

## **HELMET OPTIMISATION BASED ON HEAD-HELMET MODELLING**

**Deck C., Baumgartner B., Willinger R.**

*Université Louis Pasteur de Strasbourg IMFS – UMR 7507 ULP-CNRS France.  
willi@imfs.u-strasbg.fr*

### **ABSTRACT**

The aim of this work is to optimise a full face helmet finite element model based on the dynamic behaviour of its components against biomechanical criteria. It is well known that helmets substantially reduce head injury, although the mechanism of this protection is neither well understood nor controlled. Moreover, today helmets are designed to reduce headform deceleration and not optimised to reduce head injury. The helmet used in this study is a full face helmet with a polycarbonate thermoplastic shell and an expanded polystyrene foam, certified to BS6658A [1]. The validation of the helmet FEM corresponds to the impact test stipulated by the British Standard BS 6658A, and the ECE-R022/04 [2]. After a validation with a headform FE model as used in the experimental normative tests, the helmet model was coupled with a previously developed finite element model of the human head . This approach consists to couple the human head with the helmet FE models in order to predict intra-cranial field parameters such as acceleration, pressure and Von Mises stress. Concerning the coupling with the human head, a frontal impact has been simulated with the same boundary conditions as for the normative test with standard helmet mechanical properties. The brain pressure varied from -94 kPa to 350 kPa. These values were higher than the brain tolerance limits for visible injuries proposed in the literature, which are -180 kPa in tension and 234 kPa in compression. The highest Von Mises shear stress values in the brain were about 47 kPa which is close to the 20 kPa limit proposed in the literature for neurological injuries. The final step of the study then consists to optimise numerically the helmet mechanical parameters against biomechanical criteria such as intra-cerebral stress levels. In order to define the influence of the helmet shell and foam properties on the human head, a parametric study of the model was undertaken and all results were analyse with a PCA method. This study led then to the conclusion that the foam elastic limit has the most important influence on biomechanical head response.

# 1 INTRODUCTION

Security is of high importance in our modern way of living from both social and economical point of view. The helmet is the most current protection system of the head. The main function of the helmet is to reduce or to avoid injuries that may occur to the head during an impact. In fact, transport accidents are the main cause of head injuries. A study led on real world frontal impacts shows that the head is the most currently injured anatomical segment by considering serious injuries. The issue of such an impact is usually catastrophic to fatal for the victim.

The main mechanical parameter that causes head injuries is assumed to be its linear acceleration. In the early seventies, the Head Injury Criterion (HIC) has been therefore developed, based on the linear acceleration sustained by the head coupled with its duration. From an experimental point of view, this criterion is determined through the measurement of the three dimensional linear acceleration of the centre of gravity of a rigid dummy head which has similar inertial properties than the human head. Even though the HIC is able to represent the global severity level of an impact, and the potential head injury level, the specialists agree to claim that the HIC is unable to predict diffuse brain injuries and subdural haematoma that are linked to the angular acceleration sustained by the head during the impact. The development of protection helmets was always led in accordance with the injury risks encountered.

Very few helmet FE models are reported in the literature. Köstner [KOS.87], and Van Schalswijk [VAN.93] were the first to propose helmet models but these were purely descriptive and were not validated. Brands et al. [BRA.96] developed a three dimensional helmet fitted to a dummy head based on elastic material for the shell and an elastoplastic material for the liner. The liner was assumed to be glued to the shell although the detailed helmet characteristics were not modelled. The model validation relied upon the headform acceleration and the general head-helmet kinematics for a frontal, top and rear impact situation. Lateral impact simulations were not possible with this model because of excessive headform rotation with respect to the helmet. Hence, although the first phase of the acceleration and peak value was acceptable, the time at which this maximum occurred and the rebound velocity were not correctly predicted, thus indicating that the energy absorption was incorrectly modelled. Nevertheless, the HIC remains the single normative parameter used to validate a helmet in terms of protection against impact. In a previous study Willinger [WIL.00b] coupled a human head model to a helmet finite element model in order to calculate the head response under normative impact. In this study, finite element modelling was used to calculate the brain pressure and Von Mises stress sustained during a frontal impact, and to compare these values to the tolerance limits proposed in the literature.

