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Motorcycle helmets—A state of the art review

F.A.O. Fernandes, R.J. Alves de Sousa*

TEMA – Centre for Mechanical Technology and Automation, Department of Mechanical Engineering, University of Aveiro, Campus de Santiago, 3810-193 Aveiro, Portugal

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ABSTRACT

This paper tries to make an overview of the work carried out by scientific community in the area of road helmets safety. In an area that is constantly being pushed forward by market competition, self-awareness of danger and tighter standards, several research groups around the world have contributed to safety gear improvement.

In this work concepts related to head impact protection and energy absorption are explained. It also makes reference to the theories related to the development of helmets, as well as to the different existing types nowadays. The materials that are typically used in impact situations and new design concepts are also approached. In addition, it is presented a literature review of current – and most commonly used – helmet test standards, along with new tests and helmet concepts to assess the effects of rotational motion.

In a non-restrictive, and never up-to-date report, a state-of-art review on road helmets safety is done, with a special insight into brain injury, helmet design and standards.

1. Introduction

Road accidents are one of the major causes of death in the world (WHO, 2009). About 31 thousand people die and 1.6 million people are injured every year in the European Union as a direct result of road accidents (ERSO, 2012). Motorcyclists are less protected against road accidents than the users of some other vehicles because they have the safety helmet as the most effective means of protection, while car occupants, for example, are protected by safety belts, airbags and even by the body structure of the car. This is also confirmed by Koornstra et al. (2003) and by Peden (2004), and also by Lin and Kraus (2008) that report that motorcycle's riders are over 30 times more likely to die in a traffic crash than car occupants. Thus, motorcycle crash victims form a high proportion of those killed and injured in road traffic crashes, as shown in Table 1 for Portugal. In Portugal, 21% of all road accident fatalities and 24.9% of all road accident severe injuries at the year of 2011 were suffered by powered two wheelers (PTW) occupants (ANSR, 2010, 2011).

Nonetheless, motorcyclists account for 14.6% of total road-user fatalities in European Union, 12.1% in Australia, 9.4% in the USA and 9.2% of total traffic fatalities in Japan (Subramanian, 2007). These statistics show once more the low capacity of protection of this means of transport. A more recent study shows that in the European Union road accident fatalities increased 17.7% among PTW occupants involved in traffic accidents and the number of fatalities was almost 6000 in 2008 (DaCoTA, 2011). In the developing countries, where motorcycle is the main means of transport, the contribution to the total road traffic fatalities is about 90% (WHO, 2009).

As already shown, motorcyclists are at high risk of injury in traffic crashes and the head is one of the areas most subjected to severe and fatal injuries. Head injury is one of the most frequent injuries that result from motorcycle accidents, as shown in Fig. 1, where head injuries occurred in 66.7% of the cases of COST database (COST, 2001). This study also reports that the majority of these injuries were severe. Other statistics on motorcycle accidents show that between 2000 and 2002 in the USA about 51% of unhelmeted riders suffered head injuries as compared to 35% of helmeted riders (Subramanian, 2007), showing thus the importance of wearing a helmet. In the same study is shown that in 27% of the fatalities the only injury present was head injury. In 2008, 42% of fatally injured motorcyclists (822 deaths) were not wearing helmets and NHTSA estimates that the majority of these unhelmeted motorcyclists would have survived if they had worn helmets (NHTSA, 2011) and also estimates that motorcycle helmets are 37% effective in preventing fatal injuries (NHTSA, 2008). This effectiveness has increased over the years possibly due to improvements in helmet design and materials (Deutermann, 2004). Brown et al. (2011) and Sarkar et al. (1995) concluded that riding and crashing a motorcycle while unhelmeted is associated with more frequent and more severe injuries and increased mortality. King et al. (2003) showed that linear acceleration transmitted to the head is always superior

* Corresponding author. Tel.: +351 234 378150; fax: +351 234 370953.
E-mail address: rsousa@ua.pt (R.J. Alves de Sousa).

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Table 1
Number of fatalities and injuries suffered by PTW occupants in Portugal at the years of 2010 and 2011 (ANSR, 2010, 2011).

Year	Total	Minor injuries	Severe injuries	Fatalities
2010	7603	6844	556	203
2011	7454	6703	564	187

in unhelmeted cases. However, the authors stated the same degree of angular acceleration for helmed or unhelmeted head.

Other studies show that helmets reduce the risk of death in motorcycle collisions by approximately 42% and reduce the risk of head injury by 69% (Liu et al., 2008; MAIDS, 2004).

As a result of motorcycle accidents, head injury is considered a major cause of death, accounting for 70% of the total deaths where helmet usage is the most important factor in preventing it and in reducing the risk of head injuries in motorcycle crashes (Abbas et al., 2012; Liu et al., 2008; Servadei et al., 2003; WHO, 2009). Hence, a motorcycle helmet is the best protective gear that is possible to wear while riding a motorcycle, being the most effective means of protection offered to motorcyclists.

Although some issues can be pointed out about motorcycle helmets, like the fact that their usage decreases motorcyclist vision and increases neck injuries, motorcycle helmets were found to reduce the risk of death and head injury in motorcyclists that crashed, therefore helmet's benefits and its usage is advised by several studies (Abbas et al., 2012; Brown et al., 2011; Deutermann, 2004; Forman et al., 2012; Liu et al., 2008; NHTSA, 2008, 2011; Sarkar et al., 1995; Subramanian, 2007; WHO, 2009).

The following sessions will give a detailed overview on the developments carried out so far on the content of helmet safety technology. Firstly, a brief and chronological introduction about motorcycle helmet origins and evolution is presented, followed by an explanation of how a helmet system works under an impact to protect the head of the user. After explaining helmet functions, the helmet main components design (shell and liner designs) influence on the helmet behaviour under impact is discussed, from geometry to materials, their properties and thicknesses. A similar but less extended analysis is done for the rest of helmet components.

The roles of the main motorcycle helmets standards namely helmets design, manufacture and test are explained, and a summarized comparison between them is done. The standards reviewed are ECE R22.05, Snell M2010, DOT FMVSS 218 and BSI 6658. In current helmet standards tests no rotational effects are measured in the headform, despite the fact that the most frequent severe injuries in motorcycle crashes are head injuries mainly caused by rotational forces that are most commonly generated as a result of oblique impacts. Proposed oblique impact tests and rigs are analysed as well as new motorcycle helmet solutions designed to reduce the rotational acceleration that reaches the head.

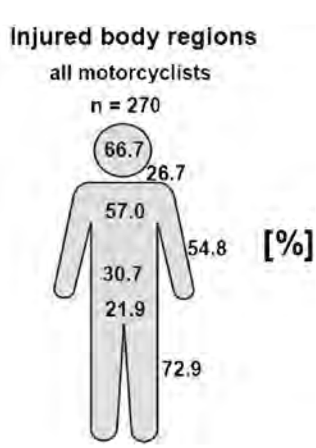

Fig. 1. Injured body regions of motorcyclists (COST, 2001).

Fig. 2. Ancient Greek Corinthian bronze helmet – 5th century B.C. (The Greek Gold).

In the end, the Finite Element Method is presented as a powerful tool. Finite Element Analysis is used to investigate and optimize helmets and it is possible to vary and study a great number of parameters using the experimental procedure. Thus, once a functioning and validated numerical helmet model is created, a great variety of information can be obtained.

2. Motorcycle helmet

2.1. Origins

Helmets have been used as a primary form of protection for a long time, by protecting the head against weapons' strikes and any kind of penetration. Thus, the primary helmet's function was the reduction of head injury mainly in combats. An example is the helmet represented in Fig. 2. Following the evolution of societies, the materials and the construction techniques used in helmet's manufacture became more advanced. Moreover, helmets evolved and diversified with the emerging of new needs of head protection against any kind of impact.

In the early 1900s, with the widespread introduction of the motorcycle, the need of a crash helmet arises. Initially, motorcycle helmets were no more than leather bonnets, first used in racing and usually worn with goggles. These skull caps were adapted from earlier aviators whose main goal was to keep the head comfort and so almost no protection was provided to the head. Thus, the problem of the non-existence of a crash helmet persisted. One example is shown in Fig. 3.

From this point, helmets evolved based on the understanding of what a helmet should do, in other words, the understanding of the biophysical characteristics of the head and the development of kinematic head injury assessment functions (Newman, 2005). Therefore, it was realized that a hard outer shell was needed to distribute the applied force and thereby reduce the localization of the impact load, improving the force distribution and thus, diminishing the likelihood of skull fracture. Moreover, the evolution of materials science was also crucial to helmets evolution.

The concept of a hard shell dates back to ancient Greek time, as shown in Fig. 2. However, as already referred, the first bonnets used as motorcycle helmets had no hard shell. After these bonnets, a

Fig. 3. Leather bonnet (P&K Military Antiques).

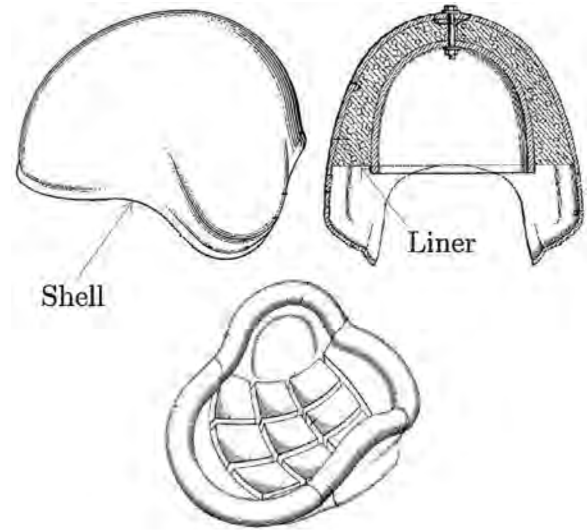

Fig. 4. Roth's and Lombard's crash helmet.
Adapted from Roth and Lombard (1953).

helmet that was constituted by some individual hard leather pieces was created, usually sewn to a hard fibre material crown section and lined with felt or fleece which a few years later was replaced by an inner suspension. This new feature increased the capacity to absorb and distribute impact's energy more effectively than the previous ones (Newman, 2005). This new device was the solution to the need of introducing a good absorbing impact energy material to reduce the inertial loading on the head and thus, reduce the probability of injuries due to induced accelerations.

In the early 1930s, the first hard shell of modern motorcycle helmets was constructed and it was made of several layers of cardboard glued and later it was constructed by impregnating linen with varnish resins, which allowed the cure into the desired solid shape (Newman, 2005). In 1939, the first helmet with moulded plastic shell was introduced by Riddell and it was used to practise football. Although the application is not the same, there was almost no difference between these two types of helmets until the middle of the 20th century, when it was recognized that, in the case of motorcyclists, they deal with one-time life threatening blow that can occur easily in a fall (Newman, 2005).

A few years later, Holbourn (1943, 1945), performed an important study where it was understood that non-penetrating head injuries are caused by short-duration accelerations acting on the head and its contents. These acceleration injuries are the most common and dangerous form of injuries for motorcyclists and are often caused by blunt impact rather than by penetration (van den Bosch, 2006).

Turner and Havey (1953) introduced the padding of modern helmets, which consisted in resilient closed cell rubber foam that was placed in the interior of the shell to dissipate impact energy effectively. However, this design was very heavy. At the same time, Roth and Lombard (1953) presented modern helmet as it is known today, represented in Fig. 4. Its hard shell was constructed by 4 layers of fibre glass and several materials were used as padding material, such as expanded polystyrene (EPS) foam or polyurethane (PU) foam. At first, PU foam was mostly used, but due to better properties of EPS foam (cheap, readily available, relatively easy to manufacture and a good crushable energy absorbing material), it is still the currently most used material as foam liner.

One year later, a new helmet covering more head area was created, as observed in Fig. 5. At the time, this helmet was believed to be among the most protective helmets ever designed. More details about the history of motorcycle helmet can be found in Newman (2005).

From here, helmets evolution was restricted through the creation of standards. Snively (1957), founder of the Snell Memorial Foundation, had a profound impact on modern helmet design and performance by showing that the only helmet that did not allow a life threatening skull fracture was the helmet made by Lombard and Roth's company, the only helmet that had EPS as padding material. Other important studies on the evolution of motorcycles helmets were Cairns (1941), Cairns (1946), where in the last one, Cairns concluded that there was a need of adoption of a crash helmet standard to compel all civilian motorcyclists to wear helmets, which would result in a considerable number of lives saved.

