INFLUENCE OF THE BODY ON KINEMATIC AND TISSUE LEVEL HEAD INJURY PREDICTORS IN MOTORCYCLISTS ACCIDENTS

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ABSTRACT

A number of Finite Element Analyses (FEA) were performed on a helmeted Hybrid III dummy and the helmeted detached head of the dummy both making impacts onto a flat anvil, in order to investigate the effect of the body on head injury predictors founded on the kinematics of the head as well as the predictors defined at tissue level. For the latter, an FE model of the human head was employed. It was shown that for the impact velocity at which the helmet had been drop tested, and certified as per ECE 22.05, the helmet was unable to protect the head when the entire body was used in the virtual impact test.

Keywords: Helmets, Energy Absorption, Drop tests, Hybrid III, Finite Element Method

MOTORCYCLE ACCIDENTS have received much attention in recent years due to the fact that motorcyclists are among the most vulnerable road users. According to statistical investigations, motorcycles comprise only 6.1% and 2.4% of all motorised vehicles in Europe and the US (ACEM, 2006) respectively; nonetheless, motorcyclists account for 16% of total road-user fatalities in Europe and 9% of total traffic fatalities in the US (COST327, 2001, NCSA, 2004). The most important piece of personal protective equipment that is intended to protect motorcyclists from severe or fatal injuries is the safety helmet. Results of a statistical investigation on motorcycle accidents in the US from 2000 to 2002 revealed that about 51% of unhelmeted riders suffered head injuries as compared to 35% of helmeted riders (NCSA, 2007).

Safety helmets should pass standard tests prior to being sold in market. Almost all standards follow the same concepts for evaluating the protective capability of a helmet, even though the details of their procedures are different (Ghajari, et al., 2008). Among these procedures, assessing the energy absorption capacity of a helmet is still the most challenging, and requires more fundamental studies. The impact absorption capacity of a helmet is determined by recording against time the acceleration imparted to a headform fitted with the helmet when it is dropped in a guided fall at a specific impact velocity upon a fixed steel anvil; the absorption efficiency is considered sufficient when the peak of this acceleration and/or a function of the acceleration, which are conventional head injury predictors, exceed(s) the relevant thresholds. There have been several criticisms and studies on various features of this standard method (COST327, 2001, Gilchrist and Mills, 1994, HIC-Workshop, 2005, SHARP, 2008, Thom, et al., 1998); however, the effect of using a detached headform, and therefore ignoring the rest of the body, on head injury indicators has received little attention.

One way to study this effect is to use anthropomorphic test dummies in drop tests. For example, Aldman et al. (Aldman, et al., 1976, Aldman, et al., 1978a, Aldman, et al., 1978b) dropped a helmeted Ogle-Opat dummy onto a surface made of asphalt concrete at 4.4 m/s and 5.2 m/s impact speeds and measured the linear and rotational accelerations of the head. In a similar research study (COST327, 2001), a Hybrid III dummy and its detached head were fitted with helmets and dropped onto flat anvils at 4.4, 5.2 and 6 m/s, and the linear and rotational accelerations of the head were recorded. These impact velocities were chosen "to simulate realistic impact conditions and to limit the risk of severe damage to the dummy". It has been concluded that "the effect of the body and the neck is thus a decrease of the measured linear acceleration values when compared with headform measurements". In spite of using helmets certified according to European standards (ECE22.05, 2002), the dummy and headform drop tests were compared at lower impact speeds than that set in ECE22.05, i.e. 7.5 m/s; therefore, conclusions were confined to a maximum impact speed of 6 m/s.

To measure head injury probability, helmet standards employ injury predictors that are based on the kinematics of the head called kinematic injury predictors. For instance, ECE 22.05 and Snell (Snell, 2005) use peak linear acceleration of the head. The former assigns a limit of 275g to the acceleration and the latter assigns a limit of 300g to it. According to COST (COST327, 2001), the peak linear acceleration of 260g corresponds to serious head injuries (AIS3), which is comparable to the limit of 280g reported in literature for the same injury severity. Since the development of finite element models of the human head that contain different tissues of the head, such as the brain, novel head injury mechanisms for tissue level injuries have emerged. For example, the first principal strain and Von Mises stress in the brain were found to be the best predictors of concussion and diffuse axonal injury (Willinger and Baumgartner, 2003, Zhang, et al., 2004). King and co-workers believe that focus should be changed from kinematic to tissue level injury predictors (King, et al., 2003). They argued that safety standards were developed when only linear acceleration could be measured.

In the present study, for the first time the Finite Element Method (FEM) was used to simulate drop tests of a Hybrid III dummy fitted with a commercial helmet. It is shown that when the whole body is involved in drop tests, increasing the impact speed from 6 m/s to 7.5 m/s drastically increases the head injury probability predicted by both kinematic and tissue level indicators. This suggests that body inertia is an important parameter and, perhaps, should be considered in evaluating the protective capability of safety helmets. On the basis of closed-form solutions obtained from a one-dimensional formulation of the standard drop test, it is demonstrated how helmet standards can be modified, in a simple and economical way, to take into account this effect.