Main objective of the present work is a parametric study to optimise helmet performance against biomechanical criteria. A principal component analysis is used to analyse head responses as a function of helmet characteristics, and to compare the conclusion with helmet optimisation against HIC and using a standard headform.

## 2 HELMET MODELLING AND VALIDATION

The helmet used in this study was a full face helmet with a non-reinforced polycarbonate thermoplastic shell and an expanded polystyrene foam, and certified to BS6658A[BRI.85]. The geometry was determined by digitising the external shell surface and the helmet shell was meshed with shell elements. Brick elements, obtained by "extrusion" of the shell surface, were used to model the foam.

Concerning material properties summarize in table 1, characteristics for the protective foam liner were obtained from dynamic compression tests on foam samples by Willinger and al.[WIL.00a]. In order to determinate shell Young's modulus, and to validate the shell global dynamic behaviour, an experimental and numerical analysis of the shell was performed (Willinger and al.[WIL.00b]).

Component	Material	Model	E [Gpa]	$\nu$	$\rho$ [kg/m <sup>3</sup> ]	Comment
Outer shell	thermo-plastic	linear-elastic	1.5	0.35	1055	Thickness 4mm
Protective padding	expanded polystyrene	elasto-plastic	1.5e <sup>-3</sup>	0.05	25	Thickness 40mm yield stress = 0.35 MPa
Headform	aluminium	rigid	27	0.3	-	Mass= 4.27 kg

Table 1 : Material and modelling data for the full-face helmet model.

The validation of the helmet FEM corresponds to the frontal impact test stipulated by the British Standard BS6658A [BRI.85] and the ECE-R022/04 [BRI.94]. For this purpose a headform FE model was coupled with this helmet and launched freely against a rigid anvil with an initial velocity of 7.5 m/s.

For the helmet validation, the results of the numerical simulation were compared with those of the experimental tests in terms of linear acceleration of the headform (figure 1-a) and the force-displacement time histories (figure 1-b) produced by the headform. The shape, the delay and the magnitude of the acceleration produced by the model agrees very well with the experimental data.

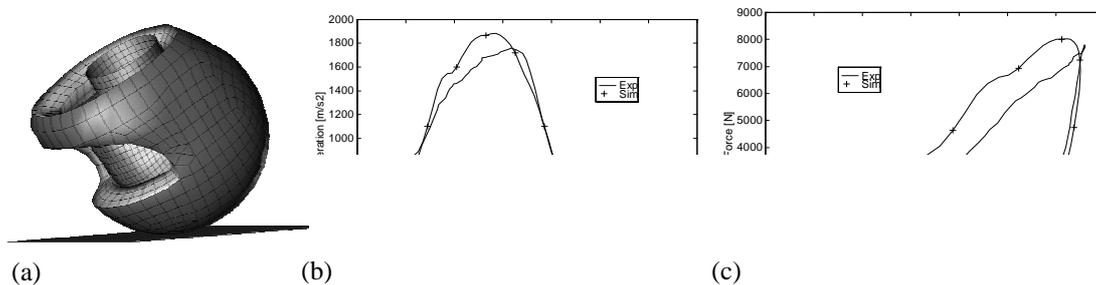


Fig 1 : Headform-helmet coupling (a), Numerical headform acceleration resultant (b) and force-displacement resultant curve (c) against experimental data.

