

Comparing the Kinematics of the Head and Spine between Volunteers and PMHS: a Methodology to Estimate the Kinematics of Pediatric Occupants in a Frontal Impact

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Abstract

Traumatic brain and skull injuries are the most common serious injuries sustained by children in motor vehicle crashes. The spine dictates the position and orientation of the head during the impact. Research has shown major discrepancies between the spinal kinematics of current pediatric Anthropomorphic Test Devices and humans during frontal impacts. The paucity of pediatric experimental data requires of applying scaling methods to predict the pediatric response using adult data. This paper presents data on four different experimental data sets corresponding to non-injurious low-speed impacts with pediatric and adult volunteers and to low-speed and high-speed impacts with Post Mortem Human Surrogates (PMHS). Two published scaling methods (mass scaling and SAE scaling) were assessed using volunteer data and failed to predict the actual pediatric displacement. A new scaling method was developed to improve the prediction of the pediatric response at low speed. The new method was then applied to the high-speed PMHS data to provide an approximation of the displacements of the head, thoracic spine and pelvis of a 6-year-old occupant in a 40 km/h frontal impact. The limitations of the method are discussed in the paper.

Keywords Frontal impacts, head and spine displacement corridors, pediatric response, scaling.

I. INTRODUCTION

Traumatic brain and skull injuries are the most common serious injuries sustained by children in motor vehicle crashes regardless of age group, impact direction or type of restraint ([1],[2]). The position and attitude of the head during the crash is dictated by the spine. The motion of the spine also determines how the seatbelt interacts with the torso of the occupant [3].

Design of restraint systems depends largely on the results of simulated crashes using anthropomorphic test devices (ATD) as human surrogates. Previous research has shown that, even if ATD can predict the maximum forward human head excursion during a frontal impact, there exist critical differences in spinal kinematics when ATD are compared to humans ([4]-[6]). These studies have suggested that the compliance of the human spine may be the cause of these differences. The thoracic spine of the frontal impact ATD (Hybrid III) is essentially rigid and unable to exhibit the flexion/extension characteristic seen in humans. Pediatric ATD are not an exception and the Hybrid III 6YO spine has been shown to be much stiffer than that of a real child [5]. The stiffness difference has been suggested to cause the ATD to predict unrealistic values for the loads transmitted to the neck that are rarely seen in real crashes [5].

The lack of knowledge on the kinematics of pediatric occupants during a frontal impact precludes the development of more biofidelic pediatric ATD. A review of the literature identified a total of 15 full-scale sled tests performed with 11 pediatric PMHS ([7]-[12]) and three full-scale sled tests with an adult PMHS having the size of a 10-year-old ([6],[13]). Given the paucity of experimental pediatric data, pediatric response for impact research has been obtained through scaling adult Post Mortem Human Subject (PMHS) responses ([14], [15]).

In 2007, The Children's Hospital of Philadelphia and the University of Virginia-Center for Applied Biomechanics launched a collaborative project with the goal of characterizing the kinematics of the pediatric spine during low speed impacts as well as to provide a unique kinematic dataset that allowed exploring new

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methods of scaling between adults and pediatric occupants in frontal impacts. The project involved pairs of tests performed in matching impact conditions between pediatric and adult volunteers at a sub-injurious speed as well as between Post Mortem Human Surrogates (PMHS) at two different speeds. Some subsets of these tests ([16],[19]) have been already published, but this is the first publication in which the whole set of matching experiments are presented and discussed as a whole.

The goal of this paper is to provide an estimation of the head, spinal and pelvic displacement of a 6-year-old (6YO) occupant in the sagittal plane during a high-speed frontal impact. The prediction of the kinematics of the 6YO is obtained by scaling the response of adult PMHS at high speed. More specifically, this paper presents:

- The development of corridors for the head, spinal and pelvic displacements of pediatric and adult volunteers at 9 km/h and of PMHS at 9 and at 40 km/h.
- An assessment of the mass scaling [20] and the SAE scaling ([14],[15]) methods to predict the response of a 6YO in a low-speed frontal impact.
- A proposal for a scaling method based on energy considerations to predict the trajectories of the head and spine of a 6YO in the same impact conditions.
- An approximation of the kinematic response of a 6YO in a 40 km/h frontal impact, based on the previously mentioned scaling methods.

The results for the prediction of the kinematics of the pediatric occupant obtained by the different scaling methods are compared and discussed with the aim of providing insight to guide the process of scaling between adult and pediatric subjects in frontal impacts.

II. METHODS

The study required performing non-injurious low-speed tests with pediatric and adult volunteers as well as low-speed and high-speed tests with PMHS. The test matrix is shown in Table I. Although the original study included pediatric subjects from different age groups, the pediatric tests discussed here correspond exclusively to those within the 6-year-old group [16]. Also, only a sample of five adult volunteers was selected to be used in this study. The third group of test subjects consisted of three cadavers nominally of the size of a 50th male percentile. The main anthropometric characteristics of each of the subjects are shown in Table II.

TABLE I
TEST MATRIX

	Pediatric volunteers	Adult volunteers	PMHS
9 km/h	PED1, PED2, PED3, PED4	AD1, AD2, AD3, AD4, AD5	PMHS1,PMHS2,PMHS3
40 km/h			PMHS1,PMHS2,PMHS3

The protocol of the study was reviewed and approved by the Institutional Review Boards at The Children's Hospital of Philadelphia and the Oversight Committee of the Center for Applied Biomechanics – University of Virginia.

Volunteer study

The Children's Hospital of Philadelphia designed and conducted non-injurious low speed frontal sled tests on pediatric and adult human volunteers. From a total of 30 male subjects enrolled in the study (20 children and 10 adults), four pediatric volunteers within the 6-year-old group and five adult subjects were chosen for this study. Each subject was exposed to six trials on a custom designed sled. The acceleration pulse was chosen after a meticulous study considering the safety, comfort and protection of the volunteers involved in the study. The crash pulse for the pediatric group consisted of a peak of 3.62 g over 124.6 ms, while that experienced by the adult group was slightly greater (3.82 g over 119.2 ms). Volunteers were asked to remain relaxed during the tests. Forward motion of occupants was restrained by a three-point conventional seatbelt equipped with a retractor. The initial torso and knee angles were set to 110 degrees by adjusting the foot and back rest position. To minimize the effect of initial head angle, the subjects were asked to initially position their head by focusing on a point placed directly in front of them at the level of their nasion. The same buck was used for all the volunteers. The height of the seatbelt D-ring was adapted to each occupant so as to maintain homologous loading conditions between subjects.

