SCALING METHODS APPLIED TO THORACIC FORCE DISPLACEMENT CHARACTERISTICS DERIVED FROM CARDIOPULMONARY RESUSCITATION

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ABSTRACT

Motor vehicle crashes are the leading cause of death for children and adults for every year of age from 3 to 36 years in the United States. Anthropomorphic Test Devices (ATDs) and computer models are key tools for evaluating the performance of motor vehicle safety systems, yet current data available for the validation of pediatric ATDs and computer models are derived from adult data through scaling or from sparse PMHS experiments. Recent measurement of large datasets of cardiopulmonary resuscitation (CPR) on children and adults provides valuable information for validating the aforementioned models. Thus, the objective of this work was to: a) evaluate the changes in the elastic force-displacement properties of the chest across the pediatric and young adult age range, and b) apply three published methods to estimate the composite modulus of the chest and scale the elastic force-displacement properties of the 8 to 10 year old to the 6 year old. In general, the data show a gradient of increasing stiffness (i.e. higher force at any given displacement) with age. CPR subjects in the 20 to 22 year old and 17 to 19 year old age ranges showed similar force-displacement behavior as did subjects in the 11 to 13 and 14 to 16 year old age ranges. The scaled elastic force-displacement curves for the 6 year old were quite similar for the femur and skull based modulus, but the CPR based curve was lower in stiffness. Elastic force-displacement properties for chests of subjects 8 to 22 years old are provided, along with similar data for 6 year old subject scaled from 8 to 10 year old subjects. These data are useful for validation of ATDs and computer models of the human pediatric chest.

INTRODUCTION

Motor vehicle crashes are the leading cause of death for children 5 to 14 years old in the United States (Xu et al. 2010). The Anthropomorphic Test Device (ATD) is a key tool for the evaluation and optimization of automotive restraint systems for occupant protection. The ATD thorax interacts with the restraints within the vehicle, and must do so in a biofidelic or human-like manner to ensure that restraint designs protect humans. More recently, computer models of the human chest have been developed and require data for validation of their force-displacement.

Post-mortem human subjects (PMHS) are the common surrogate used to represent live adults in biomechanical testing to validate ATD and computer model biofidelity. Owing to the paucity of pediatric PMHS, previous researchers have scaled adult impact data to estimate the response of the child subject (Irwin and Mertz 1997; Van Ratingen et al. 1997). More recently, pediatric PMHS have become available (Ouyang et al. 2006; Kent et al. 2009) but the number of subjects at a single age range is quite limited.

Recent measurement of large datasets of CPR events on children and adults provides valuable information for validating the aforementioned models. CPR, which involves the displacement of the sternum toward the spine to induce cardiac blood flow, provides a means to conduct mechanical "testing" of the human chest. Various electro-mechanical devices have been developed over the past three decades to improve the quality of CPR and to study the effect of CPR mechanics on clinical outcomes (Tsitlik et al. 1983; Gruben et al. 1990; Aase and Myklebust 2002). Recently, these devices have been extended to the pediatric and young adult population (Maltese et al. 2008; Sutton et al. 2009) and provide for the measurement and recording of the forces applied to the sternum during CPR and the calculation of resulting deformation (sternal displacement) of the chest though integration of accelerometer data. These CPR studies directly assess thoracic stiffness using human subjects, and can provide valuable guidance for the design and performance certification of the human surrogates such as finite element computer models or physical ATDs.

Thus, the objective of this work was to: a) evaluate the changes in the elastic force-displacement properties of the chest across the pediatric and young adult age range, and b) apply three published methods to estimate the composite modulus of the chest and scale the elastic force-displacement properties of the 8 to 10 year old to the 6 year old.

METHODS

Thirty-nine CPR events from three data sources were used to develop the 6 year old elastic force-displacement characteristics. Four CPR events reported by Tsitlik et al (Tsitlik et al. 1983), 18 events reported by Maltese (Maltese et al. 2008), and 17 new events collected by the Children's Hospital of Philadelphia and analyzed using methods previously reported (Maltese et al. 2008) were gathered into a single dataset (Table A1). The research reported herein was approved by the Institutional Review Board of the Children's Hospital of Philadelphia.

