THREE YEARS OLD CHILD HEAD-NECK FINITE ELEMENT MODELING. SIMULATION OF THE INTERACTION WITH AIRBAG IN FRONTAL AND SIDE IMPACT.

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Paper Number : 09-0392

ABSTRACT

This study proposes to assess the interaction between the 3 years old child Head-Neck system and a typical airbag, a protective system frequently used in the automotive field. Two separated models (Head and Neck) developed at the Strasbourg University (UDS) were coupled in order to estimate the injury risk during this type of impact. The first model developed is a three years old child Finite Element neck Model (FEM) based on a realistic geometry (Meyer et al. 2008). This FEM was validated in four directions against an original method based on scaling method (Irwin et al. 1997). The second FEM is a 3 years old Head FE model published by Roth et al. in 2008. This model proposed an injury criterion in terms of Von Mises stress in the brain for moderate neurological injuries. After a coupling of these two FE models two impacts a frontal and lateral impact configuration is simulated. These impacts consisted of an airbag deployment at different gaps in order to calculate and estimate child brain injury risks.

INTRODUCTION

The growing demand for greater mobility in Europe has made individual transportation an essential and even inevitable feature of modern leaving. Children are more and more often conveyed in cars or other modes of road transportations. With this increased travels, the risk for children, of becoming involved in an accident as occupant has consequently increased. Based on the above accident data, it is obvious that in spite of the significant improvements in recent years in vehicle safety, the current number of deaths and casualties added to the social and economic costs is still unacceptable. Fatalities and injuries, especially to children, shall be reduced by all the available ways: public regulation, prevention/education of road users, road infrastructure, compatibility between vehicles, active, passive and tertiary safety devices.

As regards children, it is very difficult to obtain figures for fatalities or severely injured children in the 27 European Countries, but if we consider the EU 15 countries, where the use of child restraint is mandatory since a long time, approximately 600 children are killed in cars on the European roads and 80 000 are injured (data source: IRTAD).If there has been a hudge effort on human adult FE modeling only very few attempts exist as long as children are concerned.

Due to ethical reasons, there is paucity in experimental data concerning the child's head and neck characterization. As a consequence, there is a considerable difficulty for the validation of children FE models. For the neck validation one solution is to use the Scaling method's established by Irwin's and Mertz 1997. This method permits to calculate a theorical experimental corridor based on on the adult experimental data, in terms of displacement and acceleration. The mechanical properties such as the mass density of the cervical vertebrae, the rigidity both for the intervertebral discs and the ligament are calculated with this scaling method.

One way to investigate child Head injury criteria using numerical models is to simulate real world head trauma. Well documented accidents can help to understand child injuries in comparing numerical mechanical parameters with what really happened, distinguishing biofidelic behavior of a child numerical head and the ability to have an injury predicting tool. Indeed, even if the biofidelic behavior of child models cannot be checked, based on classical experimental versus numerical validation process, investigations of child injury mechanisms can be performed by developing an injury predicting tool, studying numerical

simulation of a large number of real accidents and to correlate mechanical parameters outputs with observed injuries. In the present work these previous published Head and Neck models are coupled to a simplified thorax in order to investigate child Head-Neck response under frontal and lateral airbag deployment as a function of initial distance between airbag and Head.

MATERIALS AND METHODS

Three years Old Child's Neck FE model

The neck model used in the present study has been previously published (Meyer et al. 2008) and will therefore be presented very shortly.

A three year old male child head and neck was scanned (figure 1) in order to base this study on a realistic human geometry, and to integrate the detailed vertebrae anatomy. The surfaces were reconstructed, so that the cervical vertebrae could be completely meshed.

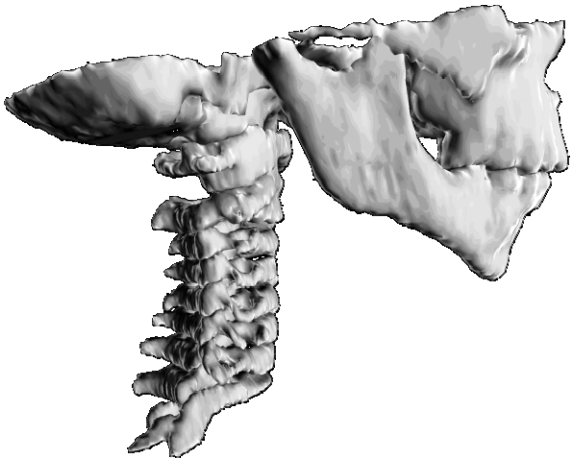

Figure 1. Reconstruction of the cervical spine based on scanner section.

