method 1 of fulfilling sub-system 2	Method 2 of fulfilling sub-system 2	Method 3 of fulfilling sub-system 2
Sub-system 3	Method 1 of fulfilling sub-system 3	Method 2 of fulfilling sub-system 3	Method 3 of fulfilling sub-system 3

B. Idea generation process

The effects database system was integrated with morphological analysis to provide expert guidance in defining the information scope of helmet related design aspects. The means that fulfill each of these systems or sub-systems were generated based on information gathered from the defined scope. The whole idea generation process (shown in Figure 3) was classified into three steps: keyword conclusion, related effects analysis, and idea generation.

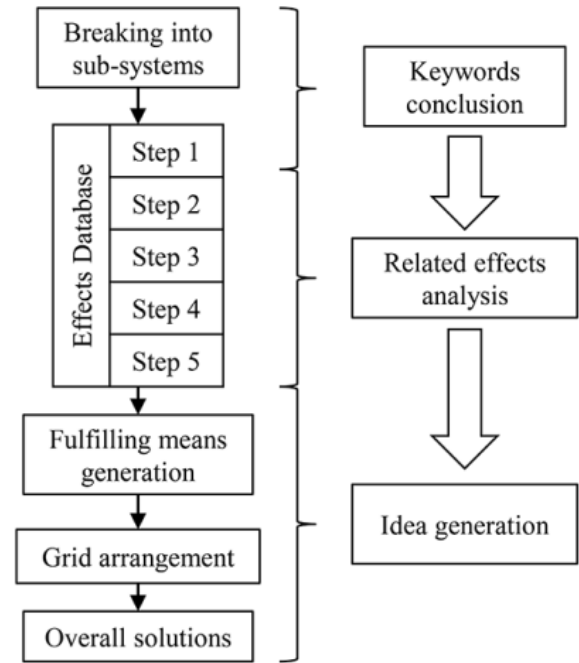


Figure 3: The idea generation process of morphological analysis integrating the effects database

1) Keywords conclusion

Following the working flow of the effects database system illustrated in Section III, the initial step was to conclude task related keywords from task background information.

Initially, the project description was analyzed and valuable information extracted (shown in Table III). Then, based on this information and with reference to design requirements, the design task was broken into several sub-systems. According to the working functions of each sub-system, its related keywords were concluded. The results are shown in Table IV.

TABLE III: VALUABLE BACKGROUND INFORMATION

Valuable background information
tropical countries; penetration resistance and shock absorbing capacity; aerodynamics; reliability; comfort; product aesthetics; light weight and good thermal characteristics; typical ambient temperature in the range 20 °C to 40 °C, combined with high levels of humidity

TABLE IV: SUB-SYSTEM OF THE DESIGN TASK AND KEYWORDS

System	Sub-systems	Keywords
Helmet design project	Outside force protection	Force
	Impact energy absorption	Force, Energy, Absorption
	Helmet fastening mechanism	
	Temperature control system	Energy, Absorption, Temperature

In Table IV there is no keyword for the sub-system ‘Helmet fastening mechanism’ as a buckle and belt design is the default approach used to tighten the helmet on drivers’ head in the helmet design. The keywords for the remaining three sub-systems were analyzed. For the sub-system ‘Temperature control system’, the internal temperature change is related to the energy change of the internal environment. Moreover, the thermal energy released from the surface of the drivers’ head must be absorbed to maintain cooling. Therefore, the keywords related with this sub-system were ‘energy’, ‘absorption’ and ‘temperature’.

2) Related effects analysis

Once all related keywords of each sub-system were concluded, the steps identified in Figure 2 were followed. Each keyword was entered into the effects database system and all keyword-related effects and principles were selected. The results of these keywords are illustrated in Table V. After further exploration of the detailed information of each effect, sub-system related items were classified. The related results are listed in Tables VI to VIII. For example, in Table VI, for the keyword ‘Force’, effects, such as ‘“Centrifugal” force’, ‘Magnus effect’ and ‘Thomson effect’ were ignored because of low relationship with the sub-system ‘Outside force protection’. Effects, such as ‘Complex molecules’, ‘Elastic deformation’ and ‘Elastic materials’, were classified into the related results. So that they could be used to generate means to fulfill this sub-system. In Table VII, the related results cell of keyword ‘Absorption’ was empty, as none of the searched results had relationships with the sub-system ‘Impact energy absorption’ after their details were explored.

TABLE V: RESULTS FROM KEYWORD RELATED EFFECTS

Key-words	Searched results
Force	“Centrifugal” force, Acoustic emission, Complex molecules, Elastic deformation, Elastic materials, Electric field, Flow of gases, Friction, Johnson–Rahbeck effect, Layering, Magnetic separation, Magnus effect, Proximity, Seebeck effect, Shape-changing objects, Shrinking, Thermal convection, Thomson effect,
Energy	Chemical bonding of gases, Chemiluminescence, Electric heating, Electro-chemical reactions, Emission of electrons, Endothermic reactions, Energy conversion, Exothermic reactions, Induced radiation, Input-process-output model, Irradiation, Isolated system theory, Luminescence, Nuclear magnetic resonance, Photoelectric effect, Radiation absorption, Radical recombination luminescence, Radioactive ray, Shape-changing objects, Thermal conduction, Thermal convection, Transport reactions, Thermo-chemical reactions, Tunnel effect, Vibration frequency sensitivity to thermal expansion, Wave movement, Zeeman effect
Absorption	Chemical reactions with light measurement, Franze-Keldysh effect, Hygroscopic effect, Luminescence, Mossbauer effect, Paramagnetic resonance, photoacoustic effect
Temperature	Changes in the optical-electromagnetic properties of materials, Condensation, Curie point, Diffusion melting, Endothermic reactions, Exothermic reactions, Explosion, Flow of liquids, Joule – Thompson effect, Magneto – caloric effect, Mossbauer effect, Rankine cycle, Seebeck effect, Self-propagating high-temperature synthesis, Super conductivity, Thermal – electrical phenomena, Thermal expansion, Thermal processing, Thermo-chromatic reactions, Thermomagnetic effect, Thomson effect, Use of strong oxidizing agents, Use of thermite mixtures, Vibration frequency sensitivity to thermal expansion

TABLE VI: RELATED EFFECTS OF THE SUB-SYSTEM 'OUTSIDE FORCE PROTECTION'

Feature	Outside force protection
Keyword	Force
Related result	Complex molecules, Elastic deformation, Elastic materials, Flow of gases, Friction, Layering, Shrinking

TABLE VII: RELATED EFFECTS OF THE SUB-SYSTEM 'IMPACT ENERGY ABSORPTION'

Feature	Impact energy absorption		
Key-word	Force	Energy	Absorption
Related result	Elastic deformation, Elastic materials, Flow of gases, Shrinking	Luminescence, Shape-changing objects, Transport reactions	

TABLE VIII: RELATED EFFECTS FOR THE SUB-SYSTEM 'TEMPERATURE CONTROL'

Feature	Temperature control system		
Key-words	Energy	Temperature	Absorption
Related results	Electro-chemical reactions, Endothermic reactions, Exothermic reactions, Energy conversion, Exothermic reactions, Induced radiation, Luminescence, Thermal conduction, Thermal convection, Thermo-chemical reactions	Condensation, Endothermic reactions, Exothermic reactions, Explosion, Flow of liquids, Joule – Thompson effect, Magneto – caloric effect, Movement of the chemical balance with temperature changes, Rankine cycle, Thermomagnetic effect, Use of strong oxidizing agents	Hygroscopic effect, Sorption

3) Idea generation

a) Morphological analysis chart

After all the sub-system related effects had been classified the knowledge scope of each sub-system was specified. The means that fulfilled each sub-system were then generated. The morphological analysis chart of the helmet design project is found in Table IX. For example, to fulfill the sub-system "impact energy absorption", the means were 'Foam' generated from effects 'Elastic deformation', 'Elastic materials', and 'Spring' generated from effects 'Shape-changing objects' and 'Shrinking', 'Air/Liquid cushion' generated from effects 'Flow of gases' and 'Elastic deformation', and 'Using disassemble material' generated from effect 'Shape-changing objects'. Including those means generated from related effects, some means were directly provided from gathered information. For example, the means 'liquid cooling system' mimicked the cooling system of the Sydney Opera House. The means 'Velcro' is widely used to attach two surfaces together.

b) Overall solutions

Solutions for helmet configurations were formulated by selecting means for each sub-function and combining selected means together. Some typical examples are illustrated in Figures 4 to 7.

