Bertil ALDMAN Award Lecture

A Public Health Priority For Only Ten Percent Of The Car Occupant Population: Why Focus On Children And How Are They Different Biomechanically?

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Abstract The 2014 Bertil Aldman lecture provides a state of the art review of pediatric biomechanics, beginning with defining the scope of the problem of child occupant protection and then describing the recent explosion of data in this field and identifying future research needs in this area.

I. SCOPE OF THE PROBLEM

Worldwide there has been a growing recognition of the importance of protecting the young from injury. In May 2011, the WHO adopted its first ever resolution on child injury prevention urging the development of science-based policies to prevent child injury as well as support of relevant research, capacity building and resource mobilization [1]. This resolution elevated the issue of child injury by recognizing it as a "major threat to child survival and health". Given the high proportion of road traffic injuries in all pediatric injuries, protection of children on the roads has received similar attention. In response to the urgent need to develop traffic safety solutions and the desire to catalyze global activities, the WHO and the United Nations (UN) General Assembly has proclaimed 2011-2020 the Decade of Action for Road Safety, with a corresponding framework for countries and communities to increase action to save lives on the world's roads [1]. The Decade of Action organization has prioritized children, identifying pediatric road safety as the focus for the 2015 Decade of Action Global Road Safety Week in May, 2015.

However an examination of field data suggests that children represent only 10-15 percent of the overall traffic fatality burden in the United States questioning the focus on child occupant protection. These numbers are replicated worldwide where children 0-14 years represent 5-13% of all road traffic deaths depending on the income level of the country [1]. The fundamental issue however is that protection of children on our roads is not only a transportation safety issue but also a public health issue.

Motor vehicle crashes (MVCs) remain a leading cause of death and disability for children and young adults. With the exception of those under one year of age, unintentional injury is the leading cause of death, serious injury, and acquired disability for children, youth and young adults up to 24 years of age in the United States [2]. (Table 1). For children age one to 14 years, motor vehicle crashes (MVC) represented 38 percent of unintentional injury deaths in 2010 outweighing many other causes of death for this age group (Figure 1). This burden extends globally in that road traffic injuries are the leading cause of death among young people aged 15 to 19 years and the second leading cause of death for 5-14 year olds [3]. Worldwide, each year nearly 400,000 people under age 25 sustain fatal injuries on the world's roads – averaging more than 1000 deaths per day [4]. Fatalities represent only the tip of the motor vehicle crash problem for children. In 2011, an average of 3 children age 14 and younger were killed and 469 were injured every day in the United States in motor vehicle crashes [5].

Moreover, their exposure to motor vehicle risk is significant because they travel in motor vehicles nearly as much as adults. On a daily basis, children under age 16 spend nearly as much time in motor vehicles as adults, averaging 3.4 trips per day for a total of 45 to 50 minutes [6].

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< 1year	1-4 years	5-9 years	10-14 years	
Congenital	Unintentional	Unintentional	Unintentional	
Anomalies	Injury	Injury	Injury	
(5,107)	(1,394)	(758)	(885)	
Short Gestation	Congenital	Malignant	Malignant	
(4,148)	Anomalies	Neoplasms	Neoplasms	
	(507)	(439)	(477)	
SIDS (2063)	Homicide	Congenital	Suicide (267)	
	(385)	Anomalies		
		(163)		
Pregnancy	Malignant	Homicide	Homicide	
Complications	Neoplasms	(111)	(150)	
(1,561)	(346)			
Unintentional	Heart Disease	Heart Disease	Congenital	
Injury	(159)	(68) Anomalies		
(1,110)			(135)	

Table 1: Top Five Leading Causes of Death by Age Group, United States - 2010



Figure 1: Top Five Leading Causes of Unintentional Injury Death for Children, United States – 2010

II. ROLE OF BIOMECHANICS

Development of solutions for traffic safety for the young (i.e children, adolescents and young adults) is complex occupying several domains of emphasis: individual, social, vehicular, environmental and medical. (Figure 2) Change in one of these domains often alters the actions in the others and a multi-disciplinary approach to create the scientific foundation is needed. In the design of safety systems, engineers must work in concert with behavioral scientists and epidemiologists to create a comprehensive view of injury and its mitigation [7].