Currently, modern helmets are capable of distributing the impact loading over a large area of the head and reducing the total force on the motorcyclist's head as much as possible. Besides that, modern helmets developed for motorcycles are able to resist to very strong impacts and have helped the human head to become less and less vulnerable (Nemirovsky and van Rooij, 2010). Nevertheless some substantial improvements are still possible (COST, 2001).

However, the evolution of actual helmets is not side by side with the evolution of the understanding of head injury mechanisms, but follows the evolution of standards, which means that if a standard is outdated, nothing requires improvements in helmets. Thus, an improved standard means improvements in helmets (COST, 2001). Newman (2005) highlighted these same issues, such as the lack of progress, the actual use of old fashioned test methods that do

Fig. 5. One of the first open face Bell helmets, the 500-TX (Bell helmets).

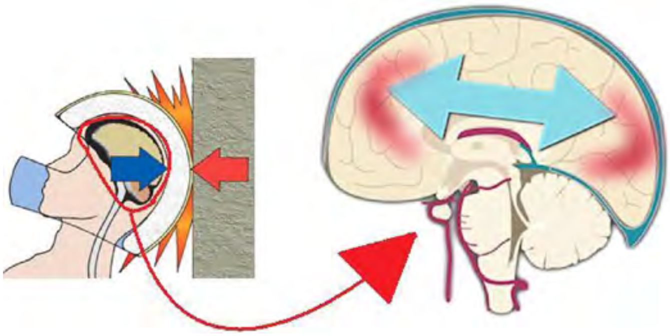

Fig. 6. Closed head injury.
Adapted from: Arai helmet (2011) and Brain Injury Association of America (2011).

not properly reflect the real life circumstances of accidents, like the biofidelity of the headform, the nature of the failure criteria, as well as the manner by which the movement of a test helmet is constrained.

2.2. Function

A motorcycle helmet is the most common and best protective headgear to prevent head injuries caused by direct cranial impact (Chang et al., 2003).

Primarily, the helmet purpose is understood as head protection against skull fractures, and modern helmets are usually efficient in this sense. Another main purpose of motorcycle helmets is the prevention of brain injury, since brain injuries are often very severe and result in permanent disability or even death. Thus, the purpose of protective helmets is to prevent head injury by decreasing the amount of impact energy that reaches the head, reducing the severity or probability of injury (Deck et al., 2003a; Liu et al., 2003).

Besides protecting the head in motorcycle crashes, helmets keep the head comfort by cutting down the wind noise and acting like a shield against wind blasts, bad weather conditions and any kind of object. They protect the head in case of accident by absorbing the impact and cushion the head to extend the time of impact. In order to have a perception of how a helmet behaves during an impact it is necessary to understand all the mechanisms involved.

Helmets can be divided into two major parts depending on the main role of each one. There is a hard outer shell that distributes the impact force on a wider foam area reducing the localization of the impact load, increasing the foam liner energy absorption capacity and consequently reducing the total force that reaches the head and the likelihood of injuries like skull fractures (Shuaeib et al., 2002b). Besides resistance to penetration, the helmet is the initial shock absorber in an accident (Liu et al., 2003). The other main part is the inner liner, which is generally made of an excellent absorbing impact energy material to reduce the inertial loading on the head (by slowly collapsing under impact) and thus, reducing the likeliness of injuries, especially brain injuries, due to induced accelerations. An example of these type injuries is the closed head injury type which is the most common type of head injury in motorcycle accidents: the skull is not fractured but the great head acceleration may cause brain injuries due to the relative movement of the brain inside the skull. For example, when an impact to the back of the head occurs, the brain moves forward inside the skull, squeezing the tissue near the impact site and stretching the tissue on the opposite side of the head. Successively, brain rebounds in the opposite direction, stretching the tissue near the impact site and squeezing the tissue on the other side of the head. Fig. 6 shows the mechanism behind the closed head injury as explained above.

This brain behaviour is explained by its consistency, which permits a movement inside the skull, within the cerebrospinal fluid (CSF). In this sense, when an impact occurs and the helmet's energy absorption capacity is not enough, the skull stops suddenly but the brain keeps the movement – due to inertia – until colliding against the skull's interior. From these collisions and other relative motions of the brain, it may occur severe injuries such as shearing of the brain tissue to bleeding in the brain, or between it and the dura mater, or even between the dura mater and the skull. This bleeding and consequent inflammation causes brain swelling, causing harder pressure against the inside part of the skull and more damage to vital regions.

2.3. Design

The helmet's mechanical response during an impact is mostly affected by its design (Aare, 2003).From what was reviewed in Section 2.2, liner softness and thickness are important variables so that the head can decelerate at a mild rate as it crushes the liner during the impact. Thicker foams remain in the plateau regime of the stress-strain curve for longer compression lengths (Kim et al., 1997). However, a helmet cannot be too thick due to practical and aesthetic constraints (Shuaeib et al., 2002b), which brings implications in the softness of the liner, so its thickness (typically between 20 and 50 mm) is limited by comfort and shape constraints (Yettram et al., 1994). In addition, the use of a thicker liner increases both the volume and mass of the helmet, with obvious disadvantages with respect to loading of the cervical spine (Huang, 1999; Huston and Sears, 1981).

Shuaeib et al. (2007) indicated foam density and foam thickness as the most contributing factors in preventing head injury. Therefore, it is important to find the perfect balance between the softness and the thickness of the inner liner, taking into account their limits. For example, when the liner is too soft the head may crush it completely upon impact and since outside the liner is the hard shell, the head suddenly stops, which results in high accelerations induced to the brain causing brain injury. On the contrary, if impact speed is lower than the one for which it was designed, the head will be decelerated a little more abruptly than was actually necessary given the available space between the inside and the outside of the helmet. Thus, an ideal helmet liner is stiff enough to decelerate the impact to the head in a smooth and uniform manner just before it completely crushes the liner. However, the required stiffness depends greatly on the impact speed of the head (Chang et al., 2000, 2003; Gilchrist and Mills, 1994a; Mills and Gilchrist, 1991; Yettram et al., 1994) and also in criteria used to optimize the protective padding liner (van den Bosch, 2006). Shortly, the best protection guaranteed by a helmet is for the impact speed which it was designed for. Mills (1993) carried out a simple mathematical approach about helmet foam liner thickness design based on impact velocity.

Thus, one of the issues of helmet's design is the doubt of how strong a helmet should be to provide the best possible protection, where the shell stiffness and the liner density are important parameters.

In practice, motorcycle helmet manufacturers design the helmets based on the speed used in energy absorbing tests in order to meet the specifications set out in standards. However, this is a costly choice (Mills, 1993, 1994; Miyajima and Kitahara, 1999; Shuaeib et al., 2002b). For example, the energy absorbing based on ECE R 22.05 are done at the velocity of 7.5 m/s. Richter et al. (2001) reported that the range of the most common head impact speed in real crashes is 5.83–8.33 m/s, which means that helmets are currently designed to the most common impact speed reported. Mills (2007) agree that real crashes occur at a range of impact velocities, most frequently at relatively low velocities, and helmets cannot prevent all injuries, as some crashes are too severe for any wearable helmet. Bourdet et al. (2012) reported that current motorcycle helmets are very effective for moderate speed impacts, but its protection reaches its limits at higher energies, where helmet deformation reaches its limits. This is supported by the analysis conducted in the COST 327 project (COST, 2001), which shows that serious injuries occur at impact speeds above 13.89 m/s, almost the double of those considered on standard tests. Bourdet et al. (2012) and Mellor and StClair (2005) postulated that if helmets could be made to absorb more energy, the number of injuries and its severity can be reduced.

Furthermore, it is shown by van den Bosch (2006) that the optimal protective padding liner density depends on the impact site, where the protective padding liner density should be lower for the front and rear regions and should be higher for top region impact. Gilchrist and Mills (1994a) demonstrated that shell geometry has influence on the shell stiffness, as helmet shells are stiffer when loaded at the crown, since that site has a double-convex curvature and is distant from any free edges. Hence, the soft liner should be located in the crown region with the objective of compensating high shell stiffness and attempting to make helmet impact response site-independent (Mills et al., 2009). Besides geometry, the exterior finish of the shell is also important, influencing the friction against the impact surface, which has a tremendous effect on the rotational acceleration (Halldin et al., 2001; Mellor and StClair, 2005; Phillips, 2004).

Therefore, motorcycle helmets' design is affected by the requirements of each standard, which is reported in several studies, such as those performed by Chang et al. (1999b), Gilchrist and Mills (1987), Hopes and Chinn (1989), Kostopoulos et al. (2002) and Yettram et al. (1994) which showed the influence of some standards requirements (such as the penetration test in Snell M2010 (Snell, 2010) and in BSI 6658 (BS, 1985)), forcing helmets to be designed with an enough stiffer shell to pass the test, leading to higher accelerations values. In fact, this could result in a helmet with a thicker shell that typically weights about 6–8 times more as compared to the foam liner (Shuaeib et al., 2002b). A motorcycle helmet shell is typically 3–5 mm thick, for the current materials used (Mills, 2007). This aspect was also criticized by Hume et al. (1995), since the frequency of motorcycle accidents involving sharp objects is extremely small, and this test causes the outer shell of the helmet to be excessively thick, which results in a heavy helmet. Otte et al. (1997) conducted a statistical study and his findings supported the conclusions of Hume et al. (1995). Mills (2007) concluded exactly the same. However, some standards do not require this type of test, such as ECE R22.05 (ECE Regulation, 2002).

Helmet improvement is also achieved by defining an adequate material behaviour (Bourdet et al., 2012). The force generated when a helmeted head strikes something, or as the head strikes a padded surface, depends on the crushing characteristics of the impacted material (Zellmer, 1993) and also on the material strength and loaded area size. Therefore, one of the primary objectives of a good helmet design is to maximize the padding area that can interact with the head during impact.

Statistical results pointed out that helmets are effective in reducing fatalities and severe injuries (Shuaeib et al., 2002a). However, the injuries that result from accelerations or decelerations are still a problem, mostly the rotational acceleration that remains underestimated (Johnson, 2000; Richter et al., 2001; Willinger and Baumgartner, 2003a,b), especially by main helmet standards. Nowadays, some researchers criticize this aspect in standards and also some of their outdated requirements. In helmet optimization studies, Deck et al. (2003a) and Deck and Willinger (2006) affirmed that nowadays helmets are designed to reduce headform deceleration or, in other words, helmets are designed to pass the standard requirements and not optimized to reduce head injury. Thus, there are still needs of improvements respecting helmet design. A helmet designer must have a thorough and comprehensive understanding of head impact biomechanics and a helmet should be defined in terms of how it should function rather than how it was styled or manufactured. The main way by which biomechanics has influenced helmet design is not so much in the understanding of different injury mechanisms, but rather in a better appreciation of biophysical characteristics of the head and the development of kinematic head injury assessment functions. This insight has provided better ways to test the impact capabilities of a helmet without first placing it on a human being and a means to judge how well one might expect it to work in actual use (Newman, 2005).

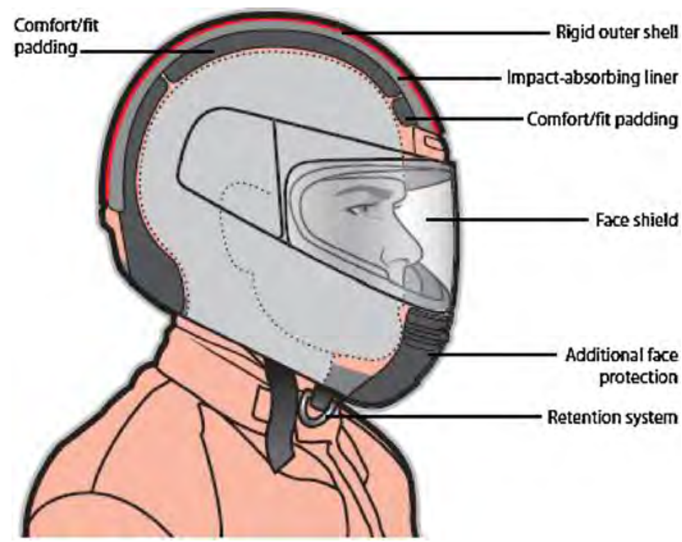

Fig. 7. Helmet components – basic construction (MSF, 2002).

