Protection capability of bicycle helmets under oblique impact assessed with two separate brain FE models

Deck C., Bourdet N., Halldin P., DeBruyne G., Willinger R.

Abstract The present study proposes a bicycle helmet evaluation under oblique impact based on a coupled experimental versus numerical test method using two separate brain FE models. For each of the 17 helmet types three oblique impacts have been conducted and the 6D headform acceleration curves have been considered as the initial conditions of the brain injury risk assessment based on the FE simulation. The study gives a new insight into helmet protection capability under oblique loading and shows that adequate protection is offered by most of the helmets when impacts leading to rotation around X and Y are concerned. However when impact leads to rotation around Z axis the protection is critical for nearly all helmets. The study considers two separate brain FE models for the assessment of brain injury risk and thus permits a comparative analysis of brain FE modeling. When impact induces rotation around X and Y axis the computed results are comparable. However when rotation around Z axis are concerned significant differences are observed which demonstrate that further efforts are needed in the domain of model based brain injury criteria harmonization.

Keywords bicycle helmets, test method, oblique impact brain FEM, Head injury criteria.

I. INTRODUCTION

The head represents the most vulnerable part of the human body and one of the most important to be protected. At EU level, recent statistics show that 2200 to 2400 fatalities occur from bike accident every year in Europe (European Road Safety Observatory http://www.erso-project.eu/). Between 21 to 61% of the victims of all bicycle accidents seeking medical care in New Zealand have a head injury as reported by Collins et al. in 1993 [1] and Eilert-Petersson & Schelp in 1997 [2].

According to Rizzi et al 2013 [3]and Amoros et al 2012 [4] including 55200 reported accidents in Sweden and 13797 accidents in France, the most common accident situation for a bicyclist is the so called single accidents i.e. the bicyclist impacts the ground and not a car. A recent accident survey conducted by the Insurance Institute for Highway Safety in US reported 722 fatal bicycle accidents in 2012. Only 17% of this population worn a helmet. Previous studies have outlined the effectiveness of wearing bicycle helmets [3-4] and also the possibility of the improvement of currently used helmets design [5-6]. It is well known in the scientific community that head rotational acceleration is a critical head loading which can lead to brain injury. Concerning neurological injuries, Holbourn (1943) [9] suggested that the rotational acceleration induced by a given impact causes high shear strains in the brain, thus rupturing the tethering cerebral blood vessels, neo and subcortical tissue. This author was the first who pointed the importance of rotational acceleration in the appearance of cerebral concussion. In 1967, Ommaya et al. [10] 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 expect to corresponds to an angular acceleration of 7 500 rad/s² for human.

Ommaya et al. (1968) [11] studied the effect of whiplash injury on rhesus monkeys and showed that if the head was subjected to a rotational acceleration above a threshold value, subdural and subarachnoid injuries were obtained. In a study based on primates, Gennarelli et al. (1982) [12] suggested that a rotational acceleration exceeding 17 500 rad/s² would produce SDH in the rhesus monkey. With the objective to investigate 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 with and without

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considering the angular rotation by Deck et al. 2007 [13]. 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 intra-cerebral loading, supposed to lead to neurological injuries. In this study the effect of angular acceleration was found to increase the intra cerebral shearing stress for all accident cases considered of about 50% whatever the impact severity was. Kleiven et al. 2007 [14] as well as Zhang et al. 2001 [15] 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.

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 accident in order to demonstrate that a tangential loading of the head does exist in addition to the normal impact velocity. Mills et al. (1996) [16] showed that oblique impacts are the most common situations in motorcycle crashes. More recently Bourdet et al. (2012) [17] quantified the head tangential component of the head impact, by reconstructing real world and virtual bicycle accidents.

Despite this widely recognized understanding of head tangential loading on one hand and the effect of the induced rotational acceleration to the brain on the other hand, no standard head protection is currently considering head rotational acceleration. Only ECE R22.05 EU [18] 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 [16], [19]–[21] but none of them consider the time evolution or rotation direction of this parameter. To the author's opinion, the only way to integrate the complexity of brain geometry and brain material properties is to progress towards tissue level brain injury criteria as proposed in existing FE model based head injury criteria [14], [15], [22]. In 2004 Deck demonstrates that helmet optimization strongly depends on the head substitute and injury criteria taken into account. In the domain of bicycle helmet optimization Milne et al 2012 [23] and 2013 [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 2013 [25] in the context of the development of an advanced 'honeycomb" bicycle helmet.

In order to progress in the field of helmet protection against tangential impacts a number of experimental attempts were proposed in the literature. Aldman et al. 1976 [19] dropped a helmeted headform fixed to a dummy neck against a rotating steel disc. In 2001 Halldin et al. [20] designed an 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. 2011 [21] 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 itself fixed to a 20 kg mass which drops against a sliding plate. Advanced model based head injury criteria have also been suggested in recent attempts to improve bicycle helmet test methods (Deck et al 2012) [26]. More recently Willinger et al. [27] and Halldin et al. [28] proposed new experimental test including oblique impacts.

It is this later test method which has been applied by Stigson et al. [29] in the context of a bicycle helmet consumer test published by FOLKSAM. A total of 17 different helmet types have been impacted under oblique impact conditions and the headform response has been recorded in terms of linear and rotational acceleration versus time. In this context the aim of the present study is to apply the coupled experimental vs numerical helmet test method to a significate number of existing helmets in order to express the level of protection offered by current helmets, based on advanced brain injury criteria using two state of the art brain FE models The experimental 6D headform responses are considered in the present paper in order to drive the KTH (KTH-Royal Institute of Technology, Stockholm) brain FE model reported by Kleiven et al. [14] as well as the SUFEHM (Strasbourg University FE Head Model) model published by Sahoo et al. [22] and to assess the brain injury risk with these two separate brain FE models. As a whole the objective of the present study is to assess the protection capability of existing helmets under complex loading and to contribute to the development of novel helmet test methods.