One of the key characteristics of the protection of child occupants is that the protection principles change with developmental age. There are different restraint recommendations for toddlers than young school-age children. Fundamental to these recommendations are differences in biomechanics across the age range. One cannot simply divide the population into 'pediatric' and 'adult' but rather must consider that variations within the pediatric age group. The pediatric structural components of the human body develop and attain maturity through diverse physiologic processes that proceed at varied pace well into young adulthood. These

developmental changes translate into differences in how the child's body withstands and responds to load compared to an adult.



Figure 2: Domains of traffic safety solutions for youth

Historically, the field of biomechanics has estimated the pediatric response utilizing quantitative scaling relationships based on anatomical and material differences. This was limited however because in contrast to the wealth of biomechanical data on the adult response to trauma, pediatric biomechanical data are relatively sparse. The last decade, however, has seen a tremendous increase in contributions to the biomechanics literature based upon pediatric subjects – volunteers, post-mortem human subjects, and animal models - thus increasing our knowledge of how to design injury mitigation systems to protect the young. The remainder of this manuscript will focus on describing the highlights of that recent work and identifying future research needs in this area. Emphasis will be on three primary body regions: the head, the neck and the thorax. This emphasis is based on epidemiological data which highlights the importance of head injury mitigation as the primary prevention priority for children. Injuries to the brain and skull are among the most common injuries sustained by children through a variety of mechanisms. Spine and thoracic biomechanics are not only important from an injury mitigation standpoint but they govern the kinematics of the head during traumatic loading and therefore play a role in head injury protection. Following the biomechanics discussion, I provide several priorities for the future.

III. BODY REGION-SPECIFIC BIOMECHANICS

Brain and skull

Traumatic brain and skull injuries are the most common serious injuries sustained by children in motor vehicle crashes regardless of age group, crash direction, or restraint type [8-13]. Head injuries are responsible for one-third of all pediatric injury deaths [14-15] and are particularly relevant clinically as the developing brain is difficult to evaluate and treat, and even mild brain injuries in childhood can lead to deficits that remain long after the injury [16]. Understanding the mechanisms of brain injury sustained by children requires quantification of biomechanical response – both at the brain tissue level as well as the global head kinematics.

The young brain and skull is a fundamentally different structure than that of an adult. For the youngest children, incomplete development of skull bones, deformability of the sutures and presence of fontanels leads to change in shape with direct impact, often without fracture. That deformation transmits the loads to the brain itself which has higher water content and less myelination than that of an adult. Complete fusion of the bony plates and disappearance of the sutures occurs around 20 years of age when the skull has reached its full definitive size. By age two, the brain has reached nearly 80% of its adult volume and by age five the brain is adult-size.

The emphasis in the last decade has been on quantifying differences in mechanical properties of the brain and skull within the pediatric age range and in comparison to adults. Substantial focus has been placed on the very young. Coats and Margulies tested human infant cranial bone in three-point bending at high rates similar to a fall and determined the stiffness and ultimate stress increased with age [17]. Davis et al examined different regions of a 6 year old calvarium and quantified variations in biomechanical response throughout the skull regions [18]. Brain material properties have been quantified using porcine, rodent and human pediatric tissue [19-22]. These results suggest that at small strains the infant brain is softer than the adult, but at large strains most relevant for moderate and severe TBI, the adult is softer than the infant. More research is required to find the appropriate tissue properties for brain material at both high strain and high strain rates.

The relationship of these biomechanical differences to resulting injury vulnerability remains an area of open question. Human patient data suggest that young children who sustain severe TBI in early childhood or moderate or severe TBI in infancy may be particularly vulnerable to long term poor outcomes [23-26, 16]. Several have attempted to understand if this is a biomechanical or physiological difference with age by applying appropriately scaled inputs to animal subjects of varying ages and measuring clinical outcomes [27-31]. These studies have been contradictory – some show equivalent vulnerability between the young and old, some show vulnerability to increase with age and some show vulnerability to decrease with age. This area clearly deserves further study.

