

Comparative Evaluation of DOT vs. ECE Motorcycle Helmet Test Method

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Abstract Among existing motorcycle helmet test methods the FMVSS 218 also called DOT and the ECE R2205 are the most communally concerned. It is interesting to observe that the pass/fail criteria for these two key standards are different for the same human head. In this context the present paper suggests comparison and evaluation of these two standards in terms of human head protection level expressed in terms of computed axonal strain induced by pulses that fulfill DOT on the one hand and ECE on the other. A total of 24 pulses have been generated and simulated with a head FE model that permits assessment of the axon strain at the time of impact. The computation of axon strains and brain injury risk for the two different pulse groups showed that the axonal strains are significantly higher for the *DOT-pulses* than for the *ECE-pulses*. It is the first time that two different motorcycle helmet standards are compared and evaluated in terms of brain injury risk based on the extreme pulses that fulfil respectively one standard and the other.

Keywords axonal strain, head injury, ECE R22.05, FMVSS 2018, standard test

I. INTRODUCTION

The criteria used routinely in head protection standards are the resultant peak acceleration of the centre of gravity (CG) of the head and the Head Injury Criterion (HIC). The HIC has been introduced in 1972 by the National Highway Traffic Safety Administration (NHTSA) [1]. Its computation has been deduced by curve fitting [2] and [3] from the Wayne State University Curve [4] which shows a correlation between, on the one hand the head acceleration and impact duration and, on the other hand the occurrence of head injury. Injury risk curves for HIC have then been proposed in a number of studies such as by [5] for skull fracture and AIS $\geq$ 4 brain injury and introduced into regulations. Currently the threshold value for HIC routinely used in standards are 1,000 for car occupants and 2,400 for motorcyclist helmets with an impact velocity of 7.5 m/s as long as ECE22.05 standard is concerned.

There are several motorcycle helmet standards around the world, and this paper focusses on European and US ones, i.e. ,respectively ECE R22.05 [6] and DOT FMVSS 2018 [7]. These standards have separate pass-fail criteria for the same objective which is to assure that a given helmet on the market presents adequate safety requirements. Globally speaking, both test methods consists of the dropping of a helmeted headform against an anvil. The headforms are instrumented with accelerometer sensors allowing to record the headform acceleration pulse at the time of impact. There are differences between the two standards in terms of headform boundary conditions at neck level, impact velocity, and pass/fail criteria, and the present study will focus on the differences in terms of head injury criteria. As both standards consider the protection level of a human head, at biomechanical level it would be more appropriate to have similar pass/fail criteria, even if impact conditions may be different. Typically, both pass/fail criteria are based on the linear headform acceleration and HIC established in the 1960s.

In last decades a number of alternative head injury criteria have been reported in the literature in order to take into account not only the linear head acceleration but also the rotational acceleration (BRIC [8], RVCI [9], RIC [10]) and more interestingly combining both linear and angular metric (GAMBIT [11], HIP [12], PRHIC [10], PCS [13] and CP [14]). These metrics, based on global maximum head kinematics however do not take into account the time evolution of the loading curve or the direction of the impact. More over most of them do not propose relevant injury thresholds. Finally it should be mentioned here that current helmet test methods are not (yet) considering any head rotational motion. In addition to Global head kinematic parameters, tissue level brain injury criteria based on head FE models have been proposed as well. [15] introduced SIMon model with CSDM as the

injury metric, [16] suggested strain time strain rate based on the WSU brain model, [17] published the KTH model with maximum principal strain as a threshold and [18] suggested brain shearing stress based on SUFEHM. More recently main axon bundle were implemented in advanced brain models and brain injury criteria were extracted from head trauma simulations in terms of axonal strain by [19] and [20].

In the literature the biofidelity of criteria used in motorcycle helmet standards has been assessed by [21] who implemented DOT and ECE pulses into SIMon (Simulated Injury Monitor) human head finite element model (FEM) [15]. In addition, this study considered stress sensors between the headform and the helmet. This study showed that peak acceleration and SFC (Skull Fracture Criteria) are biofidelic criteria for skull fracture. It was also shown that a 50% probability of fracture occurs when the peak acceleration is 290 g; a value which is similar to that used in the current ECE standard (275 G, but lower than that in the FMVSS 218. (400 G)). Considering a similar methodology based on FE simulation of headform acceleration pulses, [22], concluded that the DOT protocol is not expected to be adequate for predicting acceleration-induced brain injury. These authors also concluded that allowing the headform to rotate during impact should provide a more biofidelic test method and that a single value of peak acceleration is insufficient in characterising a large number of injury modes

In order to further evaluate the biofidelity of the injury criteria considered by the DOT and ECE standards, a detailed head FEM with relevant brain injury criteria based on axon strain [20] has been used in the present study in order to compute the brain response in terms of axon strains under *DOT pulses* and *ECE pulses* and to assess the brain injury risk for both types of head acceleration pulses.

