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Novel Methods to Determine Pediatric Anterior-Posterior Thoracic Force-Deflection Characteristics

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ABSTRACT

Pediatric anthropometric test devices (ATDs) are a key tool for developing motor vehicle safety systems. Such ATDs are limited by the adult-derived biofidelity data used to guide their design. A novel force deflection sensor (FDS) for determining applied force and chest deflection during cardiopulmonary resuscitation (CPR) has been developed, but is subject to certain errors associated with deformation of the mattress and backboard upon which the patient lies during CPR. The purpose of this paper is to describe a numerical method for compensating for mattress deformation errors, and to compare the compensated chest deflections with actual measured chest deflection. A series of CPR simulations with manikins on hospital beds and stretchers was performed with the FDS sensor in place and subsequently processed with the mattress compensation technique. Average error between mattress compensated and measured maximum chest deflection was lowest on the stretcher with stretcher pad (0.3 ± 0.9 to 1.5 ± 0.5 mm) compared to two different hospital beds (0.3 ± 1.6 to 5.9 ± 1.6 mm). The methods described and evaluated herein provide a promising approach to obtain thoracic biomechanical data from live children, with the end goal to supply enhanced thoracic biofidelity requirements for development of future pediatric ATDs.

INTRODUCTION

Pediatric anthropometric test devices (ATDs) are a key tool for developing motor vehicle safety systems. Such ATDs, including the Hybrid III family and Q series, are limited by the adult-derived biofidelity data used to guide their design. That is, the design requirements that ensure the child ATD behaves like a human during an impact (generally termed “biofidelity” requirements) are based largely upon scaled data from adult cadaver and adult volunteer impact experiments, and not from pediatric-specific data. Faced with the engineering task of developing pediatric ATD’s, the developers of these scaling techniques used the best available child biomechanical data to develop pediatric ATD biofidelity requirements. More specifically, Irwin and Mertz (1997) used the Kroell thoracic impact tests (Kroell et al., 1974), where an impactor with a constant initial velocity is propelled into the torso of adult post-mortem human subjects (PMHS). The authors applied equations that yield the ratio of the chest deflection and force between the adult and child. Similarly, equations were developed that scale the mass of the impactor used in the test, which are also dependent upon elastic moduli of the thorax, with skull modulus used as a surrogate. The work of Irwin and Mertz forms the basis for the Hybrid III family of child ATD's employed in Federal Motor Vehicle Safety Standard (FMVSS) No. 208 Frontal Impact Protection and No. 213 Child Restraint Evaluation in the United States, and similar standards in Canada. Van Ratingen et al. (1997) developed similar techniques to guide the design of European pediatric ATD thoraces, and employed femur elastic modulus for scaling of thorax forces and deflections. One theme runs throughout the cited scaling papers – thoracic material property parameters figured prominently in the scaling equations, and limited available pediatric data led the authors to use data from other body regions in place of pediatric-specific thoracic material information.

Inspection of the torso maturation process suggests that the scaling techniques described previously could be enhanced if pediatric thorax biomechanical data could be obtained. The sternum consists of 6 main bones – the manubrium superiorly, followed by sternebrae 1 through 4 and the xiphoid process. The 4th sternebra appears at age 12 months, while the xiphoid process appears at 3 to 6 years. Fusing between sternebrae begins at age 4 years and continues through age 20 years. The sternum as a whole descends with respect to spine from birth up until age 2 to 3 years, causing the ribs to angle downward when viewed laterally, and the shaft of the rib to show signs of axial twist (Scheuer and Black, 2000). The costal cartilage also calcifies with age, likely influencing its flexibility. Thus, pediatric thoraces differ from the adult not only geometrically, but materially and structurally as well; these differences likely influence the mechanical response of the child to blunt impact and are not presently considered in current ATD biofidelity data scaling techniques.