### 3 HUMAN HEAD MODELING AND VALIDATION

Strasbourg University has developed the "ULP" (Université Louis Pasteur) human head FE model (Kang et al. [KAN.97]) represented in figure 2. The present ULP model includes the main anatomical features: skull, falx, tentorium, subarachnoid space, scalp, cerebrum, cerebellum, brain stem. Falx and tentorium have a layer of shell elements, skull is simulated by three layered composite shell and the others were constituted by brick elements. The finite element mesh is continuous and represents an adult human head. The subarachnoid space was modelled between the brain and the skull to simulate the cerebral-spinal fluid. This space is constituted by a layer of brick elements and surrounds entirely the brain. The tentorium separates the cerebrum and cerebellum and the falx separates two hemispheres. A layer of brick element simulating the cerebral-spinal fluid surrounds these membranes. The scalp was modelled by a layer of brick elements and surrounds the skull and facial bone. Globally, the present human head model consists of 13208 elements. Its total mass is 4.5 kg.

Material properties assigned to the different parts are all isotropic, homogenous and elastic. The Young's modulus of the brain and the subarachnoid space were found by Willinger et al. by modal analysis [WIL.95]. The viscoelastic properties assigned to the brain were scaled from Khalil [KAL. 77]. The behaviour in shear was defined by:  $G(t) = G_{\infty} + (G_0 - G_{\infty})\text{Exp}(-\beta t)$  with  $G_0$ : Short term shear modulus,  $G_{\infty}$ : Long term shear modulus and  $\beta$ : Decay constant.

The skull was modelled by a three layered composite shell representing the inner table, the diploë and the external table of human cranial bone. In order to reproduce the overall compliance of cranial bone, a thickness in combination with an elastic brittle law were selected for each layer. In order to model the material discontinuity in the case of fracture, it was necessary to use values for the limiting (ultimate) tensile and compressive stress (UTS and UCS in table 1) obtained from Piekarski 1970 [PIE.70] and integrated in the Tsai-Wu criterion.

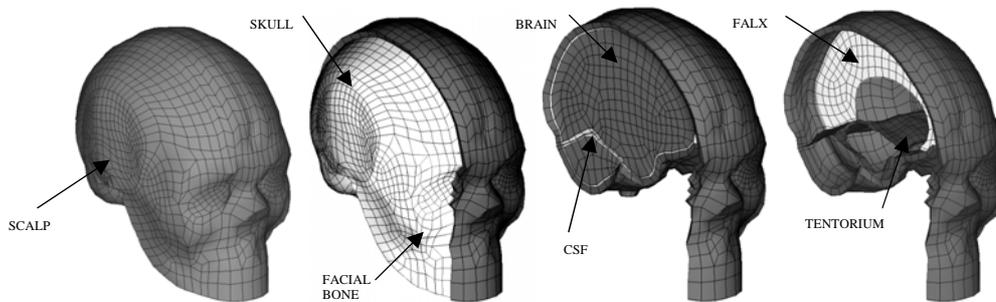


Fig 2 : Finite element model of the head (ULP model).

A total of eight instrumented cadaver impacts were reconstructed with the objective of validating the ULP model under very different impact conditions. Currently head FE models are validated against Nahum's et al. impact [NAH.77] and have moreover been validated against other experimental data as those of Trosseille et al. [TRO.92] for high damped long impact durations, and those of Yoganandan [YOG.94] for very short impact durations including bone fracture.

## 4 HEAD HELMET COUPLING IN NORMATIVE IMPACT

In this part of the study, the helmet model was coupled with the finite element model of the human head developed previously (Kang et al. [KAN.97]) as shown in figure 3-a.

This frontal impact was modelled with an initial velocity of 7.5 m/s (figure 3-b) as stipulated by the standard.