II. METHODS

In this section a brief description of the head injury criteria considered in DOT and in ECE helmets is proposed as well as the head acceleration pulses considered for the assessment of the head response under front, lateral vertex and occipital impact are presented. Finally, a short presentation of the Strasbourg University FE Head model (SUFEHM) and the related brain injury criteria.

DOT FMVSS 2018 Motorcycle Helmet Standard

Federal Motor Vehicle Safety Standard 218 (FMVSS) is the US motorcycle helmet Approval Standard. For the absorption test, the helmet is fitted to a metallic headform instrumented with a single linear accelerometer. This headform is guided vertically on a monorail with a rigid assembly. The test consists of measuring the headform acceleration at the CG of a helmeted headform at the time of impact against a flat or hemispherical anvil. Each helmet is impacted at four different points with two identical successive impacts at each point. Two of these sites are impacted on the flat anvil and the two others against the hemispherical anvil. The tests performed on flat anvil are at an initial speed of 6 m/s when those against the hemispherical anvil are at 5.2 m/s. For this standard, the pass/fail criteria for approval are as follows:

- The acceleration peak should not exceed 400 G;
- the dwell time for an acceleration of 200 G must not exceed 2 ms;
- the dwell time for an acceleration of 150 G must not exceed 4 ms.

ECE R22.05 Motorcycle Helmet Standard

Directive ECE 22 05 is the current standard in all countries of the European Union. The absorption capacity of the helmet during an impact is assessed by recording the headform acceleration when the free fall helmeted headform is impacting at a specific speed against a rigid anvil. The impact velocities recommended by this Directive are 7.5 m/s and 5.5 m/s respectively for points B (frontal), P (vertex), R (occipital), X (lateral) and for the point S (chin). The headform used during the tests is instrumented with a triaxial linear accelerometer. Two kinds of anvil are used, a flat horizontal one and a kerbstone shaped anvil. The absorption efficiency is considered to be sufficient when the resultant acceleration measured at the CG of the headform at no time exceeds 275 G, and when the Head Injury Criterion ((1) does not exceed 2,400 for the five impact points.

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

The difference between the pass/fail criteria considered by the two standards lead to critical pulses respectively acceptable for ECE R 22.05 and DOT standard as illustrated in 0. To generate critical acceleration pulses a number of acceleration pulses (both DOT and ECE) have been collected from different experiments for general inspection. Based on these qualitative observations equations of critical pulses were generated according to DOT and ECE requirements, respectively.

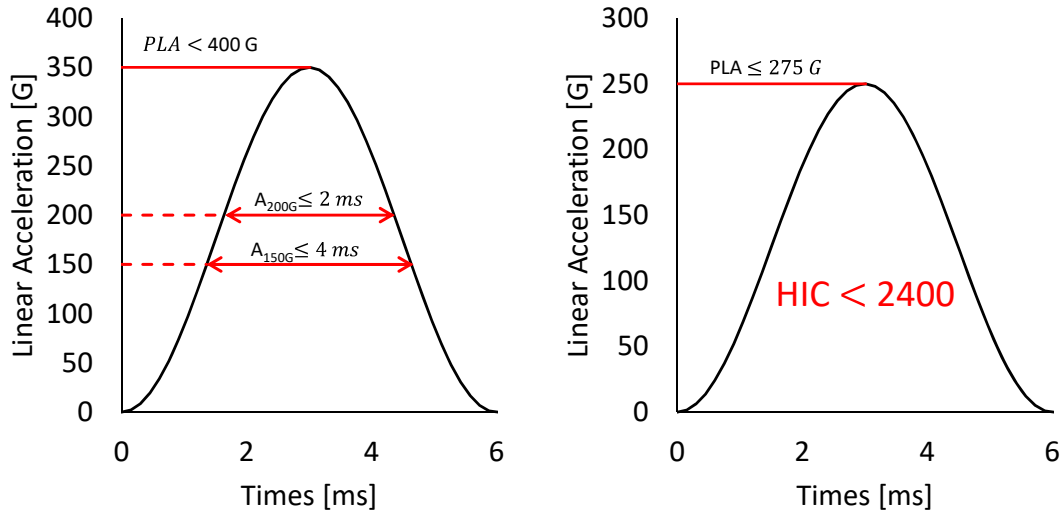


Fig. 1. Illustration of pass-fail criteria for DOT in terms of PLA and dwell time at 200 G and 150 G (left) and ECE in terms of PLA and HIC value (right). HIC is maximum for t_1 and t_2 .

$$Acc(t) = \frac{PLA}{2} \left(\sin \left(\frac{2\pi}{T} t - \frac{\pi}{2} \right) - 1 \right) \quad (2)$$