Recently, our research facility acquired the necessary equipment for measuring the applied force and sternal deflection of the thorax during Cardiopulmonary Resuscitation (CPR). In brief, a load cell and accelerometer sensor package has been integrated into a clinical monitor-defibrillator (Laerdal Medical, Stavanger, Norway) to measure chest compression and applied force during CPR. This Force-Deflection Sensor (FDS) is interposed between the palms of the hands of the person administering CPR and the sternum of the patient. The accelerometer signal is processed with a double-integration algorithm, yielding deflection.

Thus, the significance of the development of the FDS is that it provides the first means we are aware of to directly measure the force-vs-deflection characteristics of the live pediatric chest at crash-relevant deflections. The implication of knowing the force-vs-deflection characteristics is that it now becomes possible to scale the thoracic deflection response from adult cadaver blunt impact tests (Kroell et al., 1974) or belt loading tests (Kent et al., 2004) based upon thoracic biomechanical data from live children.

Extracting force and chest deflection data from CPR is subject to certain environmental and numerical errors. Our process of deflection calculation is reliant on double-integration, and thus is subject to error accumulation; its accuracy has been previously documented not to exceed 1.3 mm within a 95% confidence interval in manikins compressed on rigid surfaces (Aase and Myklebust, 2002). Chest compressions in the clinical setting are typically performed on a variety of deformable hospital bed or stretcher surfaces with CPR backboards in place. As the accelerometer measures FDS acceleration with respect to ground, as opposed to the spine of the subject, the deflection calculations are a sum of the deformation of thorax and the deformation of the bed/stretcher mattress. The emergent nature of CPR precludes the addition of a spine accelerometer to the patient, which through subtraction of integrated acceleration signals would allow for calculation of chest deflection.

Mattress compensation

We have developed a method to compensate for the mattress deflection during CPR and thus extract an estimate of the true subject chest deflection. Application of this method involves measuring the subject plus mattress deflection with the FDS during CPR. Detailed environmental information is also recorded, including FDS placement on the subject's chest, backboard orientation on the bed, patient orientation on the backboard, bed/stretcher surface type, and subject anthropometry. Within 24 to 48 hours of the CPR event, a CPR Event Reconstruction takes place. The same type of bed/stretcher used during the CPR event is used in the reconstruction, and the aforementioned environmental details are recreated. Then, a CPR Manikin of similar size to the subject of interest is placed on the backboard and chest compressions are performed. The reconstruction data are used to compensate for the mattress deflection. The purpose of this paper is to describe a numerical method for compensating for mattress deformation errors, and to compare the compensated chest deflections with actual measured chest deflection.

METHODS

Numerical Method of Mattress Compensation

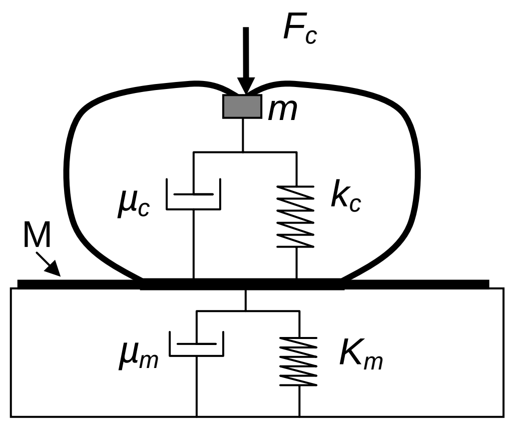

Figure 1: Mechanical model of chest atop compliant bed/stretcher.

Figure 1 shows a schematic of a mechanical model of a chest under compression, when the thorax is placed on a compliant surface such as a mattress of a hospital bed. The equations of motion for the chest and mattress surfaces can be written as

$$m(a_c + a_m) = -k_c x_c - \mu_c v_c + F_c \qquad (1)$$

for the chest surface and

$$Ma_m + m(a_m + a_c) = -k_m x_m - \mu_m v_m + F_c \qquad (2)$$

for the mattress superior surface, where

x_c , v_c and a_c are the displacement, velocity, and acceleration of the sternum relative to the superior surface of the mattress, respectively,

x_m , v_m and a_m are the displacement, velocity, and acceleration of the superior surface of the mattress, respectively,

m is the mass of the part of the torso that moves relative to the mattress surface.

