Proposal of a new motorcycle helmet test method for tangential impact

Nicolas Bourdet, Sounak Mojumder, Simone Piantini, Caroline Deck, Marco Pierini, Remy Willinger

Abstract Based on an existing in-depth accident database available at Florence University, a total of 19 motorcycle accidents were selected for reconstruction using a multibody simulation approach. These computations give access to the head impact conditions in terms of initial head velocity vector. The real-world accident simulation permitted the computation of the victim's kinematics and demonstrated that the impact velocity vector presents an important tangential component. Typically, an impact angle of 40–50° is observed. Therefore, it is suggested to consider tangential helmet impacts in three different planes with an 8.5 m/s impact speed against an anvil inclined at 45°. Further, it has been demonstrated that the 6D HIII headform response under tangential impact is reliable and can be used in helmet testing. Finally, by driving a human head FE model with the 6D acceleration versus time history, it appears that the brain injury risk can be computed. The proposed test method suggests four tangential impacts leading to rotation among the three reference axes. For this combined linear and rotational head loading, the helmet can be evaluated against brain tissue level injury.

Keywords Motorcycle helmet, head impact conditions, accident reconstruction, multibody simulation, head injury criteria.

I. INTRODUCTION

It is well known in the scientific community that head rotational acceleration is a critical head loading that can lead to brain injury. Concerning neurological injuries, in 1943 Holbourn [1] suggested that the rotational acceleration causes high shear strains in the brain, thus rupturing the tethering cerebral blood vessels, neo and subcortical tissue. Holbourn was the first to suggest the importance of rotational acceleration in the generation of cerebral concussion. In 1967, Ommaya *et al.* [2] proposed a method in order to extend the results of experiments on concussion producing head rotations on lower primate subjects to predict the rotations required to produce concussions in man. A chart of angular acceleration required to reproduce concussion in the rhesus monkey indicates that an acceleration of 40,000 rad/s² will have a 99% probability of producing concussion, which was expected to correspond to an angular acceleration of 7,500 rad/s² for humans.

In a study based on primates, Gennarelli *et al.* [3] proposed that a rotational acceleration exceeding 17,500 rad/s² would produce Sub Dural Hematoma (SDH) in the rhesus monkey. With the objective of investigating the influence of the head rotational accelerations on the intra-cerebral mechanical parameters under accidental head impact, a total of 69 real-world head trauma were simulated by Deck *et al*. [4], with and without considering the angular rotation. The numerical simulation of these head trauma, by considering linear and rotational accelerations, on the one hand, and linear head acceleration only, on the other hand, permitted it to demonstrate and to express quantitatively the dramatic influence of the rotational acceleration on both intra-cerebral loading and brain-skull relative motion, supposed to lead to neurological injuries and subdural hematoma, respectively. In this study, for all accident cases considered the effect of angular acceleration was found to increase the intracerebral shearing stress by about 50% of whatever the impact severity was. Kleiven *et al*. [5] and Zhang *et al.* [6] demonstrated that the angular kinematics of the head was the most important factor in determining the brain strain, based on numerical simulation of real-world head trauma. More recently, Takhounts *et al*. [7-8] used a head FE model in order to establish a head injury criteria for rotational acceleration, called BrIC.

In parallel with the demonstration of the critical role of head angular acceleration in brain injury, a number of studies focused on the head kinematics in real-world accidents in order to demonstrate that a tangential loading of the head does exist in addition to the normal impact velocity. Mills *et al.* [9] showed that oblique

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impacts are the most common situations in motorcycle crashes. More recently, Bourdet *et al*. [10-11] quantified the head rotational acceleration due to the tangential component of the head impact by reconstructing real-world and virtual motorcycle and bicycle accidents.

Head injury is the most common cause of severe injuries in motorcycle accidents [12]. Otte *et al.* [13] showed that oblique impacts, with a significant tangential force on the helmet, are more common than radial (normal) impacts in motorcycle crashes. They found that the mean impact angle, between the helmet direction and the horizontal, was 28 degrees. Similarly, Harrison *et al.* [14] found that, for a jockey falling from a horse in a steeplechase, typically the impact angle is about 60 degrees/normal.

An analysis of several real-world accident cases from COST 327 European project concluded that the median speed for concussion was about 47 km/h (13 m/s) and for brain injury about 60 km/h (16.5 m/s). Moreover, for head injury AIS2+, rotational motion was found to be the cause of over 60% of injuries and linear motion about 30% of injuries: 57% of impacts occurred at body angle of less than 30 degrees; 32% of impacts occurred at body angle greater than 60 degrees.