To do so, a theoretical peak-shaped acceleration is applied to generate the acceleration curves. A shifted sinusoidal function as illustrated in 0 is proposed as this shape is very close to the most common involved acceleration shapes. The sinusoid-based acceleration pulses are defined in (2). And it appears that the acceleration pulses can be reduced to just two parameters, i.e., the Peak Linear Acceleration (PLA) and its duration T . To generate the curves, T is ranging from 4 to 20 ms step of 1 ms and the range for PLA is 150 to 400 g step 10 G. Thus about 400 curves were generated, but only those where the DOT criteria only were acceptable and the ECE criteria only were acceptable and with velocity changes within the range of 6.0 m/s to 15 m/s were retained as acceptable pulses. It can be observed in Figure 2 that these changes in velocities can be higher than the initial velocities of the headform which is 6 m/s and 7.5 m/s respectively for the DOT and the ECE test but not lower. This observation is linked to a possible rebound phenomenon, which leads to a higher velocity change than the initial velocity. At the other hand rebound velocity cannot be higher as the initial velocity, so the velocity change cannot be over twice the initial velocity. As a conclusion, for the critical pulses under consideration the velocity changes range from 6.3 to 8.3 m/s for the *DOT-pulses* and from 8.9 to 14.1 m/s for the *ECE-pulses*. In Figure 2, 24 relevant pulses are represented in black corresponding to the DOT-requirements only and in grey corresponding to the ECE-requirement only. Table 1 reports the peak linear acceleration, HIC values, linear acceleration at 2 ms and 4 ms as well as change velocity including (incident and rebound velocities) for each of these curves. Curves that meet both the DOT and ECE criteria are not considered in the present study.

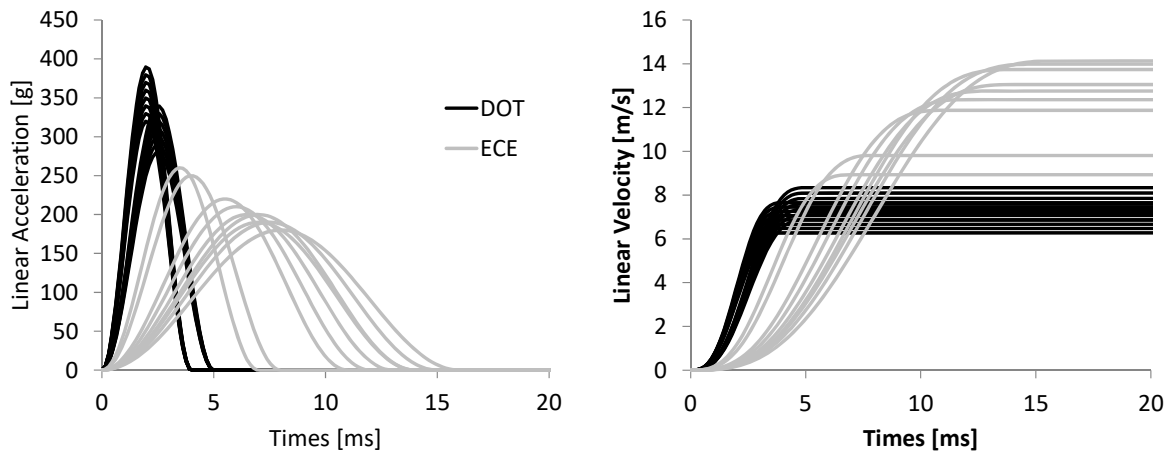


Fig. 2. Sinusoidal based acceleration pulses generated and fulfilling respectively the DOT requirements only and ECE requirements only in terms of acceleration (left) and change velocity (right).

A total of 24 curves has been selected according to the criteria defined previously: 15 DOT curves which pass DOT criteria only, and nine curves which pass ECE criteria only. Each generated head acceleration curve has been considered as the input of the head FEM for the simulation of a frontal, vertex, occipital and lateral impact as shown in Figure 3.

TABLE I

Description of DOT and ECE curves in terms of Peak Linear Acceleration, HIC values and t_1 and t_2 , Linear Acceleration at 2 ms and 4 ms, change velocity including (incident and rebound velocities) as well as pulse duration T.

