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**EFFECT OF ANTHROPOMETRY SCALING ON THE RESPONSE OF THE PIPER CHILD
SCALABLE HUMAN BODY MODEL SUBJECT TO PELVIC IMPACT**

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Abstract

The Open Source PIPER child scalable human body model was publicly released in April 2017 (www.piper-project.org) along with frontal and side impact validation conditions. The objective of this paper is to investigate the effect of anthropometry scaling on the response of the model in side pelvic impact.

Three setups from two published studies were used: (1) a lateral drop test (2) a greater trochanter impact with a rigid pendulum (3) a pelvis side impact with a flat surface. The first study used scaling assumption developed for crash test dummy design (setups 1 and 2) and the second performed tests on post mortem human surrogates. The baseline 6 years old child model was scaled using a model morphing methodology to match the stature and weight of the surrogates used in the two published studies.

Overall, the main trends observed in the three setups can be approached using the baseline model. Although the model morphing did not account for specific skeletal dimensions, it reduced some of the discrepancies between model response and reference for the drop test and flat plate impact. However, it had little effect on the pendulum test. In that case, the model response was in the corridor at low speed but above at higher speeds. Possible reasons for this difference should be further investigated.

Introduction

Human Body Models (HBMs) have the potential to provide significant insight into the pediatric response to impact in case of motor vehicle accident. The PIPER child scalable finite element model (Beillas et al., 2016) was publicly released under an Open Source license in April 2017 (www.piper-project.org). The model is scalable as a function of stature/age within the PIPER software (also available at www.piper-project.org). Several applications were performed in frontal impact with the PIPER model (Giordano et al., 2017) or a previous version (Beillas et al., 2014). In side impact, reference experimental datasets available for validation are even scarcer than in frontal impact.

For pelvic response, two sources of data are available, which both have potentially significant limitations for human body model validation:

- Corridors scaled from the adult (Irwin et al., 2002). These include two setups for the pelvis: a drop test which loads the whole region and impactor test more focused on the greater trochanter. While these are used as reference for dummy design, the scaling is based on strong assumptions (e.g. use of scale factors in length, mass, moment of inertia and stiffness to characterize the 6 YO (years old) relative to the midsize adult male) which can be difficult to verify. For Irwin, the reference dimensions for the 6 YO are 1168mm high for a weight of 20kg86 which differs from the 6 YO baseline model (1127mm, 23kg).
- PMHS (Post Mortem Human Surrogates) data: data from Ouyang were collected on PMHS, and do not rely on assumptions such as scaled curves. However, the Ouyang's PMHS differ significantly in terms of anthropometry from the model and also from Irwin. For example the two PMHS close to 6 YO are 1090 (1080) mm high for a weight of 20 (16.5) kg.

(Kim et al., 2009) developed a finite element of a 10 YO child pelvis and used the Ouyang's results from four PMHS (mean age 9.4 YO) to identify material properties. This optimization-based approach is performed on an isolated pelvis and doesn't take in consideration the effects of the mass and soft tissues.

The objective of the study is to investigate the effect of anthropometry scaling on the response of a human body model and see if a same model may approximate the response from both sources. This could help understanding, through a model, the compatibility of Ouyang's PMHS results and Irwin's scaled corridors. For the current study, the response of the PIPER child model is compared to both references, before and after morphing the model to the anthropometry of the reference.

Methods

PIPER Scalable Child HBM

The PIPER Scalable Child HBM is a finite element model developed in the Ls-Dyna explicit FE code (LSTC, Livermore, CA) that aims to describe children of various ages. The baseline model describes a 6 YO child whose main anthropometric dimensions were normalized by nonlinear scaling (kriging interpolation) to the dimensions of a 6 YO as predicted by the GEBOD regressions (Cheng et al., 1994). This baseline has a standing height of 1127mm with a total mass of 23kg. Its performance was checked against over 20 setups, mostly derived from PMHS tests, from regional (e.g. isolated head drop) to full body (e.g. sled), in frontal or side impact.