Reflective markers were placed on the head, neck, torso, upper and lower extremities. These markers were tracked in time using a 3D motion analysis system (Model Eagle 4, Motion Analysis Corporation, Santa Rosa,

CA, US) at a frequency of 100 Hz. On the spine, markers were located on the spinal process of C4, T1, T4, T8 and T12. The anterior superior iliac spines of the subjects were also tracked during the duration of the test. A comprehensive description of the instrumentation, test setup, subject anthropometry and data analysis can be found in [16].

TABLE II
ANTHROPOMETRY OF TEST SUBJECTS

	AGE (years)	GENDER	WEIGHT (kg)	HEIGHT (cm)	SEATED HEIGHT (cm) ⁺
PED1	6	M	24	122	69
PED2	8	M	35	140	73
PED3	7	M	28	133	70
PED4	8	M	29	130	71
AD1	24	M	68	165	84
AD2	22	M	107	180	97
AD3	24	M	74	169	91
AD4	30	M	81	180	93
AD5	20	M	82	178	93
PMHS1	59	F	80	167	93
PMHS2	69	M	84	178	92
PMHS3	60	M	81	191	93
Avg. 6 YO human*	6	NA	21	118	64
Adult 50 th human**	40	M	82	180	94

⁺ Measured from seat to top of the head * Malina et al. (1973) [17] ** Man-Systems Integration Standards (NASA, 1995) [18]

The specific objectives of the volunteer study were to quantify the trajectories of the head and thoracic spine of pediatric occupants during a low speed frontal impact and compare this kinematics with those of adult occupants. The results from these tests have been already published and discussed in [16] and constitute the basis for the exploratory analyses of scaling between adult and pediatric subjects discussed in the present paper.

TABLE III
MAXIMUM CHANGE IN VOLUNTEER HEAD ANGLE. DATA ARE EXPRESSED IN DEGREES (AVERAGE ± STANDARD DEVIATION). ADAPTED FROM [16].

Age group	Relative to sled coordinate system	Relative to segment joining T1-T4
6-8 years	42.9±6.4	12.4±6.5
9-11 years	38.6±6.5	12.9±4.8
12-14 years	32.1±6.6	11.7±5.8
Adults (21-37 years)	22.8±8.1	5.3±5.2

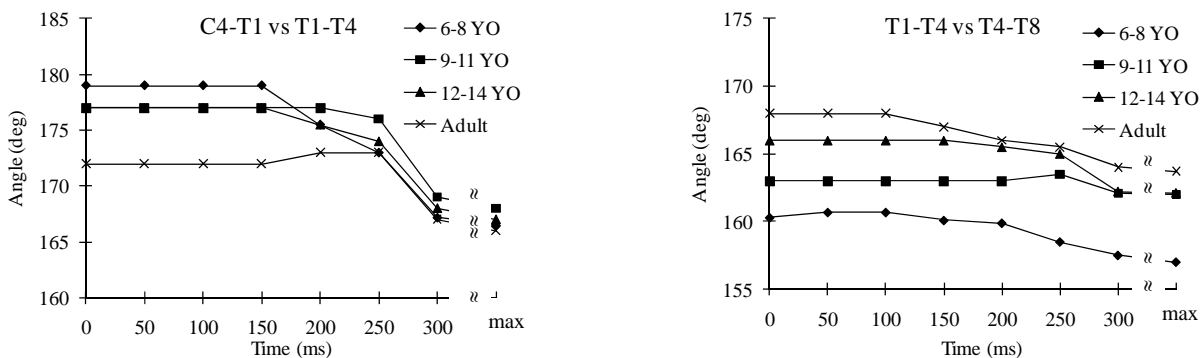


Fig. 1. Volunteer study. Change in angle between rigid links connecting C4-T1 vs T1-T4 (upper) and T1-T4 vs T4-T8 (lower). Adapted from [16].

Table III and Figure 1 show some of the most significant results reported in [16]. The significant differences found in the flexion angles between the age groups challenge the validity of the application of any scaling method based on dimensional analysis to predict the pediatric response using scaled adult data.

PMHS study

The Center for Applied Biomechanics performed a total of six sled tests on three adult PMHS. PMHS were screened before testing to confirm the absence of any infectious blood disease (HIV, Hepatitis B, Hepatitis C) and of any other pathology that could affect the impact response and injury occurrence of the subjects.

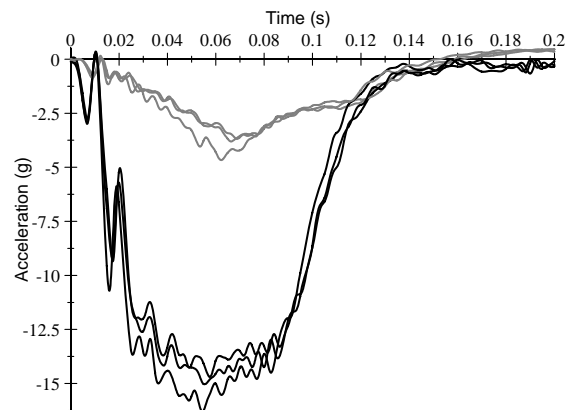


Fig. 2. Low speed (~9km/h) and high speed (~40 km/h) deceleration pulses used in the PMHS study [19]. The low speed deceleration pulse was similar to that used in the volunteer study [16].

