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# Ongoing efforts on scalable child models, associated tools and procedures in the PIPER project

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## Abstract

In Europe, different Child Restraint Systems architectures can be used to restrain older children. The specificities of the child biomechanical response during the interaction with these systems are however largely unknown and much remains to be done for the safety of children in cars. Human body models (HBMs) have the potential to describe the diversity of children both in terms of anthropometry and biomechanical development. This paper describes some of current challenges and ongoing progresses towards the development of a family of HBMs continuously scalable in the range 1.5 to 6 years-old. Significant updates of the models are ongoing to prepare them for various safety applications. These include a better description of the deformability of key anatomical structures to help improve the response in frontal impact and side impact. The updates also open the possibility of future investigations of injury mechanisms.

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## 1. Introduction

In Europe, different Child Restraint Systems (CRS) architectures can be used to restrain above 1 year. They include integral and non-integral CRS, with harness, shield, or 3 point belts. The restraint by a 3 point belt typically requires the use of a booster seat or a booster cushion to help position and guide the belt. The child occupant can therefore be subjected to interactions with restraint systems that are different and more diverse than the adult.

Currently, the dynamic performance of CRS is typically verified with crash test dummies. For children, Hybrid III, CRABI, P series (regulation R44 in Europe) and Q series (regulation R129 in Europe) are available, covering a range from 0 to 10 years-old. In the relative absence of relevant Post Mortem Human Subjects (PMHS) data, the performance requirements for the dummies have mainly been set using simplified scaling assumptions (e.g. Irwin et al., 2002). Requirements have been limited regarding kinematic behavior and the specificities of the loading configurations with different architectures are not accounted for. Because of simplifications and limitations arising from the need of a physical implementation, dummies are typically designed to match the requirements but they do not attempt to represent accurately the human anatomy and its evolution with age, even at the regional level. As a consequence, their performance in configurations that would differ from the requirements can be limited by design choices. For example, despite originally aiming for omnidirectionality, existing Q dummies such as the Q3 have shown limitations in side impact which have triggered the development of specific versions (e.g. Q3s). Also for Q dummies, the diagonal belt interaction and submarining behaviors, for which there are no specifications, have been extensively

questioned (e.g. Beillas et al., 2013, 2014). Overall, the variety of loading conditions encountered in child safety could be associated with an uneven performance of the assessment tools that could make difficult an objective evaluation of different CRS architectures.

Beyond loading configurations, child postural preferences have been found to differ from dummy seating (e.g. slouched posture, Reed et al., 2006) in ways that could affect the impact response (e.g. submarining risk). Naturalistic studies have also shown that children use a variability of postures on the road (Jakobsson et al., 2011). Such postural variations are typically difficult to simulate with existing dummies. Also, in the new regulation R129, CRS manufacturers are able to specify any child stature range for the CRS but they must verify that the CRS is able to accommodate not only an average child of that stature but also children of with other anthropometric characteristics (e.g. 95<sup>th</sup> percentile shoulder width for this stature). However, the dummies used to verify the dynamic performance of the CRS are only available in few statures/ages, corresponding to 50<sup>th</sup> percentiles of ages for all dimensions (e.g. the Q3 represents an average 3 years-old).

Human body models (HBMs), in particular Finite Element (FE) models, can more accurately represent the anatomically complex geometry and topology of the human body. They can also be relatively easily implemented in multiple dimensions. HBMs have therefore the potential to better describe the diversity of children both in terms of anthropometry and biomechanical development. Because of this better representation of the human anatomy, they could be more robust to the diversity of loading configurations. Also, such model could be used to estimate human tolerances based on accident reconstructions as it has been attempted with child dummies in the past (e.g. Johannsen et al., 2012). In the future, HBMs use could lead to a better assessment of the kinematic response, loading and potentially injury risk for restrained children.

However, while great effort has been made to generate and validate detailed models for adults (such as the commercially available models from the Global Human Body Model Consortium, and the THUMS from Toyota), fewer child HBMs have been presented in previous literature. The development of child human body models faces numerous challenges. For example, there is a need to represent at least some of the biomechanical changes due to the continuous growth and development that take place before birth and continue until the adolescence. Tools and procedures are needed to help represent these continuous changes into models. Scaling and positioning tools are also necessary to enable simulations with HBMs of different sizes for a given age, or with HBMs in different positions.

This paper describes some of current challenges and ongoing progresses towards the development of a family of human body models continuously scalable in the range 1.5 to 6 years-old (and perhaps beyond). Associated tools for personalization and positioning are also briefly described.

As a baseline, the study uses a model family (LBMC child models) that was previously developed as part of the CASPER and Proetech projects. The models were already used to study the submarining behavior (Beillas et al., 2013), the interactions between thorax and diagonal belt (Beillas et al., 2014a) and the abdominal and thoracic loading associated with shield CRS (Beillas and Soni, 2014b). Based on the analysis of the model performance and limitations, the objectives of the current effort are to:

- Update the head and neck regions of the model, due to their importance for kinematics and injury response (Durbin et al. 2003, Arbogast et al. 2004, Howard et al. 2004).
- Extend the validation, in particular to include side impact. This also requires updating some regions of the model (e.g. shoulder, pelvis) to make them deformable.
- Account for the effects of growth by implementing cartilaginous tissues for the head, spine and shoulder. The objective is to capture age-specific anatomical and functional changes.
- Prepare the model for scaling and positioning using the PIPER software framework.

This work was conducted as part of the PIPER EC funded project.

## 2. Methods

### *2.1 Description of the baseline models and their validation (LBMC Models)*

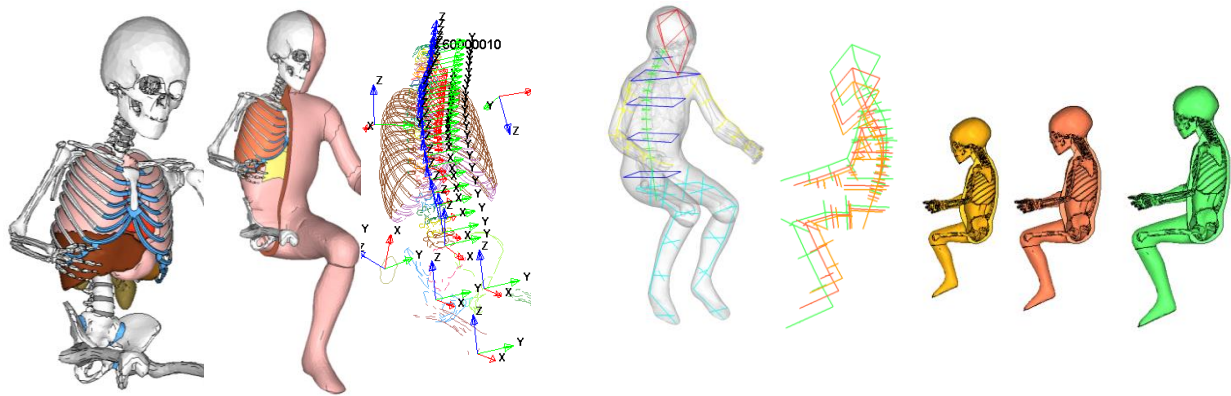
The models used for the current study were previously developed at the University Lyon 1-Ifsttar *Laboratoire de Biomécanique et Mécanique des Chocs* (LBMC). The baseline LBMC child model (Figure 1a) aimed to simulate a 6 years old (y.o.) child and to study its interactions with different types of restraint systems. Its geometry was based on CT scans from patients. As CT scans were collected in a supine position, the spine, pelvis and solid abdominal organs positions were adjusted using postural MRI data previously collected on adults (Beillas et al., 2009). Its external envelope was built by combining the skin segmented in the supine position (where relevant) with the average envelope developed by the University of Michigan Transportation and Research Institute (UMTRI) as part of the OCATD study (Reed et al., 2001). The OCATD position was also used to define the model posture for the upper and lower extremities. The ribcage (ribs, sternum and costal cartilage) was modeled as deformable while other skeletal structures were modeled as rigid bodies and articulated using 6 d.o.f. joints. Internal thoraco-abdominal structures were either modeled using tetrahedral elements (heart, liver, kidneys, spleen) or single pressure point monitored volumes (lungs, stomach, intestines). The model is covered with a solid flesh component that is continuous from the head to the distal femoral head. The lower legs and feet were modeled as rigid bodies. The model geometry was normalized using anthropometric dimensions predicted by the GEBOD software (Cheng, 1994) for an average 6 y.o. child and nonlinear scaling techniques based on Kriging interpolation. The model was also scaled to match the dimensions of 3 and 1.5 y.o. children derived from GEBOD. Illustrations of the kriging procedure and the resulting family of models are provided in Figure 1b. The models include 204 parts for 332k elements (including 47k rigid elements) and approximately 70k nodes. Material properties were adapted from previous literature studies.