Curve ID	PLAC [g]	HIC	t_1 [ms]	t_2 [ms]	A_{2ms} [g]	A_{4ms} [g]	ΔV [m/s]	T [ms]	Curve ID	PLA [g]	HIC	t_1 [ms]	t_2 [ms]	A_{2ms} [g]	A_{4ms} [g]	ΔV [m/s]	T [ms]
ID00_DOT	390	3614	1.1	3.0	166	0	7.7	3.9	ID00_ECE	260	2306	1.7	5.3	203	92	8.9	6.9
ID01_DOT	310	2555	1.2	3.8	181	24	7.6	4.8	ID01_ECE	250	2392	2.0	6.0	204	121	9.8	7.8
ID02_DOT	320	2766	1.2	3.8	186	25	7.8	4.8	ID02_ECE	220	2391	2.7	8.3	198	154	11.9	10.9
ID03_DOT	329	2987	1.2	3.8	192	26	8.1	4.8	ID03_ECE	210	2322	3.0	9.0	193	154	12.4	11.9
ID04_DOT	339	3219	1.2	3.8	198	27	8.3	4.8	ID04_ECE	200	2227	3.2	9.8	187	153	12.8	12.8
ID05_DOT	320	2204	1.1	3.0	136	0	6.3	3.9	ID05_ECE	190	2110	3.5	10.6	178	153	13.0	13.9
ID06_DOT	330	2380	1.1	3.0	140	0	6.5	3.9	ID06_ECE	200	2398	3.5	10.6	188	161	13.7	13.9
ID07_DOT	340	2564	1.1	3.0	145	0	6.7	3.9	ID07_ECE	190	2260	3.6	11.3	180	156	14.0	14.8
ID08_DOT	350	2757	1.1	3.0	149	0	6.9	3.9	ID08_ECE	180	2106	3.9	12.1	172	151	14.1	15.8
ID09_DOT	360	2958	1.1	3.0	153	0	7.1	3.9									
ID10_DOT	370	3168	1.1	3.0	157	0	7.3	3.9									
ID11_DOT	380	3386	1.1	3.0	162	0	7.5	3.9									
ID12_DOT	280	1981	1.2	3.8	163	22	6.9	4.8									
ID13_DOT	290	2163	1.2	3.8	169	23	7.1	4.8									
ID14_DOT	300	2354	1.2	3.8	175	24	7.4	4.8									

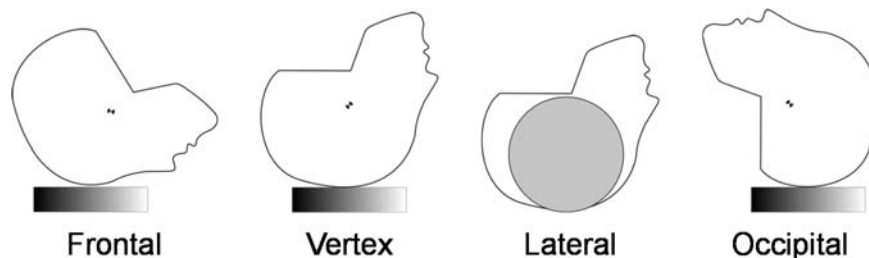


Fig. 3. Different orientations for implementing the acceleration curves into the head FEM.

Simulation of the Head Impacts

A state-of-the art finite element head model (FEHM) with enhanced brain and skull material laws [23], [24] was used for numerical computation of the 96 impacts. The head model was equivalent to a 50th percentile adult human head. The main anatomical features of this model include the scalp, skull, membranes, brain, brainstem and cerebrospinal fluid (CSF). The head model presents a continuous mesh that was made up of 13,208 elements,

including 1,797 shell elements for the skull and 5,320 brick elements for the brain. The total mass of the head model was 4.7 kg.

The skull was modelled with a three-layered composite shell representing the inner table, diploe and outer table of human cranial bone, including the modelling of fracture. More information about the implemented constitutive laws for the skull as well as the skull fracture criterion are available in [23][25] [23], [25].

The brain model was enhanced by implementing fractional anisotropy and fibre orientation extracted from medical imaging and based on diffuse tensor imaging (DTI) coming from an atlas build up with 12 healthy subjects and implemented into the brain FEM in order to mimic the main axon bundles [24] [26], as illustrated in Figure 4.

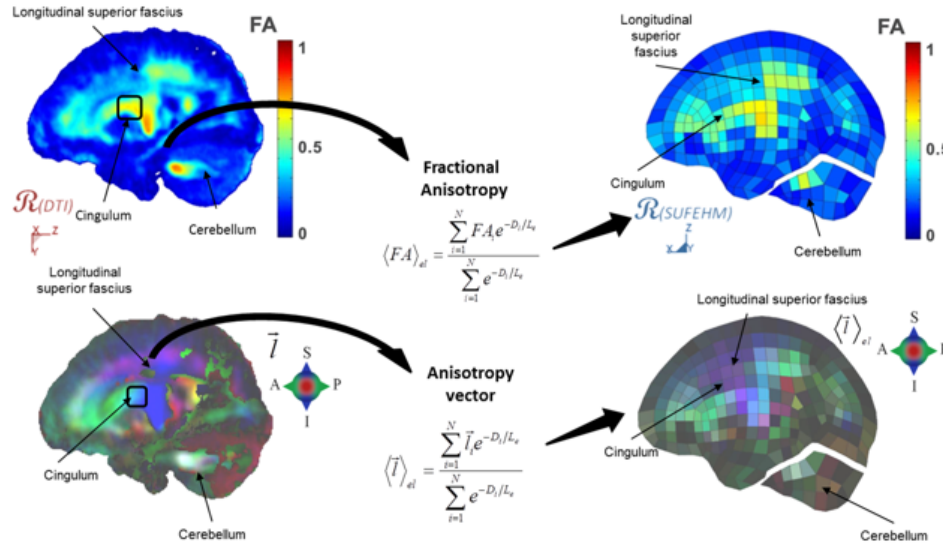


Fig. 4. Illustration of FA and anisotropy vector coming from an atlas of 12 healthy patients used to implement anisotropy in FEHM. Details on the methodology used is given in [24]