Despite this widely recognised understanding of head rotational loading and the effect of the induced rotational acceleration to the brain, no standard head protection is currently considering head rotational acceleration. Only ECE R22.05 EU [15] motorcycle helmet standard considers a tangential impact condition, but helmet evaluation is limited to the recording of the tangential force. One possible reason for the current situation is that no accepted head rotation threshold has been established yet. A number of maximum head rotational accelerations have been proposed in the literature [9], [16]–[18], but none of them considers the time evolution or rotation direction. In the authors' opinion, the only way to integrate the complexity of brain geometry and brain material properties under complex head loading is to progress towards tissue-level brain injury criteria, as proposed in existing FE model-based head injury criteria [5-6][19-20]. In 2004 Deck demonstrated that helmet optimisation strongly depends on the head substitute and injury criteria taken into account. More recently, first attempts were made [21]) to optimise new helmets against biomechanical criteria by coupling the human head FE model to a helmet FE model. Advanced model-based head injury criteria have also been suggested in recent attempts to improve bicycle helmet test methods [22].

In the domain of bicycle helmet evaluation, Milne *et al.* [23-24] suggested a new helmet assessment method using model-based head injury criteria under both linear and tangential impact conditions, exactly as Hansen *et al*. [21] did in the context of the development of an advanced "honeycomb" bicycle helmet. In a similar way, but in the context of hockey helmet evaluation, Post *et al.* [22] suggested to impact a helmeted Hybrid III head neck system and to introduce the linear and rotational accelerations into an existing head FE model in order to assess the injury risk.

In order to make progress in the field of head protection against tangential impacts, a number of attempts were proposed in the literature. Aldman *et al.* [16] dropped a helmeted headform fixed to a dummy neck against a rotating steel disc. Halldin *et al.* [17] designed a new oblique impact test for motorcycle helmets based on an instrumented free Hybrid III dummy head dropped vertically against a horizontally moving plate. More recently, Pang *et al*. [18] published a novel laboratory test in order to investigate head and neck responses under oblique motorcycle helmet impacts using a mobile anvil. This proposal is based on a test rig considering a helmeted Hybrid III head fitted to the Hybrid III neck, which itself is fixed to a 20 kg mass that drops against a sliding plate.

Concerning hockey helmets, Gerberich *et al.* [23] and Flick *et al.* [24] investigated hockey head trauma and reconstructed experimentally typical impact conditions applied to the Hybrid III head and neck system. It was shown that both linear and rotational head accelerations are significant and can potentially lead to brain injury as far as head injury criteria proposed by Zhang *et al.* [6] are concerned. More recently, Rousseau *et al*. [25-26] developed a hockey helmet test bench where the helmeted Hybrid III head was fixed to a Hybrid III neck and impacted frontally or laterally with an impactor. Linear and rotational head accelerations in the range of 100–120 g and 3–6 krad/s², respectively, were recorded. In order to further investigate these aspects, Walsh *et al.* [29] investigated helmeted Hybrid III head kinematics when fixed on Hybrid III neck and impacted at a number of points around the head from several different impact angles. First result is that even when directed along the centre of mass, rotational acceleration can be as high as 10 krad/s². Finally, Walsh *et al.* [30] consolidated these results with 20 further impacts and demonstrated that highest linear acceleration was obtained for radial directed impact for all impact points. In this study the authors plotted linear versus rotational acceleration for all impacts. Pure correlation was found (R^2 = 0.4), demonstrating that both injury parameters must be recorded during testing as one cannot be estimated by means of the other.

As far as motorcycle helmets are concerned, no improvement of current standard tests has been proposed since 2005 to the authors' knowledge. Therefore, the present paper's objective is to consolidate road observation via real accident simulation, and to endup with a proposal for a motorcycle helmet test method under tangential impact, including advanced model based brain injury criteria.

II. METHODS

The methodology applied in this study is to analyze the motorcyclist's kinematics during a real-world accident in order to extract the head impact conditions in terms of head impact point and impact velocity vector. The method is to reconstruct accidents using multi-body systems based on real-world accident data.

A total of 19 motorcycle accidents were selected from Florence University's in-depth accident database (named InSAFE) [38-39]. Each accident reconstruction was carried out using Madymo® software. The human model considered for the motorcyclist was the scalable pedestrian TNO model implemented in Madymo® package and validated against post-mortem human tests, as reported by Hoof *et al.* [33] and De Lange *et al.* [34]. The principle of solving multibody system is to define a set of rigid bodies represented by ellipsoids and connected by joints. Unlike finite element (FE) computation, contact between two bodies is not computed by deformable surfaces but rather by a penetration force defined by a function. The computational time of this multibody approach is strongly reduced in comparison with FE simulation. The models are developed using ellipsoids in such a way that the geometry, mass and inertia are respected. The vehicle models are developed using ellipsoids so that the geometry is respected, as illustrated in Fig. 1. The contact force functions used on each part of the car are extracted from the study of Martinez *et al.* [35], in the same way as modelled in *Bourdet et al.* [10].