The model response was compared to results from tests on PMHS or volunteers, or from biofidelity targets defined for dummies by scaling (e.g. abdomen specification in EECV 2008). While the validation work mostly focused on regional responses for the thorax and abdomen (where the interactions with the CRS occur), it also included global kinematics results from tests using PMHS restrained by a harness (Wismans et al., 1979) or a shield (Kallieris et al., 1976), and volunteers restrained by a belt (Arbogast et al, 2009). The model was found to be in overall agreement with the literature results, with examples of responses provided in Figure 2.

While these results were encouraging and applications involving different types of CRS were performed (Beillas et al., 2013, 2014a, 2014b), the LBMC models suffered from limitations including:

- The modelling effort was mostly focused on the thoraco-abdominal region, with other regions being very simplified (head and neck in particular).
- Most skeletal structures were modeled as rigid and some structures were too simplified for injury prediction (e.g. head, neck).
- The response was only verified in frontal loading, and remains to be checked in side impact. For the side impact scenario, the modelling assumption of rigid bony components (in particular the pelvis and the shoulder) may be problematic.
- The model did not account for age-dependency besides dimensional scaling. The ossification process (replacement of cartilage by bone) and the changes of material properties were not implemented between different models.
- Only average models in the reference posture were used, and the generation of different postures may be challenging.

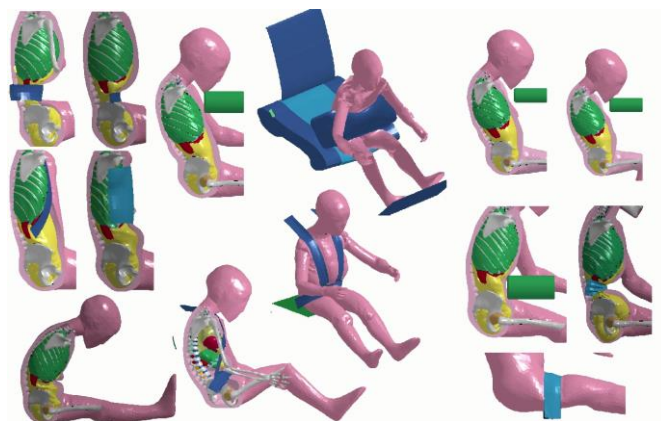
These limitations are the focus of the current study.



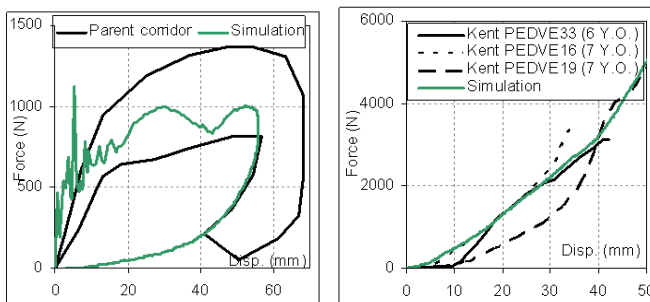
a) Baseline (6 y.o.): Mesh of the thorax, whole body and joints articulating the skeletal structures

b) Points used to drive the kriging and scaled models corresponding to 1.5, 3 and 6 y.o. child dimensions

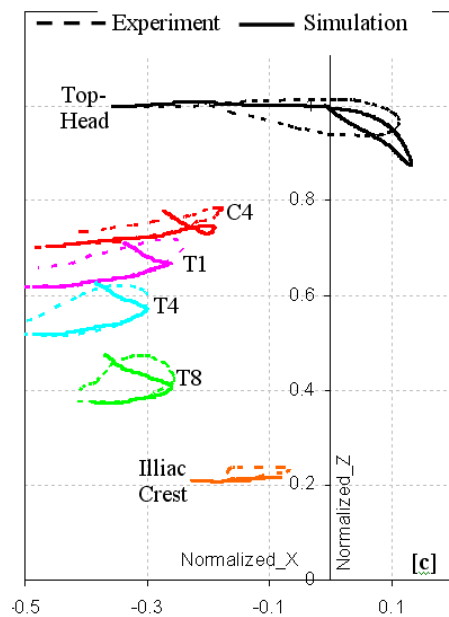
Figure 1: Overview of the LBMC models



a) Overview of test setups including regional loading applied to the thorax or abdomen and sled configurations



c) Responses for the Ouyang et al. (2006) thorax impact and for the Kent et al. (2009, 2011) diagonal belt loading



b) Trajectories of markers in the Arbogast et al. (2009) sled volunteer setup (low acceleration)

Figure 2: LBMC model responses: examples of a few setups

## *2.2 Scaling and positioning – Approach and PIPER tools*

As with the LBMC models, the ongoing developments are concurrent on three main ages (1.5, 3 and 6 y.o.). The models are transformed into one another using nonlinear scaling driven by control points corresponding to anthropometric dimensions. The scaling approach is automated and models corresponding to any age target can be generated within this range. Extrapolation outside this range is also considered (1-10 y.o.).

With the introduction of local features to describe the evolution with age and the postural changes, there is a need to update and refine the geometrical transformations. In this study, the approach selected was to conserve a control point based approach along with Kriging transformation and to refine targets to account for significant local features.

More generally, there is a need to provide child model users with methods and tools for specific safety applications, such as scaling to match population characteristics (e.g. in relation to R129) or specific postures (e.g. normal positioning in a CRS or positions from naturalistic driving). The development of such tools is the primary goal of the PIPER EC funded project. In the PIPER tools, scaling methodologies are currently based on Kriging interpolation and interactive positioning is based on lightweight physics simulation approaches (more details can be found at [www.piper-project.eu](http://www.piper-project.eu)). The tools will be made available as Open Source at the end of the project. For applications with the child models, it seems important to ensure that the model will be compatible with the software, and that the software features will allow relevant applications.

For the current study, these tools were tested with the child model as a preliminary evaluation to investigate the scaling and positioning abilities of the model. Conversely, attention is being paid in the model development process not to include features that would make scaling and positioning impossible.

## *2.3 Need to model the growth*

Besides global geometrical changes (size of different body regions as a proportion of total body height, body center of gravity shifts), child growth is also associated with structural changes (e.g. replacement of growth cartilage by bone) and material properties changes (e.g. variation of water/protein content in a tissue). Significant age-related differences in tissue mechanical response between adults and children are found in many studies (Crandall et al. 2013), with effects that vary with the type of tissue and body region. These anatomical attributes may be important to have a thorough understanding of the biomechanics of children in a range of loading situations and may have implications for vehicle safety. In the current study, ossification processes and age-specific material properties are considered for three regions of interest as detailed below: 1) the head 2) the neck and 3) the shoulder.

As potentially important in injury analysis, in the head, gray matter (GM) and white matter (WM) were modelled separately. According to Matsuzawa et al. (2001), the volume ratio WM/GM is strongly age-dependent. From 3D MRI scans, a fast WM volume increase is visible in the first 2 years of life while all major white matter fiber tracts could be identified by age 3 years. This development is ordered (central to peripheral, inferior to superior and posterior to anterior, Barkovich et al. 1998). For the 3 y.o. model, the WM elements are identified on an anatomical ATLAS (NIHPD Objective Pediatric Atlas, Fonov et al. 2009) giving a WM/GM ratio of 0.44 (Matsuzawa et al. 2001). For the 1.5 and 6 y.o. child models, the ratio is adjusted to be 0.37 and 0.5 respectively (Matsuzawa et al. 2001). In the current study, an algorithm generating a transition between gray matter to white matter elements was implemented to mimic the growth of white matter tracks and adjust WM/GM ratio with age.

