Helmet optimisation based on head-helmet modelling

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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 intracranial 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 -101 kPa to 267 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 or 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 haematomas 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 [3], and Van Schalwijk [4] were the first to propose helmet models but these were purely descriptive and were not validated. Brands et al. [5] 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 [6] 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[1]. 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.[7]. In order to determinate shell Young’s module, and to validate the shell global dynamic behaviour, an experimental and numerical analysis of the shell was performed (Willinger and al.[7]).

<table>
<thead>
<tr>
<th>Component</th>
<th>Material</th>
<th>Model</th>
<th>E [GPa]</th>
<th>ν</th>
<th>ρ [kg/m³]</th>
<th>Comment</th>
</tr>
</thead>
<tbody>
<tr>
<td>Outer shell</td>
<td>thermo-plastic</td>
<td>linear-elastic</td>
<td>1.5</td>
<td>0.35</td>
<td>1055</td>
<td>thickness= 4 mm</td>
</tr>
<tr>
<td>Protective padding</td>
<td>expanded polystyrene</td>
<td>elastoplastic</td>
<td>1.5e³</td>
<td>0.05</td>
<td>25</td>
<td>thickness= 40 mm, yield stress= 0.35 MPa</td>
</tr>
<tr>
<td>Headform</td>
<td>aluminium</td>
<td>rigid</td>
<td>27</td>
<td>0.3</td>
<td>-</td>
<td>Mass= 4.27 kg</td>
</tr>
</tbody>
</table>

The validation of the helmet FEM corresponds to the frontal impact test stipulated by the British Standard BS6658A[1] and the ECE-R022/04 [2]. 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 model. The shape, the delay and the magnitude of the acceleration produced by the model agrees very well with the experimental data.

![Figure 1](image1.png)

**Figure 1**: Numerical headform acceleration resultant (a) and force-displacement resultant curve (b) 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. [8]) 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 theses 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 [9]. The bulk modulus of these parts and the material properties of the other parts are similar to those used in WSU model (Chun Zhou [10]), excepted the skull modelled by three layered composite skull with a elastic brittle law. This skull modelling permits us to simulate the bone fracture introducing material discontinuity and then to analyse its effects on the head response in case of head impacts involving skull fracture.

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 [11] and has moreover been validated against other experimental data as those of Trosselle et al. [12] for high damped long impact durations, and those of Yoganandan [13] for very short impact durations including bone fracture.

Figure 2 : Finite element model of the head (ULP model).
4 Head helmet coupling in normative impact

In this part of the study, the helmet model was coupled with our finite element model of the human head (ULP model) as shown in figure 2-a. This frontal impact was modelled with an initial velocity of 7.5 m/s (figure 3-b) as stipulated by the standard.

(a)  (b)

Figure 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 (centre-coup). The model predicted a maximum compression of 267 kPa in the region of the impact site and a maximum tension of -101 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 14 kPa) and in the brain stem, where the maximum value calculated for the Von Mises stress was 38.5 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.

(a)  (b)

Figure 4: (a) Pressure time history in the pole and the opposite point of the brain for a frontal impact. (b) Von Mises stress time history at different locations in the brain.
5 Parametric study and helmet optimisation

In order to define the influence of the helmet mechanical characteristics on the intra-cranial response, a parametric study of the model 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. The parameters calculated for a given simulation were the intra-cranial pressure (P), Von Mises shearing stresses (VM) as well as the skull and brain accelerations at their centre of mass. A total of 16 simulations was run with a simulation protocol illustrated on table 2. The histograms on figures 5-a,b represent the maximum intra-cerebral pressure and Von Mises stress calculated for each simulation. The value ranged respectively from 19.4 kPa (S3) to 42.9 kPa (S16) for the pressure and from 25.3 kPa (S3) to 45.3 kPa (S16) for the Von Mises shearing stress. To analyze all results, a principal component analysis (PCA), Volle [14], was performed by considering eight variables (four helmet properties and four head response parameters).