M is the remaining part of the torso mass plus the backboard and half the mattress mass.

k_c and k_m are the stiffnesses of the chest and mattress, respectively,

μ_c and μ_m are the damping constants of the chest and mattress, respectively, and

F_c is the force applied to the sternum (measured by the FDS).

The stiffness (k_m and k_c) and damping (μ_m and μ_c) parameters of the mattress and chest are assumed to be functions of deflection. In general, these functions are assumed to be of the form,

$$k(x) = k_1 + k_2 x \qquad \text{and} \qquad \mu(x) = \mu_1 + \mu_2 x$$

We will assume that the mass of the sternum m is insignificant and thus,

$$Ma_m = -k_m x_m - \mu_m v_m + F_c \tag{3}$$

The total movement of the chest surface (x_t) is measured by the FDS, and is given by

$$x_t = x_c + x_m \tag{4}$$

In order to measure the actual deflection of the chest x_c, we need to quantify and subtract the mattress deflection (x_m) from the measured total movement x_t. x_m is found by rearranging Equation (2) and applying the assumption described in Equation (3).

$$x_m = \frac{F_c - Ma_m - \mu_m v_m}{k_m} \tag{5}$$

In order to solve this equation, we need to determine the stiffness (k_m) and damping coefficient (μ_m) of the mattress, as well as the mass of the patient, backboard, and mattress (M).

Determination of mattress properties

In the CPR event reconstruction, a manikin is placed on the same mattress/backboard as was used during resuscitation. The manikin is loaded to the same weight as the estimated weight of the patient on the backboard, which serves as the weight (M) in Equation 5. Compressions are performed on the manikin chest using the FDS to collect sternal force and total displacements. Unlike the CPR event itself, we are able to attach a reference accelerometer to the spine of the manikin, yielding measured values for x_m, v_m and a_m. By use of these measurements it is possible to calculate the stiffness and damping coefficients of the mattress through Equation 3. The methods for calculating these parameters are identical to those employed by Arbogast et al. (2006) in determining the stiffness and damping coefficients of the chest during CPR on rigid surfaces and are not repeated here. Thus, through the CPR event reconstruction, the stiffness and damping properties of the mattress are determined. Through Equation 5 it is now possible to determine the mattress deflection in the CPR event itself, which will lead to determination of the true chest compression from Equation 4.

Calculation of mattress compression for a CPR event

To determine an estimate of x_m for the resuscitation event, we apply Equation 5, but in this case we use the F_c from the CPR event itself, as measured by the FDS. It is clear that Equation 5 is not directly solvable, since x_m appears on both sides of the equation, and determination of x_m and v_m requires knowledge of a_m. Equation 3 must therefore be solved iteratively. This is done by first calculating a first order depth estimate x_{m1} based on an assumption of zero mattress damping and subject mass:

$$x_{m1} = \frac{F_m}{k_m(x_m)} \tag{6}$$

Damping will mainly cause a phase shift of x_m relative to F_m. To make a first order correction for damping, we calculate the delay of x_m and advance this signal so that it is essentially in phase with F_m. The phase-shifted estimate x_{m1} is now used to calculate v_{m1} and a_{m1} by time differentiation. Equation 5 is then solved again to calculate a second order estimate for mattress compression depth. To enhance the accuracy, the process can be repeated.