Nevertheless, what a helmet designer normally changes to affect helmet response is foam thickness, foam material and shell material (DeMarco et al., 2010).

Recently, Post et al. (2012) carried out a study of impacts on football helmets, where it was concluded that it is possible to influence strains incurred by the brain using design characteristics, which shows the importance of helmet's design.

2.4. Components and materials

A typical modern motorcycle helmet is composed by six basic components:

- a very thin and hard outer shell,
- a thick and soft impact-absorbing inner liner,
- a comfort padding,
- a retention system,
- a visor,
- a ventilation system.

These and other components are represented in Fig. 7 and their functions in Fig. 8.

2.4.1. Outer shell

In general, the hard outer shell is made from thermoplastic materials such as polycarbonate (PC) or acrylonitrile–butadiene–styrene (ABS), or even by composite materials such as fibre reinforced plastics (FRP) like glass reinforced plastic (GRP) or carbon reinforced plastic (CRP) or just carbon fibre or Kevlar®. The shells made of thermoplastics materials are isotropic while the FRP shells show an anisotropic material behaviour in the plane of the shell (Mills and Gilchrist, 1992). The most common FRP is GRP, which consists typically in epoxy resin reinforced with glass fibre. Commonly, thermoplastic shells are

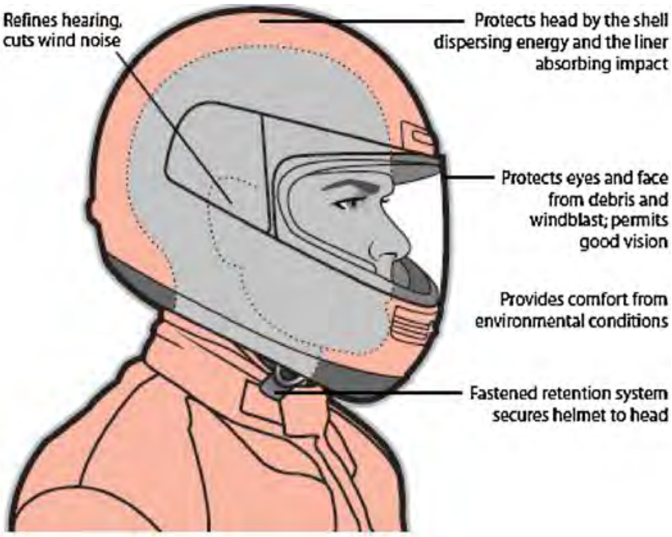

Fig. 8. Helmet components – protective/comfort functions (MSF, 2002).

relatively cheap when compared with composite ones. However, the GPR is a relatively low cost with fairly good mechanical performance (Tinard et al., 2012a). Carbon fibre and Kevlar are normally used for the most advanced helmets (Cernicchi et al., 2008).

The outer shell is responsible for:

- spreading the impact load over a large area of the helmet, therefore reducing the concentrated stresses during an impact that reaches the head and increasing the amount of energy absorbed, by having a larger area of effective energy absorbing liner;
- prevent helmet penetration by a pointed or a sharp object that might otherwise puncture the skull;
- provide a structure to the inner liner so it does not disintegrate upon abrasive contact with pavement or other impacting surfaces. This is important because the foams used as liner materials have very little resistance to penetration and abrasion as showed by Richter et al. (2001). All the helmets that showed damage in the internal lining also had a cracked shell, which increases the risk of injury (Beusenberg and Happee, 1993; Vallee et al., 1984). Thus, it can be stated that one of the shell's primary roles is to provide integrity against multiple impacts, what makes it an indispensable helmet component. Also, if protects the foam against abrasion, it also protects the head.
- absorbing the initial shock in an accident. However, just a little amount of energy is absorbed. From the literature, there are several values determined, such as 30% of the total impact energy (Mills, 1995), 10–30% of the total energy (Gilchrist and Mills, 1994a,b; Mills and Gilchrist, 1991), 12–15% in a study performed by Ghajari et al. (2009b) and 34% of the total impact energy dissipated (Di Landro et al., 2002). This value is not consistent between the studies due to the differences in the tests, such as impact velocity, the materials and their properties, etc.

Helmet shells made from advanced composite materials are progressively substituting the thermoplastics ones. However, they are generally more expensive and still being evaluated in what concerns the real benefits against increased cost.

During an impact, when the liner foam crushes completely, the unabsorbed energy will be transferred to the head and the impact forces developed will be very high. These impact forces will be reduced if there is another mechanism that could absorb energy. One possibility is to have a outer shell that absorbs some additional energy during impact (Pinnoji and Mahajan, 2010). This fact makes

Fig. 9. Helmet deformation modes with FRP (left) and PC (right) shells (Beusenberg and Happee, 1993).

composite materials desirable for helmet's shell application, as composite shells may absorb energy through damage mechanisms such as fibre breakage, matrix cracking and delamination. The main advantage of using composite outer shells lies in their capability of absorbing more energy by rupture in comparison with thermoplastic outer shells. Thermoplastics shells can also absorb energy by both buckling and permanent plastic deformation (Cernicchi et al., 2008), however still a relative little amount compared to composite shells, if the energy absorption mechanisms that relies mainly to fibre breakage of the composite are activated.

The shell stiffness has an important influence in the overall dynamic performance of the helmet. The stiffness of FRP shells is higher than the stiffness of thermoplastic shells as demonstrated by Beusenberg and Happee (1993), by comparing both experimentally, where the stiff FRP outer shell showed only minor deformation, where the energy was predominantly absorbed by foam deformation, as shown in Fig. 9. For this reason, Brands (1996) considered FRP shells more preferable. Gilchrist and Mills (1994a) studied the deformation mechanisms of ABS and GRP and concluded the same as Beusenberg and Happee (1993), that composite shells deforms less than thermoplastic ones. However, it was also reported that the impact forces with fibre composite helmet shells are much greater than those with thermoplastic shells.

However, such behaviour cannot be achieved at low energy impacts, showing a dependence of composite shells on the impact velocity, which is greater compared to thermoplastic ones (Mills and Gilchrist, 1991). Also, composite shells are much stiffer, which could leads to substantial accelerations at low energy impacts because their energy absorbing capacity relies mainly to fibre breakage. On the other hand, at higher energy impacts, composite shells provide substantial protection to the motorcyclist due to the large amount of impact energy absorbed by the helmet system until its final failure (fibre fracture) (Kostopoulos et al., 2002). Therefore, at high energy impacts, composite shells are more effective. Gilchrist and Mills (1994a) also showed that to occur delamination it is necessary a great amount of impact energy, reporting also that the impact forces with fibre composite helmet shells are much greater than those with thermoplastic shells (in a flat anvil). Mellor and Dixon (1997) carried out experiments on GRP shell motorcycle helmets with various anvil shapes to investigate the impact characteristics. They concluded that GRP shell effectively spreads the load of anvils when they are of kerbstone and edge type compared to flat type.

On the other hand, at low energy impacts, a thermoplastic shell like a polycarbonate one might be more effective, having better protective characteristics with lower stiff shells, as demonstrated by Markopoulos et al. (1999). This finding is also present in other studies (Chang et al., 2000, 2003; Gilchrist and Mills, 1994a; Mills and Gilchrist, 1991; Yettram et al., 1994). Despite of delamination mechanism is responsible for a good amount of energy absorbed by helmets with composite shells, which make them particularly desirable, the thermoplastic-shelled helmets may actually perform better than for example – the fibreglass ones – because its higher flexibility may lead to higher deformation of the EPS liner, contrary to FRP that may crush and delaminate. The stiffer FRP outer shell is often used in combination with a low-density EPS foam, whereas the softer PC and ABS outer shells (relatively poor shock absorbing capacity) compensate their compliance with a stiffer, high-density EPS foam (van den Bosch, 2006).

Moreover, fibre-based materials had a much lower rate of fracturing, whereas plastic shells fractured more often and the rebound of a helmet with a thermoplastic shell is much higher than a fibreglass helmet, which makes the thermoplastic one less effective and thus less safe (Aare, 2003).

In order to assess the impact behaviour and better understand the energy absorbing mechanisms of composite shells, several finite element models of helmet with composite outer shell were proposed. The first one was proposed by Brands (1996), a simplified model in which the outer shell was composed of resin reinforced with glass fibres and aimed to explain the dynamical behaviour of a helmet during impact. The composite material was modelled with an elastic law considering no damage during impact and random orientation of fibres in the material, without delamination and thus, no realistic behaviour was reproduced by this model.

Kostopoulos et al. (2002), with a more realistic model that considered different composite layers, modelled with an elastoplastic law couple with rupture and delamination mechanisms, studied the influence of the complex behaviour of a composite shell on the helmet's shock absorption capability. From the results of this study, Kostopoulos et al. (2002) indicates that what makes composite materials ideal for production of safety helmets is the ability to sustain extensive damage without compromising the integrity structure. In the same study, Kostopoulos et al. (2002) also showed that composite shell systems exhibiting lower shear performance provide additional energy absorbing mechanisms and result in better crashworthiness behaviour. Thus, from different composite materials tested, Kevlar® shell was the one that exhibit longer impact duration and an associated lower peak acceleration value. In other words, Kevlar® fibre shell exhibits much higher absorbed energy and the energy absorbed by the foam liner was also higher. However, mechanical properties of the composite materials used in the study were based on the literature data and not on experimental tests.

Other models were proposed by Pinnoji and Mahajan (2006) and Mills et al. (2009), where the outer shell was made of resin reinforced with glass fibres. However, the outer shell modelled with an elastoplastic (Pinnoji and Mahajan, 2006) and an elastic (Mills et al., 2009) law respectively, did not take into account delamination or rupture.

Pinnoji and Mahajan (2010) performed a study with the aim of analysing damage and delamination mechanisms of different composite outer shells of a helmet during impact and compared the results with those obtained with ABS (Pinnoji et al., 2008b). The results showed that the energy absorbed by the composite shell in helmets during damage and delamination is smaller than the energy absorbed by the plastic deformation of ABS shell. The composite shell is stiffer as compared to ABS shell in the direction of impact and gives higher impact forces on the head. Nevertheless, the composite model proposed by the authors is based on the study of Kostopoulos et al. (2002) which was not validated.

The model proposed by Kostopoulos et al. (2002) was until recently the most advanced helmet finite element model with composite outer shell with a major drawback, the model has not been validated.

Recently it was performed a study, divided in three parts, by Tinard et al. (2011), Tinard et al. (2012a), Tinard et al. (2012b) where it was developed and validated a new finite element model of composite outer shell for a motorcyclist helmet and it was assessed and optimized regarding to biomechanical criteria.

In the first one, Tinard et al. (2011), it was proposed a realistic model of a composite outer shell of a commercial helmet based on experimental tests, such as modal analysis, to obtain the elastic and rupture properties of each layer, identifying the constitutive law of the composite material used. In the second one, Tinard et al. (2012a), the helmet model was validated against experimental data under normative conditions as prescribed by standard ECE 22.05 (ECE Regulation, 2002). Nevertheless, the delamination mechanism has not been considered, which is a drawback of this model due to the importance of delamination mechanism during the crash of composite materials (Kostopoulos et al., 2002; Tinard et al., 2012a). In the third and last one, Tinard et al. (2012b) evaluated the real injury risk sustained by a detailed and validated FE head model during impacts with the approved motorcycle helmet (Tinard et al., 2012a) optimizing it against biomechanical criteria rather than standards criteria. The results showed that even if a helmet passes the tests of shock absorption required by the standard ECE 22.05, injury risks remain high.

Other studies with models of helmets with composite outer shell coupled with a FE human head model were proposed by Pinnoji and Mahajan (2006), Pinnoji and Mahajan (2008) and Ghajari et al. (2009). However, the aim of these studies was not the shell.