Displacement (x) as used here is defined as the motion of the sternum in the anterior-posterior axis of the chest, with respect to the thoracic spine. Displacement is defined as positive when the sternum moves toward the spine. Force (F) as used here is the force applied to the sternum in the anterior-posterior direction. A positive force moves the sternum closer to the spine.

Each subject's chest was modeled with a spring and damper in parallel; both the stiffness and damping coefficients were linearly dependent upon displacement. Thus, the relationship between force and displacement for the spring (elastic) component of the model is,

$$F_e = xa_1 + x^2 a_2 \tag{1}$$

where a_1 and a_2 are the stiffness coefficients of the chest as previously defined by Tsitlik (Tsitlik et al. 1983) and (Maltese et al. 2008). For each patient, F_e vs. x was plotted, stratified into the following age groups: 8 to 10, 11 to 13, 14 to 16, 17 to 19, and 18 to 22 years, thus accomplishing our first objective of evaluating the elastic force-displacement properties of the chest across the pediatric and young adult age range. In each age group, mean and standard deviation corridors were drawn.

Scaling

Using model theory (Langhaar 1951; Eppinger et al. 1984), mechanical scale factors were developed for systems of varying size and modulus of elasticity, but similar shape and density. Given two systems (labeled 1 and 2 in the equations below) of differing size and elastic modulus, scale ratios (λ) can be written for modulus (E), density (ρ) and characteristic length (L),

$$\lambda_\rho = \frac{\rho_2}{\rho_1} \qquad \lambda_E = \frac{E_2}{E_1} \qquad \lambda_L = \frac{L_2}{L_1} \tag{2}$$

Assuming the chests have similar composite densities ($\lambda_\rho=1$), by dimensional analysis scaling relationships for displacement, force and stiffness can be written as,

$$\lambda_x = \lambda_L \qquad \lambda_F = \lambda_L^2 \lambda_E \qquad \lambda_K = \lambda_E \lambda_L \tag{3}$$

Where the subscripts 1 and 2 refer to system 1, the system being scaled from (the 8 to 10 year old in this case), and system 2, the system being scaled to (the 6 year old in this case).

Inspection of equation (3) reveals that λ_L and λ_E are both unknown. λ_L was determined from mean anterior posterior (AP) chest dimension data from a robust, population-representative study of pediatric anthropometry (Snyder et al. 1975). From the aforementioned study, the AP chest dimension was found to be 133 mm for the 6 year old. For the 8 to 10 year old, the AP chest dimension of the 9.5 year old was used, which was 148 mm.

The composite modulus of the chest (λ_E), was determined by three different methods. First, skull (parietal) bone modulus data of Thibault et al. (1999), Margulies and Thibault (2000), McPherson and Kriewall (1980) and Hubbard (1971) as reported by Ivarsson (2004) was used as a surrogate for the composite modulus of the chest. Using the exponential equation fit to the skull bone modulus data reported by Ivarsson, the modulus with respect to age equation is,

$$E_{skull} = 1.7373 + 7.2928(1 - e^{-0.17003age}) \quad (4)$$

From equation 4, we found the composite modulus of the 6 year old (6.65 GPa) and for the 8 to 10 year old (7.58 GPa). (9.5 years was used as the age variable for the 8 to 10 year old in Equation 4.) Thus,

$$\lambda_{E-skull} = \frac{6.4\ GPa}{7.58\ GPa} = 0.84 \quad (5)$$

Similar to the skull bone modulus method, the second method used the femur bone modulus as a surrogate for the composite modulus of the chest. The following equation based upon work by Curry and Butler (1975) as reported by Ivarsson (2004) was used,

$$E_{femur} = -0.0029316age^2 + 0.28851age + 8.3468 \quad (6)$$

Based upon equation 6, the composite modulus based upon femur data for the 6 year old was 9.97 GPa and for the 8 to 10 year old was 10.82 GPa (age = 9.5 years was used for the 8 to 10 year old).