II. METHODS

A total of 17 bicycle helmet types was tested under vertical drop against a 45 degree impact surface as reported by Stigson et al. [29]. For each impact location only one experimental impact was conducted. The helmets were attached to a 50% HIII head and tested in different impact locations introducing rotation around the three reference axis called Xrot, Yrot and Zrot, as suggested by Halldin et al. [28] and illustrated in figure 1. In the present study, the three linear and rotational accelerations versus time were considered in order to apply a coupled experimental versus numerical test method published by (Deck et al 2012, Milne et al 2013, Willinger et al 2014, Halldin et al. 2015, Bourdet et al. 2016) and shown in figure 2. To do so, the 6D head kinematic was considered as the input for two separate brain computational models, in order to compute intra-cerebral stain and to assess injury risk separately with two models, the KTH model and the SUFEHM model.

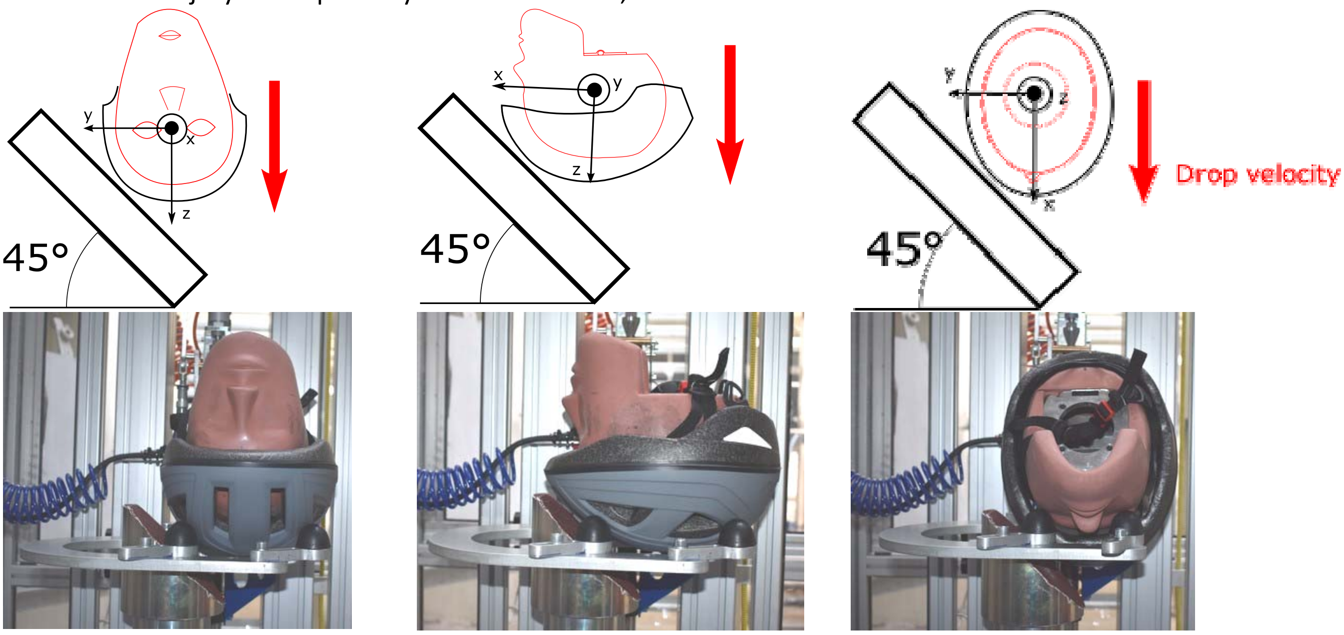

Fig. 1. Illustration of the three oblique impact tests leading respectively to rotation around Y, X and Z axis.

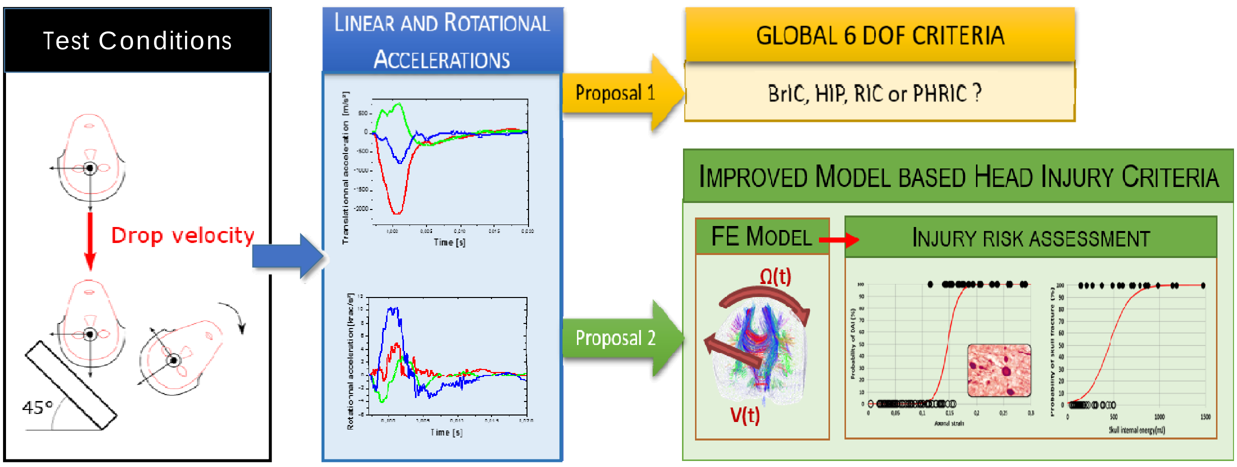

Fig. 2. Description of the coupled experimental versus numerical helmet test method.