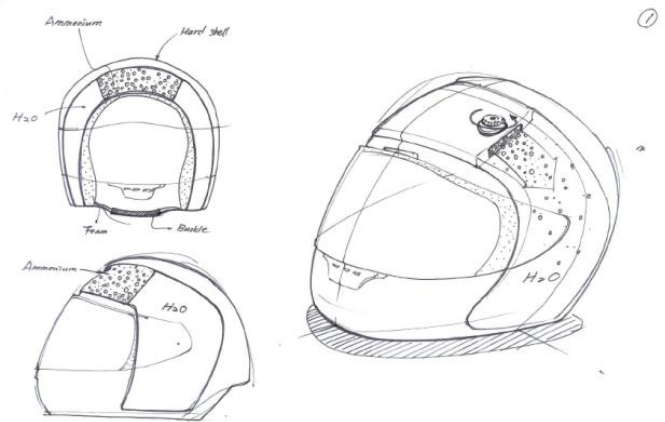


Figure 4: Combining means A1, A2, A3, B3 and A4

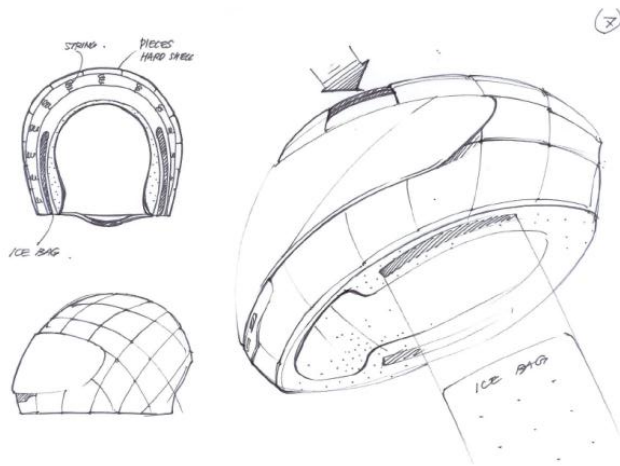


Figure 5: Combining means B1, B2, A3, B3 and B4

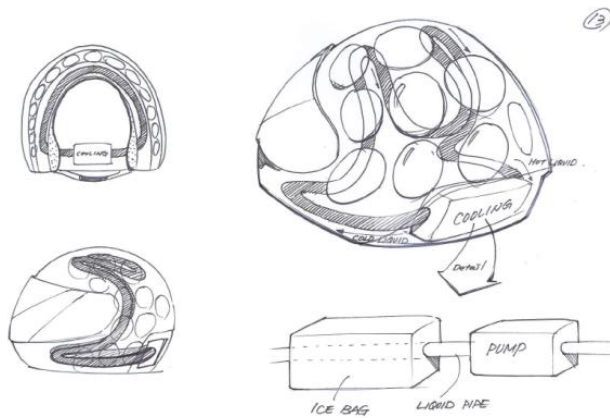


Figure 6: Combining means A1, C2, A3, B3 and D4

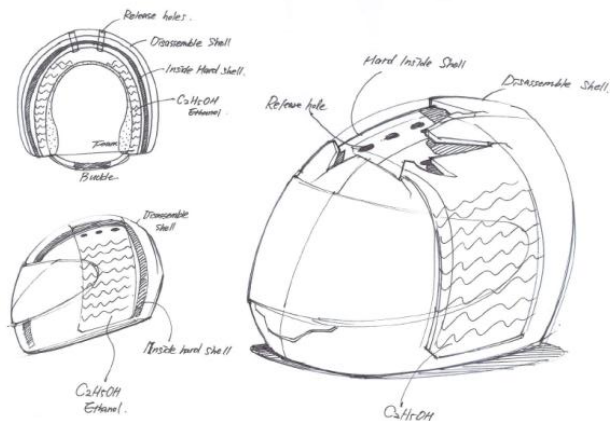


Figure 7: Combining means A1, D2, A3, B3 and C4

V. RESULT ANALYSIS

According to the design requirements of the helmet project and helmet standards, final products are evaluated according to the following aspects: penetration resistance

and shock absorbing capacity, head protection against brain rotation, aerodynamics and reliability, comfort, product aesthetics, low weight, and good thermal characteristics, as well as brand and economic considerations. To ensure that product ideas have good performance, concept sketches generated were initially evaluated according to these aspects. The examples in Figures 8-10 are used to illustrate the evaluation process.

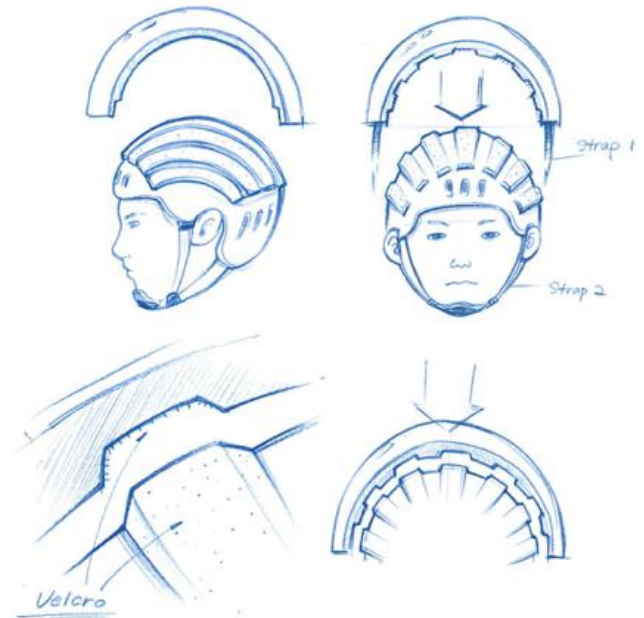
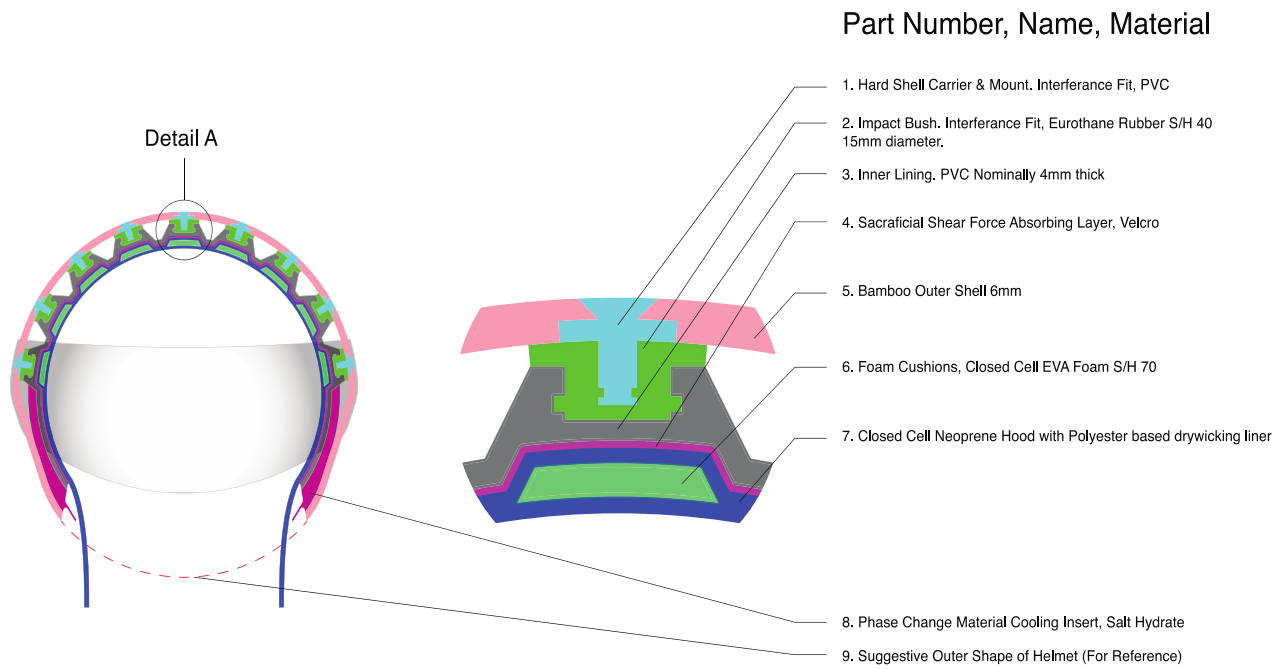