The model is comprised of approximately 546,000 elements of which 52,000 are rigid. They are distributed in 407 parts describing the main anatomical structures with material properties from the literature. With regards to the current application in side impact, the pelvis, proximal femur and ribcage are represented as deformable bones while the thoracolumbar spine is modelled using rigid vertebrae articulated using 6 degrees of freedom (d.o.f) beams. Internal organs of the thorax and abdomen are also described. Growth cartilage is described on the pelvis and proximal femur. Considering the aims of this study, the mechanical properties related to the pelvis were checked against literature sources. For a six year old, the cartilage can be considered as hyaline cartilage (Ponseti, 1978; Grissom et al., 2018) and the equivalent Young's modulus of 29 MPa seemed appropriate testing literature results (Yamada (1970); Oloyede et al., 1992; Ruan et al., 2013). Special attention was given to the elastic modulus and Young's ratio for the pelvis trabecular bone as their values (11.35 GPa and 0.3) seemed overestimated in the PIPER model: they were replaced by 60 MPa and 0.2, respectively (between the values used by (Kim et al., 2009; Dalstra et al., 1995) for the elastic modulus). An elastic law was used and failure was not considered in the current study. The mechanical properties of the pelvis are summarized in Appendix 1 (Ruan et al, 2013; Shi et al, 2015; Kim et al, 2009; Majumder et al, 2004; Dalstra et al, 1995; Parfitt et al, 2000; Meyer et al., 2011). Other body regions are also present in all simulations. They include the head (with deformable skull and brain), neck (with deformable vertebrae, discs, ligaments, muscles) and the extremities (with all joints modelled using 6 d.o.f. beams between rigid bones except the shoulder and the hip which are deformable). Superficial muscles and subcutaneous fat all around the model are modelled using a solid mesh that is covered by a shell representing the skin. Their properties are also provided in Appendix.

All simulations were run using the R7.1.3 of Ls-Dyna, with a time step of 0.32 μ s obtained with marginal mass scaling (15 grams added).

Side pelvic impact setups

The Irwin's lateral drop tests setup consisted in a free fall of the model from heights of 0.5 and 1m onto rigid impact surfaces (Figure 1a). Sacrum acceleration was saved as output of the simulation. (Irwin et al., 2002) proposed also corridors for drop heights of 2.0 and 3.0 m. However the corresponding setups were not simulated in the present study because the characteristics of the padded surfaces were unknown.

The setup of the Irwin's pendulum pelvis lateral impacts was described in the ISO-TR9790 (1999). It consisted in impacting the greater trochanter region by applying to the pendulum an initial velocity varying between 4.5 and 11.5 m/s, as illustrated in Figure 1b. The impactor was scaled by (Irwin et al., 2002) from the 17.3 kg rigid impactor used for the midsize adult, resulting in a 3.89 kg pendulum for the 6 YO dummy (used in the current study). Contact force calculated between pelvis and impactor was saved.

The Ouyang's setup consisted in delivering the impact on the pelvis using an impactor with a flat surface covered with Sorbothane® while the opposing side is against a fixed support covered with neoprene foam, as described in (Ouyang et al., 2003). The Ls-Dyna *MAT_VISCOELASTIC law was used to represent their behavior with material parameters from the literature (Al-Rumaih, 2009; Jeong et al., 2000; Sharma, 2002; Sorbothane, 2018, see Appendix 2). The loading conditions were simulated by applying the initial velocity to the impactor of the same mass (3.24 kg) as the one used in the tests (Figure 1c). Small variations of the orientation of the support and impactor (less than 10 degrees) were also simulated to evaluate if they have an effect in the response of the model. As the velocity varied between PMHS, the exact value was used for the comparisons

between baseline and scaled models (see below). Impactor displacement and contact force between pelvis and impactor were saved as outputs of the simulations.

Anthropometric scaling using the PIPER software

The baseline model was scaled using the PIPER tool to match the stature and weight of the 6 YO anthropometry selected by Irwin et al. (that was used for the 6YO dummy) and of five PMHS (3 and 6 YO children in the Ouyang's setup). All scaling was performed within the PIPER software (v1.0.0, Open Source software available at www.piper-project.org).

In the PIPER software, scaling tools are based on Kriging interpolation (Jolivet et al., 2015). Targets are defined from anthropometric dimensions which are represented by a network of selected control points which are then used to morph the model. Several options are available in PIPER to compute the anthropometric dimensions and corresponding control points and transform the model. They were used for the current study in a three steps approach.

First, the baseline 3 YO model was generated using the PIPER Child scaling module. In that module, child models with external dimensions close to GEBOD (Cheng et al., 1994) can be generated based on a set of control points provided with the PIPER model and Kriging interpolation. Many of these points are on the skin in locations that allow associating them with anthropometric measurements defined in GEBOD. The position of these points is then updated depending on the target age using regressions from GEBOD and used for Kriging (see Appendix 5 for more details). The resulting models, called baseline 6 YO and 3 YO, were used for subsequent scaling for children of similar ages.