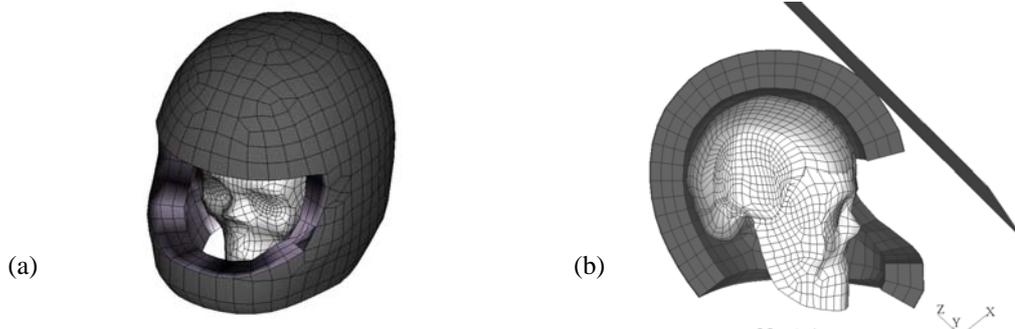


Fig 3 : (a) Coupling of the human head and the helmet FE models .  
(b) Configuration of the frontal impact.

The frontal and occipital pressure time histories are shown in figure 4-a. The pressure was uniformly distributed across the brain, with compression in the frontal region (coup) and tension in the occipital region (contre-coup). The model predicted a maximum compression of 350 kPa in impact area and a maximum tension of -94 kPa at the opposite point. From figure 4-b, it appears that high Von Mises stresses occurred in the occipital area, in the base of the brain (about 31 kPa) and in the brain stem, where the maximum value calculated for the Von Mises stress was 40 kPa at some locations. The distribution of the Von Mises stress may be related to the topology of the skull, particularly the base which is a very irregular surface.

This simulation leads to intracerebral loading above tolerance limits which shows that today helmets are designed to resist to impacts which the human head probably doesn't support without injury.

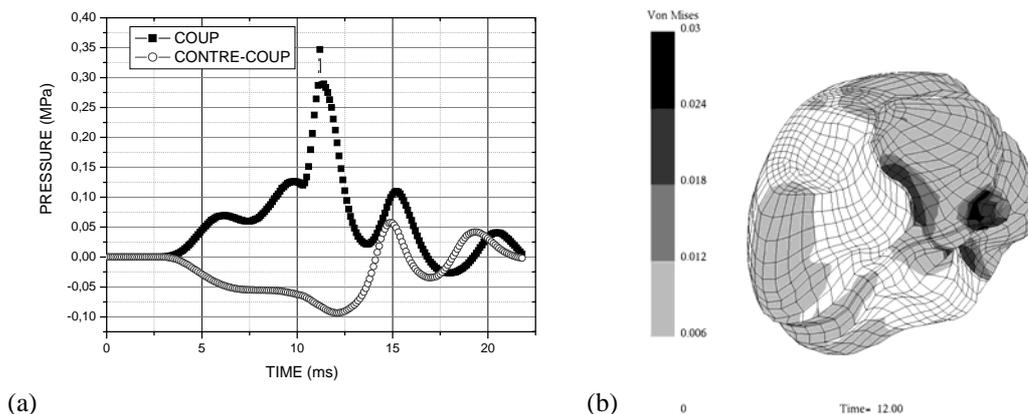


Fig 4 : (a) Pressure time history in the pole and the opposite point of the brain,  
(b) Brain Von Mises stress field 12 ms after the beginning of the impact.

## 5 PARAMETRIC STUDY AND HELMET OPTIMISATION

After a first simulation of a frontal impact under standard conditions which permits it to estimate the intra-cranial parameters under standard impact, we proceeded to the helmet optimisation study. In order to define the influence of the helmet mechanical characteristics on the intra-cranial response, a parametric study was undertaken. Four mechanical parameters have been varied: the foam elastic limit (D) and Young modulus (A), the thickness of the shell (B) and its Young modulus (C). Each parameter has been set on three different values: the reference value used in the model validation, a high (+30%) and a low (-30%) value. The tests used for the parametric study remain the drop test on a flat anvil in frontal impact situation (figure 5) at 7.5 m/s initial velocity as illustrated in figure 5.