KTH model (figure 3) was comprised of 19350 nodes, 11454 eight-node brick elements, 6940 four node shell and membrane elements, and 22 two-node truss elements. The total mass of the head is 4.5 kg. Mooney- Rivlin hyperelastic constitutive law was used for the isotropic brain model with addition of second order Proney series to account viscosity. This model was used for the simulation of 58 American football impacts by Kleiven et al 2007 [14]. With this isotropic brain model a threshold for a 50 % risk of mild concussion has been established for a critical value of the Maximum Principal Strain (MPS) of 26% for the white matter [14]. In this earlier study mild concussion was estimated in the context of American Football player for which concussion was identified by sport medics on the field. This mild concussion was associated to MAIS 1 brain injury.

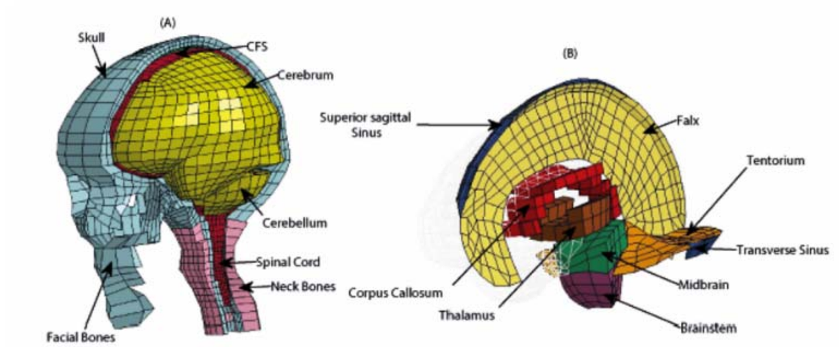

Fig. 3. Illustration of the isotropic KTH head model [6]

SUFEHM is another full validated head FE model that presents an anisotropic brain structure. Concerning the cerebral structure, the 3D directions of the main axon fibers have been implemented into the brain model, based on MRI medical imaging, and more precisely on Diffuse Tensor Imaging (DTI) as shown in Figure 4. It is important to mention that model permits the computation of axon elongation in case of impact. In order to establish brain tolerance limits a total of 125 well-documented real world head trauma have been simulated with this second head model in order to establish brain injury thresholds in terms of axon strain. The regression analysis demonstrated that the critical value (for a 50% risk of short term coma) is an axonal strain of 15%. In this previous study brain injury was characterized by the reporting of loss of consciousness on the accident scene as within the accident databases. The proposed investigation distinguished between existing loss of consciousness and no loss of consciousness and corresponds to an AIS2 brain injury. More details concerning this second model and the related brain injury criteria can be found in Sahoo et al. 2016 [22].

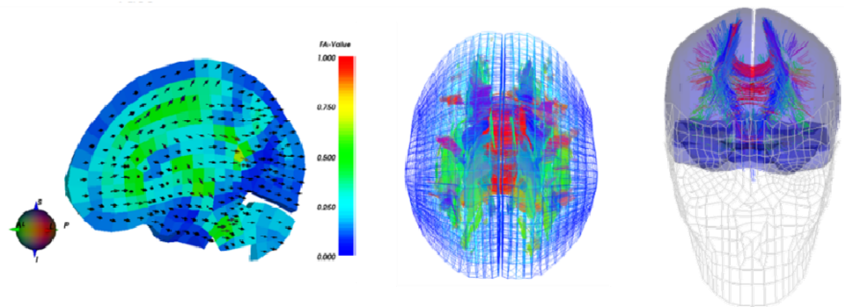

Fig. 4. Illustration of the SUFEHM and the main axon fiber bundles implemented into the anisotropic brain model.

Even if the brain injury criteria proposed by both models are model-dependent the assessment of brain injury risk for each impact can be expressed by a relative risk expressed by equation 1

$$R = BS/BSmax \qquad (1)$$

Where BS is brain strain and BSmax is the maximum brain strain for a given brain model and leading to a 50% risk of brain injury.

Therefore R will be expresses as Rk and Rs respectively for KTH and SUFEHM as expressed in in equation 2 and equation 3.

The advantage of the form of the expression of brain injury risk is to relate the computed

$$Rk = MPS/26 \qquad (2)$$

$$Rs = MAS/15 \qquad (3)$$

The advantage of this expression of the brain injury risk is that the output of two different brain models can easily be compared in terms of brain injury risk and that the result for a given impact can be R=1 (50% risk of injury) as the brain strain is equal to the critical brain strain and vary from R=0 (brain strain is half of critical brain strain) up to R=2 (brain strain is twice the critical brain strain)

III. RESULTS

For the 51 impacts (17 helmets * 3 impact locations), the brain response computed separately with both FE models lead to brain strains ranging typically from 10% to 20%, with some extreme values of 35 %, as reported in figures 5 to 7.

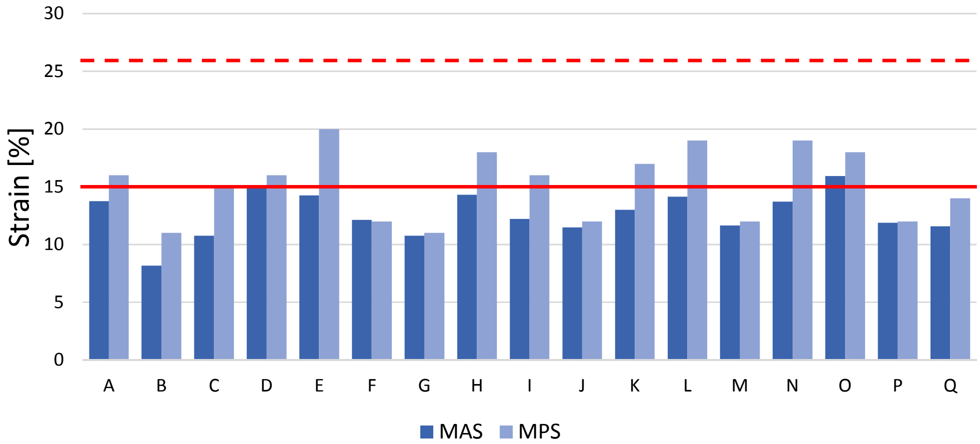

Fig. 5. Superposition of the MAS and MPS computed respectively with SUFEHM and KTH brain models under Xrot Impact. Red line show the MAS-15 and dashed red line MPS-26 critical threshold for SUFEHM and KTH models respectively.