$$\lambda_{E-femur} = \frac{9.97\ GPa}{10.82\ GPa} = 0.92 \quad (7)$$

The third method of determining λ_E was based upon extrapolation of CPR data in the 8 to 22 year old age range to the 6 year old. Briefly, the stiffness of the chest (k) in the 8 to 22 year old age range was determined by (Maltese et al. 2010) to be,

$$k = \frac{16.7(age) - 54}{L(0.15)} \quad (8)$$

where L_c is the anterior-posterior chest dimension for the subject. Writing the ratio of stiffness between systems 1 and 2,

$$\lambda_K = \frac{16.7(age_2) - 54}{16.7(age_1) - 54}\frac{L_1}{L_2} = \frac{16.7(age_2) - 54}{16.7(age_1) - 54}\left(\frac{1}{\lambda_L}\right) \quad (9)$$

and then incorporating the equation for λ_k in (3) above,

$$\lambda_E = \frac{16.7(age_2) - 54}{16.7(age_1) - 54}\left(\frac{1}{\lambda_L^2}\right) \quad (10)$$

From Equation 10, using the AP chest dimension of 133 mm for the 6 year old, and 148 for the 9.5 year old, the composite modulus scale factor for CPR is

$$\lambda_{E-CPR} = 0.55 \quad (11)$$

The three moduli of elasticity ($\lambda_{E\text{-skull}}$, $\lambda_{E\text{-femur}}$, and $\lambda_{E\text{-CPR}}$) were each applied in equation (3) above, yielding three force-displacement curves for the 6 year old.

RESULTS

Figures 1 through 5 show the force-displacement curves for the CPR subjects in the 8 to 10, 11 to 13, 14 to 16, 17 to 19, and 20 to 22 year old age ranges, along with the mean and standard deviations.

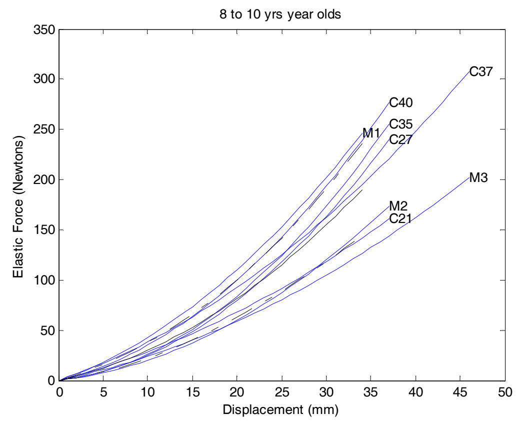

Figure 1 –Elastic force-displacement for the 8 to 10 year old subjects. Refer to Appendix Table A1 for data source for each curve. Dashed solid black line is the mean while the dashed black line is the standard deviation.

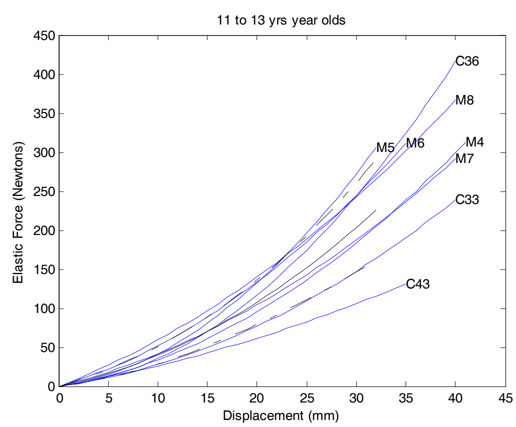

Figure 2 –Elastic force-displacement for the 11 to 13 year old subjects. Refer to Appendix Table A1 for data source for each curve. Dashed solid black line is the mean while the dashed black line is the standard deviation.