The mechanical behaviour of the brain model has been extensively validated for brain pressure against intracranial pressure data from [27] and brain strain against local brain motion data from [28], [29].

Further, 109 real-world head trauma cases have been reconstructed with this head model to derive brain injury criteria. To do so, a number of intra cerebral parameters were computed such as brain axonal strain rate, axonal strain, first principal strain, Von Mises strain, first principal stress, Von Mises stress, CSDM (0.10), CSDM (0.15) and CSDM (0.25). For all parameters maximum value occurring in one element were taken into account. Based on an in-depth statistical analysis using logistic regression, it appeared that axonal strain presented the highest Nagelkerke R^2 , which demonstrated that this parameter was the most adequate metric to predict DAI. The proposed brain injury tolerance limit for a 50% risk of DAI has been established at 14.65% of axonal strain [20]. An illustration of the FEHM used for the present study as well as its associated injury risk curves in terms of skull fracture prediction and moderate diffuse brain injury are shown in Figure 5.

For each of the pulses (DOT or ECE) generated in the previous section the maximum axon strain was computed and compared to the critical threshold value.

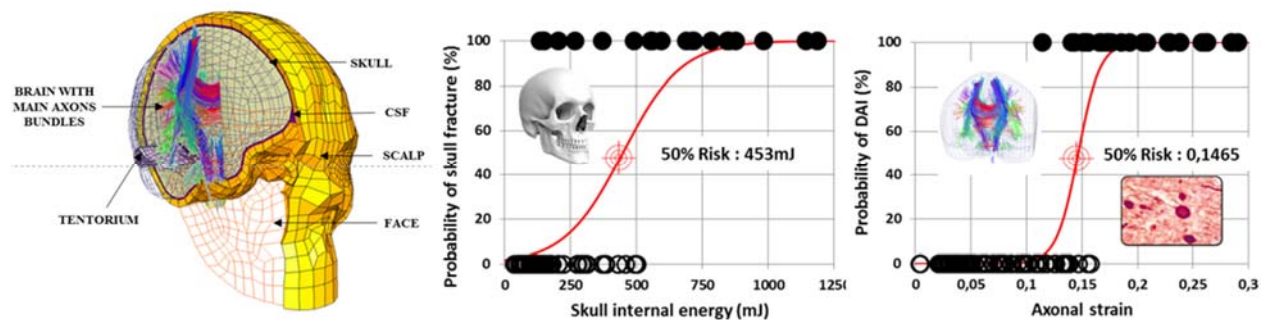


Fig. 5. Illustration of the human head FE model used for the simulation of the head impacts including the white matter tracts and associated injury risk curves to predict probability of skull fracture by addressing skull strain energy and moderate diffuse axonal injuries by addressing axonal strain.

III. RESULTS

A total of 15 extreme pulses that fulfil the DOT requirements in terms of maximum linear acceleration and pulse durations were generated and called *DOT pulses*. On the other hand nine extreme pulses that fulfil the ECE requirements in terms of maximum linear acceleration and HIC have been generated and called the *ECE-pulses*. Finally, the 24 pulses were considered as inputs for the simulation of the brain response under frontal, lateral, occipital and vertex head impact. For each of these 96 impacts, the brain injury risk was assessed based on the computed axon strain and their comparison to the threshold value for a 50% risk of injury. Results are reported in Figures 6 to 8 in terms of maximal axonal strain respectively for Frontal, Vertex, Occipital and Lateral impact configurations.

The computation of axonal strains for the two different pulse groups showed quasi systematically that the axon strains are higher for the *DOT-pulses* than for the *ECE-pulses*. However, some of the ECE-pulses applied under occipital impact lead to more severe brain loading than the DOT.

Coming to the assessment of brain injury risk it appears globally that all pulses lead to risk of moderate brain injury (AIS 2+), as most values are over the red line which illustrates the 50% risk for this injury. However, the *DOT-pulses* lead often to higher risk as the *ECE-pulses*.

A more detailed analysis of results per impact configuration shows that the Frontal and Occipital impacts (0 and Figure 7) lead to extreme brain loading for the DOT-pulses and to brain loading close to the threshold for the ECE-pulses. However, for the *ECE-pulses*, the occipital impact lead to higher risk than the DOT-pulses. The Vertex and the Lateral impacts (Figure 8 and Figure 9) lead to less critical brain loading, as *DOT-pulses* present risks close to 50% and *ECE-pulses* nearly no risk at all.