Table 2: Simulation protocol indicating for each of the 16 simulations, the helmet characteristics retained: +/- stand +30% or -30% of the reference helmet properties. The foam elastic limit (D) and Young modulus (A), the thickness of the shell (B) and its Young modulus (C), the pressure (P), Von Mises (VM), Brain acceleration (\(\Gamma_{\text{Brain}}\)) and skull acceleration (\(\Gamma_{\text{Skull}}\)).

<table>
<thead>
<tr>
<th></th>
<th>S1</th>
<th>S2</th>
<th>S3</th>
<th>S4</th>
<th>S5</th>
<th>S6</th>
<th>S7</th>
<th>S8</th>
<th>S9</th>
<th>S10</th>
<th>S11</th>
<th>S12</th>
<th>S13</th>
<th>S14</th>
<th>S15</th>
<th>S16</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>-</td>
<td>+</td>
<td>-</td>
<td>-</td>
<td>+</td>
<td>-</td>
<td>+</td>
<td>-</td>
<td>+</td>
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<td>-</td>
<td>+</td>
<td>-</td>
<td>-</td>
<td>+</td>
<td></td>
</tr>
<tr>
<td>B</td>
<td>-</td>
<td>-</td>
<td>+</td>
<td>-</td>
<td>-</td>
<td>+</td>
<td>+</td>
<td>-</td>
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<td>+</td>
<td>+</td>
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<td>-</td>
<td>+</td>
<td></td>
</tr>
<tr>
<td>C</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>+</td>
<td>-</td>
<td>-</td>
<td>+</td>
<td>+</td>
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<td>+</td>
<td>+</td>
<td>+</td>
<td>+</td>
<td></td>
</tr>
</tbody>
</table>

Figure 5: Maximum intra-cerebral pressure (a) and Von Mises stress (b) calculated for each simulation (REF = reference helmet).
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 A, B and C have low correlation with all output variables (P, VM, \( \Gamma_{\text{Brain}} \), \( \Gamma_{\text{Skull}} \)). On the other hand, the foam elastic limit (D) is highly correlated with brain and skull accelerations as well as pressure (0.85).

**Table 3:** Correlation matrix between the N=8 variables: the foam elastic limit (D) and Young modulus (A), the thickness of the shell (B) and its Young modulus (C), the pressure (P), Von Mises (VM), Brain acceleration (\( \Gamma_{\text{Brain}} \)) and skull acceleration (\( \Gamma_{\text{Skull}} \)).

<table>
<thead>
<tr>
<th></th>
<th>A</th>
<th>B</th>
<th>C</th>
<th>D</th>
<th>P</th>
<th>VM</th>
<th>( \Gamma_{\text{Brain}} )</th>
<th>( \Gamma_{\text{Skull}} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>1</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0.247</td>
<td>0.075</td>
<td>0.142</td>
<td>0.262</td>
</tr>
<tr>
<td>B</td>
<td>0</td>
<td>1</td>
<td>0</td>
<td>0</td>
<td>0.003</td>
<td>0.333</td>
<td>0.147</td>
<td>0.098</td>
</tr>
<tr>
<td>C</td>
<td>0</td>
<td>0</td>
<td>1</td>
<td>0</td>
<td>0.253</td>
<td>0.299</td>
<td>0.224</td>
<td>0.207</td>
</tr>
<tr>
<td>D</td>
<td>0</td>
<td>0.003</td>
<td>0</td>
<td>1</td>
<td>0</td>
<td>0.823</td>
<td>0.499</td>
<td>0.896</td>
</tr>
<tr>
<td>P</td>
<td>0.247</td>
<td>0.333</td>
<td>0.253</td>
<td>0.823</td>
<td>1</td>
<td>0.487</td>
<td>0.949</td>
<td>0.981</td>
</tr>
<tr>
<td>VM</td>
<td>0.075</td>
<td>0.147</td>
<td>0.299</td>
<td>0.499</td>
<td>0.487</td>
<td>1</td>
<td>0.581</td>
<td>0.436</td>
</tr>
<tr>
<td>( \Gamma_{\text{Brain}} )</td>
<td>0.142</td>
<td>0.098</td>
<td>0.224</td>
<td>0.896</td>
<td>0.949</td>
<td>0.581</td>
<td>1</td>
<td>0.921</td>
</tr>
<tr>
<td>( \Gamma_{\text{Skull}} )</td>
<td>0.262</td>
<td>0.367</td>
<td>0.207</td>
<td>0.855</td>
<td>0.981</td>
<td>0.436</td>
<td>0.921</td>
<td>1</td>
</tr>
</tbody>
</table>