Brain and skull material property data such as those described above are needed to more accurately scale injury criteria [32] or as material properties in finite element models [33-34], for the purpose of predicting skull deformation or brain strain during loading. Relating these biomechanical measures to a diverse set of physiological, pathological, histological, and behavioral outcomes will provide quantitative data upon which to more effectively design safety devices specific for children.

Neck and spine

During the human developmental process, extensive tissue and morphological changes occur in the cervical spine. At birth, the cervical vertebrae are composed of several bony masses which do not fully fuse until age 4 to 6 years and the cervical spine ligaments are lax compared to the adult. The uncovertebral joints of C3-C7, formed between the lateral edges of the superior surfaces of one vertebral body and the inferior surface of the vertebral body directly above, do not form until age 6 years [35] and the facets are predominantly horizontal, thus providing limited restriction of anterior-posterior shear (subluxation) at the facets [36-37]. Changes continue throughout the pediatric age range until the early teenage years. Kasai et al. documented an increase in vertebral body diameter from age 10-14 years, an increase in the initially horizontal facet angle through age 10 years, and changes in the cervical lordosis angle (angle between C3-C7) through the ages of puberty [38]. The thoracic kyphotic angle (angle between T4-T9) has been reported to decrease until approximately 10 years and then increase through young adult hood [39]. Skeletally mature adult vertebral anatomy in the human does not occur until later in the second decade of life. These anatomic changes influence the kinematics and kinetics of the spine in response to traumatic load as well as the trajectory of the head in the case of a restrained occupant.

Age-based differences in kinetics of the cervical spine have been quantified using whole post mortem human subjects (PMHS) pediatric cervical spines as well as isolated vertebral segments from both human and animal specimens. A limited sample of whole pediatric PMHS cervical spines (age 2-12 years; n=10) were subject to sagittal plane bending and tensile load to failure [40] and demonstrated increasing tensile failure load with increasing age. Considering the kinematics of a restrained occupant in which the cervical spine is loaded in tension, further quantification of the biomechanical response in this orientation was critical. Luck et al performed a comprehensive study on 18 PMHS spines (whole spines as well as cervical segments) ranging in age from 20 weeks gestation to 14 years [41-42]. Tensile load to failure sharply increased with increasing age for all spinal levels. Tensile stiffness increased as spinal level became more caudal however no differences in tensile strength were found across spinal level.

Post mortem animal subjects, including porcine, baboon, and caprine specimens, have also been utilized to understand the structural response of the spine [42-50]. In general, with increasing age, bending stiffness, compressive failure load, tensile stiffness and tensile failure load increased. (Table 2)

It has been well documented in the clinical literature that the pattern of cervical spine injuries shifts from the upper cervical spine to the lower cervical spine at approximately 9 years of age. The biomechanics supports this age-based variability in injury patterns. In the baboon, a decreasing trend in tensile failure load was observed when moving from the upper to lower cervical spine except for the youngest specimens who demonstrated no difference across spinal level [45-48]. For the youngest specimens, the upper cervical spine had the lowest tolerance; while for those specimens greater than 8 human equivalent years, the lower cervical spine was the weakest [48].

Age	Tension	Tension	Tension	Tension	Tension	Tension			
group	Failure –	Failure –	Failure – [50]	failure –	Stiffness –	Stiffness –			
	caprine (N)	baboon (N)		FMVSS 208	caprine (N-m)	baboon (N-			
	[43]	[45]		(N)	[43]	m) [45]			
1 year old	0.10		0.25	0.24	0.13				
3 year old	0.16	0.33	0.30	0.34	0.18	0.54			
6 year old	0.30	0.55	0.37	0.45	0.38	0.71			
12 year old	0.62	0.66			0.66	0.76			
Adult male	1.00	1.00	1.00	1.00	1.00	1.00			

Table 2: Tensile scale factors developed from several biomechanical studies

While quantifying the response of isolated cervical spine segments is an important part of the biomechanical puzzle, understanding how those parts work together in combination with muscles and other supporting anatomy is ultimately the question to answer when designing a safety system for children. Our research group at Children's Hospital of Philadelphia, through non-injurious frontal sled tests on pediatric human volunteers, quantified the kinematic response of the restrained child's head and spine and compared it to that of the adult [51]. Normalized forward and downward excursion of the spine significantly decreased with age (Figure 3). The majority of the spine flexion occurred at the base of the neck not in the upper cervical spine and the magnitude of flexion was greatest for the youngest subjects. Additional flexion occurred in the thoracic spine as well. Using inverse dynamics, we quantified the forces and moments at the upper cervical spine and demonstrated that axial force decreased with increasing age while flexion moment increased with increasing age [52].