FE MODELS

For this study, FE models of a full-face commercial helmet, human head, Hybrid III dummy and its head were prepared in the Ls-Dyna crash code (Hallquist, 2007) format and coupled together as appropriate to simulate helmeted headform and helmeted dummy drop testing at different impact speeds. Kinematic injury predictors were extracted directly from these simulations, while the Strasbourg University FE head model (SUFEHM) was employed to calculate tissue level predictors.

The two most important parts of the helmet in terms of energy absorption were considered in the FE modell: the liner and the shell; the pertinent geometries, which were provided by Dainese s.p.a, were prepared and meshed using Hypermesh. At the end, the liner had 41384 single integration 4-node tetrahedral solid elements, and the shell had 19090 single integration 4-node quadrilateral shell elements. The selected average sizes of the elements were those suggested by Cernicchi et al. (Cernicchi, et al., 2008), who performed a convergence study on the element sizes of the foam liner and composite shell.

The EPS (Expanded Poly-Styrene) liner was modelled using the *Crushable Foam* material model of Ls-Dyna. The shell was made of composite laminates with different laminas and lamination in the front-top-rear region, sides and chin guard (Fig. 1). It was modelled using the *Laminated Composite Fabric* material model of Ls-Dyna. For each lamina, one through thickness integration point was defined and relevant fibre orientation and material properties were assigned to it. The material properties were obtained from conventional mechanical tests performed by Alessandro Cernicchi during his PhD at Imperial College London.
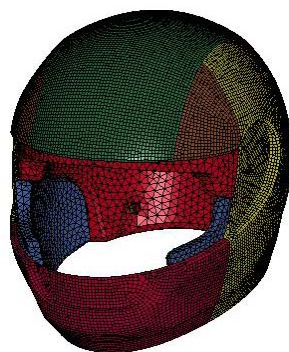

Fig. 1 – FE model of a Commercial Full-Face Helmet

The FE model of the 50th percentile male Hybrid III dummy, released on 30 October 2008, was downloaded from the LSTC website. The dummy was in the sitting posture. According to accident investigations (COST327, 2001, MAIDS, 2004), in motorcycle accidents the most frequent opposite objects are passenger vehicles. In an impact with a car, the motorcyclist usually hits the car shortly after motorcycle/car collision, which means the rider does not have enough time to change the posture considerably. Therefore, using a dummy in the sitting posture represents a number of body positions. The dummy was calibrated by simulating neck extension and flexion and thorax impact tests as per FMVSS 572 (LSTC, 2008). For this study, the dummy scalp was defined as rigid in order to decrease the number of variables that could have affected results. A copy of this model was modified such that only the head remained. Then, both the dummy and its detached head were fitted with the helmet.

Two impacts were simulated:
1. dropping the helmeted Hybrid III dummy onto a flat anvil
2. dropping the helmeted detached head of the dummy onto a flat anvil.

An accident investigation (COST327, 2001) showed that 50% of motorcyclists impacted the opposite object at body impact angles (body impact angle is the angle between the body longitudinal axis and the surface of the opposite object) in the range of 0°-10°; in addition, more than 25% of the helmets were impacted in the frontal side. Therefore, the dummy was positioned so that its body axis was parallel to the anvil surface and the blow occurred in the front site of the helmet (Fig. 2).The helmeted headfrom was dropped in the same configuration. The defined impact velocities were 6 m/s, which is the same as that in the previous experimental study of COST 327, and 7.5 m/s as in the ECE 22.05 regulation.
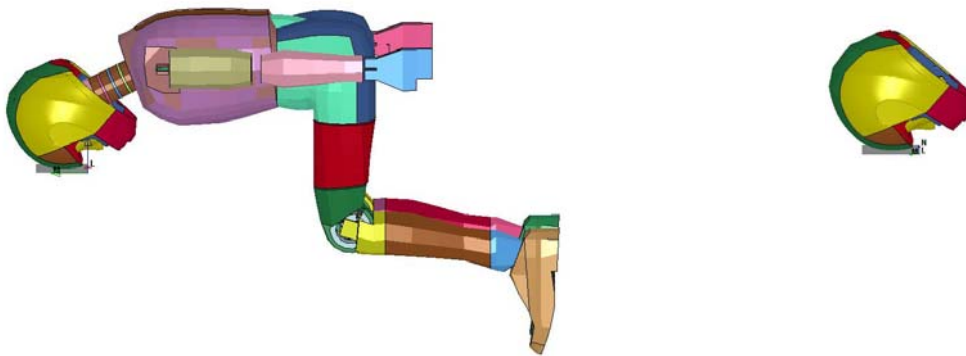

Fig. 2 – Virtual Drop Tests Using Hybrid III Dummy and its Head

In addition to the parameters that are measurable during the above simulations, a comparison between the dummy and headform impact tests was made based on tissue level injury predictors by using the Strasbourg University head injury prediction tool. This tool is a modified version of the Strasbourg University FE head model (SUFEHM) described below; the scalp is removed and the skull is defined as rigid (Deck, et al., 2007). The model was driven by prescribing acceleration components versus time at the centre of gravity of the head, which were recorded during the dummy and headform impact simulations.