Figure 8: Elaboration of detail from Figure 5



The Helmet Cross Section

Detail A

Figure 9: Elaboration of detail from Figure 5

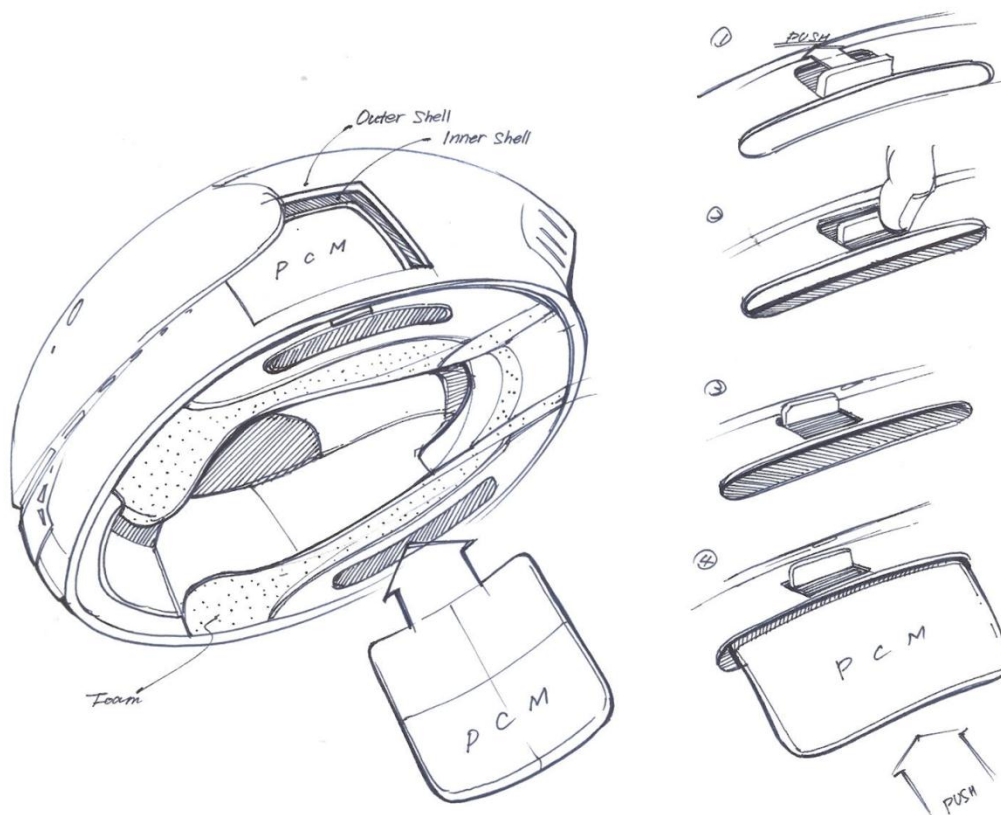


Figure 10: Elaboration of detail from Figure 5

TABLE IX: GENERIC MORPHOLOGICAL ANALYSIS CHART FOR HELMET DESIGN

		A	B	C	D
Features		Means			
1	Outside force protection	Outside hard shell (Complex molecules)	Elastic material (Elastic deformation, Elastic materials)	Force generator (Flow of gases)	Dynamic design to change the direction of the force (Friction)
2	Impact energy absorption	Foam (Elastic deformation, Elastic materials)	Spring, Bushing (Shape-changing objects, Shrinking)	Air/Liquid cushion (Flow of gases, Elastic deformation)	Using disassemble material (Shape-changing objects)
3	Helmet fasten mechanism	Buckle	Tight foam	Velcro	Magnets
4	Temperature control system	Chemical reactions absorb surround energy (Endothermic reactions)	Cold material (Thermal conduction)	Gas release from liquid phase to gas phase to absorb surround energy (Condensation)	liquid cooling system

The helmet illustrated in Figures 8-10 has two separate parts: the outer shell and inner hat (Figure 8). Users don the inner head hat. The outside of the inner hat, and the inside of the outer shell, are covered with Velcro so the two parts combine together where they meet. The mechanical tooth construction of the connecting surface assists in ensuring that the two parts combine correctly. The straps from the inner and outer parts are joined to the same chin buckle. The inner hat is made of comfortable soft material such as dry wicking fabric to fully fit the user's head shape. Its core is filled with a polyurethane based shear thickening polymer, similar to the D30 products [13].

The outer shell has three layers: outer shell, spring layer and inner shell (Figure 9). Thin bamboo plates are bonded together by hard shell carrier (part no.1) to create the outer shell of the helmet. Beneath the outer shell, the bamboo plates are mounted into the main impact bushing (part no. 2). The impact bushing is 15 mm in diameter with length 11 mm and is designed to interface with (part no. 3) the inner shell. The inner shell is made of elastic material making full contact with the inner hat. Two chambers are placed at the front of the helmet so that the cooling system using phase change material can cool the inside temperature (Figure 10). Phase change materials are pushed into the chambers through the entrance holes. New phase change materials can be easily reloaded.

Penetration resistance and shock absorbing capacity

Bamboo had been demonstrated to possess strength superior to some fibre-glass composites under certain conditions [12]. The bamboo outer shell will therefore be able to shield a certain level of penetration force when motorcyclists fall during accidents.

Head protection against brain rotation

The spring layer between the outer and inner shell of the helmet is designed to work in the same way as the layer of slippery cerebrospinal fluid sits as protection between the brain and the skull. When a rotational force impacts, this layer allows the outer shell to move slightly relative to the inner shell that tightly fits onto the motorcyclist's head. Much of the twist motion is eliminated before it reaches the brain. Bushing has been widely used as the vehicle suspension system. It can effectively absorb shocks from many angles and is likely to reduce damage to motorcyclists' heads [14].

Aerodynamics and reliability

The outer shape of the helmet is designed to mimic the outer shape of helmets on the market which have already demonstrated effective aerodynamic function. Straps of the two parts are joined in the same chin buckle and the two parts are tightly combined by Velcro. The design has good reliability during accidents.

Thermal characteristics

The cooling system using phase change material has demonstrated good performance in tropical ambient environments. The phase change materials are easily reloaded. Depending on their journey plans, motorcyclists could bring moderate amount of PCM encapsulated in bags. Therefore, this design has potentially good ventilation characteristics.

The aspects 'product aesthetic', 'light weight' and 'comfort' need to be tested with full-scale physical prototypes. The 'product aesthetic' largely depends on the helmet's outer shape design. If a helmet has good heat

transfer characteristics, it will be more comfortable to wear than a helmet with poor heat transfer characteristics.

It can therefore be concluded that the proposal has plausible design merits. Because this overall solution is formulated by combining selected attributes from the morphological analysis grid in Table IX, by replacing some sub-systems in this formulation, additional ideas may also demonstrate good performance. In all 18 concepts have been developed to date using this approach based on Table IX. The particular example illustrated in Figures 8-10 and described in this paper has been presented as it represents well explored intellectual property territory and therefore does not involve commercially sensitive issues.

VI. CONCLUSIONS

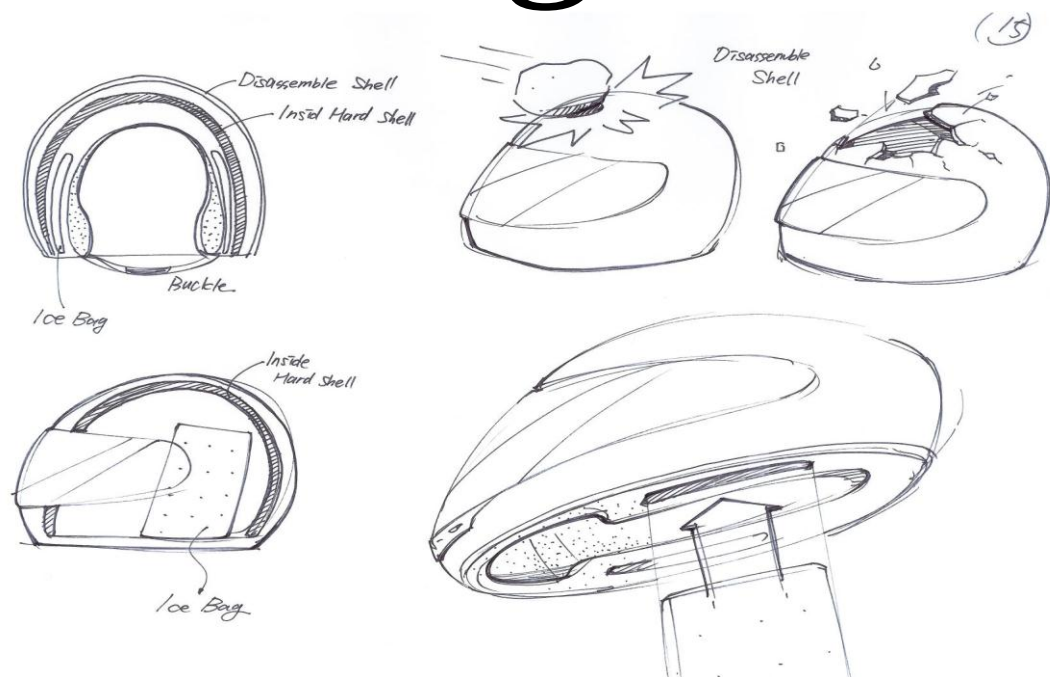
Morphological analysis integrated with an effects database system has been used to generate sketch ideas for helmet concept designs. Existing five star motorcycle helmets have high performance in functional aspects such as safety, comfort and aesthetics. Because of poor ventilation characteristics, the problem of excessive heat and sweat formation can make these helmets uncomfortable to wear in environments with high ambient temperatures and humidity. The objective of the project was to design motorcycle helmets with advanced heat transfer characteristics while ensuring performance functionality in other attributes.