The head response for a given simulation was calculated in terms of intra-cerebral pressure and Von Mises shearing stresses. A total of 16 simulations were run with a simulation protocol illustrated on table 2. An identical study was performed by replacing the human head model by a standard headform, as illustrated in Figure 1a, in order to identify the influence of helmet properties on headform acceleration or HIC value. This second investigation was designed to define the optimum helmet against the classical HIC criterion and to compare this result with the previous study where biomechanical were criteria considered. The histograms show in figures 5-a,b represent the maximum intra-cerebral pressure and Von Mises stress calculated for each simulation. Concerning the coupling of the helmet model with the finite element model of the human head, the maximum values ranged respectively from 150 kPa (S8) to 1550 kPa (S5) for the pressure and from 17.7 kPa (S8) to 76.1 kPa (S6) for the Von Mises shearing stress. Figure 5-c is relative to the parametric study using a standard headform.

To analyze all results, a principal component analysis (PCA), Volle [VOL.97], was performed by considering seven variables (four helmet properties and three head response parameters).

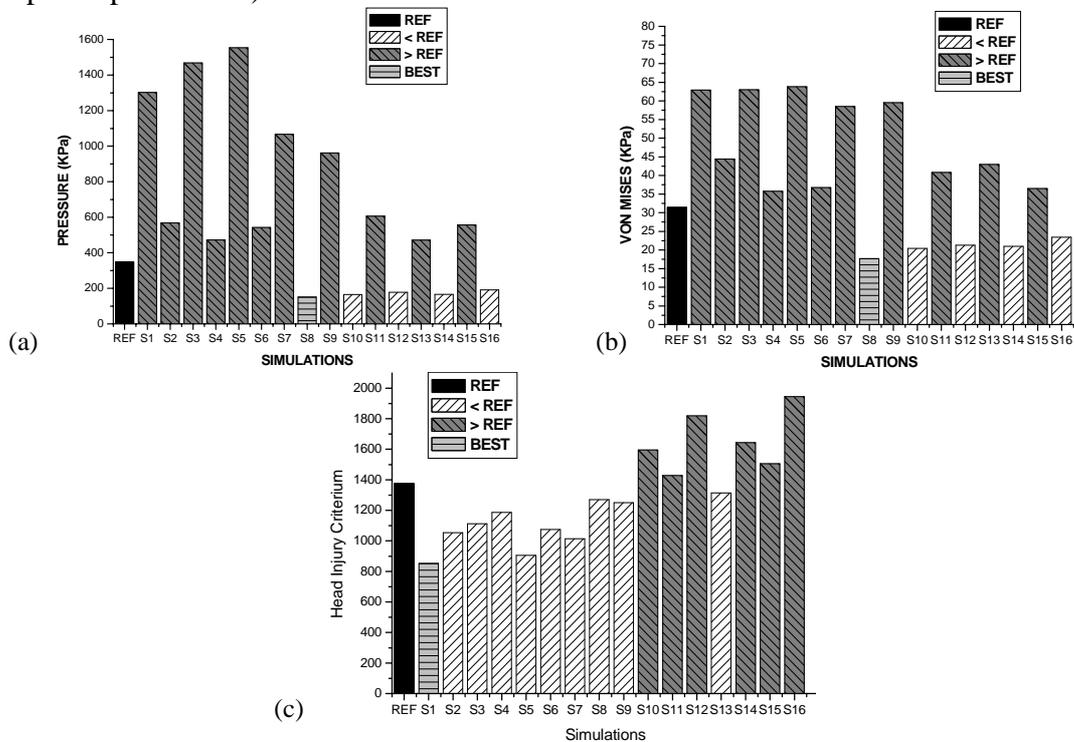


Figure 5 : Maximum intra-cerebral pressure (a), Von Mises stress (b) and HIC value (c) calculated for each simulation (REF = reference helmet).

Table 2 : Simulation protocol indicating for each of the 16 simulations, the helmet characteristics retained: +/- stand +30% or -30% of the reference helmet properties.