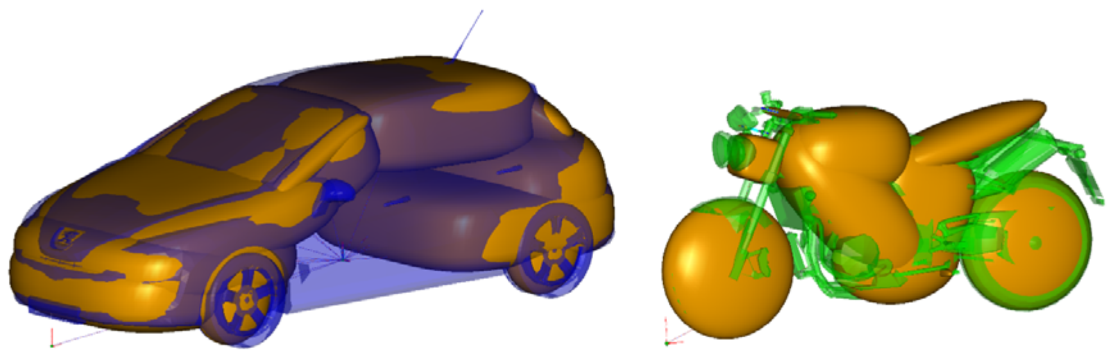


Fig. 1. Car and motorcycle modelling with superimposition of ellipsoids on car geometry.

A motorcycle accident reconstruction can be divided into three main steps: accident data collection and analysis; victim's kinematic reconstruction; and extraction of head impact conditions.

The first step consists of gathering accident data and evaluating the initial conditions of the accident, either by using the momentum and energy balance or Virtual Crash® and PC-Crash® software, as well as studying the victim medical report. This step was carried out by the MOVING team from Florence University. The second step is to compute the kinematics of the motorcyclist during the accident. This reconstruction is based on a parametric analysis that permits the extraction of the most realistic accident constellation.

III. RESULTS

A total of 25 cases of motorcycle accidents from the InSAFE accident database of Florence University (Italy) were collected as part of the European projects RASIF and COST TU1407 "Safe2Wheelers". Among these cases, the most frequently observed collision configurations were fronto-lateral (head-on-side) and stroke side (sideswipe). The powered two-wheeler (PTW) type most often involved in accidents from this database is the scooter, as shown in 0. This database also allows the extraction of the distribution of injured segment depending on the severity of injuries based on the AIS scale, as illustrated in 0. From this database a total of 19 cases have been selected to be reconstructed based on the available accident details.

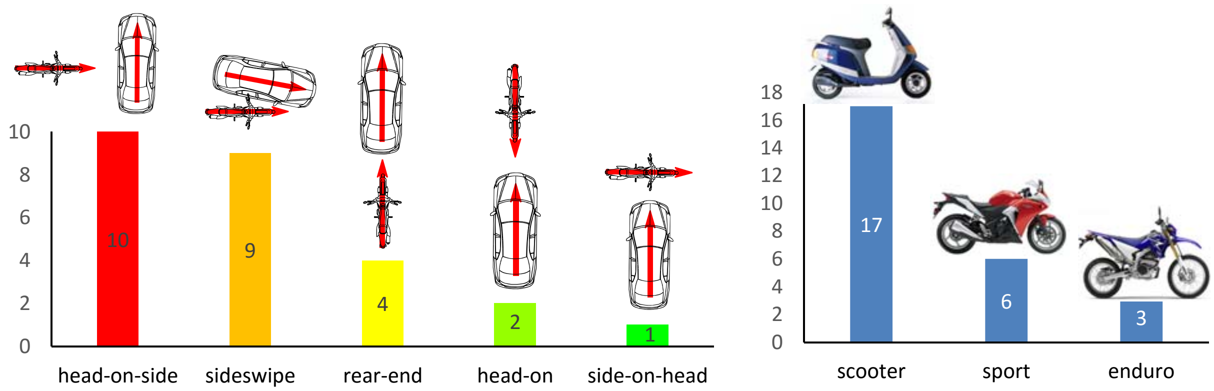


Fig. 2. Distribution of accident configurations (*left*), and distribution of PTW type involved in the 25 selected accident cases (*right*).

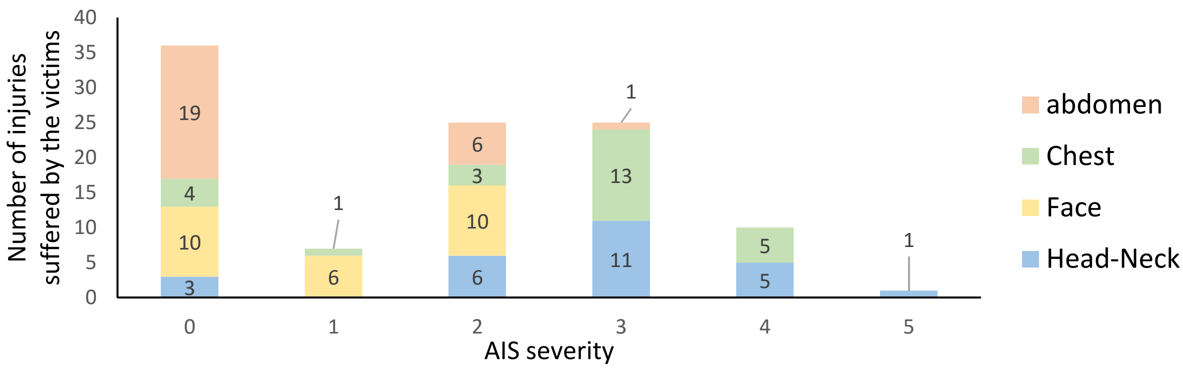


Fig. 3. Distribution of injured segments depending on the severity of the injury, expressed as AIS.