A tool to enhance idea generation called the effects database has been developed and applied to helmet design. The idea generation process was classified into three steps: keyword conclusion; related effects analysis; idea generation. Assisted by information from the scope, each sub-system was found to fulfill several functions. Through combining selected means, 18 sketch ideas were generated. The evaluation process was illustrated by judging one example (shown in Figures 8-10) selected from the sketch ideas. This idea is designed to have good performance in the following aspects: penetration resistance and shock absorbing capacity, head protection against brain rotation, aerodynamics and reliability, and good thermal characteristics. The remaining aspects: comfort, product aesthetics, and light weight must be tested with physical prototypes.

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Helmet Performance and Design



Applying Problem Structuring Methods to the Design Process for Safety Helmets

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ABSTRACT

Problem Structuring Methods (PSMs) offer a useful set of decision support devices when integrated into the design process, particularly providing input on mitigation for a wide range of factors. The activity of design can typically start out as an unresolved problem and initiates exploratory creativity so that the early stages can involve the complex task of developing a variety of intangible concepts. PSMs are especially suited in developing scenarios where behavioural factors impact perceived physical/causal consideration, particularly at the early stages of a project, by reducing the number of “blind alleys” the designer may be induced to follow. Using the example of safety helmet design this paper presents a framework to illustrate how PSMs, including morphological analysis, the analytic hierarchy process and group facilitation methods, can be integrated into design projects. It is posited that helmet design can be explained as a function of the physical design parameters that include varied and complex technical considerations on the one hand, and contextual and behavioural factors on the other. The example presented shows how a potential large number of parameters and variables derived from the initial design scoping exercise design can be transformed into a handful of objectively established scenario which can then be worked upon by the design team.

Keywords: *problem structuring; decision support; morphological analysis; analytic hierarchy process; Facilitation; multi-methodologies*

I. INTRODUCTION

Design is multi faceted and multi-dimensional, consisting of both the physical/technical (or functional)

areas of design, as well as factoring in behavioural and contextual responses to the designed object by users of the end product. Weber and Condoor (1998) [1] state that

“...there are difficulties identifying independent design functions and determining the synergistic compatibility of combining solution alternatives. The concept generation phase is an important stage of the product realization process”.

It can be argued that a reductionist approach be used for many of the physical and functional aspects of a design – such as materials, structural integrity under stress, bio-mechanical response to external and internally generated stimuli. On the other hand, Design, inflected by contextual and behavioural considerations, is also open to a more holistic or systems approach. Often seen as two opposing epistemological approaches, both reductionism and holism can be accommodated and indeed integrated within a broader design system.

Physical design methods and the behavioural responses to design (many of which are not quantifiable), are highly complex, exacerbated by high levels of interconnectivity. This is not just due to the variety of parameters that have to be considered in the design process (physical complexity), but to intangible factors inherent within the nature of individual and group behaviour in response to designed objects and systems. At the contextual/ behavioural level many of these factors are; inherently non-quantifiable, contain non-resolvable uncertainties, cannot be causally modelled or simulated in a meaningful way. Instead, a judgemental approach must be taken [2].

With both tangible and intangible components, systemic design problems can be addressed by integrating a number of Problem Structuring Methods (PSMs).

This paper presents first an overview of PSMs in general and how they have developed as a core set of generic methodologies within the broader decision support domain. In addition the value of combining a number of such methodologies – as multi-methodologies, is discussed. Using helmet design the paper illustrates in outline how problem structuring methods can be applied to design.

The central part of the paper examines in greater detail how a number of PSMs, General Morphological Analysis (GMA) and the Analytic Hierarchy Process (AHP), when supported by PSM sourced facilitation tools, can generate not only a host of possible ideas but reduce them to a few viable options. The overall process is then presented as a three stage, 15 phase sequence of actions in the form of a conceptual framework. The safety helmet example is then used to illustrate in outline how an overall design issue can be transformed into a handful of objectively established scenarios. Finally areas for more empirically based research are highlighted.

II. PROBLEM STRUCTURING METHODS (PSMS)

PSMs have been developed to structure issues, problems and decision situations, rather than solving them [3]. Prior to their development it was seen that traditional Operational Research (OR) methods tended to restrict themselves to well-structured problems yielding finite answers which could be accurately monitored. In addition traditional OR addressed systems which were complicated, often having a large number of parts. However whilst such problems can be deemed to be complicated this does not imply that such systems have high complexity. Complexity can be seen as a function of multiple information sources which are linked and interdependent and are often uncertain or intangible [4]. In other words it was identified that there were whole categories of problems where traditional OR methods could not justify such a causal approach.

PSMs are largely used where a problem is poorly defined or unstructured. Such methods were developed independently from the mid-1960's onwards in both the US by academics such as Rittel and Weber [5], Kunz and Rittel [6], Ackoff [7, 8], and Nelson [9]. Rosenhead [3] clustered a number of mixed

academic/practitioner generated methods under the title "Problem Structuring Methods". In the UK PSM's have tended to be defined in fairly narrow terms based on, principally, British sourced methods such as Soft Systems Methodology (SSM), Strategic Choice Approach (SCA) and Strategic Options Development and Analysis (SODA).

The authors of this paper suggest that the term PSM can include a broader band of methodologies, which seek to address issues such as systems thinking, including systems dynamics, incorporating significant schools of thought by academics such as Stafford Beer [10] and Russell Ackoff [e.g. 7, 8] (see [11 and 12]).

Mingers and Rosenhead [13] state that unstructured problems are characterised by the existence of:

- Multiple actors,
- Multiple perspectives,
- Incommensurable and/or conflicting interests,
- Important intangibles,
- Key uncertainties.

Such criteria are highly relevant to a much broader spectrum of problems than that originally postulated as being a PSM, and can be expanded to embrace additional criteria such as volatility, interconnectivity, ambiguity, and complexity. PSMs have a vital role in "mitigating" rather than "solving" problems and are especially suited to scenarios where behavioural factors impact perceived physical/causal solutions.

III. THE ROLE OF MULTI-METHODOLOGIES

The very nature of unstructured problems would indicate that, such are the uncertainties inherent therein, no single methodology is likely to "solve" such a problem. Indeed, such a condition of initial ambiguity as to a defined outcome creates fuzziness, encouraging the introduction of assessing problems with more than one methodology – "multi-methodology". Mingers and Brocklesby [14] contend that to address such complexity as exists in the real world, it can be beneficial to combine different methodologies from different paradigms, and that practitioners are doing this already. They argue that in the real world individual methodologies "*differ in the degree to which they assist throughout the various stages through which interventions typically proceed*".

The concept of design embraces a large number of characteristics and that design of an object, system or idea, can be seen as being a problem waiting to be structured. Indeed one of the risks in the design

process is for the designer to be overly prescriptive and subjective, particularly at the initial conceptual and creative phases. A problem structuring approach, at the early stages of a project, can facilitate the design process by reducing the number of “blind alleys” the designer may be induced to follow.

The argument that a problem structuring approach is too structured, especially for a creative discipline such as Design, can be mitigated on five main counts:

1. PSMs only offer “decision support” – with the emphasis on “support”, and as such offer a framework for action rather than a specific solution.
2. PSMs work in the domain of the Uncertain, and thus by definition do not aim to prescribe any one particular solution – unlike traditional OR methods.
3. The process of problem structuring is essentially iterative, allowing for numerous individual, team and group interventions at any one stage – in effect PSMs encourage on-going review and evaluation of a design.
4. The highly complex, interconnected nature of the design process, combined with multiple variables often requires more structured methods (including computerisation), to help unblock creative thought.
5. PSM support-group based facilitation allows for additional creative inflexion for an idea.