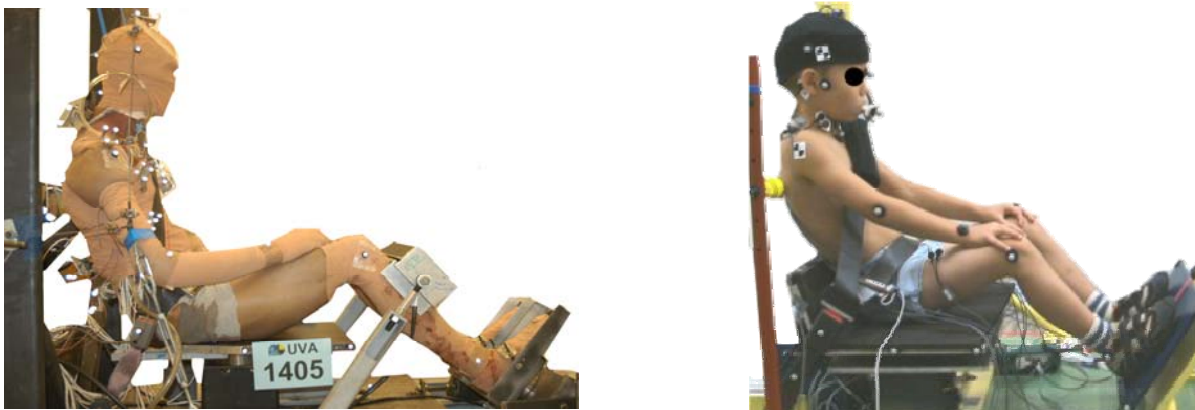


Fig. 3. Comparison between the initial position of a PMHS and a pediatric volunteer.

Kinematic data were obtained using a 16-camera Vicon MX™ motion capture system operating at 1000 Hz. The cameras tracked the motion of spherical retroreflective targets within the cameras' collective viewing volume. Clusters of four markers were rigidly attached to the head, right acromion, T1, T8, L2, L4, pelvis, 4th and 8th ribs bilaterally and sternum. The use of 3 or more non-collinear markers allowed the 6 DOF position and orientation of each structure to be determined [21]. An extensive description of the designed and attachment of the hardware to the anatomical structures is discussed in [22].

Each PMHS was first subjected to a low speed (approximately 9 km/h) frontal impact comparable to that of the volunteers. Then, a high speed test (approximately 40 km/h) was performed on the same subject. Figure 2 shows the crash pulses used in these tests. The forward motion of the PMHS was restrained by a three-point seatbelt of similar characteristics to the one used in the volunteer study. Belt position on the torso of the PMHS as well as the position of the seatbelt anchor points with respect the seat matched also the conditions used in the volunteer tests [16]. Initial position of the PMHS was set to replicate the volunteer initial position. PMHS head initial angle was set to zero degrees. A comparison between the initial position of one of the PMHS and one pediatric volunteer is shown in Fig. 3.

Normalization of results and corridor creation

Before undertaking any scaling effort, the responses from each of the subjects were combined into corridors representing each of the four datasets considered in the analysis: pediatric volunteers at 9 km/h, adult

volunteers at 9 km/h, adult PMHS at 9 km/h and adult PMHS at 40 km/h. The method described in [23] was used here to preserve the characteristic shape of individual trajectories in the sagittal plane. Each group of subjects was normalized to the 50th percentile within their age group. The anthropometric characteristics of the 50th percentile 6-year-old and the 50th percentile adult male are included in Table II. The method provides the average response of the subject and a corridor that includes the standard deviation of the response in both the X and Z axes.

Scaling methods

Mass scaling. The method was first used in [20]. It is based on dimensional analysis and assumes that there is geometrical and dynamic similarity between the two systems related by the scaling factors. The three fundamental magnitudes used to scale all the other magnitudes in the system are length (subscript *L*), mass density (subscript *ρ*) and modulus of elasticity (subscript *E*). A length scale factor is calculated as the cube root of the ratio of the mass (subscript *m*) of a standard-sized subject to the actual subject, assuming that the densities are equal between the subjects. The difference in material properties due to tissue development are accounted for in this method according to the value λ_E proposed in [15]. In the original study λ_E was assumed to be 1, since it dealt with scaling between different sizes of adults. Reference [15] reported that the ratio of elastic moduli of bone ($\lambda_E=0.88$) between a 6-year-old and an adult mid-size male can be used to scale the mechanical response of the adult. Though this value was proposed to be used within the SAE scaling paradigm, it has been considered also here to account for the differences in tissue properties due to development. The following relationships can be derived using dimensional analysis:

$$\lambda_t = \lambda_E^{-1/2} \lambda_m^{1/3} \tag{1}$$

where *t* is the time and *m* is the mass.

SAE scaling. This method was originally applied to scale the Hybrid III 50th percentile to the small female and large male ATD ([14],[15]). The method involves a length scale factor (given by the erect seated height of the subjects) and a mass scale factor (total body mass). Again the assumption of equal density is imposed in the method, so that the length scale factors in the x- and y- directions are given by (2).

$$\lambda_x = \lambda_y = \sqrt{\frac{\lambda_m}{\lambda_z}} \tag{2}$$

The SAE scaling method also considers the differences in the material properties of the tissue due to maturation. These differences are given by the scaling factor for the modulus of elasticity λ_E shown in Table IV [15]. In the original work published in [14], λ_E was assumed to be equal to 1 since the scaling was applied to adult subjects. In this study, the moment-scaling factor has been updated to reflect the age-changing properties of the tissue as shown in Table IV ($\lambda_E=0.88$).

TABLE IV
SCALING FACTORS BETWEEN SPECIMENS M320-T7T9 AND F470-T7T9 AS GIVEN BY THE MASS SCALING AND THE SAE SCALING METHODS

	λ_L	λ_ρ	λ_E	λ_m	λ_t
Mass scaling	$\lambda_x=\lambda_y=\lambda_z=0.64$	1	0.88	0.26	0.68
SAE scaling	$\lambda_z=0.68, \lambda_x=\lambda_y=0.62$	1	0.88	0.26	1

Scaling based on energy considerations. As an alternative to the previously discussed methods, we explored a parallel scaling method that would consider the transfer of the kinetic energy of the buck into the forward motion of the restrained occupant. The assumption is that the energy of the buck equals the work done by the seat belt restraining the forward motion of the occupant (equation (3)) [24].

$$\frac{1}{2} m_i \Delta v^2 = \int_0^{s_{\max}} F_i ds_i$$

$$i = \text{pediatric, adult}$$

$$(3)$$

where m is the mass of the occupant, Δv is the speed change in the considered time interval, F is the seatbelt force and s is the displacement. Assuming that the pediatric and adult subjects are exposed to the same Δv , then the following expression can be obtained:

$$\frac{m_{ped}}{m_{ad}} = \frac{\int_0^{s_{\max}} F_{ped} ds_{ped}}{\int_0^{s_{\max}} F_{ad} ds_{ad}} \Rightarrow \int_0^{s_{\max}} F_{ped} ds_{ped} = \frac{m_{ped}}{m_{ad}} \int_0^{s_{\max}} F_{ad} ds_{ad}$$

$$(4)$$

And the next step consists of approximating the value of the integral as the difference between the value of the product $F_i s_i$ at the initial and at the final position. Thus, if the calculations are performed until the time of maximum forward head excursion (which coincides approximately with peak shoulder belt force [19]), expression (4) can be simplified to obtain:

$$s_{ped} = \frac{m_{ped}}{m_{ad}} \frac{F_{ad_{peak}}}{F_{ped_{peak}}} s_{ad}$$

$$(5)$$

Equation (5) relates the length of the path of the pediatric and adult subjects. To apportion this path to the X and Z axis, the real trajectory of the pediatric volunteers can be used so that the intrinsic differences between the two groups are considered in the scaling.