Then, the stature and weight of each PMHS were used to predict other anthropometric dimensions using regressions derived from the Snyder database (Snyder et al., 1977) in the anthropometric module. A corresponding network of control points defined using a simplified scalable model (provided with the PIPER software) was then used to represent these anthropometric dimensions on the model and both skeleton and skin were scaled together in the Kriging module.

Finally, the weight was adjusted – typically reduced – by adjusting the external circumferences in regions with large flesh thickness (e.g. legs, arms) using the simplified scalable model. The transformation was performed with the resulting skin envelope and fixed bones in the Kriging module.

The illustration in Figure 2 shows sections and segments dimensions with control points defined in the “Scaling Constraints” PIPER module to reduce the weight of the model

Quantitative comparisons of simulations with experimental results

Curves comparison between the Ouyang's tests and simulations was quantified using two approaches. For the first approach, although it is typically used for time histories, the CORAplus (CORrelation and Analysis) software CORA (2014) was applied to the force displacement signals (as the time histories are not available for the tests). Only the part of the signal with increasing displacement was kept. The ISO 18571 standard ISO/TS 18571 (2014) cross-correlation method was used to generate three subratings for phase, magnitude and slope. A second approach consisted in computing directly the peak values, displacement at peak force, initial slope until the peak force and area under the curve.

Results

Scaling

The models scaled to the Irwin's 6 YO stature and weight and the five Ouyang's PMHS are shown in Figure 3, with heights and weights summarized in Table 1. Segment masses for the Irwin scaled model and Irwin's 6 YO are compared in Appendix 3. [The models will be made available on the web].

Response in Irwin's setups

Acceleration of the sacrum resulting from the free fall simulations for heights of 0.5 and 1m are shown in Figure 4 for the baseline and the Irwin's scaled child models. The response of the model is within Irwin's acceleration limits for the scaled model and below the lower limit for the baseline.

Figure 5 represents the pendulum pelvis force vs the impactor velocity for the baseline and Irwin's scaled models compared to the Irwin's force corridor. The forces are inside the corridor for the 4.5 m/s impact and above for all other velocities. The difference between the baseline and scaled models was small (less than 0.26 kN at 11.5 m/s). For comparison the result of the experimental testing performed by Yaek (2017) with the lateral impact child dummy prototype Q6s is also plotted in Figure 5 for an impact velocity of 6 m/s.

Response in Ouyang's setup

The impactor engaged the pelvic region, resulting in a compression of the soft tissues, followed by a pelvic translation, some spine lateral flexion, and a compression of the soft tissues on the fixed side.

For both experiments and simulations, the force increases almost linearly with the impactor displacement until reaching its peak. Then, the force decreases while the impactor displacement continues to increase (Figure 6). However, in the simulations, the peak force occurred at a larger displacement than in the tests, and the maximum impactor displacement was too small.

The simulations performed with the scaled models reduced the discrepancies observed between the baseline model and the experiment. The scaling improved the peak forces in the three cases with the highest discrepancies (PMHS 5, 8, 9), and improved the displacement at the peak force in all cases. Regarding the Phase, Magnitude and Slope computed with CORA, it was found that morphing improved the magnitude and slope for all five subjects, and phase for 4 of the 5 subjects (or an improvement for 14 out of 15 measures, detailed results in Appendix 6). Average improvements were 0.26, 0.11, 0.16 for the Phase, Magnitude and Slope, respectively. These results were consistent with the direct calculation of peaks, slope and energy on the curves: morphing improved the peak force, displacement at peak force, slope and area under the curve by 5.7%, 49.8%, 20.7% and 2.2%, respectively. The scaling had limited effect on the peak displacement, which remained consistently smaller than the reported experimental results.

Figure 7 illustrates how small rotations (10 degrees or less) of the impactor and the support around the longitudinal axis may affect the shape of the curve for the subject 9: a rotation of the impactor (left) affects mainly the slope of the curve until the peak force while a rotation of the support (right) affects mainly the peak displacement. However the changes of impactor angle or support had more limited effect on the comparison metrics. Positive rotation angles of the impactor and support generally led to small improvements (increase of 0.111, 0.015 and 0.017 for phase, magnitude and slope, respectively) while negative angles degraded them slightly (decreasing of 0.33, 0.03 and 0.01).