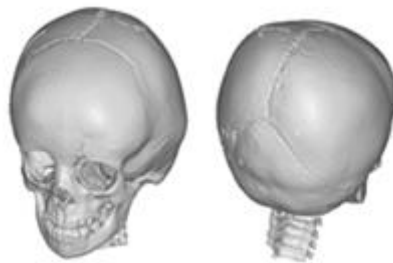
In the skull, structural changes are mainly due to the closure of fibrous joints called sutures (Scheuer et al. 2004). Softer structures in the skull allow broader movements (deformation), making the cranial bone partially compressible and capable to slightly change shape. In the current study,



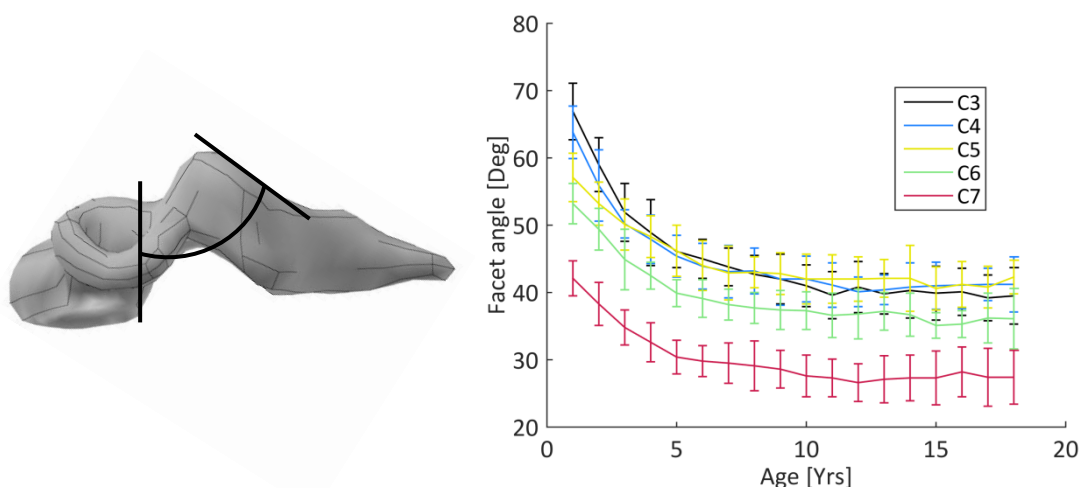
sutures were modeled as cartilaginous tissue. They were segmented manually on the CT-scans of a 3 y.o. and 1.5 y.o. child in the 3DSlicer software package (Figure 3). The scaled 6 y.o. model does not contain fibrous sutures (they were considered almost closed and negligible).

In the vertebrae of the cervical spine, the replacement of growth cartilage by bone occurs within each individual vertebra, changing material compositions as well as the local topology (Yoganandan et al., 2002). These topological changes, together with the change in facet joint angle (Kasai et al., 1996), illustrated in Figure 4, have previously been shown to be important for local spine kinematics (Kumaresan et al., 2000). They were taken into account in the current study by adding local control points used in the scaling from the baseline model.

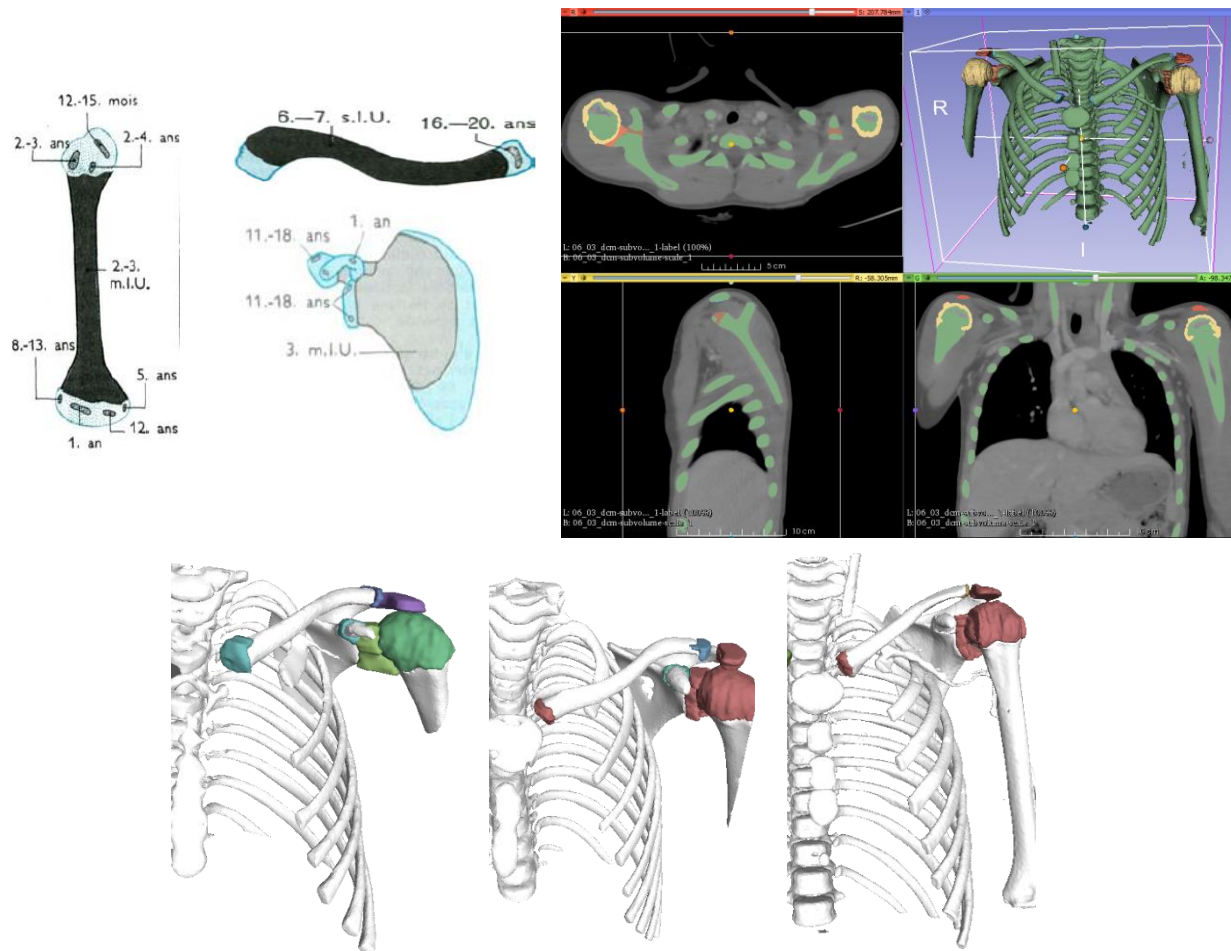
For the shoulder, one important change with age seems to be presence of growth cartilage (e.g. Kahle (1978) which could affect the stiffness response in side impact. For the current study, the cartilage of shoulder bones was segmented manually on the CT-scan of a 6 y.o. child in the 3DSlicer software package (Figure 5). The surfaces extracted by segmentation were used as a basis for the mesh development of the shoulder model. As the upper extremity positions derived from the OCATD study did not seem the most relevant for crash testing in narrow CRS, the shoulder and upper extremities were also re-positioned along the body.



**Figure 3: Identification of the cartilaginous structures of the skull: the segmented surfaces were obtained by segmentation using a high threshold to enhance the location of the sutures.**



**Figure 4: Change in facet joint angle as a function of age for the C3 to C7 vertebra (Kasai et al., 1996) with the facet joint angle illustrated in the left image.**



**Figure 5: Shoulder and ossification process. Top Left: illustration of the ossification process for the humerus, clavicle and scapula. Illustrations from Kahle (1978). Top Right: manual segmentation of the humerus and scapula of a 6 y.o. CT-scan to serve as a basis for the model geometry. Bottom: segmentations of 1.5, 3 and 6 y.o. CT scans (from left to right).**

### 3. Results

#### 3.1 Scaling algorithms and positioning tools

In order to import and interpret LBMC model in the PIPER framework tool (currently under development), metadata files were developed, which contain information about elements, part and joint definition for the model. The results from the scaling protocol used for the LBMC models (custom Scilab Scripts) were compared to the Kriging module developed as part of the PIPER tool. Results are shown in Figure 6a. The element quality (scaled Jacobian) obtained with the new tool (72-18) was compared to the element quality of the model obtained by the older approach (Scilab script, 72-18-ref) and the original model. Differences between models are marginal, suggesting that the new PIPER tool can be used to replace the former scripts in future work. In particular, kriging target points need to be updated to better capture local geometrical changes of important features (as it will be illustrated in for the neck).