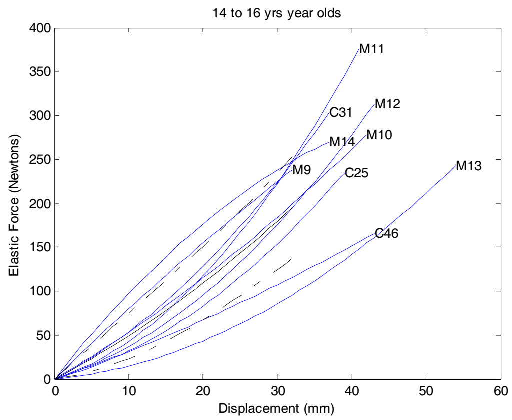

Figure 3 –Elastic force-displacement for the 14 to 16 year old subjects. Refer to Appendix Table A1 for data source for each curve. Dashed solid black line is the mean while the dashed black line is the standard deviation.

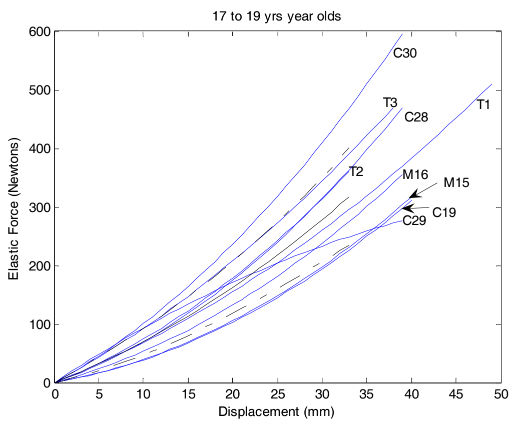

Figure 4 –Elastic force-displacement for the 17 to 19 year old subjects. Refer to Appendix Table A1 for data source for each curve. Dashed solid black line is the mean while the dashed black line is the standard deviation.

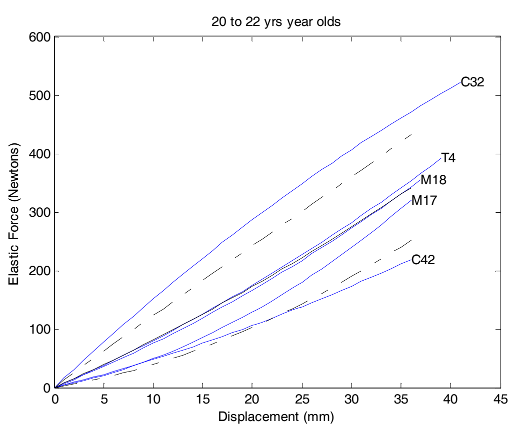

Figure 5 –Elastic force-displacement for the 20 to 22 year old subjects. Refer to Appendix Table A1 for data source for each curve. Dashed solid black line is the mean while the dashed black line is the standard deviation.

Figure 6 shows the mean elastic force for each of the age groups, as well as the scaled elastic force for the 6 year old using the three (femur, skull, and CPR) methods for finding the composite modulus of elasticity of the chest. In general, there is a gradient of increasing stiffness (i.e. higher force at any given displacement) with age. However, CPR subjects in the 20 to 22 years and 17 to 19 years age ranges showed similar force-displacement behavior as did subjects in the 11 to 13 and 14 to 16 year old age ranges.

Scaling of the 8 to 10 year old data to the estimate the 6 year old force-displacement based on femur and skull bone modulus produced similar curves with forces of 137 and 126 N at 30 mm displacement, respectively. However, scaling based on extrapolating the 8 – 22 year old CPR stiffness data produced significantly lower force of 81 N at 30 mm displacement. These differences are proportional to the differences in λ_E values in equations 5,7, and 11.