For the cervical vertebrae, shell elements offer the possibility of strictly respecting the anatomical surface. The upper and the lower ligamentary system were reproduced with springs elements and the intervertebral discs with bricks elements (3 layers).

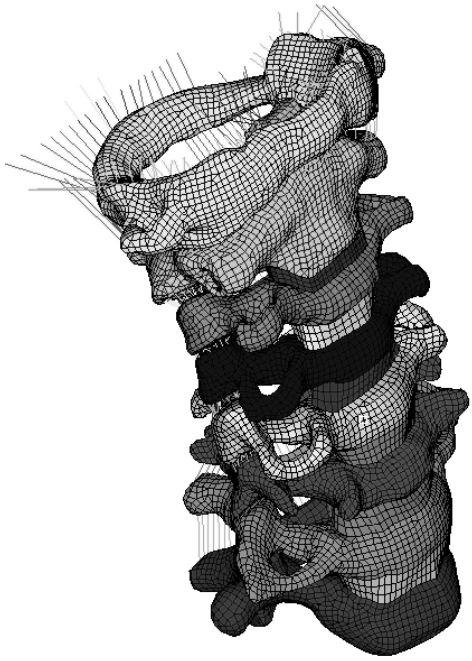

Figure 2. Surface meshing of the cervical column (C1-T1) with its ligamentary system. (See also Meyer et al. 2008 for further details).

Finally the model of the three year old child neck contains a total of 2 826 brick elements and 44 758 shell elements and 712 springs.

Finite element models of adult neck are typically validated against experimental data carried out by the N.B.D.L (Van der Horst M.J. 2002, Meyer et al. 2004), with frontal, oblique, lateral impacts (Ewing et al. 1968, Ewing et al. 1977). Unfortunately, for ethical reasons, it is not possible to perform similar tests on children so no data exist in the literature for dynamic validation of a paediatric neck model. In the present study, inputs for the three-year-old-child model correspond to those used in the NBDL tests (Frontal, Lateral, Oblique) but outputs, i.e., head accelerations and displacements ' corridors , are scaled down in accordance with Irwin's method (1997). An example of the frontal validation is illustrated in figure 3 where the superimposition of experimental response corridors obtained with the scaling method, and numerical curves obtained with the finite element model of the child neck is reported.

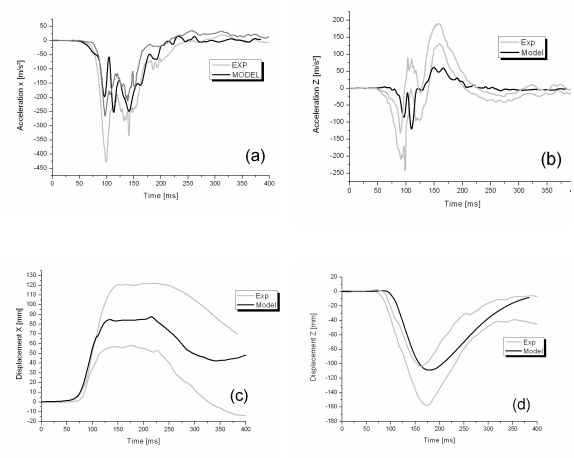

Figure 3. Results under frontal impact: X-axis (a), Z-axis (b) linear head acceleration, X-axis (c), Z-axis (d) head displacement and kinematic response of the whole head/neck system (e).

Child's head model and injury criteria.

The head model wich will be coupled to the neck was published by Roth et al. in 2008. Hereafter a short presentation is re-called.

As illustrate in figure 4, the developed three years old head model takes into account the main anatomical features of a three year old child. It includes the scalp, the skull, the sutures (sagittal, coronal, lambdoid), the face, the cerebro spinal fluid (CSF), the falx, the tentorium and the brain. Finally, the whole model of the three year old child head (a) contains a total of 23000 brick elements and 3500 shell elements.