The SUFEHM includes almost all of biomechanically important parts of the head, e.g. the scalp, skull, brain, cerebrospinal fluid (CSF), tentorium and falx (Fig. 3). This model was developed by Willinger et al. (Willinger, et al., 1999) using the Radioss code, and validated against cadaveric experiments. Through replicating 61 real world accidents with this model, Willinger and Baumgartner (Willinger and Baumgartner, 2003) found that Von Mises stress in the brain, the strain energy of the CSF and the strain energy of the skull are the best predictors of diffuse axonal injury, subdural haematoma and skull fracture respectively. The head model was converted from the Radioss to Ls-Dyna format to make it consistent with the existing FE models of the helmet and dummy. It was subsequently validated through replicating two cadaveric experiments to guarantee that the conversion did not reduce the capability of the model to reproduce experimental data (Ghajari, et al., 2009).

Before presenting the simulation results, a one-dimensional analytical model of the impact response of helmeted headforms is explained. This model results in simple closed-form equations that

reveal the role of input variables (such as the initial velocity) on output parameters (such as the linear acceleration of the head). The equations demonstrate how the standard drop test can be adapted to consider the effect of body inertia.
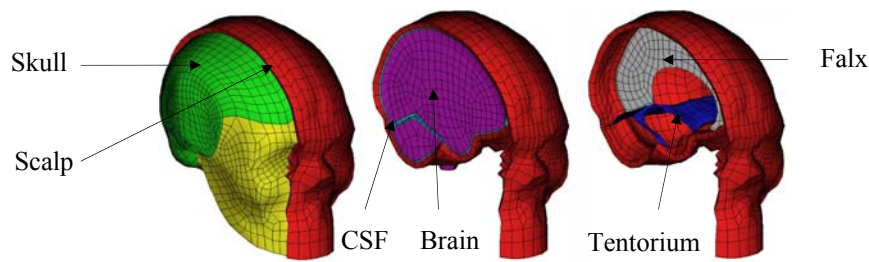

Fig. 3 – Strasbourg University Human Head FE Model (Willinger and Baumgartner, 2003)

ANALYTICAL MODEL

Two parts of a helmet absorb impact energy: the liner and shell. The liner is usually made of expanded polystyrene (EPS) whose typical stress-strain curve is shown in Fig. 4. Gilchrist and Mills (Gilchrist and Mills, 1994) assumed a constant yield stress (S_Y) for the liner under compression and derived the following relation between the normal force on the helmet (F) impacting a flat anvil, and the central deflection of the foam (y):
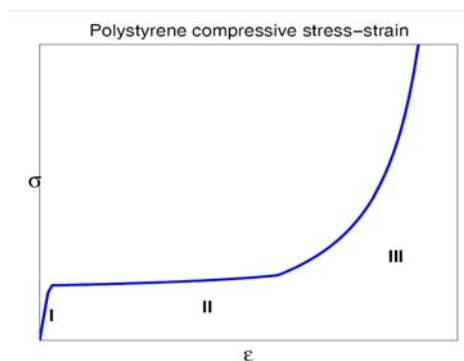
$$F = 2\pi R S_Y y \tag{1}$$

Fig. 4 – Stress-Strain Curve of a Typical EPS (Cernicchi, et al., 2008)

The helmet was assumed to be locally spherical with radius R. This equation was found to give a good approximation of the impact behaviour of thin-shelled helmets such as bicycle helmets. However, the relatively stiff shell of motorcycle helmets increases the contact area for impacts with kerbstone or spherical anvils, and it absorbs part of the impact energy. Fig. 5 shows the internal energy histories of
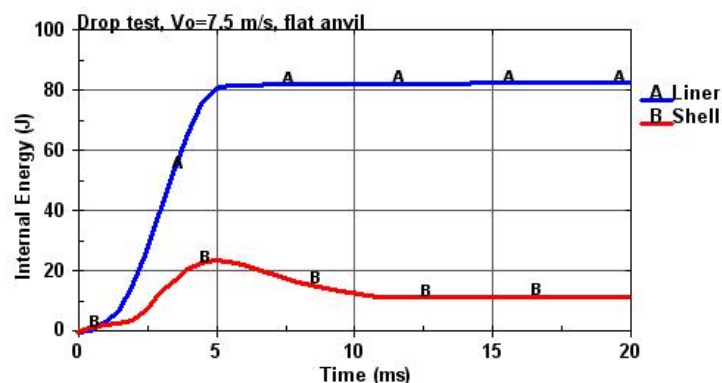

Fig. 5 – Internal Energy of Helmet Parts from FEA

the liner and shell calculated from the FEA of a standard drop test. After 15 ms, the internal energies of the liner and shell remain constant at 83 *J* and 12 *J*, respectively. These constant energies are equal to the absorbed energies. The contribution of the shell to energy dissipation is, therefore, 12%, which is a considerable percentage.