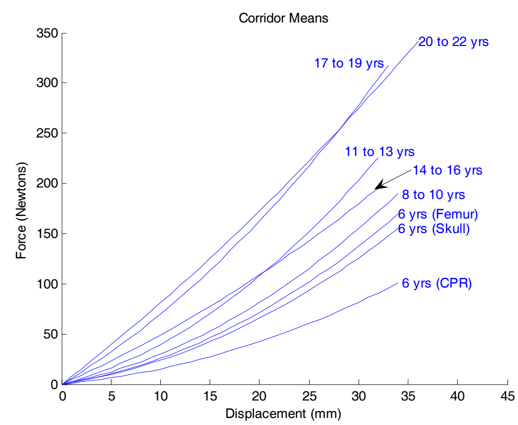

Figure 6 –Mean elastic force-displacement for the CPR subject age groups, and scaled data force-displacement for the 6 year old using the composite modulus of elasticity derived from skull, femur, and CPR.

Table 1 – Force (Newtons) at select displacements (mm) for Femur, Skull, and CPR-based Corridor means for the 6 year old (based upon curves in Figure 6).

Cell value is Force (N)		λ_E Data Source		
		Femur	Skull	CPR
Displacement (mm)	10	26	24	15
	20	72	66	42
	30	137	126	81

DISCUSSION

This paper presents average elastic force-displacement curves for the human chest in the age range of 8 to 22 years, based upon data collected during CPR. Understanding the force-displacement data of pediatric human subjects is an essential first step for validating ATD and computer models of the human body. This paper also employs model scaling laws to estimate the force-displacement of the chest for subjects close to, but outside, the age range of the CPR data collected which offers a potential improvement over current scaling methods that scale data from the elderly adult to the child.

The scaled estimates for force-deflection response of the 6 year old vary significantly depending on the method used. The force-deflection and stiffness estimates based on skull and femur modulus are roughly 50% higher than stiffness derived through extrapolating the CPR data. Until CPR stiffness data is available for ages overlapping and including the 6 year old, computer model validation could consider utilizing a force-deflection curve that falls roughly midway between the femur modulus and CPR data extrapolation based curves.

The data show a general increase in the force-displacement properties of the chest with age, though between the ages of 17 and 22 years old and 11 and 16 years old the force-displacement curves were quite similar. It is reasonable to assume that the mechanical properties of the chest of a 17 year old would be similar to that of a 22 year old, given that the chest has achieved adult size at that age. Conversely, the similarity observed in the force-displacement curves of the 11 to 16 year old age range cannot be explained by size similarity (since the chest is growing larger during this time period). Rather, inspection of the torso maturation process reveals dramatic material and morphological changes in this age range. Fusing between sternebrae begins at age 4 years and continues through age 20 years (Scheuer and Black 2000). The costal cartilage also calcifies with age, likely influencing its flexibility. The timing of these tissue changes during maturation may vary between subjects of the same age. Thus, age may not be the only explanatory variable for changes in thoracic stiffness during development, and another metric of skeletal maturity should be developed and employed.

The magnitude of chest deformation during CPR is in the range of relevance for motor vehicle crashes. At the time of data collection in this study, clinical resuscitation guidelines prescribed targets for CPR chest compressions: 38 to 51 mm of sternal displacement for the adult, or one-third to one-half the anterior-posterior (AP) chest depth for the child (American Heart Association 2005). In terms of displacement, the CPR compression target for the 6 year old child is 47 to 72 mm, assuming an AP chest depth of 143 mm (Irwin and Mertz 1997). For comparison with the general chest displacements observed in impact experiments, chest displacements in hub impact testing with PMHS range from approximately 50 to 70 mm in adults (Lobdell et al. 1973) and from 31.5 to 73 mm overall deflection in children (Ouyang et al. 2006). Thus, chest compression magnitudes during CPR are similar to motor vehicle crash (MVC) events, however CPR rates of chest compression (0.25 m/s) (Maltese et al. 2008) are an order of magnitude lower than those observed in belt loading thoracic compression experiments with PMHS (1 to 2 m/s)(Kent et al. 2004). The low rate of compression during CPR is an advantage for determining thoracic stiffness, since the inertial forces are negligible and viscous forces are quantifiable (Bankman et al. 1990).

LIMITATIONS

It is important to note that CPR, unlike a car crash, loads the chest repeatedly (often hundreds of cycles). The consequences of this repeated loading are not fully understood. It is possible that the chest changes stiffness as the CPR chest compressions continue, though our current data do not show any consistent trend of increasing or decreasing stiffness.