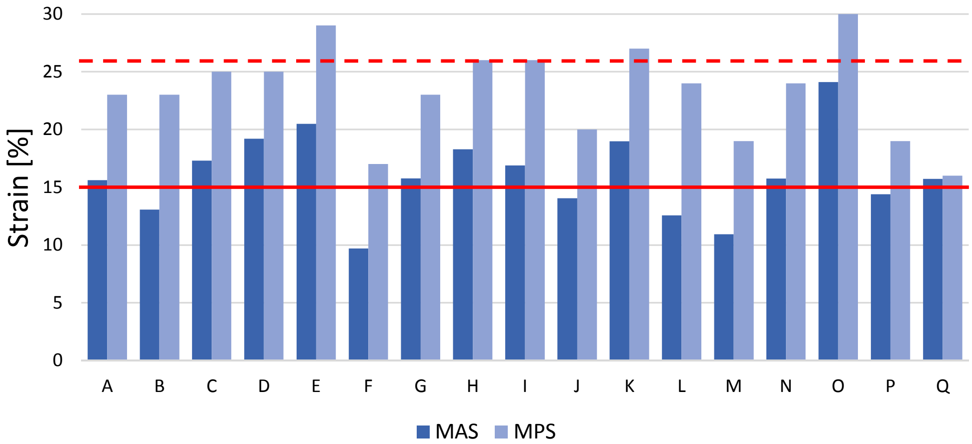

Fig. 6. Superposition of the MAS and MPS computed respectively with SUFEHM and KTH brain models under Yrot Impact. Red line show the MAS-15 and dashed red line MPS-26 critical threshold for SUFEHM and KTH models respectively.

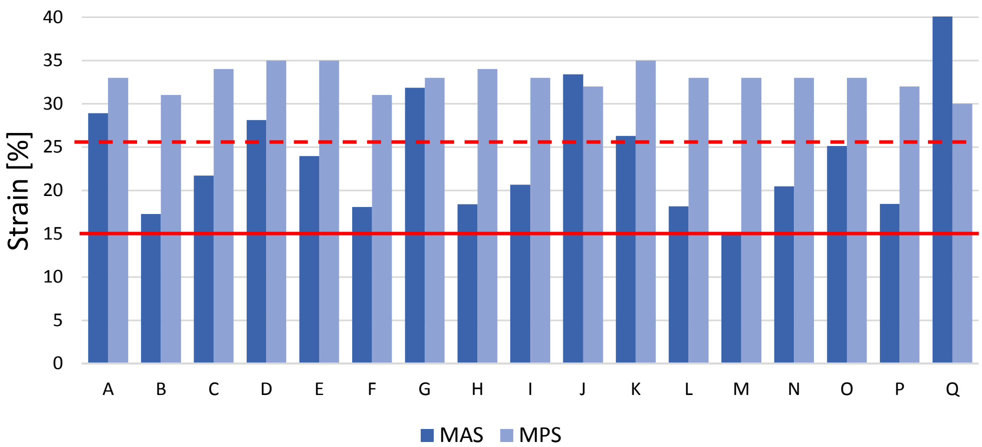

Fig. 7. Superposition of the MAS and MPS computed respectively with SUFEHM and KTH brain models under Zrot Impact. Red line show the MAS-15 and dashed red line MPS-26 critical threshold for SUFEHM and KTH models respectively.

A more detailed analysis of figures 5 to 7 permits an analysis per rotation direction and shows following result in terms of MAS for SUFEHM and MPS for KTH model.

Xrot : MAS 7% to 16% and MPS 12% to 20%

Yrot : MAS 9% to 24% and MPS 16% to 30%

Zrot : MAS 16% to 40% and MPS30% to 35%

For the reported helmet impacts and the induced brain loading, the results show that quasi systematically the computed MPS values are significantly higher than MAS levels. Further it can clearly be observed that impact direction leads to increased brain strains when ranging from Xrot to Yrot and Zrot. The results also show that important differences exist between helmets as for example the Yrot impact (figure 6) reports lowest MAS (respectively MPS) of 9% (respectively 15%) to highest of 24% (respectively 30%).

Further the combined KTH vs SUFEHM helmet evaluation can be analyzed in terms of relative risk of brain injury called R and computed with both, KTH and SUFEHM models. Figure 8 to 10 report the relative risk computed for all impacts with both FE models. Results show that for lateral impacts leading to rotation around X axis (Xrot), the nearly all helmets show brain injury risk being clearly below a 50% risk for both head FE models (Just one presents a MAS slightly over 15%) . For frontal impact (Yrot) leading to rotation around Y axis (left-right), most of the helmet lead to injury risk close to 50% for both head FE models. Finally the lateral impacts, leading to rotation around vertical Z axis (Zrot) were the most critical ones for most of thehelmets as the injury risk was clearly over 50% for both head models. Only helmet M presents a MAS close to 15%.

Regarding this relative risk, more detailed analysis of figure 8 to 10 leads to following results per impact direction:

For Xrot, the impacts leading to rotation around X axis, all relative risks are under 1 as shown if figure 8, illustrating that the protection level of the different helmets is acceptable for this kind of impact. The mean relative risk for both brain FE models is 0.7 which means that typically the brain injury risk is under 50%. A further analysis of figure 8 also reveals that the relative risk computed with SUFEHM (called Rs) is higher compared to Rk computed with KTH model. Mean values are respectively 0.8 and 0.6 respectively for Rs and Rk. Finally a linear regression analysis showed that the correlation factor between Rs and Rk is 0.79, which demonstrates that both brain FE models lead to comparable results.