	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	S12	S13	S14	S15	S16
A	-	+	-	+	-	+	-	+	-	+	-	+	-	+	-	+
B	-	-	+	+	-	-	+	+	-	-	+	+	-	-	+	+
C	-	-	-	-	+	+	+	+	-	-	-	-	+	+	+	+
D	-	-	-	-	-	-	-	-	+	+	+	+	+	+	+	+

The principle of the PCA is to research the best data representation with the less possible dimensions to reduce the number of variables or the initial space dimension number. This consequently allows to explain and to display data with a reduced number of axes in order to facilitate the interpretation of synoptic results. The first result is the correlation matrix reported in table 3. From this, we can observe that some of the variables are highly correlated which means that they move together. We can mention for example that input variables B and C have low correlation with all output variables (P, VM, HIC). On the other hand, the foam elastic limit (D) is highly correlated with HIC criterion (0.811) and pressure and von Mises shear stress are highly correlated with A (respectively -0.741 and -0.785) coupled with D (-0.516 and -0.448 respectively).

Table 3 : Correlation matrix between the N=7 variables.

	A	B	C	D	HIC	P	VM
A	1	0	0	0	0.442	-0.741	-0.785
B	0	1	0	0.003	0.322	-0.151	-0.219
C	0	0	1	0	0.075	-0.136	-0.17
D	0	0.003	0	1	0.811	-0.516	-0.448
HIC	0.442	0.322	0.075	0.811	1	0.751	-0.772
P	-0.741	-0.151	-0.136	-0.516	-0.751	1	0.957
VM	-0.785	-0.219	-0.17	-0.448	-0.772	0.957	1

The next step is then to calculate the principal components. Here the correlation matrix (table 3) is considered in a mathematical point of view. For this symmetric matrix (7\*7) the eigenvalues and eigenvectors are then determined. These eigenvalues reflect the quality of the projection from the N-dimensional initial table (N=7 in this study) to a lower number of dimensions. Each eigenvalue corresponds to a factor which is a linear combination of the initial variables, and all the factors are un-correlated ( $r=0$ ). The eigenvalues and the corresponding factors are sorted by descending order of how much of the initial variability they represent (converted to %). The eigenvector associated with the largest eigenvalue has the same direction as the first principal component. The eigenvector associated with the second largest eigenvalue determines the direction of the second principal component. The sum of the eigenvalues equals the trace of the square matrix and the maximum number of eigenvectors equals the number of rows (or columns) of the correlation's matrix. These axes are defined by linear forms (1) and (2):

$$AxisF1 = -0.372A - 0.129B - 0.074C - 0.331D - 0.474HIC + 0.501P + 0.505VM \quad (1)$$

$$AxisF2 = 0.591A - 0.256B + 0.156C - 0.625D - 0.336HIC - 0.145P - 0.190VM \quad (2)$$

Table 4 : Eigenvalues.

	F1	F2	F3	F4	F5	F6	F7
Eigenvalue	3.6276	1.1818	1.003	0.9996	0.1490	0.0327	0.0063
Variance %	51.82	16.88	14.33	14.28	2.13	0.47	0.09
cumulated	51.82	68.7	83.03	97.31	99.44	99.91	100

Ideally, the first two or three eigenvalues will correspond to a high percentage of the variance, ensuring us that the maps based on the first two or three factors are a good quality projection of the initial multi-dimensional table. In this study, the first two factors allow us to represent 68.7% of the initial variability of the data. These factors are given in table 4.

The correlation circle represented in figure 6-a is useful in interpreting the meaning of the axes. It shows a projection of the initial variables in the factors space. In this study, the horizontal axis which represent 51.82% of the variability (first eigenvalue equals 3.6276), is linked with HIC criterion (0.474), pressure (0.501), and Von Mises shear stresses (0.505). Along F2 which describes 16.88% of the variability, the main important parameters are: Young modulus of the foam (0.591) and its elastic limit (0.625).