For positioning, illustrations are provided in Figure 6b. The process can be done interactively using the lightweight physics positioning module. Work is still ongoing on the element quality of the transformed models.



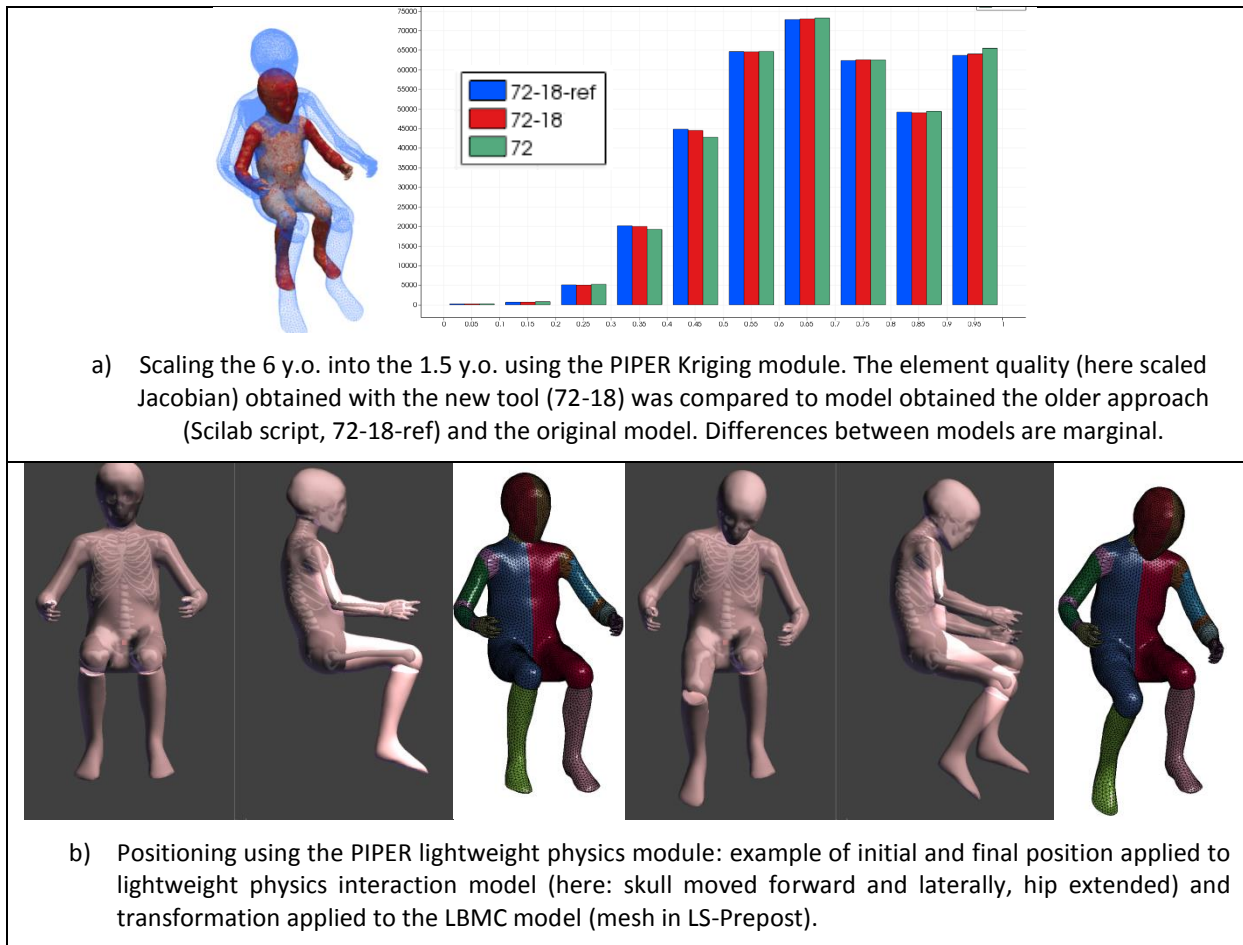


Figure 6: PIPER tools to position and personalize human body models: preliminary applications with the LBMC model

### 3.2 Model improvements and regional developments

For the time being, model improvements include mainly the development of head, neck and shoulder models as well as the update of the flesh.

#### 3.2.1 Head Modeling

A 3 y.o. child FE model of the head, continually scalable in the range 1.5 to 6 y.o., was developed in LS-DYNA code. The baseline model (3 y.o.) was based on computer tomography of a single subject and includes the scalp, the skull, the cerebrum, the cerebellum, the meninges, the falx, the tentorium and the cerebrospinal fluid (Figure 7). A Kriging scaling procedure was adopted to scale the model in order to obtain dimensions close to the average of the 3 y.o. group (Cheng 1994). The multi-block approach available in ANSYS ICEM 15.0 (ANSYS, Canonsburg, PA, USA) was used to mesh solid and shell elements. The inner and the outer layer compact bone of the skull were modeled by shell elements; the diploe layer of the cranium, the cerebrum, the cerebellum and the cerebrospinal fluid were represented by solid elements. Finally, the dura mater, the tentorium and the falx were modeled by membrane elements. A typical spatial resolution of 3 mm was chosen to capture fundamental anatomical structures but, at the same time, having acceptable computational cost for the simulations.

The new 3 y.o. head model was updated with the associated skin (Park and Reed, 2015). The 1.5 y.o. and 6 y.o. models were then obtained by scaling using the PIPER Kriging tool (Figure 8). Head length, head breadth and head to chin height were specified as targets, according to data from GEBOD database (Cheng et al. 1994). These targets were chosen to account for variation in head segments proportions with age.

Sutures of the skull were modeled as cartilaginous tissue. The surfaces extracted by segmentation were overlaid with the model and used to generate a transition of elements from the bony skull to fibrous tissue (Figure 9). In future studies, the closure mechanism of the suture will be made continuous with age by interpolation.

Separate representations of gray and white matter were also implemented. For the algorithm, the pattern described by Barkovich et al. (1998) is followed to generate the element transition (Figure 10). WM element neighbors are identified by 8-connectivity and the target ratio is adjusted with growth priorities (Figure 10b). Resulting models for the three ages are shown in Figure 9c. The process can be applied using age as a continuous variable.

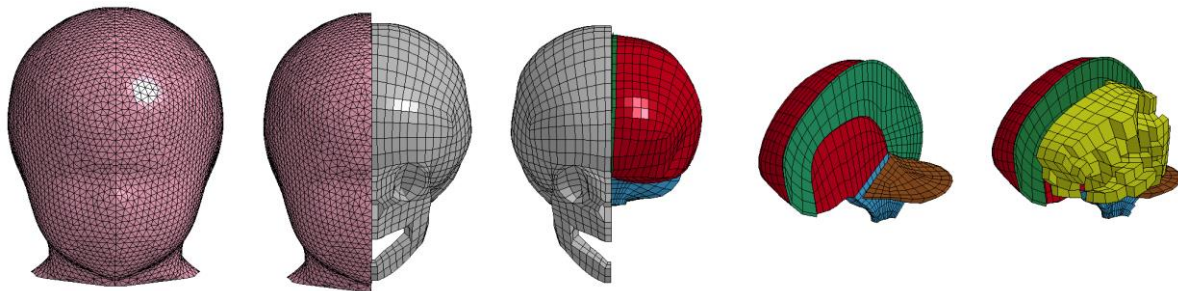


Figure 7: The new head model. On the left the isometric view of the head model is presented. In the middle, the brain is exposed. On the right, internal view of the model with membranes exposed.



Figure 8: The 1.5 y.o (right) and 6 y.o models (left) obtained by scaling using the PIPER Kriging tool. The head model is attached to the new neck model and the LBMC thorax model.