Compensation evaluation

Having described the theory behind mattress compensation, the work in this paper focuses on the objective evaluation of the mattress compensation method. To evaluate the above method for mattress compensation, several manikin experiments were conducted with the FDS. The purpose of these experiments was to perform "CPR" measurements on a manikin, measure the actual chest deflection in the manikin by using a reference accelerometer, and compare this to the chest deflection estimated using the FDS and methods described in the theory section above. For these experiments, three beds were tested – a Steris Emergency Department Stretcher with mattress (Stretcher) (STERIS Corporation, Mentor, Ohio) a manual hospital bed with maxifloat mattress (Bed #1) (Hill-Rom Batesville, Indiana), and a Hill Rom Advanta ICU Bed with maxifloat mattress (Bed #2) (Hill-Rom Batesville, Indiana) (Table 1). These mattress/beds represent a typical range of surfaces one might encounter in clinical CPR. For each stretcher/mattress surface type, two manikins were tested – one Original Resusci-Anne (OMan) manikin and one slightly modified version of this manikin (ModMan), adapted to measure simultaneous compressions and ventilations. The two manikins are both adult thorax-sized manikins designed and manufactured by Laerdal, and have approximately the same chest stiffness. The manikin subject was placed atop a CPR backboard on the mattress surface. 50 compressions were performed while logging the compression force and movement of the chest surface by the FDS. In addition to the FDS, a second reference accelerometer was placed on the spine plate of manikin at the approximate location of T5 to track the compression of the mattress through double integration.

Table 1. Test matrix showing number of compressions delivered to two different manikins and three surface types. All tests were conducted with a CPR backboard under the manikins.

Number of Compressions		**Manikin**	
		Modified Resusci-Anne Manikin (ModMan)	**Original Resusci-Anne Manikin (OMan)**
Surface Type	**Stretcher**	50	50
	Bed #1	50	50
	Bed #2	50	50

To assess the accuracy of the mattress compensation method, the *estimated* chest deflection was determined by the method outlined in the Numerical Method section of the Introduction to this paper. The *measured* chest deflection was calculated by first double integrating the FDS and reference accelerometer signal, yielding their respective displacements. The reference displacement was then subtracted from the FDS displacement, yielding the measured chest deflection. Two methods were employed to calculate the maximum deflection of any given compression cycle. The *max deflection* is the maximum displacement during a compression cycle. The *max-to-min deflection* is the difference in displacement magnitude between the maximum of each compression cycle minus the previous minimum (Figure 2). The error in millimeters (mm) between the measured and calculated maximum deflections for each compression cycle was then determined, and then averaged for all 50 compressions, yielding the average error. Also calculated was the average maximum chest and mattress deflection across all 50 compression cycles.

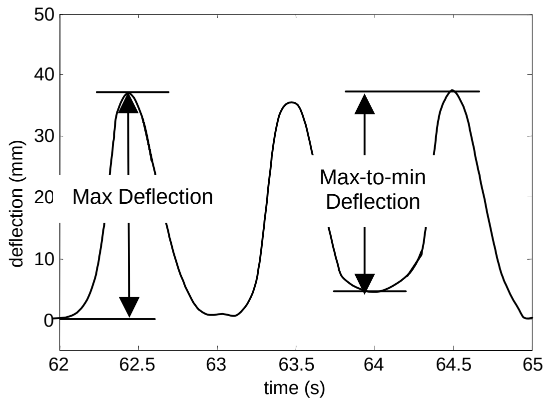

Figure 2: Methods of calculating maximum deflection, and estimated and measured deflection. Max deflection is simply the magnitude of deflection. Max-to-min deflection is calculated as the maximum of each wave minus the previous minimum.

RESULTS

Mattress and chest deflection time-histories exhibited sinusoidal shapes (Figure 3). Measured average maximum chest deflection delivered to the manikins ranged from 36.2 ± 4.2 to 47.4 ± 2.5 mm, suggesting loading magnitude consistency between manikins.