Analysis

Although the kinematic response of the volunteers and PMHS tests has been partially published elsewhere ([16],[19]), we have augmented the content of these publications with the development of corridors according to the method proposed in [21]. Since the kinematics of the test subjects were the basis for the analyses of the scaling methods discussed in this paper, the corridors developed for the trajectories of the head and spinal segments of the different test subjects are included here as the first results of this investigation. Therefore, the following four analyses were performed in this paper:

- a) Creation of corridors for the trajectories of the head, thoracic spine and pelvis in the sagittal plane (XZ) for each group of test subjects. These corridors are based on the kinematic data published in [16] and [19].
- b) Assessment of two commonly used scaling techniques (mass scaling, SAE scaling) to predict the trajectories of the head and spine in the sagittal plane of a pediatric occupant in a low speed frontal impact using data from adult occupants.
- c) Formulation of a new scaling procedure based on energy to improve the prediction of the previously described trajectories.
- d) Extrapolation of the scaling paradigm to the case of a high speed impact.

All the previous analyses were performed until the time of maximum forward (X) excursion of the head.

III. RESULTS

Displacement corridors of the head, spine and pelvis in the sagittal plane.

The displacements in the sagittal plane of the individual subjects within each group were combined to produce the corridors shown in Figures 4, 5 and 6. The kinematic responses within each group have been thoroughly discussed in [16] and [19]. However, none of the previous publications had addressed the development of corridors for a 50th percentile subject within each group.

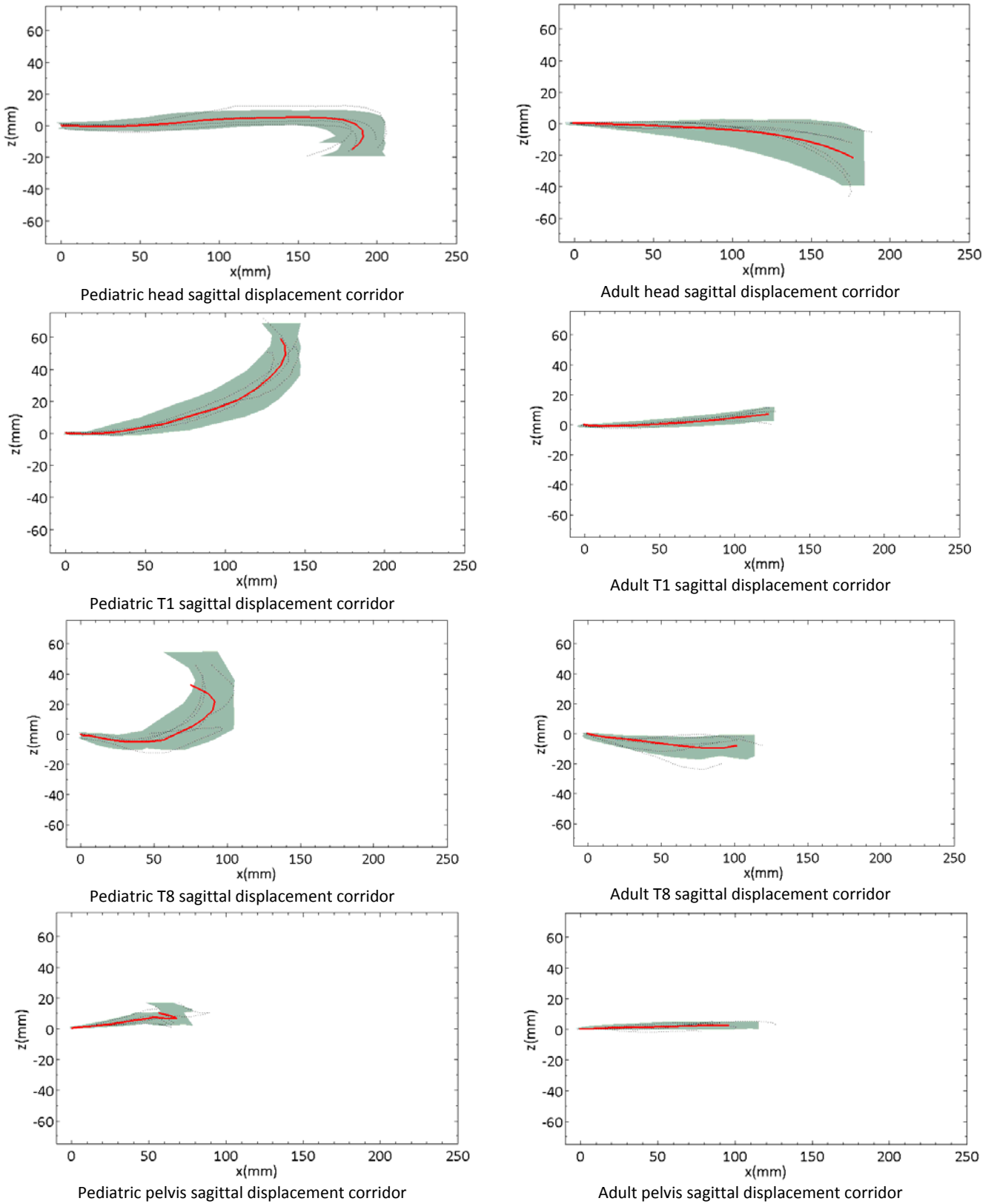


Fig. 4. Comparison of the trajectories and corridors between the pediatric (right) and adult volunteers (left).