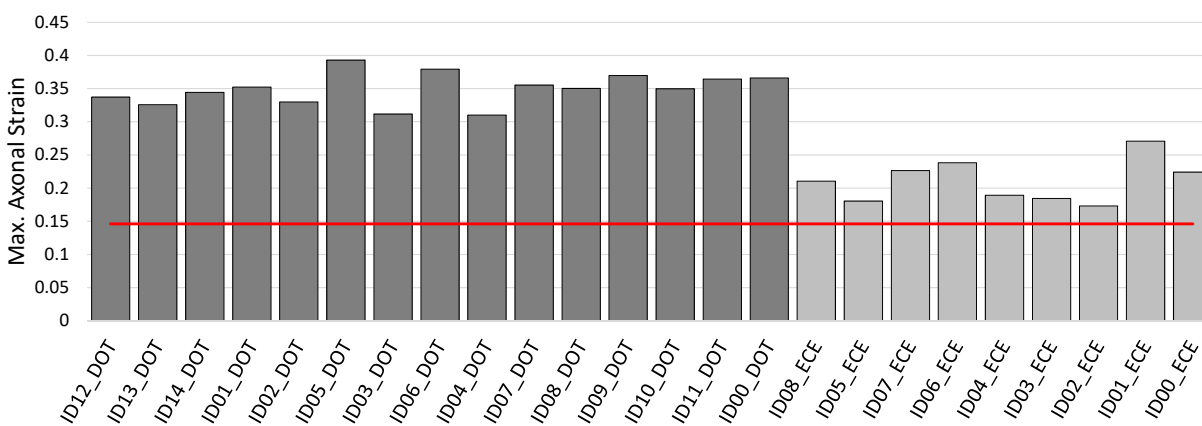


Fig. 6. Maximum axonal strain for different pulses for frontal impact (red line represents 50% of AIS2+).

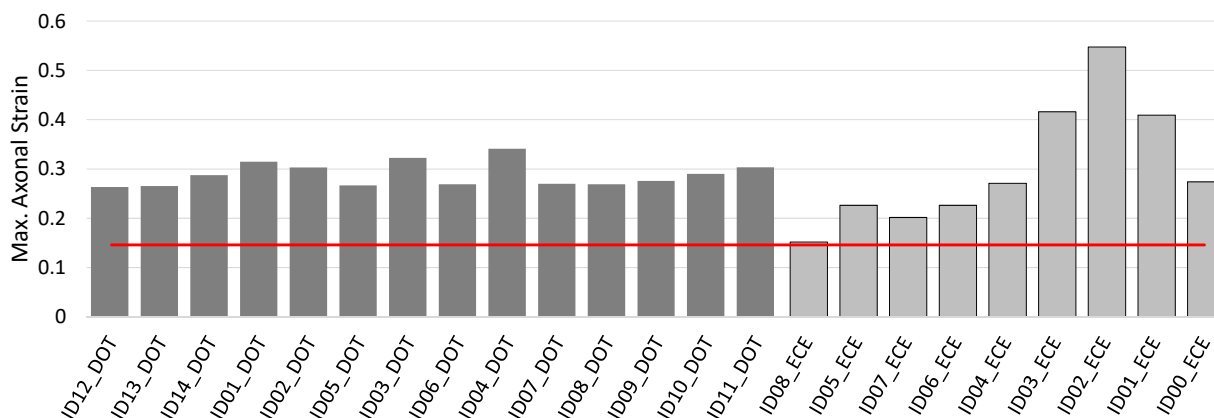


Fig. 7. Maximum axonal strain for different pulses for occipital impact (red line represents 50% of AIS2+).