Examples of Figure 8 show how small rotations of both impactor and support affect the response of the scaled model, in some cases bringing to closer to the experimental curves. However, the peak displacements remain too small compared to the experimental values.

Discussion

The performance of the PIPER child HBM in pelvis side impact was evaluated using three test setups derived from scaling adult or paediatric experimental data. With the anthropometric scaling based on regressions between dimensions, the mass of the scaled Child HBM were close to those of the PMHS used in the Ouyang's tests (within 0.9 kg) and of the 6 YO described in (Irwin et al., 2002) (difference of 0.34 kg). Furthermore, for the scaling to the anthropometry of Irwin's 6 YO, the masses of the body segments were also approached. For example, the mass of the torso is 10.74 kg for the Irwin's 6 YO and 11 kg for the scaled model (Appendix 3). Two types of impactors were simulated: the pendulum impactor; which engages mainly the greater trochanter (Irwin et al., 2002) and the flat plate impactor, which engages both the iliac wing and the greater trochanter

(Ouyang et al., 2003). The model captured most trends observed in the three setups, and model scaling helped reduce the discrepancies between test and simulation results in most cases. These results were confirmed using the CORA software as well as the calculation of peak force, slope, ... of the curves. Although the CORA methods are not designed to be used with force displacements data, the scores are coherent with the results of the other approach. In particular, model scaling helped bring the accelerations within the corridor defined for Irwin's drop tests. For a 1m drop height, the pelvis acceleration is below the lower boundary of 91 g for the baseline model (76.3 g) and within the limits 91-111 g for the scaled model (100.3 g). This suggests that this setup is very sensitive to the anthropometry.

For the Ouyang's configuration, anthropometry and mass changes also improved the shape of the response curve and the timing of the peak when compared to the baseline model. Peak forces were also improved but the effect on the peak displacement very limited. As the PMHS weights were low considering their age, this suggests scaling accounting for the weight should be considered to develop response corridors. Regarding the peak displacements, it is worthwhile to note that, in the Ouyang's tests, the initial kinetic energy of the impactor is lower than the area under the force vs. displacement curve, by up to 28% for the PMHS 8 (see Appendix 4). This may suggest either (1) a measurement issue or (2) an issue with the pneumatic device that was not completely decoupled after contact and that continued to transfer some energy to the impactor during the compression. However, this seems difficult to assess based on the information available in the Ouyang's publication.

To the contrary of Ouyang's plate impact and Irwin's drop test setups, the subject mass had limited effect in the simulation of Irwin's pendulum impact: at 11.5 m/s, the maximal value of the impactor force is 5.72 kN for the baseline model and 5.46 kN for the scaled model. While the reason is not certain, it can be observed that for that setup, the load is applied locally to the greater trochanter area and transmitted to the acetabulum through the femoral head while the loading was more distributed for the other setups. Overall, while the response of the HBM at 4.5 m/s is within the Irwin's corridor, it is above for higher velocities but closer to the corridor than the Q6s dummy (Figure 5). The difference between the model response and the corridor increases with the velocity. Based on Irwin et al. (2002), the 6 YO corridor has slopes of 0.24 and 0.30 kN/(m.s⁻¹) for the lower and upper boundaries, respectively. In the simulation, the force also increases with the velocity but with a higher slope of 0.41 kN/(m.s⁻¹) (obtained by linear regression). The reasons for this mismatch are not clear and could be coming from the model itself or from the simplified scaling assumptions. Irwin's slopes are scaled from the midsize adult male corridor (slopes of 0.74 and 0.91 kN/(m.s⁻¹)) defined in ISO-TR9790, (1999) using a force scale factor R_F (defined by Mertz (1984)). That factor depends on the masses of the impacted body segments (taken here as the whole lower torso and the lower extremities), the impactors masses, the bone moduli (based on parietal bone) and the erect sitting heights, all estimated on the 6 YO and the adult. The scaling does not account for local parameters such as the width of the pelvis, the cartilage or the bone thickness. The various assumptions should be investigated in the future.