Figure 4 compares the response between the pediatric and adult volunteers. The solid line indicates the average displacement expected for a 50th percentile subject within each group (6-YO and adult). The shaded area represents the corridor developed based on the standard deviation in both the X and Z axes. A comparison between the average displacements and the corridors shows that the forward displacement of the tracked segments was similar between the two groups and even slightly greater within the pediatric group in the case of the head and T1. Only the pelvic forward motion is substantially greater in the adults. As for the displacement

in the Z axis, the pediatric spine and pelvis moved superiorly producing a marked curvilinear trajectory of the bony structures. Although a slight vertical component in the +Z axis can be seen in the adult T1 displacement, T8 moved inferiorly instead of superiorly and the adult pelvis remained almost parallel to the X axis throughout its whole motion. In both groups, the head moved initially almost parallel to the X axis and eventually started to undergo a curvilinear trajectory in flexion. The head flexion was more pronounced in the pediatric group.

The differences in the magnitude and nature of the displacements between the pediatric and adult groups have been discussed at length in [16]. The former study indicated that the primary factor governing the differences in the head and spinal kinematics between the age groups was decreasing head-to-neck girth ratio with increasing age, although muscle response and cervical vertebral structural properties are suggested as potential influencing factors as well.

Figure 5 presents the corridors obtained for the head and the spine in the PMHS experiments at 9 km/h. . At 9 km/h, the forward displacements of the head and spinal segments of the PMHS were substantially greater in magnitude than the ones of the adult volunteer group. On the contrary, the volunteers exhibited a greater forward pelvic displacement. All the cadaveric tracked segments exhibited also displacement in the positive Z direction (superior) that was nonexistent in the case of T8 and pelvis in the adult volunteers. On the contrary, the PMHS head and T1 after moving anteriorly and superiorly, started to undergo flexion at both speeds as seen in the adult volunteer group.

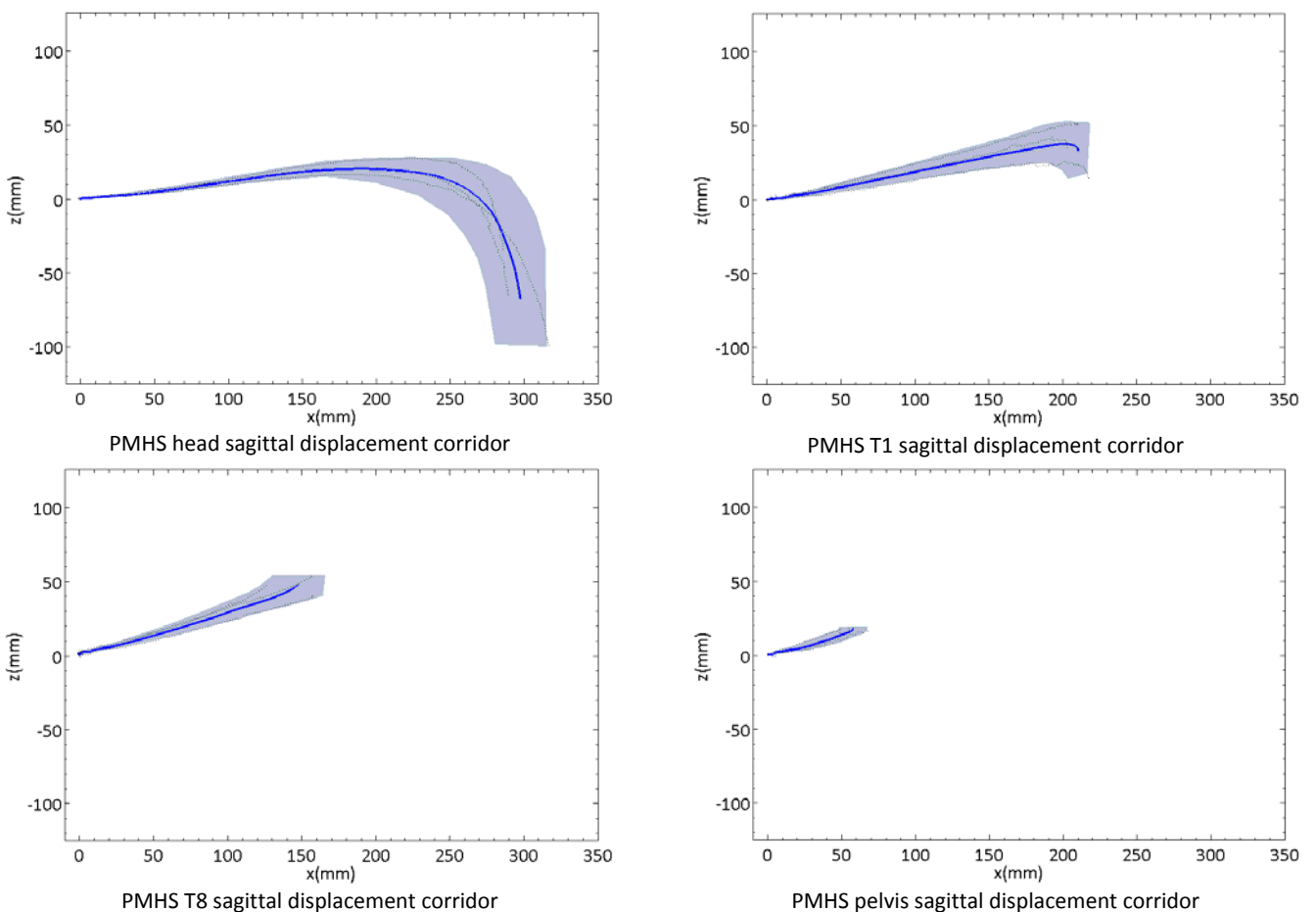


Fig. 5. Normalized trajectories and corridors of the adult PMHS at 9 km/h.

There were not substantial qualitative differences of the PMHS displacements between the two tested speeds. The corridors corresponding to the 40 km/h impact are shown in Figure 6. The magnitude of the anterior displacement of the tracked segments was greater than in the case of the 9 km/h impact, but the nature of the motion did not change, with the exception of that of the pelvis. The vertical component of the displacement was not significantly different from the low speed case.