The scaling laws used herein are subject to certain limitations as result of the assumptions made in their application. First, the composite modulus scale factor was applied solely in the anterior-posterior direction, which is the primary direction of chest compression during CPR and during frontal crashes. However, in doing so we assumed that the chest dimension scales equally in all directions ($\lambda_L = \lambda_x = \lambda_y = \lambda_z$) which most likely is not the case.

We presumed that skull and femur bone moduli were suitable surrogates for the composite modulus of the chest. Other researchers have used skull bone modulus as a material modulus for a component of the chest (i.e. rib) and then determined the composite stiffness scale factor based on a rib-hoop under compression using seated height as the length scale dimension (Irwin and Mertz 1997). Of note, aside from the choice of length scale dimension, our derived equation for chest stiffness (Equation 3) was the same as that which was derived by Irwin and Mertz.

The force-displacement data presented here are intended to be used to validate models subjected to *low-rate* loading. It is clear that the inertial and viscous properties of the human chest during high-rate loading will generate forces that are orders of magnitude higher than what we report herein. However, the value of our data is that we provide force-displacement data of subjects very close to the age range of interest (6 years), and data on such young subjects is limited in the literature (Ouyang et al. 2006; Kent et al. 2009).

Similarly, while the data presented herein are theoretically applicable to both physical ATDs and computer models, application of the data to the ATD is currently limited by the ability of the ATD to accurately represent both low- and high-rate loading conditions. Indeed, current ATDs are validated exclusively in the dynamic range similar to car crashes and, although validating the ATD to low-rate data presented here would expand the applicability of the ATD, such expansion may not be possible given the materials currently used to construct the ATD and the cost and durability requirements.

CONCLUSIONS

Herein we have provided low-rate thoracic force-displacement properties for subjects from 8 to 22 years old, and provided estimates for the 6 year old subject. The force-deflection or stiffness estimates for the 6 year old subject based on skull and femur modulus are roughly 50% higher than stiffness derived through extrapolating the CPR data. These data are useful for validation of computer models of the human pediatric chest.

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APPENDIX

Table A1 – CPR event data.

Event	Age	sex	x_{max}	a_1	a_2
	years		mm	N/m	N/m^2
M1	8	F	34	3062	122712
C35	8	F	37	1018	158887
C40	8	F	37	2026	147394
M2	9	F	37	1009	99622
C21	9	F	37	2284	55657
C27	9	F	37	1036	146994
M3	10	F	46	1839	55344
C37	10	M	46	3042	78972
M4	12	M	41	2019	136757
M5	12	M	32	1714	245459
M6	12	M	35	3699	147971
M7	13	F	40	3119	104030
M8	13	F	40	4853	107487
C33	13	M	40	1723	106291
C36	13	F	40	1284	228117
C43	13	M	35	2177	45029
M9	14	M	32	8199	-23276
M10	14	F	42	4937	40001
M11	14	M	41	2681	158551
M12	15	F	43	2609	108554
C31	15	F	37	4288	106059
C46	15	F	43	2924	21665
M13	16	M	54	786	68821
M14	16	F	37	10608	-89390
C25	16	F	39	2184	98204
M15	17	F	40	2612	132907
T1	17	F	49.2	5920	91400
C19	17	F	40	2537	130901
T2	18	M	33.5	5900	152000
C28	18	M	39	5248	174036
C29	18	M	39	10037	-75481
C30	18	M	39	8205	181297
M16	19	F	39	3997	131141
T3	19	M	38.1	8150	111000
C32	20	M	41	15878	-77227
C42	21	M	36	4317	48838
M17	22	F	36	3412	151735
M18	22	M	37	6821	75158
T4	22	M	39.8	7350	68800

*Note: M indicates data from (Maltese et al. 2008), T indicates data from (Tsitlik et al. 1983), and C is new data collected at the Children's Hospital of Philadelphia and analyzed using the methods of (Maltese et al. 2008).