To expand on the comparison between curve scaling and model morphing approaches, it would have been interesting to scale the response curves from Ouyang to account for the anthropometric differences. However, this is not straightforward as the setup is not a simple pendulum loading since the pelvis excursion is limited by a support plate. A simple pendulum scaling approach would have a marginal effect on the response curve (example in Appendix 7-figure H) but its validity is questionable beyond the beginning of the loading when significant loads are carried by the support plate, (i.e. 2.97kN for the subject 8). A more adequate scaling methodology could be further investigated.

Regarding the Irwin setups, scaling the model response using Irwin coefficients had a marginal effect for both lateral drop and pendulum, while model morphing had much larger effect for the lateral drop (figure F and figure G in Appendix 7).

Scaling Limitations

In Ouyang's study, the relative width of the PMHS at the greater trochanter and pelvic wings were not provided. More or less prominent iliac wings or greater trochanters may lead to different loading paths when impacted by a plate. This was documented for the adult by (Lebarbé et al., 2016). Rotations of up to 10 degrees about X in

different directions were applied to the impactor and support to simulate a difference of loading path. This led to curve shape changes and, in some cases, it reduced the difference between simulation and experiment (e.g. PMHS 4 and 8, Figure 8).

More generally, due to the absence of detailed anthropometry of the PMHS, predictions had to be made using average external dimensions derived from the Snyder database. This is an important limitation of the scaling as it does not account for the PMHS specificities, as for example changes of soft tissues thickness or skeletal dimension. With the objective to improve scaling, it would be interesting if such geometrical parameters could be considered for future work, either through subject specific modelling or through the use of statistical models for the skeleton.

Another limitation concerns the PIPER software framework where the scaling is performed only on the anthropometry of the model without considering variations of the material properties with the age.

Conclusion

In order to study its response in side impact, the PIPER child model was scaled to approach the size of the PMHS used in Ouyang's study or reference dimensions for the 6 YO as specified by Irwin using the PIPER software framework. The response of the scaled models demonstrates an effect of the anthropometry on the side impact response. The effect may explain some of the discrepancies when comparing the baseline child model with the Ouyang and Irwin's responses, and does not support specific modifications of the baseline model.

Discrepancies remain mainly on the effect of impact velocity in Irwin's pendulum lateral pelvic impact. While the model is within the corridor at 4.5 m/s, it is above for higher velocities, although closer to the corridor than the Q6s. The reasons for this mismatch should be further investigated.

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Figure 1: Pelvis lateral impact used to assess the performance of the PIPER Child HBM. (a) Irwin's drop test setup (b) Irwin's pendulum setup (c) Ouyang's setup with support (blue) and impactor (red)

Figure 2: Source (white) and target (blue) sections and segments with control points defined in the PIPER application and used to drive the morphing by Kriging interpolation

Figure 3: Scaling to Ouyang's PMHS anthropometry. See Table 1 for the characteristics

Figure 4: Sacrum acceleration vs drop height for the baseline (blue) and scaled (orange) models compared to the Irwin's corridor (Irwin et al., 2002). The PIPER baseline model was scaled to the Irwin's 6 YO stature and weight.

Figure 5: Pendulum pelvis force vs impactor velocity for the baseline (blue) and scaled (orange) models compared to the Irwin's corridor (Irwin et al., 2002)

Figure 6: Response comparison between the baseline (blue), scaled (orange) and test PMHS (dashed) for five PMHS tested in (Ouyang et al., 2003). The 3 YO model is used as the baseline for the first three PMHS (top) and the 6 YO is used for the last two (bottom)

Figure 7: Response of the model scaled to the PMHS 9 impacted with small variations in the orientation of the impactor (left) or support (right). The initial configuration curve (orange), is compared to the modified impactor conditions (green) and the test (black).

Ouyang 2003: Subject 4, impactor and support rotation

Ouyang 2003: Subject 8, impactor and support rotation

Figure 8: Response of the model scaled to the PMHS 4 (left) and PMHS 8 (right) impacted with small variations in the orientation of the impactor and support. The initial configuration curve (orange) is compared to the modified impactor and support conditions (green) and the test (black).