Recently, Pinnoji et al. (2008a) tested the possibility of outer shells made of aluminium foams, which have high strength, light weight and good energy absorption capabilities. The aim of Pinnoji was reduce the helmet weight without changing its dynamic performance. As a result, it was observed that the resultant force on the head is less with metal foam shell and the helmet weight is reduced by 30% as compared to ABS helmet. The headform acceleration was also lower than the ABS outer shell. However, due to the permanent (plastic) deformation of metal foam, it might not behave well to a second impact in the same region. Though, this is only a possibility that was tested and motorcycle helmets market still is dominated by thermoplastics and fibre reinforced shells. More recently, further developments were done in this issue by Pinnoji et al. (2010). Different metal foam shell densities were tested and the helmet was validated. The ULP FE head model was also used to assess the helmet against biomechanical criteria. The better results were obtained with the metal foam shell of density 150 kg/m^3 which corresponds to approximately 73% reduction in mass compared to that of ABS shell and also the impact forces on the head were lower in both front and top impacts. Von Mises stresses in the brain were within the injury tolerance limits at 7.5 m/s impact velocity for all cases, except for ABS helmet in top impact. In front and top impacts, von Mises stresses in the brain were reduced by approximately 25% and 22%, respectively for helmet with low-density metal foam shell compared to the ABS helmet. It was also observed that the resultant force on head was less with lower density metal foam helmet as compared to the ABS helmet.

2.4.2. Inner liner

The purpose of the inner liner foam is to absorb the remaining force of the impact that was partially absorbed (a small amount of energy) and dispersed by the outer shell, by crushing during the impact and thereby increasing the distance and period of time over which the head stops, reducing its deceleration, absorbing most of the impact energy and so reducing the load transmitted to the head. In the studies performed by Deck et al. (2003a) and Deck and Willinger (2006), one of the conclusions was that the elastic limit of the foam used as inner liner has the most important influence on Head Injury Criterion (HIC) response but its Young's modulus has the most important influence on biomechanical head response. The liner density is also an important property because the yielding stress at which the foam crushes is directly related to it (Gibson and Ashby, 2001). Currently, the most common liner material in protective helmets is EPS foam, which is a synthetic cellular material with excellent shock absorbing properties and a convenient cost-benefit ratio (Di Landro et al., 2002), whose mass density applied in helmets varies from approximately 30 to 90 kg m^{-3} (Brands, 1996; van den Bosch, 2006). EPS absorbs the energy during the impact of the helmet through its ability to develop permanent deformation, by crushing (foam collapsing), providing the required protection to the motorcyclist. Again, the impact velocity is an important variable since the normal velocity component largely determines the amount of EPS liner crushing (Mills, 2007). It can be concluded that high-density EPS are able to absorb larger amounts of energy than low-density EPS can do, but transfer higher accelerations and forces localized at the impact point (Di Landro et al., 2002). Although this type of foam has an excellent first impact performance in case of a subsequent impact in the same area, the protection level offered by EPS would be minimal since the material deforms permanently without elastic recovery (Gilchrist and Mills, 1994b; Shuaeib et al., 2002b,c, 2007). Thus, its energy absorption capability is significantly decreased after a first impact, particularly in high energy impacts. This is one of the reasons why, if a helmet is damaged in an accident, it will have little protective value in the occurrence of a subsequent event (Liu et al., 2003). To overcome this issue, some materials were proposed, such as:

- expanded polypropylene foam (EPP) by Shuaeib et al. (2007);
- micro-agglomerate cork (MAC) by Alves de Sousa et al. (2012).

The EPP is very similar to the EPS, presenting similar peak accelerations and impact durations for a same helmet with EPS, as verified by Shuaeib et al. (2007). The micro-agglomerate cork has a good energy absorption capacity and high viscoelastic return and its capacity to keep absorbing energy is almost unchanged after the first impact, mainly due to its viscoelastic behaviour, which is a characteristic desired in multiple impact situations. This characteristic is also important for helmets approval by some well accepted standards that require a test with two impacts on the same helmet point, for example Snell M2010 (Snell, 2010). However, for the same volume of EPS, the MAC is a heavy solution, which is a problem for helmet approval and increases the risk of injury. In this sense, hybrid EPS/MAC paddings can give a better compromise (Coelho et al., 2013). However, Pedder (1993) found that multiple impacts do not occur on the same helmet site in crashes, occurring at different sites as helmets rotate between the impacts. Also, the ECE R22.05 standard does not demand double impacts to the same site as some of its previous versions, such as ECE R22.03 (Mills, 2007).

A closed cell foam based on Vinyl Nitrile Polymer was also tested as energy absorbing liner by Goel (2011). This material was selected because of its good energy absorbing characteristics. Over multiple impacts, ski helmets with the novel liner showed substantially lower peak accelerations compared to helmets with EPS liner. From the drop tests performed, the helmet with the new liner had around 20% less peak acceleration on first impact compared to common EPS liner. This difference increased after each impact, showing the good capacity of this material as an energy absorbing liner material. The reduction in foam thickness was also considerably lower compared to the standard case.

2.4.2.1. Novel configurations. In addition to new materials, several configurations have been proposed in order to enhance the energy absorption properties of motorcycle helmets, which can lead to an improvement of the safety levels provided by current commercial helmets.

Caserta et al. (2011) replaced part of the helmet's liner by layers of hexagonal aluminium honeycombs as reinforcement material to the energy absorbing liner of a commercial helmet, as shown in Fig. 10. The results showed that this new configuration provides better protection to the head from impacts against specific surfaces than the original EPS liner. Best results were obtained for impacts against the kerbstone anvil. However, the results obtained for impacts against the most impacted surface, the flat anvil, revealed some limitations and at some impact points the results were even worse than the commercial helmet.

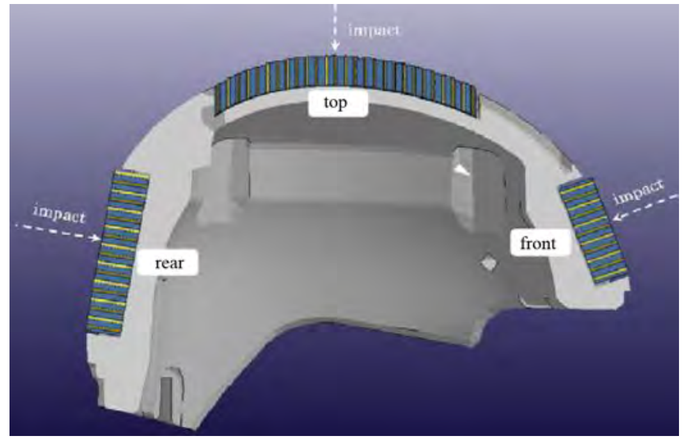

Fig. 10. Schematic section of the prototype liner proposed by Caserta et al. (2011).

Also, the thickness of the liner necessary to accommodate honeycomb layers is extremely limited, so in a real accident scenario excessively thin layers of EPS foam could be easily broken by the honeycombs during impact and thus, the honeycomb could penetrate the scalp causing head injuries.

Recently, Blanco et al. (2010) proposed an innovative helmet liner that consists of an ABS lamina with deformable cones in it, as shown in Fig. 11. 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. Experimental and numerical tests were performed and the model was validated. No optimization was done, leaving a gap for further improvement, but the results from the model validation show high accelerations induced to the head. Although this concept was developed to ski helmets, it could easily be applied to motorcycle helmets.

Other configuration is the cone-head shock absorbing foam liner, developed to absorb impact force more effectively and thus, protecting the head more effectively from intra-cranial injury. This concept proposed by Morgan (1993) consists in a motorcycle helmet foam liner made of two density layers, as shown in Fig. 12. The outer layer, which is the black part, is made of high-density foam and has truncated cones facing inwards. The inner layer, the grey one, which is close to the head, is made of softer low-density foam and has cones facing outwards.

Fig. 11. ABS cone liner proposed by Blanco et al. (2010).

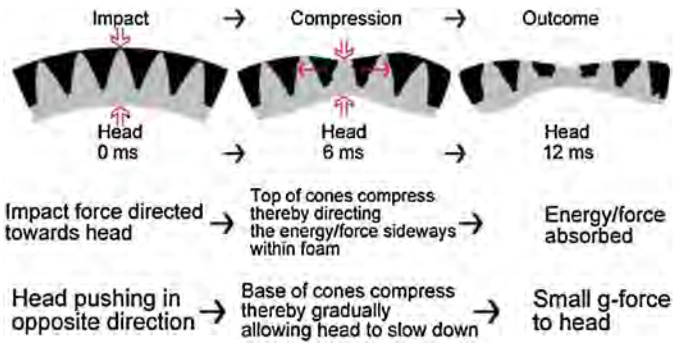

Fig. 12. The mechanism of cone-head compression liner (Morgan, 1993).

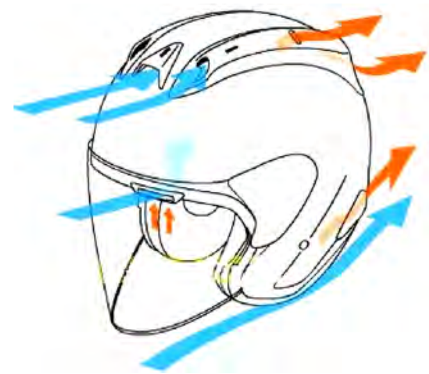

Fig. 13. Ventilation system (Helmet Boys).

When an impact occurs, the impact force pushes towards the head and causes the lower density cones to compress. The collapsing of the cones causes the energy to spread sideways within the thickness of the foam liner instead of towards the head. The dispersion of the energy sideways prevents the impact energy to be translated through until it reaches the brain and the area of effective energy absorbing liner is increased by this mechanism. As a result, the head causes the basis of low-density foam cones to compress. Also, the head will experience a gradual deceleration because of the crushing/compression of the cones, minimizing the energy induced to the head. The cones reduce the deceleration of the head and the impact time of interaction is longer or the head stopping time is longer. Hence there is a reduction in the forces translated across the thickness of the new shock absorbing liner to the skull.

This concept is the most promising from the ones presented and is already used in commercial helmets. This lighter liner helps also to reduce rotational acceleration of the head during impact.

2.4.3. Comfort liner

The comfort padding consists in sufficiently firm foam covered by a fabric layer that contacts and surrounds the head. This inner comfort foam is generally made of soft and flexible foams with low density as open-cell PU or polyvinyl chloride (PVC) (Brands, 1996; Chang et al., 2003; Gilchrist and Mills, 1993; Mills, 2007; van den Bosch, 2006). It keeps the comfort and the adequate helmet fitting by distributing the static contact forces (Gilchrist et al., 1988; Gilchrist and Mills, 1993; van den Bosch, 1998). The static contact force distribution is important to avoid headaches (Gilchrist et al., 1988). Other materials have been proposed, such as wool and Tunisian alpha fibre used by Taher Halimi et al. (2012) to develop a novel comfort liner for a motorcycle helmet. According to the authors, this new design with these natural fibres and phase change material improve sweat absorption and perception of thermal comfort. In other words, it can facilitate breathability and evaporative transfer of heat in the safety helmet, increasing comfort and well-being.

As a result of the low stiffness, the comfort foam does not contribute significantly to the energy absorbing properties as it crushes completely without absorbing any relevant amount of energy and, therefore, has no injury reducing effect (Beusenberg and Happee, 1993; Cernicchi et al., 2008). Manufacturers generally produce different sizes for every model adding different thicknesses of comfort liner to two different sizes of shell and energy-absorbing liner. This is important, as showed by Chang et al. (2001), that assessed the effect of the fit between the head and the energy-absorbing liner and concluded that the fitting influences the acceleration induced to the head.

It is rare to find a study where these features were modelled with success. For example, none of these studies modelled the comfort foam (Khalil et al., 1974; Köstner and Stöcker, 1987; van Schalkwijk, 1993; Yettram et al., 1994; Liu et al., 1997, 1998; Liu and Fan, 1998). Brands (1996) has modelled the comfort liner and concluded that it is significant for the shape of the headform acceleration peak, however, due to the low stiffness of this liner in the model that causes problems with numerical stability, it was removed. Also, Pinnoji and Mahajan (2010) affirmed that this foam is often very soft and thin to absorb energy and is used only for fitting different head sizes, not having influence on headform response during an impact (Tinard et al., 2012a,b). Therefore, the majority of the cases do not model this foam because it is difficult to model and from a cost-benefit point of view, the improvements in the fitting of the helmet on the head are not worth it.