For the Yrot impacts, leading to rotation around Y axis, the relative risks computed for the different helmets are close to 1 as shown if figure 9, illustrating that current helmets protect reasonably well for this kind of impact. The mean relative risk for both brain FE models is 1, which means that typically the brain injury risk is close to 50%. A further analysis of figure 9 also reveals that the relative risk computed with SUFEHM (called Rs) is higher compared to the one computed with KTH model and called Rk. Mean values are respectively 1.1 and 0.9 respectively for Rs and Rk. Finally a linear regression analysis showed that the correlation between Rs and Rk is 0.80, which demonstrates that both brain FE models lead to comparable results.

For the Zrot impacts leading to rotation around Z axis, the relative risks of the different helmets are clearly higher than 1 as shown if figure 10, illustrating that the protection capability of current helmets is critical for this kind of impact. The mean relative risk for both brain FE models is over 1, which means that typically the brain injury risk is over 50%. A further analysis of figure 10 also reveals that the relative risk computed with SUFEHM (called Rs) is higher as the one computed with KTH model (called Rk). Mean values are 1.5 and 1.3 respectively for Rs and Rk. Finally a linear regression analysis showed that the correlation between Rs and Rk is very low and close to 0.1, which demonstrates that both brain FE models lead to quite different risk evaluations.

As a whole, when considering all of the 51 impacts and the related Rs and Rk value, a correlation coefficient of 0.82 was found between the outcomes computed with both brain FEMs. Even if absolute values are different for both models, the trend is typically respected as shown in figures 8 to 10.

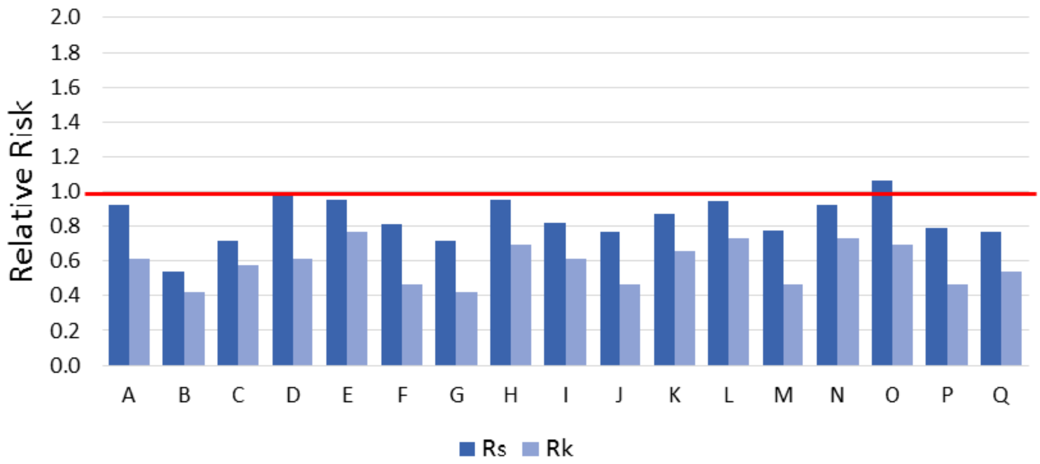

Fig. 8. Superposition of the Relative Risk of brain injury based on SUFEHM and KTH computations for XRot impacts, for the whole set of 17 helmets.

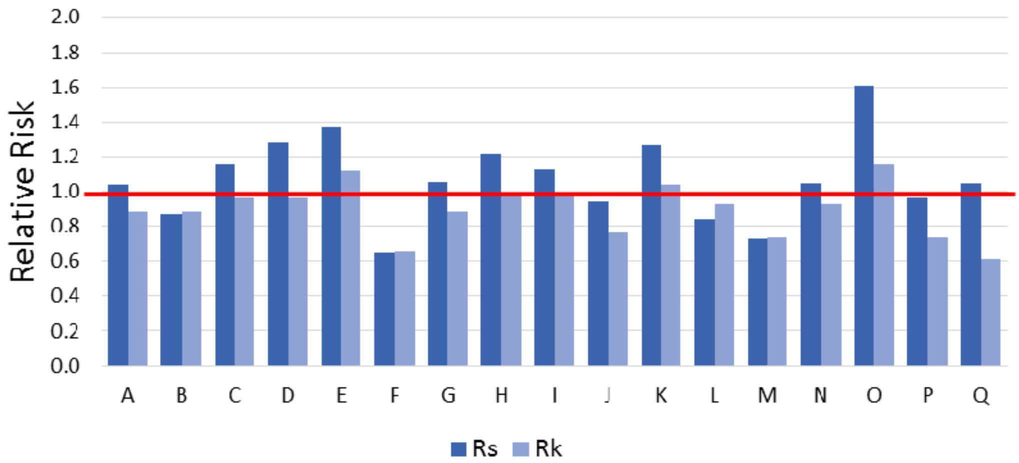

Fig. 9. Superposition of the Relative Risk of brain injury based on SUFEHM and KTH computations for YRot impacts.