For impacts on flat anvils, we neglect the effect of the shell on increasing the contact area. In addition, we assume that the shell and liner absorb the impact energy sequentially. Therefore, the impact of a helmet with a flat anvil is equivalent to the same impact at reduced velocity when the shell is removed. To calculate the reduced velocity, the energy conservation principle is employed as follows:

$$\frac{1}{2}mV_0^2 = DE_{shell} + DE_{liner} \tag{2}$$

where *m* is the mass of the helmet-headform and *DE* is the dissipated energy. By using the ratio of the total dissipated energy to that of the liner (α) in this equation, we have:

$$\frac{1}{2}mV_0^2 = \alpha DE_{liner} \tag{3}$$

or

$$DE_{liner} = \frac{1}{2}m(\frac{V_0}{\sqrt{\alpha}})^2 \tag{4}$$

Thus, the reduced velocity is:

$$V_{0,r} = \frac{V_0}{\sqrt{\alpha}} \tag{5}$$

To calculate the acceleration of the centre of gravity of the headform, we assume that the helmet and headform are one rigid body whose centre of gravity is located at the centre of gravity of the headform. By using the Newton's second law and substituting for force from Eq. 1, we have:

$$m\ddot{y} = -2\pi RS_Y y \tag{6}$$

The earth's gravity is not considered in this equation because it is negligible compared to the accelerations expected in helmet drop tests. Assuming $y(0) = 0$, the solution to the differential equation (6) is:

$$y(t) = \frac{V_{0,r}}{\omega}\sin \omega t, \quad \omega = \sqrt{\frac{2\pi RS_Y}{m}} \tag{7}$$

The derivation of the peak linear acceleration of the headform (*PLA*), the maximum normal force on the anvil (*NFA*) and the maximum compression of the liner (*Td*) is straightforward from equations (1), (6) and (7):

$$PLA = \frac{V_{0,r}}{\sqrt{m}}\sqrt{2\pi RS_Y} \tag{8}$$

$$NFA = \sqrt{m}V_{0,r}\sqrt{2\pi RS_Y} \tag{9}$$

$$Td = \frac{\sqrt{m}V_{0,r}}{\sqrt{2\pi RS_Y}} \tag{10}$$

These closed-form equations are written in a way to illustrate the effect of the impact variables on the outputs.

RESULTS

As mentioned before, to investigate the influence of the body on head injury predictors a helmeted dummy and the helmeted head of the dummy were virtually dropped onto a flat anvil. The FEA results of the dummy and headform impacts at speeds of 6 m/s and 7.5 m/s are shown in Fig. 6 and Fig. 7, respectively. In these figures, the linear acceleration is the resultant of the linear acceleration components at the centre of gravity of the head, and the rotational acceleration is the acceleration component corresponding to the rotation about the ear-to-ear axis; the nose-to-head vertex direction

defines positive rotation. The other components of the rotational acceleration were negligible because the impact occurred in the symmetry plane of the models.

The comparison between the dummy and headform drop tests at an impact speed of 6 m/s, in Fig. 6, shows that when the dummy was used, the peak linear acceleration of the head was lower but the normal force was higher. These phenomena are the consequences of the body inertia affecting the head through the neck, and are consistent with the experimental results reported in COST (COST327, 2001). The time traces of the rotational accelerations of the dummy impacts have one positive pulse (peak) and one negative pulse (valley) before the linear acceleration and external force reduce, as compared to just one positive pulse for the headform impacts. At the peak of linear acceleration and normal force, the rotational acceleration of the dummy head is minimum whereas that of the headform is zero. The second peak in the rotational acceleration history of the dummy head is the consequence of the potential energy accumulated in the neck when the normal force increased. The sudden decrease in this force allowed the energy to be released, which in turn caused the high rotational acceleration of the head. As this behaviour is attributed to the stiff neck of the dummy, the second peak of the rotational acceleration is not considered when comparing the dummy and headform drop tests.
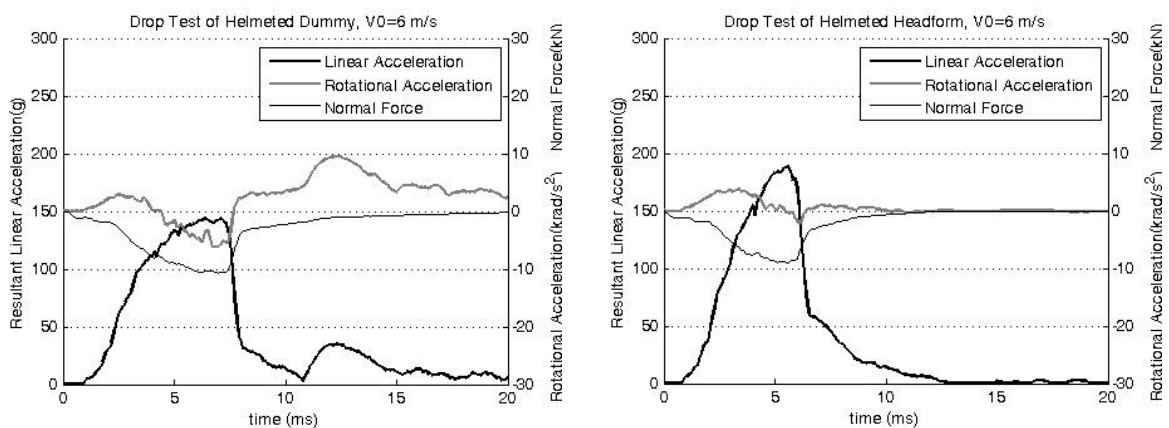