Table 1: Summary of statures and masses for the PMHS and corresponding scaled models

		Age (years)	height (mm)	weight (kg)	Scaled model	
					height (mm)	weight (kg)
Ouyang PMHS	#3	3	850	10.5	850	11.35
	#4	3	930	13.5	930	13.7
	#5	3	910	10	910	10.9
	#8	6	1090	20	1080	19.8
	#9	6	1080	16.5	1080	16.1
Irwin	6 YO	6	1168	20.86	1168	21.2
Baseline Child HBMs	3 YO	3	952	16		
	6 YO	6	1127	23		

Appendix 1

Mechanical properties for the pelvis of the PIPER child HBM

The pelvis is meshed with deformable elements describing the cancellous and cortical bone (tetrahedral and shell elements), as well as the growth cartilage based on the CT scans (**Erreur ! Source du renvoi introuvable.**). Mechanical properties are summarized in **Erreur ! Source du renvoi introuvable.**

Figure I : Pelvis modeling with bones in grey, joint in green and growth cartilage in blue

Table B : Summary of the pelvis properties for the PIPER child HBM. Source of the original properties: EC funded CASPER project - deliverable D2.5 (Meyer et al., 2011). References are only provided for comparison purpose. As large differences were observed for the trabecular bone, its properties were changed and the original value was provided in footnote.

Tissue	Material	Young E, Bulk K, Shear G modulus [GPa], Decay constant β [ms^{-1}], Poisson ν , shell thickness e [mm]	References
Pelvic cartilage	Visco-elastic	$K=0.083$, $G_0=0.01$, $G_i=0.0025$, $\beta=0.25$	(Ruan et al., 2013): $K=0.053$, $G_0=0.009$, $G_i=0.096$, $\beta=0.0015$ Corresponding modulus $E=0.029$, between values from (Oloyede et al., 1992): $E=0.021$ and (Yamada, 1970) : $E=0.050$
Pubic symphysis cartilage	Visco-elastic	$K=0.083$, $G_0=0.01$, $G_i=0.0025$, $\beta=0.25$	(Ruan et al., 2013): $K=0.053$, $G_0=0.009$, $G_i=0.096$, $\beta=0.0015$ Corresponding modulus $E=0.029$: between values from (Oloyede et al., 1992): $E=0.021$ and (Yamada, 1970) : $E=0.050$
Sacro-Iliac joint cartilage	Elastic	$E=0.1$, $\nu=0.3$	(Shi et al., 2014) : $E=0.054$, $\nu=0.4$
Pelvis trabecular bones (3D)	Elastic ¹	$E=0.06^2$, $\nu=0.2^3$	E between values from (Kim et al., 2009) : 0.0448 and (Majumder et al., 2004) , (Dalstra et al., 1995): 0.070
Pelvis cortical bones (2D)	Elastic	$E=15$, $\nu=0.3$, $e=1$	E between values from (Kim et al., 2009) : 11.6 and (Dalstra et al., 1995): 17 (Parfitt et al., 2000) : $e=0.8$

¹ Instead of an elastoplastic material as defined in the PIPER Child HBM

² Instead of the value of 11.35 GPa defined in the PIPER Child HBM

³ Instead of the value of 0.29 defined in the PIPER Child HBM

Appendix 2

Mechanical properties for the sorbothane impactor and neoprene support

Sorbothane and neoprene are modelled as viscoelastic materials with the LS-Dyna *MAT_VISCOELASTIC law. Material properties are given in Table C.

Table C: Parameters of the LS-Dyna *MAT_VISCOELASTIC law for sorbothane and neoprene

References	Sorbothane (Sorbothane, 2018), (Al-Rumaih, 2009) or assumed	Neoprene (Jeong et al., 2000)
Density (10^{-6}kg/mm^3)	1.3	1.23
Elastic Bulk modulus (GPa)	4	0.0785
Short-time shear modulus (GPa)	9e-04	8.3e-03
Long-time shear modulus (GPa)	4e-04	1.8e-03
Decay constant (ms^{-1})	0.05	0.53

An alternative source for sorbothane properties was also considered: the experimental engineering stress-strain curve published by (Sharma et al., 2002)(Figure J) was used as input the LS-Dyna *MAT_HILL_FOAM law. The effect on the response was marginal (Figure K) and the viscoelastic properties were kept for the final simulations.

Figure J: Engineering stress-strain plot for sorbothane extracted from (Sharma et al., 2002)

Figure K: Simulation of the response of the HBM (PMHS 8 of Ouyang) by modelling sorbothane covering the impactor with a `*MAT_VISCOELASTIC` or a `*MAT_HILL_FOAM` law.