2.4.4. Retention system

The retention system or chin strap keeps the helmet attached to the head all the time. However, there are records of a considerable number of roll off helmets even with the chin strap intact and closed Richter et al. (2001) after a crash, leaving the head unprotected from any following impact. All types of helmets have a retention system and the chin-strap is usually made of polyethylene terephthalate (PET) or nylon. The retention system generally consists of a strap bolted to each side of the outer shell. Mills et al. (2009) concluded that chin-straps and also the foam inside the chin bar affect helmet rotation on the head.

2.4.5. Visor

The visor is made of a strong and transparent material like PC and is equipped with water and scratch proof coating to protect the face from any object that impacts that region and from weather conditions and also to provide a clear vision.

2.4.6. Ventilation system

The ventilation system ensures that fresh air is conducted into the helmet and exhaled air and humidity are vented out, decreasing temperature inside the helmet. A ventilation system is represented in Fig. 13. Besides having a multi-impact protection performance, the EPP foam is a resilient material, which is pointed by Shuaeib et al. (2007) as a material that has potential as liner material for ventilation system improvement because its resiliency allows for the ease of ventilation holes and channels moulding without the foam breakage at the stage of mould extraction. Moreover, EPS foam is brittle in its nature, which makes the introduction of ventilation channels in the foam more difficult. A study performed by Pinnoji and Mahajan (2006) indicates that the ventilation channels grooved in liner foam are not detrimental to the dynamic performance of the two-wheeler helmet.

A complete description of the manufacturing process of each motorcycle helmet component can be found in Shuaeib et al. (2002c).

Fig. 14. Full face helmet by CMS helmets (CMS helmets, 2011).

Fig. 15. Modular helmet by CMS.

2.5. Types of helmets

Nowadays, there are several configurations of helmets available in the market, which can be classified into four basic types of helmets for motorcyclists. From the most to the least protective, helmet types are:

- Full face helmet;
- Modular helmet (also known as "flip-up" helmet);
- Open face helmet (also known as "three-quarters" helmet);
- Half helmet.

2.5.1. Full face

Full face motorcycle helmets are by far the most common type of helmet (MAIDS, 2004), being the most worn type of helmet (Richter et al., 2001). A full face helmet covers the entire head, with a rear that covers the rear of the skull at the top of the neck, and a protective section along the cheekbones to encompass the jaw and the chin, denominated chin bar. The fact that full face helmets cover the entire head means that they are the safest option among all types of helmets since they reduce the risk of head injury providing extra strength around the entire skull. Fig. 14 shows a full face helmet.

However, the fact that full face helmets involve the entire head has some disadvantages like the increased interior heat, the sense of isolation and the reduced peripheral vision. Also, they are one of the heavier types of motorcycle helmets due to the padded interior and mainly due to the shell, which covers a larger area compared to other types of helmets. This aspect can be detrimental in a crash because it can cause injuries on the neck and on the brain due to acceleration or just increase neck fatigue in an ordinary ride (Huang, 1999; Huston and Sears, 1981).

Nevertheless, the COST 327 final report (COST, 2001) and Richter et al. (2001) showed that 15.4% and 16% respectively of total helmet damages were located at the chin guard, which shows the important protection offered by full face helmets at this area. Otte (1991) concluded that impacts on the face and jaw areas are common in motorcycle crashes. In addition, Chang et al. (1999a), Chang et al. (2000), concluded that the chin bar provided by this type of helmets offers an essential protection and that the energy-absorbing capability of them could be improved by the introduction of the energy-absorbing liner in this area, plus the comfort liner. Actually, the chin bar contains a rigid foam to absorb energy Mills (2007). Also, Mills et al. (2009) in the case of frontal impacts concluded that the chin bar foam came into play, protecting the face (Mills, 1996). Thus, wearing a helmet with less coverage eliminates that protection and so the less coverage the helmet offers the less protection is provided to motorcyclist's head. Mills (2007) emphasizes that chin bar prevents the lower part of the forehead and temple being struck as the helmet rotates.

According to the COST 327 final report (COST, 2001) and Aare (2003), full-face helmet offer better protection than the others to the entire head. However, Shuaeib et al. (2002a) alerted that the extent of coverage in helmets like these might lead to helmets with weaker lateral protection represented in helmets with thin shells at the sides, which may constitute a weak point on helmet lateral protection. Also, the side is the weaker area as compared to other helmet areas due to edge flexibility resulting from lower stiffness associated to the larger shell curvature at the edge. The impacts to temporal regions are an important issue in real motorcyclist accidents because impacts to this region represent a considerable number of total impacts (39.5%, 12.8% and 18.3%, respectively (Hopes and Chinn, 1989; Hurt et al., 1981; Otte et al., 1997)) and the side of the skull represents the weaker area as regarding human tolerance for skull fracture due to the lower skull thickness at this region.

2.5.2. Modular helmet

A modular helmet is basically a combination between full face and open face helmets. It combines the safety of full face helmets with the openness of open face helmets. When fully assembled and closed, it resembles full face helmet by having a chin-bar for absorbing impacts on that area. Its chin-bar may be pivoted upwards to allow access to the face, as in an open face helmet, which is a great advantage in terms of comfort and practicability, as shown in Fig. 15. However, this same mechanism makes this type of helmets the heaviest type.

Although modular helmets do look the same as full face helmets, even when the front is down, they might offer a little less protection in the chin area. Nevertheless, there are not wide scientific studies that assess the protective capacity of the pivoting or removable chin bar of modular helmets. Thus, the doubt of how protective this helmet is, is still an issue, leaving an opportunity for future work. The actual state of the standards contributes somehow to this. The DOT standard does not require chin bar testing. The ECE 22.05 allows the certification of modular helmets with or without chin bar tests, since it is only indicated if the helmet protects or not the chin area. However, the Snell tests helmet's chin bar, and modular helmets are not an exception. Recently, Snell certified a modular helmet for the first time, the Zeus ZS-3000, in 2009 (Web Bike World).

2.5.3. Open face helmet

The open face helmet covers almost the entire head, except part of the face (especially the lower chin-bar), leaving this area unprotected, as shown in Fig. 16. Thus, an open face helmet provides the same protection as a full face helmet, except when the impact is to the face (COST, 2001), even in non-crash events like eye injuries due to the fact that some of these helmets do not have a visor to protect the users from dust.

Hitosugi et al. (2004) observed that people with open-face helmets were significantly more likely to have sustained severe head

Fig. 16. Open face helmet by CMS.

injuries, especially brain contusions, than people with full-face helmets.

2.5.4. Half helmet

Half helmet has essentially the same front design as an open face helmet but without a lowered rear in the shape of a bowl, as shown in Fig. 17. The half helmet barely provides the minimum coverage generally allowed by some standards, by covering only the top half of the cranium, and offers no protection for the face from the ears down. This issue is also highlighted by Shuaeib et al. (2002a), where the half-shell helmet is considered the most vulnerable to impacts at lateral and back head regions. Thus, a half-shell helmet offers less protection simply because it covers less area and also does not contain much padding, absorbing less energy. In addition, this type of helmets is known to come off of the motorcyclist's head in some accidents which, allied with all other factors that proves the inferiority of these helmets, led to the prohibition of the use of half helmets in some countries (DeMarco et al., 2010). A recent study evaluated the effectiveness of different styles of helmets, including half-coverage, open-face and full-face (Yu et al., 2011). The riders involved in crashes wearing half helmets were twice more likely to have head injuries than riders wearing full face helmets or even open face helmets.

3. Helmet safety standards

Motorcycle helmet standards were created after the widespread introduction of motorcycles. The first biomechanical studies suggested that use of motorcycle helmets should be mandatory due to the significative increase on head protection (Cairns, 1941, 1946; Snively, 1957).

Helmet standards have been established in many countries to evaluate the protective performance of helmets against head injuries. Some standards are regulated by governments, like in Europe and North America, but in other countries they are issued by private organizations. Almost all standards are different from each other, but similar in their primary goal: assessing the helmet

Fig. 17. Half helmet (Smith Family Powersports).

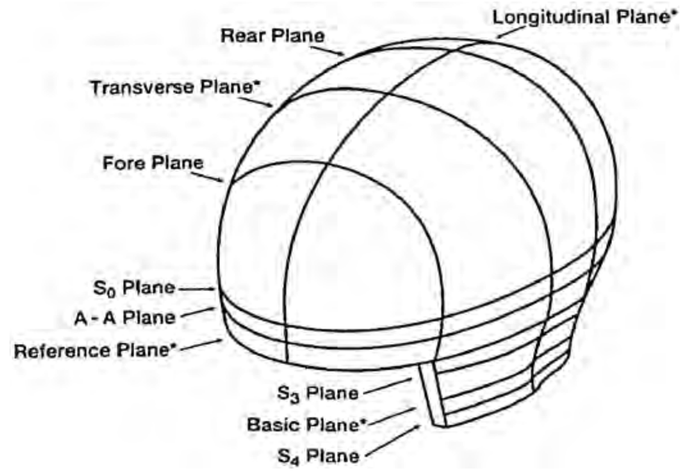

Fig. 18. ISO Head form – ISO DIS 6220-1983 (International Standards Organisation, 1983).

impact energy absorbing capability. These standards prescribe a number of tests to ensure that a helmet satisfies safety requirements.

Standards also evaluate parameters like comfort, ventilation, weight, fit, cost, appearance and availability. Because it is impossible to create a helmet for all impact conditions, designers are challenged to create a helmet capable to resist to the higher number of possible situations.

In fact, all motorcycle helmets nowadays available in the market were designed, manufactured, and tested to meet standards. Therefore, the performance tests required by any standard eventually influence helmet design. Nevertheless, none of the standards are able to precisely replicate the threats that a motorcyclist may experience in a crash. This is justified by the need of reliability and repeatability in the testing environment.

Nowadays, it is well known that helmets substantially reduce head injury, being safer to a motorcyclist to wear a helmet rather than none. Nonetheless, today helmets are designed to reduce headform deceleration and not optimized to reduce head injury (Aare et al., 2003; Deck et al., 2003a; Forero Rueda et al., 2011; Kleiven, 2007; Tinard et al., 2012b). Gimbel and Hoshizaki (2008) carried out a study where the authors concluded that headform mass played a significant role in the helmet materials performance. The mass and size of the headforms specified by the standards are nearly the same. For instance, the dimensions of the ECE R22.05 and the latest version of Snell (Snell, 2010) headforms are based on the ISO-DIS-6220 standard (International Standards Organisation, 1983) (Fig. 18), with their mass increasing with their size. Their dimensions are given by each standard. These standards test headforms comprises the entire head rather than the partial headform employed by DOT FMVSS-218 (U.S. Department of Transportation, 2012).

The head impact speed is another important variable in helmet impact study. Current standards impact speeds range up to 7.75 m/s although higher velocities are achieved riding a motorcycle. Nevertheless, the perpendicular impact speed of the helmet is usually not the same as the riding speed. When a motorcyclist falls, the impact is commonly oblique, which means that the impact speed is decomposed into two components, perpendicular and tangential to the road surface. The range of impact speeds used in motorcycle helmet standards in their energy absorbing tests aims to include velocities that are more common in real life (Richter et al., 2001). But it is also worthy referring that the tangential component is not assessed by current standards.

Table 2
Overview of motorcycle helmet standard tests.

Standard	ECE R22.05	Snell M2010	DOT FMVSS 218	BSI 6658
Impact	×	×	×	×
Penetration		×	×	
Retention	×	×	×	×
Roll off	×	×		×
Rigidity test	×			
Friction test	×			×

In summary, no helmet designed for a particular standard or standards can provide the maximum protection in all types of crashes and no helmet can protect the rider against all impacts.

3.1. Standards comparison

Almost all the standards follow the same concepts in evaluating the effectiveness of the helmets during accidents, which are:

- the helmet has to be able to absorb enough impact energy;
- it has to remain on the head during the accident;
- it must resist to penetration by sharp objects.

European motorcyclists have to wear helmets that meet ECE 22.05 regulation even if in some cases this standard offer less protection than DOT or Snell. In order to find a solution for this problem, Snell engineers developed the Snell M2010 standard tempting to approach the DOT and the ECE 22.05 requirements. The last DOT update was also made in that sense. Nevertheless, the ECE 22.05 (ECE Regulation, 2002) rating is the most widespread helmet standard, required in over 50 countries worldwide (Pratellesi et al., 2011). Regarding the current and future developments, the motorcycle standard ECE 22.05, represents the state of the art in performance specifications, and that this fact is partly due to biomechanics considerations (Newman, 2005).