Fig. 6 – Results of Virtual Drop Test at 6 m/s Impact Speed

The effect of including the body on the drop test results at a speed of 6 m/s was a larger normal force on the helmet (Fig. 6), thus further crushing the liner foam. The FE simulation indicated that the maximum compressive strain of the liner in its crushed region was 68% while this quantity was 52% using the headform. Increasing the impact speed from 6 m/s to 7.5 m/s caused more deformation of the liner such that its maximum compressive strains in the crushed region were 87% and 70% for the dummy and headform impacts, respectively. The dummy drop test results at an impact speed of 7.5 m/s, shown in Fig. 7, indicate that the linear acceleration of the head rose suddenly after 6 ms and exceeded that of the detached headform shown in the same figure; this is in contrast to the expected behaviour shown in Fig. 6 and that reported in previous experimental studies (Aldman, et al., 1976, Aldman, et al., 1978a, Aldman, et al., 1978b, COST327, 2001). This phenomenon is the consequence of the bottoming out of the foam liner. Foams are usually designed for a maximum volumetric strain between 70-80% to confine their response to the plateau region of their stress-strain curve (Zone II in Fig. 4), whereas the FEA indicated the maximum volumetric strain in the crushed region of the helmet liner was 87%. It is concluded that the energy absorption of the helmet was not sufficient for the dummy drop test at 7.5 m/s.
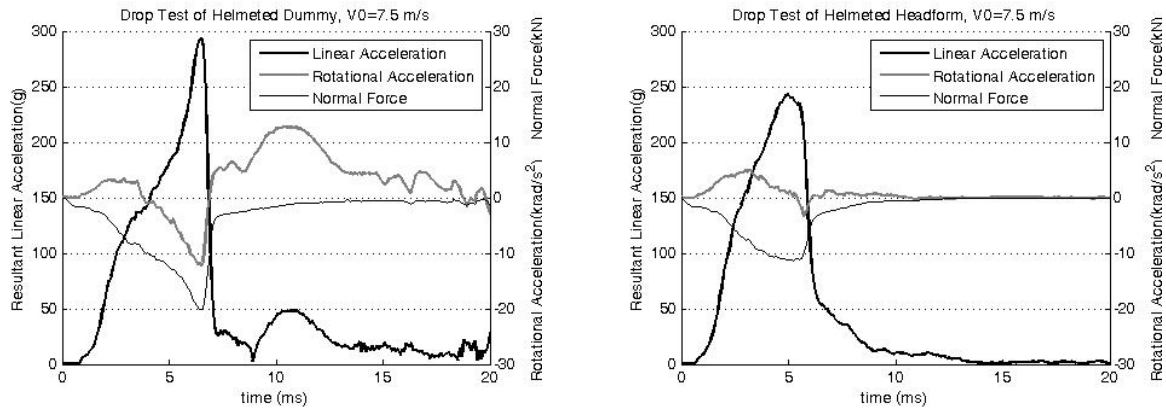
Fig. 7 – Results of Virtual Drop Test at 7.5 m/s Impact Speed

The dummy and headform drop tests are compared in relation to kinematic and tissue level injury predictors, which are presented in Table 1 along with their injury limits. The limits of the *PLA* and *HIC* are those set in the ECE 22.05 standard. The *PLA* threshold is in the range of 260g-280g for serious head injuries (AIS3); however, the limit of *HIC* is very high compared to the threshold of 1500 for AIS3 (COST327, 2001). The rotational acceleration limit was obtained by Margulies and Thibault (Margulies and Thibault, 1992) and the skull fracture limit was measured in impact tests on cadavers by Yoganandan et al. (Yoganandan, et al., 1995). In order to avoid obtaining spurious results because of the poor aspect ratios of the elements and coarse meshes, the maximum Von Mises stress in the brain at each sampling time was the mean of the Von Mises stresses of the 10 elements that had the largest value among all.

The thresholds of the tissue level injury predictors have been found by Marjoux et al. (Marjoux, et al., 2008) using the Radioss format of the SUFEHM. When different FE packages are used to solve an identical FE model, they generate slightly different magnitudes for the same mechanical output; hence, the limits assigned to the *VMB* and *IECSF* are not precise injury thresholds for the Ls-Dyna format of the SUFEHM. Nonetheless, these thresholds were used to predict the probability of the injuries because they were the best available limits.