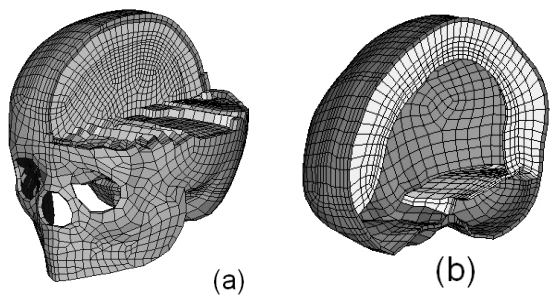

Figure 4. Meshing description of the detailed three year old child head finite element model (a) Cross section of the HEAD FEM (b) Membranes Falx & tentorium. (See also Roth et al. 2008)

In order to investigate child injury criteria with the finite element model, 25 real world accidents involving child aged from 2.5 to 3.5 year old were collected. These accidents are free fall from different heights and are simulated with the 3YOC head in order to extract the best mechanical parameter able to predict injury occurrence. From medical files, several data are available: gender, age, height of fall, type of impacted surface, type of injury. Injuries are classified into two categories: Moderate neurological injuries (2 in the AIS scale, unconsciousness limited to few hours after impact) and severe neurological injuries (>3 in the AIS scale, with a >24hours coma). Among these 25 cases, 15 accidents induced with no neurological injuries, 8 lead to moderate neurological injuries, and 2 to severe neurological injuries.

The determination of the head injury risk curves for specific injury mechanisms is based on a correlation study between the values of the proposed candidate criteria and the neurological lesions occurrences. Maximum values of mechanical parameters are used to build a histogram. The accident cases are finally sorted according to the injury classification, *i.e.* moderate or absence of neurological injuries. When the injury predictor candidate is adequate, a distinction is visible between the low values of the uninjured cases and the high values of the computed for the injured cases. This threshold can accurately be calculated since it is the value leading to a 50% risk an injury. For the statistical approach, the modified maximum likelihood method is chosen. It is a logistic regression method developed and described by Nakahira et al. (2000). The quality of the regression is thereby given by the negative estimator *EB* which should be as close to zero as possible. For each of these parameters: Von Mises stress, peak linear acceleration , maximum pressure, peak angular acceleration and HIC value, EB were calculated in order to identify the most relevant injury parameter (figure 5).

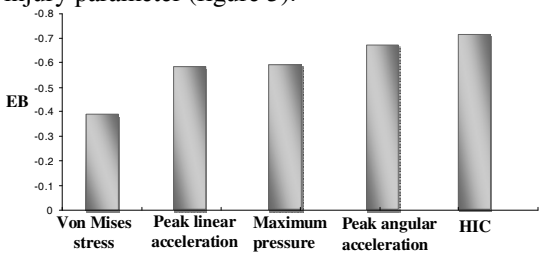

Figure 5. EB regression parameters for several candidates for moderate neurological injury criterion.

As a result of numerical reconstructions of real world cases, shear distribution in term of stress appear to be an interesting predicting candidate for neurological lesions. These parameters had also been used for prediction of neurological injuries in adult head finite element model by Deck et *al*. (2008). As a conclusion a Von Mises brain shearing stress of 48 Kpa will be retained for neurological injuries.

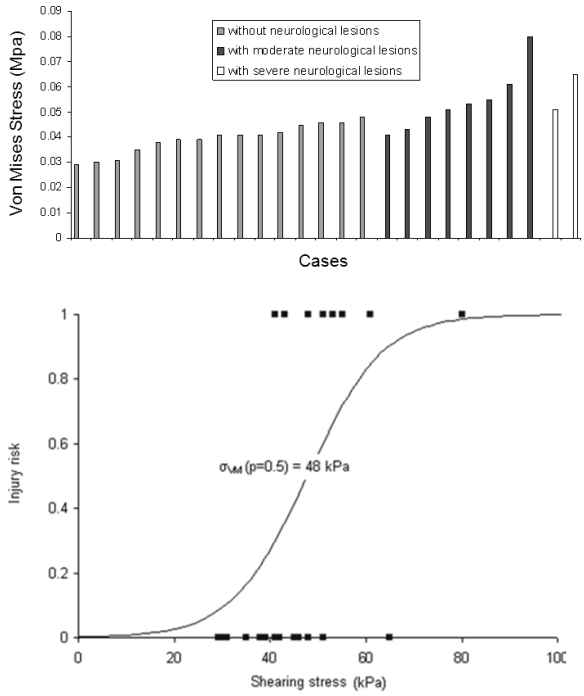

Figure 6. Histogram illustrating the correlation between the best mechanical parameter candidate (brain Von Mises stress) computed with 3 YOC FE model and corresponding injury risk curves.