In order to report the results of these accident simulations, a case study is depicted in depth as an example. This accident occurred between a motorcyclist and a car in a head-on situation crash. 0 shows an example of a detailed accident case, with pictures illustrating the final position of each vehicle and damage sustained. Initial conditions are evaluated with Virtual Crash® software and are given as input base for the multibody simulation of the victim's kinematics. The initial conditions of the impact, implemented in Madymo®, as well as the pictures of the kinematics are shown in 0.

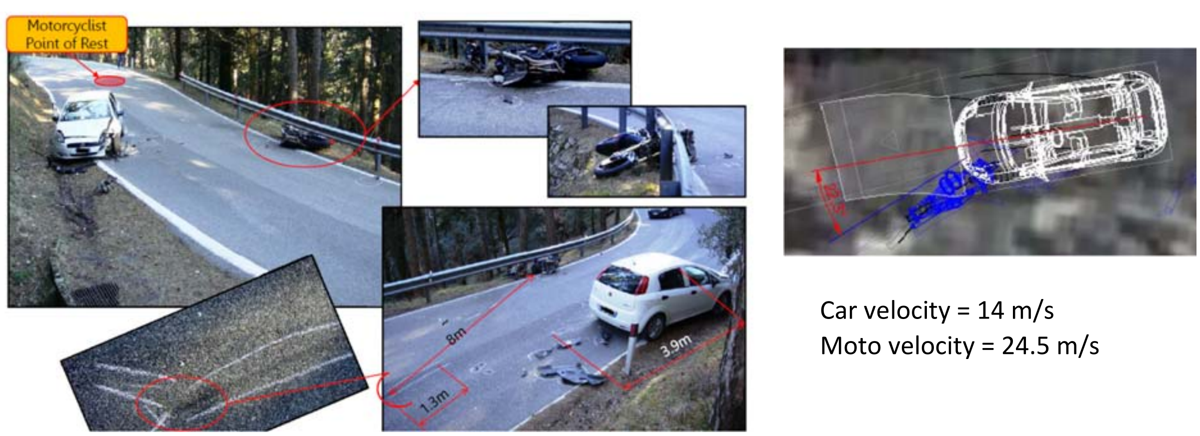


Fig. 4. Scene pictures of the accident case and first rigid body simulation allowing the extraction of accident configuration and velocities.

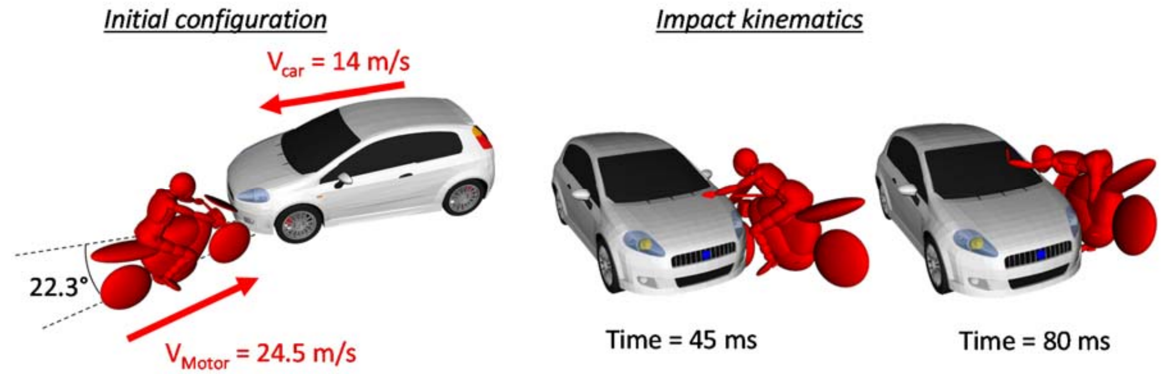


Fig. 5. Illustration of the initial conditions of the impact and extraction of the victim kinematics.

A Python script was used to extract the head impact conditions in terms of location of impact on the head and on the car as well as in terms of head impact velocity vector, as illustrated in 0. For the presented case, resultant head velocity is 18 m/s and normal and tangential velocities are 10.1 m/s and 14.9 m/s, respectively.