Table 1. Results of Drop Test Simulations

		Indicator of	Injury Threshold	Helmeted Headform		Helmeted Dummy	
	$V_0 (m/s)$	-	-	6	7.5	6	7.5
Kinematic Injury Predictors	*PLA* (g)	Head Injury	275	188	243	144	293
	HIC	Head Injury	2400	1057	1904	820	1872
	PRA (krad/s2) – $\Delta\omega$ (rad/s)	DAI	10 – 100	3.7 – 10.5	5.0 – 13.1	6.1 – 57.8	12.3 – 71.2
	NFH (kN)	Skull Fracture	11	7.9	10.1	9.2	18.1
Tissue Level Injury Predictors	*VMB* (kPa)	Severe Neurological Injury	39	36	48	35	51
	IECSF (J)	Subdural Haematoma	4.2	8.0	12.4	4.2	6.9

PLA: Peak Linear Acceleration
PRA: Peak Rotational Acceleration
$\Delta\omega$: Maximum Change in Rotational Velocity
NFH: Maximum Normal Force on Head
VMB: Maximum Von Mises Stress in Brain
IECSF: Maximum Internal Energy in CSF

$$HIC = (t_2 - t_1)[(\int_{t_1}^{t_2} a(t)dt)/(t_2 - t_1)]^{2.5}$$

$a(t)$ is the resultant linear acceleration, in g unit and t_1 and t_2 are, respectively, any starting and ending time in impact pulse duration

In spite of fulfilling the requirements of the ECE 22.05 standard with regards to the *PLA* and *HIC*, when the dummy was used at the same impact conditions, the helmet failed to protect the head. The *PLA* exceeded its threshold and skull fracture was expected. Although the *PLA* was higher for the dummy impact at 7.5 m/s, *HIC* was lower, which was probably the consequence of a smoother distribution of the detached headform linear acceleration (Fig. 7).

The *VMB*s of the dummy and headform impacts are comparable for both impact speeds, but their time traces are completely different, as shown in Fig. 8. For the headform impacts at 6 m/s and 7.5 m/s speeds, the *VMB* peaks occur around 9 ms whereas the head is exposed to the maximum linear and rotational accelerations around 5 ms. The viscous properties of the brain probably postpone the peak of the distortional stress. For the dummy impacts, similar peaks are recognizable around 10 ms, which may be the consequence of the linear acceleration peak and rotational acceleration valley around 6 ms; however, these peaks are much less than those of the headform, even though the accelerations of the dummy head were higher for an impact speed of 7.5 m/s. This is probably due to the interaction between linear and rotational accelerations, which has alleviated the stress inside the brain. However, this interaction may exacerbate the brain stress (Ueno and Melvin, 1995). For instance, the effect of the low continuous linear acceleration of the dummy head after 11 ms added to the effect of the rotational acceleration second peak (these accelerations are both the consequences of the neck energy release) to produce the maximum of the *VMB* for the dummy impact at around 18 ms.
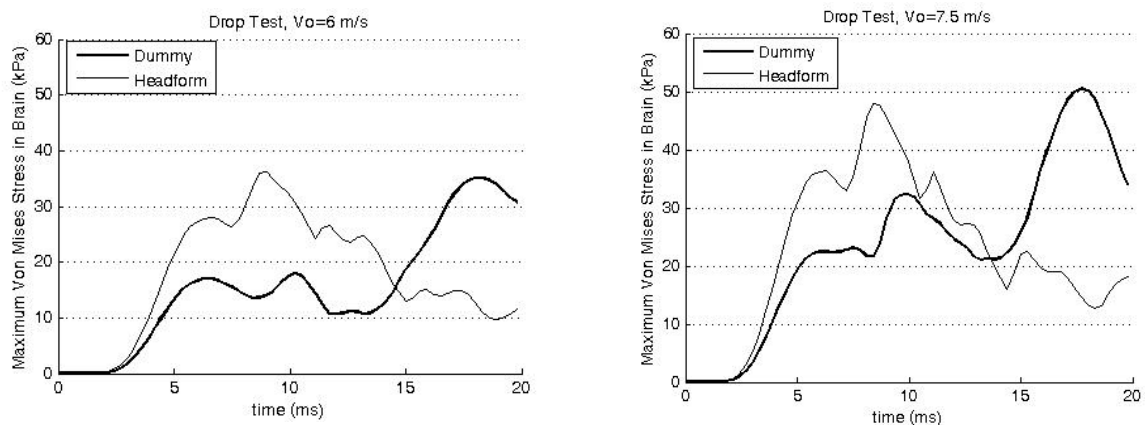

Fig. 8 – Results Obtained from the Head Injury Prediction Tool

The *IECSF* is larger in the headform drop tests compared to the dummy drop tests. More investigation of the time history of the CSF internal energy reveals that its maximum and the maximum of linear acceleration occur at the same time for all drop tests. At this time, the rotational velocity of the head of the dummy is about zero, but the rotational velocity of the headform is close to its peak value. When these velocities are applied to the rigid skull of the Strasbourg University head injury prediction tool, owing to the rotational inertia of the intracranial parts, higher relative skull/brain displacement occurs in the headform case, which in turn results in more distortion of the CSF. It is commonly acknowledged that the relative displacement between the skull and brain is the main cause of SDH (Horgan, 2005).

MODIFIED HEADFORM

It was shown that the presence of the whole body in helmet drop tests resulted in further crushing of the liner; as a result, when a helmet is designed to pass a standard drop test using a headform, its liner may bottom out when the whole body is present at the same impact velocity as that set in the standard. This indicates the important role of the body, which should be considered in energy absorption tests. Since using a dummy in standards has drastic impacts on the price of helmets, other measures should be found that are simple and economical.