The stretcher mattress pad exhibited the lowest mattress compression (8.5 ± 2.2 to 8.7 ± 2.1 mm for the two manikins) compared to Bed #1 and Bed #2 (22.8 ± 7.7 to 25.8 ± 8.0 mm), as seen in Table 2. Of note, average error between estimated and measured max-to-min chest deflection was lowest on the stretcher (0.9 ± 0.8 to 1.5 ± 0.5 mm) compared to the beds (3.8 ± 1.4 to 5.9 ± 1.6 mm). Estimated chest deflections were more accurate at maximum value than at minimum value for Bed #1 and Bed #2. (Figure 3 – right.) This is reflected in the maximum chest deflection data (Table 2), where the Max-to-min deflection measurement methods showed higher error (0.9 ± 0.8 to 5.9 ± 1.6) than the Max error (0.3 ± 0.9 to 2.1 ± 2.1).

Table 2. Average ± standard deviation of measured maximum deflection for chest and mattress, and average maximum chest deflection error, stratified by surface type and manikin. Both Max-to-min and Max deflection calculation techniques are employed as in Figure 2.

			Average Max-to-min Measured Deflection		**Average Maximum Chest Deflection Error**	
			Chest	**Mattress**	**Max-to-min**	**Max**
		Surface Type	**(mm)**	**(mm)**	**(mm)**	**(mm)**
Manikin	**SCV**	**Stretcher**	43.2 ± 2.7	8.5 ± 2.2	1.5 ± 0.5	0.6 ± 0.6
		Bed #1	40.8 ± 2.3	23.2 ± 4.5	4.3 ± 1.6	1.7 ± 1.4
		Bed #2	41.1 ± 2.6	25.8 ± 8.0	4.3 ± 1.5	2.1 ± 2.1
	RA	**Stretcher**	47.4 ± 2.5	8.7 ± 2.1	0.9 ± 0.8	0.3 ± 0.9
		Bed #1	37.5 ± 5.0	22.8 ± 7.7	3.8 ± 1.4	0.4 ± 1.2
		Bed #2	36.2 ± 4.2	23.6 ± 4.7	5.9 ± 1.6	0.3 ± 1.6