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 (8*8) the eigenvalues and eigenvectors are then determined. These eigenvalues reflect the quality of the projection from the N-dimensional initial table (N=8 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 axis are defined by linear forms (1) and (2):

\[
\text{Axis}F1=0.091A+0.107B+0.132C+0.436D+0.479P+0.301VM+0.477\Gamma_{\text{Brain}}+0.475\Gamma_{\text{Skull}}
\]

\[
\text{Axis}F2=0.397A+0.644B-0.313C-0.092D+0.141P-0.509VM-0.062\Gamma_{\text{Brain}}+0.194\Gamma_{\text{Skull}}
\]
Table 4: Eigenvalues.

<table>
<thead>
<tr>
<th>Eigenvalue</th>
<th>F1</th>
<th>F2</th>
<th>F3</th>
<th>F4</th>
<th>F5</th>
<th>F6</th>
<th>F7</th>
<th>F8</th>
</tr>
</thead>
<tbody>
<tr>
<td>4.164</td>
<td>1.289</td>
<td>1.011</td>
<td>1.001</td>
<td>0.457</td>
<td>0.07</td>
<td>0.014</td>
<td>0.004</td>
<td></td>
</tr>
<tr>
<td>Variance %</td>
<td>52.06</td>
<td>16.12</td>
<td>12.64</td>
<td>12.51</td>
<td>5.58</td>
<td>0.87</td>
<td>0.18</td>
<td>0.05</td>
</tr>
<tr>
<td>cumulative</td>
<td>52.06</td>
<td>68.18</td>
<td>80.82</td>
<td>93.33</td>
<td>98.91</td>
<td>99.78</td>
<td>99.96</td>
<td>100</td>
</tr>
</tbody>
</table>

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% 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 52% of the variability (first eigenvalue equals 4.164), is linked with foam elastic limit (0.436), pressure (0.479), brain (0.477) and skull’s (0.475) accelerations. Along F2 which describes 16% of the variability, the main important parameters are: Von Mises (0.509) and thickness of the shell (0.644).

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 right (less value) to the left (high value) along F1 from 1 to 8. It corresponds to the influence of foam elastic limit.

Figure 6: PCA correlation circle of the 8 variables (a), Factorial plane (b).

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 intracranial stresses higher than the tolerance limits. It's the reason why we used a parametric study to optimise the helmet performance against these biomechanical criteria. The PCA method used to analyse results,
led to the conclusion that the foam elastic limit has an important influence on biomechanical response. Moreover, this study permitted us to propose an optimisation in terms of pressure with the configurations S1, S3, S5 and in terms of Von Mises stress with S5 (figure 7-a). Nevertheless, the HIC remains the single normative parameter used to homologate a helmet in terms of protection against impacts. So the same parametric study was performed with a headform's standard. Histogram in figure 7-b shows the maximum acceleration of the centre of mass with this headform. The analysis of these results shows that the best optimisation is the configuration S3.

So, an optimisation with 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.

![Figure 7](image)

(a) Maximum Von Mises stresses in frontal impact with ULP model.

(b) Maximum acceleration of the centre of mass in frontal impact with headform’s standard model.

**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 267 kPa in the region of the impact site and a maximum tension of -101 kPa at the opposite point. Moreover Von Mises stress which is a good indicator of concussion reached a maximum value of 38.5 kPa. These values are higher than tolerance limits proposed by Willinger [6] (20 kPa). 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 have gives the same results. Moreover, this study led then to the conclusion that the foam elastic limit has the most important influence on biomechanical head’s response.
Acknowledgement

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References