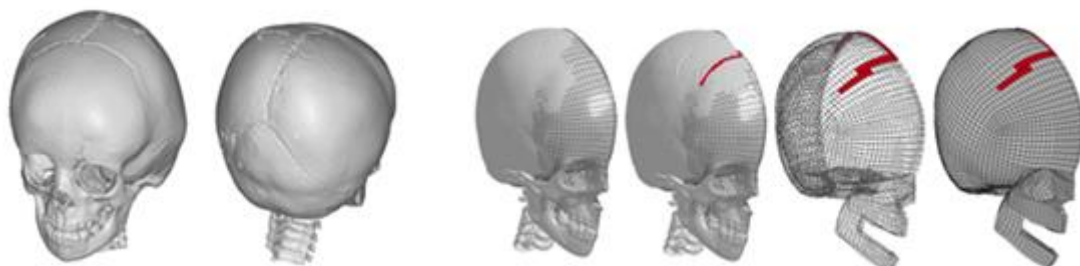
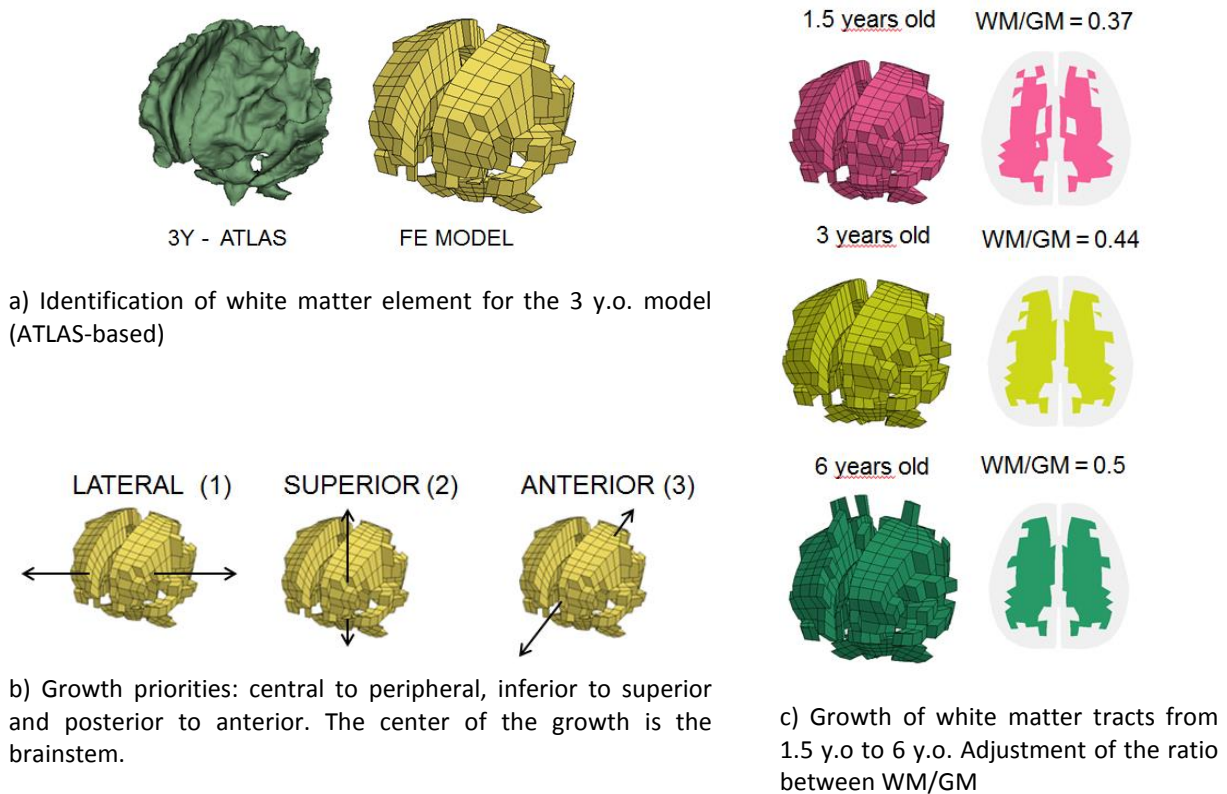


Figure 9: Identification of the cartilaginous structures of the skull. On the left, the segmented surface (high threshold) enhances the location of the sutures. On the right, the segmentation is superimposed onto the model.



**Figure 10: Results of the growth algorithm to adjust the ratio between white matter and gray matter with age.**

### 3.2.2 Neck Modeling

The same computer tomography of the 3 y.o. subject was used to construct the cervical spine baseline model. The cervical vertebrae were modeled using solid elements (hexas) and a layer of shell elements (quads) for the cortical bone. A continuous mesh approach was adopted for the intervertebral disks by extrusion of the elements between the vertebral bodies. The discs include the nucleus pulposus and annulus ground substance, with the annulus fibrosis modeled with fiber reinforced shell elements with two fiber directions superposed on the solid elements of the ground substance. The mayor part of the ligaments were modeled using non-linear spring elements except two ligaments at the upper spine responsible for restricting the motion of the Atlas, modeled with shell elements. An illustration of the new neck model can be seen in Figure 11.

To capture the local anatomical changes occurring during cervical maturation, the target points used in the Kriging scaling were updated in order to better describe the local anatomy, such as facet angle and vertebral body size, illustrated in Figure 12. With the use of known anatomical changes from the literature at each vertebral level, it is then possible to continuously scale the baseline 3 y.o. model to the desired age.

### 3.2.3 Shoulder Modeling

Solid meshes were developed for the humerus, clavicle and scapula. These bones were simulated as deformable using a shell layer (quads) for the cortical bone and solid elements (hexas) for the bones and growth cartilage. Joints were simulated using contacts, ligaments and capsules (as opposed to 6 d.o.f. in the previous model) based on literature descriptions. The posture was also updated to position the arms closer to the body. Illustrations of the new shoulder are provided in Figure 13.

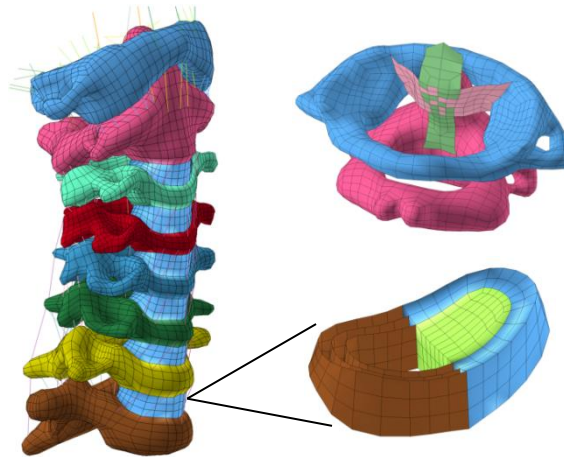


Figure 11: The new neck model to the left, and illustration of the intervertebral disk and the upper cervical spine ligaments to the right.

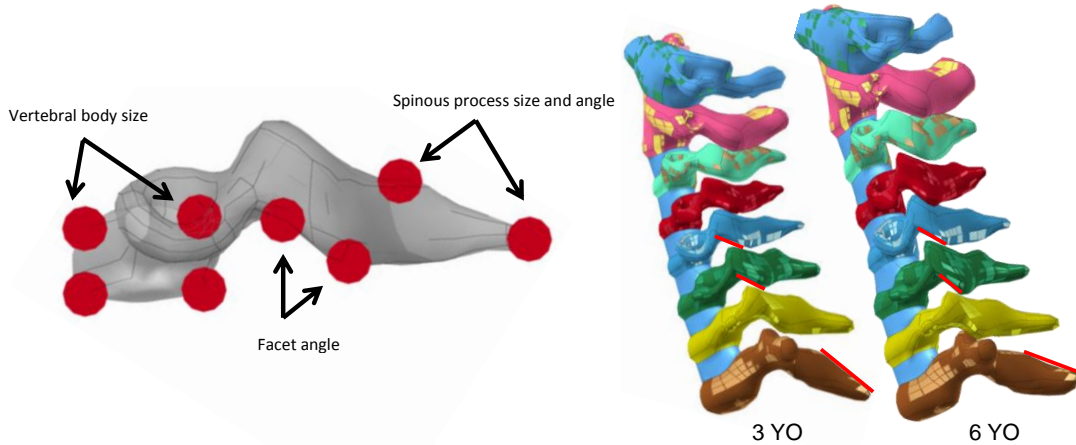


Figure 12: Update of the Kriging targets points for continuously scalable vertebral geometry. Left: point selection. Right: illustration of the 3 and 6 y.o. cervical spines.