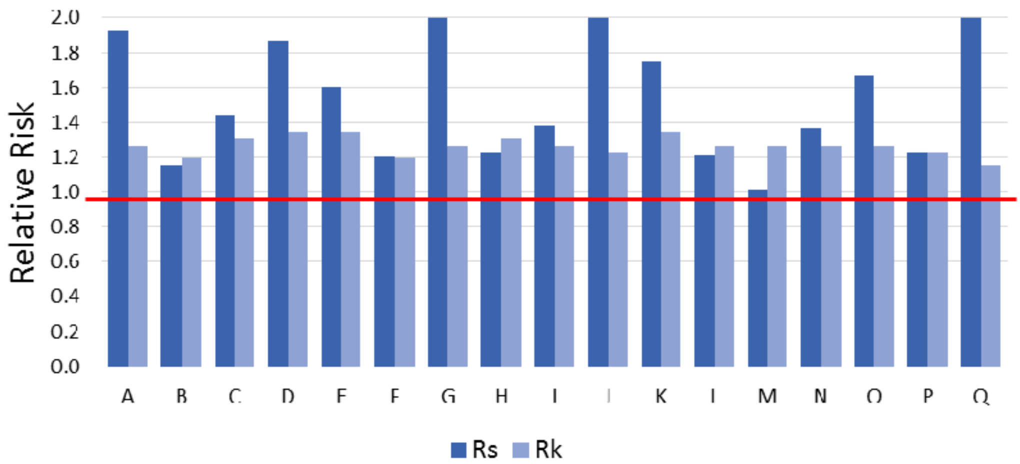

Fig. 10. Superposition of the Relative Risk of brain injury based on SUFEHM and KTH computations for ZRot impacts. , for the whole set of 17 helmets.

IV. Discussion

The present study proposes a helmet protection assessment under oblique impact based on a coupled experimental versus numerical test method using two separate brain FE models. It therefore permits an evaluation of helmet performance but also a critical analysis of the test method as two separate brain FE models are used.

At experimental level it should be mentioned that each impact was conducted only once by Stigson et al [29]. The unique 6D acceleration head response was than considered as input data for the computation of both Brain FE models in order to assess the brain injury risk. It is obvious that a consolidated helmet test method should involve at least three or more tests per impact location.

For the reported helmet impacts and the induced brain loading, the results show that quasi systematically the computed MPS values are significantly higher than MAS levels. This result can be explained by the fact that MAS is computed with an anisotropic “axon-reinforced material” when MPS is based on an isotropic brain constitutive law. With an earlier version of SUFEHM (Deck et al 2008 [30] critical MPS was estimated at 31% instead of 15% for MAS. Recently the KTH model was improved by implementing anisotropic brain constitutive laws by Giordano et al. 2014 [31] and similar results were shown, i.e. a critical MAS value of 10 to 15%. It would be important in a further step to also consider this updated model. A further critical aspect of the present study is that the injury criteria proposed by KTH model is related to mild concussion (AIS 1), when SUFEHM proposes a threshold for initial loss of consciousness which corresponds to AIS2. This demonstrates that furth harmonization is needed for the development of advanced helmet test methods.

Coming to results in terms of relative brain injury risk, both brain FE models lead to comparable results for all impact direction but not for the impact leading to rotation around Z axis. For this type of impacts the assessment of brain injury risk was significantly different for both models, with a very low correlation coefficient. Considering SUFEHM, the relative risk presents a very wide discrepancy between the different helmets (relative risk ranging from 1.1 to 1.9) when KTH model proposes a smaller spread out with relative risks that are very close to 1.3 for all the helmets. This observation could lead to suggest that SUFEHM is too sensitive or that KTH model is not enough sensitive to this kind of loading. A possible explanation of this observation is that SUFEHM is an anisotropic model with main axon bundle located in the central area of brain when KTH model considers an isotropic brain constitutive law. To conclude on this issue, clearly further investigation and harmonization efforts are needed.

The limitations exposed in the present discussion, especially when the rotation around Z axis is concerned does not permit a definitive conclusion concerning helmet evaluation for this kind of impact. It also does not permit to solve the brain computation issues that may exist for one or the other or even both of the models. Even if this aspect represents a limitation, the present paper reports a first effort towards model based brain injury criteria harmonization.

So far it was demonstrated that typically helmets protect adequately against the Xrot and Yrot impacts, but that the protection against the Zrot is very critical. It is expected that this critical protection is linked to the helmet geometry and helmet mechanical properties, but also due to the human head geometry and internal brain structure.

This study shows that important differences exist in brain loading induced by the different helmets, which will definitively lead to different head protection capability. It therefore demonstrate the need of advanced test methods for new helmet test standards as well as for comprehensive helmet rating methods. ”.

V. CONCLUSIONS

The protection capability of a set of 17 bicycle helmets available on the market has been assesed under oblique impact according to model based brain injury criteria. The experimental 6D acceleration versus time headform responses recorded by FOLKSAM (Stigson et al [29] were transmitted to KTH-University Stockholm and University Strasbourg for a separate computation of the brain response with their brain models known respectively as KTH and SUFEHM models.

This coupled experimental versus numerical helmet test method involving oblique impacts has been applied to the 17 bicycle helmets. Being impact at three different locations in order to induce rotation around the three reference axis. With both brain FE models the brain injury risk was assessed. Main conclusions are:

- Maximum Principal Strains (MPS) computed with KTH model are higher than Maximum Axon Strains computed with SUFEHM
- For helmet impacts leading to rotation around X and Y directions both models demonstrate an acceptable protection as the brain injury risk are around or below 50%.

- For helmet impacts leading to rotation around the vertical Z axis the protection of the head seems to be critical as both models demonstrate brain injury risk over 50% for most of the helmets.
- Correlation coefficients between results computed with KTH model and SUFEHM are close to 0.8 for the Xrot and the Yrot impacts. However it is very low when the Zrot impacts are concerned.

This study shows that helmets performance varies significantly from one helmet to the other.