Similarities between standards are well accepted and useful for manufacturers that have the possibility to sell the same helmet in countries regulated by different standards, without deep design changes. However, differences are still visible and it is possible to have a helmet approved by one standard and rejected by another. An example is the double impact required by Snell M2010 and DOT against the single impact required by ECE 22.05. It can be argued that double impacts are not typical of accident events, but the requirement is an acceptable procedure which provides a margin of safety for the user (Thom et al., 1998).

A short summary of the tests performed from each standard is presented in Table 2.

Penetration tests have been criticized by Hume et al. (1995) since the frequency of motorcycle accidents involving pointed objects is extremely small and this test causes the outer shell of the helmet to be excessively thick leading to heavier helmets. Otte et al. (1997) conducted a statistical study and his findings supported the conclusions of Hume et al. (1995).

A comparison between current standards from the impact point of view is summarized in Table 3. The anvil typically used in current standards is the flat anvil. This can be justified by the fact of being the most common type of object found in real crashes (flat and rigid) (Gilchrist and Mills, 1994b; Shuaeib et al., 2002a; Vallee et al., 1984), usually the road surface. More anvils are used for test purposes, such as the kerbstone anvil (ECE R22.05), the hemispherical steel anvil (DOT FMVSS 218 and Snell M2010) and the edge anvil (Snell M2010).

The acceleration-based head injury criteria used by the standards to access the helmets performance in the impact absorption tests are explained in Sections 3.1.1 and 3.1.2. Nevertheless, the HIC and the Peak Linear Acceleration (PLA) remain as the only normative parameters used for helmet homologation in terms of protection against impacts. This means that no standard assess the rotational motion that a motorcyclist is subjected, neither the local tissue thresholds. However, the rotational acceleration occurs in all motorcyclists accidents (Johnson, 2000) and has a tremendous effect in brain injuries. Also, the current trend is to design helmets to pass the standards with no consideration from the biomechanical point of view (Shuaeib et al., 2002a; Tinard et al., 2012b). So, optimization based on biomechanical criteria (for example strain and stress based head injury criteria) is different than the optimization with HIC criterion, which is correlated with acceleration of a rigid headform's centre of mass as used for helmet's homologation.

A more detailed comparison between current standards can be found in Thom (2006). The current Snell standard is an updated version from the one used in that study.

Recently, Pratellesi et al. (2011) tested uncertainties that are related to the homologation procedure in ECE 22.05. Finite-element simulations that have been conducted revealed that the HIC value, which is relevant for the homologation of motorcycle helmets in Europe, is changed by up to 30% by testing uncertainties that are in total agreement with the corresponding homologation standard. This fact casts the credibility of those homologation standards into doubt as a deviation of up to 30% is certainly not within the tolerable range of a security issue.

3.1.1. Peak linear acceleration (PLA)

PLA is basically the maximum acceleration value measured at the centre of gravity of the headform during impact and used in most current standards. Usually, it is stated as a number multiplied by the gravitational acceleration constant g ($1\,g = 9.81\,\mathrm{m/s^2}$). This method ignores the duration of the impact. However, standards take into account the impact duration through the HIC and also limit the duration of the impact. Moreover, some studies present a

Table 3
Standards comparison.

Standard	M2010 (Size J headform) Velocity	DOT Velocity	BSI 6658 Velocity (flat or hemi anvil)	ECE 22.05 Velocity
Impact criteria				
1st impact	7.75 m/s	6.0 m/s	7.5 m/s or 7.0 m/s	7.5 m/s
2nd impact	6.78 m/s	5.2 m/s	7.0 m/s or 5.0 m/s	–

Standard	M2010 (Size J headform)	DOT	BSI 6658	ECE 22.05
Failure criteria				
Peak	275 g	400 g	300 g	275 g
150 g	–	4 ms	–	–
200 g	–	2 ms	–	–
HIC	–	–	–	2400

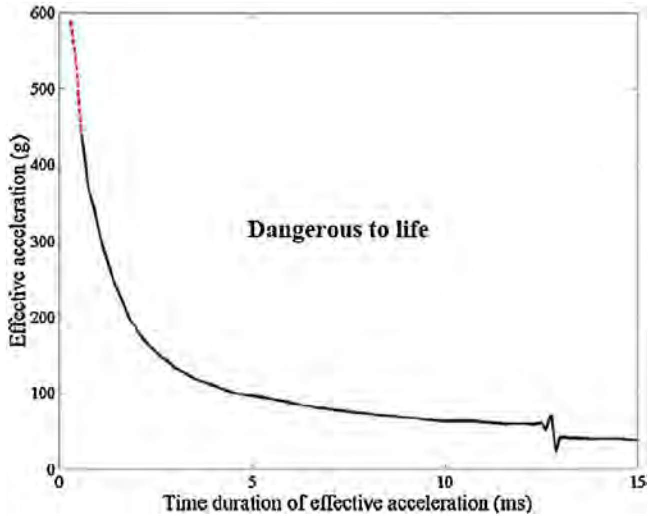

Fig. 19. The Wayne State Tolerance Curve (Kleiven, 2002).

limit of 80 *g* for a duration that shall not pass 3 ms (Got et al., 1978; Stalnaker et al., 1971; Versace, 1971) to not occur any type of head injury. Mertz et al. (1997) estimates a 5% risk of skull fractures for a peak acceleration of about 180 *g*, and a 40% risk of fractures for a peak acceleration of 250 *g*.

More recently, King et al. (2003), in a numerical study, estimated the MTBI (mild traumatic brain injury) tolerance for head linear acceleration, where there is a probability of MTBI occurrence of 25%, 50% and 75% for head linear acceleration of 559 m/s^2, 778 m/s^2 and 965 m/s^2, respectively. Other values were determined by Peng et al. (2012), predicting a 50% probability of AIS 2+and AIS 3+head injury risk for 116 *g* and 162 *g* respectively.

3.1.2. Head injury criterion (HIC)

The most commonly acknowledged and widely applied head injury criterion is the HIC, which is based on the assumption that the head linear acceleration is a valid indicator of head injury thresholds. However, it does not take into account head kinematics nor impact direction and rotational acceleration (Gennarelli et al., 1982; Newman, 1980; Ono et al., 1980), even though rotational acceleration is believed to be the cause of several head injuries as already referred. In consequence, the validity of HIC is intensively debated and there is reason to believe that safety developments could be made more efficiently by taking into account the effect of rotational kinematics into current safety procedures (Bellora et al., 2001; Deck et al., 2003a; Feist et al., 2009; Fenner et al., 2005; Hopes and Chinn, 1989; Kim et al., 1997; Kleiven, 2003, 2005; Marjoux et al., 2008; Newman, 1980, 1986; Viano, 1988).

The HIC is the result of the evolution from the Wayne State Tolerance Curve (WSTC), developed in the pioneering work of Gurdjian and his co-workers (Gurdjian et al., 1953, 1955) and was firstly presented by Lissner et al. (1960), which established the relationship between average translational accelerations and durations of average acceleration pulses, and also by creating a boundary that separates the “skull fracture” zone from the “no skull fracture” zone (Nahum and Melvin, 1993), becoming useful as a criterion for determination of concussion and onset of brain injury. Further works were also developed (Gurdjian et al., 1963), until the final form of WSTC was published by Gurdjian et al. (1966), shown in Fig. 19, where skull fracture and concussion were used as the failure criterion. This relation between concussion and skull fracture was also observed by Melvin and Lighthall (2002), where 80% of all observed concussion cases also had linear skull fractures. In the final form, the WSTC was developed by combining results from a wide variety of pulse shapes, cadavers, animals, human volunteers, clinical research, and injury mechanisms.

Therefore, the head can withstand higher accelerations for shorter durations and any exposure above the curve is considered an injury, while below does not exceed human tolerance. The WSTC is also supported by experiments conducted by Ono et al. (1980) in primates and scaled to humans, which led to the Japan Head Tolerance Curve (JHTC) that is very similar to the WSTC. Nevertheless, the WSTC is based only on direct frontal impact tests and it was not applied to non-contact loading conditions and other impact directions.

By plotting the WSTC in a logarithmic scale, it becomes a straight line with a slope of −2.5, which was used by Gadd in his proposed severity index called Gadd severity index (GSI) (King, 2000; Nokes et al., 1995). Gadd (1966) introduced the concept of a severity index to provide a rational and consistent basis for comparing the severity of various head impacts, based on the WSTC and on the long pulse duration tolerance data by means of the (Eiband, 1959) test data and given by this empirical expression:

$$GSI = \int a(t)^{2.5} dt \tag{1}$$

where a is the instantaneous head acceleration in *g*’s and t is the time duration of the acceleration pulse in seconds. The initial value to this failure criterion from Gadd’s point of view, was initially set to a 1000, as a threshold for concussion for frontal impact. Later, Gadd (1971) suggested a threshold of 1500 for non-contact loads on the head.

Over the years, this criterion was reviewed and several modified forms were proposed. One of those reviews was made by Versace (1971), who analysed the relationship between the WSTC and GSI and proposed a mathematical approximation of the WSTC that is based on average acceleration, Eq. (2).

$$VSI = \left[\frac{1}{T}\int_{t}^{t} a(t)dt\right]^{2.5} \tag{2}$$

Later, the National Highway Traffic Safety Administration (NHTSA) proposed the head injury criterion (HIC) (NHTSA, 1972), a new criterion to identify the most damaging part of the acceleration pulse by finding the maximum value of the same function. This form is known at present as HIC:

$$HIC = \left(\left[\frac{1}{t_2 - t_1}\int_{t1}^{t2} a(t)dt\right]^{2.5}(t_2 - t_1)\right)_{max} \tag{3}$$

where $a(t)$ is a resultant head acceleration in *g*’s, the interval $t_2 - t_1$ are the bounds of all possible time intervals defining the total duration of impact that must be less or equal to 36 ms and t_1 and t_2 are any two points of the acceleration pulse in time, in seconds.

A HIC value exceeding 1000 is considered to cause severe head injury (however a helmet could be approved by a standard with higher HIC values). Hopes and Chinn (1989) indicated that there is an 8.5% probability of death at an HIC value of 1000, 31% at 2000 and 65% at 4000. Marjoux et al. (2008) predicted a 50% risk of skull fracture, SDH, moderate neurological injury and severe neurological injury using HIC as criterion, obtaining these values respectively: 667, 1429, 533 and 1032. Other values were determined by Peng et al. (2012), predicting a 50% probability of AIS 2+and AIS 3+head injury risk for HIC values of 825 and 1442 respectively.

The HIC takes into account acceleration and impact duration. The linear acceleration, a, is the resultant acceleration measured by a triaxial accelerometer array positioned in the headform centre of gravity, which has similar inertial properties than the human head. This approach claims that two parameters (acceleration and duration of the acceleration over the time of impact) rather than a

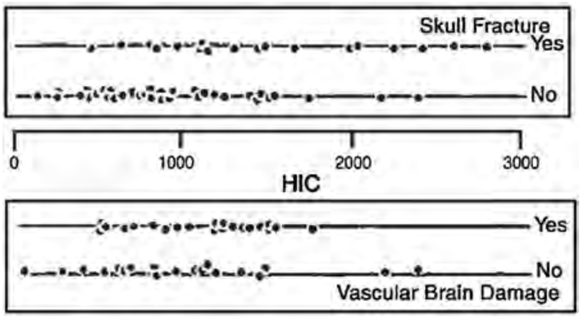

Fig. 20. Relationship between measured HIC and the occurrence of the skull fracture or the extravasations of fluid from blood vessels (Viano, 1988).

single parameter (peak acceleration) consists in an improvement in terms of criteria assessment (Newman, 1980). However, still does not take into account variations in human tolerance and it is based on the assumption that human brain is a viscoelastic medium (Kim et al., 1997).