419 **Appendix 3**

420 **Segment masses and total body mass for the PIPER HBM scaled to the Irwin's 6 YO stature and weight.**

421 *Table D: Segment masses and total body mass for the Irwin's 6 YO and the scaled model (units: kg)*

Segment	Irwin's 6 YO	Scaled model
Head + Neck	3.97	3.82
Torso	10.74	11
Upper Extremities	1.87	2.1
Lower Extremities	4.28	4.24
Total Body Mass	20.86	21.16

422

Appendix 4

Ouyang's setup: comparison between the initial kinetic energy of the impactor and the work done by the contact force

For each Ouyang's PMHS, the work done by the contact force between the HBM and the impactor is evaluated from the integration of the curve force vs impactor displacement (Figure L).

Figure L: Contact force vs impactor displacement to evaluate the work done by the force.

Table E: Comparison between the initial kinetic energy of the impactor and the work done by the HBM-impactor contact force (mass of the impactor = 3.24 kg)

PMHS	Age years	Impact velocity m/s	Impactor kinetic energy Joules	Work done by the contact force Joules	variation Joules	variation %
3	3	8	103.7	120	16.3	15%
4	3	7.9	101.1	115.5	14.4	13%
5	3	7.7	96.0	108.8	12.8	12%
8	6	9.1	134.2	177	42.8	28%
9	6	7.1	81.7	94.2	12.5	14%

Appendix 5

Scaling using the child scaling module

This module is dedicated to the scaling of the PIPER scalable model as a function of the age or stature. A set of 291 control points (source points in orange) provided with the PIPER metadata are defined on the skin surface or on the near spine. The points on the skin are associated with anthropometric dimensions used in GEBOD (Cheng et al. 1994). Using regressions linking age (in this example the target age is 3 YO) to the anthropometric dimensions, the control point positions can be updated, generating target points (in blue). For the points near the spine, separate assumptions are used (e.g. linear scaling) in order to avoid creating local deformities. Model morphing (kriging) is then driven by the control points. In this example a preview of the scaled model (3YO) is shown in red.

Figure M : scaling of the PIPER 6YO child model using the child scaling module

Appendix 6

Quantitative comparisons of simulations with experimental results

1. Quantitative comparisons using the CORA software

Table F : Cora scores for the curves of figure 6

Ouyang PMHS	Child model	Phase	Magnitude	Slope
#3	Baseline	0.1	0.813	0.724
	Scaled Model	0.6	0.845	0.843
#4	Baseline	0.5	0.873	0.684
	Scaled Model	0.9	0.897	0.841
#5	Baseline	0.455	0.685	0.523
	Scaled Model	0.091	0.869	0.779
#8	Baseline	0.5	0.726	0.557
	Scaled Model	0.6	0.842	0.747
#9	Baseline	0.111	0.694	0.65
	Scaled Model	0.778	0.864	0.718
average improvement scaled Model/Baseline		0.26	0.11	0.16

Table G : Cora scores for the curves of figure 7

	Scaled Model	Phase	Magnitude	Slope
Subject 9	wo Impactor or Support Rotation	0.778	0.855	0.725
	Impactor Rot. +10deg	1	0.87	0.774
	Impactor Rot. -10deg	0.111	0.812	0.783
	Support Rot. +5deg	0.778	0.87	0.71
	Support Rot. -5deg	0.778	0.842	0.642
	average rotations >0	0.111	0.015	0.017
	average rotations <0	-0.3335	-0.028	-0.0125

Table H: Cora scores for the curves of figure 8

Ouyang PMHS	Scaled Model	Phase	Magnitude	Slope
#4	wo Impactor or Support Rotation	0.778	0.855	0.725
	Impactor and Support Rot. +5deg	1	0.87	0.774
#8	wo Impactor or Support Rotation	0.6	0.846	0.725
	Impactor and Support Rot. -5deg	0.9	0.827	0.677

2. Quantitative comparisons computing peak force, displacement, slope and area under the curves for figure 6 of the manuscript

Table I : Peak force, impactor displacement at peak force, slope and area under the curves for figure 6 of the manuscript

		Peak Force value [kN] Fmax	Displ. [mm] at Fmax	Slope [kN/mm] at Fmax	Area under the curve [J]
Subject 3	Experiment	2.81	22	0.128	90
	Baseline	2.91	34	0.086	96
	Scaled Model	2.99	21	0.143	100
Subject 4	Experiment	3.00	26	0.116	91
	Baseline	2.86	34	0.084	93
	Scaled Model	2.62	27	0.097	99
Subject 5	Experiment	2.16	21	0.103	77
	Baseline	2.76	34	0.081	95
	Scaled Model	2.41	26	0.093	82
Subject 8	Experiment	4.36	18	0.242	140
	Baseline	3.29	37	0.089	128
	Scaled Model	3.85	32	0.120	131
Subject 9	Experiment	3.18	16	0.198	86
	Baseline	2.31	33	0.070	77
	Scaled Model	2.62	16	0.163	79