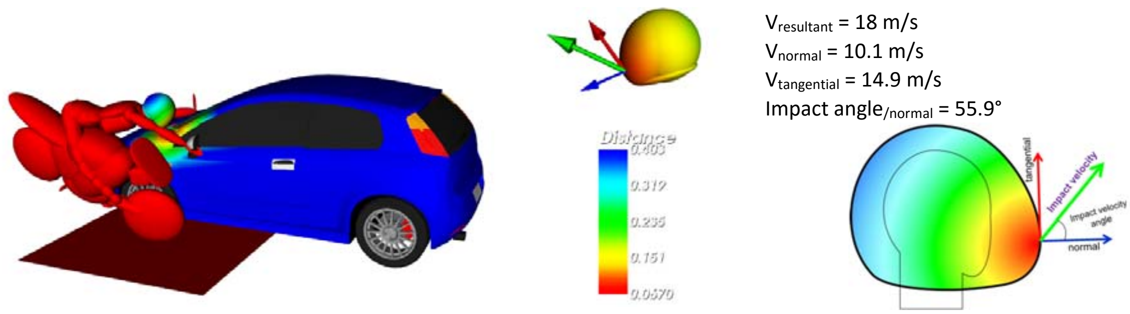


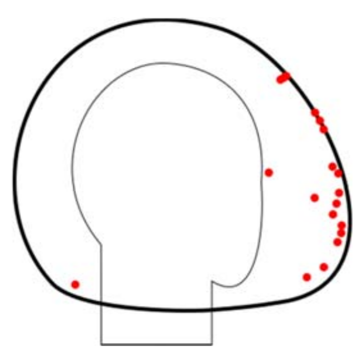
Fig. 6. Illustration of the head impact conditions and extraction of the velocity vector (case 1).

The extraction of the head impact conditions was carried out for the 19 accident cases as illustrated in 0. The results in terms of head impact velocities and impact angles are reported in Table I. The mean resultant head impact velocity is of 10.1 ± 4.6 m/s, with components in terms of normal and tangential velocities of 7.0 ± 3.2 m/s and 6.1 ± 5.1 m/s, respectively. This leads to an impact velocity angle of 36.5 ± 23.3 deg.

TABLE I

HEAD IMPACT CONDITION OF THE 19 ACCIDENT CASES

Case ID	Resultant Velocity [m/s]	Normal Velocity [m/s]	Tangential Velocity [m/s]	Impact Velocity Angle [deg]	Head Impact Points
1	18.0	10.1	14.9	55.9	
2	19.9	14.2	14.0	44.7	
3	4.8	4.5	1.8	21.8	
4	4.8	4.8	0.5	6.2	
5	4.7	4.7	0.8	9.7	
6	12.4	6.6	10.4	57.6	
7	8.9	8.7	1.4	9.3	
8	14.4	9.5	10.8	48.8	
9	3.1	1.8	2.5	54.1	
10	13.9	10.8	8.7	39.0	
11	12.4	12.1	2.8	13.3	
12	11.0	8.2	7.4	42.3	
13	10.4	10.2	1.7	9.5	
14	18.6	4.4	18.1	76.3	
15	11.2	5.1	10.0	63.1	
16	9.7	4.0	8.9	65.6	
17	6.5	6.5	0.1	1.0	
18	6.6	3.5	5.6	58.3	
19	7.7	6.1	4.6	37.2	



The results from the reconstruction of the 19 accident cases can be added to the previous accident reconstructions carried out during several projects conducted in collaboration with BASt, CEESAR and IFSTTAR. A total of 56 real-world motorcycle accidents have been reconstructed, for which mean value of impact velocity in terms of resultant, tangential and normal components, as well as impact angle, can be evaluated. It was observed that the mean impact angle is 44 ± 22 degrees, with an impact velocity of 11.2 ± 6.2 m/s, which show the importance of tangential velocity during real accident cases.