Figure 3: Age-based differences in normalized forward excursion for restrained human volunteers in a lowspeed frontal sled test. Skeletal landmarks include head top (HT), C4, T1, T4, and T8. (Reprinted with Permission The Stapp Association from [51])

Our research group furthered this line of research and examined the response of human volunteers in lateral and oblique far-side loading [53]. The effects of age on overall resultant excursion were not clear however we did observe that in the same loading condition, younger subjects demonstrated less lateral displacement and more forward displacement than adults suggesting that the kinematic coupling of the spine is influenced by the anatomic differences with age highlighted above.

The benefit of human volunteer tests is that the response incorporates physiological muscle response and analyses do not require considering a transfer function between an animal model and the human. However, for subject safety such testing cannot be performed at injurious rates. The human body has been established to have a rate-dependent viscoelastic response so it is likely that these findings would differ at higher rates. To address this limitation, Lopez-Valdes combined the human volunteer data of Arbogast et al. with adult PMHS tests, animal surrogate tests and in vitro bending tests of sections of the pediatric and adult thoracic spine to develop innovative scaling methodologies to predict the response of the child at higher speeds [54]. This work is an encouraging start but more research is needed in particular understanding the whole body response of children using a PMHS model (see below).

<u>Thorax</u>

Radiological assessment of torso shape has documented variations with age [55-58] that may lead to agebased differences in action and timing of a seat belt in controlling the kinematics of the torso. Specifically, for the young, the sternum is multi-segmented, the ribs are more flexible, the connection of the ribs to the other thoracic skeletal structures is more cartilaginous, and the orientation of the ribs is more horizontal [35].

These anatomical characteristics have translated into differences in biomechanical response. Ouyang et al. [59] tested 9 pediatric cadavers aged 2 - 12 years, subjecting them to Kroell-style [60] frontal sternal impact. These data were then re-analyzed by Parent [61] to correct for previous analytical limitations. He demonstrated that compared to the existing scaled biofidelity corridors, the three-year-old response corridor shows similar deflection but higher force, while the six-year-old response shows similar force but higher deflection.

Our research group has pursued alternative methods for obtaining pediatric thoracic force and deflection data and have used cardiopulmonary resuscitation (CPR) in the clinical setting to determine stiffness response of the pediatric chest. Specifically, we have utilized a load cell and accelerometer Force-Deflection Sensor (FDS) (Philips Healthcare, Andover, MA) which has been integrated into a patient monitor-defibrillator to provide visual and audio feedback on the quality of CPR chest compressions [62]. The FDS is interposed between the hand(s) of the person administering CPR and the sternum of the patient, and it records force and acceleration data during each compression cycle. Maltese et al. described the force-deflection response of eighteen subjects ages 8 to 22 years who received CPR chest compressions [62] and compared them to similar data from adults [63-64]. This analysis demonstrated that the elastic stiffness of the thorax increases from age 8 to age 40, but decreases after age 40 reaching similar levels to the pediatric subjects by age 50. (Figure 4). This bimodel relationship with age has been replicated, albeit with smaller sample sizes, with pediatric and adult PMHS specimens [65].

These biomechanical differences with age translate into clinical differences in injury patterns. Clinically, children frequently receive lung contusions in the absence of rib fracture in contrast to adults where rib fractures are the dominant thoracic injury. This has implications for development of thoracic injury criteria for children. Criteria and thresholds developed from adult PMHS impact data and scaled to the child will predict best the injuries present in those experiments – rib fractures with the occasional soft tissue injury, but thoracic injury criteria for children must do the opposite - predict primarily soft tissue injury, with the occasional rib fracture. This dichotomy deserves further study.