As mentioned in the introduction a number of early attempts were made in order to improve helmets test methods. In 2011 Ghajari et al. [32] conducted linear and oblique helmet impacts by considering separately head alone and full body models. In this study authors concluded the upper body effect could be taken into account by adding mass and inertia to the headform. However this study also mentioned that the added mass or inertia are very impact direction dependent and that an added mass could lead to more rigid helmet design. Klug et al 2015 [33] applied a new helmet testing protocol to a set of helmets by adding mass and inertia in order to replicate the influence of upper body and neck. These authors tested a number of helmets under oblique impact (30° inclined anvil) and extracted no less the 11 head injury criteria based on global kinematic parameters. It was concluded that it was not possible to recommend a specific set of injury criteria and that focus should be on rotational acceleration and body mass effect in further studies. As a continuation, Feist et al. 2016 [34] conducted a parametric study using three different helmeted full human body FEM under oblique impact. Simulations showed that head loading not only depends on neck kinematics but also on thoracic and even lumbar spine impact responses. This means that an accurate neck model is not enough for accurate helmet testing. These simulations also showed that the effective impact mass is 10–15% higher if the full body simulations in comparison to head-only simulations and that the inertia must not be changed to replicate rotational peak velocity. To replicate rotational peak acceleration, though, the inertia needs to be increased by 20% about y-axis, and decreased by 20-40% about x-axis. Finally these authors mentioned that the added mass or inertia would also depend on the considered injury criteria.

In a similar way, Fahlstedt et al. 2016 [35] investigated the influence of the neck and the upper body on head kinematics and brain injury risk for unhelmet and helmeted impacts based on a parametric study conducted with a FE model of the helmeted (or unhelmet) THUMS-KTH human body model. Results showed a variation among the impact situations and helmet designs. The studied cases showed for the first 15 ms an average ratio between head only versus head plus upper body and neck for peak brain tissue strain of 1.04, for peak linear acceleration of 1.06, for peak angular acceleration of1.08 and for peak angular velocity of 1.05. The study also showed that the influence of neck and body is highly dependent on impact direction, helmet design and neck muscle activation. In this context of very complex positive or negative upper body and neck effect it was decided in the present study, but also in the standard body's working group CENT-TC158WG 11 to consider head alone impact against 45° inclined anvil. Even if this option is not totally able to reproduce complex real world accident loading it is hypothesized that it constitutes an optimal helmet loading which permits to evaluate its protection capability under linear and tangential loading.

In US, where rigid-guided headform boundary conditions are considered, Becker et al 2015 [36] compared Snell M2010 requirements to DOT-FMVSS2018 standard. If it is possible to compare helmet by considering headform linear acceleration only, it must be mentioned that global head kinematic parameter do not take into account the impact direction, the complex time evolution or the combined effect of linear and rotational loading in case of oblique impact. It is therefore suggested in the present study to focus on tissue level injury criteria related to local brain tissue loading whatever the 6D head kinematic is.

Nowadays a number of state of the art head FE models exist and an increasing number of them are used for real world head trauma simulation in order to derive tissue level injury criteria. Arguments for tissue level brain injury criteria are:

- Threshold for moderate, reversible brain injury
- Under linear impact this criteria permits to distinguish frontal impact from lateral and occipital impacts.
- Under oblique impact this criteria takes into account the 6D time evolution of the brain loading, i.e. the combined rotational plus linear effect for the different impact directions.

To the author's knowledge it is the first time that helmet evaluation under oblique impact is conducted based on model based injury criteria derived from two brain models. This study is therefore a first step towards novel experimental versus virtual helmet test methods.

VI. ACKNOWLEDGEMENT

The authors which to thank FOLKSAM for the provision of the experimental helmet impact tests results.

Fondation MAIF is acknowledged for their support in the context of EuroNcasque project as well as the European Union –SmartHelmet Project.

VII. REFERENCES

[1] Collins Ba, Langley Jd, and Marshall Sw, "Injuries to pedal cyclists resulting in death and hospitalisation.," *N. Z. Med. J.*, vol. 106, no. 969, pp. 514–516, Dec. 1993.

[2] E. Eilert-Petersson and L. Schelp, "An epidemiological study of bicycle-related injuries," *Accid. Anal. Prev.*, vol. 29, no. 3, pp. 363–372, 1997.

[3] M. Rizzi, H. Stigson, and M. Krafft, "Cyclist injuries leading to permanent medical impairment in sweden and the effect of bicycle helmets," in *Chalmers Publication Library (CPL)*, 2013, pp. 412–423.

[4] E. Amoros, M. Chiron, J.-L. Martin, B. Thélot, and B. Laumon, "Bicycle helmet wearing and the risk of head, face, and neck injury: a French case--control study based on a road trauma registry," *Inj. Prev. J. Int. Soc. Child Adolesc. Inj. Prev.*, vol. 18, no. 1, pp. 27–32, Feb. 2012.

[5] M. Öström, U. Björnstig, K. Näslund, and A. Eriksson, "Pedal Cycling Fatalities in Northern Sweden," *Int. J. Epidemiol.*, vol. 22, no. 3, pp. 483–488, Jun. 1993.

[6] R. S. Thompson, F. P. Rivara, and D. C. Thompson, "A Case-Control Study of the Effectiveness of Bicycle Safety Helmets," *N. Engl. J. Med.*, vol. 320, no. 21, pp. 1361–1367, May 1989.

[7] S. Thomas, C. Acton, J. Nixon, D. Battistutta, W. R. Pitt, and R. Clark, "Effectiveness of bicycle helmets in preventing head injury in children: case-control study," *BMJ*, vol. 308, no. 6922, pp. 173–176, Jan. 1994.

[8] A. S. McIntosh, D. Kallieris, R. Mattern, N. L. Svensson, and B. Dowdell, "An Evaluation of Pedal Cycle Helmet Performance Requirements," SAE International, Warrendale, PA, SAE Technical Paper 952713, Nov. 1995.

[9] Holbourn A.H.S., "MECHANICS OF HEAD INJURIES," *The Lancet*, vol. 242, no. 6267, pp. 438–441, Oct. 1943.