Figure 6-b is the ultimate goal of the PCA. It permits to look at the data on a two-dimensional map, and to identify trends. We can see that simulations, which represents the maximum of pressure and Von Mises, are classified from the left (less value) to the right (high value) along F1 from 1 to 17 (17 represents the reference). We can note that the best simulations in terms of pressure and von Mises are localize in the portion of space describes by  $F2 \leq 0$ . It corresponds to the influence of the young modulus of foam and its elastic limit.

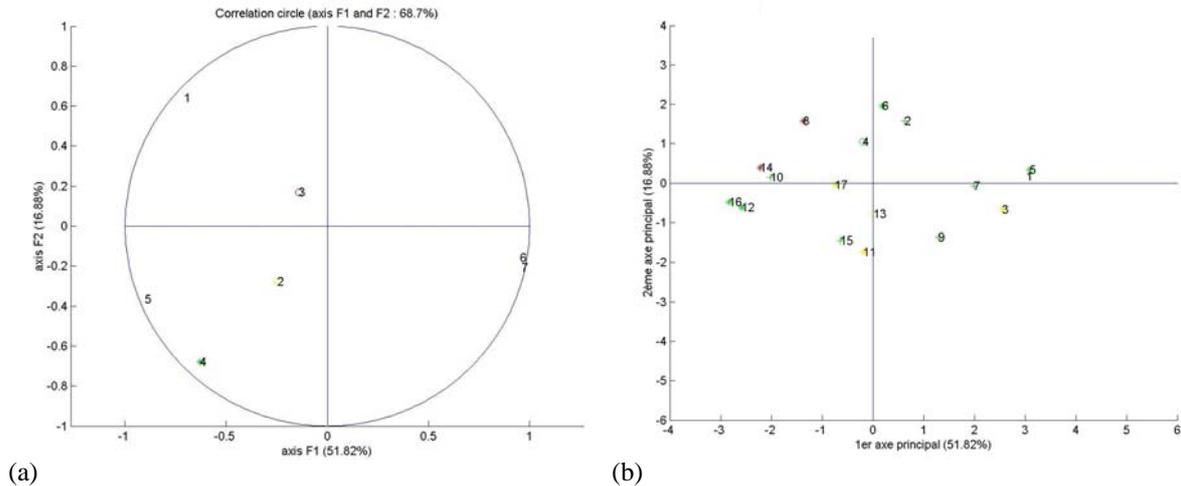


Fig -6 : PCA correlation circle of the 7 variables (a), Factorial plane (b).

## 6 DISCUSSION

The helmet model described in this paper was validated in frontal impact configuration. The coupling of this helmet and the finite element model of the human head predicts intra-cerebral stresses higher than the tolerance limits under normative impact conditions. In this study we propose to optimise the helmet performance against biomechanical criteria. The PCA method used to analyse results, led to the conclusion that the foam elastic limit has an important influence on HIC criterion but it's the young modulus of foam and its elastic limit which have an influence on biomechanical head's response. Moreover, this study permitted us to propose an optimisation in terms of intra-cerebral pressure and von Mises stress with the configurations S8, S10, S12, S14 and S16 (figures 5-a, 5-b). Nevertheless, the HIC remains the single normative parameter used for helmet homologation in terms of protection against impacts. So the same parametric study was performed with a standard headform. Histogram in figure 5-c shows this maximum value for each simulation. The analysis of these results shows that the optimum helmet is the configuration S1.

So, an optimisation based on biomechanical criteria is different than the optimisation with HIC criterion which is correlated with acceleration of headform's centre of mass and used for helmets homologation.