IV. SAFETY HELMET DESIGN – APPLYING A PROBLEM STRUCTURING APPROACH

Design of safety helmets offers an interesting case of how PSMs can be applied to the early stage development of a design strategy. It is posited that helmet design can be explained as a function of the physical design parameters that include varied and complex technical considerations on the one hand, and contextual and behavioural factors on the other. Physical/technical components of the design can be addressed by a more reductionist approach whereas the contextual considerations are more suited to a holistic approach. Combined, the design will need to integrate both approaches. It is here that multi-methodological problem structuring processes can be applied.

For the purposes of this paper, six (non-exclusive) contextual scenarios have been identified which segment helmet usage, namely:

- Military - where the threat can consist of

high velocity projectiles such as bullets and shrapnel as well as fire and blast.

- Building Construction Head protection – where the threat is usually objects falling from a height.
- High Speed Motor Sports – such as high velocity impact either from falling off or being thrown out of the vehicle or from impact generated from other vehicles or stationery objects.
- High Speed Non-Motor Sports (downhill skiing, lugging, bobsleigh, cycling) – similar to above but where accidents caused by other vehicles are less likely.
- Low – speed sports (US football, cricket, ice hockey, lacrosse, horse-riding, rock climbing) – either contact sports or where a thrown object can cause injury.
- General purpose head protection – for example dangers to head caused from working in confined spaces.

Each of these scenarios require that the designer takes into account different physical and contextual factors when exploring design possibilities. In order to demonstrate the proposed methodology the helmet type selected is for a Motor Cyclist (High Speed Motor Sport). It should be highlighted at this point that the parameters and variable proposed are for illustrative purposes only and are not proposed as specific and definitive items for consideration in a full product development exercise.

Physical/Technical Parameters

Physical and technical design requirements might include, for example:

- Material use and performance
- Stress analysis of the physical components in a helmet
- The bio-mechanics of those parts of the anatomy to be protected by the helmet and the subsequent stresses the body is put upon under a wide range of impact scenarios (speed of deceleration, angle of contact in accident situations, etc.)
- Nature and type of force directed onto the helmet.

Expanding on the above additional parameters and variables within each of the parameters can be represented in a sample matrix format, as shown in Table 1

TABLE 1: MORPHOLOGICAL ANALYSIS MATRIX

Impact Type	Impact Provenance	Speed/ Force of Impact	Physical Design Weight	Primary protective Material	Manufacturing Complexity	Cost of Manufacture
Casque Penetration	Head-on/frontal	Over 60 mph	Over 5 kg	Foam	Can use current production line	Same as current models
Shock Absorption	Full side impact	40-60 mph	3.5-5 kg	Aluminium Honeycomb	Limited line modification	20% higher than current
Brain Rotation	Glancing side	20-39 mph	2-3.5 kg	Kevlar	Major line modification	50% higher than current
Compression	Rear	0-19 mph	Under 2 kg	Ceramic	New line required	Double
Whiplash effect	Crown (falling object)					
	Crown (head as projectile)					

Contextual/ Behavioural Parameters

There has been considerable research into the physical aspects of helmet design mechanics and components (especially into bio-mechanics), but this needs to be tempered by behavioural and psychographic factors as regards user uptake, which is likely to vary across helmet use scenarios – motor cyclist, construction worker, recreational sports

involving risk to head injury etc.

Examples of 8 main parameters are shown in Table II, which could be included in Behavioural (including Psychographic) and Environmental responses by target users and which can influence design considerations. Again it is highlighted that the parameters and variable proposed are for illustrative purposes only.

TABLE II: MAIN PARAMETERS

Age	Sex	Occupation	Fashion Preference	Peer Group	Individual attitude to risk	Cultural attitude to risk	Regulatory Environment
Over 60	Male	Retired	Functional & safe	Other Bikers	Very careful	Safety conscious	Highly regulated
40-60	Female	Management	Functional & Fashionable	“Fashionistas”	Moderately careful	Risk/reward	Increasingly regulated
30-39		Blue Collar	Fashionable only	Average Commuter	Basic safety	Fatalistic	Basic regulations
20-29		Student		Fun rider	“what the heck”		Unregulated laissez-faire
Under 20		Unemployed					

In our physical/technical matrix example there are 7 parameters or dimensions (P) identified as column

headers. Within each of these parameters discrete states or conditions (S) are then included. Later in the

paper we shall discuss in more detail how these parameters and states are determined.

The matrix represents a problem space (called a morphological matrix) for the selected safety helmet scenario and consists of some 30720 possibilities – made up of 5 variables or states in parameter column 1 times each subsequent column with its individual states.

The matrix generates, a large number of combinations or possibilities. As such it highlights the potential complexity of dealing with multivariate fields. In the contextual/behavioural matrix made up of 8 parameters, the problem space in the example consists of 28800 possibilities (5x2x5x3x4x4x3x4).

The design problem is exacerbated when, in addition to the problem space of the physical/technical aspects, are included the contextual/behavioural factors which may impact the former. By combining the contextual/behavioural factors with the physical/technical array it is just that, “combinatory” and not additive. The potential full problem space when both the physical and behavioural possibilities are combined (30,720 x 28800) is nearly 900 million! This is obviously far too great a number to work practically with – so how can we structure the problem better?

V. A PROBLEM STRUCTURING APPROACH: REDUCING THE PROBLEM INTO A SOLUTION SPACE

Problem space reduction can be undertaken as two separate stages – one for each of the matrices using a PSM called General Morphological Analysis, which is a general form of non-quantified, dimensional analysis [15]. The activity of design can be viewed as an unresolved problem and initiates exploratory creativity. General Morphological Analysis (GMA) is a methodology that can be used for idea and concept generation phases of the design process [1]. Generating concepts from a morphological matrix began over fifty years ago, pioneered by the Swiss-born astrophysics professor Fritz Zwicky (1898-1974), whilst at the California Institute of Technology. It is still used today as an important step in the engineering design process.

GMA, extended by the technique of cross consistency assessment, is a method for structuring and investigating the internal properties of inherently non-quantifiable problem complexes, which contain any number of disparate parameters. It encourages the investigation of boundary conditions and it virtually

compels practitioners to examine numbers of contrasting configurations and policy solutions. As a method for identifying and investigating the total set of possible relationships or “configurations” contained in a given problem complex, GMA’s primary task is to generate ideas with the aim of providing as many opportunities as possible.

Morphology is a methodology that enables organization of alternative solutions for each function of a system and combines them to generate a great number of solution variants each of which can potentially satisfy the system-level design need. The basic format consists of a matrix in a grid of columns and rows. The main aim of this technique is to single out the most important dimensions of specific problems and then examine all the relationships between them. “Morphology” means structure, and the technique seeks to explore all the possible “sub-structures” that a multi-dimensional matrix yields: hence its inclusion in the problem structuring method category. Consequently it is an exploratory approach and attempts to identify opportunities (or possibilities) and allow the user to “structure” a problem rather than solve it.

This methodology is not a replacement for creative thinking but a structured means for developing as well as documenting, design alternatives. It allows the designer to consciously explore design alternatives without confining them to human short-term memory limitations. The GMA variant applied here includes an audit trail mechanism to help the recording of assumptions made for the selected parameter headings and the states or conditions contained selected within each parameter and for each subsequent decision made at the assessment stage.

The complicated-sounding term morphological analysis belies a very simple method. In its most basic form it is no more complex than an ordinary matrix. If one takes a matrix of two axes with, say, ten items on each axis, one finishes with $10 \times 10 = 100$ combinations. If this is done as part of an idea-generation exercise one can produce 100 ideas with the greatest of ease. The novel concept that Zwicky propagated was that by adding a third dimension it was possible to multiply the output of the idea-generation process by the number of items on that dimension. Therefore if one can think of a useful third dimension and place on it ten items one will finish with as many as $10 \times 10 \times 10 = 1,000$ ideas. This is a larger output than any brainstorming session could ever hope to produce. A three-dimensional “morphology” can still be shown

in pictorial form as a cube. However if one is prepared to develop a non-pictorial morphology, one can add as many dimensions as one wishes. ($10 \times 10 \times 10 \times 10 = 10,000$ cells). This represents, literally, a mass-production unit for ideas – an “idea factory”.