Coupled three years child Head-Neck-Thorax model under impact.

In the framework of the present study the Neck FE model was coupled to the Head FE model. The connection between the Head and the Neck is made through the ligamentary system. The existing upper ligaments were connected to the Head FEM and the contact between the atlas and the occiput was reproduced with a sliding interface. The objective of the coupling between the Neck and the Thorax was to take into account the mass and inertia effect of the thorax in case of an airbag impact. The geometry was taken from an adult thorax and scaled down in accordance with Irwin's method. As the thorax has only an inertial effect in the context of this study a very simplified thorax model is proposed. The whole three years old coupled model is illustrated in figure 7.

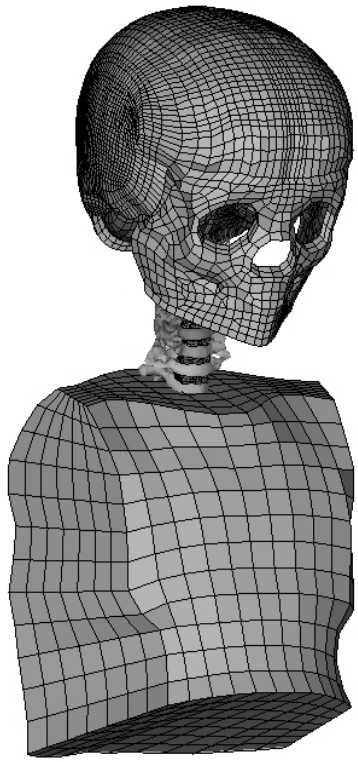

Figure 7. Coupling of the three years old Head-Neck FEM to a simplified thorax model.

In order to provide realistic inertia, the thorax was meshed with bricks elements and the density was adjusted in order to have a mass of 6.61 Kg (Irwin et al 1997) and an inertia of Ixx=3.15*10^7g.mm², Iyy=2.73*10^7g.mm², Izz=2.36*10^7g.mm². Finally T1 vertebra was associated to the thorax as a key element for the coupling of this segment to the head-neck complex.

In order to simulate child airbag interaction during airbag deployment two impacts conditions are suggested, a frontal and a lateral one. For each case the child is supposed to be seated statically without seat back, i.e. without any restrain of his thorax and with no initial velocity. For the frontal impact five distances between the chin and the airbag are proposed i.e. 6, 8, 10, 12 and 13.5 cm. For the lateral impact five distances between ear and airbag are suggested as well, being 6, 8, 10, 12 and 13.5 cm. For these ten impacts the injury parameters at head will be computed in order to express the injury risk for each case. Figure 10 presents more details relatively to the initial conditions of this impact simulation for the frontal configuration, as the

airbag center of mass is set at 4.2 cm below the child head center of mass.

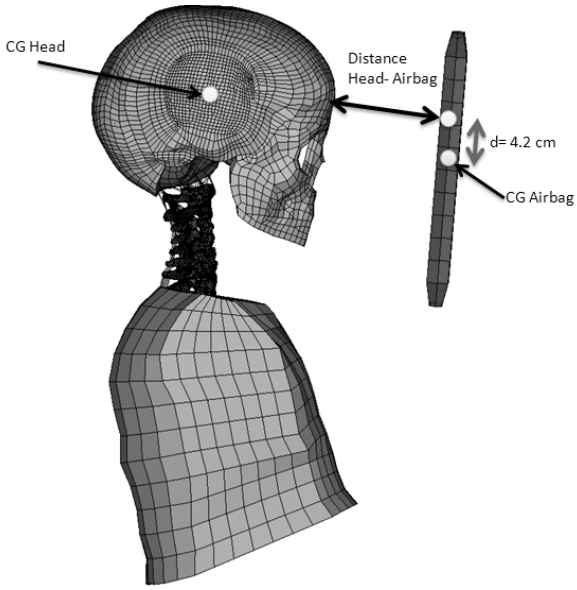

Figure 8. Child under airbag deployment under frontal configuration.

RESULTS

In this section results are reported separately for the frontal airbag deployments as the lateral ones.

Frontal Impact

Figure 9 represents the maximum of forces calculated per head/airbag distances. It can be observed that there is no significant correlation between head/airbag distance and calculated maximum interaction force.