Figure 4: Elastic force at 15% compression of the pediatric chest during cardiopulmonary resuscitation. (Reprinted with Permission The Stapp Association from [62]).

IV. FUTURE PEDIATRIC BIOMECHANICS PRIORITIES

While there has been tremendous progress in the last 10 to 15 years in generating pediatric biomechanics data, the range of what is known even at the material level is dwarfed by what remains unknown. Several future biomechanics research priorities have been identified.

1. Emphasize pediatric brain injury biomechanics

Biomechanical efforts need to continue to focus on prevention of brain injuries. Worldwide, the incidence of TBI has been estimated at 500 million new cases annually (circa 1985), and due to increasing global automobile use and declining deaths due to infectious diseases, TBI is becoming the global dominant source of mortality and morbidity [66]. TBI is particularly devastating to the young. Hospitalization costs associated with pediatric TBI are estimated to exceed \$1 billion annually [67], and post-TBI neuropsychological sequelae include altered behavior and diminished academic achievement, increased family strain and burden of care, and quality of life similar to pediatric oncology patients, all extending for years following the injury event [16].

As the risk of head injury in motor vehicle occupants increases with increasing child age [9], there is need to develop TBI mitigation strategies that protect the child that has outgrown the add-on child restraint. To do so, research is needed to improve computational models of the pediatric brain and to develop accurate injury criteria and understand how such criteria change within the pediatric age range.

2. Quantify the whole body response of children in potentially injurious impact loading situations.

Knowledge of the whole-body response of children in potentially injurious impact loading situations remains essentially undefined. Such data are critical for determining the loads to which organ-level materials are exposed during the complex external loading environment of a car crash. Therefore, while recent advances that define material-level behavior have been useful for quantifying the risk of failure when a material is loaded, further research must quantify the relationship between whole-body dynamics and loads to internal tissues in order to maximize the utility of pediatric tissue material property data.

There has been limited dynamic study of whole body pediatric PMHS. To our knowledge, 15 pediatric whole body PMHS sled tests have been conducted to date. Subjects range in age from 2 to 13 years at speeds from 31 to 50 km/hr [68-72]. These studies examine a diverse set of restraint conditions and the instrumentation, documentation and injury coding reflect the fact that these studies were conducted several decades ago and therefore limit the contemporary value of the data to be obtained. Future advances in pediatric crash

protection must be informed by the impact testing of intact, pediatric and adolescent PMHS. The biomechanics and medical communities must work together to build consensus, identify impediments to the successful procurement and testing of pediatric PMHS, develop best practices, and determine appropriate mechanisms for disseminating this information to stakeholders. Impediments to the successful procurement and testing of pediatric PMHS include a lack of knowledge by the broader research community regarding the necessity of this research, general sensitivity to the use of pediatric PMHS for impact testing, the logistical challenges of locating appropriate pediatric PMHS and linking the specimens with qualified research labs, the lack of consensus regarding the processes and procedures for acquiring and using pediatric PMHS, and the lack of resources for this type of research.

3. Develop biomechanical research tools specific for children

The primary tools by which the automotive industry and biomechanics research community evaluate occupant protection are anthropomorphic test devices (ATD) and computational models of those ATD. The effectiveness of these efforts is directly related to how well the ATD mimic actual humans in their kinematics and injury prediction ability. Pediatric ATD have largely been size scaled versions of adult ATD as pediatric-specific biomechanical data have been lacking. As described above, the last 10 years has seen an increase in fundamental biomechanics data that quantify the child and adolescent's component level response to dynamic loading (for a comprehensive summary see [73]). It is critical that these data be incorporated into ATD design in a timely manner.

As is the case in all areas of engineering design and evaluation, computational assessment of occupant protection is gaining in importance and value. Computational models of the ATD were initially introduced to fill this need. There has been tremendous effort to develop these models for adults, most notably through the Global Human Body Modeling Consortium (www.ghbmc.com), the development of THUMS (Total Human Model for Safety) and initiatives within the EU child safety project CASPER. The development of pediatric human body models must be part of these efforts. Furthermore, the field is benefitted by harmonization of these models rather than having, for example, several human body models simulating a 10 year old child. Integration of these efforts and knowledge into a single model is not trivial and presents logistical challenges but in the end is the most resource efficient approach. Lastly, it is important that these models have the capability to be modified parametrically to account for natural variations in anthropometry, structural geometry and material properties.