The approach begins by identifying and defining the dimensions (or parameters) of the problem complex to be investigated, and assigning each of these a range of relevant “values” or conditions. A morphological box - fittingly known as a “Zwicky box” - is constructed by setting the parameters against each other in an n-dimensional parameter space. Each cell of the parameter space contains one particular value or condition from each of the parameters, and thus marks out a particular state or configuration of the problem complex.

Whilst an excellent concept for generating ideas derived from multiple dimensions or parameters it does create a practical problem in how to analyse all the configurations generated by the model. In our earlier example for a motor cycle safety helmet we calculated that the first matrix, made up of physical and technical components, generated over 30k combinations or possibilities – an additional 29k combinations for the contextual/behavioural space: resulting in a combined number of combinations, when the two matrices were brought together of nearly 900 million.

The solution to reducing this vast number is to *“examine the internal relationships between the field parameters and to reduce the field by identifying, and weeding out, all mutually contradictory conditions”* [15]. This is carried out for each matrix by an exercise called cross consistency assessment (CCA), where all of the parameter values in the matrix field are compared with one another on a pair-wise basis – similar to a cross impact matrix. As each pair of conditions is explored a judgement is made to see if the pair can co-exist. Using this approach Ritchey [16] notes that it is important to understand that there is no reference to causality – only to mutual consistency. Through this process a typical morphological field can be reduced by up to 99% (or more). The outcome is a matrix converted into a “solution space” and becomes an interactive inference model where any parameter or state, can be selected as an input and any others as an output.

For the purpose of the helmet design and given the application of the CCA exercise, our first problem space – the physical/technical matrix with 30720

combinations – can be reduced to a “solution” space of some 300 – which is much easier to work with. Similarly the contextual/behavioural matrix consisting of 28800 combinations can be reduced to a solution space of around 280. Even so – when combined the two solution space matrices would still generate some 84000 possible solutions – reducing the impact of the method as a practical decision support tool.

It is important to realise that the solutions generated via the GMA and its supporting CCA process make no assumptions about preference of solutions. The output of the model identifies purely sets of internally consistent solutions in a multi-criteria environment, and one in which any one individual or group of individuals is unlikely to be able to identify what multiples are totally consistent with one another. It is at this stage that other multi-criteria methods can be introduced. Here a method known as the analytical hierarchy process (AHP) has been used [15]. AHP processes subjective and personal preferences of an individual or a group in making a decision. With the AHP, hierarchies or feedback networks are constructed, then judgments made on pairs of elements with respect to a controlling element to derive ratio scales (similar to the pair-wise approach in CCA). These judgements are then synthesized throughout the structure to select the best alternative. One can see that the role of facilitation insinuates itself into such a process and although judgements can be made by any one individual, a consensus driven approach based on heterogeneous group composition, will yield a more rounded and objective response.

Fundamentally, AHP works by developing priorities for alternatives and the criteria used to judge the alternatives. Usually the criteria, whose choice is often made by a hierarchically positioned decision-maker, are measured on different scales, such as weight and length, or are even intangible for which no scales yet exist. As for GMA both quantitative and qualitative criteria can be used in the process. Measurements on different scales, of course, cannot be directly combined. First, priorities are derived for the criteria in terms of their importance to achieve the goal, followed by priorities for the performance of the alternatives on each criterion. The process of prioritization solves the problem of having to deal with different types of scales, by interpreting their significance to the values of the user or users. Finally, a weighting and adding process is used to obtain overall priorities for the alternatives as to how they contribute to the goal. There is a danger here that for

certain practitioners of the method might interpret the weighting in purely discrete terms. This is erroneous as the method really aims to identify “relative” differences between criteria rather than absolute differences. With the AHP a multi-dimensional scaling problem is thus transformed to a single dimensional scaling problem.

As for many methodologies in the Problem Structuring domain the AHP, like GMA is readily suited to work with non-quantitative criteria. Both methodologies employ a pair-wise comparison approach and thus certain learning economies of scale can be applied when using GMA, and AHP. There is the issue of sequence when using such methods. GMA should always be used first as – looking at the possibilities. GMA assesses from an “either/or” standpoint whereas AHP considers “ands” (or rather groups of “ands”).

For the safety helmet application the physical/technical problem space can be reduced by CCA to some 300 possible solutions. These 300 solutions can be converted into a hierarchy enabling application of AHP. The AHP provides an additional filtering process so that, for example, the top 5 places in the derived hierarchy are selected to represent scenarios set against the contextual/behavioural matrix solution space. However prior to this the contextual/behavioural solutions can be subject to the AHP – so that again, say, the top 5 solutions are selected. The next iteration will be to run a GMA exercise combined core parameters and variables from each of the two original matrices. Again the resultant solution space would be subject to the AHP so that the final top 5 in the hierarchy are selected for more detailed scenario development. Iteration is an essential part of the process of applying these methodologies to arrive at a more objective short-list of viable ideas for design purposes.

There is one further refinement activity that needs to be integrated into the above process and this introduces the role of facilitation and other consensual methods for decision making.

VI. THE ROLE OF FACILITATION

Problem Structuring Methods can be seen to act in a key decision support role when dealing with Design problems and their inherent complexities. Both GMA and AHP are methods suited to collective concept exploration creativity and the development of collective understanding of complex problems.¹ Groups can be used to bring together actors and

stakeholders with differing viewpoints, to help develop new designs for products and services and to help co-ordinate complex projects [17]. Ritchey [15] identifies that ample evidence exists indicating that group interaction repeatedly outperforms individual capacity in the area of concept exploration and creativity for open-ended problems. The principal role of the facilitator is to help the group to increase its effectiveness by acting as a neutral, particularly where problems are of a complex nature and where the stakeholders within the group can take different personal and organisational positions.

The composition and behavioural profile of the group being facilitated is important. Each of the members of the group should be subject matter specialists in their respective area and that such expertise be composed of heterogeneous function experts and where duplicate competences are reduced to a minimum. In other words it is unadvisable to have too many group members from the same discipline if “group think” is to be avoided. Where specialists can be brought together on-site and in one location then workshops can be run as on a continuous basis and where such workshops can last up to two or more days, depending on the complexity of the problem being addressed.

Such are the demands on people’s time in the workplace that it can be extremely difficult to get such a varied group of people in one place for lengthy periods of time – especially in geographically dispersed organisations. Nevertheless there are other approaches which can be employed to mitigate such difficulties and which themselves form part of those methods under the PSM panoply. They operate by offering a dispersed facilitation option. The two most well-known methods which allow for various stakeholder opinion to be accommodated and co-ordinated are IBIS (Issue-Based Information Systems) and the Delphi method.

In this paper the authors will be referring mainly to IBIS type methods as the Delphi approach, (originally developed by the Rand Corporation in the 1950’s), although an accepted form of dispersed group input to complex problems, can be seen as a more structured and rigid form of achieving consensus. For readers wishing to explore the Delphi method in more detail then we would refer them to [18]. IBIS was originally developed by Kunz and Rittel in the 1960’s and 70’s and is an argument based scheme. Subsequent variants have been developed by [19] using the term Dialogue Mapping with a more design focussed variant being

developed by Imperial College, called DesignVUE [20]. The advantage of the latter two methods is that they are available in a computerised format.

Issue-Based Information Systems (IBIS) are used as a means of widening the coverage of a problem and hence is highly suitable to engaging a dispersed group of specialists and stakeholders. By encouraging a greater degree of participation, particularly during the initial stages of a design process, the lead designer is able to increase the opportunity that issues of his or her proposed solution, in dispersed locations will be discovered and discussed by others. Since the problem may be a symptom of another higher-level problem, this approach also increases the likelihood that someone will attempt to provide alternative views which can help evolve the dialogue. Issue-Based Information Systems also help to make the design process “transparent” through the use of the all important audit trail (how did we get here?).

Within the design domain “designVUE” is particularly suitable as a dispersed facilitation dialogue tool as it addresses both individual and collaborative design practice. The application has been enhanced to provide support for a range of design and engineering information processing activities including requirement capture and justification, design rationale capture, functional modelling and decision making. The tool supports Issue-Based Information System (IBIS) argumentation and bi-directional hyperlinking between designVUE files. With this functionality it is possible to capture design thinking including issues, solutions and pros and cons while easily importing supporting evidence such as images, web-links, documentation and multimedia files enabling users to capture sources of inspiration, integrate supporting evidence and visualise design decisions.