## 7 CONCLUSION

A finite element model of an existing helmet has been proposed and validated in frontal impact configuration. The coupling with an anatomic head was performed and a first impact has been simulated with the same boundary conditions as for the normative test. The model predicts a maximum compression of 350 kPa in the region of the impact site and a maximum tension of -94 kPa at the opposite point. Moreover Von Mises stress which is a good indicator of concussion reached a maximum value of 40 kPa. These values are higher than the 20 kPa tolerance limits proposed by Willinger [WIL.00]. We can therefore conclude that normative impact led some lesions. The presented model, allow parametric study that may be used to optimise helmet performance against biomechanical criteria. A principal component analysis was performed to analyse head responses as a function of helmet characteristics. A similar study was conducted by replacing the human head model by a headform model and showed that optimisation against headform's response and human head's response does not lead to the same results. Moreover, this study permitted us to conclude that the foam elastic limit has the most important influence on HIC response but it's the young modulus of foam and its elastic limit which have an influence on biomechanical head's response.

## REFERENCES

- [BRA.96] Brands, D.W.A., Thunissen, J.G.M., Wismans, J.S.H.M., Modelling head injury countermeasures : a 3D helmet model, *AGARD Specialists meeting on Impact Head Injury : Response, Mechanisms, Tolerance, Treatment and countermeasures*, Mescalero, New Mexico, USA, 1996.
- [BRI.85] British Standard Specification for Protective Helmets for Vehicle Users, BS 6658, 1985.
- [CHU.95] Chun Zhou, Khalil T. B., King A. I., A 3D human finite element head model for impact injury analyses. *Symposium Proc. Prevention Through Biomechanics*, pp. 137-148, 1995.

- [KAN.97] Kang, H.S., Willinger, R., Diaw, B.M., Chinn, B.P., Validation of a 3D anatomic head model and replication of head impact in motorcycle accident by finite element modelling, *41<sup>st</sup> Stapp Car Crash Conf.*, Miami, Florida., pp. 329-338, 1997.
- [KHA.77] Khalil T.B., Hubbard R.P., Parametric study of head response by finite element modelling, *J. of Biomechanics*, Vol. 10, 1977, p119-132.
- [KOS.87] Köstner, H., Stöcher, U.W., Mathematische analyse der stossabsorption im schutzhelmmaterial, *VDI-bericht*, Vol. 657, pp. 211-244, 1987.
- [NAH.77] Nahum, A.M., Smith, R., Ward, C.C., Intracranial pressure dynamics during head impact, *Proceed. of the 21st Stapp Car Crash Conf.*, SAE Paper 770922, pp. 339-366, 1977.
- [PIE.77] Piekarski, Fracture of bone. *J. Appl. Phys.* 14, N°1, 215-223 (1970).
- [TRO.92] Troselle, X., Tarrière, C., Lavaste, F., Guillon, F., Domont, A., Development of a F.E.M. of the human head according to a specific test protocol, *Proceed. of the 36th Stapp Car Crash Conf.*, pp. 235-253, 1992.
- [UNI.94] United Nations, "Regulation N° 22", Geneva, CH, 1994.
- [VAN.93] Van Schalkwijk, R., Helmet shock simulation with MARC using a hypo-elastic foam model, *MARC Analysis Research Corporation*, MTR-9304, 1993.
- [VOL.97] Volle, M., *Analyse des données*, 4<sup>ème</sup> édition Economica, 1997
- [WIL.00a] Willinger, R., Baumgartner, D., Chinn, B., Neale, M., Head tolerance limits derived from numerical replication of real world accidents, *Proceed. of IRCOBI Conf.*, pp. 209-221, 2000.
- [WIL.00b] Willinger, R., Baumgartner, D., Guimberteau, T., Dynamic characterization of motorcycle helmets : modelling and coupling with the human head. *Journal of Sound and Vibration*, vol. 235, pp. 611-625, 2000.
- [WIL.95] Willinger R., Taleb L., Pradoura P., Head biomechanics from the finite element model to the physical model. *Proc. IRCOBI*, pp 245-260, BRUNNEN, 1995.
- [YOG.94] Yoganandan, N., Biomechanics of Skull Fracture, *Proceed. of Head Injury 94 Symposium*, Washington DC, 1994.