Other researchers have criticized the use of the HIC as a suitable predictor for head injury, for example Feist et al. (2009) that criticized HIC because it is solely based on translational acceleration and does not take rotational acceleration into consideration. In Fig. 20, it is possible to observe the occurrence of head injuries even for the cases where HIC values are below the limit. In other words, the occurrence of skull fractures and brain damage was also observed at relatively low HIC values. In this figure, it is not noticeable the existence of a clear HIC limit to these injuries, because there are cases of injury and no injury through the analysed HIC range. In addition, Viano (1988) added that reliable predictions should not be expected from a measurement of a resultant translational acceleration of the head and analysis by a mathematical routine that gives results in a single HIC value. However, Hopes and Chinn (1989) also reviewed HIC drawbacks made by other researchers and concluded that HIC still could be an useful predictor for comparing energy absorbing safety devices in impacts where the death frequently occurs without skull collapse. Also, Deck et al. (2003a) concluded that HIC is able to represent the global severity level of an impact and the potential head injury level, however HIC is unable to predict diffuse brain injuries and SDH that are linked to the angular acceleration sustained by the head during the impact. It was also highlighted that an optimization based on biomechanical criteria is different than the optimization with HIC criterion (Deck et al., 2003b). In a study about HIC, Fenner et al. (2005) criticized HIC for not being sensitive to impact direction. Newman (1980) stated the same opinion. Kleiven and von Holst (2002) also criticized HIC, once it does not predict the size dependence of the intracranial stresses associated with injury and it does not take into account head sizes. However, limits for some sizes of the human head were proposed for HIC_{36} by Kleinberger (1998) and for HIC_{15} by Eppinger et al. (2000) using the HIC scale factor proposed by Melvin (1995). The higher proposed limits were the ones relative to adults, 1000 and 700 for HIC_{36} and HIC_{15}, respectively. However, these limits only take into account the skull material properties.

In overall, HIC is considered to be not enough to predict head injuries because it does not take into account the injury type, the rotational motion and the impact direction and also has nonsensical units (Newman, 1975).

Thus, HIC only treats the resultant translational acceleration and the duration of the impulse and no consideration is made for the direction of the impulse or rotational acceleration components (Bellora et al., 2001; Kleiven, 2003, 2005).

Despite all the criticism, HIC is the most disseminated injury criterion as it is adopted by current helmet's standards for helmet certification, such as the ECE R22.05. Thus, virtually all helmets available in the market were assessed according to this criterion. The PLA is used together with the HIC by the majority of the standards. ECE Regulation (2002) is an example of a standard that use both criteria. In the case of ECE 22.05, the peak linear acceleration is limited to 275 *g* and the HIC value should be inferior to 2400 in order to be approved. However, Shuaeib et al. (2002a) concluded that a severe but not life-threatening injury can occur if HIC reaches or exceeds 1000. Limits for HIC were suggested by Horgan (2005) for HIC values of 1000 and 3000 which were defined as 16% and 99% probability of life threatening injuries, respectively.

King et al. (2003), in a numerical study, estimated the MTBI (mild traumatic brain injury) tolerance for HIC_{15}, where there is a probability of MTBI occurrence of 25%, 50% and 75% for HIC values of 136, 235 and 333, respectively. Zhang et al. (2004) proposed a linear acceleration of 85 *g* with an impact duration ranging between 10 and 30 ms and a HIC value of 240 as the injury tolerance for mild TBI.

Thus, HIC and proposed acceleration thresholds do neither take into consideration rotational and translational loads, nor directional dependency. There is therefore a need for more complex injury assessment functions, accounting for both translational and angular acceleration components as well as changes in the direction of the loading (Kleiven, 2005).

4. Oblique impact

The most frequent severe injuries in motorcycle crashes are head injuries, mainly caused by rotational forces (Aare et al., 2004; Gennarelli, 1983) that are most commonly generated as a result of oblique impacts (Otte et al., 1999).

Head rotational force results in large shear strains arising in the brain, which has been proposed as a cause of traumatic brain injuries like DAI by the tearing of neuronal axons in the brain tissue and SDH by rupturing bridging veins (Gennarelli, 1983; Margulies and Thibault, 1992). Thresholds were proposed in these studies and many others reviewed. However, many of these thresholds were proposed based on pure rotational motion and DiMasi et al. (1995) and Ueno and Melvin (1995) concluded that these thresholds probably have to be decreased, if an angular motion is combined with a translational motion, which is typical in oblique impacts.

In current helmet standards tests, no rotational effects are measured in the headform, partly because there are no accepted global injury thresholds for a combination of rotations and translations and there is no realistic test capable of reproducing impacts similar to the most commonly observed impacts in real life motorcycle accidents.

One of the reasons for such lack is that the criteria used, PLA and HIC, only assess linear motion. The ECE R22.05 shock absorption test allows headform rotation during impact, but, unfortunately, rotational accelerations are not measured. There is other test that allows headform rotation. However, this is used only to assess external projections against helmet's surface, to check that helmet projections do not cause excessive tangential forces. One of the reasons why there are no helmet standards measuring rotational effects is because there are no globally accepted injury tolerances for helmet impacts that include rotations (Aare, 2003; Aare et al., 2004). Several criteria were proposed over the years but none was globally well accepted.

This ignores the fact that in real life almost always external load results in both translational and rotational head accelerations and both determine the total deformation pattern of the brain. In addition, the effects of rotational acceleration are believed to be the main cause for specific types of traumatic brain injury, such as DAI and SDH (Ho, 2008; Glaister, 1997; Gennarelli, 1981,

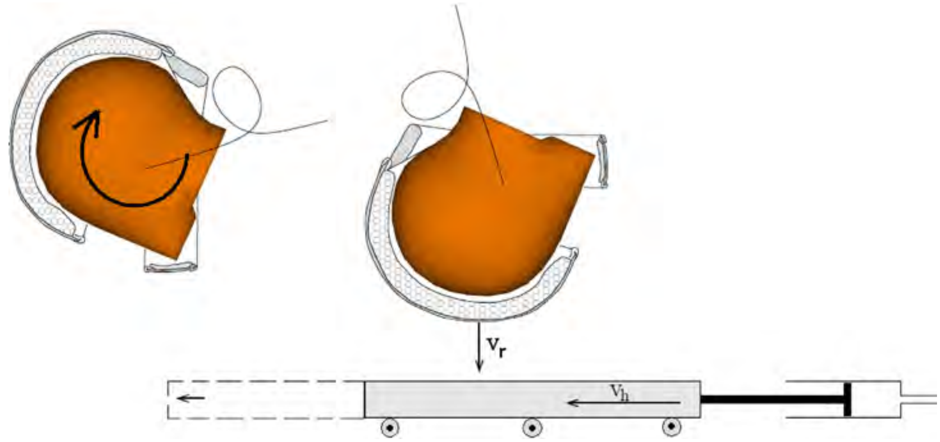

Fig. 21. Oblique impact.

Adapted from: (van den Bosch, 2006)

1983; Gennarelli et al., 1987; Kleiven, 2005, 2007b; King et al., 2003; Ommaya, 1988; Viano and King, 1997; Holbourn, 1943; Ommaya and Hirsch, 1971; Unterharnscheidt and Higgins, 1969; Unterharnscheidt, 1971; Hodgson and Thomas, 1979; Ono et al., 1980; Margulies et al., 1990; Ueno and Melvin, 1995; Miller et al., 1998; Zhang et al., 2003; Aare et al., 2004; Mordaka et al., 2007; Oehmichen et al., 2006; Bandak, 1997; Aare, 2003; Bandak and Eppinger, 1994).

Halldin et al. (2001) recognize rotational accelerations to be a major cause for head injury in motorcycle accidents, in particular SDH and DAI. Since oblique impacts, with a significant tangential force on the helmet are more common than radial (normal) impacts in motorcycle crashes, Otte (1991) and Otte et al. (1999) developed an oblique test procedure to assess the helmet's ability to reduce rotational acceleration of the head during impact. In this test, a free falling helmeted headform impacts an horizontally moving steel plate covered with grit grinding paper as shown in Fig. 21, in order to be similar to an impact against road surface.

The oblique impact test proposed by Halldin et al. (2001) consists in a free falling headform that impacts an horizontally moving steel plate moved by a pneumatic cylinder of 1 m stroke. It is possible to perform an oblique impact at a desirable impact velocity by controlling the radial helmet velocity and the tangential velocity of the plate. A rough road surface was simulated by a grit grinding paper, bonded to a steel plate, which slides on flat PTFE (polytetrafluoroethylene) bearings. In the headform centre of gravity an accelerometer capable of recording linear and rotational acceleration components was positioned. Further developments were made by Aare (2003).

In this study it was also concluded that higher angular accelerations are found in rougher surfaces. The basic idea behind this configuration is shown in Fig. 21 and was first presented by Harrison et al. (1996). After the oblique test proposed by Halldin et al. (2001), others emerged, such as the ones proposed by Aare and Halldin (2003) and Pang et al. (2011). Aare and Halldin (2003) recently proposed a new method to test helmets for oblique impacts, but they did not propose any injury tolerances for such a test. Later, in a following work, Aare et al. (2004) proposed a threshold based on both types of motion, where the authors concluded that this threshold makes good predictions of brain injuries.

More recently, Mills et al. (2009) found that the peak headform rotational acceleration was shown to be a function of three main parameters: the impact velocity component normal to the road, the friction coefficient between the shell and the road, and the impact site/direction. It was relatively insensitive to the tangential component of impact velocity, where no relation between tangential force and rotational acceleration was found. Several oblique impacts were performed with different friction coefficients between the headform and the inner liner and it was observed that raising the friction, the head angular acceleration also raised proportionally. Mills et al. (2009) also showed that side impacts induce larger rotational accelerations to the head than impacts at other sites of the helmet.

The friction coefficient between the shell and the road was also identified by Finan et al. (2008) as the parameter with more influence in oblique impacts, where reducing the friction between these two surfaces reduced the peak rotational acceleration and vice-versa. In the COST programme, it was concluded that the frictional behaviour of helmet shells, either observed in a friction test or in an oblique impact test, depends on material and surface properties as well as on deformation of the samples (COST, 2001). In this study it was also concluded that the maximum external force coincides with peak linear and rotational acceleration, where their peaks happen roughly at the same time.

Ghajari et al. (2013) carried out a study where it was investigated the effects of the presence of the body in helmet oblique impacts. The THUMS human body model presented previously (Iwamoto et al., 2002, 2007), was used. A comparison between full-body impacts and those performed with an isolated headform showed that the presence of the body modified the peak head rotational acceleration by up to 40%. In addition, it had a significant effect on head linear acceleration and the crushing distance of the helmet's liner. The authors also suggested including the effect of the body on head rotational acceleration in headform impacts, modifying inertial properties of the headform. The modified inertial properties were determined by a severe and frequent impact configuration, where the results of helmet impacts obtained by using the modified headform were in very good agreement with those of full-body impacts.

4.1. Advanced motorcycle helmets

In an oblique impact there are three different types of slip that are important to address with regard to absorption of rotational energies during head impacts:

- the first is between the impacting surface and the outer helmet shell;
- the second is between the shell and the liner;
- the third is between the helmet and human head.

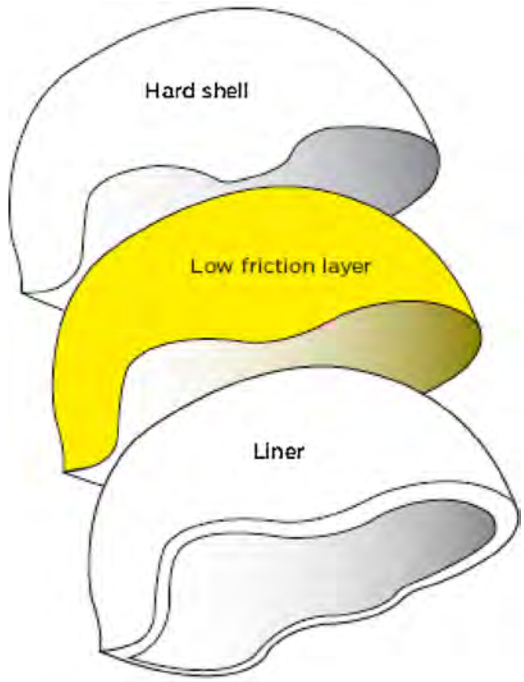

Fig. 22. Multi-direction Impact Protection System (MIPS) – construction (MIPS).

Helmets are already very smooth on the outside to reduce friction between the impacted object and the helmet. Nevertheless, since the helmet has to fit human head in order to avoid other injuries, the slip between the shell and the liner is the only place where a significant improvement is possible. Based on this, Halldin et al. (2001) presented a new helmet, the Multi-direction Impact Protection System (MIPS).