4. Study realistic postures and positions child occupants assume and use this data to quantify biomechanical response

Restraint performance is evaluated using ATD positioned in prescribed, optimal seating positions. Human occupants, children in particular, have been observed to assume a variety of additional positions that involve changes in posture and alterations in seat belt placement and geometry which are assumed to potentially affect restraint system performance (e.g. [74-75]).

There have been several efforts to describe these position and postural differences using naturalistic, observational methods. These methods typically involve a parent driving an instrumented vehicle with no special instructions, no experimenter present, and data collection via unobtrusive instrumentation. Initially these efforts, while being critically important for defining the nature and magnitude of the problem, have been largely qualitative and/or dichotomize the out-of-position as present or not (i.e. leaning forward out of the restraint as a yes/no variable). More recent studies are underway that use emerging technology to quantify these positions and postures [76-77].

These data provide a unique window into the dynamic kinematics of pediatric occupants in near crash events such as emergency braking and evasive swerving maneuvers and in limited numbers, actual crashes. Consideration of how to use these data for obtaining biomechanics knowledge represents an intriguing methodological challenge.

V. PARTNERSHIPS WITH OTHER DISCIPLINES

Biomechanists cannot work in isolation. They must collaborate with epidemiologists, behavioral scientists, policy makers, and industry. Biomechanics resources are limited and therefore research must data driven. Rigorous data can provide the scientific context and real world grounding for a diverse set of biomechanics activities and guide priority setting. Two pediatric research priorities in other disciplines that influence the future of pediatric biomechanics are:

1. Continuation of crash injury surveillance specific to children in a way that is nationally or regionally representative

We need effective crash surveillance systems that ensure high quality child-focused data will be collected on a sufficient number of children and adolescents to be representative of the geographic area of study. There is compelling evidence that such data are critical for continued advancements in occupant protection for this age group. Field data should dictate the direction of future biomechanics activity.

2. Adapt the Abbreviated Injury Scale (AIS) to include varied outcomes such as long-term disability and cost. Quantify the influence of age on the interpretation of specific injury codes.

The Abbreviated Injury Scale (AIS) was originally created by the Association for the Advancement of Automotive Medicine (AAAM) as a threat-to-life scale and has been used as a standardized benchmark of injury severity. As motor vehicle crash-related deaths are decreasing in high-income countries, particularly for children, there is an increased need to understand the non-fatal burden of injuries including functional disability (physical, cognitive, behavioral) and quality of life. In addition, it is also critical to measure the cost of fatal and non-fatal injuries to society including acute and chronic medical care expenses, wage and productivity losses, and reduced quality of life costs. Since pediatric occupants differ from adults (and children of various ages differ from each other) in terms of injury tolerance and recovery, age-specific differences in injury risk and outcomes should be further quantified and systematically categorized. Incorporating disability outcomes, cost, and age-specific measures into the AIS can further enrich this ubiquitous and long-standing tool for injury severity, which will benefit injury prevention (through more informed injury risk curves) and treatment (through improved injury triage, pre-hospital and acute care, and rehabilitation).

VI. CONCLUSIONS

Partial success has been achieved in the field of child occupant protection. In the US, MVC fatalities for those 12 and under have decreased 43% in the past decade, but still more than 650 children age 12 and under died in crashes in 2011 [78]. In the European Union, child occupant fatalities (0-14 years) have decreased from approximately 1800 in 2001 to approximately 800 in 2010 [79]. Future gains will be more difficult as the problems have become more challenging. The explosion of pediatric biomechanical knowledge in the last decade will facilitate development of better child ATD and computer models and therefore improve the tools available to design better interventions to keep children safe. Future pediatric biomechanics research must focus on defining anthropometrics, age-dependent injury tolerance and material and whole body response across the entire pediatric age. These data should be quantified at loading rates relevant to injurious conditions, and across all impact directions. Effort should be placed on validation or further development of the scaling methodologies – both across age and across species – in order to benefit from the richness of existing animal model, PMHS and human volunteer data.

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