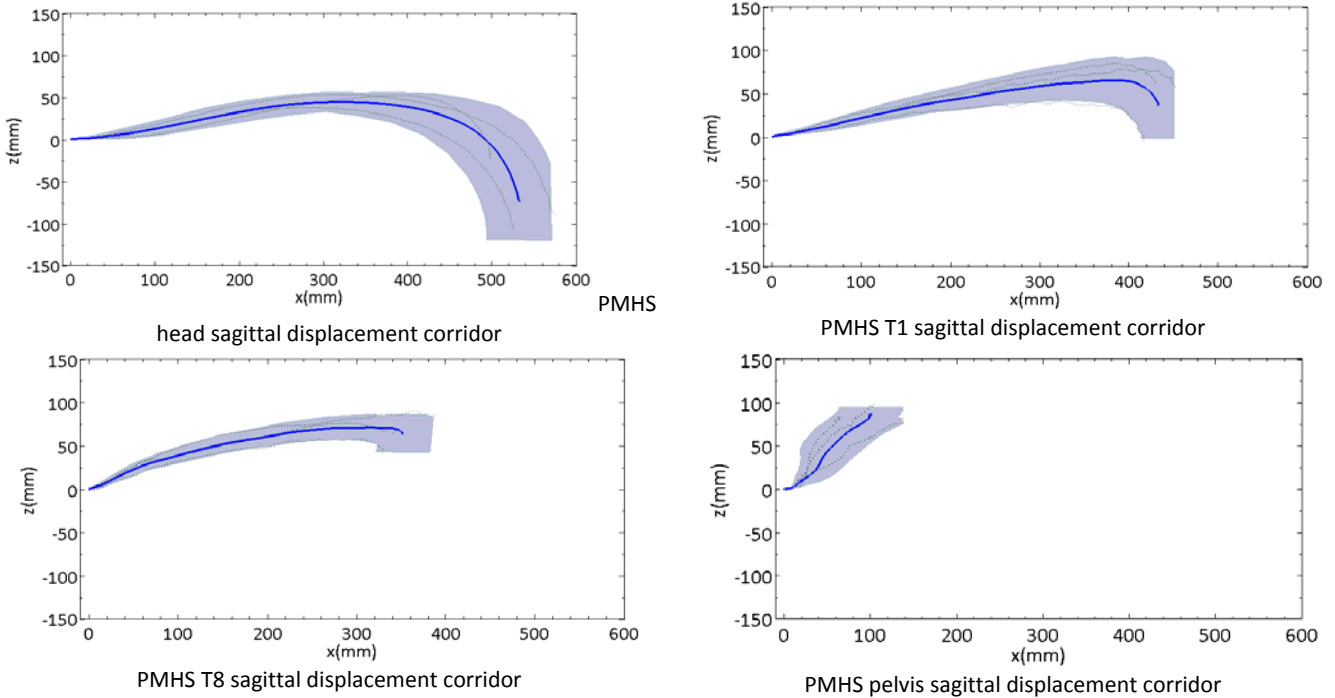


Fig. 6. Normalized trajectories and corridors of the adult PMHS at 40 km/h.

Assessment of scaling methods to predict the pediatric response

To assess the performance of the mass scaling and SAE scaling methods in predicting the kinematic response of a pediatric subject, the adult volunteer data was scaled according to the derivations presented in the methods section. The length-scaling factors showed in Table IV were applied to the X and Z components of the head and T1 displacements of the average 50th percentile adult male. The results corresponding to each method are presented in Figure 7.

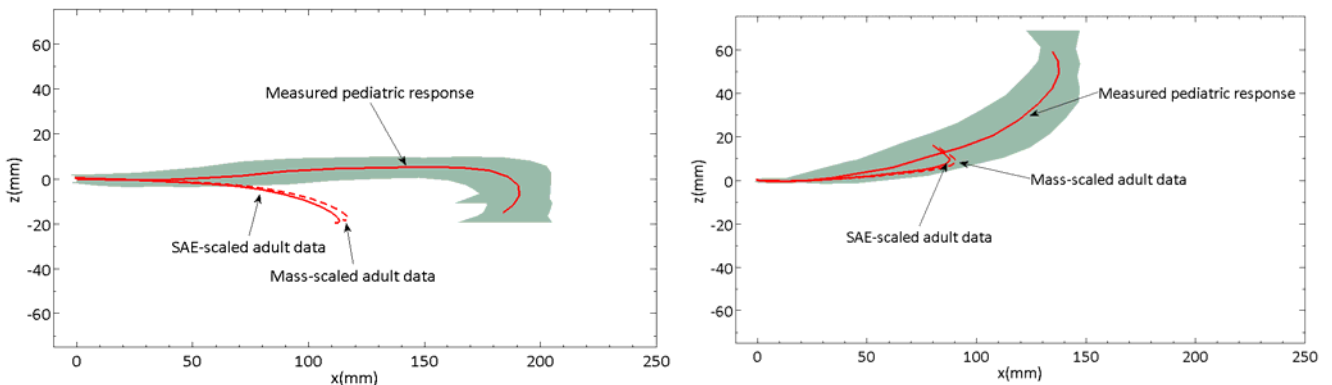


Fig. 7. Prediction of the average trajectories of the head (left) and T1 (right) as given by the mass scaling and SAE scaling methods.

Both methods failed to predict the peak forward excursion of the head of the average pediatric subject. Despite Figure 4 showing that the magnitude of the displacements was similar between the two age groups, the length-scaling factors proposed by either of the two methods were significantly smaller than 1, and therefore the methods under-predicted the forward motion of the pediatric subjects.

New scaling method based on energy. Assessment of the method at low speed.

The proposed scaling method showed in equations (3)-(5) was subjected to the same assessment as the mass scaling and SAE scaling methods. The calculation of the scaling factor between the lengths of the paths followed by the anatomical structure under study (s_{ped} , s_{ad}) required knowing the ratios between the masses of the subjects and the peak belt forces. Reference [16] reported the value of peak belt forces in both groups (610 N

for the adults and 203 N for the pediatric volunteers) and thus, expression (5) can be calculated as follows:

$$s_{ped} = \frac{m_{ped}}{m_{ad}} \frac{F_{ad_{peak}}}{F_{ped_{peak}}} s_{ad} = \frac{23\ 610}{77\ 203} s_{ad} = 0.91 s_{ad} \quad (6)$$

According to equation (6), the length of the path of the pediatric subject was 91% of the length of the adult. The known proportion between the X and Z components of the pediatric displacement were used to calculate the predicted X and Z components of the scaled displacement. Figure 8 shows the comparison between the observed pediatric trajectories and the ones predicted using this method.