As shown earlier, when the helmet liner is loaded below its energy absorption capacity, the peak linear acceleration of the head is lower using the dummy, but the maximum force on the anvil and the maximum compression of the liner are greater compared to the headform drop test. Referring to the

equations (8), (9) and (10), the only parameter that influences the impact outputs of a helmeted headform drop test in a similar way is the mass of the falling object. Since the peak of force and acceleration occur at the same time (Fig. 6 and Fig. 7), the equivalent mass of the helmeted headform can be obtained from the following relation:

$$m_e = \frac{NFA_{dummy}}{PLA_{dummy}} \tag{11}$$

which is the ratio of the maximum force on the anvil to the peak linear acceleration of the head measured in a dummy drop test. For instance, this ratio for the dummy impact at 6 m/s is (10.7kN/144g) 7.597kg. After subtracting the helmet mass from this value, the equivalent mass of the headform will be 7.051kg.

In order to examine the accuracy of this approach, two drop tests at 6 m/s and 7.5 m/s were simulated using a modified headform, and the results were then compared with the dummy drop tests. The modified headform was a revised version of the dummy detached headform; the mass and moment of inertia were multiplied by the ratio of the new mass of the headform to its original mass (4.769 kg), i.e. 1.479.

The time trace of the modified headform linear acceleration plotted in Fig. 9 follows that of the dummy very well for both impact speeds. The error in the predicted *PLA* is less than 10%. It has been underscored that both the acceleration level and its dwell time are indicators of head injury thus it is remarkable that the linear acceleration traces vs. time are comparable. As a result, *HIC*, which is a function of linear acceleration vs. time, was predicted accurately (Table 3). The maximum normal force on the head, which is an indicator of the skull fracture, was also predicted precisely using the modified headform. The most important output parameter is the maximum crush distance of the foam liner, which is replicated successfully by utilizing the modified headform; the error is less than 1.5 %.
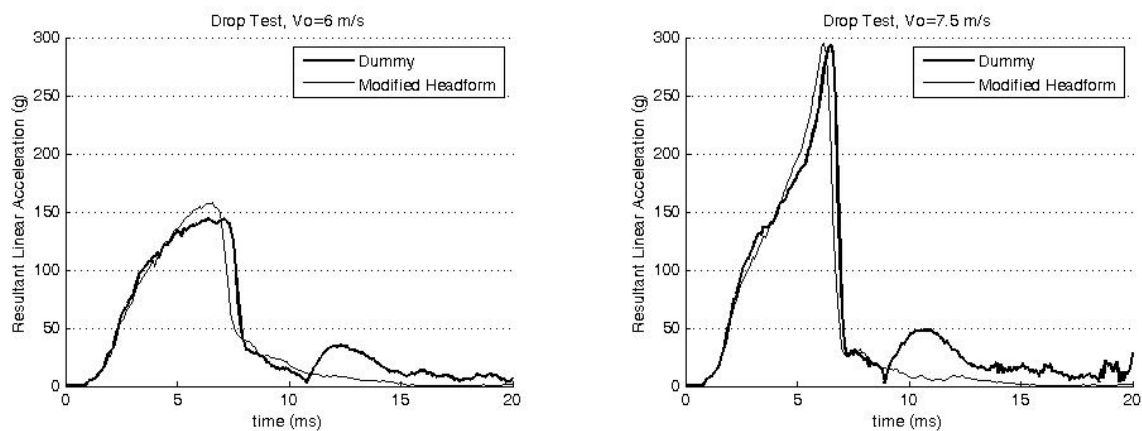

Fig. 9 – Linear Acceleration of the Head

Table 2. Results of Drop Test Simulations

	Helmeted Dummy		Helmeted Modified Head		⎮Modified Head Output-Dummy Output⎮/Dummy Output	
$V_0(m/s)$	6	7.5	6	7.5	6	7.5
PLA (g)	144	293	158	295	9.7%	0.1%
HIC	820	1872	796	1801	2.9%	3.8%
NFH (kN)	9.2	18.1	9.6	18.7	4.3%	3.3%
Td (mm)	17.8	22.2	17.6	22.1	1.1%	0.1%

To sum up, the closed-form equations and the FE simulations using the modified headform suggest that as long as the linear acceleration of the headform is the criterion for assessing the impact

absorption capacity of helmets, increasing the mass of the headform is a precise and economical way of considering the effect of the rest of the body.

DISCUSSION

It was indicated that the presence of the whole body in helmet drop tests reduced the *PLA* but increased the *NFA* and *Td* when the crushed region of the liner foam did not enter the densification zone of the foam's characteristic stress-strain curve (zone III in Fig. 4). These results are similar to the experimental results reported in COST (COST327, 2001) for a maximum impact speed of 6 m/s. However, an increase in the impact velocity from 6 m/s to 7.5 m/s revealed the catastrophic effect of the liner bottoming out on the *PLA* and *NFA*, which is a phenomenon that happens when the body is attached to the head. These results raise doubts about standard helmet testing procedures, which employ a detached headform in drop tests.