[10] A. K. Ommaya, P. Yarnell, A. E. Hirsch, and E. H. Harris, "Scaling of Experimental Data on Cerebral Concussion in Sub-Human Primates to Concussion Threshold for Man," SAE International, Warrendale, PA, 670906, Feb. 1967.

[11] A. K. Ommaya, F. Faas, and P. Yarnell, "Whiplash injury and brain damage: an experimental study," *JAMA J. Am. Med. Assoc.*, vol. 204, no. 4, pp. 285–289, Apr. 1968.

[12] T. A. Gennarelli and L. E. Thibault, "Biomechanics of acute subdural hematoma," *J. Trauma*, vol. 22, no. 8, pp. 680–686, Aug. 1982.

[13] C. Deck, D. Baumgartner, and R. Willinger, "Influence of rotational acceleration on intracranial mechanical parameters under accidental circumstances," in *Proceeding of IRCOBI Conference*, Maastricht, The Netherlands, 2007.

[14] S. Kleiven, "Predictors for traumatic brain injuries evaluated through accident reconstructions," *Stapp Car Crash J.*, vol. 51, pp. 81–114, Oct. 2007.

[15] L. Zhang, K. H. Yang, and A. I. King, "Comparison of brain responses between frontal and lateral impacts by finite element modeling," *J. Neurotrauma*, vol. 18, no. 1, pp. 21–30, Jan. 2001.

[16] N. J. Mills and A. Gilchrist, "Response of helmets in direct and oblique impacts," *Int. J. Crashworthiness*, vol. 2, no. 1, pp. 7–24, 1996.

[17] N. Bourdet, C. Deck, R. P. Carreira, and R. Willinger, "Head impact conditions in the case of cyclist falls," *Proc. Inst. Mech. Eng. Part P J. Sports Eng. Technol.*, vol. 226, no. 3–4, pp. 282–289, Sep. 2012.

[18] "ECE 22.05, Uniform provision concerning the approval of protective helmets and their visors for driver and passengers of motor cycles and mopeds." .

[19] B. Aldman, B. Lundell, and L. Thorngren, "Non-Perpendicular Impacts - An Experimental Study On Crash Helmets," in *IRCOBI Conference Proceedings*, Amsterdam (The Netherlands), 1976.

[20] P. Halldin, A. Gilchrist, and N. J. Mills, "A new oblique impact test for motorcycle helmets," *Int. J. Crashworthiness*, vol. 6, no. 1, pp. 53–64, Jan. 2001.

[21] T. Y. Pang *et al.*, "Head and neck responses in oblique motorcycle helmet impacts: a novel laboratory test method," *Int. J. Crashworthiness*, vol. 16, no. 3, pp. 297–307, 2011.

[22] D. Sahoo, C. Deck, and R. Willinger, "Brain injury tolerance limit based on computation of axonal strain," *Accid. Anal. Prev.*, vol. 92, pp. 53–70, Jul. 2016.

[23] G. Milne, C. Deck, R. P. Carreira, Q. Allinne, and R. Willinger, "Development and validation of a bicycle helmet: assessment of head injury risk under standard impact conditions," *Comput. Methods Biomech. Biomed. Engin.*, vol. 15, no. sup1, pp. 309–310, 2012.

[24] G. Milne *et al.*, “Assessment of Bicyclist Head Injury Risk under Tangential Impact Conditions,” in *IRCOBI Conference Proceedings*, Gothenburg (Sweden), 2013.
[25] K. Hansen *et al.*, “Angular Impact Mitigation system for bicycle helmets to reduce head acceleration and risk of traumatic brain injury,” *Accid. Anal. Prev.*, vol. 59, pp. 109–117, Oct. 2013.
[26] C. Deck, N. Bourdet, A. Calleguo, P.R. Carreira, and R. Willinger, “Proposal of an improved bicycle helmet standard,” in *International Crashworthiness Conference Proceedings*, Milan, Italy, 2012.
[27] R. Willinger, C. Deck, P. Halldin, and D. Otte, “Towards advanced bicycle helmet test methods,” in *International Cycling Safety Conference*, 2014, pp. 18–19.
[28] P. Halldin, C. Deck, and R. Willinger, “Proposal of a new bicycle helmet test method.pdf,” in *International Cycling Safety Conference*, Hannover, Germany, 2015.
[29] H. Stigson, “Bicycle helmet test 2015 by Folksam.” Folksam, 2015.
[30] C. Deck and R. Willinger, “Improved head injury criteria based on head FE model,” *Int. J. Crashworthiness*, vol. 13, no. 6, pp. 667–678, 2008.
[31] C. Giordano and S. Kleiven, “Evaluation of axonal strain as a predictor for mild traumatic brain injuries using finite element modeling,” *Stapp Car Crash J.*, vol. 58, p. 29, 2014.
[32] M. Ghajari, S. Peldschus, U. Galvanetto, and L. Iannucci, “Evaluation of the effective mass of the body for helmet impacts,” *Int. J. Crashworthiness*, vol. 16, no. 6, pp. 621–631, 2011.
[33] C. Klug, F. Feist, and E. Tomasch, “Testing of bicycle helmets for preadolescents,” *Int Res Counc. Biomech InjIRCOBI*, pp. 136–155, 2015.
[34] F. Feist and C. Klug, “A Numerical Study on the Influence of the Upper Body and Neck on Head Kinematics in Tangential Bicycle Helmet Impact,” in *IRCOBI Conference Proceedings*, 2016.
[35] M. Fahlstedt, P. Halldin, V. S. Alvarez, and S. Kleiven, “Influence of the Body and Neck on Head Kinematics and Brain Injury Risk in Bicycle Accident Situations,” in *IRCOBI Conference Proceedings*, 2016.
[36] E. B. Becker, D. V. Anishchenko, and S. B. Palmer, “Motorcycle Helmet Impact Response at Various Levels of Severity for Different Standard Certifications,” in *IRCOBI Conference Proceedings*, 2015.