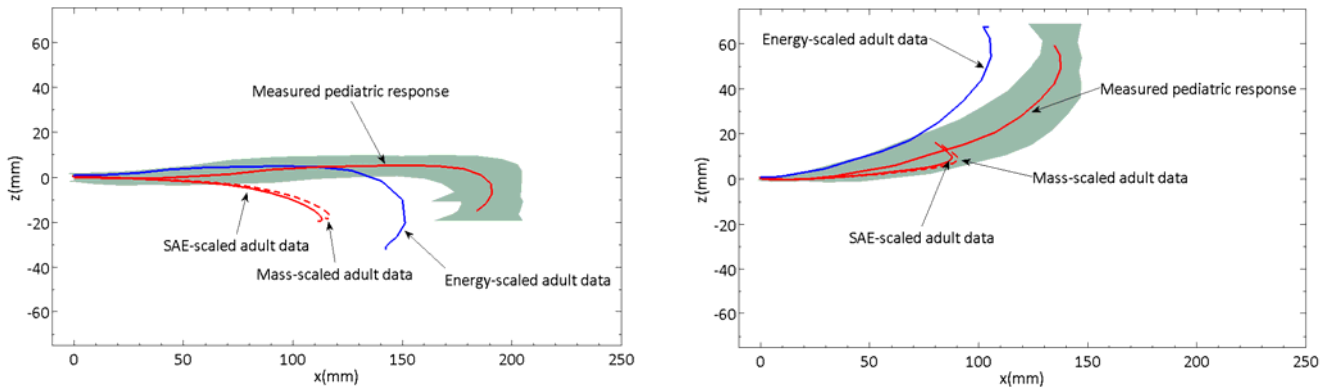


Fig. 8. Prediction of the average trajectories of the pediatric head (left) and T1 (right) as given by the energy based method.

Again the method failed to predict correctly the peak forward displacements of the pediatric head and T1. However the results given by this method improved the ones obtained using either the mass scaling or the SAE scaling method.

Approximation of the displacement of a 6YO in a 40 km/h frontal impact

The last calculation consisted of applying the above explained energy-based method to scale the displacements of the adult PMHS head, spine and pelvis at 40 km/h to provide an approximation of the pediatric trajectories at that speed. Figure 9 shows the results obtained for this approximation.

Two underlying assumptions were used to calculate the scaled adult trajectories. First, it was assumed that the ratio observed in the peak shoulder belt forces between the adult and pediatric volunteers at 9 km/h also held in the case of a higher speed. Second, once the length of the path was scaled as given by equation (6), the X and Z components of the displacements were apportioned using the relation between the X and Z adult PMHS observed in the 40 km/h tests.

IV. DISCUSSION

To our knowledge, the results presented here compare the kinematic responses of pediatric and adult volunteers with adult PMHS for the first time in the literature. The test matrix, test fixture and impact parameters of this study were chosen so that matching pair of tests with different surrogates could be compared.

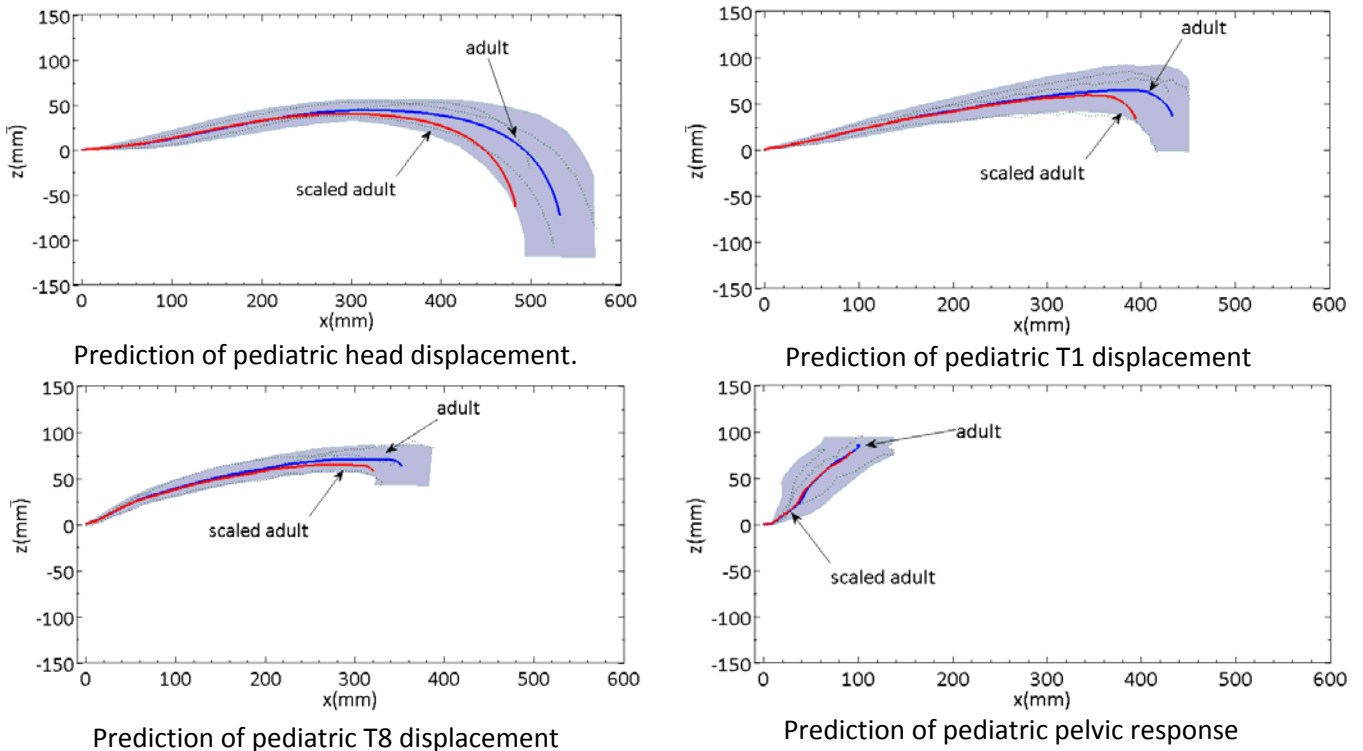


Fig. 9. Adult scaled displacements using the energy method to approximate the displacements of a 6YO in a 40 km/h frontal impact.