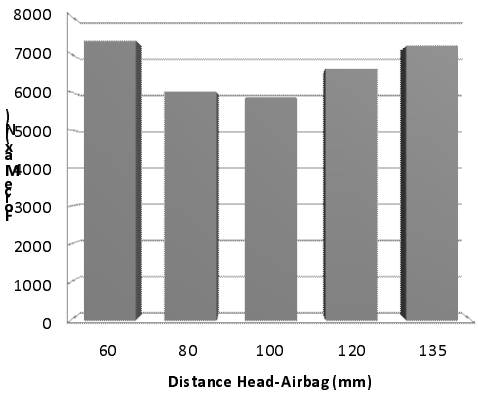

Figure 9. Maximal interaction force calculated for the five head-airbag distances (d=6 cm; d=8 cm; d=10 cm; d=12 cm; d=13,5 cm).

For the five airbag distances simulated the intra-cerebral Von Mises stress calculated (location and time evolution curves) are reported in figure 10. The locations of these maxima are similar in the five cases (at the vertex area). Five bricks were considered to obtain a mean value of the time evolution of Von Mises stress at these maxima location.

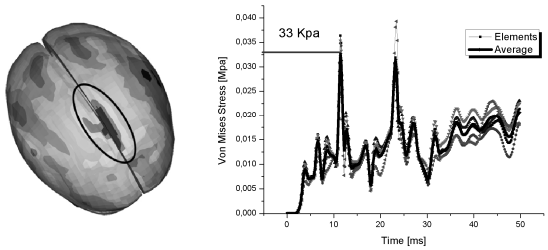

Figure 10. Illustration of the intracerebral Von Mises stress computed for the distance 100 mm (location of these maxima on left and time evolution on right) in frontal impact configuration.

Maxima of Von Mises stress are obtained after the total airbag deployment (after 10ms) and all results are summarized in figure 11.

For the head-airbag distance of about 60mm, an intracerebral Von Mises stress of 59kPa wich is higher than the tolerance limits calculated for a 50% risk of moderate neurological injuries (48kPa) is obtained. It is interesting to observe that the intra-cerebram Von Mises stresses are much more correlated with the head-airbag distance then the interaction force is.

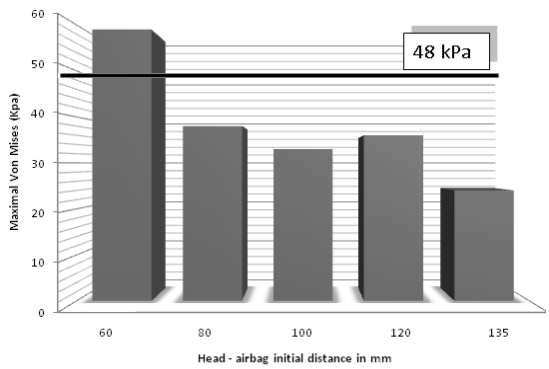

Figure 11. Maxima of intracerebral Von Mises stress computed within for the five head/airbag distances under frontal impact configuration.

Lateral impact

The conditions of the lateral airbag impact are similar to the frontal impact i.e. a free thorax boundary conditions boundary conditions. An overall view of the kinematics under this lateral airbag deployment (d=100 mm) is illustrated in figure 12.

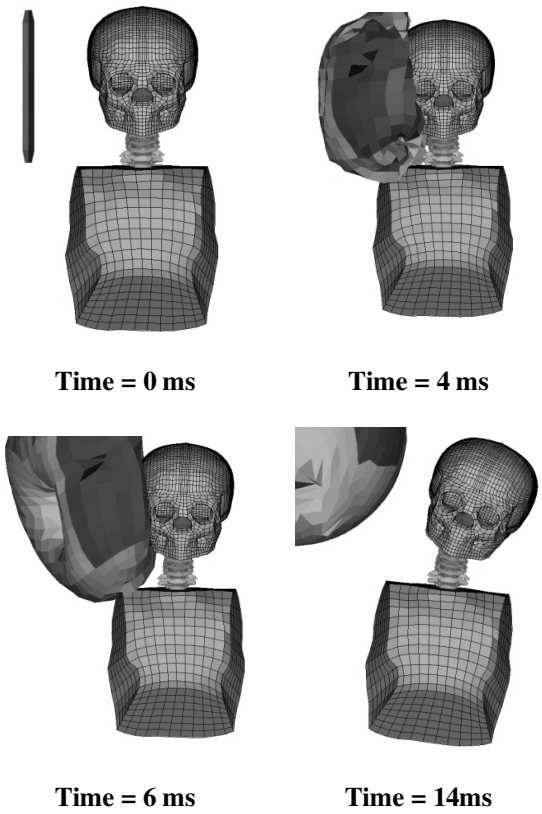

Figure 12. Overall kinematics of the Head FEM under lateral airbag impact.