Table J: Variations (%) of peak force, displacement at peak force, slope and area under the curves for baseline and scaled models vs experimental testing (figure 6 of the manuscript)

Ouyang PMHS	Curve	Fmax Variations [%]	Displ. Variations [%] at Fmax	Slope Variations [%] at Fmax	Energy variations [%]
#3	Experiment				
	Baseline	-3.8	-54.5	32.9	-6.8
	Scaled Model	-6.6	4.5	-11.7	-10.7
#4	Experiment				
	Baseline	4.7	-30.8	27.2	-2.5
	Scaled Model	12.6	-3.8	15.9	-9.3
#5	Experiment				
	Baseline	-27.7	-61.9	21.1	-23.3
	Scaled Model	-11.3	-23.8	10.1	-6.8
#8	Experiment				
	Baseline	24.6	-105.6	63.3	8.4
	Scaled Model	11.6	-77.8	50.2	6.3
#9	Experiment				
	Baseline	27.3	-106.3	64.7	10.6
	Scaled Model	17.6	0.0	17.6	7.3
Averages	Baseline	17.6	71.8	41.8	10.3
	Scaled model	11.9	22.0	21.1	8.1
	Scaled model/Baseline	5.7	49.8	20.7	2.2

Appendix 7

Scaling the 6YO PIPER model results

Irwin's coefficients (Irwin et al., 2002) were computed to scale the response curves of the 6YO PIPER child model (baseline):

- to the weight and size of the PMHS #8 in Ouyang's setup using a pendulum scaling approach ;

- to the weight and size of Irwin's 6YO in Irwin's setups (pendulum and drop tests).

They are described in the tables below for pendulum tests (Table K) and drop tests (Table L).

Table K : Evaluation of the Irwin's coefficients for pendulum tests to scale the 6YO PIPER child model (baseline) to the Ouyang PMHS #8 (Ouyang's setup) and to the Irwin's 6YO.

Target	Ouyang PMHS #8	Irwin 6YO
Age	6	6
Target mass M_t	20.00	20.86
Baseline mass M_b	23.00	23.00
Target effective mass M_{te}	10.7	10.43
PIPER baseline effective mass M_{be}	12.305	12.305
Pendulum mass M_{pt} for target	3.24	3.89
Pendulum mass M_{pb} for the PIPER baseline	3.24	3.89
$\lambda_{mt} = M_t/M_b$	0.8696	0.8476
$\lambda_{mp} = M_{pt}/M_{pb}$	1.0000	1.0000
$\lambda_{ms} = (M_{te} + M_{pt}) / (M_{be} + M_{pb})$	0.8968	0.8842
$\lambda_{me} = \lambda_{mt} * \lambda_{mp} / \lambda_{ms}$	0.9697	0.9586
λ_k	1.0	1.0
λ_v	1.0	1.0
$R_f = \lambda_v * (\lambda_{me} * \lambda_k)^{0.5}$	0.9847	0.9791
$R_d = \lambda_v * (\lambda_{me} / \lambda_k)^{0.5}$	0.9847	0.9791

Table L : Evaluation of the Irwin's coefficients for drop test to scale the 6YO PIPER child model (baseline) to the Irwin's 6YO.

Target	Age	Target mass M_t	Baseline mass M_b	λ_k	λ_v	$\lambda_{mt} = M_t/M_b$	$R_a = \lambda_v * (\lambda_k / \lambda_{mt})^{0.5}$
Irwin 6YO	6	20.86	23.00	1.0	1.0	0.9070	1.0500

The effect of the scaling applied to the 6YO PIPER model results (Figure 4 to figure 6 of the manuscript) is shown on the following figures.

Figure N : Effect of the Irwin's scaling on the sacrum acceleration vs drop height

Figure O : Effect of the Irwin's scaling on the pendulum pelvis force vs impactor baseline

Figure P : Effect on the Irwin's scaling on the pelvic lateral force vs the impactor displacement for the Ouyang test with subject 8