Some portions of these tests have been already published. Arbogast et al. published a thorough comparison between the pediatric and adult volunteers [16]. Reference [19] described the corresponding low-speed cadaveric tests as well as the results observed in the 40 km/h tests. This paper consolidates all these results, proposes corridors for a 50th percentile subject within the 6YO and the adult male groups and assesses two commonly used scaling methods. As a result of the performance of both methods, we developed and implemented a new approach. Although it improves considerably the pediatric response prediction over the other two methods, the energy-based method still fails to describe completely the pediatric behavior at 9 km/h. Finally, this method is used to scale the response of adult PMHS in a high-speed frontal impact to approximate the response of a 6YO at high speed.

Caution must be exerted to understand correctly the results provided to describe the kinematics of the head and the spine of a 6YO. First, the approximation given for the 6YO displacements is only valid under the test configuration (seat and belt geometry, belt characteristics, occupant initial position) described with detail in [16] and for the crash pulse of the 40 km/h impact showed in Figure 2. Also, the relationship between the peak belt force experienced by the 6YO subject and that of [19] the adult PMHS at 40 km/h [19] was assumed to be a similar ratio as that found between the pediatric and adult volunteers [16]:

$$\frac{F_{PMHS_{peak}}}{F_{ped_{peak}}} = \frac{610}{203} \approx 3 \quad (7)$$

This assumption was based on the data presented by [25] comparing the response of several dummy sizes at two different speeds (29 and 48 km/h) in the same impact environment. Also, in the prediction of the X and Z components of the trajectories, it has been assumed that the ratio between the two components for the pediatric response at 40 km/h matched that shown by the adult cadavers (i.e. the scaled curve and the original one shared the same shape) at 40 km/h. This is probably questionable looking at differences exhibited by the pediatric and adult volunteers at 9 km/h (specifically T1 and T8), but it was the only possibility of retrieving the trajectories from the path in the absence of any pediatric data at 40 km/h. Another limitation of the study is the absence of muscular activity in the PMHS tests: even if the volunteers were asked to not tense their muscles during the tests, it can be expected some amount of muscle tensing influencing the displacements observed in these tests. Other relevant factors such as belt and seat geometry, and initial position were matched as close as

possible between the two types of surrogates.

The comparison between the pediatric, adult and cadaveric occupants in the 9 km/h tests provided interesting results. Although [16] had already shown that the forward displacement of the head and the spine of the pediatric occupants normalized by seating height were greater than the one of the adults, the corridors provided in Figure 4 show that the forward head and T1 displacement of the average 6YO would be greater than that of the average adult. Whether a bigger sample size in each group would have modified this result is something to be explored in future studies. The kinematics of two pediatric thoracic vertebrae tracked exhibited a curvilinear concave trajectory that was inexistent in the adult volunteers. Similar findings had been discussed also in [16]. As for the kinematics of the PMHS at the same speed, the anterior displacements of these surrogates were consistently greater than those of the volunteers, except in the case of the pelvis. Also the PMHS displacement exhibited a vertical component that, even if it was not as significant as the one observed in the pediatric volunteers, was substantially greater than that of the adult volunteers. These differences between PMHS and volunteers could have been caused by the lack of muscle tone, despite the fact that the volunteers were asked to remain relaxed during the tests.

Although the pediatric group was exposed to a slightly greater acceleration pulse, [16] showed that there was no influence of the differences in delta-v on the increased forward displacement found in the pediatric group. Reference [16] attributed the difference to the decreasing head-to-neck girth ratio with increasing age. Other aspects such as vertebral structural development or muscle activity could also be related but they were not addressed in the original study.

Since the displacements observed in the pediatric group were of similar magnitude (if not greater) than those exhibited by the adults, any scaling method proposing to use a scaling factor significantly smaller than 1 will fail to predict the pediatric response. This was the case of the mass scaling and SAE scaling methods which are based on geometrical considerations. Similar findings in the scaling between pediatric and adult subjects have been reported in the case of the estimation of the thoracic stiffness under belt load [26] as well as on the moment vs. angle characteristic of the thoracic spine [27]. As an alternative method, we explored a scaling relationship based on energy that includes a ratio between the masses of the subjects and a ratio between the peak belt forces. The new method provided a better fit to the actual pediatric data at 9 km/h than the mass scaling and the SAE scaling methods. Thus, this energy-based method was the one chosen to be applied to the PMHS data at 40 km/h to approximate the response of a 6YO in a high speed frontal impact.

As mentioned above, PMHS over-predicted the forward excursion of the volunteers (except for the pelvis) at 9 km/h. Should this be also the case at 40 km/h is unknown because of the lack of data comparing volunteers and PMHS at that speed.

The sets of experimental data discussed in this paper constituted a unique opportunity to assess the capability of different scaling methods to predict the response of pediatric subjects. Also, based on the comparisons between different surrogates at low speed and the corresponding experiments at high speed, an estimation of the displacements of the head, thoracic spine and pelvis of a pediatric occupant in a 40 km/h frontal impact have been calculated. The approximation was subjected to a series of limitations that have been discussed in the paper. Nevertheless, the estimation provided here for the kinematics of a pediatric subject may assist in the development of physical and computational models of a 6YO occupant in high-speed frontal impacts.

V. CONCLUSIONS

Four different experimental data sets corresponding to non-injurious low-speed impacts with pediatric and adult volunteers and to low-speed and high-speed impacts with adult PMHS were combined to assist in the estimation of the kinematic response of a 6YO in a high-speed frontal impact. Corridors showing the average response and standard deviation of the X and Z displacement in the sagittal plane were developed for each type of subject. Two published scaling methods (mass scaling, SAE scaling) were applied to the adult volunteer data and failed to predict the response of the pediatric volunteers at low speed. A new method based on energy conservation that improved the prediction of the pediatric displacement at low speed was proposed. This method was then applied to the response of the PMHS in a high-speed impact to provide an approximation of the head, spinal and pelvic displacement of a 6YO occupant in a 40 km/h frontal impact. The limitations of the scaling method are discussed in the paper.

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