Other prototype helmet, the Phillips Head Protection System (PHPS) proposed by Phillips (2004), aims to reduce friction outside the helmet shell, by introducing easy-shear layer, contrary to the MIPS that introduces the easy-shear layer inside the helmet. The developers argued that this would reduce head rotational accelerations.

4.1.1. *Multi-direction Impact Protection System (MIPS)*

Besides the oblique impact test proposed, Halldin et al. (2001) presented a new helmet. In that study it was tested one helmet type with three different interfaces between outer shell and protective padding liner:

- The "bonded" helmet, where the outer shell and the protective padding liner were glued together;
- The "free" helmet, the outer shell and the protective padding liner were joined by rubber strips at the bottom edge. No countermeasures for reducing friction between outer shell and protective padding were taken;
- The MIPS helmet, which was designed to reduce head's rotational acceleration. In this one, a low-friction Teflon film (a low-friction layer) is placed between the outer shell and the protective padding liner which allows the shell to rotate relative to the liner in an oblique impact, as shown in Figs. 22 and 23 . MIPS has also a release mechanism that will make the helmet feel robust in normal handling but will release when a certain load is exceeded. Comparing to a conventional helmet, the weight is increased by less than 5%, while comfort and design will not change at all with this new technology (MIPS).

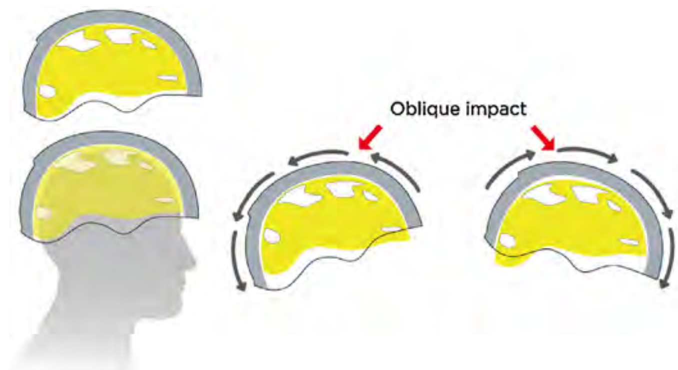

Fig. 23. Multi-direction Impact Protection System (MIPS) – function (MIPS).

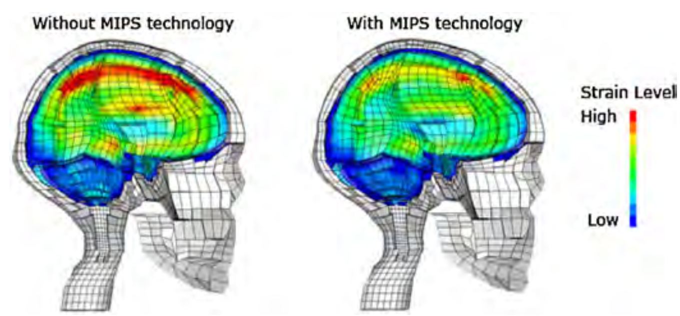

Fig. 24. Results of oblique impact simulation with KTH FEHM (MIPS).

The MIPS helmet reduced the peak rotational acceleration by 28% and 39% compared with the FREE and the BONDED helmets, respectively (Halldin et al., 2001). However, the magnitude of the reduction effect in the MIPS helmet is somewhat decreased in tests incorporating an artificial scalp on the headform (Aare and Halldin, 2003). It was also concluded that the comfort foam has influenced the results significantly, since one of the functions of the comfort foam is to provide a better fit, which is important in oblique impacts.

Basically, the MIPS mimics the brain's own protection system based on a sliding low friction layer between the head and helmet liner, brain injuries are significantly minimized in connection with angled impacts, as shown in Fig. 23. When the head is subjected to an impact, the brain slides along a membrane on the inner surface of the skull, which reduces the forces transmitted to the brain.

In numerical and experimental tests, the last version of MIPS has shown a dramatic reduction of the forces transmitted to the brain, as shown in Figs. 24 and 25 , respectively. The results showed that it was possible to reduce the forces to the brain by up to 40% at an impact angle of 45° by adding MIPS technology (MIPS). It has also been shown that helmets with MIPS technology perform well in the standard regulation test used today (MIPS).

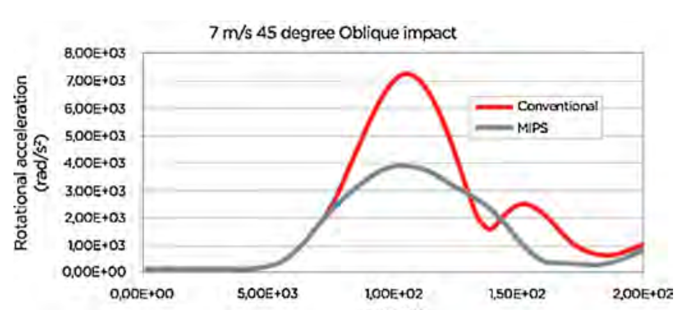

Fig. 25. Results of experimental oblique impact (MIPS).

Fig. 26. Phillips Head Protection System (PHPS) (Ask Nature).

4.1.2. The Phillips Head Protection System (PHPS)

The PHPS enhances traditional helmet design by adding a specially designed lubricated high-tech polymer membrane over the outside of the helmet (Phillips, 2004). Several materials were used, such as closed cell plasticized PVC, high density PU foam and silicon foamed rubber (Phillips, 2004). The membrane is designed to slip in a controlled manner over the inner shell of the helmet. This concept mimics human scalp, which is a natural protection to brain and skull. Thus, this is a layer on the outside of the helmet which acts exactly as the scalp does in human head: by sliding on the shell it limits rotation. An illustrative example is shown in Fig. 26.

The lubricant and elastic quality of the PHPS membrane on a crash helmet decreases this rotational force and reduces its effect by over 60% in the critical milliseconds following an impact, significantly reducing head trauma and reducing the risk of traumatic brain injury (Phillips, 2004). It decreases the friction of the helmet surface by moving and sliding over the hard shell. The PHPS high-tech polymer membrane was developed together with a specially designed lubricant. The membrane is designed to slip in a controlled manner over the inner shell of the helmet.

This technology is already commercialized by Lazer SuperSkin motorcycle helmet. Tests performed on the first commercial implementation of the PHPS by LAZER Helmets SA verified that upon head impact the LAZER SuperSkin helmet reduced the risk of intracerebral shearing by 67.5%, by reducing the mechanical effects of rotational acceleration by more than 50% (Lazer helmets). These results were obtained by Rémy Willinger through simulations performed with the FE head model of Strasbourg's University (Phillips Head Protection System). These tests show the rotational effect of the impact is significantly reduced with the addition of the PHPS membrane.

This concept was also proposed by Mellor and StClair (2005) that tried to develop an advanced helmet; several layers were tested in several tests. The sacrificial layer in the exterior of the outer shell revealed good results.

5. Finite element models of a motorcycle helmet

Initially, motorcycle helmet's design, impact behaviour investigation and optimization were based on experimental investigation, where the results were restricted by varying few impact parameters such as the impact speed or the shape of the anvil. However, varying helmet parameters experimentally was an impossible task, due to testing sample manufacturing constraints, mainly the cost and time spent with such methodology.

This was overcome with the development of analytical models. The development of mathematical models is vital to a better understanding of the helmet impact and also head injury mechanisms. The exact manner in which helmets protect the head is still not fully understood. Over the years several mathematical models have been proposed.

The earlier theoretical attempts to solve the helmeted-head impact problem were based on analogue techniques that the helmeted-head system could be approximated by an equivalent set of lumped masses, springs and dashpots (Mills and Gilchrist, 1988; Wilson and Carr, 1993; Gilchrist and Mills, 1993, 1994b), that represented helmet components and the headform. These lumped masses systems were then solved using basic dynamic and vibration theories such as modal analysis and dynamic compression (Willinger et al., 2000b). The lumped mass models considered useful in parametric studies are usually simple models capable of providing a quicker and cheaper prediction than an empirical approach and also capable of describing deformation for one specified type of loading condition. However this solution, which is usually either one or two dimensional, had limited advantages due to the approximation degree involved and the incapacity of representing most of the essential impact features encountered in real accidents (for example the helmet geometry) and – finally – with these models it is impossible to calculate the stiffness of individual helmet parts from their shapes, dimensions and material properties. For example, the influence of the helmet fit on the headform is not taken into account by almost all these models, because it is difficult to model such interaction, mainly during impact. A few authors tried, such as Willinger et al. (2000b). This means that the application of lumped mass models is very limited.

Such fact, allied to the advance of CPU power, led to the development of detailed models by using Finite Element Method, which led to more detailed results on stresses and strains not only from the impacted helmet but also from the human head. Finite Element Models do not only allow modelling the mechanical properties of the helmet components, but also include the geometry of the helmet. This allows the influence of the interaction between helmet and head to be investigated and provide much more information about the helmet's impact than a lumped mass model. The first attempt was reported by Khalil et al. (1974), performed with the concern about the biomechanical response of the head to the transient impact waves. More examples of the first simplified FE models are the ones developed by Köstner and Stöcker (1987), van Schalkwijk (1993), Yettram et al. (1994). Some of these models had some limitations, such as the non inclusion of a separate headform model and none of them took into account the effect of the soft comfort liner that provides the fit of the helmet to the head (Brands, 1996). Also, the first models were not validated, but were used for trend studies only.

Few years later, more advanced FE motorcycle helmet models were developed by Brands et al. (1996), Liu et al. (1997), Liu et al. (1998), Liu and Fan (1998), Scott (1997) and Chang et al. (2000). In these models, helmet geometry was simplified, with either spherical or regular shapes adopted. Brands (1996) and Brands et al. (1996) also validated the developed helmet FE model under standard tests in terms of head acceleration where it was highlighted the helmet behaviour during the impact.

More recently, more realistic models were developed, in order to study helmet's materials (Alves de Sousa et al., 2012; Caserta et al., 2011; Kostopoulos et al., 2002; Pinnoji et al., 2008a, 2010; Pinnoji and Mahajan, 2010; Tinard et al., 2011, 2012a), the optimization of head dummies (Willinger et al., 2001; van den Bosch, 2006), the oblique helmeted impacts (Aare, 2003; Forero Rueda et al., 2011; Ghajari et al., 2011; Mills et al., 2009), the effect of impact velocities (Chang et al., 2003), the helmet's design optimization (Deck et al., 2003a; Mills and Gilchrist, 1992; Pinnoji and Mahajan, 2006; Pinnoji et al., 2008b; Tinard et al., 2012b), the virtual modelling and simulation of impacts with motorcycle helmets (Aiello et al., 2007; Cernicchi et al., 2008; Ghajari et al., 2009; Pratellesi et al., 2011) and the biomechanics on helmeted impacts, such as the optimization against biomechanical criteria (Deck and Willinger, 2006; Marjoux et al., 2008; Neale et al., 2004; Shuaeib et al., 2002a; Tinard et al., 2012b; Willinger et al., 2000a, 2002) among many others available in the literature. Therefore, once a functioning and validated numerical helmet model is created, a great variety of information can be obtained. Such a model may be a three dimensional Finite Element Model, to account for shell vibrations and to be able to use complex anisotropic material models.

6. Conclusions

Helmets are used as the main head protection gear for a long time. There are a lot of helmet types nowadays, specific for each application. Motorcycle helmets are widely used between motorcyclists, being required in almost all countries due to motorcycle standards. This is the most effective means of protection available for a motorcyclist to protect his head during an accident. Current motorcycle helmets comprise a hard shell and an energy absorbing liner. Head injury mechanisms have been intensively studied over half a century, different theories have been proposed, different head models have been developed and a lot of head injury thresholds have been predicted. It is possible to conclude from this paper that there is nowadays a global acceptance of what is thought to be the cause of each injury. However, the criteria and associated thresholds are still being studied because there is no agreement in this matter. This represents a problem for the global community, because the current standards are outdated and there is no agreement about which is the best update. Nevertheless, rotational acceleration is generally accepted among the researcher community as the main mechanism of brain injury and none of the current standards access this type of injury in their helmet impact testing. By the time this paper is being read, new contributes are being published, conferences are happening and ideas are appearing. The authors hope that this review paper can serve the purpose of being a good kick-start survey for any scientist trying to enter in this field to keep the pace with the rapid evolution of this area.

Acknowledgements

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