There is no evidence as to whether or not helmet standards have already taken into account the effect of the body. The simplest and most economical way of including the effect of the body in impacts is to use a detached headform but to change one (or more) impact variable(s). The closed-form equations have shown that the mass of the headform is probably the best variable, and the comparison between the results of drop tests using the dummy and a modified headform confirmed this assumption when the *PLA, NFA* and *Td* were taken into consideration. The dummy applied in this study was a 50th percentile adult male whose head is 4.8 *kg* close to the 4.7 *kg* mass of the middle size ISO headform employed by standards; hence, standards do no use a modified headform.

The equivalent mass is dependent on the dummy impact configuration, i.e. the impact site and the body impact angle. Table 3 shows the values of *PLA* and *NFA* obtained by dropping a helmeted Hybrid III pedestrian dummy at different impact configurations and velocities (COST327, 2001). In this table, B/30° and R/0° refer to the impact sites B (frontal) and R (rear) and the body impact angles 30° and 0°. Although a different helmet was used in this experimental study and the dummy was in standing posture, the modified headform mass for B/30° impacts is close to that obtained by the FEA of the frontal impact. The mean value of the experimental results is 6.55 kg, which is comparable to 7.0 kg of the FEA. For the impact configuration R/0°, a different value was obtained for the modified headform mass, but it is still independent of the impact speed. Dummy virtual drop tests will be performed for other frequent impact configurations to find corresponding equivalent masses.

Table 3. Results of Dropping a Helmeted Dummy onto a Flat Anvil (COST 327,2001)

	$V_0 (m/s)$	*PLA (g)*	*NFA (kN)*	Equivalent mass (kg)	Mass of the helmet (kg)	Mass of the modified headform (kg)
B/30°	4.4	85	6.507	7.804	1.384	6.420
	5.2	99	7.736	7.965	1.384	6.581
	6	111	8.758	8.042	1.384	6.659
R/0°	4.4	122	7.013	5.860	1.384	4.476
	5.2	142	8.071	5.794	1.384	4.410
	6	165	8.940	5.523	1.384	4.139

Another impact variable that can be changed to include the body effect in the simple headform dropping is the impact velocity. This parameter, however, does not affect all the outputs in the same way as the headform mass does (equations 8, 9 and 10). An increase in the impact velocity increases the maximum force and the maximum compression of the foam, but it also augments the peak linear acceleration of the head. Nonetheless, this change may be implemented in a standard test by increasing the limit of the *PLA* and other assessment functions, such as *HIC*, otherwise the helmets cannot be designed and optimized so that their whole capacity will be used in protecting the head. For instance, the maximum compressive strain of the foam for the dummy impact at 6 m/s (68%) was comparable to that of the headform impact at 7.5 m/s (70%). In addition, the *NFH*s of these impacts were close (Table 3). However, the *PLA* and *HIC* for the former impact were 144g and 820 while those of the latter impact were 243g and 1904. It may be argued that standard limits for these injury indicators are too high, but this implies that the helmets are efficient at an impact velocity lower than the test velocity. For instance, the helmets certified by ECE 22.05 are probably efficient at the 6 m/s impact

speed and those certified by Australian/New Zealand standard (AS/NZS1698, 2006) are protective at velocities lower than 6 m/s, set in this standard.

With the aid of the Strasbourg University head injury prediction tool, it was indicated that the accelerations measured from the dummy drop test at the 7.5 m/s impact speed lead to DAI and SDH. Even when the headform was employed, DAI and SDH were probable in this impact speed. Deck and Willinger (Deck and Willinger, 2006) had the same conclusion; they showed that a helmet certified in accordance with a standard may still cause neurological lesions at the same impact conditions. This conclusion shows the inconsistency between the tissue level and kinematic indicators, which originates from their different nature. The kinematic predictors are based on the kinematics of the head disregarding the internal parts, but the tissue level predictors are defined for the injuries of tissues comprising the head, thus being more bio-faithful than kinematic predictors. Nonetheless, the tissue level predictors are outputs of FE models and cannot be measured directly during tests. Moreover, the number of studies on head injuries using kinematic predictors is higher than those using tissue level predictors. As a result, it is still very difficult to standardize FEM based injury indicators.

The Hybrid III dummy was developed to study car rear impacts. Researchers believe that this dummy is not suitable for investigating direct impacts, such as motorcycle accidents, because its neck is too stiff. A field of future research would be to employ a more realistic FE model for the human neck and to investigate its effect on the injury predictors.

CONCLUSIONS

A commercial helmet was drop tested virtually using the HIII dummy and its detached head. It was shown that the presence of the body increases the crush distance of the liner compared to a test with the headform. This effect caused complete bottoming out of the liner at the 7.5 m/s impact speed and consequently large accelerations of the head and normal force on the head. Through the 1D analytical solution and the consequent FE simulations, it was shown that increasing the mass of the headform is a simple and economical way of considering the effect of the whole body.

The investigation of the helmet standards revealed that they probably do not consider the body inertia effect. Hence, designing the liner foam as the main energy absorbing part of the helmet to pass the standard energy absorption test may bear the risk of very high accelerations of the head and skull fracture in real impact situations with head impact velocities close to impact velocities set in the relevant standards.

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