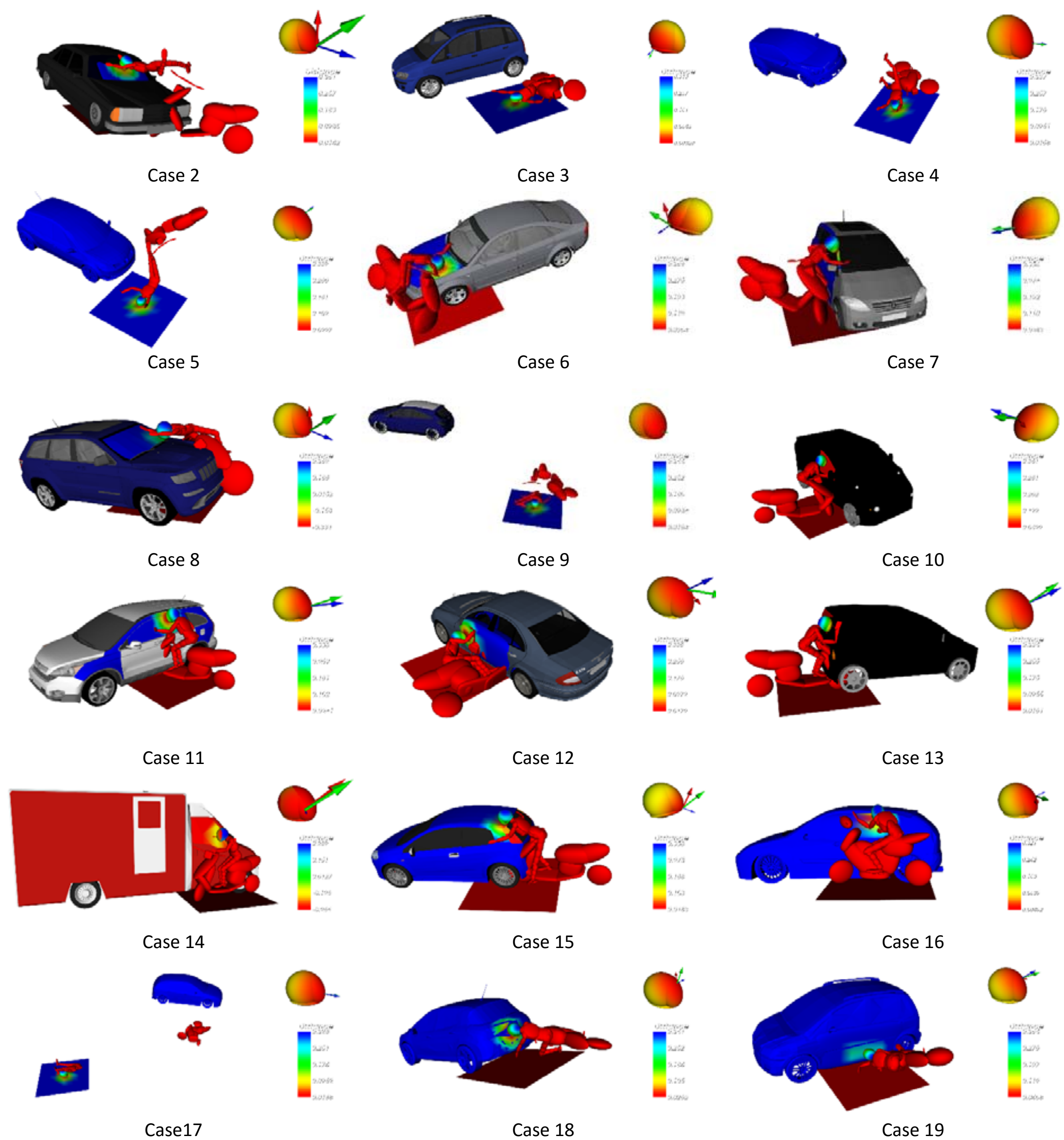


Fig. 7. Illustration of the head impact condition of all of the 19 accident cases.

IV. Discussion

In the current ECE R22.05 standard test, impact conditions are characterised by a linear impact on points B, P, R, X with a velocity of 7.5 m/s and against point S at 5.5 m/s, as illustrated in 0. Impacts are conducted at two temperatures (-20°C and +50°C) and one wet (+20°C) condition. The helmeted head impacts two anvils, a flat one and a kerbstone-shaped one. The boundary conditions of the head at neck level are characterised by a free falling head. In addition to this linear impact, a tangential impact test is also prescribed, with an initial speed of

8.5 m/s against a 75° inclined anvil. No headform response is recorded for this test as only the tangential force applied to the anvil is recorded.

If the head initial linear velocity seems to be reasonable with regard to real-world accident situations and an impact speed against which it is possible to provide protection with current helmet thickness, the fact that this velocity has only a normal component is not acceptable.

Several proposals for helmet standard evolutions have been established in previous projects, such as COST 327 [13] and APROSYS. Table II reports impact velocity and angle for motorcycle and other helmeted users, such as bicyclists or horse rider. The reported angles are defined as the angle between the anvil and vertical axis in case of drop test, which correspond to the angle between velocity vector and surface.

TABLE II

SYNTHESIS OF DROP VELOCITY AND ANGLE FROM REAL ACCIDENT CASES

Helmet	References	Drop velocity [m/s]	Anvil / Drop axis[deg]	Surface
Motorcycle	Otte *et al*. 1999	12	<30	Side of car or road
Equestrian	Mellor and Chinn 2006	9	37	Hard grass
Bike	Vershueren 2009	5.3	40	Road
	Bourdet *et al*. 2013	6.8	60	Car
	Bourdet et al. 2012	6.7	55	Road
		10.2	33	
Motorcycle	Real-world cases	11.1	44	Road and car

Accident reconstructions reported in this study demonstrate that a significant tangential velocity exists which leads to head rotational acceleration in addition to the linear acceleration. Therefore, the protecting capability of a helmet should be assessed under a combined linear and tangential impact. It is important to mention that very few motorcyclist accidents occur at temperatures as low as -20°C as such a low temperature is exceptional in the EU, so only very few motorcyclists travel in such extreme cold temperature. It is therefore no longer acceptable that helmet optimisation includes material behaviour at such low temperatures. The outcome is that current helmet design boasts optimal helmet performance at -20°, which presents no optimal protection at temperatures for which accidents actually occur.