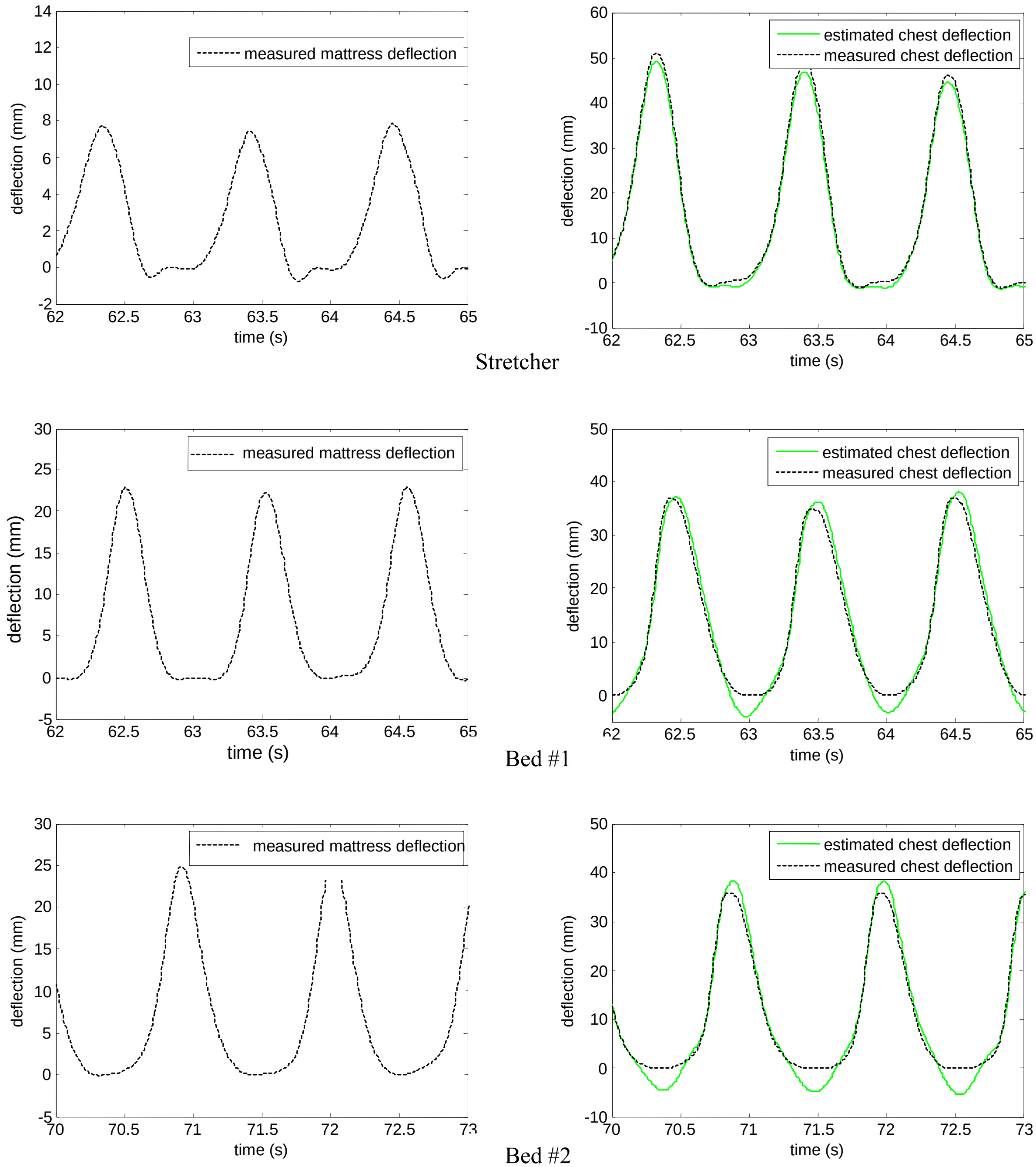

Figure 3: Exemplar deflection time-histories for the mattress and manikin chest. Manikin chest deflections (right panels) are shown with estimated deflections overlayed with measured deflections. Zero deflection indicates the resting position of the chest or mattress in the absence of any applied CPR force. Positive deflection values indicate mattress or chest compression.

DISCUSSION

Thoracic biofidelity of pediatric ATDs is based upon scaled adult thoracic impact data with minimal consideration for developmental changes during the human maturation process. The methods described and evaluated herein provide a promising approach to obtain thoracic biomechanical data from live children, with the end goal to supply enhanced thoracic biofidelity requirements for development of pediatric ATDs.

Like any experimental technique, our methods are subject to certain quantifiable errors in measurement accuracy. Because it is an inertial device, the FDS sensor alone is unable to distinguish thoracic compression and compression of the surface on which the subject lies. In our case, that surface is the mattress of a stretcher or bed, which compresses during CPR even when a conventional CPR backboard is used under the patient. We have described and evaluated a mattress compensation method, whereby with certain assumptions the chest deflection can be estimated from chest plus mattress deflection data. Our evaluation of these methods showed that the accuracy of the mattress compensation method was highest when CPR is administered on the least compliant of surfaces; the thin stretcher mattress pad, where mattress compression averaged 8.5 to 8.7 mm, yielded chest deflection estimation errors ranging from 0.9 mm to 1.5 mm. The thicker hospital bed, which compressed 22.8 to 25.8 mm, yielded errors ranging from 3.8 mm to 5.9 mm. Examination of the exemplar deflection time-histories (Figure 3) reveals the source of error in the bed surface tests – a pronounced deviation of the estimated deflection from the measured deflection near the zero-point of each compression cycle. The cause of the deviation is not clear, but perhaps further refinement of the chest model (Figure 1) will improve the accuracy of the estimated deflection.