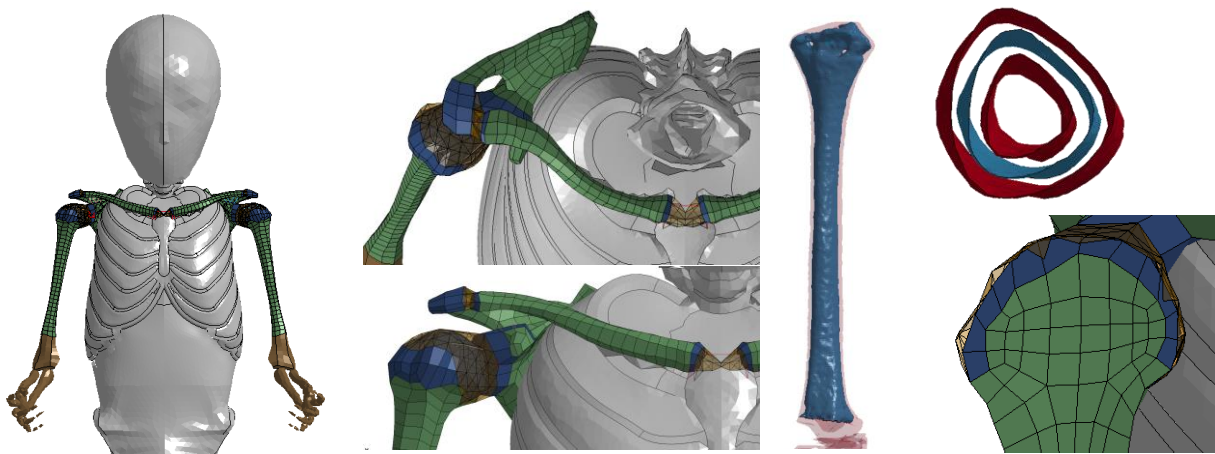


Figure 13: Shoulder update. The position of the arms was adjusted and the bones were modelled using deformable elements (in green and blue) except for the elbow and the forearm. The model accounts for both bony parts (in green) and growth cartilage (in blue). Joints are simulated with contacts, capsules (in transparent orange) reinforced by 1D ligaments (springs). For the humerus, the cortical bone is simulated using a shell located on the mid surface of the cortical component (right).



### *3.3 Preliminary responses for the updated models, at the isolated and full body levels*

While the assembly and model refinement processes are still ongoing, preliminary validation efforts for the updated components are listed below.

#### *3.3.1 Validation of the head model*

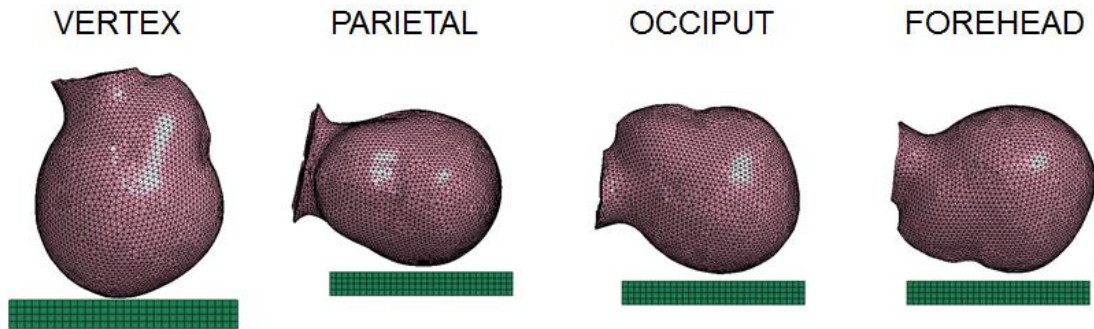
The performances of the 6 y.o. head model were verified on drop test available from Loyd (2011). The head was dropped from heights of 15 and 30 cm for 5 different impact locations: vertex, occiput, forehead, left parietal, right parietal. Figure 14 shows the results of the simulations. The kinematics of the impact is plausible and the model response is in agreement with the experimental data. Further validation will be provided in term of model response in compression test (Loyd 2011). 1.5 y.o and 3 y.o. head models will also be checked.

#### *3.3.2 Validation of the neck model*

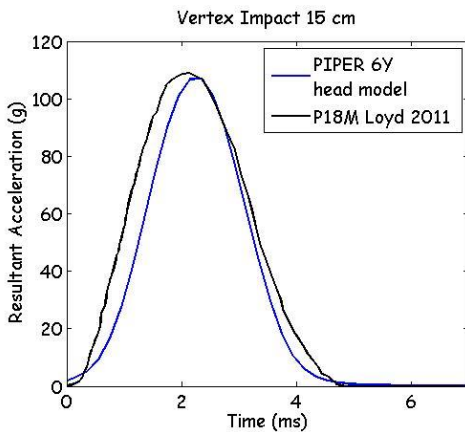
The kinematics of the 3 y.o. cervical spine model was compared to quasi-static flexion and extension bending experiments by (Ouyang et al., 2005) in 0.8, 1.6 and 2.4 Nm. In Figure 15, the response of the model is compared to the experiments as well as illustrating the deformed and un-deformed model, showing a slightly too soft response, especially in flexion which is currently under investigation. Further validation of the cervical spine will include additional local spine kinematics from Luck (2012) as well as global head kinematics in sled tests (Arbogast et al., 2009; Kallieris et al., 1976; Wismans et al., 1979).

#### *3.3.3 Side impact validation*

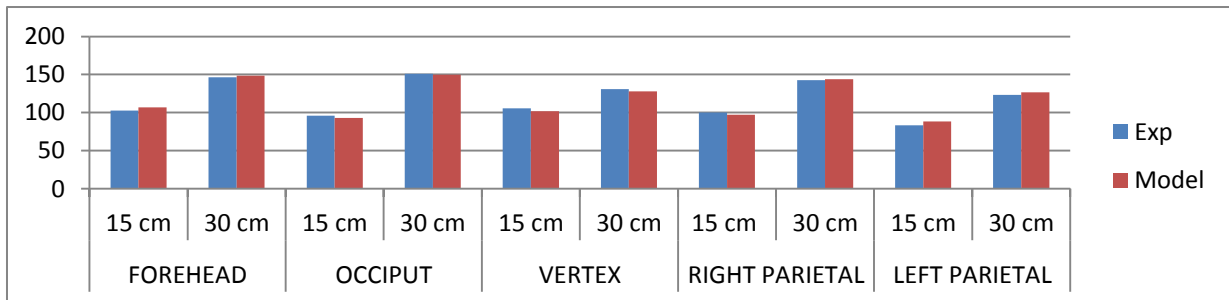
The LBMC baseline model (6 y.o.) was updated with the new shoulder model and associated skin. Most of the configurations previously investigated with the LBMC model were re-simulated without issues. Additionally, the work on the side impact validation was initiated as shown in Figure 13. Simulation results show that, overall, the kinematics of the human model are plausible, while the force developed during the impact is higher than the response corridor proposed by Irwin et al. (2002) for 6 y.o. children. Also, the duration of the impact is shorter with respect to the experimental data. These issues are further amplified for the 3 y.o. scaled model (Figure 16). The differences are currently under investigation.



a) Set-up of the drop tests. The head was dropped from a height of 15 and 30 cm to an aluminum plate. The acceleration of the head was measured.



b) Comparison of experimental and simulated acceleration for the vertex impact from 15 cm of height. Experimental results are reported in black while simulated curves are in blue.



c) Summary of the drop test results for all the impact locations. Peak acceleration are reported for both experimental data and simulated curves.

Figure 14: Preliminary response of the new head model. The experimental data is taken from Loyd 2011.