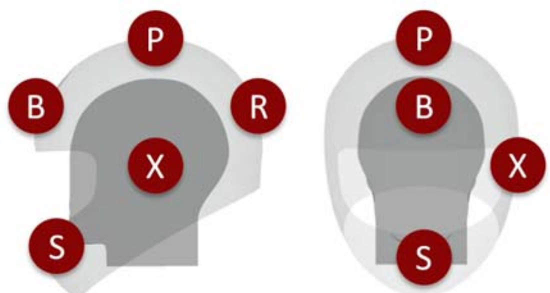


Fig. 8. Representation of the ECE R22.05 impact points with an impact velocity of 7.5 m/s for B, P,R, X and of 5.5 m/s for point S.

The current headforms are characterised by a non-deformable “head-shaped” mass. The value of the mass includes “some” neck effects and its rotation inertia is not controlled. The current ISO headform exist in several sizes and is fitted with a 3D linear acceleration amount. Simply adding rotational accelerometers would be difficult and ineffective due to the non-controlled rotational inertia of the headform. It should be mentioned that existing simple headforms, such as the Pedestrian ISO headforms, have a more realistic mass but present inertia quite far from human head inertia, as shown in Table III. Therefore, this kind of headform would not be a viable alternative. On the contrary, the Hybrid III dummy head presents a mass and inertia close to the human head, an essential aspect if rotation is considered. Moreover, a rubber skin models the deformation of the head during an impact as reported in the study of Hubbard *et al.* in 1974 [36], an essential aspect for helmet optimisation.

As mentioned earlier, a tangential impact is proposed in ECE R22.05. However, as only the tangential force at anvil level is recorded, this measure only describes the friction coefficient of the helmet, but does not permit an evaluation of the capability of the helmet to protect the brain against angular loading. Moreover, the threshold of 3.5 kN has only a very poor link to the head injury outcome.

TABLE III

SYNTHESIS OF HEADFORM INERTIAL PROPERTIES AND COMPARISON WITH HUMAN HEAD CHARACTERISTICS

	Mass [kg]	I_{xx} [kg.m²]	I_{yy} [kg.m²]	I_{zz} [kg.m²]
ISO Pedestrian	4.5	11.10^{-3}	11.10^{-3}	$110.5.10^{-3}$
Hybrid III 50th	4.5	$17.088.10^{-3}$	$18.872.10^{-3}$	$22.685.10^{-3}$
Human Head	4.5	$17.996.10^{-3}$	$18.360.10^{-3}$	$21.902.10^{-3}$
ISO Helmet	5.7		Not controlled	

Following to the previous critical analysis of head impact condition, a proposal can be made. First, two temperatures (0°C and +50°C) and one wet condition (at +20°C) are proposed. No change is suggested for the linear impact velocity, which should remain at 7.5 m/s. However, it would be important to replace the ISO headform with a fully 6D instrumented HIII dummy head in order to record any possible rotational acceleration. It is proposed to introduce a set of tangential impacts, based on results from the accident reconstruction investigations, as illustrated in 0. For each helmet it is suggested to conduct a set of four drop tests at an impact velocity of 8.5 m/s and with an anvil angle of 45° with a controlled friction paper (abrasive paper of 80 gr). The head will drop freely, which means that there will not be neck attached to the head.

The first two impacts are tangential impacts in the sagittal plan (Fy+ and Fy- points), leading to rotation around the Y-axis in extension and flexion direction. The next two tangential impacts are located at parietal level (Lx and Lz) and will be applied in the frontal plane, one introducing rotation around the X (postero-anterior) direction and one introducing a rotation around the Z (vertical ascendant) direction.

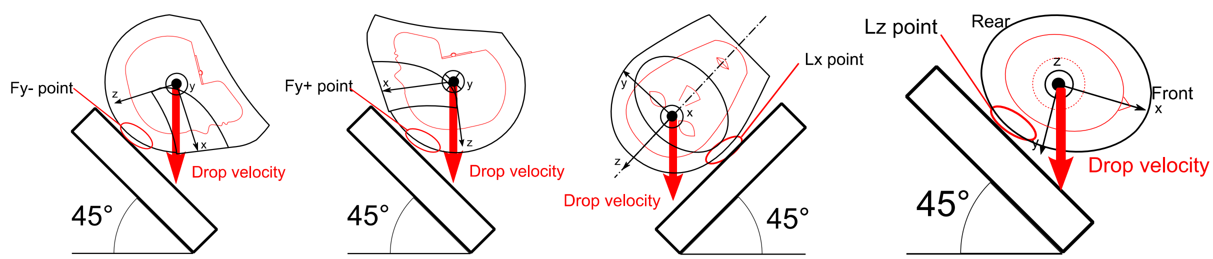


Fig. 9. Illustration of the four tangential impact conditions introducing angular acceleration around Y axes (called Fy+ and Fy-), X axes (called Lx) and Z axes (called Lz).

It is further recommended to replace ISO headforms with Hybrid III dummy head and to record linear and rotational acceleration versus time. In the present analysis, Hybrid III 50th head (dimension 597 mm) will be used, for helmets within 56–61 cm in size. A critical point that must be managed is the sizing aspect, as currently a number of sizes are available for the ISO 960 headform. This can be solved through the different versions of the Hybrid III dummy heads, as illustrated in Table III. As it is well known that sizes A, C, E, J, M and O represent 95% of sizes used in the standard test for helmet homologation, only these sizes would be of interest. From Table IV it appears that these sizes would be adequately covered by five sizes of the Hybrid III heads family.