One potential application of the data lies in the development of improved ATD biofidelity requirements. As discussed in the introduction, Irwin and Mertz (1997) developed impact corridors for pediatric ATDs. For the chest, the authors used the Kroell thoracic impact tests, where an impactor with a constant initial velocity is propelled into the torso of post-mortem human subjects (PMHS). The authors applied equations that yield the ratio of the chest deflection and force between the adult and child,

$$R_D = \lambda_x \qquad \text{and} \qquad R_F = \lambda_E \lambda_x \lambda_z$$

where λ_E is the ratio of the elastic moduli of skull bone, and λ_x and λ_z are the ratios of the characteristic lengths in the x and z directions, respectively. The ratio of R_F to R_D is the stiffness scaling factor for the chest that is based upon the modulus of skull bone. When applied to real children and adults, the FDS sensor has the potential of providing stiffness scaling factors for the whole chest of the adult and child subject. The adult data has already been presented by Arbogast et al. (2006), and data collection with the FDS on children is underway at The Children's Hospital of Philadelphia.

The methods and data presented are subject to certain limitations. First, the mattress compensation method validation was performed on manikins, yet the method itself is intended for use on humans. If the manikin produces some mechanical artifact that is not present in the patient, it is possible the accuracy of our methods would be reduced. Second, we have compared our estimated chest deflection to measured chest deflection, however the measured chest deflection was based upon double integration of accelerometer signals, and may be subject to numerical errors in the integration process. However, deflection validation testing on rigid surfaces has found the sensor to work reasonably well (Aase and Myklebust, 2002). Finally, our data will be collected on thoraces of patients in a hospital and may have certain thoracic structural abnormalities caused by disease or injury. The effect of these abnormalities on thoracic stiffness needs to be considered in future analyses.

CONCLUSIONS

A novel method for directly determining the force-deflection characteristics of the pediatric chest has been developed, using a model of CPR chest compressions with mattress compression compensation. Measurement errors ranged from 0.9 ± 0.8 mm to 5.9 ± 1.6 mm, with higher errors on more compliant mattress surfaces. When applied to children, these measurement techniques show great promise for collecting data to guide development of future pediatric ATD thoraces.

ACKNOWLEDGEMENTS

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DISCUSSION

PAPER: **Novel Methods to Determine Pediatric Anterior-Posterior Thoracic Force-Deflection Characteristics**

PRESENTER: ***Matt Maltese, The Children's Hospital of Philadelphia***

QUESTION: *Guy Nusholtz, Daimler Chrysler*
Matt, I noticed that you're going to about 60 mm on some of the compressions. How close to the spine are you getting? And, is there any indication of rib fracture or damage with that much compression? I know in adults we always get ribs breaking when we do CPR, but what about children?

ANSWER: So, you're talking about the one subject that I put up there?

Q: Yeah, I looked at the one subject you got close to 60 mm.

A: Yeah.

Q: That looks like a lot of compression.

A: That is a lot of compression. That subject was adult-sized. It was a 16 year-old.

Q: Oh, okay.

A: So, that would explain--

Q: That part—That part explains it.

A: And we don't have information on the rib fractures. This happened last Sunday.

Q: Thank you.

Q: *Joe McFadden, Vehicle Research and Test Center*
I was wondering if you're scaling for velocity at all? I didn't notice that in your equation, whether you talked about that.

A: So, you're talking about this?

Q: Yeah.

A: So this is a re-publishment—Is that a word?—re-presentation of what Mertz did and he didn't scale for velocity. He assumed the same velocity, effective velocity in his scaling methods and just dropped down the mass and tension.

Q: Okay. Will that be something you'll have to look at if you're trying to relate that to stiffness in a car impact situation?

A: So, I said before that if we have—Our data is being collected at approximately 2/10ths of a meter per second and the cars are at 4 meters per second. So yeah, there has to be something done to extract out the stiffness of the chest and try to get a sense of how it performs. We get a sense of how it performs statically. We have to do some translation to it to get a dynamic stiffness of it. That's where we'll test.

Q: Thank you.

A: At the static rates.