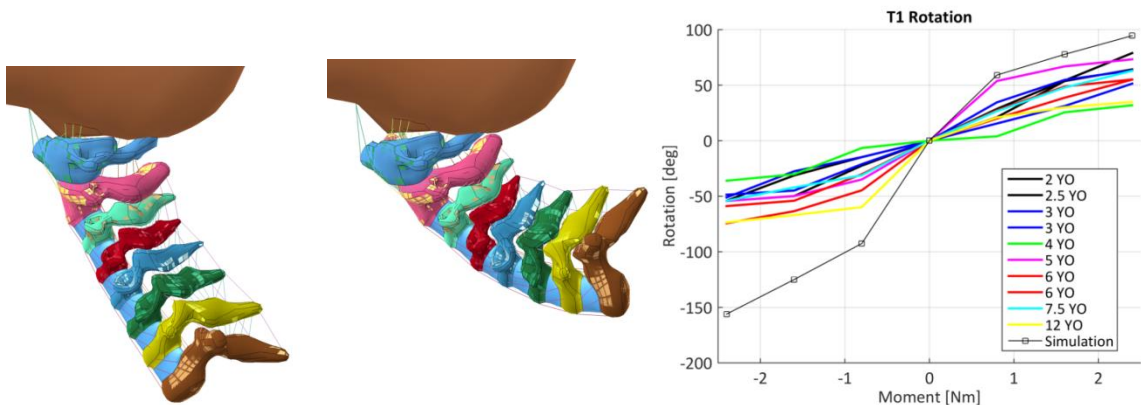
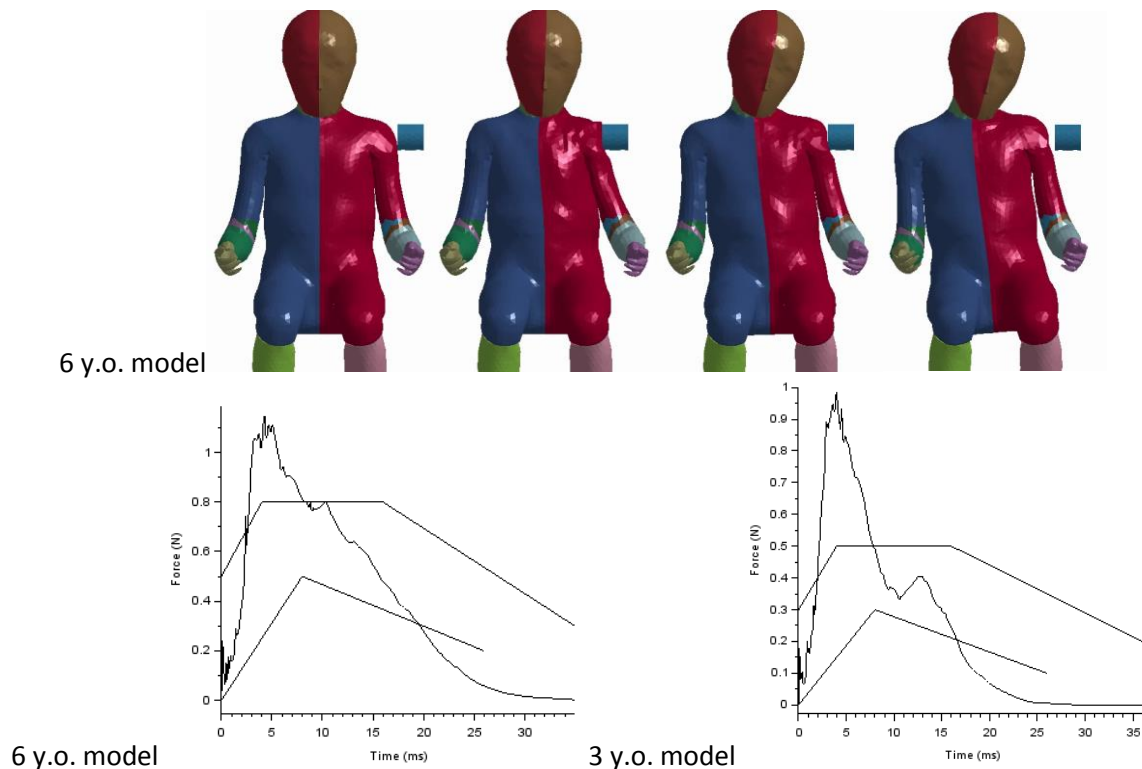


Figure 15. Response of the 3 y.o. model to extension bending, with un-deformed and deformed configurations to the left. To the right is the response of the model compared to cadaver experiments (Ouyang et al., 2005).





**Figure 16: Preliminary response of the new shoulder model. Left: kinematics against the shoulder impactor test and Right: response against scaled corridors proposed by Irwin et al. (2002) for 6 and 3 y.o. children**

#### 4. Discussion and Conclusions

Significant updates of the LBMC models are ongoing to prepare them for various safety applications. Improvements on the description of the deformability of key anatomical structures should help improving the response in frontal impact and also make possible applications in side impact. The updates should also help with future investigations of more local injury mechanisms in regions such as the head and neck (which was not previous possible with the previous LBMC models). A major improvement still needed in the short term is the replacement of the rigid pelvis and femur by deformable models for usage in side impact.

In parallel to the updates, the validation matrix has been expanded (compared to the LBMC models) and will continue being expanded. Transformation processes between models of different ages are under construction to better account for growth and biomechanical changes. The methodology being developed for taking into account the growth of white matter tracts in the brain should be applied to other regions of the body. These methods will be implemented in the PIPER tool to facilitate future applications. Work on evaluation of model scaling and positioning in the PIPER tools is ongoing, along with the model development and should ensure that the tools and models are compatible as soon as they are publicly released, enabling applications such as model scaling as in R129 or positioning as observed in naturalistic driving studies.

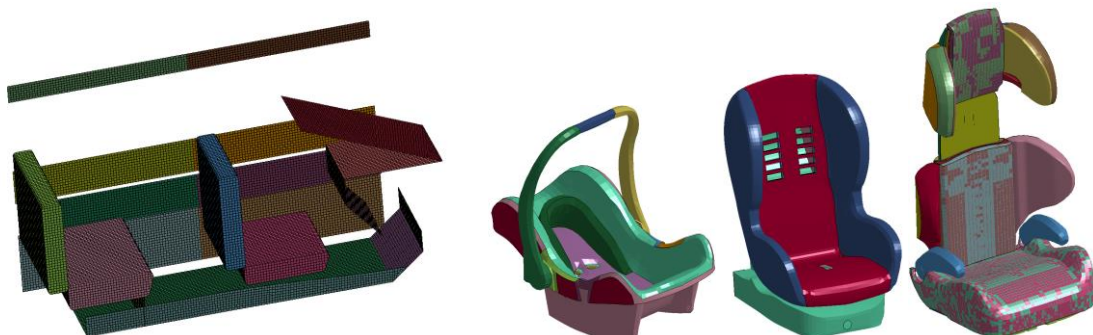
The updated model will be released under an Open Source license based on the GPL v3 one year at most after the end of the PIPER project (April 2017) so that applications and improvements can be performed by other research teams.

Once these updates are completed, one of the first research applications for the model could be related to the determination of injury tolerances based on accident reconstructions. The tests setups currently used with the model may be sufficient to verify its global or regional response to impact

(e.g. in terms of force or displacement characteristics) in a number of relevant configurations, and the use of deformable components representing the human anatomy could be sufficient to help interpolating between them for other impact conditions. However, the datasets currently available are not sufficient for the determination of injury thresholds due to the limited number of specimen (e.g. a few per study at most). One alternative which was previously used for the Q dummies is the use of accident reconstructions to attempt determining injury risk curves (Johannsen et al., 2012). Numerous accidents were reconstructed with actual vehicle in past EC Funded projects (CASPER, CHILD, CREST) which could be used as input for simulations with human body models. However, for such an approach, there will be a need to represent at least part of the vehicle to which the pulse measured in the reconstructions could be applied. In preparation for this work, simplified parametric vehicle environments are being developed as part of the PIPER project

These environment models consist of relevant components of the vehicle interior. The components are parametrized (e.g. positions, angles) and can be adjusted to describe specific interiors for the reconstructions. In the modelling process, existing information about the car-to-CRS interface are being implemented and generic CRS models that were developed during the CASPER project are being improved. Parameters of the environment models will be validated against eligible reconstruction tests using Q dummy models. Illustrations are provided in Figure 17.