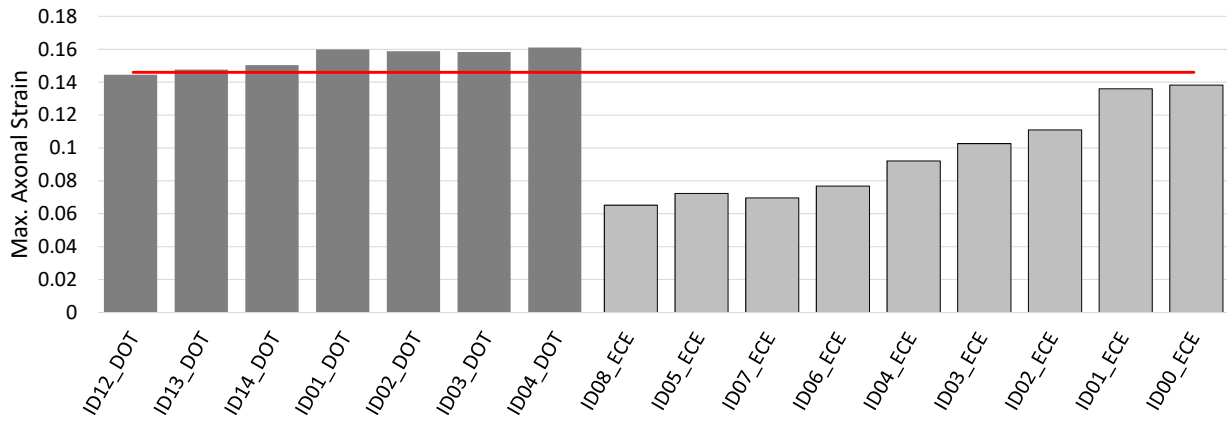


Fig. 8. Maximum axonal strain for different pulses for vertex impact (red line represents 50% of AIS2+).

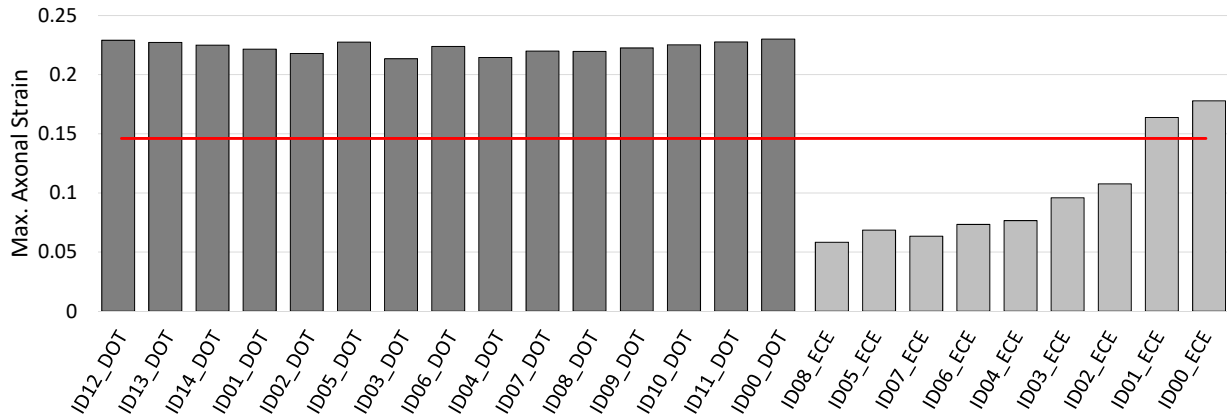


Fig. 9. Maximum axonal strain for different pulses for lateral impact (red line represents 50% of AIS2+).

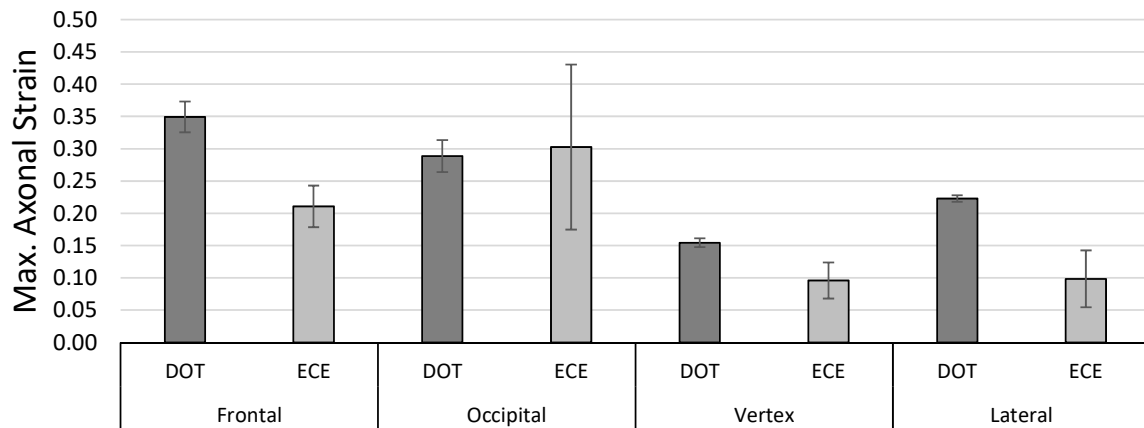


Fig. 10. Mean Axonal strain for each impact configuration and requirement pulse.

Figure 10 gives a synthetic view of results as the mean value of the computed axonal strain for the frontal, vertex, occipital and lateral impact configurations are reported for the DOT and ECE-pulses. It can be observed in this figure that the axonal strains are often higher for the DOT-pulses as for the ECE-pulses.