As for frontal impact configuration no significant correlation between head/airbag distance and maximum interaction force calculated is observed as illustrated in figure 13.

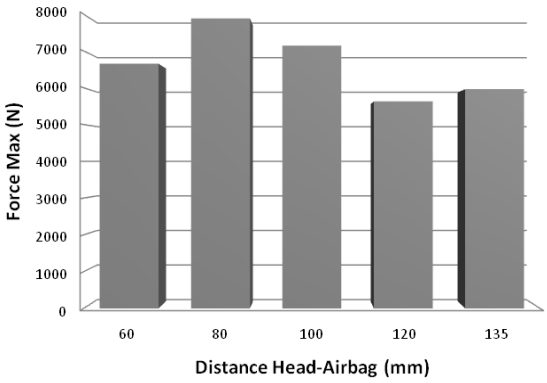

Figure 13. Maximal interaction force calculated for the five head-airbag distances (d=60mm; d=80mm; d=100mm; d=120mm; d=135 mm).

Figure 14 shows the maximum intracerebral Von Mises stress computed (location and time evolution curves) for the five airbag distances in lateral impact configuration. Location of these maxima is similar in the five cases (at the opposite side to the impacted area). Except for the 135 mm distance all simulated cases conduced to the same conclusion i.e. that it exits a risk of moderate neurological injury due to the fact that the intracerebral Von Mises stress calculated exceed tolerance limit fixed to 48kPa.

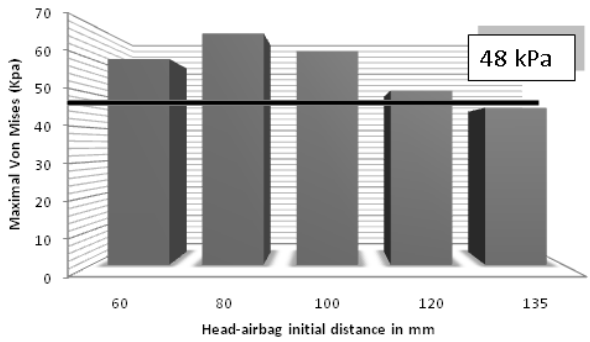

Figure 14 Maxima of intracerebral Von Mises stress computed within the brain for the five head/airbag under lateral impact configuration.

DISCUSSIONS

After the development of a three years old child Head-neck FE model and its use under airbag deployment it's important to define the limitations of this study. A number of limitation exist at biomechanical modeling level were clearly improvement are needed in the future especially as long as neck injury criteria are concerned. The boundary conditions applied aren't the same as in accident condition as the simulations don't take into account the initial velocity of the whole body, the effect of the seatbelt and the initial position influence kinematic's and the injury risk. The main originality of the proposed head-neck-thorax model is to consider a realistic and detailed geometry of the cervical spine and a FE head model who proposed tolerance limits for moderate neurological lesion. It's therefore a step towards numerical tools for the assessment of the child head and neck injury risk under airbag deployment.

CONCLUSIONS

The objective of this study was to assess the injury risk of the child head-neck system under airbag deployment for frontal and lateral configuration.

The presented work is based on a head and neck model of the three year old child developed in earlier studies as well as first head injury criteria. Proposed then is the coupling of the head-neck system to a simplified thorax model in order to assess head injury risk under frontal and lateral airbag deployment. A parametric study on head-airbag distance is finally conduced with following main conclusions.

For the frontal airbag deployment it's shown that there is a low correlation between initial distance and the interaction force. A head injury risk appears only if the initial distance between the airbag and the head is less than 80 mm.

Concerning the lateral airbag deployment configuration a similar conclusion can be made as for the frontal impact i.e. no correlation between the interaction force and the injury risk. However the brain injury risk appears to be much higher as for the frontal impact. For all distances a brain injury risk over 50% has been computed.

Even if a number of limitations exist in this child response simulation under impact first steps have been provided towards numerical tools designed for the assessment of child head-neck injury risk under deployment.

AKNOLEDGMENT

The authors wish to thank TRW for their support.

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