While the first reconstructions will be performed in the upcoming two years, it is envisioned that this work will require more fundamental investigation on issues such as the characterization of model performance, uncertainty and sensitivity the accident reconstructions. Work on the procedures is therefore also needed, which could benefit applications in virtual testing.



**Figure 17: First parametric vehicle environment model (under development) and CRS models developed during CASPER project**

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### **References**

- Arbogast KB, Balasubramanian S, et al. Comparison of kinematic responses of the head and spine for children and adults in low-speed frontal sled tests. *Stapp Car Crash J*, 2009, 53:329-372.
- Arbogast K.P., Chen I., Durbin D.R., Winston F.K. (2004). Injury risks for children in child restraint systems in side impact crashes, Proc. of International Research Conference on the Biomechanics of Impact (IRCOBI), Graz, Austria, September 2004.

- Barkovich A., Kjos B. et al. (1988). Normal maturation of the neonatal and infant brain: MR imaging at 1.5 T, *Radiology* 166 (1988) 173–180.
- Beillas P., Lafon J. and Smith F.W. (2009) The Effects of Posture and Subject-to-Subject Variations on the Position, Shape and Volume of Abdominal and Thoracic Organs. *Stapp Car Crash Journal* 53:127–54.
- Beillas P., Soni A., Renaudin F., Johannsen H. (2013) Comparison of the submarining behaviors of a 6 years old human model and a Q6 dummy in sled testing. 11<sup>th</sup> Conference on the Protection of Children in Cars, Munich, Dec 5, 2013
- Beillas P., Soni A., Chevalier M-C., Johannsen H., Müller G., Holtz J., (2014a) Q6 Dummy Thoracic Response and Diagonal Belt Interactions: Observations based on Dummy Testing and Human and Dummy Simulations, IRCOBI Conference Proceeding. Paper IRC-14-39. Berlin, 2014.
- Beillas P., Soni A. (2014b) Assessment of abdominal loading by shield CRS in frontal impact: comparison between direct loading in isolated tests and sleds. 12<sup>th</sup> Conference on the Protection of Children in Cars, Munich, Dec 4-5, 2014
- Cheng H., Obergefell L., Rizer A., "Generator of Body Data (GEBOD) manual", Report No.AL/CF-TR-1994-0051. Human effectiveness Directorate, Crew Survivability and Logistics Division, Wright-Patterson Air Force Base, Ohio, 1994.
- Crandall J., Myers B., Meaney D., Schmidtke S. (2013). *Pediatric Injury Biomechanics*, Springer, New York, 2013.
- Durbin D.R., M.R. Elliott M.R., Winston F.K. (2003). Belt-positioning booster seats and reduction in risk of injury among children in vehicle crashes. *JAMA* 289: 2835–2840
- EEVC (2008) Wismans J, Waagmeester K, LeClaire M, Hynd D, de Jager K, Palisson A, van Ratingen M, Trosseille X, Qdummies report – advanced child dummies and injury criteria for frontal impact, EEVC Document No. 514, 2008
- Fonov S., Evans A.C, McKinstry R.C., Almlı C.L. and Collins D.L. (2009). Unbiased nonlinear average age-appropriate brain templates from birth to adulthood. *NeuroImage* 47(1) S102 DOI: 10.1016/S1053-8119(09)70884-5
- Howard A., Rothman L., McKeag A.M., et al., (2004). Children in side-impact motor vehicle crashes: seating positions and injury mechanisms, *J. Trauma* 56 (6):1276–1285
- Irwin A.L., Mertz H.J., Elhagediab A.M. and Moss S. (2002) Guidelines for Assessing the Biofidelity of Side Impact Dummies of Various Sizes and Ages. *Stapp Car Crash Journal*, Vol. 46 (November 2002)
- Jakobsson L., Bohman K., Stockman I., Andersson M, Osvalder A.-L. (2011) Older Children's Sitting Postures when Riding in the Rear Seat. *Ircobi Conference 2011*. IRC-11-44
- Johannsen H., Trosseille X., Lesire P., Beillas P. (2012) Q-Dummy Injury Criteria Using the CASPER Project Results and Scaling Adult Reference Values, *Proceedings from IRCOBI Conference*, Dublin, 2012.
- Kahle W. – "Anatomie 1, Appareil Locomoteur", Flammarion Medecine-Sciences, 1978
- Kallieris D, Barz J, Schmidt G, Hess G, Mattern R. (1976) Comparison between child cadavers and child dummy by using child restraint systems in simulated collisions. *Stapp Car Crash Conference*, paper 760815, Dearborn, MI, USA.
- Kasai, T., Ikata, T., Katoh, S., Miyake, R., & Tsubo, M., 1996. Growth of the Cervical Spine With Special Reference to Its Lordosis and Mobility. *Spine* 21.
- Kent R, Salzar R, Kerrigan J, et al. Pediatric thoracoabdominal biomechanics. *Stapp Car Crash J*, 2009, 53:373-401.
- Kent R, Lopez-Valdes FJ, et al. Characterization of the pediatric chest and abdomen using three post-mortem human subjects. *Proceedings of the 22nd ESV Conference*, 2011, Washington DC.
- Kumaresan, S., Yoganandan, N., Pintar, F.A., & Maiman, D.J., 2000. Biomechanical study of pediatric human cervical spine: a finite element approach. *Journal of Biomechanical Engineering* 122, 60–71.

- Loyd A.M. (2011). Studies of the Human Head From Neonate to Adult: An Inertial, Geometrical and Structural Analysis with Comparisons to the ATD Head. Dissertation. Department of Biomedical Engineering, Duke University.
- Luck, J.F., 2012. The Biomechanics of the Perinatal, Neonatal and Pediatric Cervical Spine: Investigation of the Tensile, Bending and Viscoelastic Response. Duke University.
- Matsuzawa J., Matsui M. et al. (2001). Age-related Volumetric Changes of Brain Gray and White Matter in Healthy Infants and Children. *Cerebral Cortex* 11 (2001) 335-342.
- Ouyang, J., Zhu, Q., Zhao, W., Xu, Y., Chen, W., & Zhong, S., 2005. Biomechanical Assessment of the Pediatric Cervical Spine Under Bending and Tensile Loading. *Spine* 30.
- Ouyang, J., Zhao, W., Xu, Y., Chen, W., & Zhong, S. (2006). Thoracic impact testing of pediatric cadaveric subjects. *The Journal of Trauma*, 61(6), 1492-1500.
- Park, B.-K., & Reed, M. P. (2015). Parametric body shape model of standing children aged 3–11 years. *Ergonomics*, 1–12. <http://doi.org/10.1080/00140139.2015.1033480>
- R129 Text (2014) Regulation No 129 of the Economic Commission for Europe of the United Nations (UN/ECE) — Uniform provisions concerning the approval of enhanced Child Restraint Systems used on board of motor vehicles (ECRS). Including text up to Supplement 2. Official Journal of the European Union. EUR-Lex - 42014X0329(02). 29.3.2014.
- Reed M.P., Lehto M.M., Schneider L.W., Moss S., Nghi T., (2001) Development of anthropometric specifications for the six-year-old OCATD, Technical Paper 2001-01 -1057. Warrendale, PA: Society of Automotive Engineers Inc.
- Reed, M.P., Ebert-Hamilton, S.M., Manary, M.A., Klinich, K.D., and Schneider, L.W. (2006). Improved positioning procedures for 6YO and 10YO ATDs based on child occupant postures. *Stapp Car Crash Journal*, 50: 337-388
- Scheuer L., Black S. (2004). *The juvenile skeleton*, Elsevier Academic Press, San Diego, CA.
- Wismans et al. (1979) Child Restraint Evaluation by Experimental and Mathematical Simulation, 23rd Stapp Car Crash Conf Proc, SAE #791017
- Yoganandan, N., Kumaresan, S., Pintar, F., & Gennarelli, T., 2002. Pediatric biomechanics. In: Nahum, A., & Melvin, J. (Eds.), *Accidental Injury*. Springer New York, pp. 550–587.