VII. A CONCEPTUAL FRAMEWORK TO SUPPORT THE DESIGN PROCESS

Part 1: Creating the Physical/technical Scenarios

1. Determine the core focus of the design – what is the problem – are there two modelling environments, for example – the physical/technical (internal/strategic) or the contextual/behavioural?
2. Identify and select key stakeholders in the design process (maybe even include a representative of the end user). These stakeholders should be experts within their specified areas and broad enough to embrace those issues covered by the physical/technical

scenario and the contextual/behavioural scenario.

3. What are the availabilities of the team members – are they constrained by time and/or location? If such constraints apply then elect to use an IBIS based form of facilitation such as designVUE.
4. Using the GMA process, establish what are the main parameters or dimensions of the problem – use facilitation to determine an agreed consensus for selecting the parameters.
5. Repeat the above process to determine the discrete states (or conditions) within each of the parameters.
6. Carry out the Cross Consistency Assessment exercise.
7. Format the agreed matrix defining the physical/technical problem space.
8. Compile the internally consistent model to define the “solution” space to identify the internally consistent scenarios.
9. Use facilitated and group supported AHP to select the top five preferred solutions from the physical/technical matrix – these will become scenarios against which the contextual/behavioural matrix will be evaluated.

Part 2: Assess the Impact of Context/Behaviour Factors on the Physical/Technical Scenarios.

10. Via GMA repeat stages 4 and 5 to create the contextual/behavioural matrix.
11. Introduce the 5 AHP sourced outcomes from Part 1 as a set of discrete scenarios to the agreed matrix in 10.
12. Carry out the facilitated Cross Consistency Assessment exercise as in Part 1.
13. Format the agreed matrix defining the problem space derived from the scenarios and the contextual/behavioural matrix.
14. Compile the internally consistent model to define the “solution” space.
15. Use facilitated and group supported AHP to select the top five preferred solutions from the solution space in 14.

Part 3: Introduce Traditional Scenario Planning

The final five scenarios selected from stage 15, above, can now be used as highly objectively determined input for final evaluation by the design team using traditional evaluation techniques.

Using this sequence of PSMs it has been shown

how, in our motor cycle safety helmet example, a potential problem space consisting of over 900 million combinations can be reduced via “facilitated consensus” to a total internally consistent solution space of just a few hundred (via GMA/CCA), which in turn using the AHP, can be reduced, also by facilitated consensus, to an agreed hierarchy of just 5 scenarios – the selection of which will have a greater likelihood of being converted into a design which satisfies major stakeholders – as creative designers and contented end users.

VIII. CONCLUSIONS

The methodological framework presented in this paper brings together a number of PSMs. However it must be understood that what is being proposed is not a series of rigid methodological processes being bolted together – as identified earlier the boundaries between such methods can be imprecise and blurred and fuzzy. Facilitation methods and tools are key to supporting the decision making process within the overall framework especially as many of the issues can be occluded.

Indeed, facilitation is core to a number of these methods and complements the more analytic components contained within the methodologies concerned and where computer support has radically enhanced the practicalities of such methods. As highlighted above these analytical aspects of the methodologies need to be complemented by powerful centralised and de-centralised facilitation if consensus is to be achieved. The full potential of these methodologies is difficult to realize unless the analytical and facilitation elements, are not only combined, but treated with equanimity. An over or under emphasis on any one of these two approaches can reduce the effectiveness of the overall derived model. Facilitation helps to ensure that a more collective/consensual approach to design achieved rather than being overly subjectively based via the “genius in the workshop”.

The methods and the sample process shown in this paper are intended to introduce as much objectivity as possible into the design process and mitigate overly subjective outputs which can constrain the overall design process.

On-going research in helmet design has correctly concentrated on the more functional and technical aspects of helmet performance when under severe stress due to accidental impact. Apart from those

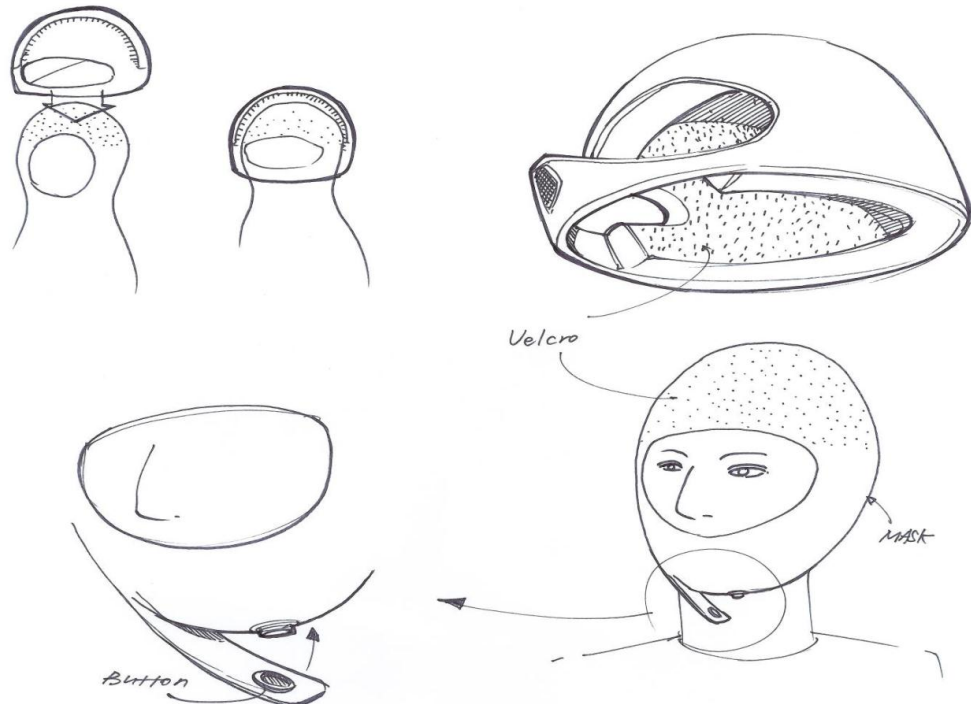
aspects of design which specifically seek to reduce injury, a major uncertainty remains when behavioural factors and attitudes are taken into account. *Is the helmet comfortable to wear - when the weather is hot or cold? Do I look “cool” in it? Do I really have to wear a helmet? How much does it cost?* These issues include a variety of intangible and often unquantifiable behavioural responses by potential wearers of the helmet. PSMs such as GMA and AHP, are suited to address such non-quantitative issues. They can provide helmet designers with additional insights and ensure that the end design is a result of a more holistic approach.

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Helmet Performance and Design



An Examination of Headform Dynamic Response for Concussive and Traumatic Brain Injuries

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ABSTRACT

Brain injury involves a spectrum from minor concussions to severe traumatic brain injuries (TBI) (Cantu, 1992). Head injury reconstruction research has been employed by a number of authors to better understand the mechanism associated with these injuries. In the past, reconstructions have largely focused on either concussions in sports or traumatic brain injuries involving pedestrian accidents (Doorly & Gilchrist, 2006; Pellman, Viano, Tucker et al., 2003). Concussions can be divided into two groups. The first, symptoms subside after a few days, weeks, or months, while the second, less studied group, are concussions resulting in persistent symptoms lasting years. There is little research describing the differences between concussions that present persistent symptoms with those that resolve after a few weeks. The objective of this research is to compare the dynamic response characteristics of accidents resulting in persistent post concussive symptoms to those describing both non-persistent concussion, and traumatic brain injury.

A total of 9 injuries were reconstructed and analyzed, 3 concussions with non-persistent symptoms were helmeted head impacts in hockey. The three concussions with persistent post concussive symptoms were unhelmeted falls and the three subdural hematoma and contusion cases (TBI). Impact reconstruction parameters such as velocity, location and the impacting surface for each accident were obtained from medical reports and video footage of the event. The test set-up used a monorail drop tower to impact the Hybrid III head and neckform in a manner that was representative of the reconstruction as established from reports.

The linear and rotational accelerations common to TBI (318 g and 23.0 krad/s²) are significantly higher than the impacts causing non-persistent concussion (146 g and 6.2 krad/s²). The dynamic response of the TBI group was higher than the persistent post-concussive group (182 g and 16.4 krad/s²) however, the linear and rotational accelerations between these groups were not significantly different. These findings suggest that levels of impact producing persistent post-concussive symptoms approach similar levels to those producing TBI. When examining the dynamic response of the individual cases, accelerations between the persistent post concussive group had responses that overlapped the TBI group. This study found that the dynamic response of head impacts resulting in persistent symptoms are similar to those resulting in TBI. The dynamic response characteristics that distinguish between persistent post-concussive injuries and traumatic brain injuries needs further investigation.