IV. DISCUSSION

This study focusses on helmet performance under impact. Therefore, penetration, oblique and climate conditioning aspects were not considered. Further, the present study considers the effect of a headform acceleration pulse on the brain response only which supposes that the skull does not sustain any deformation. The assessment of skull fracture risk is therefore not in the scope of this study. Another limitation of this study is that only single-peak shaped acceleration pulses were considered, when it is well known that more complex acceleration shapes can be recorded in real-world helmet tests. Coming to the head FE model validation, it should

be mentioned that the validation was performed against pressure according to Nahum's impact that were in a similar impact duration range. However, the validation in terms of brain deformation according to Hardy's impact was not performed for such severe and short impacts as Hardy's impacts were much longer, and this presents another limitation of our study. Moreover the reported maximum axon strains and injury risk are based on a single element bases in line with the definition of the injury criteria mentioned in the methodology and this may affect results under severe and short impacts.

Despite these limitations it was possible in this study to assess the brain injury risk for extreme pulses accepted respectively within DOT and ECE standards. Similarly, as [22] it is confirmed that maximum linear acceleration resultant is not able to predict acceleration induced brain injuries. Moreover, the comparative assessment of brain injury risk under extreme *DOT-pulses* and *ECE-pulses* globally showed a higher higher brain injury risk for DOT than for ECE. However this observation is not applicable to the occipital impact for which the *ECE-pulses* lead to higher risk. Results also show that the discrepancy in terms of axonal strain is much higher for the *ECE-pulses*. This observation is done to the fact that the range of pulses in terms of amplitude and duration is much wider for these pulses than for the *DOT-pulses* (see Figure 2).

The reason why *DOT-pulses* often lead to higher brain loading may be linked to the fact that DOT standard implies more severe impact conditions as higher headform mass and impact velocity are applied. In addition the constrain at neck level avoids any rotation, so that all energy is translational energy transmitted to the helmet. It is the authors opinion that this increased severe impact condition should not be the reason for higher pass/fail criteria as pass/fail criteria should be driven by human head impact tolerance solely. In other terms it does not make any sense to prescript severe impacts that the human head may not sustain with some of the helmets.

In his earlier study [30] conducted ECE R22 tests with an instrumentation allowing for assessment of translational and rotational movement of the headform and implemented this 6 D acceleration curve into the SIMon head FEM. Simulations were conducted with and without rotational movement indicating that to evaluate brain injury risk measures of the rotational acceleration should be recommended in the ECE R22 protocol. This analysis demonstrates that the ECE standard has a serious issue or limitations as long as the headform rotational acceleration is not recorded and processed properly. This aspect of helmet testing methods was not the purpose of the present study but it will clearly be a next step to conduct helmet tests under ECE impact conditions, to record the 6D headform response and to assess the tissue level brain injury risk by introducing the full combined linear and rotational headform loading into the numerical head injury risk assessment tool used in the present study as initiated for oblique helmet test methods by [31].

Finally the need of biofidelic pass/fail criteria in a standard could be discussed, as standard are not necessarily designed to reproduce real world accident and biofidelic pass/fail criteria. Testing standards are meant to provide a simple engineering test to evaluate the performance of a piece of equipment. Therefore head protection systems testing standards were not developed to predict intracranial biomechanics response. If test standard should remain as simple as possible, evaluation of head protective systems integrates impact biomechanics research since the 1960s with the WSU head tolerance curve from which the maximum head acceleration of 275 G comes from and later with HIC. Within the automotive environment the ISO organization introduced working groups dedicated to "Performance criteria expressed in biomechanical terms" as well as "Accident analysis" working group in order to establish accident situations and initial impact conditions. As a whole a standard is a balance between simplicity and reality that considers impact conditions, human surrogates and injury criteria and finally the protection system itself. With evolution of technology and scientific knowledge it is likely that standards will also evaluate.

V. CONCLUSIONS

It is well known that different helmet standards exist around the world with a same final objective that is to protect the human head. Different impact energies or anvil shapes and inclination may exist but pass/fail criteria should in any case consider human head tolerance limits, which clearly is not the case. In this context the present paper suggests comparison and evaluation of two well-known motorcycle helmet standards, namely FMTSS-DOT and ECE-R2205, under force respectively in the US and in EU. The evaluation was conducted in terms of head protection level or brain injury risk induced by extreme pulses that fulfil DOT on the one hand and ECE on the other. To do so critical as well as critical *ECE-pulses* were generated and applied to an existing head FEM in order to assess the brain injury risk for each of these pulses applied frontally, laterally and to the occipital plus vertex

area. Brain response was computed in terms of maximum axon strain and compared to injury risk curves derived for this metric in previous studies.

From this investigation it appears that all pulses lead to significant risk of moderate brain injury (AIS 2+). However, the *DOT-pulses* lead typically to higher axon strain, and to higher brain injury risk than the *ECE-pulses*. It was also pointed out in the discussion that even if not addressed in the present study, the ECE standard has serious limitations as long as the headform's rotational acceleration is not recorded and processed.

VI. ACKNOWLEDGEMENT

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