In the literature, head rotational acceleration limits are proposed at 8–10 krad/s². It must be recalled, however, that head tolerance limit to rotation is strongly time- and direction-dependent, so that today there is no single known limit. In addition, in case of tangential impact rotational acceleration is combined with linear acceleration so that the head sustains a complex 6D kinematic.

Today, state-of-the-art FE head models exist and have been used for the definition of injury criteria to specific injury mechanisms. These models became much more powerful injury prediction tools than HIC, so the present proposal is to implement improved, model-based head injury criteria into a new helmet impact test procedure. Based on the simulation of 120 well-documented head trauma, tolerance limits have been identified with respect to moderate AIS2+ neurological injury [37]. Human brain tolerance limits relative to neurological injuries with a risk of occurrence of 50% were established for an axon strain of 15%.

TABLE IV

COMPARISON BETWEEN THE EN 960 HEADFORMS CIRCUMFERENCES AND HYBRID III DUMMY HEADS. THE A, C, E, J, M AND O SIZES REPRESENT 95% OF SIZES USED IN GLOBAL STANDARDS AND ARE COVERED BY THE HYBRID III HEADS FAMILY

EN 960 headform size	Head circumference [mm]	Dummy model	Head circumference [mm]
A	500	Hybrid III 3 Year Old	508
B	510		
C	520	Hybrid III 6 Year Old	520.7
D	530		
E	540	H III 5th Female (or 10 years)	538.5
F	550	H III 5th Female (or 10 years)	538.5
G	560		
J	570	Hybrid III 95th Large Male	584
K	580		
L	590		
M	600	Hybrid III 50th Male	597
N	610		
O	620	Hybrid III 50th Male	597
P	630		
Q	640		

In the proposed approach, the experimental linear and rotational head acceleration versus time will constitute the inputs that will drive the head FE model, which in turn will compute the injury parameters related to neurological injury. By this methodology it will be possible to assess the brain injury risk by means of a coupled experimental versus numerical testing procedure, as illustrated in 0.

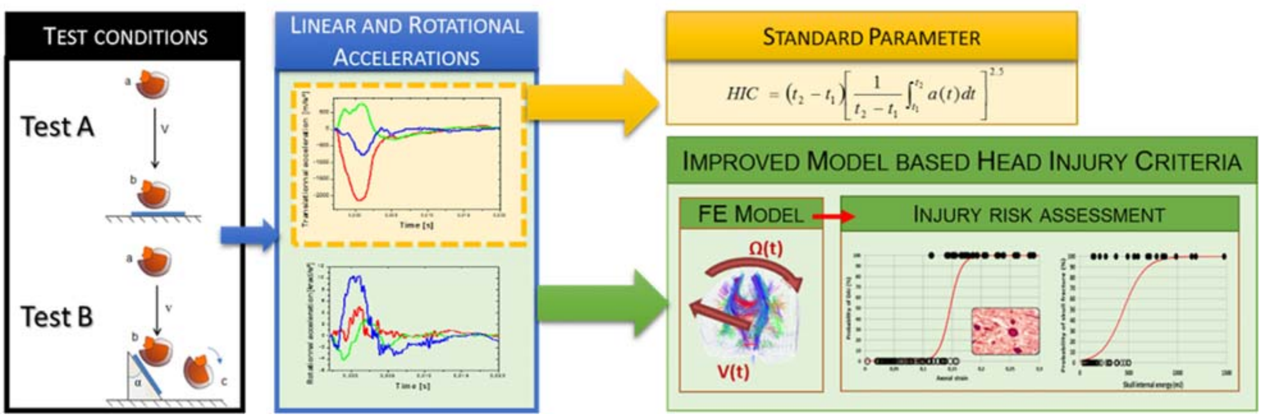


Fig. 10. Illustration of the coupled experimental versus numerical head impact test method based on novel model-based head injury criteria.

V. CONCLUSIONS

The presents study constitutes a proposal for a possible evolution of the current ECE R22.05 motorcycle helmet standard. Three key aspects have been critically reviewed: head impact conditions; head surrogate; and head injury criteria. The simulation of 19 real-world motorcycle accidents, in which the kinematics of the victim was computed in order to extract the head impact velocity vector demonstrated that the velocity vector contains a significant tangential component. Therefore, more realistic head impact conditions are proposed by implementing a tangential head impact test. Further improvement could be made with regard to the head surrogate, if the ISO headform is replaced by the Hybrid III head. This head has the advantage of more realistic inertial properties and interface characteristics between headform and helmet. A final key recommendation that is proposed concerns the assessment of the head injury risk. For this, a coupled experimental versus numeric method is proposed in order to introduce tissue-level head injury criteria. It is expected that the evolution of the standard helmet test method will enable advanced helmet evaluation and optimisation against biomechanical criteria under more realistic impact conditions.

VI. Acknowledgements

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