Impact Studies on Motor-cycle Helmets with Different Shells

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ABSTRACT

Helmets are widely used by two-wheelers for protecting the head against impact during an accident. The impact performance of helmets with shell made of composite and shell made of metal foam is studied.

Impact behavior of a composite shell for motorcycle helmet is investigated. In a composite, intra-ply damage and delamination are the principal modes of failure and energy absorption. Numerical studies are performed on helmets with outer shell of carbon/epoxy [0₃ /90₃] laminate. The damage in the lamina level is analyzed by using Hashin's failure criterion which is based on physical modes of composite failure. The delamination which is also called as interfacial fracture is examined by 'cohesive zone model' based on the fracture mechanics principles. Cohesive elements are introduced between the plies to simulate the initiation and propagation of the delamination front. Matrix tensile and compressive damage are observed at low impact velocities and followed by dynamic delamination at higher impact velocity. The increase in energy absorption in the composite shell, through marginal, is increased due to this damage and consequently force experienced by the head is reduced compared to the composite shell without the damage.

The impact behavior of helmets with metal foam outer shells is studied. Experiments and finite element analysis are carried out on metal foam helmets and a good agreement is obtained between the two. Energy absorption per unit mass is higher in aluminum foam shells with low density. The weight of the outer shell with low-density aluminum foam is reduced by approximately 70% compared to the conventional ABS shell

The Assessment of Inbound Mass Variation on the Distribution of Brain Tissue Deformation

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ABSTRACT

Although the incidence of severe head injuries has dramatically decreased as helmet use became mandatory practice, concussions remain an issue. The persistent occurrence of concussions in sport is a growing concern due to potential long-term effects. There are 225,000 new patients annually in the US that show long-term deficits associated with mTBI. The severity and recovery time of a concussion vary involving a variety of possible signs and symptoms. It is proposed that the symptoms experienced may be associated with the damaged areas of the brain (Post et al 2013). In sports, concussions are often the result of player-to-player impacts involving shoulders, arms and hands and thus different impact masses. Therefore, when determining head injury risks through event reconstruction, it is important to understand how individual impact characteristics (mass, velocity, location) influence regional brain responses. The purpose of this investigation was to examine how changes of impact mass influences the brain region with the highest magnitude of tissue deformation.

A helmeted Hybrid III headform fitted with a 3-2-2-2 accelerometer array was impacted using a pendulum system. Impacts were performed using six different inbound masses increasing by 2 kg increments (4.3, 6.3, 8.3, 10.3, 12.3, and 14.3 kg) at a velocity of 4.0 m/s. A new hockey helmet was used for each inbound mass. Six impacts were performed on each helmet, consisting of three trials for each of two impact locations representing common methods to be hit in hockey; one centric and one non-centric. Brain tissue response, peak maximum principal strain (MPS) and peak von Mises stress (VMS) was produced using the University College Dublin Brain Trauma Model (UCDBTM). Tissue deformation was measured within nine functional brain regions associated with the symptomology of concussion.

Results showed that an increase in the inbound mass produced an overall increase in the magnitude of brain tissue response. MPS increased from 12.7 – 17.0%, and 13.4 – 16.7% from an increase in inbound mass, under the centric and non-centric conditions, respectively. An increase in inbound mass also resulted in VMS changes from 4.1 – 5.5 kPa and 4.3 – 5.4 kPa under the centric and non-centric conditions, respectively. A change in inbound mass revealed a shift in which brain region experienced the highest MPS magnitude for both the centric and non-centric impact condition. The VMS peak magnitude occurred in the same brain region consistent across all inbound masses. However, the region experiencing the highest VMS was different under the centric and non-centric condition. In addition, the maximum tissue response occurred in the brain regions directly at the impact site for centric impacts, however the region experiencing the highest stress and strain under non-centric conditions were contralateral to the impact site.

This study demonstrated that an increase in the magnitude of neural tissue deformation results as the inbound mass of the impact is increased. A regional shift in the largest peak deformation magnitude was also observed, however it was dependent on the impact conditions and dependent variables being considered.

The Patent Landscape for Protective Headgear Technologies

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ABSTRACT

ClearViewIP, the IP Strategy Consultancy has analysed the headgear patent landscape during the commercialisation project of a headgear patent. While R&D in the field of head protection has taken place over a long period of time, the growth of patent filing grew dramatically in the last 20 years. Headgear technologies have followed the same pattern.

Reviewing the patent landscape can enable researchers and commercial managers to get a clear picture of who the main players are overall, from what industry segments and what their filling pattern has been.

Researchers can use patent databases to review older patents to help guide innovation and can review the latest patents or applications to get a feel for what the latest research and development themes are in their particular field.

This presentation addresses the patent analytics and key themes from the patent landscape for protective headgear technologies (filing rate, geography, key players, broad technology utilisation, and notable patents).

Our analysis shows that roughly half of all patent publications are from the US, although Japan is also a key territory for head protection innovation – primarily from motorcycle helmet manufacturers. It also appears that peaks in patent filings in this field are closely linked to the development of new legislation and safety standards. More recently there has been a growing trend towards patent filings that feature innovations in helmet comfort and integration with accessories (mp3 players / mobile phones).

Finite Element Analysis of Helmeted Impact and Head Injury Assessment of a Commercial Motorcycle Helmet

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ABSTRACT

In this work the performance of a commercially available motorcycle helmet was assessed under impact and approved by the majority of current standards. The evaluation is based on accurate reproduction of impacts that are mandatory by the current standards, required to helmet approval, using the Finite Element Method (FEM) to perform such task. The numerical framework is validated against two set of experimental data. The first concerns the constitutive model of the expanded polystyrene (EPS), the material responsible for energy absorption during impact; the second related to the headform centre of mass acceleration measured during the impacts defined in the European ECE 22.05 standard. After the validation of the Finite Element motorcycle helmet model, the energy absorbing test measured at point P was induced in a biomechanical Finite Element head model in order to predict the resultant head injuries. From this analysis, it was concluded that brain injuries such as concussion and diffuse axonal injury can occur with high probability of occurrence even with a helmet that was approved by the majority of the helmet standards. At the end, conclusion points out a strong recommendation on the necessity of improving the current head injury criteria used in motorcycle helmet standards to assess helmets performance, in order to improve the safety between the motorcyclists.

Helmet Performance and Design
Imperial College London
15th February 2013

Author Index

	Chapter
Baby, A.	A2
Blanco, D.H.	9
Brien, S.	A1
Caserta, G.D.	2, 7, 9
Cernicchi, A.	9
Childs, P.R.N.	12, 13
Coomber, R.	6
Crofton, P.S.J.	11
Cusimano, M.	A1
Dabbagh, S.	4
de Sousa, R.J.A.	A5
Deck, C.	3
Fernandes, F.A.O.	A5
Galvanetto, U.	2, 7, 9
Garvey, B.	13
Geurts, B.	A4
Ghajari, M.	2, 7, 9
Gilchrist, M.D.	10, A3
Haley, A.R.	5
Halldin, P.	1, 6
Hoshizaki, T.B.	10, A1, A3
Hoult, T.R.	11
Iannucci, L.	7
Karton, C.	A1, A3
Keane, I.,	4
Kleiven, S.	1, 6
Lanner, D.	6
Lee, H.K.	12
Mahajan, P.	8, A2

Marshall, S.	A1
McGlaughlin, D.	12
Oeur, A.	10, A1
Pangonis, R.	4
Pedder, J.	11
Plant, D.J.	11
Post, A.	10, A1, A3
Rousseau, P.	A1
Sanghi, S.	8, A2
Shishodia, B.S.	8
Smith, A.	A1
Townsend, J.	11
Walton, R.	A4
Wang, Z.	12
Willinger, R.	3, A5
Wilson, H.	4

Helmet Performance and Design

Friday 15th February 2013 was a landmark event in the area of helmet performance and design. A group of engineers, scientists, and industrialists met to consider the integration of understanding and scientific insights in head injury with protection systems.

The keynote lectures were testimony to the breadth of contribution required to make significant advances in the field and the papers presented showed that there is significant current work in incorporating the medical, scientific, engineering and design contributions to effect change in protection.

The collection of papers in this volume also show that there is much work to be done. Disciplinary silos will always exist, but by developing a common language and a dialogue there is hope that significant advances will be made.

The conference ended with a commitment by the organisers to pursue this common goal and therefore these proceedings will be followed by another volume to be produced for the second conference which will take place in India at the end of 2013."



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